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Abstract

This master's thesis introduces and analyzes a new approach for feedback control system, designed for mechanical ventilation. It is called Adaptive Volume Ventilation (AVV) and is for this particular study, aimed to be implemented into a ventilator as a part of a military project called LSTAT.

The adaptive volume ventilation controller uses three continuously measured lung physiology parameters resistance, compliance and dead space. Together with the parameter entered by the user called gross alveolar ventilation, these values are used to calculate the optimal breathing frequency and tidal volume. The optimization criterion is the minimal work of breathing, which is believed to minimize the trauma to the lung tissue, caused by the forced ventilation. The controller was implemented and tested onto a medical ventilator and a mechanical test lung.

Different techniques for evaluating the system performance are being used and the result shows a fast and accurate controller. It is also showed that the work of breathing is minimized and that the system is stable and robust. The result leads to the conclusion that a controller using the proposed technique would be safe and show strengths compared with normal control systems designed for mechanical ventilation.

RAPPORT 92

A FEEDBACK CONTROL SYSTEM FOR MECHANICAL VENTILATION

ETT REGLERSYSTEM FÖR MEKANISK VENTILERING

By
Ingmar Blomqvist
Together with
University of Utah

A MASTERS THESIS
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Sammanfattning

Detta examensarbete introducerar och analyserar ett nytt sätt att styra en mekanisk ventilator med hjälp av regler teknik. Metoden kallas adaptiv volym ventilering (engelska Adaptive Volume Ventilation) och för denna studien har malet varit att inkorporera systemet och ventilatorn i en militärt projekt kallat LSTAT.

Ett kontrollsystem enligt metoden för adaptive volym ventilation, använder sig av tre lungfysiologiska parametrar, resistans, compliance samt dead space, som samtliga mäts kontinuerligt. Tillsammans med den parameter som användare anger, den sa kallade netto alveolara ventilationen, används dessa värden för att beräkna den optimala andningsfrekvensen och tidal volymen. Det optimeringskriterium som används är andningens arbete, som ocksa tros leda till en minimering av trauman som lungvävnaden utsätts för under ventileringen. Reglersystemet är implementerat och testat pa en medicinsk ventilator samt en mekanisk testlunga.

Olika tekniker för att utvärdera systemets uppförande har använts och resultaten visar att systemet är snabbt och exakt. Rapporten visar även att arbete verkligen kommer att minimeras och att systemet är stabilt och robust. Resultaten leder till slutsatsen att den föreslagna metoden för styrning av mekanisk ventilering skulle vara säker och visar styrkor jämfört med andra metoder.

Acknowledgements

I would first like to thank Dwayne Westenskow, my supervisor at University of Utah, for his help and valuable support during this project. I would also like to thank Bernd, Kai and all other people at the bioengineering lab for your inspiration and good opinions throughout the project.

The AVV controller was first developed by Craig Flanagan and this study is based on the result from his work.

I would also like to thank Lars-Gösta Hellström, my supervisor in Sweden for his help to make things run smoothly as well as his ambitious proof reading. Hakan Elmqvist my examiner for his help with solving all organization issues.

Finally I would like to send special thanks to Dwayne Westenskow and Andras Gedeon for creating this opportunity.

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Chapter 1

Introduction

The specific aim for this project is to develop, implement and test a new closed loop control method called AVV[1]. This controller will take the lung physiology of the patient into account and calculate the optimal breathing frequency. The goal is to implement this controller to the ventilator that is incorporated in the LSTAT project. By reducing the number of setting necessary for the operator to enter, the handling proceeders can be simplified. This makes it possible for a non trained person to operate the ventilator. It is also a goal to design a system that continuously measures several respiration parameters, that will be used to evaluate the system performance.

1.1 Ventilators for medical use

Breath is, for natural reasons, vital for the human being. Lack of oxygen will cause the human body irreparable damage within minutes, and lead to death after another few minutes. The first organ to take damage is the brain, which will function for only three minutes without oxygen. It is therefore necessary to be able to assist people who can not breathe on there own. For at least a couple of hundred years, mankind has investigated these possibilities. Today a wide range of systems are available, anything from supported breathing equipment with a mask to complete life support ventilation with intubation.

1.1.1 History

Technical devices to support ventilation started to be developed during late nineteenth and early twentieth century. One of the first model, named the iron lung, consisted of a tank in which the entire body of the patient, except for the head, was encapsulated. A negative pressure was applied, which forced the chest to expand and the lungs to take in air. The pressure differences were applied by a mechanical bellow. These systems had some problems due to reliability. However physicians argued that it was more physiological

[1]Adaptive Volume Ventilation

to apply negative pressure to the lung via the chest wall rather than to exert a positive pressure in the airways. An alternative to the tank respirator was presented with the Cuirass ventilator. The principal was the same, however the intermittent negative pressure was applied either to the chest wall or the chest wall and the abdomen. Today neither the iron lung nor the Cuirass ventilator are being used in normal day to day ventilation therapy. However both these systems have shown to have advantages for chronic ventilation support and are therefore occasionally being used to treat these diseases.

The modern mechanical ventilator was first developed by Heinrich Dräger during early 20th century. During the 1940s and early 1950 the efforts to develop systems for ventilation with intermittent positive pressure via the airway were intensified. This resulted in the first automatic mechanical ventilator, the spiropulsator, developed by the Swedish company AGA. Essential for the development of ventilatory systems using intermittent positive pressure, was the improvement of the endotracheal tube and intubation.

Today there exists a wide range of different ventilators, that covers a broad range of applications. There are many different diseases and conditions that result in the need for respiratory assistance, anything from assisted breathing at home to life support ventilation in critical care units. The most commonly used method today is ventilation with positive pressure, either noninvasive with a mask or invasive with an tracheal tube.

1.1.2 Classification and definition

Today, most ventilators uses intermittent positive pressure. A good way of classifying ventilators would be in the way they control and adjust this pressure throughout the breath cycle. This way of classification will result in four major groups.

I **Pressure controlled** This system uses compressed gas to generate the force that will produce the work of breathing. The air gas flow to the patient will be relatively constant. Trigger point for changing from inspiration to expiration will be the inspiration pressure. This might cause serious problem if, due to system failure, the set inspiration pressure is never reached.

II **Flow controlled** Similar to the pressure controlled system, the force is generated using compressed gas. The flow will be controlled by an air regulator, which also triggers the change from inspiration to expiration.

III **Time controlled** This system normally uses compressed gas to drive the ventilation. However, the trigger point will be set by the time limit, given through the respiration rate.

IV **Volume controlled** Together with this system, the patient will be ventilated with a defined volume. The system will automatically adjust the flow and pressure, to deliver the set volume. There are normally two different volumes being used, either the tidal volume[2] or the minute ventilation[3].

Modern ventilators are often equipped with more than one of these control mechanisms. These mechanisms are build into different modes at which the ventilator can work at. Some of the more commonly used modes are listed below.

Controlled ventilation In this mode, the ventilator has complete control over ventilation. It will not react on any spontaneous breathing performed by the patient.

Assist-Controlled Ventilation This mode gives the patient the ability to trigger the machine to make an assisted breath. The ventilator will sensor negative inspiratory pressure during the pause time and start an assisted breath. If the patient does not trigger the machine, it will deliver the minimal minute ventilation set by the clinician.

Synchronized Intermittent Mandatory Ventilation (SIMV) This mode permits the patient to breath spontaneously while periodically receive ventilator-generated assisted breaths. These breaths will be synchronized with the spontaneous breathing of the patient and therefore cause minimal disturbance to the patient.

Continuous Positive Airway Pressure (CPAP) Assisted ventilation is not supplied but rather continues positive mean pressure. CPAP-mode permits the patient to breath spontaneous around this positive mean pressure.

Positive End Expiratory Pressure (PEEP) Similar to CPAP, however PEEP retains the positive pressure at the end of the expiratory phase. This mode is primarily used to increase the volume of residual gas in the lungs.

SIGH A feature that is installed on several ventilators today. It will cause the patient to take a breath every 100th breaths, that has double or triple the normal tidal volume.

More useful definitions

- **I:E-Ratio** The ratio of Inspiratory and Expiratory time.
- **Airway resistance** The resistance caused by a gas flow through the airway system. Measured in units of $cmH_2O/sec/l$.
- **Compliance** Compliance of the lung and thorax tissue. Measured in units of l/cmH_2O.

[2]Size of each breath including dead space; given in ml

[3]Total air gas volume delivered to the patient during one minutes ventilation. Defined as tidal volume multiplied with respiration rate and given in l/min

1.2 Closed loop control system for medical ventilators

Today closed-loop or feedback control systems can be found in many different applications in everyday life. One example would be the thermostat controlling the temperature in a normal house. If the temperature falls below a certain lower set temperature, the heater will be turned on and as the temperature reaches the higher set temperature it will be turned off. Another example would be the cruise controller of a normal car. A closed-loop control system can be described as a system that measures one or more control parameters and comperes it to a set value. If there is a difference, or error between the two values, this error will be fed back to the output variable that is being used to control the system. The change in this variable will more or less rapidly cause the error to decrease. Today a number of more sophisticated systems are available, that also will take the speed of the change into account.

1.2.1 Background

In 1957, Frumin [6] reported the first working closed-loop controller for a medical ventilation system. This system was designed to control the ventilation of patients during normal anesthesia. The end tidal CO_2 concentration was measured with a sensor placed in the breathing circuit and compared to the desired value. If the measured end tidal CO_2 concentration exceeded the value which the system was set to maintain, the inflating pressure automatically was increased. The deeper respiration ensured that more CO_2 was exhaled and the end-tidal $\mathrm{CO_2}$ rate was decreasing. The small adjustments continued until the patients CO_2 concentration reached the desired value, and there after the pressure was held constant.

Since Frumin first reported a successful use of closed-loop control systems for mechanical ventilation, there has been a number of studies showing the strength of feedback control systems for medical ventilation. It has been shown that feedback control systems perform better for several specific tasks, than a clinician. A specific end tidal CO_2 concentration can be achieved more rapidly and held more constantly with a closed-loop controller than with manual control. It seems as if computer controlled ventilators are more successful in cases where the anesthesiologist is distracted or lacks experience. However, the anesthesiologist may use other parameters and observations to reach his conclusions, observations that can be useful during rapid changes. In these cases, where there are no strict rules to follow, a closed-loop control system might run into problems. Therefore, it is much to early to say that computers handle ventilation better than clinicians. More studies have to be done.

The rapid development of the microprocessor and the widespread use of computers, have improved the closed-loop control systems fast during the last couple of decades. The development will most likely continue in the future; safety and stability for soft and hardware will be increased. This will create good opportunities for a more widespread use of closed-loop systems in the field of medical technology.

1.2.2 Ventilator control systems

CO_2 based controller This controller will adjust the patient ventilation to reach a certain value of the PCO_2. The technic was first introduced by Frumin [6] in late 1950s and has since then proven to perform well by several authors. Ohlson [5] used a microprocessor based feedback controller to adjust minute ventilation based on the end tidal CO_2 concentration. Arterial PCO_2 was continuously monitored. The results showed that the controller kept $PaCO_2$ within 1.2 mmHg of the desired value when CO_2 production increased with 44%. However the controller did not keep the $PaCO_2$ at a desired level when a significant change in the blood flow to ventilation ratio occurred. Chapman [3] described a very fast PI-controller[4], that uses the end-tidal CO_2 fraction to adjust the minute volume of a ventilator. This controller was constructed to settle to the new set point in less than 60 seconds, with a overshoot not bigger than 20%. The controller showed the ability to maintain the end-tidal CO_2within 0.1% of the set point.

Although these systems keep the end-tidal CO_2 well within 3 mmHg of the desired value, the arterial $PaCO_2$ may show a great difference from the desired value. These differences are limited for healthy persons, but with abnormal ventilation to perfusion rate, the error starts to increase. To be able to better control the arterial $PaCO_2$, a intra-arterial sensor for PCO_2 or pH can be used. This will make the results much better, however it is not justified to use invasive sensors in most clinical applications. Coon [7] has shown in his study that the pH-value can be used as the controlled variable. Coon [7] used the systemic arterial blood pH, measured by a dual function pH/P_{CO_2} intra arterial sensor. The system responded appropriately to changes in the set point pH from 7.30 to 7.50, as well as to infusion of lactic acid. Long-term ventilation with systemic arterial blood pH servo control ventilator proved to be feasible.

Adaptive Lung Ventilation As presented in the previous chapter, closed-loop control systems traditionally has been based on end tidal or mean expired CO_2. The controlled variables have usually been the tidal volume V_t and the respiratory rate RR. One of the disadvantages with this approach is that it does not take into account the size of the patient or the lung mechanics. Systems which are based on the end tidal CO_2 may therefore, not respond correctly, in cases where the lung perfusion rapidly decreases. Such a case, could for instance be a patient with a lung emboli. In this case, the end tidal CO_2 would decrease and the controller, consequently, would decrease the ventilation. This would, for obvious reasons, be the wrong thing to do; instead increased ventilation is needed. Laubscher [8] presents a new approach to closed-loop controlled ventilation called Adaptive Lung Ventilation ALV. The ALV as described is based on a pressure controlled ventilation mode where the clinician enters the Gross Alveolar Ventilation GAV[5]. The ALV then tries to achieve this goal by adjusting mechanical rate and pressure. These adjustments are based on measurements of the patient lung mechanics and the dead space. The study showed that the system works fine for different lung pathologies, regarding speed, overshoot and steady state performance.

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[4] Proportional-Integral controller

[5] $V_{gA}l/min$

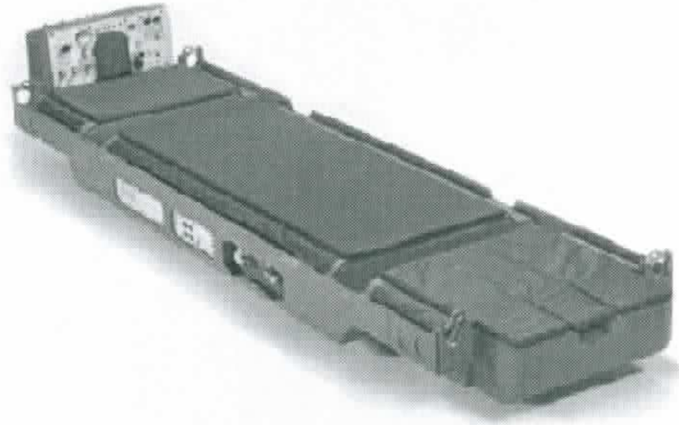

Figure 1.1: The LSTAT - Life Support for Trauma And Transportation - is a compact medicare unit. It includes advanced life support treatment onboard a normal NATO stretcher.

For the three different lung pathologies that were tested, the rise time never exceeded 81 seconds, the overshoot were below 8.0% and the steady state performance were below 5.6%. The ALV uses an optimization criterion based on minimal work of breathing, first introduced by Mead [4]. This will be described more in detail in chapter 2. This is the same criterium as the one being used in the AVV controller.

1.3 LSTAT

Life Support for Trauma And Transportation (LSTAT^{tm}) is a compact and transportable medical care unit. The LSTAT concept was developed by Northrop Grumman Corporation in conjunction with U.S. Army. The first prototype was completed in 1995 and has since then been tested and evaluated by the U.S. Army, Air Force and Marines. The basic idea of the LSTAT is to provide a more sophisticated trauma care closer to the battlefield or accident scene.

1.3.1 Background

A significant proportion of the military field medical effort is currently consumed by post surgical patients, who must, according to current doctrine, be stabilized before evacuation [1]. For critical patients this can take several days or more, before they can withstand the stress added with ground or air transportation. The attendant high ambient noise and vibration drastically degrades the care-giver's diagnostic and therapeutic capability. Heart

Figure 1.2: In April 2000, LSTAT starts to be used by the US army in Kosovo.

and lung sounds are difficult to detect, and simple therapeutic actions becomes difficult in a vibrating environment. These conditions will create hazardous situations for patients, since medical problems may not be detected, or correctly treated. The delay, created by the lack of possibilities for critical care transportation, will significantly increase the burden to the ground forces, a burden that could be taken away if the patient could be transported earlier in his resuscitation procedure. To address these critical care transport requirements, the LSTAT project was initiated in April 1993. The operational objectives were to:

- Improve field mobility
- Provide evacuation capability for transportation of unstable patients
- Decrease patient holding requirements by allowing earlier evacuation
- Optimize use of care-giver's time by automating therapeutic devices

The LSTAT have been tested and used in a number of military operations. November 1999 three LSTATs participated in the joint service Pacific Warrior 1999, Hawaii. The exercise included multiple elective surgeries performed aboard LSTAT, and included operational use in field exercises. LSTAT provided patients continuous care throughout pre-op, intra-op, and post-op phases, bringing the total number of such surgeries to nearly 30. March 2000 LSTAT training for U.S. Army MASH 212th was conducted in Germany. In April 2000 the U.S. Army begun using the LSTAT in Kosovo.

The integration of LSTAT into civilian emergency health care will possibly enhance management of acute injured persons. It would be particular useful in cases where evacuation possibilities are limited or delayed.

1.3.2 System description

The LSTAT has three basic components:

I The base unit

II A NATO stretcher

III A canopy to cover the entire patient

The base unit include several medical devices to fulfill the diagnostic and therapeutic needs. It also include capability to monitor and log medical parameters, system performance data, and patient data to an onboard CPU. The base unit includes the following components:

- Ventilator with oxygen source
- Suction pump
- Capnograph
- Defibrillator
- Environmental control
- Onboard computer with a Datapak for monitoring different parameters.
- Battery

The Datapak includes a physiological monitoring system that acquires and archives the following physiological parameters:

- ECG
- Blood pressure
- Heart rate
- Respiratory rate
- Pulse oximetry

Some of these parameters are used in servo control system built into the LSTAT.

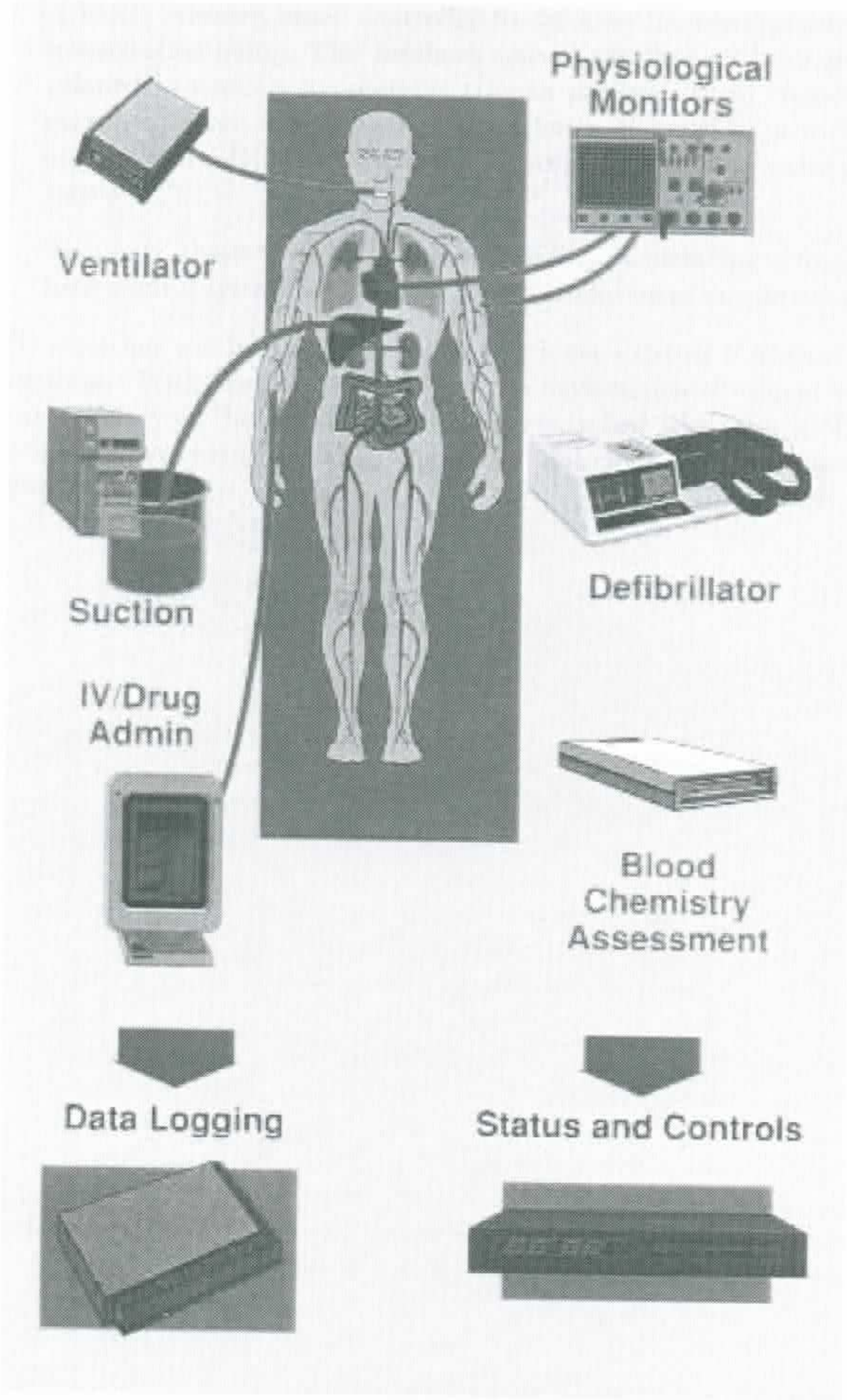

Figure 1.3: The LSTAT is based on a standard NATO stretcher. To this stretcher several devices for life support monitoring and treatment are connected.

- A blood pressure based controller to drive an intravenous fluid resuscitation pump. This feedback control system have been developed to manage hypotensive trauma patients, when trained personal are not available. The algorithm is designed to provide appropriate volume of fluid, avoiding over or under fluid resuscitation.

- This M.S. thesis investigates the possibility of including a feedback control system for controlling the ventilation of the patients.

The NATO stretcher will integrate the system with the existing NATO standards and evacuation platform. With the canopy in place, the environmental system will keep the temperature and quality of the circulating air at a controlled level. Each of the two air handling systems that are being used (the ventilator and environmental control) operates with NBC filters[6].

[6]Nuclear, biological and chemical warfare

Chapter 2

Methods

2.1 System description

A computer controlled feedback system has been implemented to control tidal volume and respiratory rate of a Impact Uni-Vent ventilator. The system can be seen in picture 2-1. It has been built and tested to investigate the possibility for using a closed-loop controller, to control the ventilator that is included in the LSTAT project. This would make the procedure of handling the ventilator much easier.

The system, shown in figure 2-1, contains four cornerstones. The control part of this system, the brains so to say, is included in a normal computer. Communication with the computer occurs over serial and parallel ports. On this computer a program using C++ was developed. This program includes communication, data acquisition, and control routines. For data acquisition a respiratory profile monitor called CO_2SMO Plus, delivered by Novamatrix, is being used. This box will communicate with the computer over a normal serial port. CO_2SMO Plus uses a flow sensor to measure the necessary respiratory values that will be used to control the ventilator. The flow sensor that CO_2SMO Plus uses, is attached to the hose that delivers the ventilation gas to the patient. By putting the sensors as close to the patient as possible, the systemic error can be minimized and the actual values for patient ventilation can be measured. If the systemic error is small, the signals that the CO_2SMO Plus reads is directly correlated to the settings of the ventilator. The ventilator is controlled by the computer. The computer gives the ventilator commands using the parallel port. These settings are read to two digital potentiometers. By regulating the resistance of these potentiometers, all different values for tidal volume and respiratory rate can be controlled. How this works and why this approach is being used will be explained further in chapter 2.1.2. The ventilator uses these and other settings, to ventilate a test lung. The test lung being used in this study is a Dual Adult TTL Training/Test Lung, delivered by Michigan Instrumentation Inc. This test lung can simulate a patients resistance and compliance.

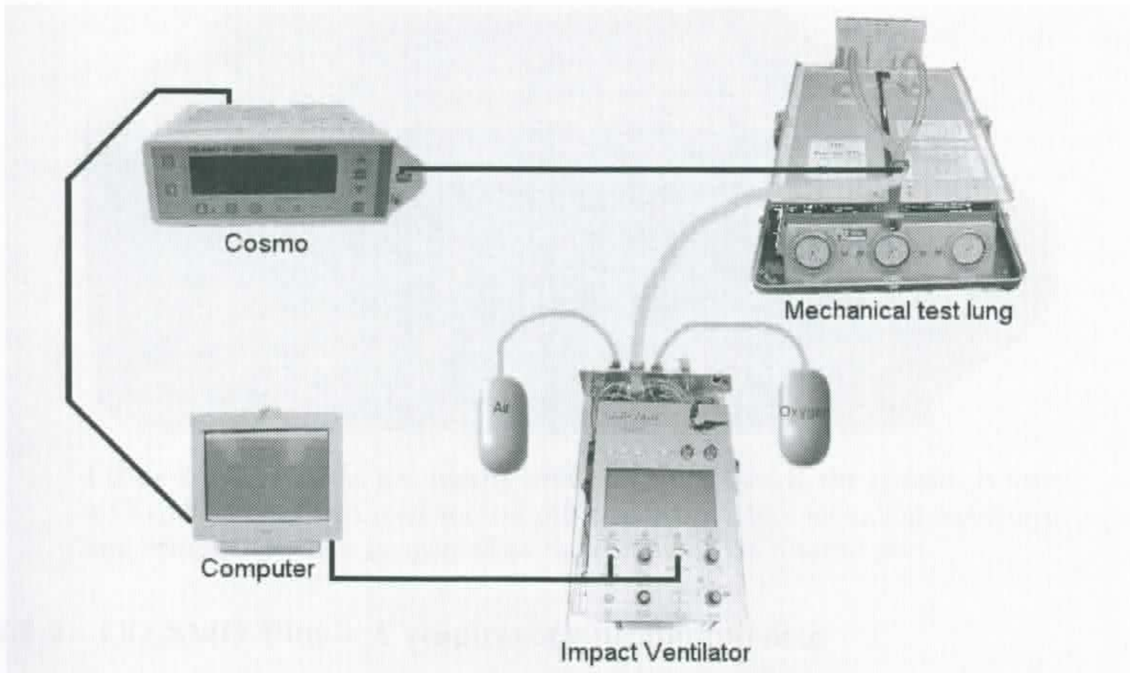

Figure 2.1: The system used contain four cornerstones. These are shown in the picture together with connections between them.

2.1.1 Computer

Most of the communication in the system is performed at a higher discrete level. Therefore the computer has a central position in the system. In this study, a normal Dell computer, with a 450 pentium III processor has been used. The computer uses two communication ports to send and receive information to the system. A 9-pin serial port is used, together with a 9 to 25-pin converter, to connect the CO_2SMO Plus. A 25-pin parallel port is used for communication to the Impact ventilator. The serial port uses a normal RS-232C. This is the classic serial port, standard equipment for nearly every computer, and so also the CO_2SMO Plus supports RS-232C. For communication with CO_2SMO Plus a bit rate of 19200 bits per second is being used. Serial ports in general, and especially RS-232C, are slow but have the advantage of being stable. The protocol allows only one pin to be used for transmitting and another for receiving data. Consequently up to 12 clock cycles will be used to send one character (8 bits). The parallel port on the other hand uses 8 different pins for sending and receiving, making it possible to send one character each cycle. A standard IBM parallel cable is utilized to send data to the two digital potentiometers, which controls the ventilator. Communication protocol for these potentiometers, will be presented in 2.1.3.

The program to run the closed-loop controller has been built with Borland C++ builder 3. This is a ANSI C++ visual development tool, that supports window building technic.

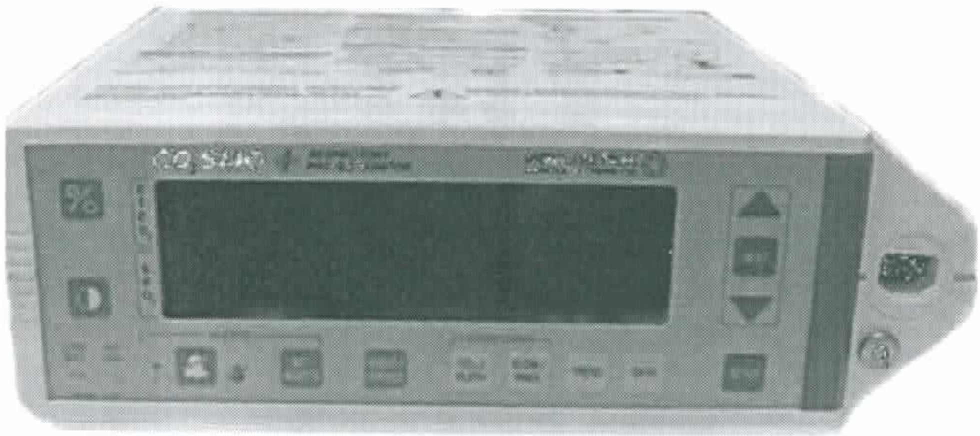

Figure 2.2: This is the respiratory profile monitor used in the system. It uses a CO_2/flow sensor to read several different parameters including resistance and compliance. It is connected to the computer via a serial port.

2.1.2 CO_2SMO Plus - A respiratory profile monitor

The feedback controller used in this study uses the resistance, compliance and dead space to calculate the correct target values for tidal volume and respiratory rate. To monitor these values a respiratory profile monitor called CO_2SMO has been used. This device uses several sensors to calculate all kinds of different respiratory values. The CO_2 sensor utilized in CO_2SMO uses the infrared absorption technique, which has been used clinically for more than two decades, but still remains the most popular technique. The principal is based on the fact that CO_2 molecules absorb infrared light energy of specific wavelengths. The absorbed amount of energy will be directly related to the CO_2 concentration. By comparing the measured light intensity to a calibrated value, the absolute CO_2 can be calculated. Flow measurements are made by a differential pressure pneumotachometer. Respired gas flowing through a flow sensor will create a small pressure drop across the two tubes connected to the sensor. The pressure drop is transmitted to a differential pressure transducer that is being used to calculate the flow. The CO_2SMO calculates and presents the respiratory values on the front display, but it also comes with the possibility to send the data to a data collecting unit over a serial port. For this study, data has been transmitted and read to a computer (chapter 2.1.1).

In this study, the CO_2SMO is used to collect data for flow, pressure, tidal volume and respiratory rate. These values are used to evaluate how accurately the system works, in other words to see if there are any differences in the set and delivered values. In this first study, no CO_2 gas is introduced and the mechanical test lung will for obvious reasons not produce any. Because the CO_2 gas are used to measure the dead space we cannot measure this parameter. Therefore the dead space added by the tubes was measured manually and implemented as a fixed value into the program.

Figure 2.3: The medical ventilator used in this study is a Impact Uni-Vent Model 754. It is a particulary robust ventilator that uses volume settings to specify the ventilation pattern.

2.1.3 Impact Uni-Vent - A medical ventilator

According to the proposal given from the LSTAT-group, a ventilator called Uni-Vent^{TM} Eagle^{TM} model 754 delivered by Impact^{TM}, will be included in the LSTAT. A modified version of this ventilator has been used in this study to implement and test the feedback control system. It is a very compact and portable ventilator that is designed to operate in all different kinds of environments. It is build with a internal compressor, an air/oxygen mixer that when attached to an external oxygen source, gives the system an oxygen range of 21 to 100 %. The ventilator is controlled by an internal microprocessor, which will continuously monitor and display pressure, flow, control settings, alarm parameters and power signals. The ventilator has three different modes ACV, SIMV and CPAP, all of which can be used together with PEEP and SIGH. It can be operated with internal batteries or with the universal AC power supply in place, which allows a wide range of power supplies to be used (90-256 V and 47-440 Hz).

A knob that actually is designed as an analog potentiometer, controls the ventilation rate. It sets the ventilation rate to any value between 1 and 150, except for the CPAP (Rate = 0). The chosen ventilation rate is shown in the LCD display. Tidal volume will also be controlled by an analog potentiometer knob. It ranges from 10 to 3000 ml, and the maximum flow is approximately 1000 ml/sec. Set and delivered tidal volume are shown in the LCD display, just above the knob.

The Impact ventilator is equipped with a standard 9-pin, female, RS-232C serial communication port. This port permits remote monitoring of ventilator function and capabilities. For this study the communication port was not used, instead the ventilator functions are monitored by the CO_2SMO box. By measuring the ventilator function closer to the patient, systemic error is smaller. The version of Uni-Vent^{TM} used in this study does not support controlling ventilator setting over the serial port. Therefore a different approach for controlling the ventilator is used. The analog potentiometers that are used to set tidal volume and ventilation rate are replaced with two digital potentiometers. These digital potentiometers have the same total resistance as the one used originally by the Uni-Vent^{TM}. They are controlled from the computer over the parallel port. In the future, as the ventilator supports direct serial port communication to the micro controller inside the ventilator, an easier strictly digital communication can be performed.

Digital potentiometer The potentiometers used are a dual digital potentiometer model Dallas DS1267-10 consist of two digital controlled solid-state potentiometers. The range is 0 to 10 kΩ and is equal to that of the original potentiometers. They have a 8-bit resolution (256 position) which is the same as the ventilator uses. Supply voltage is +5 V. The two 8-bit values and a stack select bit are read to a 17-bit I/O shift register. When transaction has been completed, the contents of the I/O shift register are loaded into respective multiplexers for setting the wiper. More information about communication protocol can be found on Dallas Semiconductors homepage.

Figure 2.4: This is the mechanical test lung used to test the systems performance. The resistance and compliance can be individually set for each of the two "lungs". This gives a good possibility to simulate different lung pathologies.

2.1.4 Mechanical test lung

For simulating and testing the feedback controller, a mechanical test lung is used. It is a "two-lung" device with individual compliance setting for each side. Attaching different plug-in resistors simulates pulmonary resistance. The gas delivery hose is attached to the test lung trachea, via the CO_2/flow sensor that is connected to the CO_2SOM. The test lung simulates only the lung mechanics of an adult. Different settings for the resistance and compliance are used to simulate different physiological cases.

2.2 Software

A program for controlling the entire system was developed in C++. It has been built using Borland C++ Builder 3, which presents a powerful C++ Window development environment for development of desktop applications. The graphical routines that builder 3 supports has been used to develop a window interface that presents all necessary data and information to the user. For serial port communication, Builder 3 tools are used. For data collection

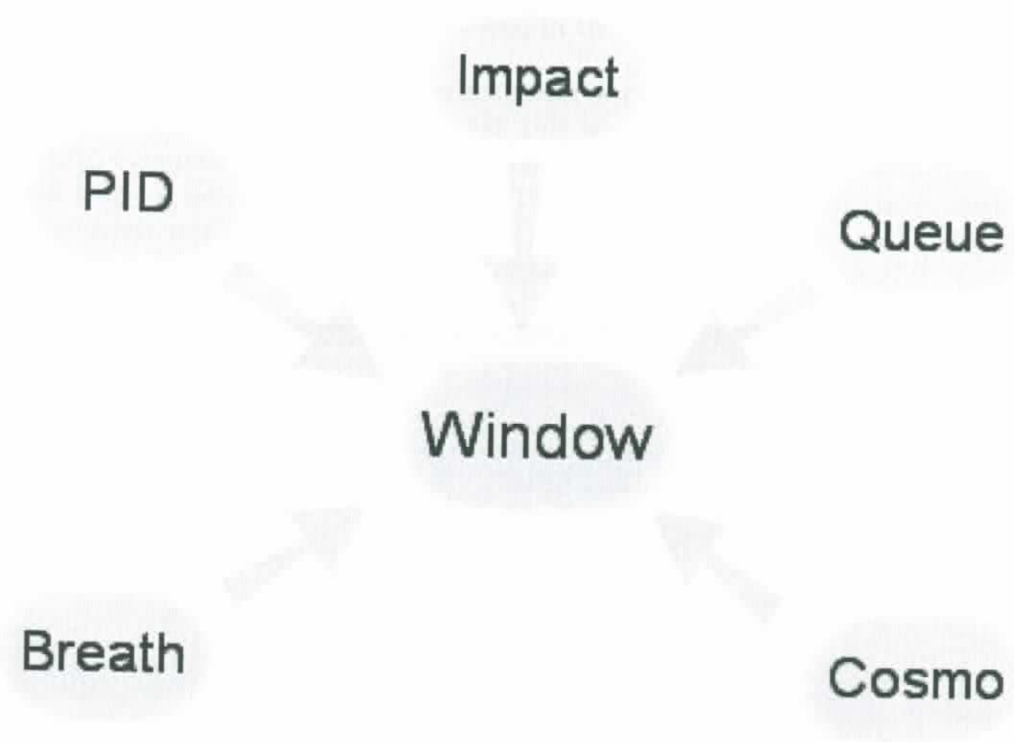

Figure 2.5: The control program, build with C++ Builder, contains six different classes, each with a specific task. The center of the program, is class Window, which controls the other five classes.

and processing, normal C++ functions have been written.

The program, as it looks now, contains six classes. One of them, the Window class, creates and controls the graphical interface. It also instantiates five objects, each of which will regard to one of the five remaining classes. These objects inherit parts of the class name into its own name. The Window object created from the Window class, also controls all events in the program through the EventHandler. This function is running continuously whenever the program is active. The Cosmo and Breath classes are being used to communicate with the CO_2SMO box, where the Cosmo object sends commands and the Breath object parses the data received from CO_2SMO. Class Impact contains functions to set values of tidal volume and respiratory rate to the ventilator and class PID calculates these values. The sixth class, called Queue, is only used as a support class. This class contains routines to create a circular queue, which is being used in the program to store data. Several of the classes include routines for writing data to an Excel file, an opportunity that is used for testing the system. The program is built strictly object oriented; consequently only the Window object sees the other objects. The other objects are completely free and only uses standard C++ libraries. This makes it possible to reuse the classes without any limits. No global variables have been used. A more thorough explanation of what the classes do, how they are created and being used are discussed in the following section.

Class Window This class contains the graphical interface, which can be seen in figure 2.6. When the program has been compiled, this window is created and active for the user to give commands and read data. Events such as clicking on a button or marking text box are taken care by the object TWindow. When the user starts the program by clicking on Start button, a function called EventHandler is called. This function is active as long as the user does not either quit the program or stop it.

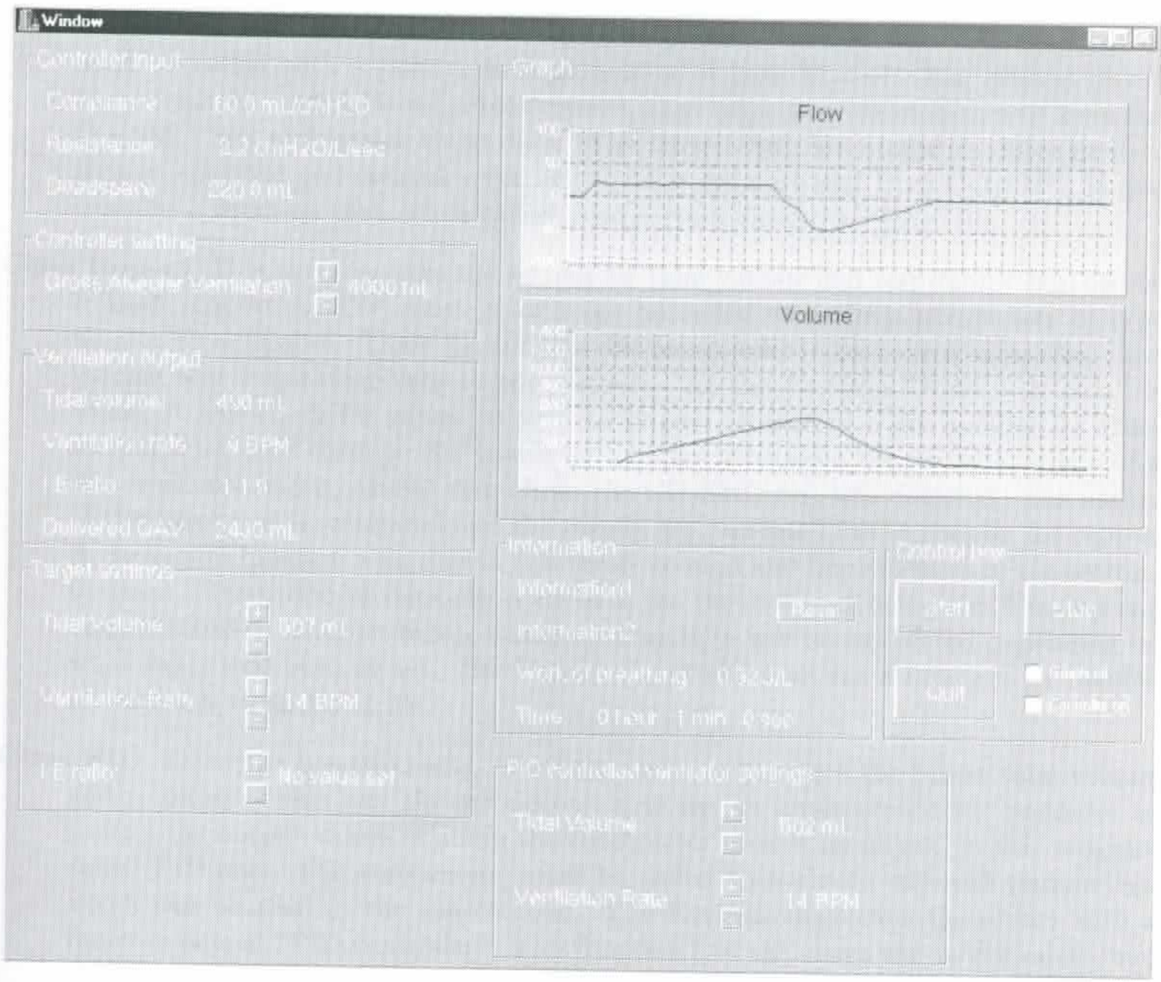

Figure 2.6: This picture shows the interface that the user sees, as the program is running. It contains several different information boxes, that gives the user all necessary information organized in a logic order.

Class Cosmo This class contains several functions to set up the communication protocol for the CO_2SMO box. It also takes care of sending information to CO_2SMO. Calling the constructor will create a Cosmo object, in this program called "cosmo_one". The constructor must be called with pointers to two parameters, or more accurately two objects, the first of type TLabel (an Builder 3 type) and the second of type TApdComPort (a Builder type that refers the serial port). The Cosmo object has two important public functions that, when called, will tell the CO_2SMO box either to start or stop sending data. What data that will be sent to CO_2SMO is being defined within these functions. The Cosmo object uses the serial port for the communication proceeder.

Class breath This class basically contains one function for parsing and storing the information sent from the CO_2SMO box. When the constructor has been called, the constructor creates a object, in this program called "breath_one". The constructor must be called with a pointer to a parameter of type TLabel. The created object has a function, called ParseCosmoProtocol, that takes a string of data sent from the CO_2SMO box and parses these data. The parsed data are stored in different data strings, specified and created according to what data are sent from CO_2SMO. It also contains a function that, when called, read certain data to an Excel file.

Class Impact This class controls the setting for tidal volume and respiratory rate on the Impact ventilator. The constructor must be called with three parameters of type pointer to a TLabel. These parameters are being used to plot error messages, tidal volume and respiratory rate to the window. As the function "Send_to_impact" is called, it will check for values to send to Impact, construct a bit number for these settings and send them to the ventilator. An error between the set tidal volume for the ventilator and measured value from the CO_2SMO box was detected. The nature of this error was investigated and the results are presented in the chapter for results. A correction function was, due to this error, created and implemented in the sending function. Transition of data are done over the parallel port using the function call outportb(0x378, outport-byte), where outport_byte will be calculated depending on what value that is to be set. The Impact object also contains a function for reading information to an Excel file.

Class PID This class has two tasks. The first one is to calculate the target tidal volume and respiratory rate and the second one is to use an implemented PI[1] regulator to control the output values. Calling the constructor creates an object, in this program called PID_one. The constructor must be called with eleven different parameters, which can be read in the source code. The EventHandler calls the object with a function named "PID_controller". This function first calculates the target values, and then implement these values to the PI controller. The controller writes the output values to two parameters that later will be sent to the ventilator. The class also include routines for writing data to an Excel file.

[1]Proportional Integral

Class Queue This class contains functions for building and handling a smart storing structure called circular queue. An queue object is created with a void call to the constructor. In this program one circular queue called queue_one is used.

2.3 Description of the AVV controller

The proposed AVV ventilator controller is based on a closed-loop controller approach first described by Laubscher [8] called Adaptive Lung Ventilation. In contrast to this controller, the AVV controller uses the volume-controlled mode available on the Impact ventilator.

The AVV is based on a model for gas exchange that is shown in picture 2-7. In this model the gross alveolar ventilation (V'_{gA}) is defined as the total tidal volume (V_T) minus the dead space (V_{dS}) multiplied with the respiration rate (RR).

$$V'_{gA} = RR_{tot} * (V_T - V_{dS}) \tag{1}$$

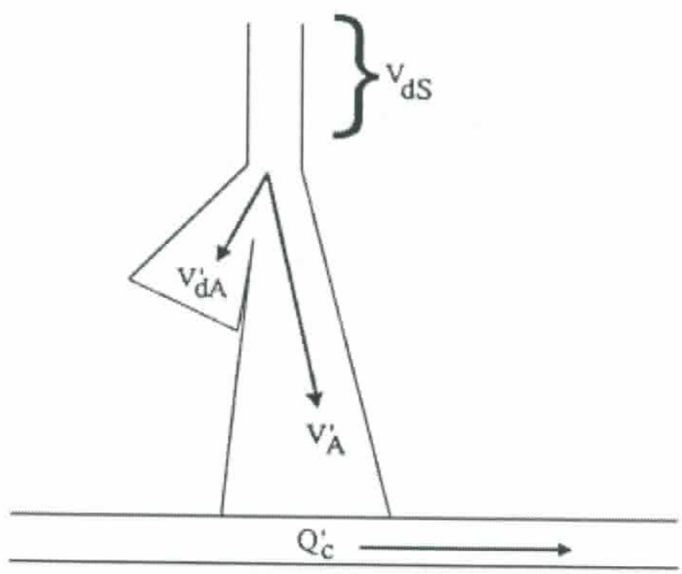

Figure 2.7: Model for gas exchange. V'_{dA} is alveolar dead space ventilation, V_{dS} is series dead space, V'_A is alveolar ventilation and Q'_c is pulmonary capillary blood flow.

The AVV consists of three parts:

I	A respiratory rate / tidal volume selector
II	A respiratory rate controller
III	A tidal volume controller

The user enters a desired V'_{gA} and the rate/volume selector determines the optimal respiratory rate (RR_{tot}^{SP}) and tidal volume (V_T^{SP}). The optimization criterion used, is minimal work of breathing (WOB) as defined by Mead [4] and Otis [2]. In order to calculate the work of breathing, Otis [2] assumed that the flow pattern is a sine wave and that all of the work is done during inspiration. This resulted in the following expression for the rate of inspiration work:

$$\dot{W} = \frac{1}{2}Kf(\dot{V}_A/f + V_D)^2 + \frac{1}{4}K'\pi^2(\dot{V}_A + fV_D)^2 + \frac{2}{3}K''\pi^2(\dot{V}_A + fV_D)^3 \qquad (2)$$

Where:

$\dot{W}$ = rate of work

$\dot{V}_A$ = minute alveolar ventilation

V_D = dead space volume

f = frequency of breathing

K = total respiratory elastance

K' and K'' = first and second order component of flow resistance

By differentiating this equation with respect to f, setting it to zero and solving it for $\dot{V}_A$ the following condition for minimal work of breathing was presented:

$$\dot{V}_A = \frac{KV_Df + K'\pi^2V_Df^2 + 4K''\pi^2V_D^2f^3}{K - aK''\pi^2V_Df^2} \qquad (3)$$

To be able to obtain a solution in terms of frequency, a first approximation of the flow resistance, with one single constant, was made. That gave the following expression:

$$\dot{V}_A = \frac{f(KV_D + R\pi^2V_Df)}{K} \qquad (4)$$

Where R is the linear approximation of the flow resistance. Solving the quadratic equation for f and replacing the elastance with its inverse (K = 1/C) yield the following expression for the minimal work frequency:

$$f_w = \frac{\sqrt{1 + 4\pi^2RC\dot{V}_A/V_D} - 1}{2\pi^2RC} \qquad (5)$$

This requires the knowledge of the alveolar ventilation (V_A) and physiological dead space (V_D). In the absence of disease, the physiological dead space (V_D) mainly consists of anatomical dead space (V_{dS}). The anatomical dead space, contrary to physiological dead space, can be measured from the flow and CO_2 signal. Therefore V_D is replaced by V_{dS} and consequently the alveolar ventilation (V'_A) must be replaced by the gross alveolar ventilation (V'_{gA}). This results in the following equation:

$$RR^{SP}_{tot} = 30\frac{\sqrt{1+4\pi^2 RCV_{gA}/V_{dS}}-1}{\pi^2 RC} \tag{6}$$

In this formula, RC is in seconds, V'_{gA} in ml/min and V_{dS} in ml. RR^{SP}_{tot} is given in breaths per minute. Figure 2.8 shows the graphical representation of how the optimal respiratory rate depends on RC, V_{gA} and V_{dS}. The rate/volume selector chooses the target RR^{SP}_{tot} for any given gross alveolar ventilation. Thereafter the corresponding target tidal volume is calculated from the following equation:

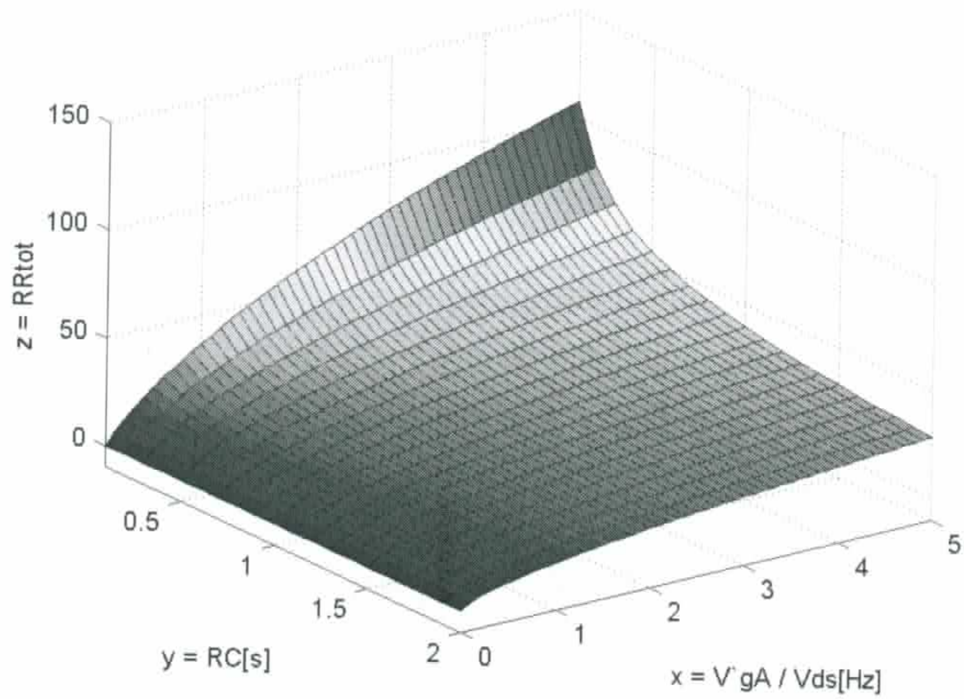

Figure 2.8: Graphic representation of the dependency of respiratory rate from time constant RC and V_{gA}/V_{dS} ratio as used by the Rate/V_T selector in the AVV controller. For most individuals, respiratory rate lies between 10 and 20 breaths per minute.

$$V^{SP}_T = V_{dS} + \frac{V_{gA}}{RR^{SP}_{tot}} \tag{7}$$

The calculated target values for respiratory rate and tidal volume, is delivered to the two implemented PI controllers. These are two standard discrete PI controllers, implemented using Euler's backwards method. They are build using a normal continuous PI-regulator:

$$u(t) = K_p e(t) + K_I \int_{t_0}^{t} e(\tau) d\tau \tag{8}$$

where

K_P and K_I = adjustable gain factors

u = control signal

$e(t) = r(t) - y(t)$, error

r = target value

y = input signal

This equation can be rewritten as

$$u(t) = K \left(e(t) + \frac{1}{T_I} \int_{t_0}^{t} e(\tau) d\tau \right) \tag{9}$$

where $T_I = \frac{K}{K_I}$.

To transform this PI controller, we extend the integral to a series using euler backward:

$$\int_{t_0}^{t} e(\tau) d\tau \approx T(e(t) + e(t-T) + e(t-2T) + ...+) \tag{10}$$

where T equals the sample interval.

This will give us the final form for the discrete PI controller:

$$u_n = K \left(e_n + \frac{T}{T_I} \sum_{k=0}^{n} e_k \right) \tag{11}$$

Both discrete PI controllers are implemented in the class PID. The first version of the AVV controller, which was implemented in a software-based ventilator/patient, contained a rule-based state machine. This state machine has not been included in the tested system.

Chapter 3

Results

The specific aim for this study is to show the feasibility of using the design approach proposed to control the Impact ventilator. The implemented feedback controller will, when implemented into the LSTAT project, simplify the ventilator operation by reducing the number of settings that the operator must enter. This makes it possible to use the ventilator even when qualified personal are not available. It has also been the goal to measure different controller parameters, to be able to compare this system with commercial available closed-loop control systems for ventilators. This study also tries to verify that the used optimization criterion for work of breathing reaches the goal and that the implemented system handles distractions in a good way.

The experimental set-up used, is described in chapter 2-1. A modified computer-controllable Impact Uni-Vent ventilator connected to a mechanical test lung, a respiratory profile monitor and a desktop were used for the testing. The respiratory profile, including resistance and compliance, were measured between the Y piece[1] and the endotracheal tube, using a combined CO_2/flow sensor. The ventilator was set to work in a assist/control mode and used external air to ventilate the patient.

The study showed that the system is, in general, well behaved. The design approach seem to be appropriate for the Impact ventilator. Measured feedback control parameters show that the system is fast and exact. The optimization criterion was verified to work appropriate.

3.1 System accuracy

The object of the first study was to investigate how accurately the set target values for the Impact ventilator correspond to those measured with the CO_2SMO. It is important that the values calculated by the computer and those set on the ventilator are accurate, otherwise the significance for especially the work of breathing study will be limited, because the work

[1] The Y piece divide the endotracheal tube to the both lungs, simulate the bronchi in the normal breathing system

of breathing is directly dependent on the minute ventilation[2]. To verify that the values given by CO_2SMO were correct, it was calibrated by simulating a breath of exactly 1000ml. The results showed that CO_2SMO read the tidal volume, with less than 2% error.

The first test was designed to test the tidal volume and respiratory rate separately. The set values were arranged to match a fix minute ventilation (range 2000 - 12000 ml/min). The respiratory rate was set at three different values 8, 12 and 16 breaths per minute and the tidal volume were calculated from these values. For each specific setting of tidal volume and respiratory rate, the outcome from the ventilator was measured for two minutes and analyzed with three parameters:

Arithmetic mean value

$$\mu = \frac{\sum_j X_j}{j} \tag{1}$$

Variance

$$\sigma^2 = \frac{\sum_j (X_j - \mu)^2}{j} \tag{2}$$

Correction coefficient

$$K_{corr} = \frac{SetValue}{DeliveredValue} \tag{3}$$

The result from this test is showed in table 1, appendix A1. A graphical presentation of the correction value for tidal volume, which actually describes the relative error, is shown in figure 3-1.

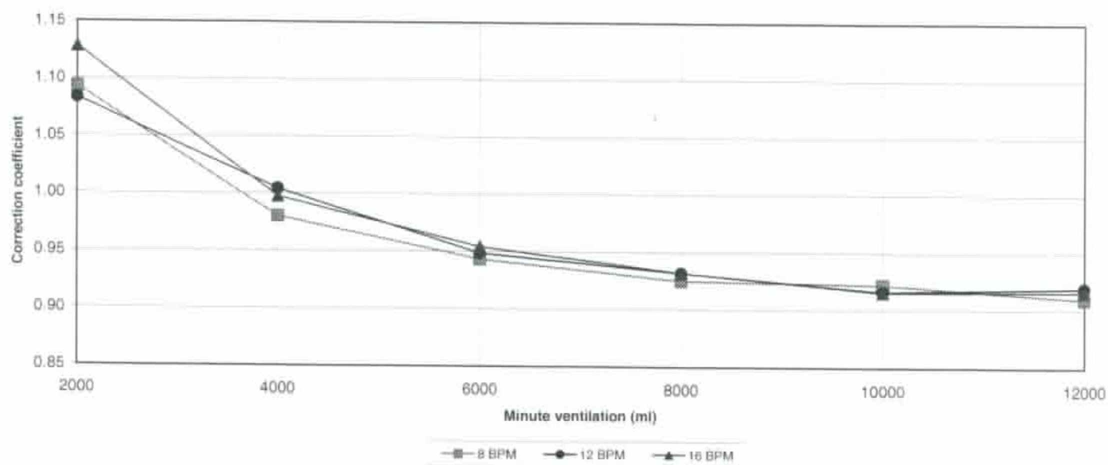

Figure 3.1: The graph shows correction value for tidal volume verses minute ventilation. This test was performed without any correction function included.

[2]Minute ventilation is equal to respiration rate multiplied with tidal volume

According to the results in table 1, the set values for respiration rate correspond very exactly to those measured with CO_2SMO. This can be explained by the fact that the ventilator actually delivers the ventilation rate that is set. We also have a problem with the resolution because the CO_2SMO box and the Impact ventilator only deals with integers for ventilation rate. In figure 3-1 the correction value for the delivered tidal volume is plotted versus minute ventilation. The result shows that the ventilator delivers very accurate tidal volume for minute ventilation of 4000 ml; however there is a problem for minute ventilation above 5000 ml and below 3000 ml. The error is not linear. To deal with this problem, a correction function was developed by interpolating a fourth order polynomial through the data points, using Matlab^{TM}.

$$K_{corr} = 1.32 - 0.154V_{MV} + 0.024V_{MV}^2 - 0.0017V_{MV}^3 + 0.000048V_{MV}^4 \quad (4)$$

As this correction function was applied to the system, a new series of testing was performed. The results are shown in table 2, appendix A1, and in figure 3-2.

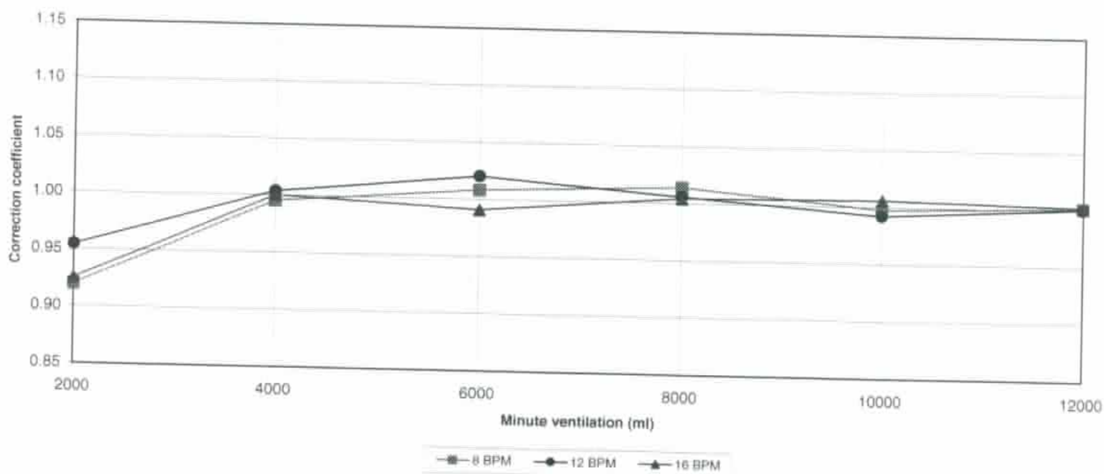

Figure 3.2: The graph shows correction value for tidal volume versus minute ventilation. This test was performed, using the correction function described above.

As shown in figure 3-2, the system performs significantly better with the correction function in place. For values around 4000 ml of minute ventilation, the system works very accurately. However there is still a relative big error at low minute ventilation. This can be explained by the small absolute value for the tidal volume, which will render to a bigger relative system error. However the significance for these values are small, because the dead space exceeds the set tidal volume for these cases and therefore these measurements are not physiological useful.

To verify system performance when the controller was used, the accuracy for set GAV[3] was tested. The ventilator was set to deliver nine different values over the range 2000-10000

[3]gross alveolar ventilation

ml. When the system had stabilized at the defined target GAV, a two minute measurement was recorded. The data was analyzed with the same parameters as for tidal volume and respiratory rate, as seen in Table 3.1.

Set GAV	Delivered GAV		
	Mean value	Std. Dev.	Error %
2000	2182	24.3	-9.10%
3000	3277	52.3	-9.23%
4000	4322	198.1	-8.05%
5000	5279	158.9	-5.58%
6000	6420	141.2	-7.00%
7000	7410	167.8	-5.86%
8000	8852	221.1	-10.65%
9000	10557	226.7	-17.30%
10000	12580	287.4	-25.80%

Table 3.1: This is the results, that was given as the AVV controller was set to deliver different values for the V_{gA}. The results shows an acceptable performance for range 2000-7000 ml V_{gA}.

The results shows that the delivered GAV is within 10% of the set value for GAV below 7000 ml/min. Higher values for GAV will usually not be used. The error for the feedback control system, according to the results, is larger than if tidal volume and respiration rate were measured separately. This can be explained by the fact that the controller tries to use floating values for the respiration rate and not only integers. Therefore the respiration rate will be higher than the controller's target value.

3.2 Speed and overshoot for different step responses

To measure and study the dynamics of the feedback control system, it is common to introduce a step change in one of the control variables and study how the system behaves. In this study three different step responses has been implemented and tested. These steps are GAV, resistance/compliance and resistance. For each step, set and delivered tidal volume and respiratory rate has been measured for 4 minutes. The data has been evaluated with three different constants:

Overshoot

$$Y_0 = \frac{Y_{max} - Y_{mean}}{Y_{mean}} \tag{1}$$

Rise time

$$T_{90\%} = T_{90\%} - T_{0\%} \tag{2}$$

Δ_{max}

$$\Delta_{max} = \frac{Max_{Tstart:Tstop} - Min_{Tstart:Tstop}}{Y_{mean}} \tag{3}$$

Delta max measures the amplitude of the oscillation of the system. All the tests have been done using normal values for resistance and compliance, that means resistance 2 $cmH_2O/l/sec$ and compliance 60 ml/cmH_2O. Also, the ventilator was set to adjust the I:E-ratio automatically.

3.2.1 V_{gA} step response

The V_{gA} step was implemented by an instant change from 4000 ml/min to 8000 ml/min. This cause an instant change of target value for tidal volume and respiratory rate, which can be observed in the pictures.

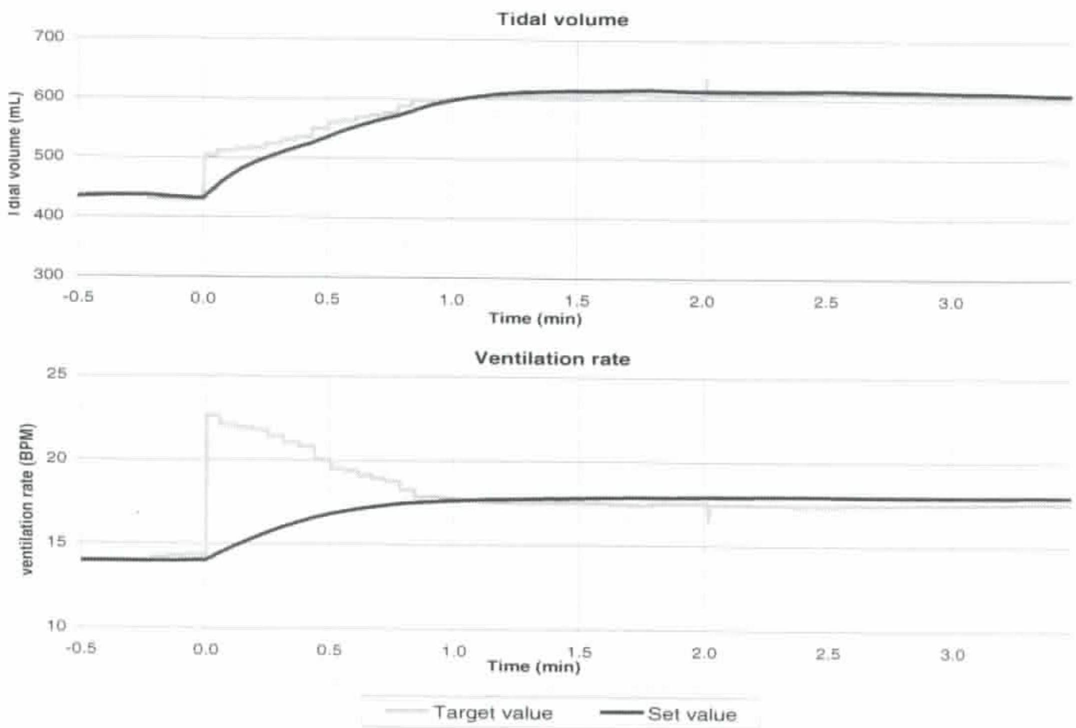

Figure 3.3: V_{gA} step response

	Tidal vol.	Resp. rate
Overshoot:	1.42%	2.75%
Rise time:	0.9 min	0.6 min
Delta-max:	1.19%	0.64%

Table 3.2: Result table

3.2.2 Resistance/Compliance step response

Step change for resistance/compliance was implemented by a sudden change from ventilating two lungs to just ventilating one lung. This resulted for a change in the resistance from 2 to 7 $cmH_2O/l/sec$ and a change for the compliance from 140 to 70 ml/cmH_2O.

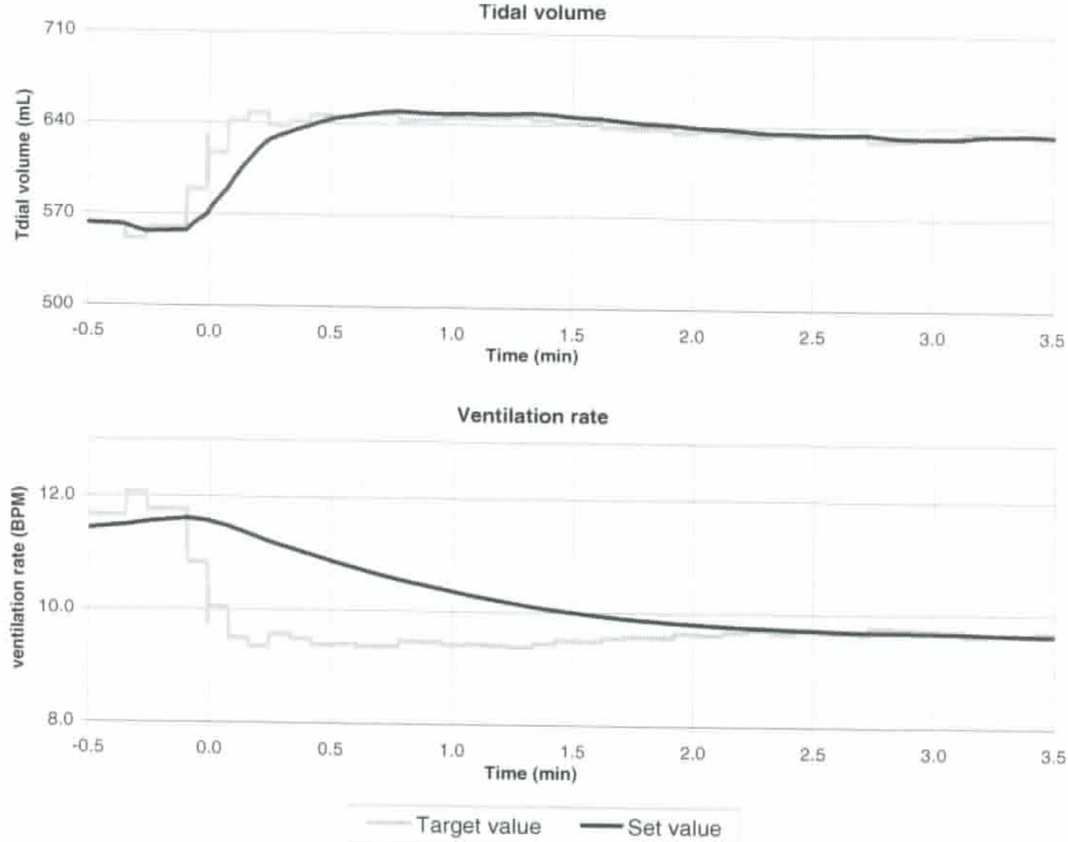

Figure 3.4: Resistance/Compliance step response

	Tidal vol.	Resp. rate
Overshoot:	0.46%	1.43%
Rise Time:	1.8 min	0.3 min
Delta-max:	0.86%	2.77%

Table 3.3: Resistance/Compliance step response, table

3.2.3 Resistance step response

The resistance step was implemented by changing the resistor used in the system to simulate airway resistance. The airway resistance changed from 2.5 to 7 $cmH_2O/l/sec$ between two breaths.

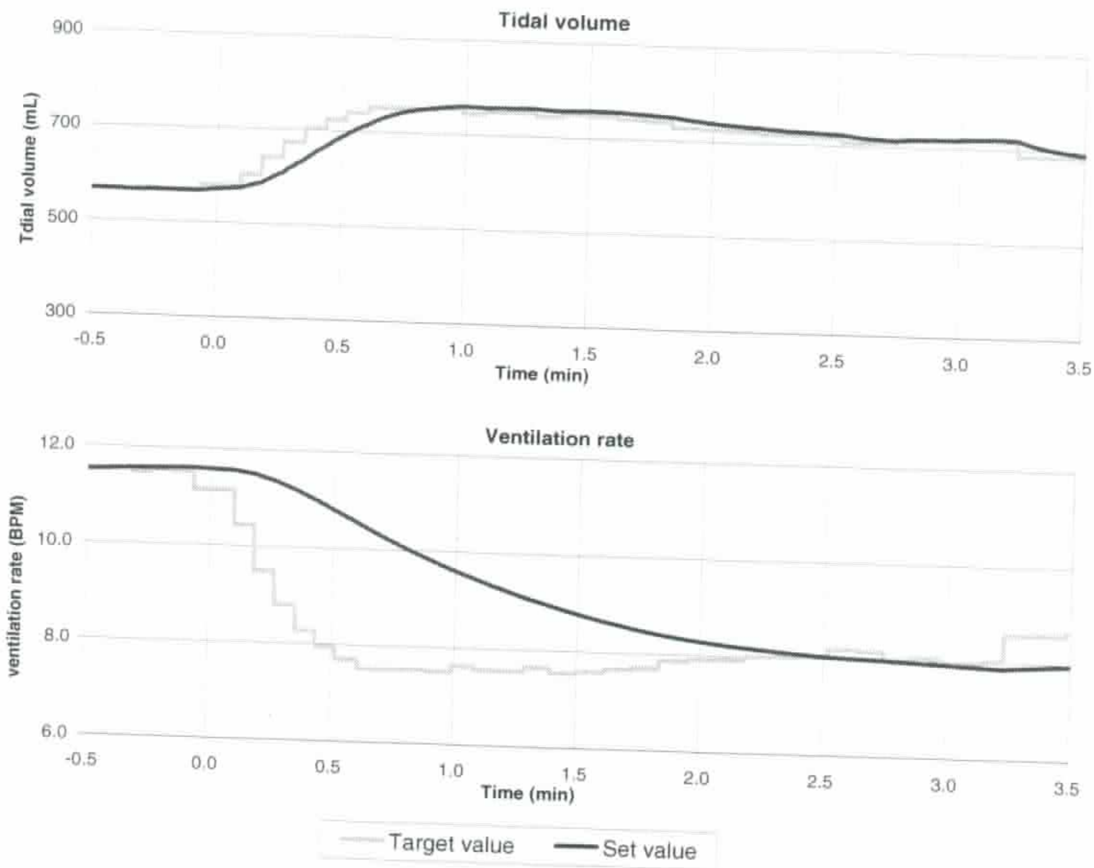

Figure 3.5: Resistance step response

	Tidal vol.	Resp. rate
Overshoot:	4.39%	-1.16%
Rise Time:	0.5 min	1.4 min
Delta-max:	8.99%	2.02%

Table 3.4: Resistance step response, table

3.3 Work of breathing

To control the Impact ventilator, an algorithm for minimizing the work of breathing is being used. This algorithm takes into account the resistance, compliance and dead space. From these parameters and the given gross alveolar ventilation, target tidal volume and respiratory rate is calculated. The goal for this test was to investigate if the controller minimizes the work of breathing, when it is being used together with the Impact ventilator. To do this, a global search for the work of breathing minimum point was performed. The work of breathing is correlated to the volume that the ventilator delivers to the patient, and therefore the GAV must be kept constant during this search. The test was performed for five different values of the GAV, from 2000 to 10000 ml in steps of 2000 ml. For each value of GAV a set of five values for the respiratory rate and tidal volume was chosen. This choice was based on an assumption for the optimal work of breathing and the values were arranged around this optimal point. The ventilator was set to work with each setting for two minutes, and then the mean value for the work of breathing was calculated. The work of breathing delivered by the ventilator was measured with the CO_2SMO box. It delivers the work of breathing for each breath, given in J/l/breath. In this study however, the minimized value is work of breathing in Joules performed by the ventilator. Therefore the measured work of breathing must be multiplied with the minute ventilation. The value is also defined to measure for one minute

$$WOB_J = WOB_{J/l} * RR_{tot} * V_T * 1min \tag{1}$$

From these values, a second order work of breathing function was interpolated. This function describes the work of breathing performed by the ventilator, and therefore the work imposed to the patient lungs. To verify that the ventilator minimize the work of breathing, the AVV feedback controller was set to ventilate the test lung with the specified GAV for two minutes. The recorded value for work of breathing and the target settings of tidal volume and respiratory rate were compared to the function given by the global search. To decrease the error within the study, all of the measured values for work of breathing were normalized to the set GAV. This was done by comparing the set and delivered GAV according to this formula:

$$WOB_{norm} = WOB * \frac{GAV_{set}}{GAV_{del}} \tag{2}$$

The results showed that the ventilator in most cases hit the target minimum point of the function.

The algorithm was tested on the test lung with three different resistance/compliance combinations:

I Normal lung, simulated by normal settings of the resistance/compliance.

II Airway obstruction, simulated by an increased resistance and normal compliance.

III ARDS[4], simulated by a normal resistance and decreased compliance.

3.3.1 Normal lung

The setting used for this test was resistance at 2.0 $cmH_2O/l/sec$ and the compliance at 60 ml/cmH_2O. The resistance was accomplished by using one R5 linear resistor. For compliance, 30 ml/cmH_2O was set to each side of the lung. The Ventilator was using the assist/control mode and the I:E-ratio was set to a fixed value 1:2. Dead space was added by means of a plastic tube and approximated to 220 ml. Picture 3.6 shows the diagram presenting the results for work of breathing study under normal conditions. The results are also shown in table 2-1, appendix A2.

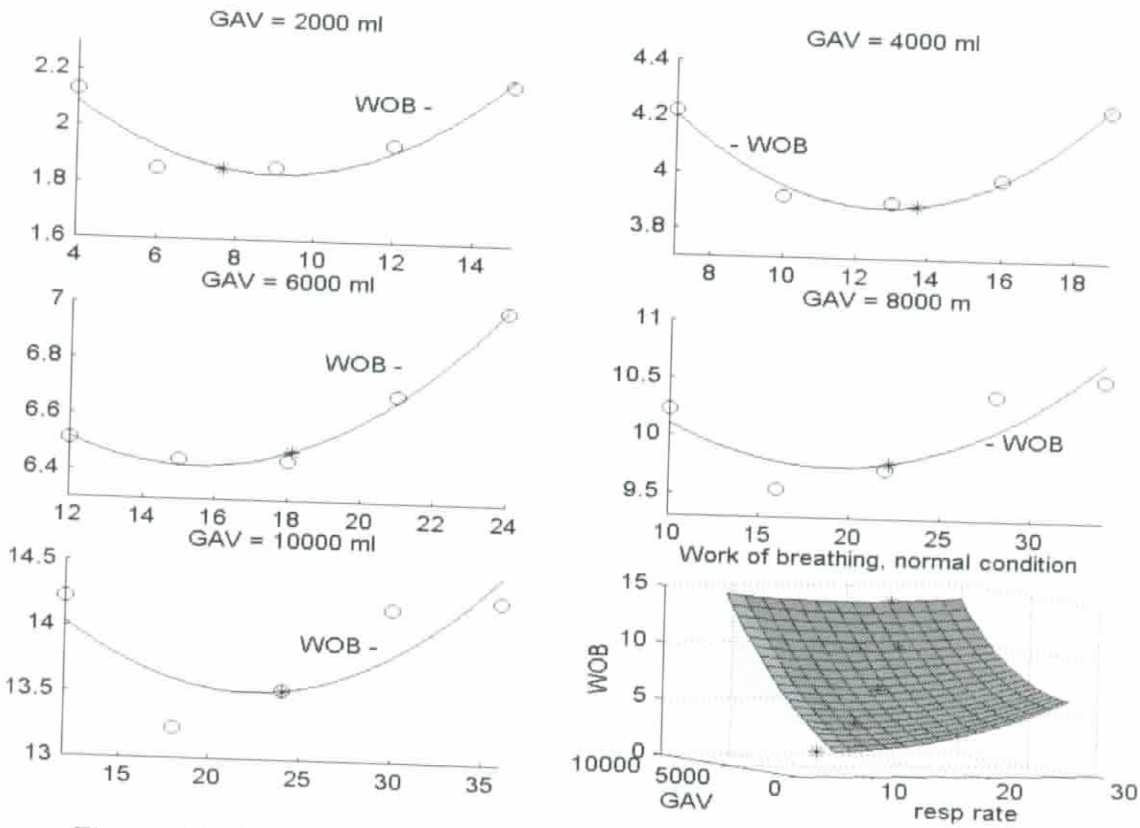

Figure 3.6: Work of breathing measured during normal condition. Circles indicates the measured values for global search of the minimal work of breathing and the star shows the value choose by the controller. The line gives the interpolated function for minimal work of breathing. The unit for the Y-axis are work of breathing and for the x-axis respiration rate

[4]Acute Respiratory Distress Syndrome

3.3.2 Airway obstruction

The setting used for this test was resistance at 27.0 $cmH_2O/l/sec$ and the compliance at 60 ml/cmH_2O. The resistance was accomplished by using one R50 linear resistor. For compliance, 30 ml/cmH_2O was set to each side of the lung. The Ventilator was using the assist/control mode and the I:E-ratio was set to a fixed value 1:2. Dead space was added by means of a plastic tube and approximated to 220 ml. Picture 3.7 shows the diagram presenting the results for work of breathing study under normal conditions. The results are also shown in table 2-2, appendix A2.

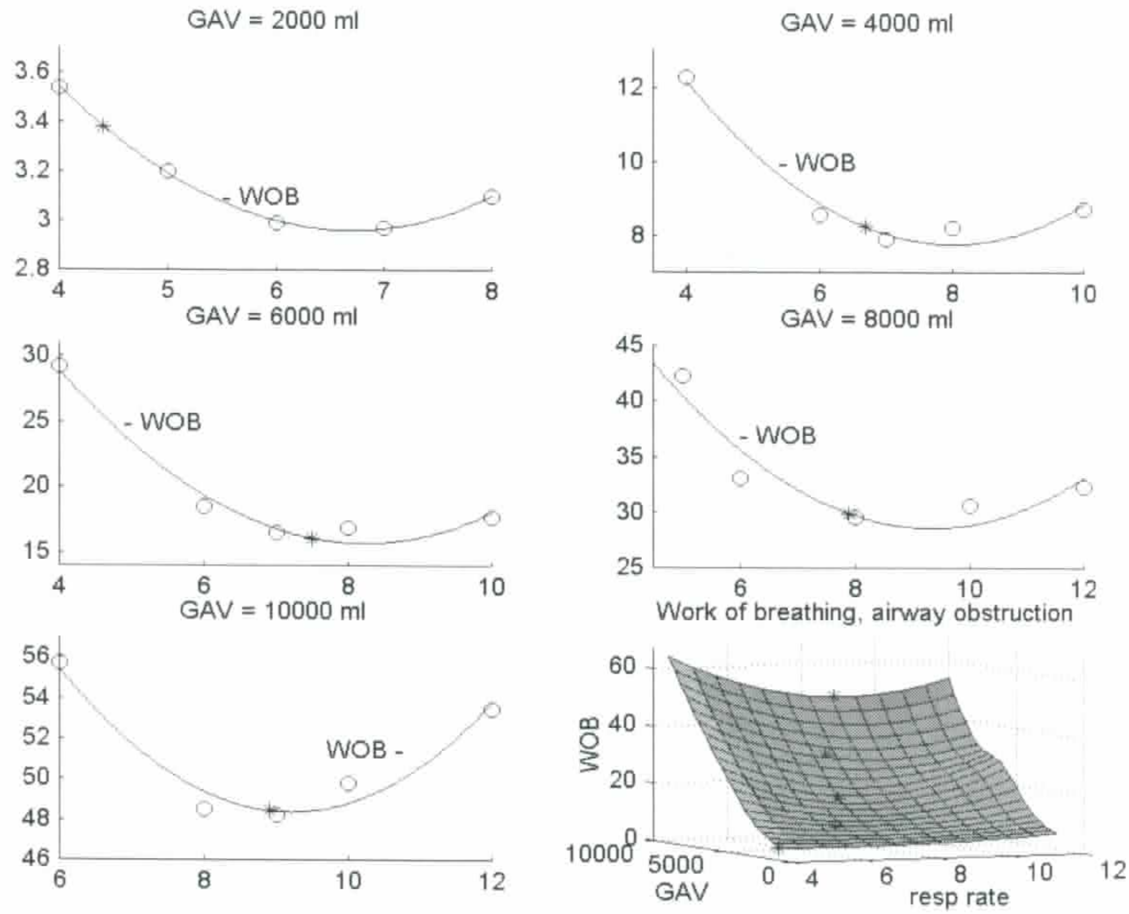

Figure 3.7: Work of breathing measured with airway obstruction. Circles indicates the measured values for global search of the minimal work of breathing and the star shows the value choose by the controller. The line gives the interpolated function for minimal work of breathing. The unit for the Y-axis are work of breathing and for the x-axis respiration rate

3.3.3 ARDS

The setting used for this test was resistance at 2.0 $cmH_2O/l/sec$ and the compliance at 20 ml/cmH_2O. The resistance was accomplished by using one R5 linear resistor. For compliance, 10 ml/cmH_2O was set to each side of the lung. The Ventilator was using the assist/control mode and the I:E-ratio was set to a fixed value 1:2. Dead space was added by means of a plastic tube and approximated to 220 ml. Picture 3.8 shows the diagram presenting the results for work of breathing study under normal conditions. The results are also shown in table 2-3, appendix A2.

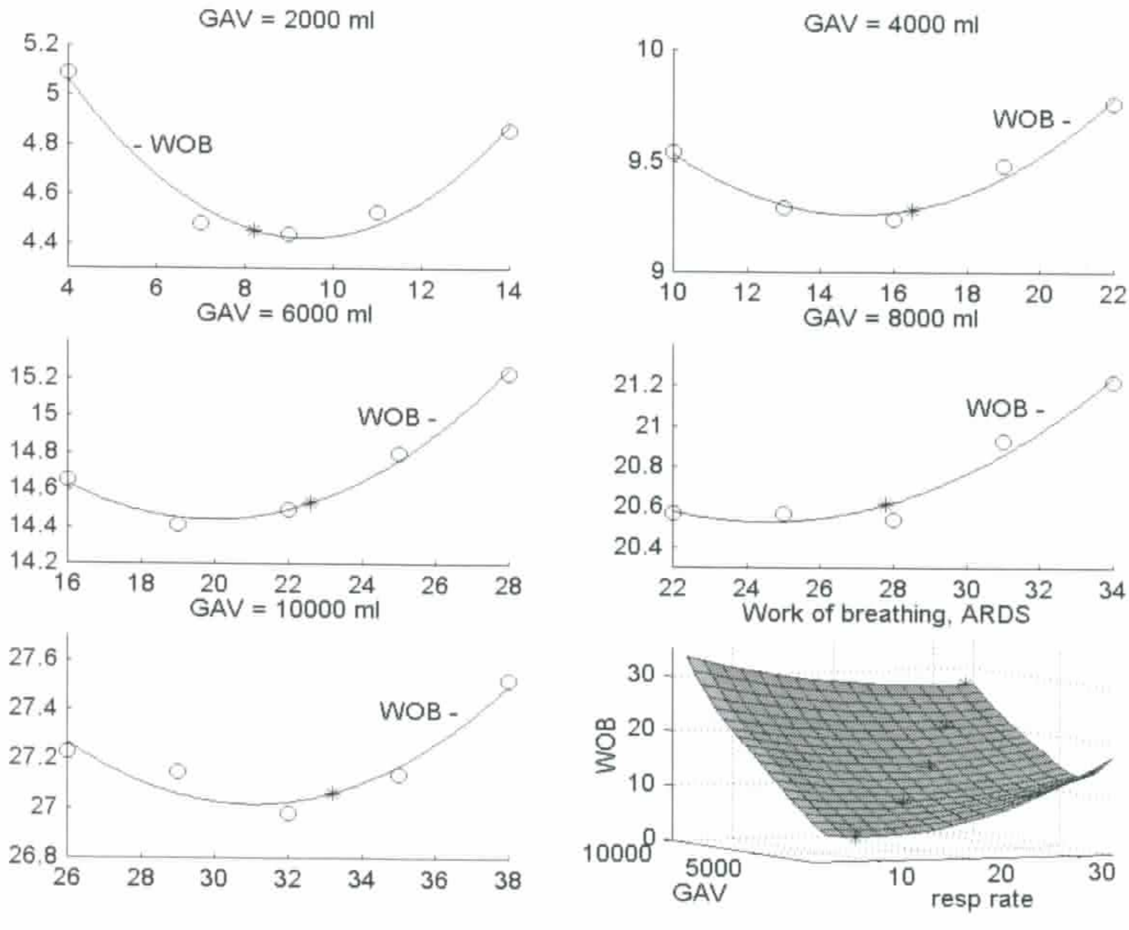

Figure 3.8: Work of breathing measured with ARDS. Circles indicates the measured values for global search of the minimal work of breathing and the star shows the value choose by the controller. The line gives the interpolated function for minimal work of breathing. The unit for the Y-axis are work of breathing and for the x-axis respiration rate

3.4 Stability and robustness

For all closed-loop control systems, stability and robustness are crucial. For medical applications, safety issues are even more important. Therefore studies of the system behavior during exceptional conditions is a necessary part in the development of medical feedback control systems. In this study, this part has been divided into two sections, the first investigating the stability and the second robustness.

3.4.1 System stability

The stability describes how well a closed-loop system behaves, when the input parameters gets out of the normal range defined for the system . In other words, if the input signals becomes very big or very small, will the system still be stable? If the system is to sensitive, that means not enough damped, it might start to oscillate. The system used in this study, is overall very well damped, that means very slow. Therefore no oscillating is expected.

To test the stability, a big change in the two input signals resistance and compliance were implemented. This was done with normal values for the resistance (5 $cmH_2/L/sec$) and compliance (80 ml/cmH_2O), except for when these values was changed. The gross alveolar ventilation was set to 4000 ml. Compliance was changed in two steps, where the new set values were defined by the range of the test lung. The resistance was also changed in two steps. In the first step the R5 resistor was changed to a resistor without any extra resistance. In the second test, a R500 resistor was implemented. No change in the dead space was implemented, and the reason for this is that the dead space can not be measured with the used test set-up

Change in compliance As the compliance was changed to a lower value, no problems occurred. The controller slowly, but steadily takes the ventilator to the target settings. When the compliance is being changed towards higher values, the CO_2SMO read range was exceeded for values above 200 ml/cmH_2O. As this happens the system will read $C = 0$, and give a warning signal, that tells the user that the RC range is exceeded. Thereafter the controller will not change the target tidal volume or respiratory rate, until a good C value has been read. In neither of the cases did the system start to oscillate. It showed to be stable, just as expected.

Change in the resistance When the resistance was changed towards zero, nothing strange happened. The system slowly reached the target values. However problems occurred as the resistance was set at a very high level. To be able to deliver the set tidal volume and respiratory rate, the ventilator needed a very high inspiratory pressure. The peak inspiratory pressure exceeded the ventilator range and a safety valve was released several times during one inspiration. This had the effect that the target tidal volume was never reached. As consequent the CO_2SMO box was not able to read the values for resistance and compliance. Therefore, once again, the controller system was shut off and a warning signal for RC out of range was signaled.

3.4.2 System robustness

The robustness of a closed-loop control system can be described as how good a system can handle systemic errors that can occur. In this study, such error could be problems with the air gas hose or data connection failure. It is important that the system can detect these errors and make the appropriate corrections.

To test the robustness, simulated errors were implemented to the system. The reaction of the system was studied for several minutes, and continuous data was collected. Three different errors were implemented:

I	Air gas hose compressed. Simulating something compressing the air gas hose, for instance if someone would, by accident, step onto the hose.
II	Computer - ventilator disconnected. Simulate a system failure caused by disconnecting the parallel communication cable.
III	Computer - CO_2SMO disconnected. Simulate a system failure where the serial port communication between CO_2SMO and computer are interrupted.

Air gas hose compressed For this test, the system was first set to ventilate the patient for two minutes. There after the hose was compressed for three minutes. Finally the system was set to work under normal conditions for one minute. The overall impression was, that the system was not affected by the extra work, that the ventilator had to perform. When the output signals were studied after the test, they showed no sign of the occurred error. Therefor it can be concluded that the system it not affected by a air gas hose compression.

Computer - ventilator disconnected For this test, the ventilator was set to ventilate the test lung under normal conditions. During this time, the communication port between computer and ventilator was disconnected. This had the effect, that both the digital potentiometers used to control the ventilator were given their default values, that is 10 kΩ. The ventilator was therefore given its minimal values for tidal volume and respiratory rate. If the system was used to ventilate a patient, this would result in a life threatening situation, because no air would be delivered. This problem is not possible to solve with the present configuration. This will be discussed more thorough in chapter 4. When the cable between computer and ventilator was connected again, the system recovered fast and started to ventilate the test lung.

Computer - CO_2SMO disconnected This test was performed like the experiment above, however the connection between computer and CO_2SMO was disconnected. As a result all information update taken from the CO_2SMO was terminated. This did not effect the ventilator, which continued to ventilate the test lung. However the closed-loop control system did not work during this time. As the cable was connected again, the system got back on line without any problems.

Chapter 4

Discussion

4.1 Summary

The specific aim for this study was to try out the behaviour of the AVV controller together with the Impact ventilator used in the LSTAT project. To do this, the Impact ventilator was implemented into a system, together with a respiratory profile monitor and a computer. The performance was tested onto a mechanical test lung. Control mechanism for the ventilator was implemented by using two digital potentiometers, which was directly digital controlled by the computer using a parallel port. Several different parameters, including step response and stability/robustness, were measured to study the system performance. The results showed that the control method used works good together with the Impact ventilator.

For control systems like these, it is important to be able to control the output variables exactly. The implemented controller uses two parameters to adjust the patient minute ventilation, respiration rate and tidal volume. Respiration rate was set quite accurately, with only smaller differences when the controller was used. In most cases the set and delivered respiration rate corresponded exact. The error for tidal volume was in general much higher throughout the entire study. The behaviour of the error was investigated by measuring the difference between set and delivered tidal volume, for several different occasions. By assuming that these data presents a good picture of the behaviour of the error, a correction function was implemented. A more thorough investigation of the nature of the error was never performed. However it was noted that the error changed its characteristic as the minute ventilation exceeded a certain level. Therefore it can be assumed that the error has its base in an internal ventilator problem, that the ventilator can not measure the delivered tidal volume accurately enough. To be able to have a big ventilation range, probably two different flow meters are used. The step change between these flow sensors, will implement a step change in the accuracy of the measurements. However it has not been the objective for this study to investigate the behaviour of the Impact ventilator. As shown in chapter three, this error was between five and ten percent for delivered GAV in a normal range of minute ventilation. This error can be explained by two limitations of the system. First the inaccuracy for tidal volume and second a problem with the discretetion of the respiration rate. The controller calculates with floating values for respiration rate, however only integers can be set to the ventilator. By changing the set tidal volume on the ventilator several times per minute, a mean value that is close to the floating value of the controller can be reached. This however will introduce one more error to the system, an error that can be seen as the controller is being used. In a future development of the system, when it is possible to control the ventilator directly with digital signals, it might be interesting to investigate the possibility to set the values only in discreet steps.

The performance of the system was tested by measuring several step responses and the result area shown in chapter 3.2. In general the system showed a good behaviour, with a fast response and no overshoot. The speed is for obvious reasons very important to avoid underventilation of the patient, something that might cause severe damage to the body. The first tissue to take damage, would be the brain, which can withstand underventilation for some three minutes. Therefore it is important that the controller has a rise time that do not exceed this limit. The system implemented to this ventilator showed under no conditions a bigger rise time than two minutes. Under normal conditions and for a step change of 50 % for ventilation volume, the rise time will be just one minute. Also big changes in the lung physiology would cause the controller to make a fast response, never having a rise time that exceeds two minutes. The system would therefore respond fast enough to most changes that might occur. A fast system will in most cases generate a big overshoot. This could cause severe damage to the patient, because of the barotraumas that the lung tissue would be exposed to during the time of over ventilation. Therefore it is important to adjust the speed so that the overshoot will be close to minimum. For normal conditions and a step in the set gross alveolar ventilation, the system showed to have no or a very small overshoot. Also the oscillations (measured with the delta max parameter) is very small for this measurement. For changes in the lung physiology, the overshoot and oscillations are bigger. This can be explained with the more unstable values for resistance and compliance. In no case are the overshoot bigger than 5 % and the oscillation are smaller than 2 %. It is therefore not likely that the system would be dangerous for a patient, regarding overshoot.

As we breathe, the body performs a work that can be measured in Joules and will be defined as work of breathing. Work of breathing will describe a second degree polynomial function with one minimum. This minimum point will be a breathing frequency, approximately where the body normally breath. This can be tested by breathing very fast or very slow, and notice that this will be harder than with a normal breathing frequency. The human body are therefore designed to use a breathing frequency that minimize the work of breathing. In this study, a equation that minimize the work of breathing, are used to select the respiration rate and tidal volume. This is believed to be the most natural way of breathing. It was therefore also important to see if we actually reach this point. The results given in chapter 3.3 shows that the controller reaches the goal. With normal conditions, the controller hits the bottom very nicely. With a airway obstruction, the controller chooses a little bit too low breathing frequency and for ARDS a little bit too high frequency. However it is not significantly different from the optimal point.

When it comes to feedback control systems for medical technology, it is obvious that the safety issues are very important. The system must be very robust and not fail to perform task or push the patient into a hazardous position in case of some problems due to chanical or human errors. The system build for this study showed to be satisfyingly ble and robust, however the handling of unexpected situations can be better. this will discussed more thorough in the section for future studies.

2 Limitations

e specific aim for this master's thesis has been to construct, build and verify the permance for a closed loop control system using the adaptive volume ventilation technique. en though the results are satisfying, there are limitations in this study, limitations that l be corrected in future studies.

One of the major limitations of this study, is of course the lack of patient studies. The tem has been tested on a mechanical test lung where the resistance and compliance adjustable. The study shows a good and safe behaviour and this makes it possible to ceed into another phase where the system will be tested on living creature. Such a study combination with this study, will present very strong results that shows the feasibility of s design approach for building a closed loop control system for mechanical ventilation. e reason that no patient study was included into this master's thesis is simply lack of ne. A lot of effort was put down into building the control and test system, such as the tware, setting up communication channels and adjusting system parameters. A patient dy has been started, and will be discussed more in chapter 4.3.

The system also have some limitations, of which the control mechanism of the ventilator ght be the biggest. It was constructed using two digital potentiometers, directly soldered to the ventilator microprocessor board. The reason for using this approach is the limited xibility that the ventilator has been constructed with. Because the ventilator is completely ntrolled by the micro controller, it would have been easier if a digital signal could have en sent directly to it from the computer. This is not possible for this model of the Impact ntilator, a feature that will be build in to later models. The ventilator is today equipped th a serial port. As a new software platform has been developed, it will be a pretty easy sk to directly set the target values to the ventilator. This will increase the accuracy of e system significantly, because the ventilator can be directly controlled and the problems th the digital potentiometers disappear.

.3 Future

his paper has been aimed to investigate the implementation of the AVV controller to a rmal ventilator during ventilation of a mechanical test lung. The results given earlier in is chapter, shows that the AVV controller works fine and that the method can be used a clinical testing. The next step will be to try out how well the used algorithm works the OR. Such a study should be aimed to show how well the respiration rate and tidal olume settings would work on a patient. It could also be interesting to see if there is a fference between the values that the controller suggests and those given by the clinician nd if there is, who would suggest the best settings out of a work of breathing stand point. uch a study are for the moment being performed at the University of Utah.

Another important improvement of the system would be to develop a direct control f the ventilator settings over a serial port. This would make it possible to build a more able and accurate system. It would also raise the question of how the ventilator settings re chosen. It might be better to limit the controller to just calculate and set integers or respiration rate and tidal volume, instead of the floating variables used today. Such a ystem has to be developed, tested and compared, to see what would be the optimal way o do it.

The impact ventilator today comes with knobs for controlling the level of oxygen and he I:E-ratio. It would be good to let the computer control theses values as well, however his propose some additional work. Especially the I:E ratio, must be controlled with the ontroller directly, something that involves the respiration rate. The tidal volume, respiraon rate and I:E ratio together defines the flow used during inspiration. Therefore in will e a kind of complicated to define a algorithm for specifying the I:E-ratio. One possibility ould be to use a state machine that check out what tidal volume and respiration rate that re chosen by the controller and thereafter adjust the I:E ratio. This would also make it ossible to build in additional safety features, by just allowing certain values for the different arameters in each state. Such a state machine was developed in the first study involving he AVV controller, however never implemented to this controller.

Appendix A

Result tables

A.1 Accuracy study

	Set value		Delivered mean value		Standard deviation		Corr. mean value	
MV	Tidal vol	Resp rate	Tidal vol	Resp rate	Tidal vol	Resp rate	Tidal vol	Resp rate
2000	250	8	229	8	2.26	0.00	1.0933	1.0000
2000	170	12	157	12	1.93	0.00	1.0834	1.0000
2000	130	16	115	16	2.58	0.00	1.1283	1.0000
4000	500	8	510	8	2.34	0.00	0.9803	1.0000
4000	330	12	329	12	2.81	0.00	1.0042	1.0000
4000	250	16	251	16	3.03	0.00	0.9975	1.0000
6000	750	8	795	8	2.23	0.00	0.9433	1.0000
6000	500	12	527	12	2.38	0.00	0.9486	1.0000
6000	380	16	398	16	2.08	0.00	0.9543	1.0000
8000	1000	8	1081	8	2.74	0.00	0.9249	1.0000
8000	660	12	708	12	1.50	0.00	0.9318	1.0000
8000	500	16	536	16	3.26	0.00	0.9320	1.0000
9000	1130	8	1238	8	2.73	0.00	0.9125	1.0000
9000	750	12	817	12	2.85	0.00	0.9184	1.0000
9000	570	16	619	16	3.04	0.00	0.9211	1.0000
10000	1250	8	1356	8	1.47	0.00	0.9220	1.0000
10000	830	12	906	12	3.29	0.00	0.9164	1.0000
10000	630	16	688	16	2.72	0.00	0.9158	1.0000
12000	1500	8	1649	8	1.80	0.00	0.9097	1.0000
12000	1000	12	1092	12	5.24	0.00	0.9195	1.0000
12000	750	16	819	16	3.81	0.00	0.9164	1.0000

Table A.1: Accuracy study for tidal volume and respiratory rate for uncorrected system

MV	Set value Tidal vol	Set value Resp rate	Delivered mean value Tidal vol	Delivered mean value Resp rate	Standard deviation Tidal vol	Standard deviation Resp rate	Corr. mean value Tidal vol	Corr. mean value Resp rate
2000	250	8	272	8	2.61	0.00	0.9197	1.0000
2000	170	12	178	12	1.88	0.00	0.9543	1.0000
2000	130	16	141	16	1.80	0.00	0.9248	1.0000
4000	500	8	502	8	5.39	0.00	0.9955	1.0000
4000	330	12	329	12	2.81	0.00	1.0037	1.0000
4000	250	16	251	16	4.15	0.00	1.0002	1.0000
6000	750	8	744	8	1.37	0.00	1.0082	1.0000
6000	500	12	490	12	3.01	0.00	1.0203	1.0000
6000	380	16	384	16	3.03	0.00	0.9909	1.0000
8000	1000	8	986	8	1.97	0.00	1.0142	1.0000
8000	660	12	656	12	2.86	0.00	1.0058	1.0000
8000	500	16	498	16	2.92	0.00	1.0044	1.0000
10000	1250	8	1253	8	2.71	0.00	0.9976	1.0000
10000	830	12	836	12	2.17	0.00	0.9923	1.0000
10000	630	16	626	16	3.00	0.00	1.0059	1.0000
12000	1500	8	1498	8	2.44	0.00	1.0012	1.0000
12000	1000	12	1000	12	2.81	0.00	1.0003	1.0000
12000	750	16	748	16	2.56	0.00	1.0021	1.0000

Table A.2: Accuracy study for tidal volume and respiratory rate for corrected system

A.2 Work of breathing

	Set mean value		Measured mean value				Delivered WOB			Delivered GAV		Flow	
GAV	Tidal vol	Resp rate	Tidal vol	Resp rate	R	C	Mean	S. Dev.	Calibrate	Mean	S. Dev.	Hi	Lo
2000	350.0	15	353.2	15	1.89	57.9	2.17	0.04	2.17	1998	59	18	-34
2000	390.0	12	396.5	12	2.08	58.6	2.07	0.03	1.95	2118	43	16	-36
2000	440.0	9	448.5	9	2.22	58.3	1.91	0.02	1.86	2056	24	16	-40
2000	550.0	6	574.0	6	2.66	58.8	1.97	0.04	1.85	2124	42	15	-48
2000	720.0	4	795.0	4	3.09	61.0	2.45	0.02	2.13	2303	18	19	-60
AVV	480.6	7.7	493.6	8.0	2.45	59.0	2.00	0.03	1.83	2189	35	14	-43
Mean	-	-	-	-	2.39	58.9	2.11	0.03	1.99	2120	37	17	-44
4000	430.0	19	432.2	19	2.24	61.9	4.27	0.08	4.24	4032	88	29	-39
4000	470.0	16	469.3	16	2.33	60.6	3.98	0.04	3.99	3989	50	25	-41
4000	530.0	13	532.2	13	2.50	60.5	3.96	0.04	3.90	4059	48	24	-45
4000	620.0	10	634.8	10	2.74	61.5	4.06	0.04	3.92	4148	46	20	-52
4000	790.0	7	812.4	7	3.26	61.3	4.37	0.03	4.22	4147	24	19	-62
AVV	513.9	13.7	519.1	14.0	2.48	60.6	4.12	0.11	3.94	4187	107	26	-44
Mean	-	-	-	-	2.61	61.2	4.13	0.05	4.05	4075	51	23	-48
6000	470.0	24	466.6	24	2.51	64	6.89	0.13	6.99	5918	125	39	-41
6000	510.0	21	509.5	21	2.63	63.1	6.77	0.10	6.68	6079	92	38	-44
6000	550.0	18	553.1	18	2.73	62.9	6.44	0.08	6.44	5996	70	35	-47
6000	620.0	15	635.2	15	2.86	62.8	6.68	0.09	6.44	6228	79	30	-52
6000	720.0	12	737.4	12	3.08	63.3	6.74	0.11	6.51	6209	80	29	-57
AVV	551.7	18.1	548.5	19.0	2.71	63.0	6.85	0.10	6.59	6241	98	36	-46
Mean	-	-	-	-	2.76	63.2	6.70	0.10	6.61	6086	89	34	-48
8000	460.0	34	461.9	34	2.88	66.8	10.71	0.35	10.53	8135	231	50	-42
8000	510.0	28	518.1	28	2.92	66.7	10.83	0.21	10.38	8345	195	48	-45
8000	580.0	22	578.6	22	2.90	64.7	9.59	0.13	9.72	7890	125	45	-49
8000	720.0	16	724.0	16	3.15	63.0	9.62	0.10	9.54	8064	72	41	-57
8000	1020.0	10	1017.9	10	3.63	63.5	10.20	0.12	10.23	7979	70	37	-71
AVV	592.9	22.2	609.0	22.6	3.06	66.1	11.18	0.66	10.16	8801	389	44	-52
Mean	-	-	-	-	3.10	64.9	10.19	0.18	10.09	8083	139	44	-53
10000	500.0	36	503.6	36	3.35	74.1	14.35	0.72	14.24	10076	470	58	-45
10000	550.0	30	556.2	30	3.23	66.4	14.29	0.24	14.17	10086	206	52	-47
10000	640.0	24	641.0	24	3.23	66.4	13.66	0.26	13.53	10097	209	49	-53
10000	780.0	18	775.0	18	3.39	63.9	13.22	0.16	13.23	9991	112	44	-61
10000	1050.0	12	1062.0	12	3.85	63.2	14.37	0.09	14.22	10105	61	44	-75
AVV	636.2	24	690.5	23.8	3.44	66.3	16.02	0.68	14.29	11214	372	53	-56
Mean	-	-	-	-	3.41	66.8	13.98	0.29	13.88	10071	212	49	-56

Table A.3: Work of breathing study during normal condition

	Set mean value		Measured mean value				Delivered WOB			Delivered GAV		Flow	
GAV	Tidal vol	Resp rate	Tidal vol	Resp rate	R	C	Mean	S. Dev.	Calibrate	Mean	S. Dev.	Hi	Lo
2000	470.0	8	476.7	8	18.17	56.7	3.18	0.04	3.10	2054	24	11	-18
2000	510.0	7	511.5	7	18.44	58.7	3.03	0.04	2.97	2040	24	11	-19
2000	550.0	6	559.5	6	19.31	59.7	3.05	0.05	2.99	2037	32	11	-20
2000	620.0	5	625.9	5	21.03	59.8	3.25	0.05	3.20	2030	19	13	-21
2000	720.0	4	714.5	4	22.61	58.4	3.50	0.02	3.54	1978	3	15	-23
AVV	662.8	4.4	716.9	5	22.65	59.8	4.41	0.17	3.55	2484	63	15	-23
Mean	-	-	-	-	19.91	58.7	3.20	0.04	3.16	2028	20	12	-20
4000	620.0	10	623.5	10.0	23.30	67.5	8.79	0.15	8.71	4035	36	19	-21
4000	720.0	8	727.2	8.0	23.87	62.5	8.32	0.06	8.20	4057	14	18	-23
4000	790.0	7	786.4	7.0	24.08	61.2	7.83	0.04	7.90	3965	15	17	-24
4000	890.0	6	893.0	6.0	24.35	62.8	8.65	0.08	8.57	4038	25	18	-26
4000	1220.0	4	1223.0	4.0	32.37	67.8	12.31	0.02	12.27	4012	11	25	-30
AVV	814.1	6.7	827.5	7.0	24.62	63.3	8.79	0.28	8.27	4252	82	18	-25
Mean	-	-	-	-	25.59	64.4	9.18	0.07	9.13	4021	20	19	-25
6000	820.0	10	823.7	10.0	28.89	77.6	17.81	0.11	17.70	6037	14	25	-25
6000	970.0	8	977.3	8.0	29.92	72.7	17.04	0.18	16.88	6058	38	24	-27
6000	1080.0	7	1068.7	7.0	30.40	71.2	16.36	0.16	16.52	5941	28	23	-29
6000	1220.0	6	1224.2	6.0	32.89	67.1	18.56	0.07	18.48	6025	12	25	-30
6000	1720.0	4	1723.9	4.0	41.54	73.7	29.27	0.04	29.19	6016	8	35	-36
AVV	1038.6	7.5	1038.6	8.0	30.78	72.1	19.81	0.79	18.15	6549	133	26	-28
Mean	-	-	-	-	32.73	72.5	19.81	0.11	19.76	6015	20	26	-29
8000	890.0	12	887.3	12.0	34.30	79.2	32.34	0.13	32.31	8008	31	33	-26
8000	1020.0	10	1015.9	10.0	34.44	83.7	30.41	0.25	30.57	7959	37	31	-28
8000	1220.0	8	1219.1	8.0	34.87	73.1	29.50	0.11	29.53	7993	19	30	-30
8000	1550.0	6	1547.2	6.0	38.23	72.6	32.91	0.16	33.06	7963	25	31	-34
8000	1820.0	5	1820.9	5.0	43.76	78.7	42.29	0.18	42.26	8005	6	37	-37
AVV	1220.6	7.9	1228.7	8.0	35.05	80.2	30.01	1.55	29.75	8069	199	31	-31
Mean	-	-	-	-	37.12	77.5	33.49	0.17	33.55	7986	24	32	-31
10000	1050.0	12	1058.6	12.0	41.55	78.1	53.78	0.34	53.44	10063	32	39	-30
10000	1220.0	10	1225.5	10.0	41.01	75.8	50.08	0.12	49.81	10055	15	37	-31
10000	1330.0	9	1325.8	8.0	40.42	66.6	47.99	0.15	48.22	9953	17	36	-32
10000	1470.0	8	1469.0	6.0	40.90	74.3	48.48	0.10	48.52	9992	14	36	-34
10000	1890.0	6	1888.7	6.0	45.19	66.9	55.78	0.03	55.71	10012	5	38	-38
AVV	1349.9	8.9	1359.8	9.0	41.48	70.0	51.56	3.16	50.26	10258	290	38	-33
Mean	-	-	-	-	41.81	72.3	51.22	0.15	51.15	10015	17	37	-33

Table A.4: Work of breathing study during obstructive condition

	Set mean value		Measured mean value				Delivered WOB			Delivered GAV		Flow	
GAV	Tidal vol	Resp rate	Tidal vol	Resp rate	R	C	Mean	S. Dev.	Calibrate	Mean	S. Dev.	Hi	Lo
2000	360.0	14	360.9	14	2.40	21.5	4.79	0.08	4.86	1972	55	19	-65
2000	400.0	11	403.6	11	2.59	21.5	4.57	0.05	4.53	2019	49	16	-70
2000	440.0	9	431.5	9	2.93	21.6	4.22	0.05	4.44	1903	31	15	-73
2000	510.0	7	512.9	7	2.98	21.6	4.59	0.07	4.48	2050	39	13	-74
2000	720.0	4	725.1	4	4.06	21.8	5.14	0.03	5.09	2020	18	16	-107
AVV	458.9	8.2	455.8	9.0	2.89	21.6	4.73	0.06	4.46	2123	48	14	-77
Mean	-	-	-	-	2.99	21.6	4.66	0.06	4.68	1993	38	16	-78
4000	400.0	22	403.9	22	2.36	21.4	9.87	0.19	9.76	4046	103	31	-71
4000	430.0	19	432.2	19	2.51	21.6	9.55	0.19	9.48	4031	110	30	-75
4000	470.0	16	467.4	16	2.68	21.6	9.15	0.18	9.24	3959	86	28	-78
4000	530.0	13	533.6	13	2.77	21.7	9.47	0.11	9.29	4077	48	26	-86
4000	620.0	10	617.8	10	3.19	21.8	9.49	0.10	9.54	3978	39	21	-96
AVV	471.2	16.5	463.3	17.0	2.65	21.7	9.56	0.26	9.25	4136	119	27	-78
Mean	-	-	-	-	2.70	21.6	9.51	0.15	9.46	4018	77	27	-81
6000	430.0	28	438.5	28	2.57	22.0	15.53	0.35	15.23	6117	191	43	-76
6000	460.0	25	459.3	25	2.63	21.8	14.76	0.33	14.80	5983	164	37	-79
6000	490.0	22	487.5	22	2.69	21.6	14.21	0.30	14.49	5885	140	38	-81
6000	540.0	19	58.8	19	2.80	21.7	14.55	0.18	14.41	6058	107	36	-87
6000	600.0	16	601.6	16	2.88	21.7	14.91	0.16	14.65	6106	89	35	-95
AVV	485.5	22.6	477.0	23.0	3.65	21.7	14.39	0.36	14.60	5912	169	35	-81
Mean	-	-	-	-	2.71	21.8	14.79	0.26	14.72	6030	138	38	-84
8000	460.0	34	461.1	33	2.69	22.5	21.26	0.55	21.22	8014	238	52	-79
8000	480.0	31	483.7	31	2.72	22.4	21.39	0.43	20.93	8174	196	52	-81
8000	510.0	28	508.2	28	2.83	22.3	20.72	0.46	20.54	8070	206	47	-84
8000	540.0	25	549.7	25	2.93	22.3	20.92	0.51	20.57	8136	218	46	-90
8000	580.0	22	591.4	22	2.99	22.1	21.01	0.40	20.57	8171	148	42	-94
AVV	503.8	27.8	517.2	28.0	2.87	22.4	21.58	0.50	20.75	8321	160	47	-86
Mean	-	-	-	-	2.83	22.3	21.06	0.47	20.77	8113	201	48	-86
10000	480.0	38	487.5	38	2.77	22.8	27.59	0.69	27.52	10027	276	62	-82
10000	510.0	35	508.7	34	2.83	22.3	26.64	0.47	27.14	9816	200	60	-85
10000	530.0	32	532.3	32	2.87	22.6	26.96	0.50	26.98	9993	203	57	-87
10000	560.0	29	573.8	29	2.95	22.4	27.86	0.45	27.15	10260	173	56	-93
10000	600.0	26	611.0	26	3.08	22.6	27.89	0.42	27.23	10243	167	49	-97
AVV	519.0	33.2	555.2	33.0	2.95	22.3	30.68	0.74	27.74	11061	277	61	-90
Mean	-	-	-	-	2.90	22.5	27.39	0.51	27.20	10068	204	57	-89

Table A.5: Work of breathing study during ARDS condition

Bibliography

[1] *Joint tactics, techniques and procedures for patient movement in joint operations*, Joint Pub, 4–02.2.

[2] Otis AB, Fenn WO, and Rahn H, *Mechanics of breathing in man*, J Appl Physiol (1950), vol 2, pp. 592–607.

[3] Chapman FW, Newell JC, and Roy RJ, *A feedback controller for ventilatory therapy*, Ann Biomed Eng (1985), 13:359–372.

[4] Mead J, *Control of respiratory frequency*, J Appl Physiol (1960), 15(3), 325–336.

[5] Ohlson KB, Westenskow DR, and Jordan WS, *A microprocessor based feedback controller for mechanical ventilation*, Ann Biomed Eng (1982), 10:3548.

[6] Frumin MJ and Lee ASJ, *A physiologically oriented artificial respirator which produce* $n_2o - o_2$ *anesthesia in man*, J Lab Clin Med (1957), 49:617.

[7] Coon RL, Zuperku EJ, and Kampine JP, *Systemic arterial blood ph servocontrol of mechanical ventilation*, Anesthesiology (1978), 40:201–204.

[8] Laubscher TP, Heinrichs W, Weiler N, Hartmann G, and Brunner JX, *An adaptive lung ventilation controller*, IEEE Trans on Biomed Eng (1994), 41(1)51–58.