BreathMet : non-obstructive breath volume measurement on horses

van der Zee, Milo G.

Award date:
1997

Eindhoven University of Technology
Faculty: Electrical Engineering
Section: Medical Electrical Engineering

BreathMet

Non-obstructive breath volume measurement on horses

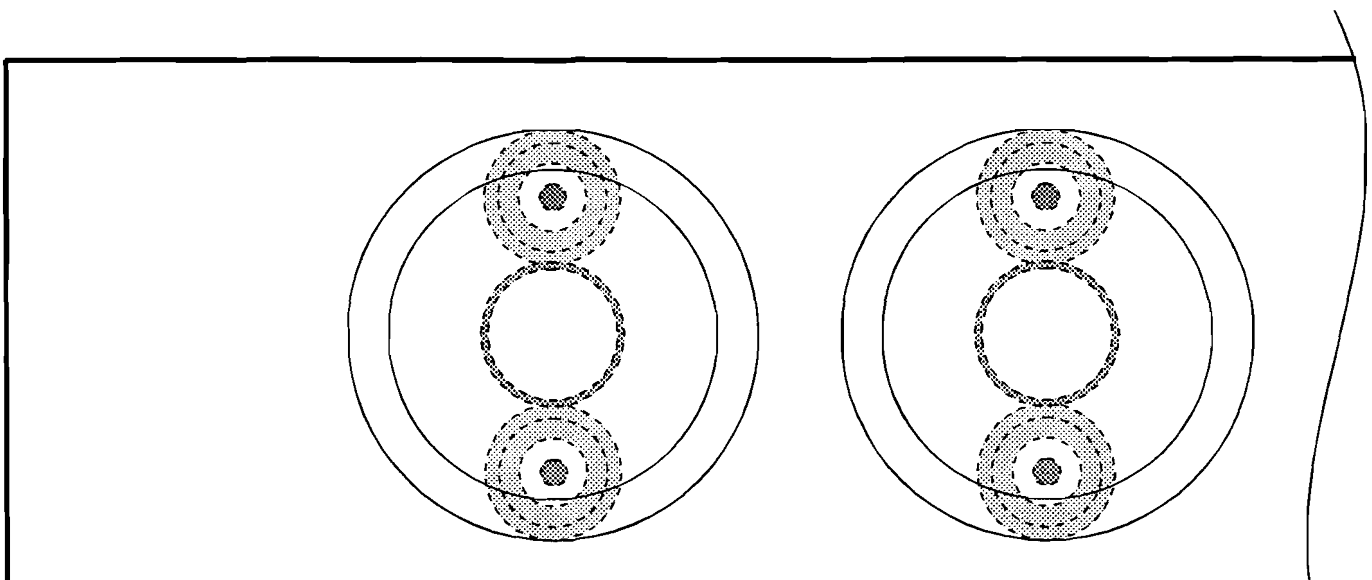
Ing. Milo G. van der Zee

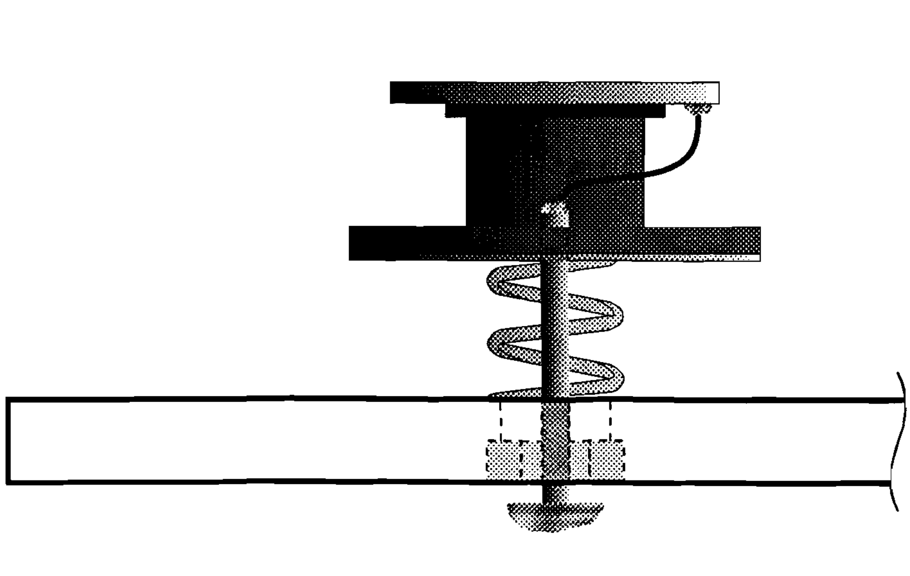
Graduation Report

Period:	*1 September 1996 – 16 October 1997*
Author:	*Ing. Milo G. van der Zee (Stud. Number:387298)* *Kerkstraat 143* *1521 JG WORMERVEER*
Mentor:	*Dr. Henk C. Schamhardt* *University Utrecht, Veterinarian Studies, Functional Morphology* *Yalelaan 1* *3584 CL UTRECHT*
	Dr. ir. P.J.M. Cluitmans / Ir. W. Leliveld *Eindhoven University of Technology, Electrical Engineering, Medical Electrical Engineering* *P.O. Box 513* *5600 MB EINDHOVEN*
Professor:	*Prof. Dr. ir. P.P.J. v/d Bosch*

BreathMet

Non-obstructive breath volume measurement on horses





1. Summary

The breathing volume taken by a horse informs about the condition the horse is in. Under anaesthesia it can be used as an indication of the depth of anaesthesia and during training the breath volume indicates the level of fatigue of the horse. The methods normally used to measure the breath volume make use of a breathing mask. One of the disadvantages of this breathing mask is that it obstructs the airways and so changes the breathing pattern. Besides this, the horse tends to get restless from the equipment put on his nose. These problems with existing methods resulted in a need for a new non-obstructive method for measuring the breath volume taken by the horse.
The method described in this report makes use of 40kHz ultrasound to follow the lung margin of the horse. From the changes in lung-margin location it seems to be possible to follow the changes in breathing volume.
Ultrasound is unable to penetrate through air-tissue interfaces and this complicates the use of ultrasound with horses. The horse has a thick fur that consists of a lot of air. This fur reflects a large amount of the ultrasound sent by the transmitters and results in very noisy measured signals because of the large amplifications needed to still be able to see anything. Because of the large amplifications the equipment is extremely sensitive for movements of the sensors what makes the fixing of the sensors to the skin a very important task. One way to ease the problems is to shave the horse. This makes it possible to reduce the amplification and so decreases the sensitivity to movement of the sensors. Because one of the demands on the equipment was that shaving the horse was not allowed the conclusion is that this method with ultrasound is not usable to measure the breath volume with horses. Because humans do not have a fur like horses the 40kHz ultrasound can be used for other measurements on humans. In one of the tests the beating of the heart is clearly seen on the computer screen.

2. Preface

To graduate at the Eindhoven University of Technology I had to complete a traineeship of at least 9 months. Most of the students complete their period at the university it self, but I wanted to go to an other University because in my opinion, completing my traineeship at an external location adds something extra to the traineeship. I also think it broadens my view to other disciplines in addition to the technical ones learned in Eindhoven. Because I am very interested in medical technology I decided to search for a graduation place in that line of work. My girlfriend studies at the University of Utrecht and mentioned to me that it would probably be a nice idea to complete my traineeship over there at the Faculty of Veterinarian Studies.

A graduating student from Eindhoven gave me the idea for my own project. He designed some equipment to measure the breathing frequency of humans during training. This frequency indicated the level of fatigue. From this my first idea for a project goal was to construct the same equipment to work with horses. During the very first visit to Utrecht it became clear that the breathing frequency is of no use in the case of horses. Instead the volume of breathed air has to be measured, to be able to say anything about the horse's level of fatigue. This was much more difficult than to just measure the frequency but we still wanted to initiate the project.

Without saying anything about details of the project I would like to state that it was much harder to measure the breath volume than expected, but this only made the project more interesting. And because a totally new method was chosen all stages from research and design until testing were gone through. This gave me the chance to show a lot of my disciplines. I had to research a subject I did know nothing about, design and manufacture analogue and digital electronics, write a program in C combined with assembly, conduct tests and at the end I had to write a report in the English language.

Besides the main project a lot of other interesting projects were going on and I often took a look at the testing sides. Also I was granted permission to install a Novell Intranetware server directly connected to the Internet. This gave me the chance to gain a lot of experience with networks and the protocols involved. Because I am very likely going to work in the networking business, this was of much use to my further career.

From all this I personally conclude that it was a very nice and interesting place to complete my traineeship. Thanks to all who helped in making it such a nice place to be. Especially I would like to thank Henk Schamhardt, Wim Leliveld and Nicole Schuttevaêr.

Greetings,

Ing. Milo G. van der Zee

Contents

1. **SUMMARY** 3

2. **PREFACE** 5

3. **INLEIDING** 9

4. **INTRODUCTION** 9

5. **DEMANDS ON THE BREATHMET** 11

6. **ANATOMICAL RESEARCH** 13

6.1. GENERAL ANATOMY OF THE HORSE 13

7. **EXISTING METHODS** 15

7.1. CONSULTED RESOURCES 15
7.2. IMPEDANCE PLETHYSMOGRAPHY 15
7.3. SONIC TRANSILLUMINATION 16
7.4. ULTRASOUND 16
7.5. PROJECTIVE IMAGERY 16
7.6. TOMOGRAPHIC RADIOGRAPHY 16
7.7. OPEN-CIRCUIT AND CLOSED-CIRCUIT METHODS 16
7.7.1. Open circuit Single-breath 17
7.7.2. Open-circuit Multiple-breath 18
7.7.3. Closed-circuit Multiple-breath 20
7.8. BODY PLETHYSMOGRAPHIC 21
7.8.1. Description 1 21
7.8.2. Description 2 22
7.9. THE CAPNOGRAM 23
7.10. ^{133}XE 23
7.11. CALORIMETRIC FLOWMETER 24
7.12. DOPPLER EFFECT FLOWMETER 24
7.13. WRIGHT RESPIROMETER 24
7.14. SPIROMETRY 25
7.15. DOUBLE-PARTITION SPIROMETER 26
7.16. INTEGRATED PNEUMOTACHOGRAPHY 28
7.16.1. Description 1 28
7.16.2. Description 2 29
7.17. PERCUSSION 30
7.18. CONCLUSION ON EXISTING METHODS 31

8. **METHOD CHOICE** 33

8.1. IMPEDANCE PLETHYSMOGRAPHY 33
8.2. PERCUSSION 33
8.3. ULTRASOUND 33
8.3.1. Ultrasound of 40kHz 33
8.3.2. Ultrasound of 5MHz 34
8.4. CONCLUSION 34

9. **EQUIPMENT DESIGN** 35

9.1. EQUIPMENT DESCRIPTION 35
9.2. SCHEMATIC DESIGN 36

9.2.1. Input block ... *36*
9.2.2. Output block ... *39*
9.2.3. I/O block ... *40*
9.2.4. A/D Converter block ... *40*
9.3. CIRCUIT BOARD DESIGN ... 42
9.3.1. Output board ... *42*
9.3.2. Input board ... *42*
9.3.3. I/O board ... *42*
9.3.4. A/D converter board ... *42*
9.4. MECHANICAL CONSTRUCTION ... 43
9.5. TESTING OF THE EQUIPMENT ... 45

10. SOFTWARE ... 47

10.1. READING THE A/D CONVERTER ... 47
10.2. INITIALISATION OF THE DELAY LENGTH ... 48
10.3. DISPLAYING THE DATA ... 48

11. EXPERIMENTS ... 51

11.1. MY OWN HEART ... 51
11.2. HORSE LUNGS ... 51
11.2.1. Test runs 1.x ... *52*
11.2.2. Test runs 2.x ... *52*
11.2.3. Test runs 3.x ... *52*
11.2.4. Test runs 4.x ... *53*

12. CONCLUSIONS AND RECOMMENDATIONS ... 55

APPENDIX A: SCHEMATICS ... 57

INPUT BLOCK ... 57
OUTPUT BLOCK ... 60
I/O BLOCK ... 62
A/D CONVERTER BLOCK ... 63

APPENDIX B: CIRCUIT BOARD LAYOUTS ... 65

INPUT BLOCK ... 65
OUTPUT BLOCK ... 66
I/O BLOCK ... 68
A/D CONVERTER BLOCK ... 69

APPENDIX C: LITERATURE ... 71

LIST TYPE 1 ... 71
LIST TYPE 2 ... 72

3. Inleiding

Bij paarden is het ademvolume een grootheid die vaak veel zegt over de toestand waar het paard zich in bevindt. Tijdens de training kan het een indicatie zijn voor de vermoeidheid, en tijdens operaties geeft het aan of de anesthesie diep genoeg is. Op dit moment kan het ademvolume alleen betrouwbaar via de luchtwegen gemeten worden door een masker dat op de neus geplaatst wordt. In veel gevallen is het echter beter om de luchtwegen vrij te laten zonder er meetapparatuur tussen aan te moeten brengen.
In het onderzoek beschreven in dit verslag zal gezocht worden naar een andere methode om het ademvolume en de ademfrequentie te meten. Indien die gevonden wordt zal, indien nodig, de benodigde apparatuur ontworpen en gemaakt worden zodat de methode getest kan worden.
Het verslag begint met het onderzoek naar de anatomie en de modelvorming van de long van het paard. Zodra de benodigde kennis aanwezig is om tot een nuttige dialoog te komen worden er gesprekken gehouden met dierenartsen en eventueel andere doelgroepen. Als extra doelgroepen kan gedacht worden aan trainers voor de professionele en amateur paardenraces. Uit deze dialogen worden de wensen en eisen gedistilleerd waarna begonnen kan worden met het kiezen van een meetmethode. Voor de gekozen meetmethode wordt een schakeling ontworpen die gebruikt wordt tijdens de testen die moeten uitwijzen of de methode bruikbaar is. Afsluitend volgen de conclusies met betrekking tot de haalbaarheid en of het eventueel commercieel interessant is om een produktie op te zetten.

4. Introduction

With horses the breath volume is a good indication of the condition where the horse is in. During training it can be an indication of wear and during surgery it indicates the depth of anaesthesia. At the moment the breath volume can only reliably be measured by making use of a mask placed over the nose. In many cases it is better to leave the nose free without placing any measuring equipment in the airways so the horse can breath freely. In the research described in this report there will be searched for other methods to measure the breath volume and breath frequency. If it seems to be possible, the needed equipment will be designed and tested.
The report starts with some anatomical information that is needed to model the lung of the horse. When enough terms are known to communicate with veterinarians they are asked about their ideas and wishes. If possible some other potential users will be contacted too. For potential users one could of course think of veterinarians but also everybody who wants to know more about the condition of his horse. From these dialogues the wishes and demands are distilled, from where the measuring method can be chosen. The chosen method is then used to design an appropriate device capable of doing the job. Evaluating the chosen method and trying to predict if the product is of commercial interest will be done in the final part of this report.

5. Demands on the BreathMet

The Breathmet, a non-obstructive breath volume meter for horses, should be a device that makes it easy to measure the lungvolume of a horse in exercise. The method used should obstruct the horse as little as possible and must be easy to use. This ensures that as many people as possible will be able to use it. The price must be held as low as possible to make it interesting for everybody who has even the slightest interest in the breath volume of his/her horse.

6. Anatomical research

To be able to communicate with veterinarians I have to know the names of the organs of the horse. Studying the anatomy of a mammal helps to get to know the names needed for conversations with veterinarians. In my case only the anatomy of the horse has to be studied.

6.1. General anatomy of the horse

First the global anatomy of the horse must be known [ana1, 130-144]. Focussing onto the anatomy of the thorax and lung area, not the anatomy of the whole horse. This is sufficient because in discussions with veterinarians only this region will be discussed. It is not very likely that the feet of a horse have anything to do with the respiratory system.

7. Existing methods

Before starting to design a new method to measure the lung volume of a horse, as much as possible existing methods are evaluated in the following paragraphs. The evaluations primarily describe the technical background of the methods and will only superficially go into the actual implementation. Because of two reasons, the methods used with humans are also described and evaluated. First there is very little information on methods used in the equine field of work and second the methods used with humans always use the same physical principles as the methods used with horses.
If an existing method should fulfil the special demands for horses there will be no need to invent a new one. But if none of the methods is good enough it is justified to put a considerable amount of time in designing a new one or improving an existing method.

Note: This chapter is a literature study and so all descriptions are 'as found' in the literature. Only minor changes are made to the original descriptions. This chapter is only used as an introduction to the next chapters where the information found in literature is used to design a new method.

7.1. *Consulted resources*

To find as much as possible existing methods to measure the lung volume, all available resources should be consulted. In Utrecht there are both human and veterinarian studies with their own libraries. This meant that a very thorough search is possible while limiting the resources on local availability in Utrecht. Almost all available resources like books and articles are indexed in two computer databases accessible through the computer network used at the university of Utrecht. The first is the UBU system that is used to search for books and the second is MedLine which contains all articles and some abstracts of the important ones. MedLine is also available through Internet (http://muscat.gdb.org/) which version is user-friendlier then the local version. The benefit of the local version is that with every article found an indication is given if it is in a library in Utrecht. The Internet version doesn't know where the articles can be read. The MedLine version of the University can only be used for recent articles (1990-1996) while the internet version searches all articles from 1969 till now.

7.2. *Impedance plethysmography*

This method is an indirect method that uses the lungs as variable impedances.
The impedance is measured by applying a small (2mA) alternating (70kHz) current across the thorax by means of two electrodes connected to a constant current generator, which maintains the current constant at a given value whatever the intervening impedance may be [imp1, 114-117].
As the lung fills with air the impedance of the thorax changes because air is virtually an insulator relative to the surrounding tissue. The change of impedance associated with the lung volume is a small percentage (1 to 2 percent) of the total impedance across the thorax. Therefore the change in voltage across the thorax is also small and the measurement equipment must be very accurate.
Because this method measures the effects of a change in lung volume and not the lung volume itself it is very sensitive for disturbances. From the measured value it is not possible to say if the change is due to a change of lung volume or due to a change in any other part surrounding the lung. The method only gives relative lung volumes. To determine the absolute volume some sort of reference must be

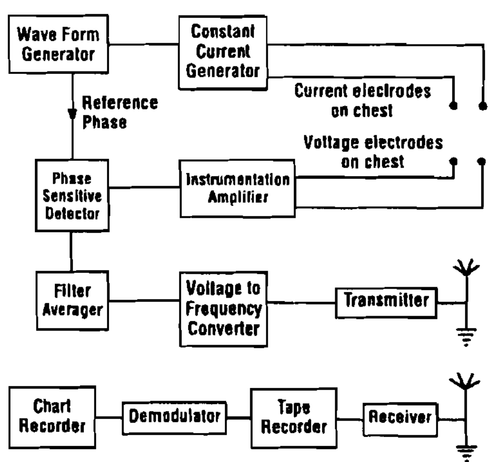


Figure 1

measured. That reference is most likely different for each horse and not constant for one horse over a longer period of time.
If the horse starts to sweat the contact between the electrodes and the skin changes and the values measured lose a lot of their accuracy. When the contacts move the impedance is also changed. So the contacts must virtually be implanted into the horses skin to get accurate results.

7.3. Sonic transillumination

In the sonic transillumination technique, the chest area is illuminated with sonic radiation in the frequency region at 100 to 300 Hz [tst1, 157]. A small loud speaker is placed at one end of the subject and a microphone on the other end. In case of a horse: on the right and left side with the lungs in between.
High attenuation indicates a well-filled lung and low attenuation suggests fluid or other matter has filled in the normal air cavities.

7.4. Ultrasound

Ultrasound is sound with a frequency that can not be heard by humans. For diagnostic examinations ultrasound in the range from 2 to 10 MHz [ult1, 3-18] is used depending on the anatomic region. The principle of ultrasonic imaging is that a burst of ultrasound is sent and at each interface between two tissues a portion of the energy is reflected.

Interface	*Reflection (%)*
Blood-brain	0.3
Kidney-liver	0.6
Blood-kidney	0.7
Liver-muscle	1.8
Blood-fat	7.9
Liver-fat	10.0
Muscle-fat	10.0
Muscle-bone	64.6
Brain-bone	66.1
Water-bone	68.4
Soft tissue-gas	99.0

Table 7.4.1: Sound reflection at various interfaces.

The reflections are recorded with a time marker. From the delay between the sending and receiving of the sound the distance to the interface can be calculated. If this is done for several directions the two-dimensional representation of the anatomy can be shown on a screen.

7.5. Projective imagery

The technique is a conventional one wherein a point source of X-radiation is used to irradiate the patient and photographic film used to record the two-dimensional pattern of transmitted radiation.

7.6. Tomographic radiography

The computerised tomography (CT) technique works by illuminating the patient with a narrow focussed 'pencil-beam' of X-ray from various angles and positions [tst1, 161]. A point detector is placed on the opposite side of the patient and is used to record the beam strength at each angle and position. The collected data is submitted to a computer for image reconstruction. The formed image is a cross-section of the subject.

7.7. Open-circuit and closed-circuit methods

Residual volume and functional residual capacity are often measured by either an open- or a closed-circuit method [phy4, 13].

7.7.1. Open circuit Single-breath

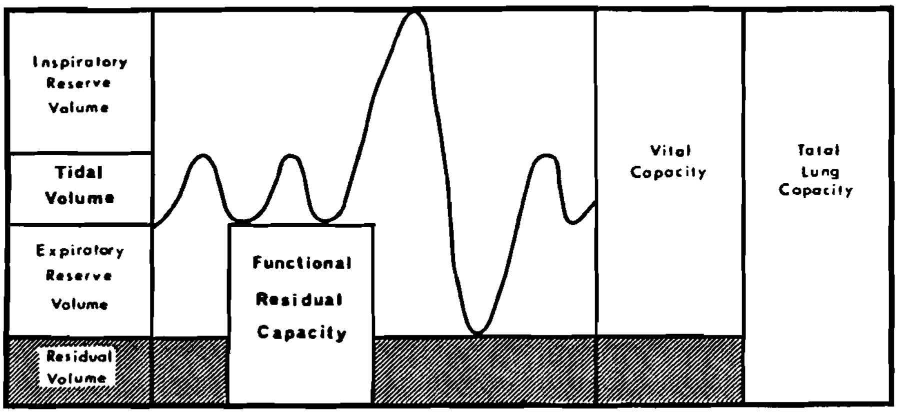


Figure 2: Definition of lung volumes

7.7.1.1. Nitrogen technique 1

Measuring the repiratory dead space can be done with a N_2 meter [phy4, 37]. A N_2 meter is used for the continuous and almost instantaneous analysis of the N_2 concentration of gas entering or leaving the subject's mouth.

The subject, who previously had been breathing room air, inspires a single breath of pure O_2; a N_2 meter records 0% N_2 during inspiration. At end-inspiration, the dead space is filled with O_2 which has just been inspired. At the beginning of expiration, the first gas sample from the mouth is pure O_2 (phase A) which has entered and left the dead space without any mixture with alveolar gas; therefore, the N_2 meter continues to record 0% N_2. Toward mid-expiration, pure alveolar gas is exhaled, uncontaminated with dead space gas (phase C); its concentration here is recorded as 40% N_2, indicating that the inspired O_2 diluted the alveolar N_2 to half of its original concentration. The actual concentration depends on the volume of O_2 inspired and the volume of the Functional-Residual-Capacity (FRC) just before inspiration. Between phases A and C is phase B. In this phase, the N_2 concentration of the expired gas rises rapidly; this represents the remainder of the dead space gas being washed out by alveolar gas , which has a N_2 concentration of 40%. The concentration of N_2 in the alveolar gas at the very beginning of phase C is used for the calculation of anatomic dead space.

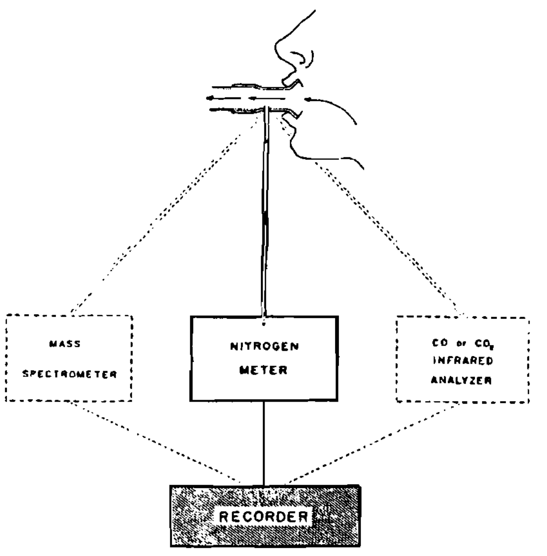


Figure 3

7.7.1.2. Nitrogen technique 2

This method is used to find lung diseases like emphysema [phy4, 65] where the distribution of gas is uneven throughout the lung. The patient inspires a single breath of O_2 and then expires slowly and evenly into a spirometer or flow meter while the N_2 meter records continuously N_2 concentration. No measurements are made on the first 750-ml of expired gas because in some patients the last part of this may contain some dead space gas. However the increase in N_2 concentration is measured over the next 500-ml of expired gas, which is certain to be alveolar gas. In healthy young adults, distribution is not perfectly uniform, but the N_2 concentration does not rise more than 1.5% throughout the expiration of this 500-ml of alveolar gas. In older healthy individuals, the N_2 concentration rises more. However, in patients with severe emphysema, the N_2 concentration may rise as much as 16%.

7.7.1.3. Nitrogen technique 3

In the single breath nitrogen test, a vital capacity (VC) inspiration of 100 per cent oxygen is followed by examination of the nitrogen in the vital capacity expirate [met3, 136]. The residual volume (RV) is calculated from the dilution equation

$$(VC \cdot F_I) + (RV \cdot F_O) = TLC \cdot F_E$$

Where F is the concentration of N_2 in the inspired gas (I), in the lungs at the onset of inspiration (O), and in the mixed expired gas (E). The mixed expired or mean N_2 concentration (F_E) is determined by collection of the total expirate in a container, or by planimetric or electrical integration of the moment-to-moment area under the expiratory N_2 concentration curve. In this test RV is calculated. Since the concentration of nitrogen in the inspired gas (F_I) is zero (i.e., 100 percent oxygen is inhaled) then

$$RV \cdot F_O = TLC \cdot F_E = (VC + RV) \cdot F_E = VC \cdot F_E + RV \cdot F_E \Rightarrow$$

$$RV \cdot F_O - RV \cdot F_E = RV \cdot (F_O - F_E) = VC \cdot F_E \Rightarrow$$

$$RV = VC \cdot \frac{F_E}{F_O - F_E}$$

7.7.1.4. Helium technique

The alveolar volume in the lungs is the volume of the lung that does not contain any dead space. The alveolar volume does actually participate in the gasexchange between the blood and the air in the lungs. The dead space does not partisipate in this process. The alveolar volume can be determined from a single inhalation of a helium-air mixture [met3, 135]. Usually it is calculated as a by-product of the single breath estimation of the diffusing capacity using carbon monoxide. In this case, a vital capacity inspiration of the gas is carried out and the TLC is calculated

$$(VC \cdot F_I) + (RV \cdot F_O) = TLC \cdot F_E$$

Since the concentration of helium in the lungs at the onset of the inhalation (F_O) was zero, the equation for calculation of TLC is

$$TLC = VC \cdot \frac{F_I}{F_E}$$

where VC is the vital capacity inspired and F is the concentration of helium in the inspired (I) and the expired gas (E).

7.7.2. Open-circuit Multiple-breath

There are several methods in this group but they all are base on the rate of washout of pulmonary N_2 when O_2 is breathed [phy4, 66].

7.7.2.1. Nitrogen technique 1

The simplest of these tests, the "pulmonary N_2 emptying rate" [phy4, 66], requires only that the patient breaths O_2, usually for 7 min, and then, by forced expiration, delivers a sample of alveolar gas at the end of the 7 min period. If the inspired O_2 is distributed evenly to all alveoli during the 7-min period, the N_2 will be washed out of these evenly and the final alveolar gas sample will contain less than 2.5% N_2. However, if some areas are markedly hypoventilated during normal breathing, they will still have a high N_2 concentration at the end of 7-min and the alveolar gas sample, obtained by forced expiration, will empty these so that the N_2 concentration of this gas is more than 2.5%. For the sake of simplicity, no measurements of functional residual capacity, tidal volume, frequency of breathing or dead space are made. However, the completeness of N_2 washout during the 7-min period does depend, among other things, on these factors. For example, patients with pulmonary disease may hyperventilate even at rest, and this tends to reduce alveolar N_2 to normal values at the end of a 7-min period even though some areas are poorly ventilated. This was demonstrated in a patient who had a large emphysematous bulla and decidedly uneven ventilation by the single-breath N_2 meter test (7% increase in expired alveolar N_2 concentration over a 500 ml volume); his 7 min "pulmonary N_2 emptying rate" was normal, because his minute volume breathing was about three times normal ventilation during the test (16.5 l/min).

The second and more difficult but quantitative test is one in which the stepwise decrease in the alveolar P_{N_2} is followed continuously, breath by breath, by rapid electrical analysers [phy4, 68]. This method is used to find if there is uneven ventilation of the alveolar gas. The extent of uneven ventilation is indicated by the deviation from a straight line of the actual values obtained in the patient. This test is seldom used because it is time-consuming and requires very precise measurements.

7.7.2.2. Nitrogen technique 2

In this method the nitrogen in the subject's lungs is washed out by successive inspirations of oxygen over a 7 minute period, while the expirations are collected in either a spirometer or a bag (open circuit) [met3, 134]. The subject is switched into a circuit that enables him to Inspire 100 per cent oxygen at the end of a normal expiration. From knowledge of the volume of oxygen breathed during the test, the amount of nitrogen that is washed out of the lungs, and the concentration of nitrogen left in the lungs at the end of the test, the volume of gas in the lungs at the onset of the measurement (FRC) can be derived

$$FRC = \frac{(V + V_{DS})(F_E - F_I)}{F_O - F_A} - C$$

where V is the volume of air expired; V_{DS} is the dead space of the spirometer, the tubing and mouthpiece; F is the concentration of N_2 in the expired gas (E), in the
inspired oxygen (I), and in the alveoli at the onset of the test (O), and the end of the test (A). C is a correction factor for the N_2 excreted from the blood into the lung during the oxygen breathing.

7.7.2.3. Nitrogen technique 3

This *open-circuit* method is based on the following principle [phy4, 13]. The volume of gas in the patient's lung is unknown. It is known, however, when the patient is breathing air, that this gas contains about 80% N_2. If the amount of N_2 in his lungs could be determined, the total volume of alveolar gas could be calculated easily. The amount of N_2 is determined by washing all the N_2 out of the lungs and measuring it. This is achieved by having the patient inspire O_2 (N_2 free) and then expire into a spirometer (previously flushed with O_2 so that it is N_2-free). He continues to do this for some minutes. The expired gas is collected in the spirometer so that its volume and N_2 concentration can be measured.

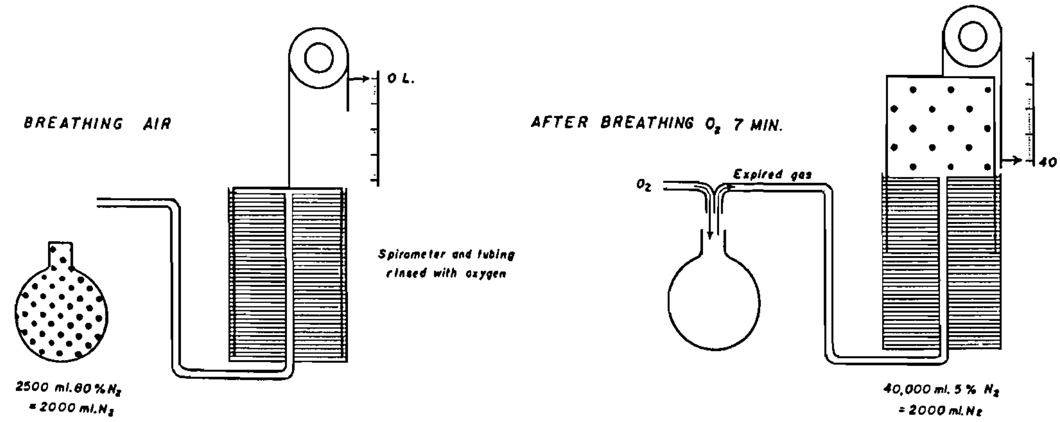


Figure 4

7.7.3. Closed-circuit Multiple-breath

7.7.3.1. Helium technique 1

The principle of the closed-circuit method uses a bellow with some known percentage of He.

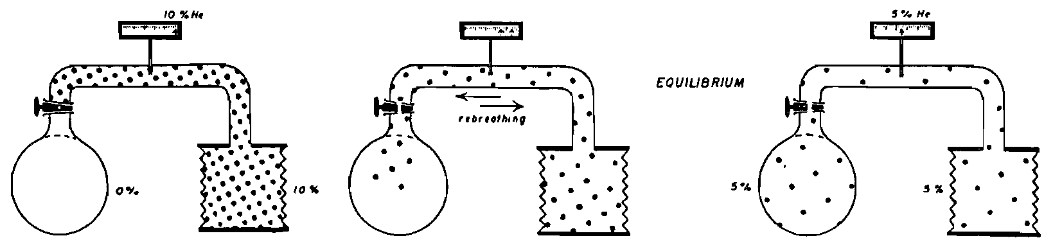


Figure 5

The patient then starts breathing the gas in the bellow in a closed circuit. So his expired breath is mixed with the air in the bellow. After some time the concentration of He in the bellow stabilises. The residual concentration is then used to calculate the total volume of the bellow added with the volume of the lungs.

7.7.3.2. Helium technique 2

If a precisely measured quantity of a relatively insoluble foreign gas such as He is added to a closed circuit, the curve of its dilution by alveolar gas, during rebreathing, gives an index of distribution [phy4, 69]. The rate may be an exponential one, indicating uniform distribution, or it may be rapid initially because of contributions from the well-ventilated areas and then slower because of less rapid exchange with poorly ventilated areas. The closed-circuit apparatus is of considerable value because of its versatility. During rebreathing with a gas such as He to an equilibrium point, the functional residual capacity can be measured. When the time to attain equilibrium is measured, an index of the volume of alveolar ventilation is obtained. When the curve of attaining equilibrium is analysed, an index of distribution of inspired gas is obtained.

7.7.3.3. Helium technique 3

In clinical laboratories where a body plethysmograph is not available, lung volume is often determined by a closed-circuit technique in which helium is utilised as the reference gas [met3, 133]. In this test the subject inspires and expires from a spirometer containing 8 to 10 per cent helium in air and a carbon dioxide absorber. The test continues until equilibrium is reached (i.e., the concentration of helium is the same in the lungs and in the spirometer) and has remained stable for at least one minute. Sufficient oxygen (usually about 250 to 300 ml/min) is added to the system throughout the determination to keep the end-expiratory lung volume constant. Knowing the volume of gas in the spirometer at the beginning of the test (V_S), the dead space of the spirometer and its tubing and the mouthpiece (V_{DS}), its helium concentration (F_I), and the helium concentration after equilibrium has been reached (F_E), one can calculate the volume of gas (V_{TG}) in the lungs from the following dilution equation

$$(V_{TG} + V_S + V_{DS})F_E = V_{TG} \cdot F_O + (V_S + V_{DS})F_I \Rightarrow$$

$$V_{TG} \cdot (F_E - F_0) = (V_S + V_{DS}) \cdot (F_I - F_E)$$

Since the concentration of helium in the lungs at the beginning of the measurement (F_0) was zero, and if the initial inspiration from the spirometer began from the FRC, then

$$V_{TG} \cdot F_E = (V_S + V_{DS}) \cdot (F_I - F_E) \Rightarrow$$

$$FRC = \frac{(V_S + V_{DS}) \cdot (F_I - F_E)}{F_E}$$

7.8. *Body plethysmographic*

7.8.1. Description 1

This method is used to measure the thoracic gas volume [phy4, 19]. This is defined as the volume of gas in the thorax, whether in free communication with the airways or not. The principle of its measurement in the body plethysmograph is based on Boyle's law, PV=P'V', which states the relationship between changes in pressure and volume of a gas, if the temperature remains constant. The patient sits within the body plethysmograph and the air-tight door is closed. He breathes the air about him through a mouthpiece. At the desired point in the respiratory cycle (end-expiration, if one wishes to compare thoracic gas volume with functional residual capacity as measured by O_2 or He techniques), the mouthpiece is occluded by an electrically controlled shutter; the patient continues to breathe against this obstruction. At end-expiration, we know that alveolar pressure, P, is equal to atmospheric pressure, because there is no gas flow; V, or thoracic gas volume, is unknown. The airway is then occluded. During the succeeding inspiration, the thorax enlarges and so decompresses the intrathoracic gas; this creates a new thoracic gas volume (V, original volume, plus ΔV, the increase in volume caused by decompression) and a new pressure, P'. The new pressure, P', is measured by the gauge in the airway between the patient's mouth and the occluded airway. The increase in thoracic volume (ΔV) is determined by noting the rise in plethysmographic pressure which is detected by a very sensitive electrical gauge (enlargement of the thorax compresses the air around the patient). Knowing P, P' and ΔV, we can solve for the unknown volume V, the original thoracic gas volume.

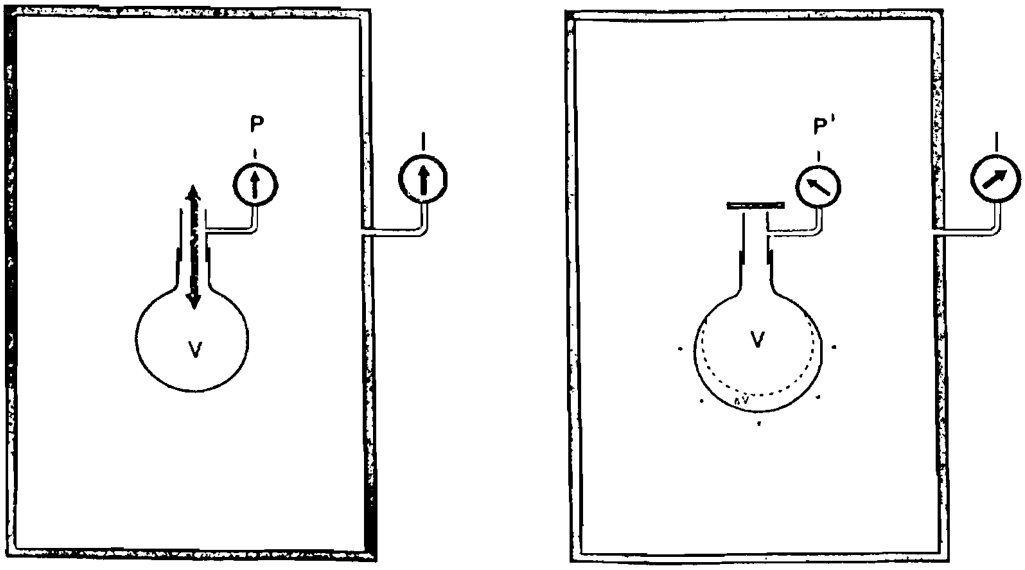


Figure 6

7.8.2. Description 2

Whole body plethysmography involves placing the animal inside an airtight, inexpansible box and having it breathe through a mouthpiece leading to the outside, a situation that is not easily adaptable to the equine temperament [met1, 177]. The method does not permit measurements of thoracic gas volume, airway resistance, tissue resistance, pulmonary blood flow, and compliance.

Boyle's law states that the volume of gas in a container varies inversely with the pressure to which it is subjected [met3, 130]. This principle is applied in practice when the volume of gas in the lungs is determined in an airtight chamber (body plethysmograph). The thoracic gas volume is determined with the subject sitting in the body plethysmograph and breathing through a mouthpiece shutter system. Since the air in the chamber heats up and is humidified by the subject's expirations, the pressure within the box rises rapidly after he first enters it. The pressure must be vented to the outside on repeated occasions until the pressure drift is minimal before measurements can be made. When there is little or no drift in the box pressure (indicating that the end-expiratory level is stable) the measurement of lung volume can be undertaken. The principle behind the calculation of lung volume is simple: while the subject is breathing quietly through the mouthpiece, the gas in the subject's lungs is at atmospheric pressure at the points of end-inspiration and end-expiration where there is no airflow. If the shutter is closed at the end of a normal expiration, the gas within the chest will be trapped at that lung volume. The subject is then instructed to pant and make gentle inspiratory and expiratory efforts against the obstruction at a rate of approximately 120/min (while holding his hands

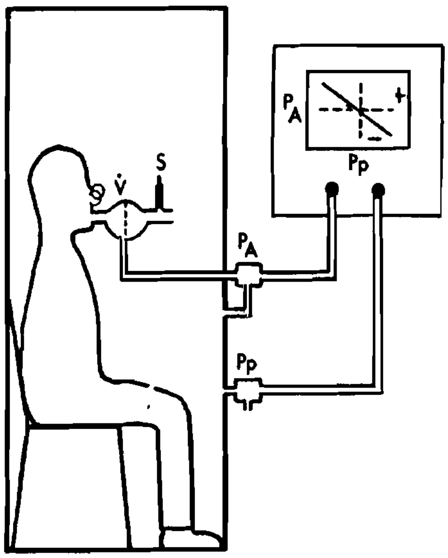


Figure 7

against his cheeks to prevent gas movement into and out of the mouth). When this is done, the air in the chest will be alternately compressed and decompressed. During the compression and decompression, the relationship between changes in pressure measured at the mouth (which are equal to alveolar pressure) and changes in Thoracic gas volume (as reflected by changes in pressure within the box) are observed continuously on an oscilloscope, the slope of this line being $\Delta P/\Delta V$. Unsatisfactory tracings will be obtained if the subject closes his glottis, or if the cheeks are allowed to balloon out or be sucked in during the breathing motions.

7.9. The capnogram

The capnogram is a diagram in which the concentration of CO_2 in the expired air is plotted against time.
The CO_2 concentration can be measured by means of a Kataferometer [met 6, 9]. The expired air is sucked up by a vacuum pump and blown along the head of a Kataferometer. The Kataferometer is a wire of platinum placed in a Wheatstone bridge. When air with a high concentration of O_2 passes along the wire, the bridge is in balance because O_2 is a good heat conductor and the wire stays relatively cold. Because CO_2 conducts heat much less then O_2 the wire is heated as the concentration CO_2 rises and the bridge is shifted away from the balance.

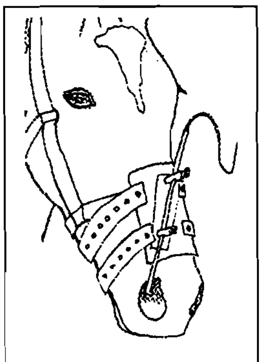


Figure 8

7.10. ^{133}Xe

This common investigative technique was introduced by Milic-Emili et al. (1966) whose exploitation of then ^{133}Xe method is one of the most important contributions to the radioactive gas literature [tec1, P404/19]. Measuring regional ^{133}Xe concentrations they applied gas dilution principles to single breath ^{133}Xe data. In principle, subjects inhaled uniformly tagged breaths beginning at any given lung volume (V_0) and terminated at TLC, where regional count rates were measured during a breathhold. The amount of isotope inspired equalled the amount of ^{133}Xe in the region at TLC:

$$U_b = F_b \cdot V \cdot \lambda$$

$$U_e = F_I \cdot V \cdot \lambda$$

$$F_b = F_I \cdot \frac{U_b}{U_e}$$

$$(TLC - V_0) \cdot F_I = F_b \cdot TLC$$

$$TLC = \frac{F_I}{F_I - F_b} \cdot V_0$$

$TLC-V_0$	Volume inspired by the region
F_I	Concentration of ^{133}Xe in the breath, measured with a counter in the spirometer
F_b	Regional alveolar ^{133}Xe concentration at TLC
TLC	Regional TLC
U_b	Regional count rate
U_e	Regional count rate at equilibrium
V	Regional volume
λ	Absorption and geometry

Thus regional lung volume as a fraction of regional TLC, could be measured at any overall lung volume between RV and TLC.

This method is also known as pulmonary scintigraphy [met2, 8-9]. It is a very informative, non-invasive and safe technique that has been currently used for several years in human medicine for diagnostic purpose.

In particular, radioactive noble gases have been successfully used in the past. The technique of labelled aerosol inhalation has been introduced rather recently and is now widely applied. Aerosol inhalation/perfusion lung scintigrams provide only localisation of ventilation or perfusion defects in projection. However by quantifying the information from the scintigrams, real functional data can be obtained.

7.11. Calorimetric flowmeter

A miniaturised heated wire is introduced into the gas stream [tec1, P413/6]. Its cooling is a function of mass flow. It is referred to as a hot wire anemometer and is quite sensitive to the heat conductive properties of the gas and is not easily calibrated if the experimental protocol includes the use of more than one gas mixture. The probe maybe either an electrical resistance with a high thermal coefficient or a metal film coated on some non-conductive material [tec1, P413/15]. The probe is, in general, minuscule and the device is properly a velocity meter since gas velocity at the site of the probe is sensed. Under certain restrictions of geometry and flow range, the measured velocity is representative of the average velocity in the measuring cross-section, and the meter may be utilised to measure flow. The theory predicts that the flow is related to the meter output E by:

$$\cdot \rho = A + B \cdot E^4$$

The anemometer is a mass flowmeter with strongly alinear output. Electrical linearization is usually provided with modern models. The meter is basically dependent on heat conductivity of the gas and needs to be calibrated and linearized for each gas mixture. It is not convenient if used when gas composition is frequently changed. The probe is heated to 200-400 °C and temperature variation of the gas in the physiological range is of little influence on the reading. The hot wire anemometer has a very high frequency response. The response time is about 0.001 s. It is therefore the flowmeter of choice in studies involving high frequency oscillating flows. At such high frequencies gas inertance is an important part of the overall impedance, and gases may be appreciable compressed and expanded during each oscillation cycle. The hot wire anemometer is then a useful alternative for measuring flow since it measures mass flow.

7.12. Doppler effect flowmeter

A laser beam is reflected with a Doppler frequency shift from particles seeded in the gas stream [tec1, P413/6]. The advantage of this technique is that it measures volume flow independently of physical properties of the gas and does not involve any restriction to flow such as in orifice or capillary tubes. Present technology allows us to concentrate the laser beam on a very small area (0.1 mm across). Just as the anemometer, the instrument is therefore a velocity meter which can be converted into a flowmeter. It has a high frequency response (up to several kHz) and is a true volume flowmeter.

7.13. Wright respirometer

This device is a miniature air turbine with moving parts of very low inertia [met5, 174]. The revolutions are recorded by means of a gear train and dial of the type used in wristwatches. The instrument indicates directly the number of litres that have passed between two successive readings. The respirometer responds to gas flow in one direction only and may therefore be used with tidal flow. The internal volume is only 22 ml and the patients may breath to and from through the apparatus with negligible increase in resistance and dead space.

In common with all inferential gas meters the response is dependent on the flow rate, since slip occurs to a greater degree as the flow rate is decreased. The instruments are adjusted to give correct readings at 20 l/min continuous flow, or 7 l/min reciprocating flow. Above this figure the response increases to become steady at about 10% high for large minute volumes. At lower flow rates the instrument reads low, being about 10% low at continuous flow of about 10 l/min or reciprocating flow of about 3 l/min (sinusoidal flow). However, the performance at low minute volumes is markedly improved when the respiratory waveform departs from sinusoidal (as it usually does during anaesthesia) and when nitrous oxide is inhaled.

7.14. Spirometry

The capacity of most commercially available spirometers is too small for use in large horses [met1, 172]. However, a suitable bell shaped spirometer can be built without too much difficulty or expense. Water-sealed bell spirometers offer a number of potential advantages: they provide the most accurate method of measuring minute volume and are the standard against which the accuracy of other techniques is assessed, minimal recording apparatus is required; calibration is independent of gas composition; corrections are required only for the difference between body and spirometer temperature; and a gas sample can readily be obtained for analysis.

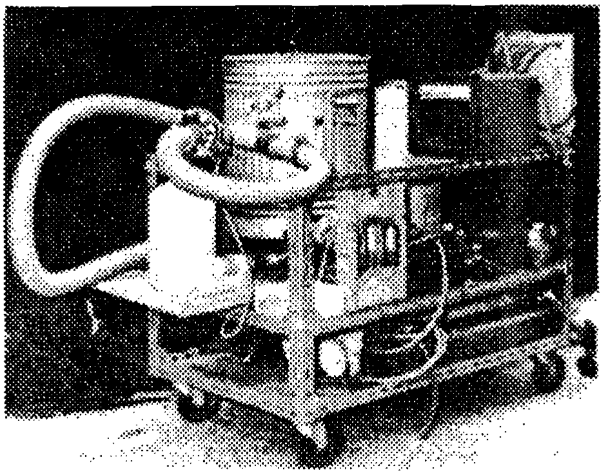


Figure 9

Spirometry can readily be performed using a face mask, nasal endotracheal tube, or chronic tracheostoma. However, the information provided does not justify the use of the cumbersome apparatus. If used for rebreathing studies, the system requires circle valves and a CO_2 absorber. The slope of a spirometer trace provides some indication of flow rates, but the high inertia of spirometry equipment precludes accurate measurement of instantaneous gas flow rates. In large animals, bell spirometers are most useful for studies of oxygen uptake, helium gas dilution, and breath-by-breath recording during general anaesthesia when adapted to a "bag-in-a-box" system.

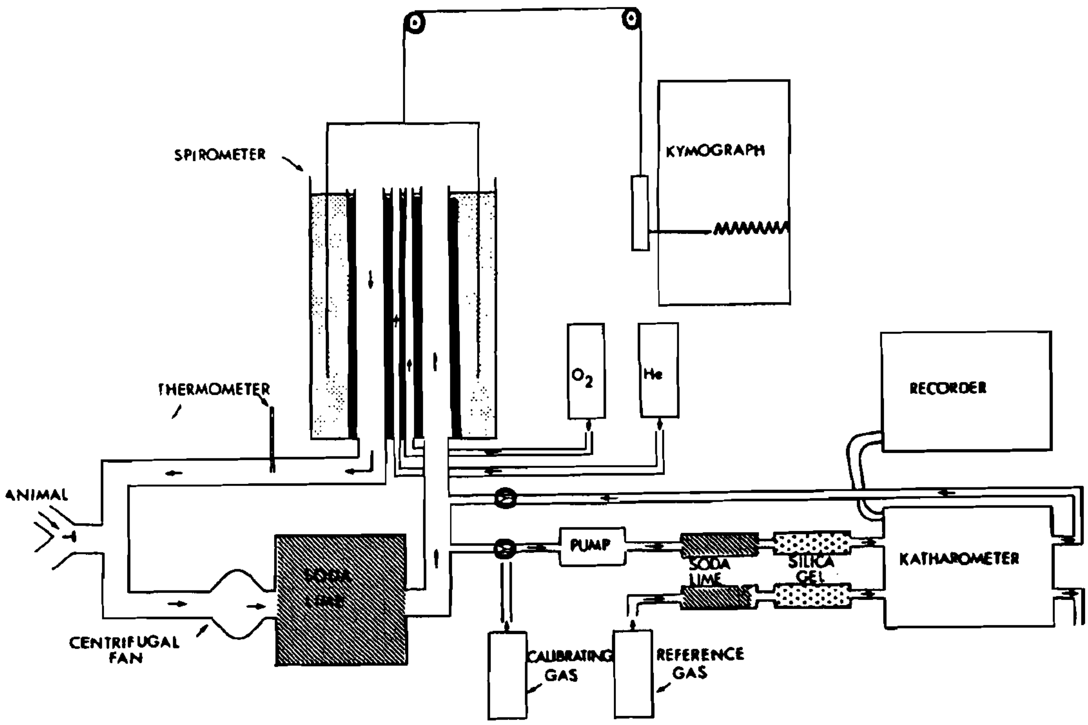


Figure 10

7.15. Double-partition spirometer

The study, the development and the construction of apparatus for the determination of the metabolism at rest and under working conditions has induced some researchers to construct a spirometer on a new principle, the "double-partition spirometer" which presents considerable advantages over an ordinary spirometer. Let us consider the spirometer in the closed circuit system shown in Figure 11. Either the mask M is placed tightly over the face of the patient or the mouthpiece is introduced between his lips and teeth. A continuous stream of air produced by the pump P passes through the mask M after it has been freed from CO_2 in the absorption container Ab. The air enters the mask M through the corrugated rubber tube Ti, leaves it through a similar tube Te and then passes through the spirometer Sp before returning to the pump P. When air is inhaled, part or all of it entering the mask M through Ti goes into the lungs so that there is less air or even none at all which enters the spirometer Sp through the expiratory tube Te. Since the pump P draws in a constant volume per unit time, the spirometer bell C, balanced by its counterweight CP, is lowered. At expiration the expired air is added to the constant air stream produced by the pump P; the output passing through the tube Te is therefore increased and the spirometer bell C is raised. In order to avoid any rebreathing the speed of the air current produced by the pump P must be at least as high as the maximum respiratory speed so that all of the expiratory tube Te. At an easy rate of respiration the peak respiratory output reaches 0.5 l/sec. In order to avoid rebreathing, therefore, the output of the pump P must at least attain 0.5 l/sec = 30 l/min. During intensive physical work a respiratory speed of as much as 8 l/sec or even more is reached, so that pump P must work at the rate of 480 l/min. Such a high output presents two disadvantages: it is more difficult to absorb CO_2 at higher air speeds, and the resistance to a fast air stream is considerable. The resistance in the tube Te at expiration becomes particularly large and troublesome. To the pump output of 8 l/sec we must add the output of 8 l/sec expired so that 16 litres of air per second pass through the tube Te at expiration. The great resistance at this speed increases the pressure in the mask M; proper expiration is thus impeded and, even worse, the mask M may lift off the face and begin to leak. We must not forget that turbulence exists in the whole system so that doubling the speed multiplies the pressure by approximately four. These objections are overcome in the "double-partition spirometer". Its principle and its operation are evident from Figure 12.

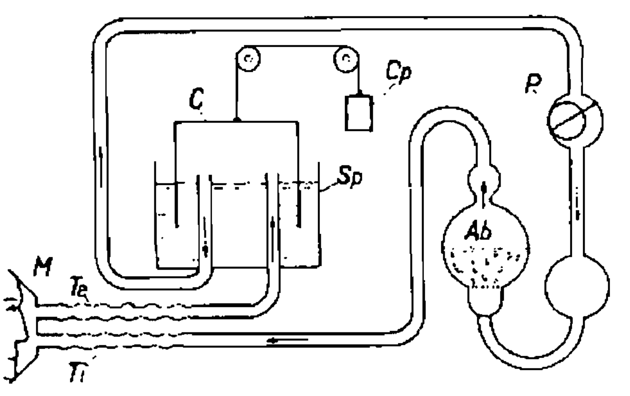


Figure 11

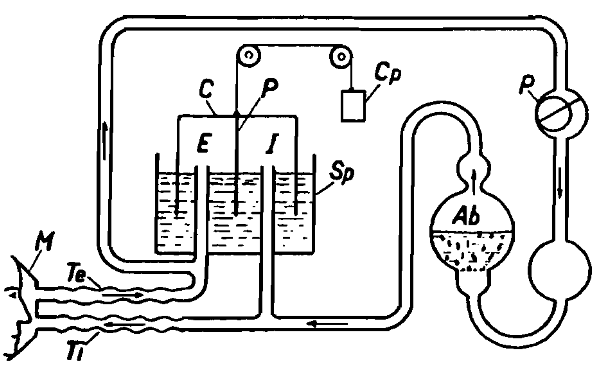


Figure 12

The spirometer bell C is divided by the wall P into two approximately equal compartments E and I. E forms the expiration chamber connected to the tube Te, I is the inspiration chamber connected to the tube Ti. This division of the spirometer bell into two connected partitions produces a new and surprising action. When the subject is not breathing, the air goes through the pump P and the absorption container Ab, passes directly through the tube Ti into the mask M and then returns through Te to the pump P. No air therefore passes into or out of the two partitions E and I of the spirometer. Assume that the pump P discharges 4 l/sec, that is, only half the maximum respiratory output. The patient inspires at the maximum rate of 8 l/sec. This air reaches him through the tube Ti, but since the pump only delivers 4 l/sec, the other 4 l/sec are taken from the chamber I. The bell C is therefore lowered thus decreasing the contents of the chamber I by 4 l/sec. During this inhalation at the rate of 8 l/sec, there is no air current in the tube Te, so that there is no rebreathing. The pump P draws from the chamber E a volume of air corresponding to its delivery, that is 4 l/sec. The content of the chamber E therefore decreases by 4 l/sec, a rate equal to the reduction in the content of the chamber I. At expiration, which we will assume to be at the same rate of 8 l/sec, the expired air passes through the tube Te. But since the pump P will only draw 4 l/sec the volume of the chamber I increases by 8 - 4 = 4 l/sec. At the same time the Pump Y delivers 4 l/sec into the chamber I, which, since the spirometer bell C is raised, increases in volume at the rate of 4 l/sec. During this time no air movement takes place through the Tube Ti. Let us compare the maximum air discharge of a normal spirometer (Fig. 11) on the one hand and of the double-partition spirometer (Fig. 12) on the other.

		Normal spirometer [l/sec]	Double-partition spirometer [l/sec]
	Delivery required of pump P	8	4
Inspiration	intake of the patient	8	8
	delivery of the tube Ti	8	8
	delivery of the tube Te	0	0
Expiration	intake of the patient	8	8
	delivery of the tube Ti	8	0
	delivery of the tube Te	16	8

It is therefore seen that the double-partition spirometer presents 'the following advantages with respect to an ordinary spirometer:

- For a given speed of respiratory current, the pump P is only required to provide half the output necessary for an ordinary spirometer, without any possibility of rebreathing.
- This halving of the output of the pump P improves the absorption of CO_2 in the absorption container Ab since the air current passes through it more slowly and since the CO_2 content is doubled.
- While in an ordinary spirometer the air current reaches a speed of 16 l/sec in the tube Te, it only attains 8 l/sec in the double-partition spirometer. The respiratory resistance is therefore reduced to one quarter so that pressure variations in the mask are also decreased.

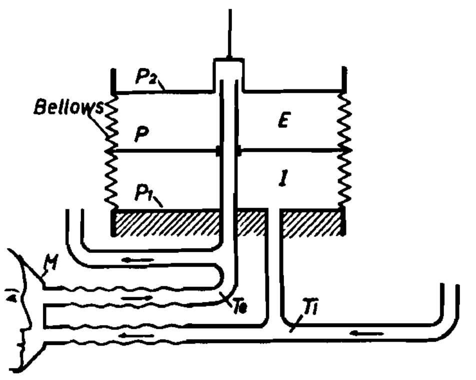


Figure 14

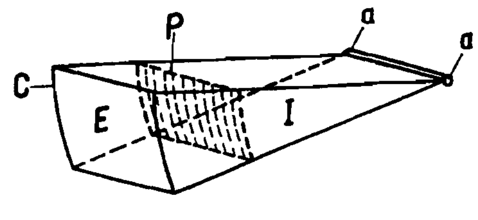


Figure 13

If the partition of the spirometer bell C is transverse as is indicated in Figure 12, the bell will have a tendency to heel over since the pressure in the chamber I will always be higher than in the chamber E. This tendency can be eliminated by using a double spirometer constructed of two concentric bells, whose volumes are the same. If the spirometer pivoting on the axis a-a (Fig. 13) is used instead of the cylindrical bell, the partition P can be placed transversely as indicated in the figure, or longitudinally. Of course, the spirometers shown in Figures 11, 12 and 13 must be sealed off with water. This is the easiest way to ensure airtightness. On the other hand, water is easily spilt when the spirometer is moved so that one now prefers apparatus without water in which the spirometer consists of a rubber bellows. As is shown in Figure 14, it is also possible to make a double-partition spirometer of this type.

The bellows made of a corrugated rubber tube are fixed at the lower end by a metal base P_1. The upper part is fixed to the light metal plate P_2. The plate P_2 is raised or lowered as the volume of the bellows is increased or decreased. In order to obtain the effect of the double-partition spirometer, a transverse partition P is placed in the middle of the bellows so that the volume is divided into two approximately equal parts. E is the expiration chamber and I is the inspiration chamber, similar to the chambers E and I in Figure 12. All the rest of the apparatus is identical to that shown in Figure 12 and works in the same manner. It is not necessary that the partition P should be hermetically sealed around the vertical tube Te passing through it. A slight passage of air from the inspiration chamber I, where the pressure is higher, to the expiration chamber E, actually has the advantage of rinsing chamber E which contains expired air rich in CO_2. All corrugated rubber bellows have the disadvantage of sagging at the lower end due to their weight. The partition wall P prevents this sag; the pressure in chamber I is in fact slightly higher than in chamber E and so supports the weight of the plate P and of the lower part of tape bellows.

7.16. Integrated Pneumotachography

7.16.1. Description 1

Integrated pneumotachography has been used in almost all bovine and equine ventilatory studies on conscious subjects [met1, 173]. When properly calibrated the accuracy of the integrated pneumotachograph approaches that of a spirometer. The pneumotachograph is a fast response measuring system that consists of an air resistance, either a fine-mesh screen or a bundle of fine tubes, and a differential pressure transducer to detect the pressure drop across the resistance occurring during gas flow. When flow is laminar the pressure difference across the resistance is proportional to the flow rate and the viscosity of the gas, with conversion of flow rate to volume accomplished by electronic integration of the flow signal. The flow and volume signals are displayed on a suitable oscilloscope or rapid response recorder. A very small pressure gradient is required and the resistance to breathing and the dead space are negligible if a suitable size of pneumotachograph head is used. The facemask used with pneumotachographs must be of a suitable size and must be designed to avoid air leaks and minimise dead space. It is possible with fibreglass moulded over a stainless steel dish and a piece of piping with a diameter equal to that of the pneumotachograph head. This end piece of the facemask provides a firm attachment for the pneumotachograph head and it does not collapse over the nostril area when positioned on the horse. The flexible portion of the mask is made from a 3-mm stiff silicone rubber sheet cut and sloped to accommodate the shape of a horse's head. A 2-cm thick, 3-cm wide piece of foam rubber is glued to surround completely the inner upper edge of the face mask. For calibration, an air source (the exhaust port of a vacuum cleaner powered by a rheostat) is coupled to a calibrated rotameter and the pneumotachograph.

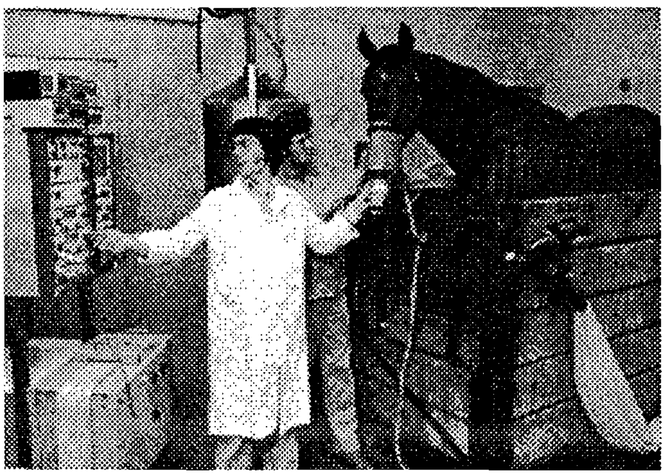


Figure 15

Horses and humans produce considerable quantities of saliva and water vapour when breathing into a face mask and pneumotachograph head. Despite the heater in the pneumotachograph head used to prevent condensation, this material can collect and change the calibration of the instrument. While testing an animal the foam rubber is smeared *with petrolatum, pack a piece of petrolatum-soaked sponge into the intermandibular space, and add more petrolatum around the upper edge of the mask to ensure an airtight seal. If there are breaks in the seal, it is easy to recognise the breaks by observing the petrolatum. The masks are likely to be leakproof since the same masks are used in the nitrogen washout studies in which the presence of leaks is easily confirmed by failure of the breath-by-breath nitrogen trace to return to zero with each breath.*

7.16.2. Description 2

This description is one of the first available and forms the basis of all Pneumographs. The article is not quite up to date but it gives a good indication on how the pneumographs were developed and evolved to the current standards.

In order to study the various phases of the respiratory cycle the in the volume of air breathed, measured by a spirometer, has often been recorded [met4, 53]. The spirogram indicates accurately the frequency and the regularity of the respiratory movements as well as the volume of air displaced with each breath, but it is very difficult to determine from it such characteristics as the exact duration of a respiratory phase or the variations in the speed of the air flow. It is thus often better to record not the integral curve of the respired air but rather its differential curve, that is the curve showing the speed of the airflow at each instant. The speed curve brings out details which are not apparent from the integral curve. The only method which indicates how the breathed air circulates and the magnitude of the force exerted by the respiratory muscles during the different phases of respiration, is the recording of the speed of the air flow; this method will also tell us whether there is a pause at the end of the inspiration or of the expiration. For these reasons many authors had attempted to record the speed of the airflow by different methods without, however, obtaining satisfactory results (at the time of this article). Already 35 years ago the first instrument is designed and described for recording under satisfactory conditions the speed of the respiratory airflow, this instrument is named the "Pneumotachograph" and the curve it gives the "Pneumotachogram". This first instrument answered all the essential requirements: minimum respiratory resistance, rapid reaction and existence or a proportional relationship between the speed of the airflow and the amplitude of the deflection of the light spot (see Figure 16). The value of the instrument has been confirmed by Silverman and Whittenberger who wrote in 1950: "Many devices have been proposed but the only previous type which has met with the requirements is the capillary tube meter of Fleisch.". The principle of the Pneumotachograph rests on Poiseuille's Law that states that in a straight rigid and narrow tube, flow delivery is proportional to pressure loss per unit length. The continuous measurement of the pressure loss, that is the measurement of the pressure difference between two points along the length of the tube, gives a differential curve whose ordinates represent the speed of the air flow, and thus the volume per unit time. This direct proportional relationship between speed and pressure difference is already used in the construction of the differential pressure rheometer for measuring the flow of a liquid. The measurement of the pressure loss for a gas is more delicate because gases, more easily than liquids, give rise to conditions of turbulence; now direct proportionality between speed and pressure difference only occurs under laminar conditions. The Pneumotachograph avoids the creation of turbulence by dividing the air into a large number of straight parallel tubes. Since we introduced the Pneumotachograph a large number of these instruments have been used for both scientific research and particularly in the last few years, for examining pulmonary function. The increasing interest evoked by the Pneumotachograph has induced a number of authors to construct instruments for the same purpose based on a similar or slightly different principle. These imitations have often given results that are inferior (The original writer of this article also sells the product he describes). Thus the good proportionality between the speed and the measured displacement, an

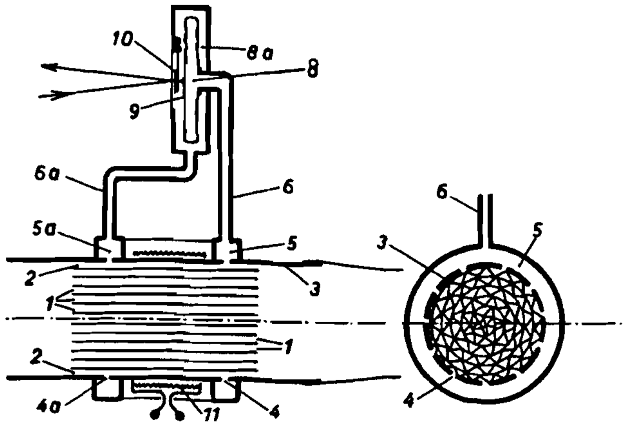


Figure 16

integral feature of the Pneumotachograph, is lost. Bouhuys has described in a recent (1958) treatise all the Pneumotachograph systems that had been perfected up to 1956. It is thus unnecessary to repeat these descriptions. I will only mention the two instruments that can give reasonable results. Silverman and Whittenberger use the resistance of a fine Monel metal sieve to produce the pressure difference. Lilly's instrument uses the same method. The difference in pressure between the entry and exit of such a sieve is only proportional to the flow provided it is very small (less than 3 mm of water). This small pressure difference indicating the speed of the air flow, cannot be measured directly and requires a complicated and expensive system of electrical amplification (Now, 1997, it is as cheap as a leave of bread). Furthermore, the proportionality between speed and displacement of the spot of light is only valid for outputs less than 1.5 l/sec., whereas intense human respiratory output can reach 5 to 7 l/sec. The great interest currently enjoyed by pneumotachography has induced us to improve our old instrument which separated the airflow into a number of parallel tubes 130-mm long and 1.8-mm interior diameter. The length of the air passages is now reduced to 32-mm, thus greatly decreasing the dead space. To improve the proportionality the inner diameter of the air passages is reduced to 0.8 mm.
The system of air passages is obtained by rolling together a thin corrugated nickel strip and a flat strip (Fig. 16). The outer corrugated layer 2 is used for the pressure tappings. The brass tube 3 has two rows of small holes 4 and 4a distributed over its whole circumference and communicating with the annular ducts 5 and 5a. The pressure is then transmitted through the tubes 6 and 6a to the differential manometer 8. If the air flow passes through the Pneumotachograph from right to left the pressure will be greater in 5 than in 5a: the membrane 3 will therefore move to the left and the ray of light will be defected upwards by the mirror 10. If on the other hand the current passes from left to right, the pressure will be greater in 5a than 5, the membrane will move to the right and the ray of light will be defected downwards. We would like to stress the fact that the membrane will only respond to differences in pressure, not to the absolute pressure. When the speed of the airflow does not exceed a certain value the air particles move in the Pneumotachograph under laminar conditions, thus ensuring strict proportionality between speed and the deflection of the ray of light. Once this limit has been passed a state of turbulence ensues, easily recognised by the slight oscillations of the light spot.

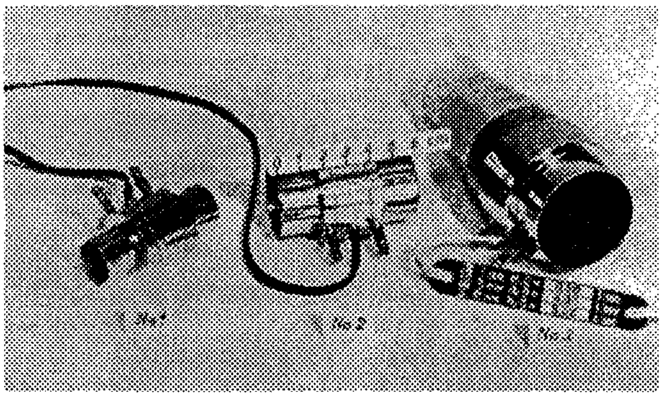


Figure 17

The metallic membrane manometer connected to the Pneumotachograph has a natural frequency of about 20 cycles per second, since the physiological oscillations of the airflow are slower, a rapid reaction and a reliable recording are assured. This new manometer has perfect stability. Even after long journeys the zero setting remains constant thus ensuring continuous strict proportionality between the speed and the deflection of the spot. A differential pressure of 10 mm of water, such as exists at the highest air speeds through the Pneumotachograph, gives, with this differential manometer, a displacement of about 40 mm on the photokymograph placed at a distance of one meter from the Frank's old differential manometer, whose light rubber membrane has a higher natural frequency, can also be used. Of course one can also use an electric differential manometer (transducer) which, beside the tachogram, can also give its integration and thus the volume breathed. To prevent water vapour condensation in the system, the apparatus is heated for 5 minutes before and also during the experiment by a small heating element built round the apparatus (6 volts, 1 amp.).

7.17. Percussion

Striking a part to produce audible sound or to bring out pain is percussion [dia1, 23]. Setting the parts in vibration produces the sound. For large animals such as horses a hammer and pleximeter is used to initiate the vibration. The nature of the produced sound indicates the type of structure that produces the sound. Because lungs produce a very different sound then the surrounding tissue it is very easy to find the borders

of the lungs by means of percussion. The borders move with the inspiration and expiration and so the position of the border could possibly have a high correlation with the volume of air in the lungs.
Today percussion is not used for determination of the lung volume but only to detect illnesses like pneumonia, pleritus and pleuropneumonia.

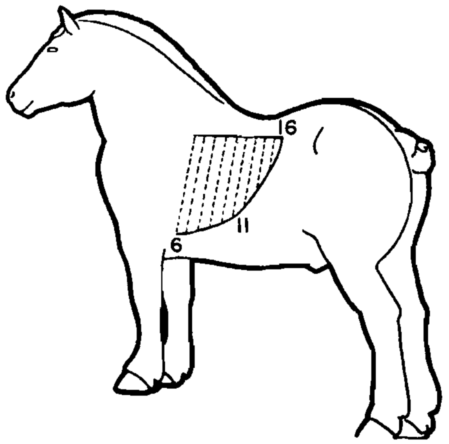


Figure 18

7.18. Conclusion on existing methods

From the previous paragraphs it can be seen that there has been a lot of research on methods to measure lung parameters. Most of the research is done on humans and only some of the methods are tried on horses or other animals. Only one method is used on a regular base with horses, the Pneumotachograph. This is a method that, like almost all methods, makes use of a facemask. According to [met7] it is seen that the use of this equipment is not optimal suited for horses, it changes the pressures in the airways considerably.
My conclusion is that the development of a new method is very much wanted by a large amount of users and potential users like trainers.

8. Method choice

This chapter uses the methods mentioned in the previous chapter to develop a new one. All methods that make use of a breath mask can't be used because of the obstruction of the airways. This limits the promising methods to very few. Because of the limited count of promising existing methods some new and very experimental methods picked up in conversations will be used in the development to see if they can be used and what the benefits eventually could be.

8.1. Impedance plethysmography

This method suffices from the problem that the impedance of the thorax changes, not only from the lung volume, but also due to the bad contacts of the electrodes with the skin. If a model of the thorax could be made, where the skin to sensor interface is not important it would result in much more accurate results. That it is possible to make such a model is not very likely because the change in impedance due to breathing is almost entirely resistive (imp2, page 168). Also it can be calculated that the dielectricum in the lung is much less then that of the surrounding tissue. So it would be virtually impossible to distinguish both impedance contributions from each other.

8.2. Percussion

Percussion is normally used to approximate the location of the lung margins. Assuming that there is a high correlation between the location of the lung margin and the breath volume this method could be used to determine the breath volume. This assumption can be made plausible by considering the way a horse breathes.

The horse breathes by first pulling the diaphragm caudal (backwards) and flattening it. The lungs follow the diaphragm and the lung margins move alongside the ribs over a distance of about 10cm. After the diaphragm is flattened almost completely, the thorax and lungs are enlarged even further by pulling the ribs lateral (sideways).

Now that the relation between the movement of the lung margins and the volume in the lung is made acceptable it is time to look for a way to mechanise the percussion method. During the mechanising of the method, it should also be made more accurate. The normal manual percussion method is just an approximation. Because the location needs to be known as accurate as possible it has to be improved. This means that just mechanising the method won't be enough. One of the possibilities to enhance the accuracy is to use higher sound frequencies and try to focus them to specific locations where the lung margins are expected. These two enhancements lead to another approach that is explained in the next chapter.

8.3. Ultrasound

Ultrasound is sound with frequencies beyond the frequency band the human ear can hear. With ultrasound it is possible to look inside a body without disturbing the tissue. It is often used for looking at foetuses to scan for abnormalities or just for fun. Ultrasound is also used for distance measurements and remote controls. For these two types of use different frequencies are used, about 40kHz for remote controls and starting at 1MHz for tissue scanning.

8.3.1. Ultrasound of 40kHz

By sending pulses of ultrasound with a frequency of 40 kHz into the lung region and picking up all reflected pulses it should be possible to see the lung margins move under the transducers. The sound wave is sent into the thorax and returned to the receiver after going 2 times through about 4 cm of tissue (Figure 19).

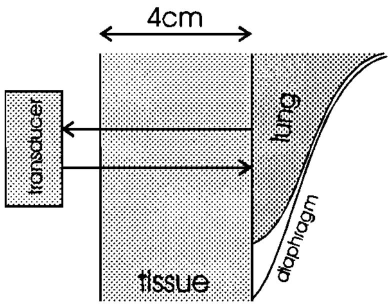


Figure 19

For simplicity let us assume that V_{sound} = 1500 m/s (Speed of sound in water-rich tissue).

$$t = 2 \cdot S_{tissue} / V_{sound} = 8 \cdot 10^{-2} / 1500 = 53 \mu s$$

The period of one sinus sent by the transducer takes $1/40kHz = 25\mu s$ and so the reflected signal is easily separated from the sent pulse that also is received.
From this calculation it is concluded that ultrasound with a frequency of 40kHz has short enough pulses to be used to follow the lung margin. Ofcourse the ultrasound pulse first has to penatrate the fur of the horse what results in a conciderable energy loss. From pilot experiments it was concluded that the amount of energy that enters the horse is large enough to result in detectable echos. Even without shaving.

8.3.2. Ultrasound of 5MHz

With high frequencies of pulsed ultrasound it is possible to get a very clear picture of the tissue directly under the skin. Tissue located deeper in animal can not be seen with high frequencies while with increasing frequencies the possible depth decreases. Because the lungs are quite close to the skin surface these frequencies could be used. The principle of low and high frequency ultrasound is the same and so a test was conducted. The test was done with high frequent ultrasound because in that case the existing equipment used for ultrasound scans could be used. The difference between 5MHz ultrasound and 40kHz ultrasound is that 5MHz ultrasound absolutely can not penetrate the fur of the horse. 40kHz is able to penetrate the fur a little better.
In the pilot experiment conducted with pulsed 5MHz ultrasound the lung margin was clearly visible and the breathing could easily be followed. One of the disadvantages of using this equipment is that it is necessary to shave all the hairs of the horse at the location where the censor is placed. In the final design this is very unwelcome and will be avoided in any way possible.

8.4. Conclusion

Because of the promising results of the conducted experiment it is expected that with ultrasound the lung margins can be followed, even with the simpler 40kHz ultrasound. Question remains if the correlation between lung margin and breathing is high enough to use the method.

9. Equipment design

First a thorough description and a clear picture of the needed equipment must exist before anything should be bought or designed. After a clear understanding of the needs is gained the equipment can be searched for. All equipment that is not bought can be designed, simulated and manufactured. At the end of the designing manufacturing process the equipment is tested to check if it functions as designed.

9.1. Equipment description

From the method some basic functions (Fig. 20) of the equipment can be derived:

- Send pulses of ultrasound (Output block).
- Receive the reflected pulses (Input block).
- Convert the received pulses to a digital notation (A/D converter block).
- Connection to a computer for data acquisition and evaluation (I/O board block).

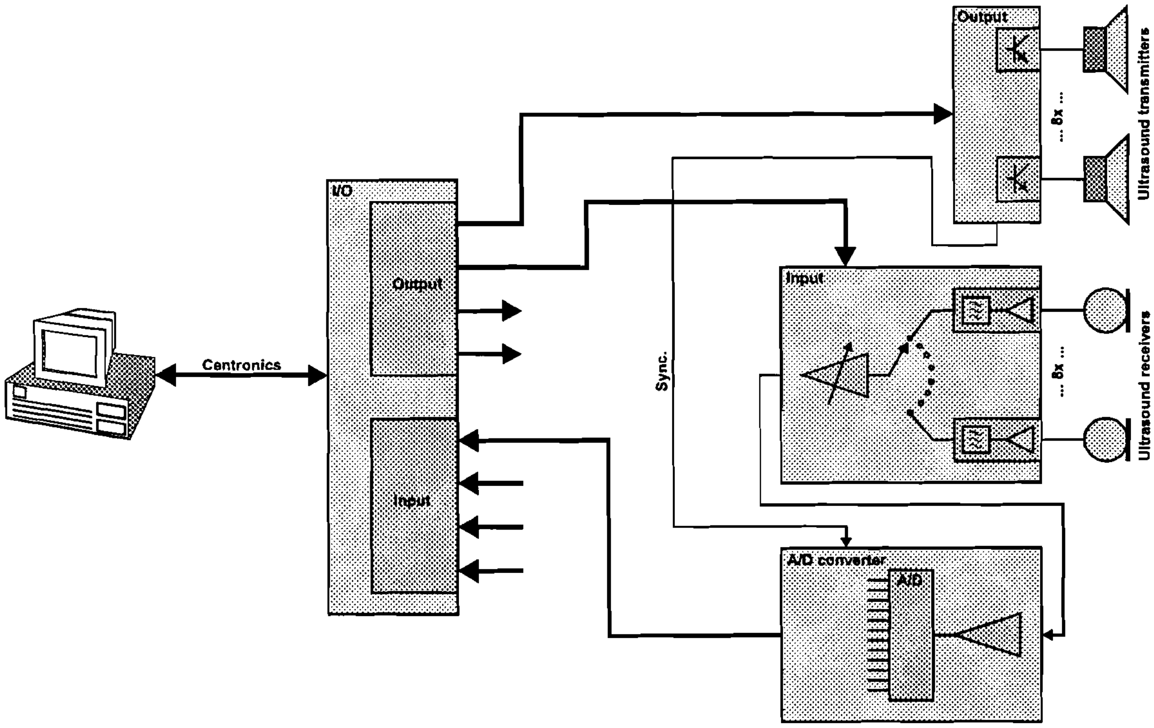


Figure 20

The *output block* sends pulses into the tissue at eight different locations. The hardware allows it that all transmitters are pulsed at once. But the software written for this equipment pulses the transmitters one at a time. This means that it is known from which transmitter the received pulses originated. The multiple locations are needed to be able to follow the movement of the lung margin. To follow the lung margin the transmitters are placed in a way that the lung margin passes as many transmitters as possible

The pulses injected into the tissue are partly reflected by tissue interfaces. The reflected signal is recorded by the ultrasound receivers from where the signal is amplified and filtered in the *input block* to get a clear signal with as little noise as possible. To correct the large changes in input signal strength the amplification of the signal must be adjustable in a large range. As with the transmitters there are also eight receivers to be able to follow the lung margin in the same way as with the ultrasound transmitters.

From the input block the filtered signal is lead to a fast enough A/D converter in the *A/D converter block*. The digital signal from the A/D converter is routed to the *I/O block* that is connected to the controlling PC. The A/D converter is a 12bit type to be able to follow even very small changes of the input signal. Because the PC uses an eight-bit bus some multiplexing has to be done in the *A/D converter block.* To receive consistent data the transmitter and the A/D converter should be synchronised. Otherwise even with a constant input signal the digital data would fluctuate due to shifting of measurement timings (Fig. 21).

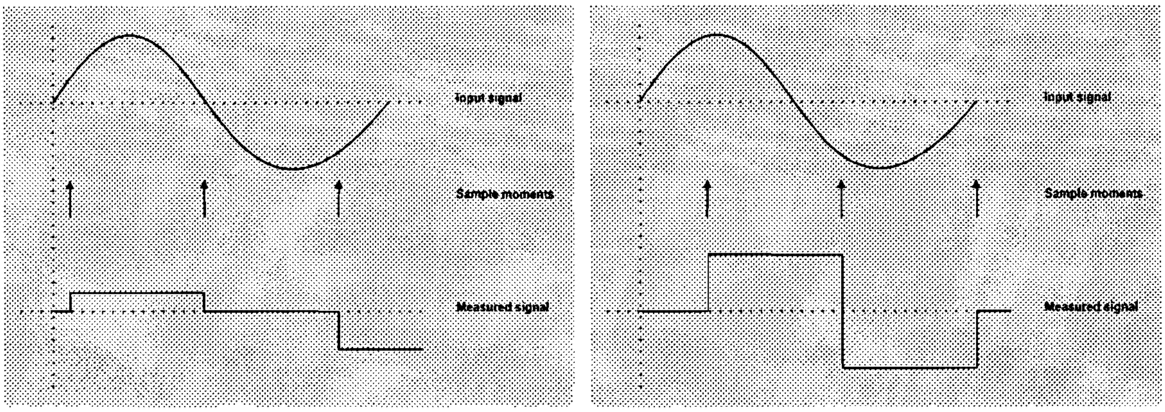


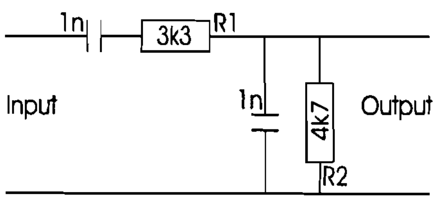
Figure 21

9.2. *Schematic design*

From the previous chapter the separate circuit blocks are clear. Because of the very small amount of interconnecting lines between the circuits the blocks are kept separate on different circuit boards. This also makes modifying the design and prototypes much easier.

9.2.1. Input block

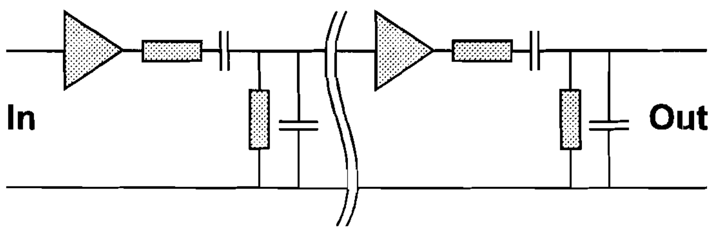
The input block amplifies and filters the signal from the ultrasound receivers to a level that the A/D converter can handle. Because the input signal is in the range of millivolts with a frequency of 40kHz the amplification should be up to 2.500 times to get the needed amplitude of minimal 2.5 Volts. To get rid of the noise and 50Hz buzz (from the mains-voltage cabling) a very sharp band-pass should be added.
Oscillation problems are very likely to occur and so some special actions have to be taken. One of the things that can be done is to use several low-gain stages in stead of one high-gain amplifier. The disadvantage is that every stage adds some noise and thus the number of stages must be as little as possible but still large enough to remove the oscillation problems. The amplification of one stage is limited by the GBW of the OpAmp. A GBW of 2MHz is feasible for common cheap IC's with four OpAmps inside. This means the maximal amplification for each stage is $2\text{MHz}/40\text{kHz} = 50$. With four stages per IC this means the total maximal amplification is $50^4 = 6.25 \cdot 10^6$. But because behind each stage a first-order band-pass filter is placed that has an amplification of 0.37x the total possible amplification is much less, $50^4 \cdot 0.37^4 = 117 \cdot 10^3$.
The amplification of 0.37x from the first-order band-pass filter is calculated in the following way.



$$|H|_{40\,kHz} = \left| \frac{\dfrac{1}{j \cdot \omega \cdot c + \dfrac{1}{R_2}}}{\dfrac{1}{j \cdot \omega \cdot c} + R_1 + \dfrac{1}{j \cdot \omega \cdot c + \dfrac{1}{R_2}}} \right| = 0.37$$

The input impedance should be large to let the signal of the ultrasound receiver enter the first OpAmp without much distortion and weakening. This means that the input signal has to go straight into the first

OpAmp without any filtering. The first OpAmp can not amplify this signal very much because this should mean that noise could easily overload the first stage. If the first stage is overloaded the rest of the stages can not repair it and so the signal becomes useless. To overcome this problem the first stage only amplifies the signal 2x what never leads to an overload. But because the first stage only amplifies the signal 2x the total amplification is limited to $2 \cdot 50^3 \cdot 0.37^4 = 4.685 \cdot 10^3$. With this amplification the maximum GBW is used and no margin is left. This means the signal is distorted and it is better to limit the amplification for each stage to approximately 30x. With this last restriction the maximum amplification for a four-stage amplifier with a forth-order band-pass filter is $2.2 \cdot 31.91^3 \cdot 0.37^4 = 1.34 \cdot 10^3$. This is too less to amplify the receiver signal up to the A/D input voltage range. To amplify the signal into the right range there are two more OpAmps in the signal path to the A/D converter, from which one is adjustable by software from 1/7x up to 213x. The other amplifier is adjustable with a potentiometer to optimise the input range. The first OpAmp is placed on the input board and the last one on the A/D converter board. The total amplification from ultrasound receiver to the input of the OpAmp place on the A/D converter board thus is 190 up to 285k times. This means that even without the last OpAmp from the A/D converter even very small signals from the ultrasound receiver are transformed into measurable voltage levels.

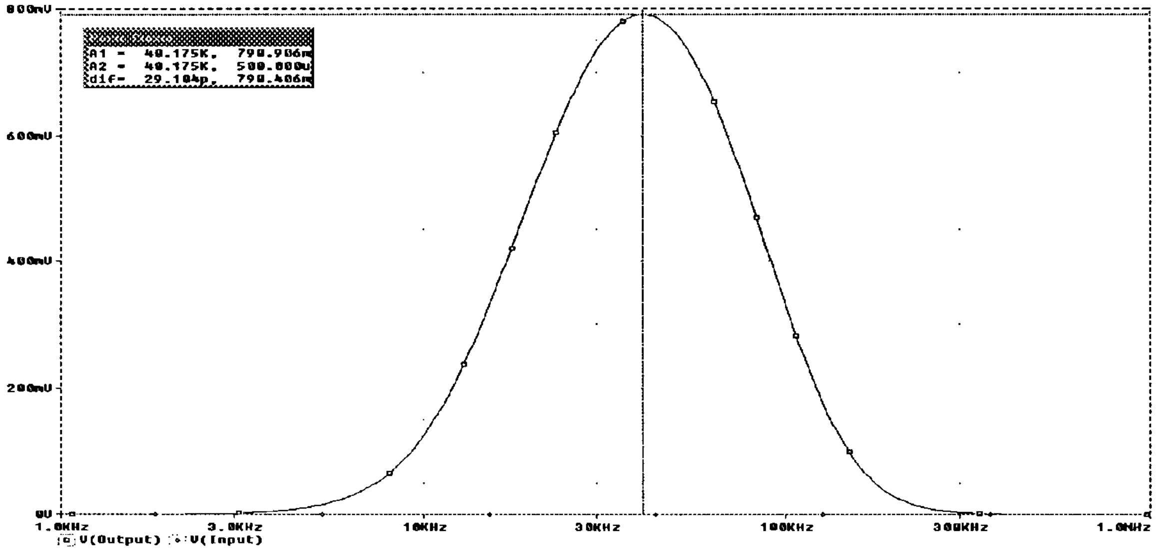


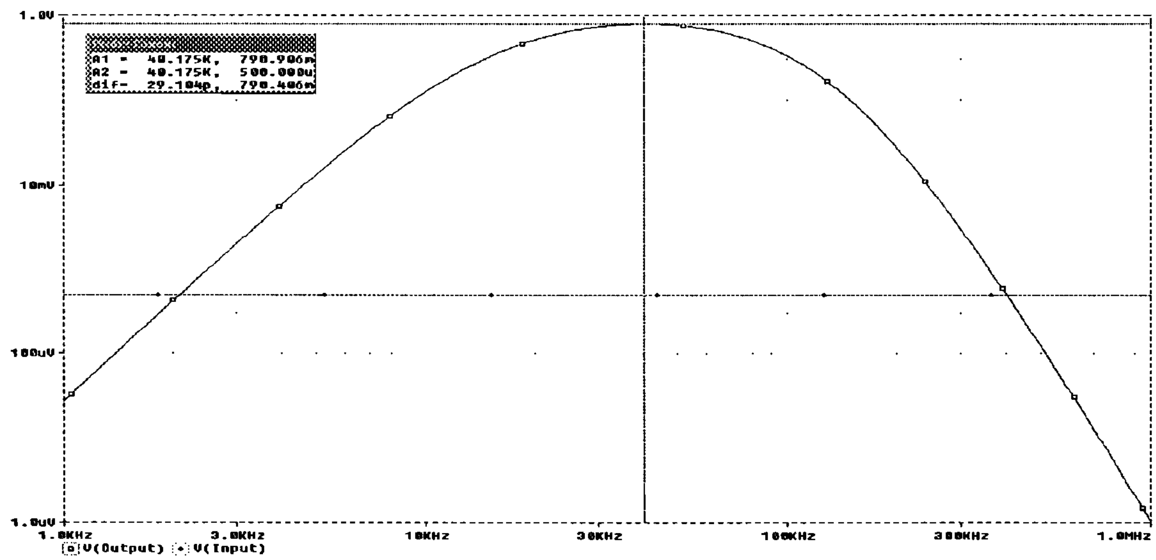
Circuit level description

The ultrasound receivers are connected to the input-connectors (K1...K8) on the top-level sheet (bm_01, page 57) of the input block. The signal of each receiver then enters the amplifier sheet (bm_01a?, page 58) where the signal goes through the first OpAmp (U1A). The amplification of the first OpAmp is small because otherwise the risk of overloading is considerable as mentioned in the previous paragraph.
The input impedance of the amplifier is 10k//47k=8k25. This impedance is empirical optimised for best impedance matching with the receiver. The 47k resistor is, besides the impedance optimisation, also added for stability reasons.
The output of the first OpAmp is connected to three exactly the same filter/OpAmp stages. Each stage consists of a simple first-order band-pass filter and an OpAmp with a total amplification of $0.37 \cdot 31.91 = 11.81$.
All outputs that come from the eight amplifier sheets (bm_01a) enter the selection and regulation sheet (bm_01b, page 59). The eight signals enter the eight-input analogue switch (ADG508F) from where the single output goes through the last slightly different filter stage. It is slightly different because it has to take into account that the output impedance of the analogue switch is approximately 300Ω. The last filter is placed behind the switch to eliminate the additional noise that is added by that switch. This last band-pass filter also eliminates most of the cross talking of the eight signals. The resulting signal enters the OpAmp (U10C) with variable amplification. The analogue switch that is placed in the feedback loop changes the amplification to the by software chosen value. With this switch the feedback resistor value is changed from 330Ω up to 1MΩ resulting in amplification from 1/7 up to 212 times (taking the output impedance of the analogue switch of 300Ω into account). The output of the adjustable OpAmp is connected to the output through a 50Ω resistor to eliminate ringing that might result from a very low impedance output of the OpAmp. Because the OpAmp output-impedance often exceeds the 50Ω, in most cases the ringing will not occur.

Simulation

With PSpice the circuit is simulated to be sure it will function as expected. Not all simulations are included in this report but only those of special interest. In this case only the complete input amplifier stage will be simulated from the ultrasound receiver to the output of the last band-pass filter. The simulation will be done without the analogue switches.





Both pictures show the same simulation results. The top picture with a linear scale and the bottom picture with a logarithmic scale. The simulated resulting amplification is $791 \cdot 10^{-3} / 500 \cdot 10^{-6} = 1.942 \cdot 10^{3}$. This differs from the calculated amplification because PSpice uses much more accurate models then the ideal models used in the calculations. Another reason for the difference is the frequency used. In the calculations a frequency of exactly 40kHz is used. PSpice results are at a frequency of 40.175kHz instead of the 40kHz used in the calculations. But with both the calculation and the simulation resulting in approximately the same results it is very likely that the actual physical circuit will also give comparing results.

9.2.2. Output block

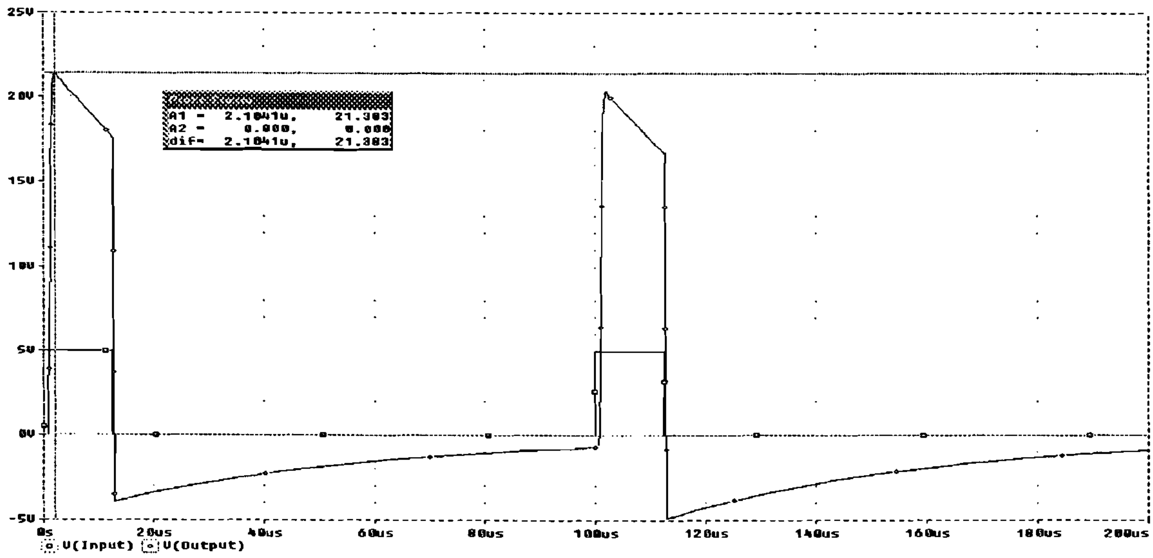
The output block transforms the digital 0...5V level from the I/O board into a signal switching between –12V and +12V. This high amplitude is needed to give the ultrasound transmitters a good kick in the ass. The higher the amplitude to the transmitters the higher the output signal will be, and thus the higher the energy contained in the ultrasound pulse. And the higher the output signal is the higher the reflected signal picked up by the receivers is. Thus the output signal must be as high as possible. The ultrasound transmitters and receivers are electrical equivalent to a band-pass filter. This means that although the input pulse to the transmitter has a rectangular shape, the pulses sent into the tissue are sinusoidal.
A higher output amplitude than 24V would be impractical because our power supplies do not support them. Our power supplies range from –12V till +12V.
To be able to synchronise the A/D converter all inputs of the output stages are combined to result in one signal. This signal is active when at least one of the outputs is active. This active signal stops the A/D converter. On the signal going inactive the A/D converter starts collecting data.

Circuit level description

Eight data lines from the I/O board enter top-sheet (bm_02, page **60**) of the output block. These inputs are connected to the eight-input AND-port (74hct30) and from there to the inverter (74hct04). The output of the inverter is used by the A/D-converter to start and stop the conversion process. All inputs to the output stages are high in the rest position and pulse low to transmit an ultrasound signal. This means that in rest the output of the inverter is high and pulses low just like the input signals to the output stages.
Each of the eight input data lines enters an output stage (bm_02a?, page **61**). The output transistor T4 follows the input TTL signal and is in saturation if the input signal is +5V. As soon as the TTL input signal falls to ground the output of C4 drops 24V. This 24V signal is transformed by the ultrasound transmitters into a ultrasound pulse with a lot of energy.
The schematics of the output stages are very simple and there even do not have to be any complex calculations. The output capacitor C4 is calculated for low resistance at 40kHz and the resisters make sure all the transistors function correctly.

Simulation

The output stage is also simulated with PSpice to be surer it will function like expected. The input signal is connected at the input (Gate) of the output stage. The output of the output stage is terminated with a 50Ω resistor. The ultrasound transmitter has an impedance of much more than 50Ω so the output voltages will actually only be better. This simulation can be seen as worst-case.



From the simulation it is seen that the amplitude of the output signal is little less than 24V. This is not surprising due to the fact that both the FET T2 and the N-P-N transistor T4 are in series with the output and both drop the voltage. The signal also can not be pulled to 0V because the P-N-P transistor T5 takes a minimum of 0.6V to stay in saturation.

9.2.3. I/O block

The I/O block connects the IBM-PC with the rest of the equipment and controls all functions on the rest of the blocks. It connects to the IBM-PC through the parallel-printer port. The parallel port can handle data speeds of up to 600 kBytes/s in standard mode. The printer port has eight bi-directional data lines and minimal four handshake lines. The I/O block used in this equipment uses multiple output and input bytes. This means that the single bi-directional printer port has to be multiplexed to additional registers. The handshake lines are used to select the active registers that the printer port is connected to.

Circuit level description

The computer printer port is connected to the rest of this block through a 25 port D-connector (J1). All handshake signals must be pulled to V_{cc} because the computer uses open collector ports to drive them and so can only pull them to ground. The data and handshake lines use series-resistors (R1...R12) to remove any possible ringing on these signal lines and to protect the ports of the printer port and the I/O block.
The eight data lines are connected directly to the output registers and the input buffers. The data lines are a bit mixed up because the design of the board was much easier this way. The disadvantage of this mixing up is that the software becomes a little bit more complex because it has to take into account which register is addressed.

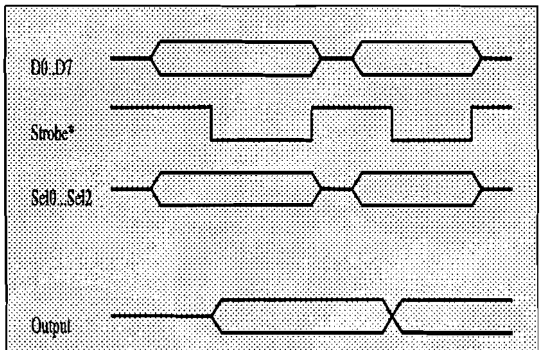


Figure 22: Output cycle

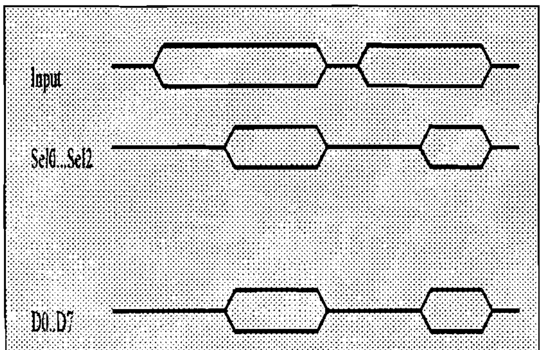
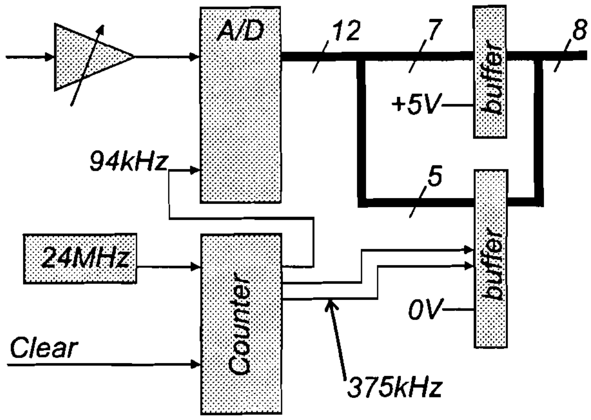


Figure 23: Input cycle

The addressing of the port is done with the handshake lines (sel0...sel2). These handshake lines are de-multiplexed to eight data signals by the 3to8 line decoder (U1). Four outputs go to the input buffers and the remaining four are used to enable the triggering of the output-registers by four OR-ports (U10). The fourth handshake line (strobe*) is used to clock the output of the eight data lines into the correct output register.

9.2.4. A/D Converter block

The A/D converter block converts the analogue signal from the input amplifiers into a 12 bit digital representation. The 12 bits are multiplexed to two groups of bits, one group of 7 bits and one group of 5 bits. These groups are, together with some synchronisation signals, connected to the I/O block through two 8-bit buffers that switch between the two groups.
Before the input signal enters the A/D converter one adjustable OpAmp amplifies the signal to the correct level for the converter. This OpAmp is added because of two reasons. One because this extra OpAmp makes the A/D converter block more universal so it can be used for other projects too and two because the input signal can now be optimally regulated to the input range of the A/D converter. It is important to use the input range of the A/D converter optimally because optimal accuracy can only be achieved by using the full input signal range.



The computer printer port is not fast enough to use full handshaking and still keep up with a data rate of 80 kSample/s needed for sampling a 40kHz signal. To solve this problem the computer is synchronised with the automatically delivered data flow. The computer now only has to read a byte from the printer port at the right moment and does not have to bother with starting the A/D converter and the multiplexing of 12 data bits to the two groups explained earlier in this chapter. The synchronisation is accomplished by accompanying the data with additional state information from the A/D converter. From the state information the computer can reconstruct the time when the valid data was on the registers and so can synchronise with the data flow delivered from the A/D converter.

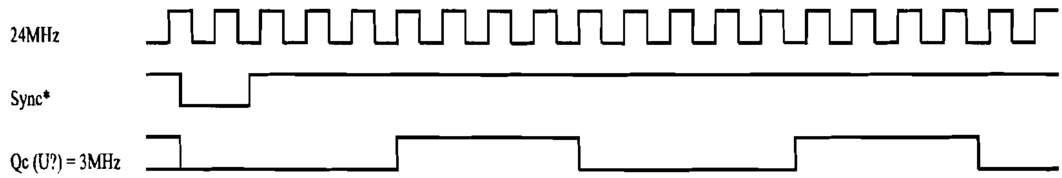


Figure 25

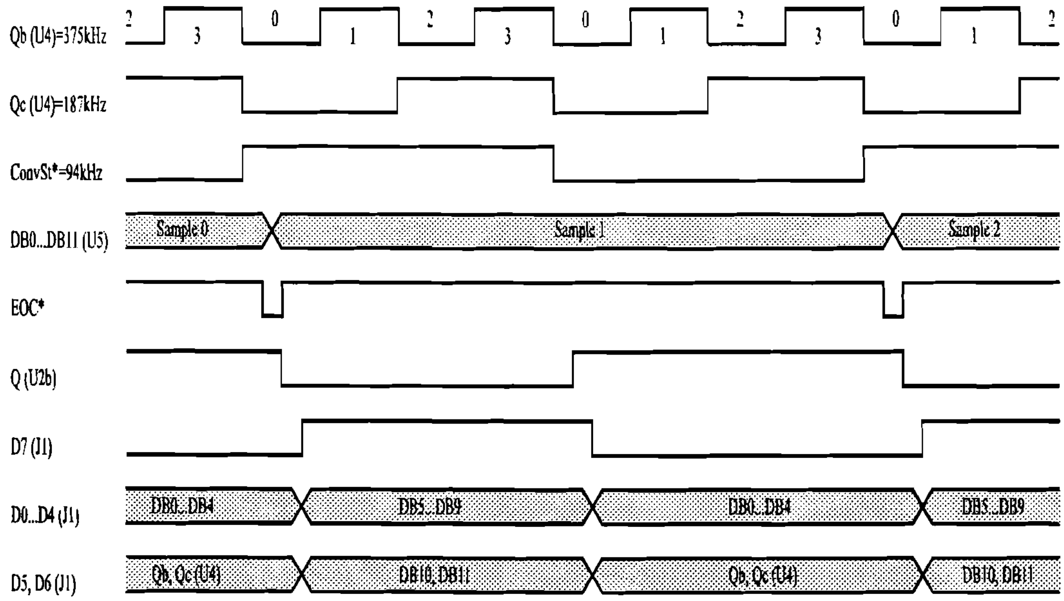


Figure 24

Circuit level description

The Counter (U?), flip-flop (U2B) and counter (U4) form a frequency divider that can be cleared by an external signal. The Qd output of the counter (U4) starts oscillating at a frequency of 94kHz as soon as the synchronisation from the ultrasound output block goes high. Because these dividers are reset with the synchronisation signal the 94kHz signal always has the same phase to the ultrasound pulse. This results in that the A/D converter samples the input signal always at the same time-offset to the sending of the pulse (resulting in a very steady signal).

Each time the A/D converter (U5) receives a rising signal on the ConvSt* input it starts converting the analogue input signal to the digital 12 bit representation. The output EOC* (End Of Conversion) pulses low as soon as the conversion is complete. The flip-flop U2A is used as an inverter of the ConvSt* signal (Figure 24). A normal inverter could have been used but this flip-flop was not used elsewhere and by using this flip-flop one extra IC is saved. The output of the flip-flop switches between the two output-buffers (U6 and U7).

9.3. Circuit board design

The equipment is divided into four distinct blocks. It is possible to integrate them all on one circuit board but that should make the equipment less flexible. Also the circuit board would get quite large with all the manufacturing problems associated with large boards like film accuracy and etching equipment size restrictions. The conclusion is that it is best to hold on to the four design parts.

To keep the boards as small as possible some SMD components are used. They also reduce wire lengths and so the picked up noise from external transmitters. An extra benefit of using SMD is that the number of drilling holes is largely reduced and so the time spent to drill them. A disadvantage is that when changes have to be made the de-soldering of SMD components is very difficult.

9.3.1. Output board

The output board in not very sensitive to noise and other disturbances. This means that no special precautions must be taken to guarantee good functioning of this board.

9.3.2. Input board

The input board is much more sensitive to picked-up disturbances than the output board due to the large amplification values. The large amplification values also make oscillation very likely on a badly designed board. Using very short wires and a large ground plane largely reduces the chances of oscillation. With the OpAmps it is very important not to route output wires along inputs of the same OpAmp. Best is to place a ground plane under the OpAmp but in the case of the input board that was not possible.

Cross talking of the separate input amplifiers occurs when they can 'see' each other electrically. To prevent this they are shielded from each other by large amounts of ground plane completely surrounding the amplifiers.

Each output from the amplifiers that goes to the input selector uses quite long tracks. These tracks carry relatively high currents and so they must be shielded in some way to avoid cross talking and oscillation problems. To limit these problems all long tracks are routed in a way that on the other side of the board there is a ground plane and all tracks cross each other rectangular.

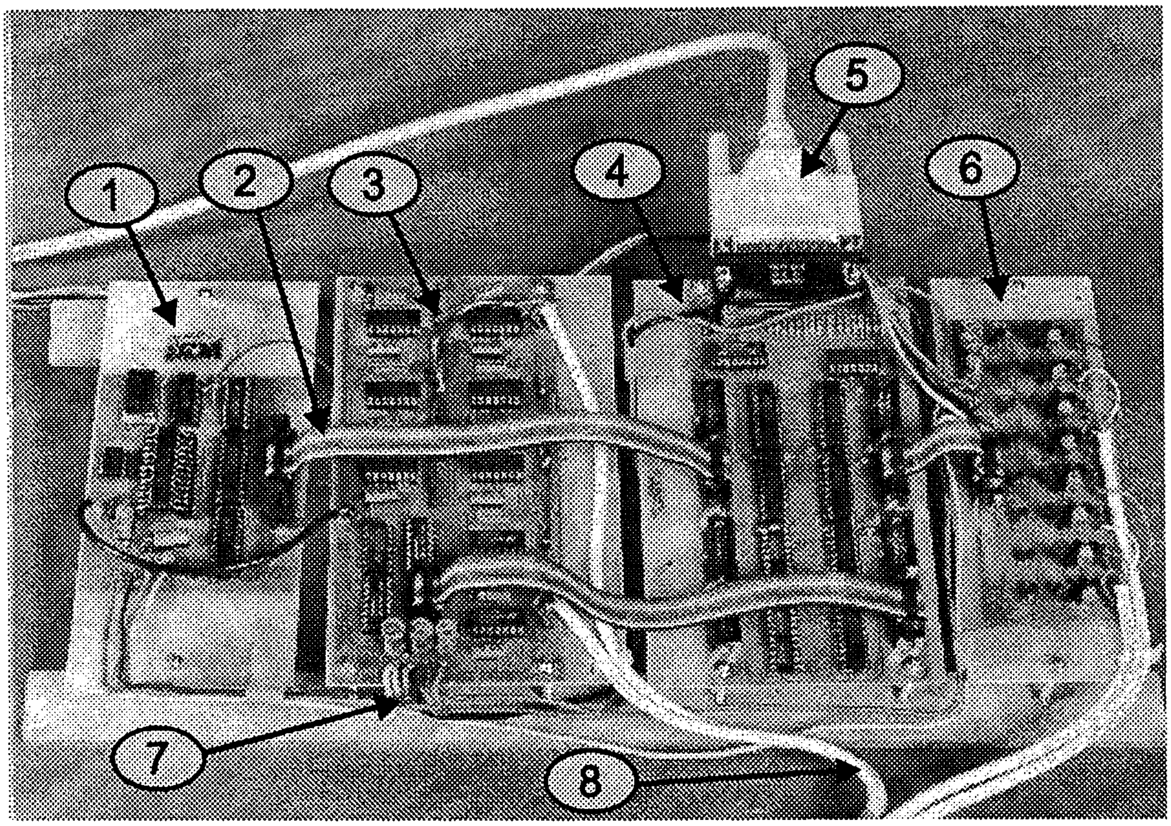
9.3.3. I/O board

Just like the output board there are no special considerations concerning the design of this circuit board. All signals are digital, so noise and disturbance is not very important.

9.3.4. A/D converter board

The A/D converter board houses a very sensitive A/D converter and so disturbances must be limited as much as possible. With minimal added distortion on this board the resulting accuracy is maximal.

The number of wires needed on this board is minimal and so the amount of ground plane can be large. This automatically highly limits the amount of distortion and noise. The large amount of ground plane is very important on this board because of the combination of high-frequent digital signals and analogue signals.



Description:

1:	*A/D converter board*
2:	*Flatcables interconnecting the different boards*
3:	*Input amplifiers board*
4:	*Input / Output board connecting the equipment to the computer*
5:	*Cable connecting the I/O board to the computer*
6:	*Output buffers board*
7:	*Power cabling*
8:	*Shielded cables connecting the boards to the sensor strip*

9.4. Mechanical construction

The problem with the transducers that are used with 40kHz is that the sound leaves the crystal and then first goes through 5mm of air (Fig. 26). So before entering the tissue the sound goes through some air and thus is reflected for 99% by the tissue. The 1% entering the tissue is for 99% reflected by the tissue-air interface at the other side. The 0.99% returning to the transducer is again reflected by the tissue-air interface before leaving the tissue to be recorded by the receiver. So the amount of energy received is about 0.0099% of the transmitted energy. The receiver is not sensitive enough to detect such low energy levels.

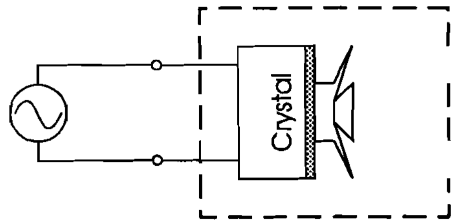


Figure 26

To overcome the problem of going two times through the air-tissue interface the transmitter and receiver are physically changed. The housing and funnel are removed (Fig. 27). Thus the layer of air is removed and a much higher level of energy is injected into the tissue. By also making use of special acoustic gel a large enough energy level is reached to be useful for measurements.

By using the modified transducer it is possible to send one pulse with a length of 25µs into the tissue and wait for the pulse to return after about 53µs or later if no lung is in it's path.

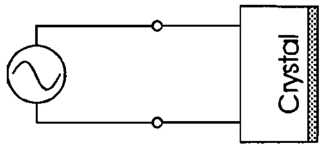


Figure 27

The eight ultrasound-transmitters and eight ultrasound-receivers must be mounted in such a way that it is easy to place them on a horse. At first it was tried with an elastic band that pushed the sensors against the skin of the horse. The problem was that the pressure on the sensors was much to low to receive any usable reflections. A way had to be found to apply enough pressure to all the sensors, preferably the same pressure to all sensors at once.

To do this all sensors are placed on an aluminium strip (Figure 29 and 28) and the pressure of each sensor is applied by separate springs. To keep the sensors from being ejected from the aluminium plate, a bubble of tin at the end of the wires locks them. Because each sensor is placed on a spring, each applies approximately the same pressure. Also each sensor moves with a certain amount of freedom to always make optimal contact with the skin. The aluminium strip is easily bendable to follow the shape of the thorax. To further improve the contact to the skin a special ultrasound conducting gel is used.

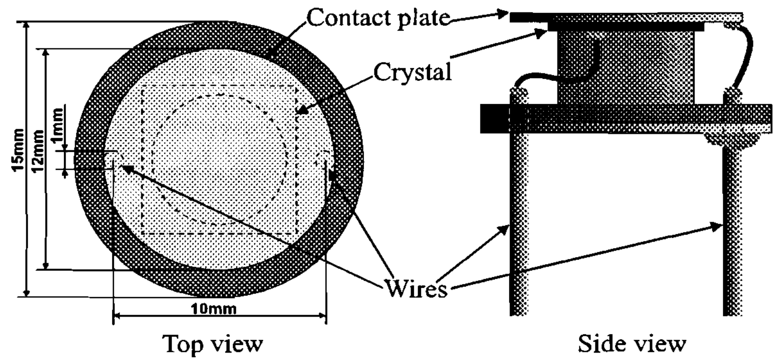


Figure 29

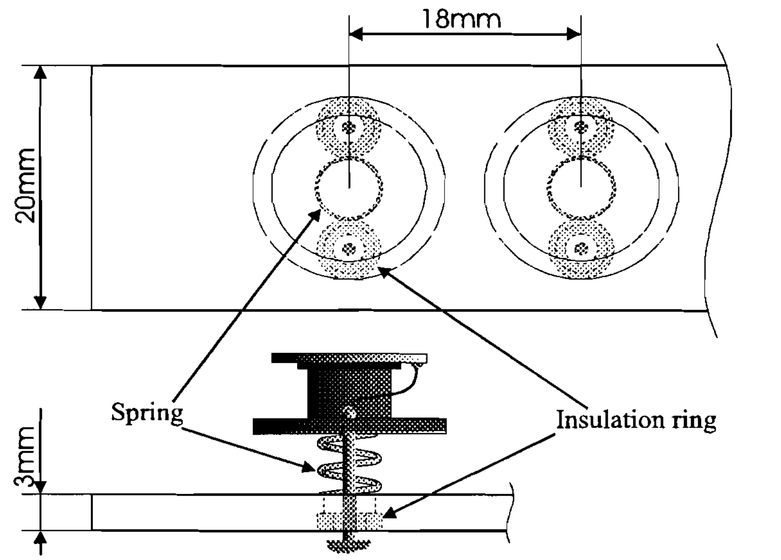
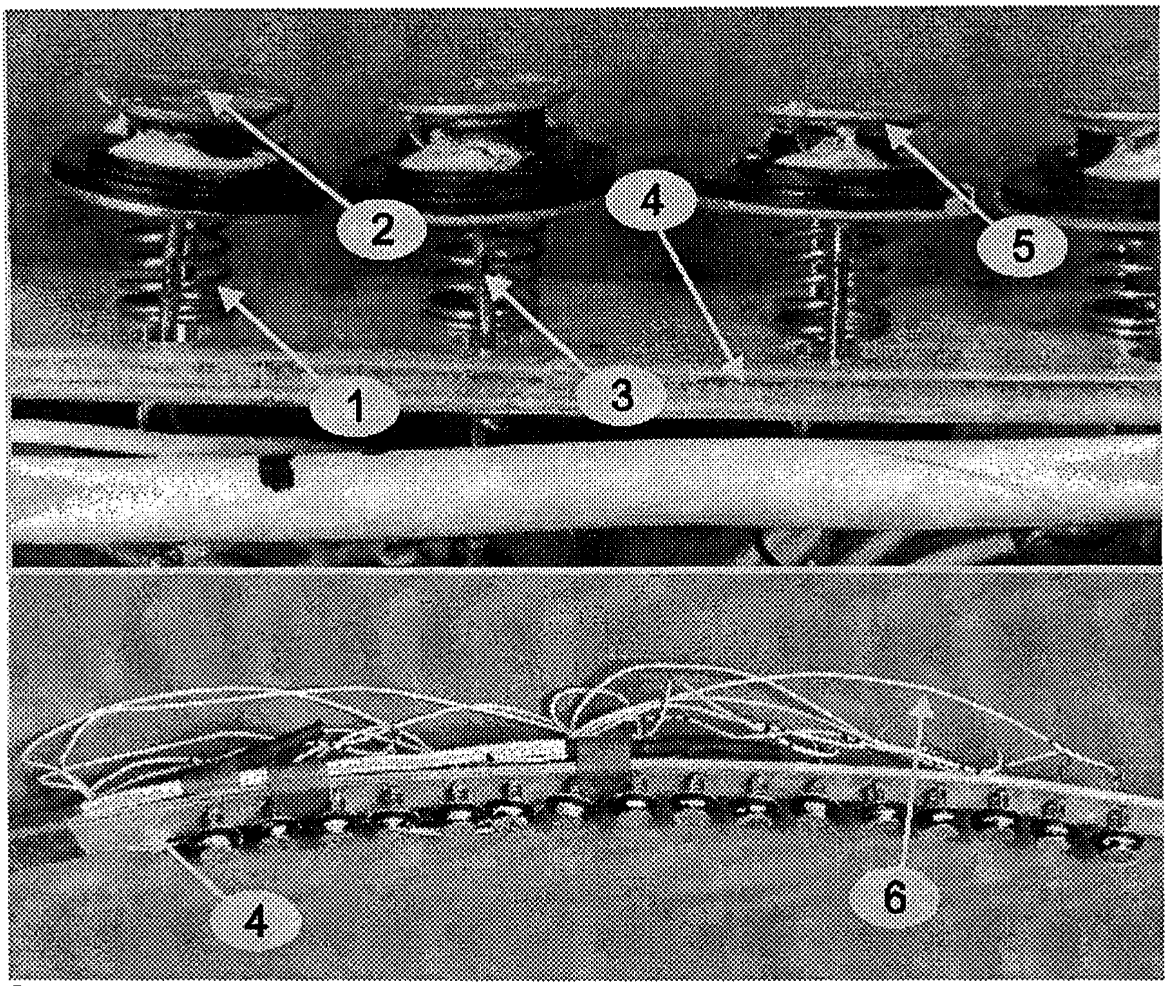


Figure 28



Description:

1:	*Spring to deliver equal pressure to all receivers and transmitters*
2:	*Contact plate between the crystal and the skin of the horse*
3:	*Signal wire connected to transducers*
4:	*Aluminium mounting strip*
5:	*Crystal that sends and receives the 40kHz ultrasound pulses*
6:	*Shielded signal wire connecting the transducers and the electronics*

9.5. Testing of the equipment

A program is written that continuously outputted random data to the registers and tried to read that data back through the input buffers. This program ran for a couple of hours without any read-back error. This meant that the I/O board was working properly.

Secondly the outputting of ultrasound pulses was tested. This gave some problems because the pinning of the FET's was different from that what my layout program thought. The result was some defective transistors and FET's. As soon as the defective components were replaced in the correct way the outputting of pulses worked like predicted.

The next block to be tested was the input block. Some oscillation problems were expected but did not arise. It did become clear that the wires from the ultrasound receivers had to be shielded up to the point where they entered the OpAmps. Otherwise they picked up a lot of distortion from the digital control wires. The input selection and amplification regulation turned out to work correct.

The last block to be tested was the A/D converter block. As with the rest of the equipment no problems arose and the design turned out to be ok.

From these tests the conclusion was drawn that the use of PSpice resulted in robust designs without the need for much modifications in the resulting prototypes. Additionally it must be mentioned that the designs were simple and so not much problems were expected, even without the use of PSpice.

10. Software

A complete description of the software does not fit in the goal of this report while the main purpose of this report is to explain the new method used to measure breath volumes. Just the most important parts of the software will be explained in full depth.

All software was written from scratch in Borland C that was combined with some Assembly code. No external libraries and no copied code fragments from other programmers were used. The code that handles the movement of the mouse and the menus was not especially written for this project. I wrote that part of the code a few years ago for another project.

10.1. Reading the A/D converter

Reading the A/D converter is quite challenging because of the special way the 12 bits are multiplexed and delivered to the printer port of the computer. The computer has to be exactly synchronised to the delivery rate of the A/D converter. Reading data is done by reading one byte, waiting, reading the second byte, waiting and start all over again (Figure 30). The waiting has to be done for a very accurate amount of time. Because not every computer has the same computing speed, the delays have to be optimised by the software. The delay can never be exactly correct and the data rate from the A/D converter also shifts a little. Using three different lengths of delay compensates for these inaccuracies. The delay lengths are short, normal and long which are used depending on the state the A/D converter was in when the data was read by the computer. Masking the state bits from the rest of the byte just read results in the state. Bit D7 always must be 0 because this indicates the correct part of the multiplexed data is read from the data bus. D5 and D6 combined forms the state counter. From the timing diagram in a previous chapter it is concluded that D6 must be 1 and D5 must be 0 at the moment the byte is read from the port. In all other cases the program is too early or too late.

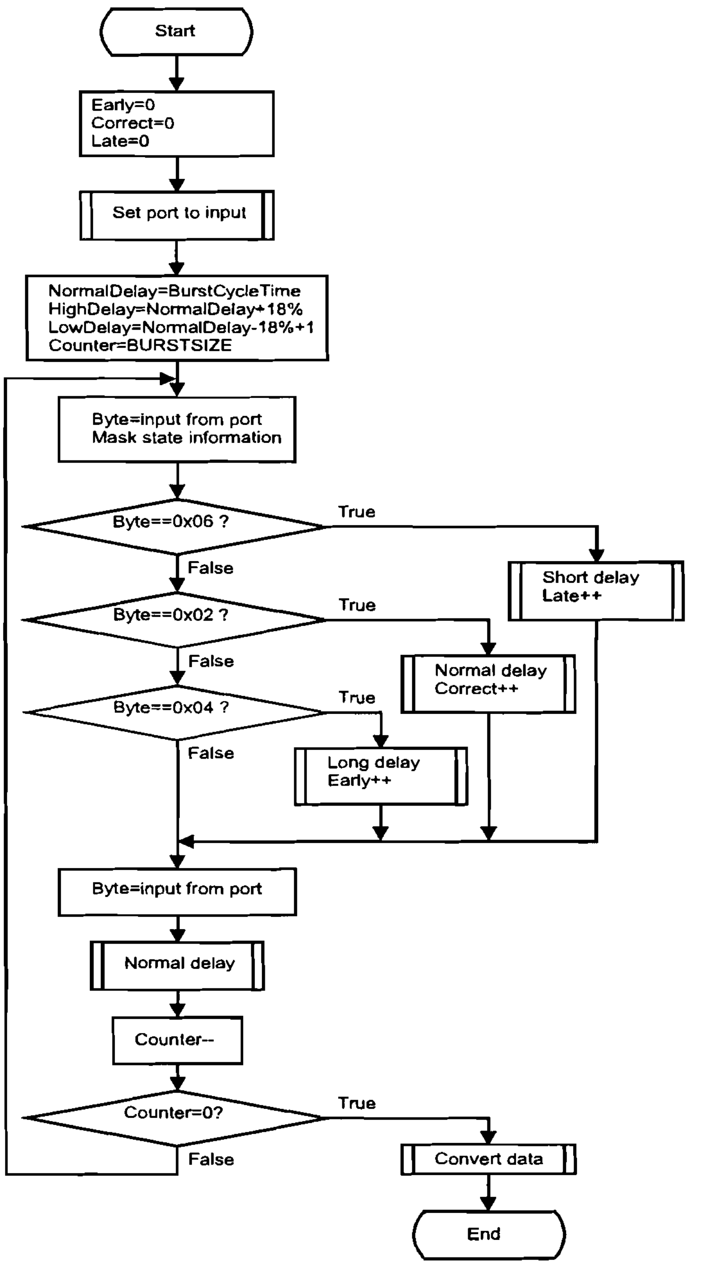


Figure 30: BurstInput()

To be able to use this same routine for synchronisation the accuracy of the length of delay is monitored. Each time a choice of delay length is made the corresponding counter is incremented by one. The result is that at the end of the burst of reads they indicate the accuracy of delay length used. In optimal situations about 80% of all samples must be exactly correct timed. The remaining 20% of samples that are sampled a little too early or too late form no problem. Even the early and late samples carry the correct data. The first few samples always are incorrect because the software needs some samples to synchronise. Typically 3 samples are needed.

10.2. *Initialisation of the delay length*

The delay length has to be initialised by the software to compensate for the different computing speeds of the computers used. In case of just one computer the software could be optimised for that computer and the code would be a lot simpler. But because the software has to function on a multitude of computers the optimisation has to be done by the software.
The A/D converter reading software checks the accuracy of the delay used. These results are used by the optimising routine to find the optimal delay length. The optimising routine just checks a lot of delay lengths and stores the best one.

10.3. *Displaying the data*

The data in the buffer returned by the BurstInput() routine has to be converted to information that carries the valuable information. This is done by taking the average of several consecutive points in one data burst and also averaging them with previously read data at the same moment. This can be seen as horizontal and in-depth averaging. Where horizontal indicates the consecutive samples and the in-depth dimension stores previous samples of the same moment but in previous bursts. Taking the average of the data slows down the shown information in following the signal strength from the receiver. This means that rapid changes become invisible if too many samples are averaged. When too few samples are averaged the shown data will follow each change in measured signal rapidly and soon becomes useless. An optimum must be found between speed and usability of the resulting shown data. This is very dependent on a lot of external factors and must be continuously changeable by the operator of the program.

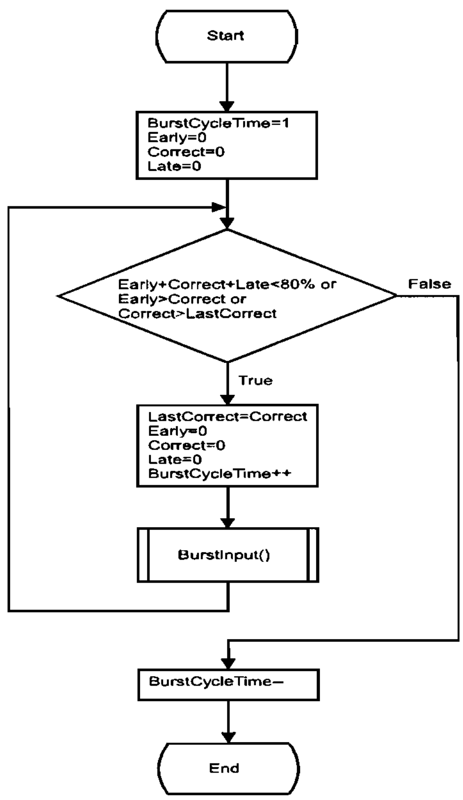


Figure 31: InitializeBurstInput()

The in-depth averaging is actually not real averaging, instead the formula used takes one previous sample and not multiple previous samples to average. The formula used is $YVal = YVal + PrevVal \cdot Weight$. After this formula is executed the result is stored back in YVal and is used by the next round where it is used as PrevVal. This means that even extremely old values are used in the resulting value and changes are followed very slowly with large values of the Weight value. The result of this way of averaging looks like low-pass filtering.

Two modes for displaying the data are possible. One mode is used for normal operation and one additional mode that only is used for testing purposes. The normal mode shows the eight input channels together on one screen. Each input channel is displayed at a different vertical offset. At the left side of each channel the amplification of the input amplifier is shown that is used when the input selector selects that channel. This makes it possible to compensate for the large differences in signal strengths between the receivers.
The test mode shows one input signal where no calculations have been performed on. This can be used to optimise the amplification of that channel and to check the functioning of the equipment and software. In this mode also the accuracy counters are shown in the menu bar, left of the author's name. In the shown

example (Figure 32) all counters are 0 because the shown data is read from disk where it was stored during a previous session.

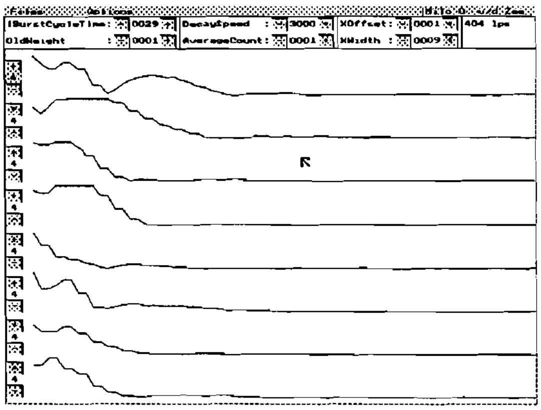


Figure 33: Normal mode

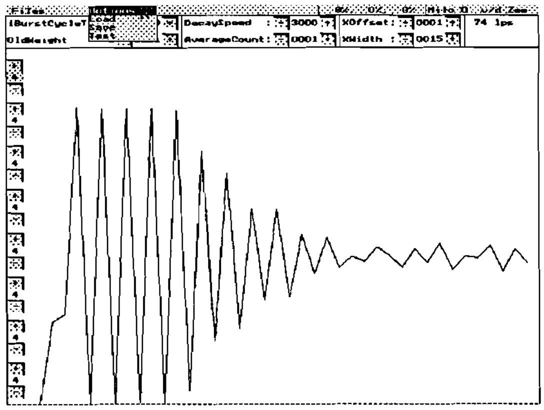


Figure 32: Testing mode

In both modes the speed of displaying is shown in the upper right corner beneath the author's name. This speed indicates the number of Lines Per Second that is displayed on the screen. In real time operation this value is limited by the speed at which the A/D converter delivers the data to the computer. With a burst size of 100 samples the maximal speed would be $\frac{94 \cdot 10^3}{100} = 940 lps$. As seen in the example this speed is even in off-line mode by far not approached. This difference comes from the slow down of the computer resulting from displaying the lines on the slow video screen through a Video-Graphics-Array adapter. VGA adapters are very slow in comparison to the rest of the computer and as soon as they get used everything has to wait. Though it should be noted that the speed is high enough to accurately display the input signal with a high enough refreshing rate and no speed increment is necessary.

11. Experiments

Several experiments were conducted. Starting with my own heart to check the software and ending with tests on horses.

11.1. My own heart

The tests on my self were very promising. A clear picture of moving tissue appeared on the computer screen.

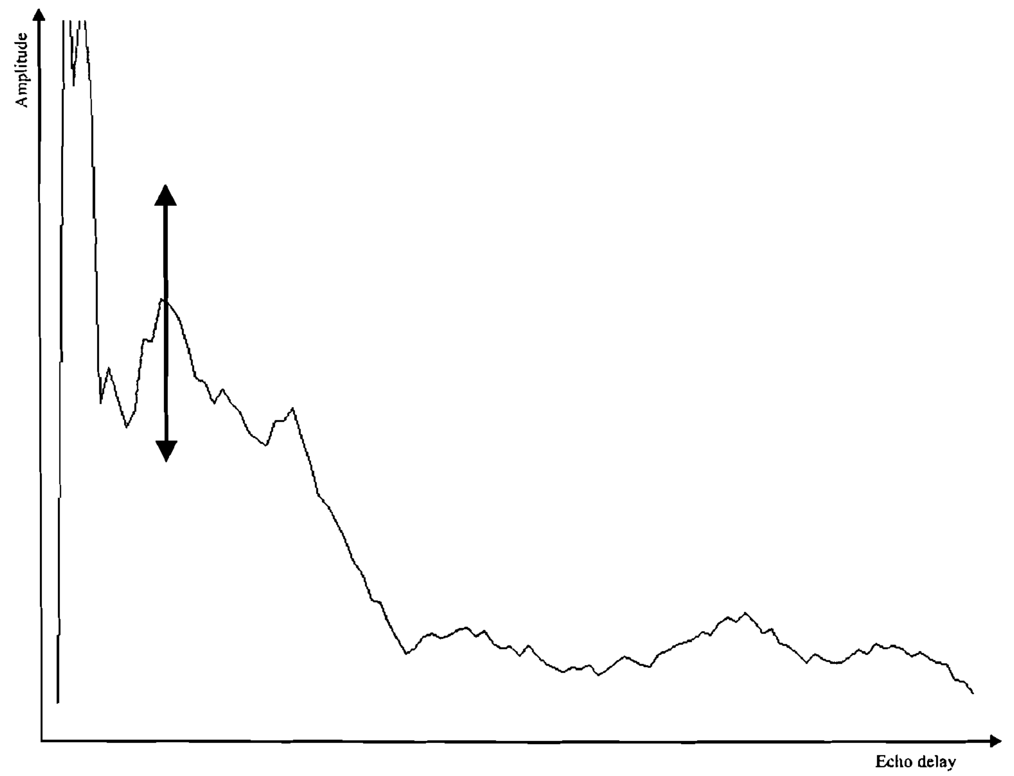


Figure 34

Figure 34 shows the result of this test. Horizontal is the delay from sending the ultrasound pulse into the tissue, and vertical the resulting received signal amplitude. The high peak at the left is the directly received transmission of the pulse itself. The next peak shows, on the real time computer screen, the changing amplitude of the received ultrasound pulse that returned from the heart chambers. The amplitude of the signal at this time offset follows the human heart rate. Even the different heart chambers can be seen in the form of two pulses following each cycle. This use is not the purpose of this equipment but it sure gives an idea of the possibilities.

11.2. Horse lungs

After the promising results obtained from my own heart the equipment was tested on horses for several times.

Date	*Horse*	*Test run*	*Results*
December 3,1997	Bram		Tested with ultrasound of 5MHz. The lung margin is clearly visible. The horse was shaven very thoroughly.
April 1, 1997	Haspel		The first test with 40kHz ultrasound. The pulses were sent with a generator and the received signal was directly measured by a scope. A lot of noise was visible but no lung-margin.
June 18, 1997	Godelief	1.1 and 1.2	Amplification of the input amplifiers is to low.
June 19, 1997	Godelief	2.1	Amplification is correct but there still is very little to see.
		2.2	
		2.3	
		2.4	By pressing the 7th sensor with extra force to the thorax the lung margin is visible.
July 8, 1997	Bram	3.1	A lot of reflected signal is visible with the new sensor strip, but still not a very clear lung margin.
		3.2	By pressing one of the sensors with extra force against the skin it was possible to follow the lung margin on the computer screen.
September 29,1997	Godelief	4.1	The thorax was shaven for optimal contact between sensors and skin. Two testmethods are used. The normal 5MHz and my 40kHz method.
		4.2	
		4.3	The sensor array is used rectangular to the ribs.
		4.4	Both methods are used in one test run.

Test runs 1.1 until 2.4 were done with the elastic band holding the sensors. Test runs 3.1 and later were done with the aluminium strip and resulted in better results due to the better contact between sensor and skin.

11.2.1. Test runs 1.x

This test was the first one done on a horse. During this test an elastic band was used to press the sensors against the skin of the thorax. It was very easy to place the sensors between the ribs because the band gave all the sensors a high level of space to move. And because the elastic band the sensors move freely it was extra easy to feel the ribs for locating the sensor exactly between two ribs.
After placing the sensors at the right location the computer screen showed very little signal. This came because the equipment was optimised during tests done on the human heart. During those tests the contact between sensor and skin was optimal because of the absence of a fur on the human thorax. This resulted in high signal strengths and the equipment amplification was optimised for those signal levels. The fur of the horse's thorax absorbs a considerable amount of the signal and the resulting signal strengths are much less then during tests on human skin.

11.2.2. Test runs 2.x

During these tests the amplification of the input amplifiers was raised to levels suitable for use with horses. Still the lung margin was not visible on the computer screen. Only by applying additional force to a transmitter/receiver combination the underlying lung margin became visible. This indicated that the force applied to the sensors by the elastic band was too little. After closer inspection of the elastic band with the sensors it became clear that only the outermost sensors got enough pressure to function correctly. The inner sensors did not get any pressure because the outermost sensors lifted the band from the skin.

11.2.3. Test runs 3.x

From the two previous tests it was concluded that the pressure to the sensors must be enlarged. This was done by the aluminium strip where each sensors has it's own spring to apply the same pressure to all

sensors. The result of this new construction was a much better signal of all sensors and the ability to evenly distribute the pressure to all sensors by bending the aluminium strip to the contours of the thorax.
Still the results were not satisfactory. With this aluminium strip it was very difficult to place all the sensors exactly between the ribs and there constantly were some sensors that shifted to the wrong location. Even with the sensors that were at the correct spot the signal was not very useful. Even the slightest movement of the sensor gave enormous signal changes picked up by the receiver.

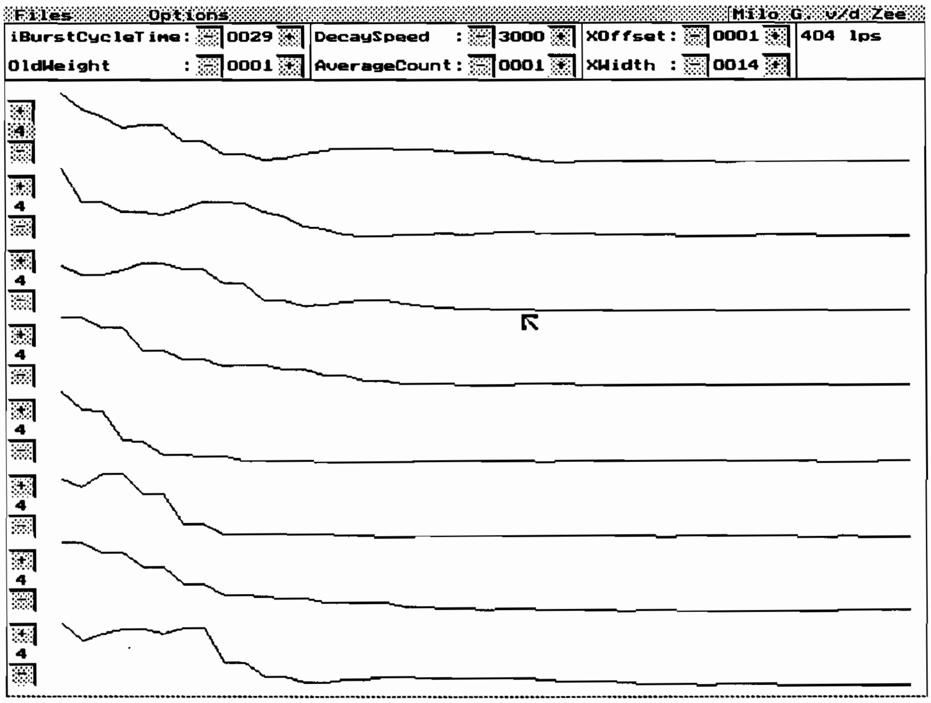


Figure 35

In figure 35 the results are shown when the third sensor from the top is manually pressed against the skin. From the moving picture on the computer screen it was possible to see the lung margin pass the sensor. It was immidiatly seen that the horse does not breathe constantly like humans. Humans breath around a relatively constant average lung volume. That results in constant lung movements. With horses that is totally different. A horse sometimes takes one deep breath and then only expires and inspires very little air from that average. After a while the horse expires some more air and from there starts breathing small amounts of air.

11.2.4. Test runs 4.x

During these tests all the possible was done to optimise the results. A very skinny horse was taken to easily locate the ribs. This horse was shaven at the location where the sensors where placed on the skin to get optimal contact between transducer and the skin. Especially the shaving of the horse was in contradiction with the original goals of the project where shaving was out of the question. But because all previous tests were very disappointing we wanted to exactly locate the problems. From these results it should be possible to enhance the circuits to overcome those located problems.

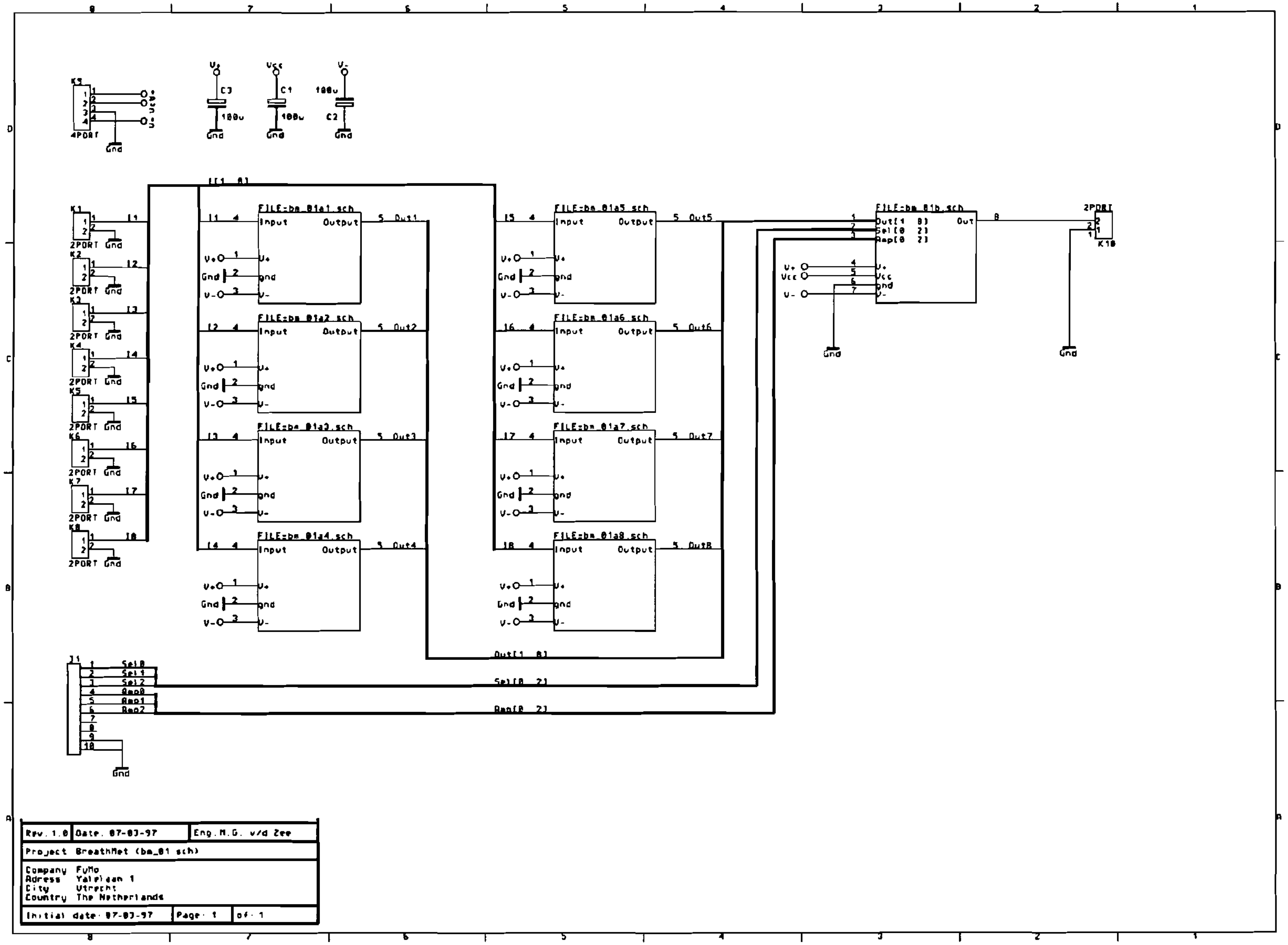
First the location of the lung margin was exactly located with the 5MHz ultrasound method. Then with the 40kHz method it was tested if this new method also sees the same lung margin. The result was that the lung margin was visible with the new method. This means that the lung margin is seen by the array. One previously found problem came across again. The tested horses breathe in a very irragular way. The frequency of breathing changes very often and the average lung volume also shifts considerable.

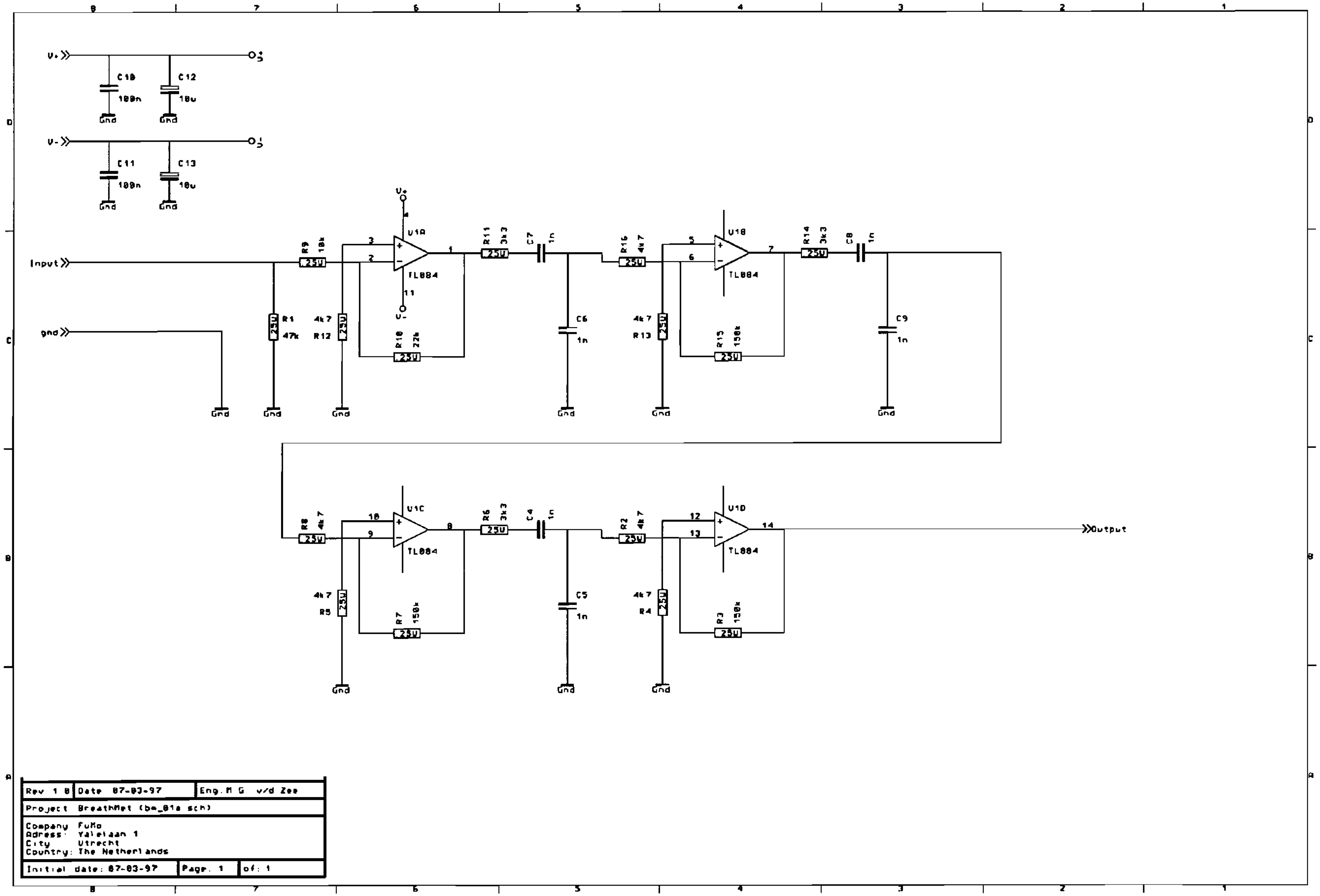
12. Conclusions and recommendations

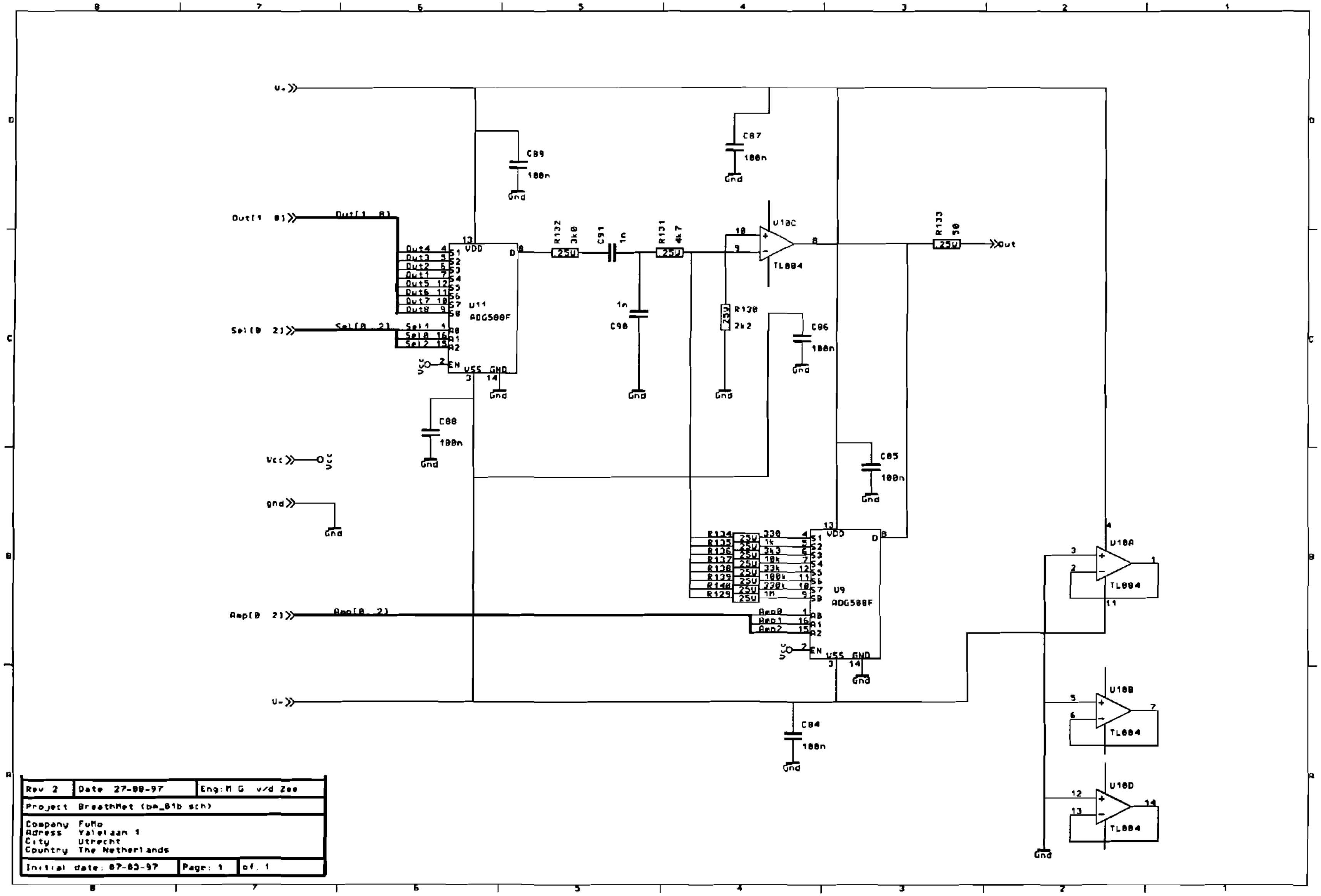
Two kinds of conclusions can be drawn from the project. One kind concerning the research and one kind of conclusions from the designs and tests.
Concerning the research the conclusion is that a lot of methods do exist to measure the breathing volume of humans. A small amount of those methods is suitable for horses. All of the used methods used on horses make use of some kind of breathing mask and so all obstruct the airways. Tests done by others indicate that these breathing masks change the breathing pattern and make some horses very restless. We concluded that a new non-obstructive method is very welcome.
From my own tests with ultrasound of 40kHz I conclude that it is going to be very difficult to invent a new method that does not obstruct the airways of the horse. The method tested is not suitable to measure the breathing volume with horses. It is hardly possible to measure the breathing frequency with this method. The method is very sensitive to the movement of the sensors and the lung-margins move much too slow and irregular. The irregular movement makes it difficult to locate the lung-margins in the noisy signal. The slow movement makes it virtually impossible to distinguish the signal changes due to the moving lung margin from the artefacts due to horse movement. Every movement of the horse transforms to a huge signal sweep in the measured signal.
It probably can only be made to work when the demands on the equipment are relaxed. Chances of success are much higher if the horse is shaven and if the sensors are tightly fixed to the horse with for example screws or glue.
From the tests done on my own body the results were much better. The heart was clearly vissible. With humans the use of low frequency ultrasound results in very cheap equipment. The usability of the results from the use of low frequency ultrasound could be worth researching.

Appendix A: Schematics

Input block

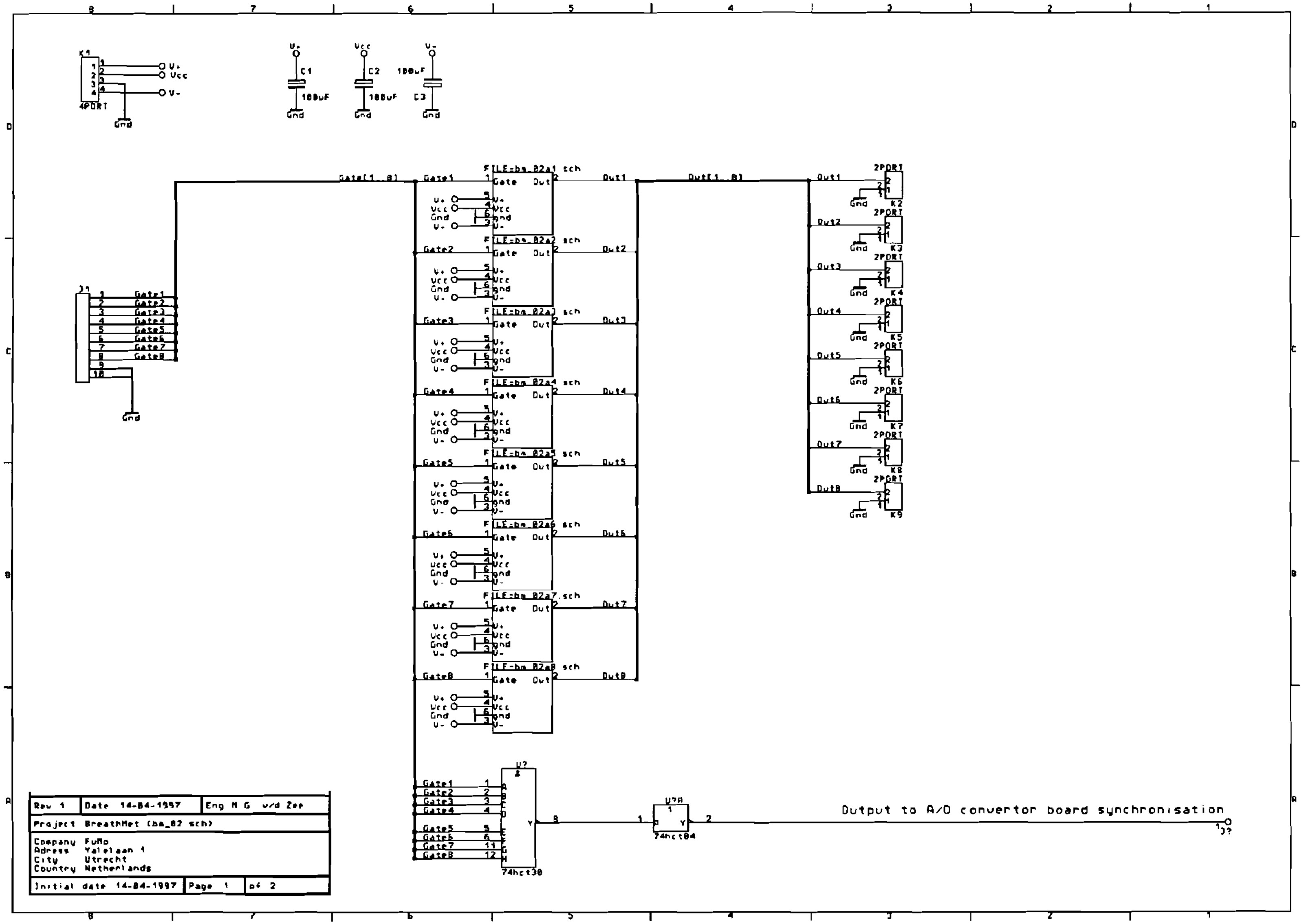


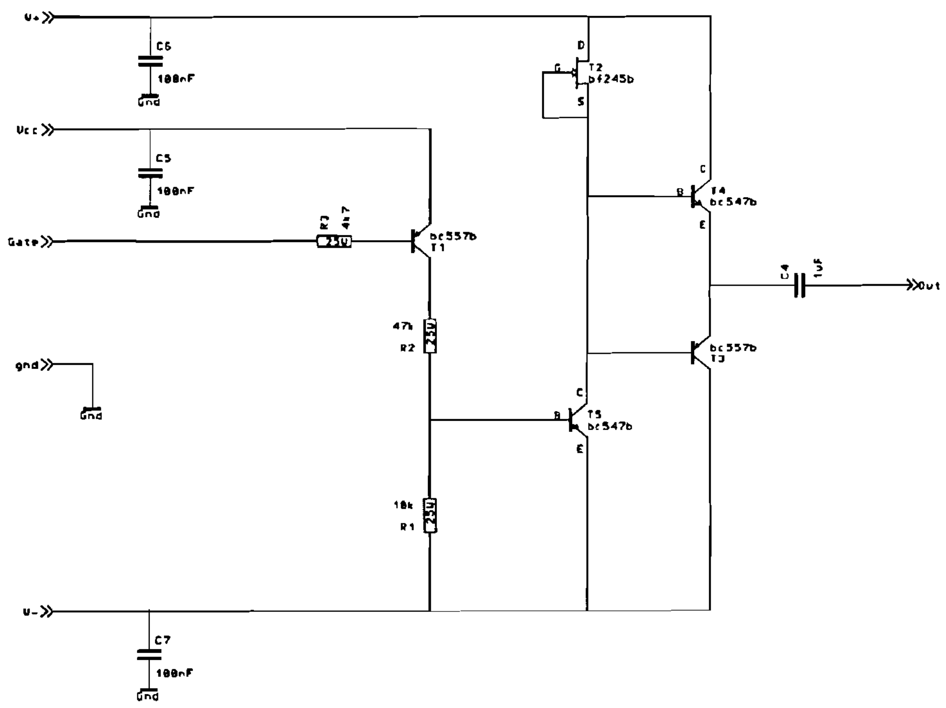

V+
C10 100n
C12 10u
V-
C11 100n
C13 10u
Gnd
Input
gnd
R9 10k
R1 47k
4k7 R12
U1A
TL084
R10 22k
R11 3k3
C7 1n
C6 1n
R16 4k7
U1B
TL084
4k7 R13
R15 150k
R14 3k3
C8 1n
C9 1n
R8 4k7
U1C
TL084
4k7 R5
R7 150k
R6 3k3
C4 1n
C5 1n
R2 4k7
U1D
TL084
4k7 R4
R3 150k
Output
Rev 1.0
Date 07-03-97
Eng. M.G. v/d Zee
Project BreathMet (bm_01a.sch)
Company FuMo
Adress: Yalelaan 1
City Utrecht
Country: The Netherlands
Initial date: 07-03-97
Page. 1
of: 1



V+
Out[1..8]
Sel[0..2]
Vcc
gnd
Amp[0..2]
V-
Out
U11
ADG508F
U9
ADG508F
U10C
TL084
U10A
U10B
U10D
C89 100n
C87 100n
C88 100n
C86 100n
C85 100n
C84 100n
C90 1n
C91 1n
R132 3k0
R131 4k7
R130 2k2
R133 50
R134 330
R135 1k
R136 3k3
R137 10k
R138 33k
R139 100k
R140 220k
R129 1M
Gnd
Rev 2
Date 27-08-97
Eng: M G v/d Zee
Project BreathMet (bm_01b sch)
Company FuMo
Adress Valetaan 1
City Utrecht
Country The Netherlands
Initial date: 07-03-97
Page: 1
of. 1

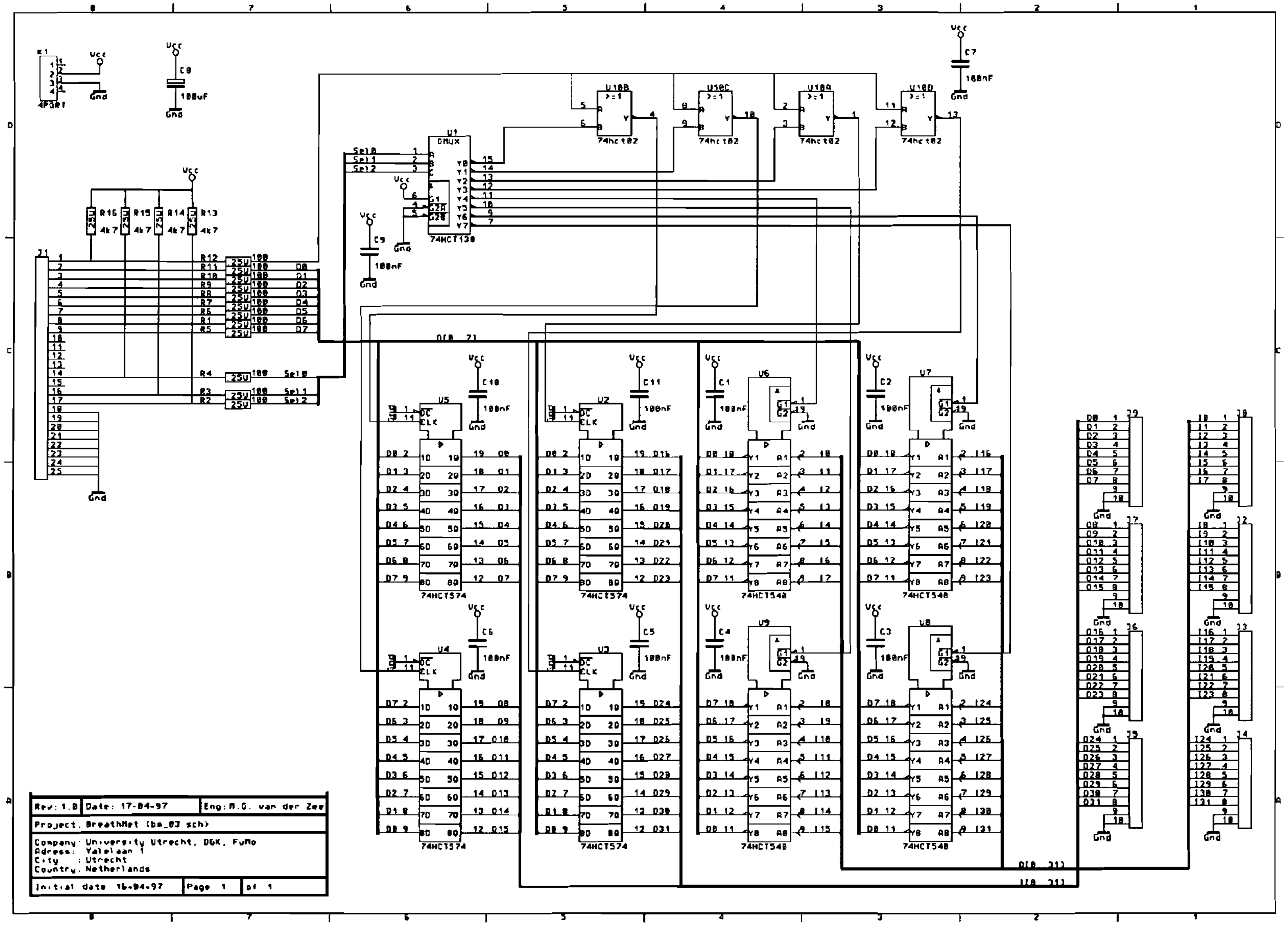
Output block



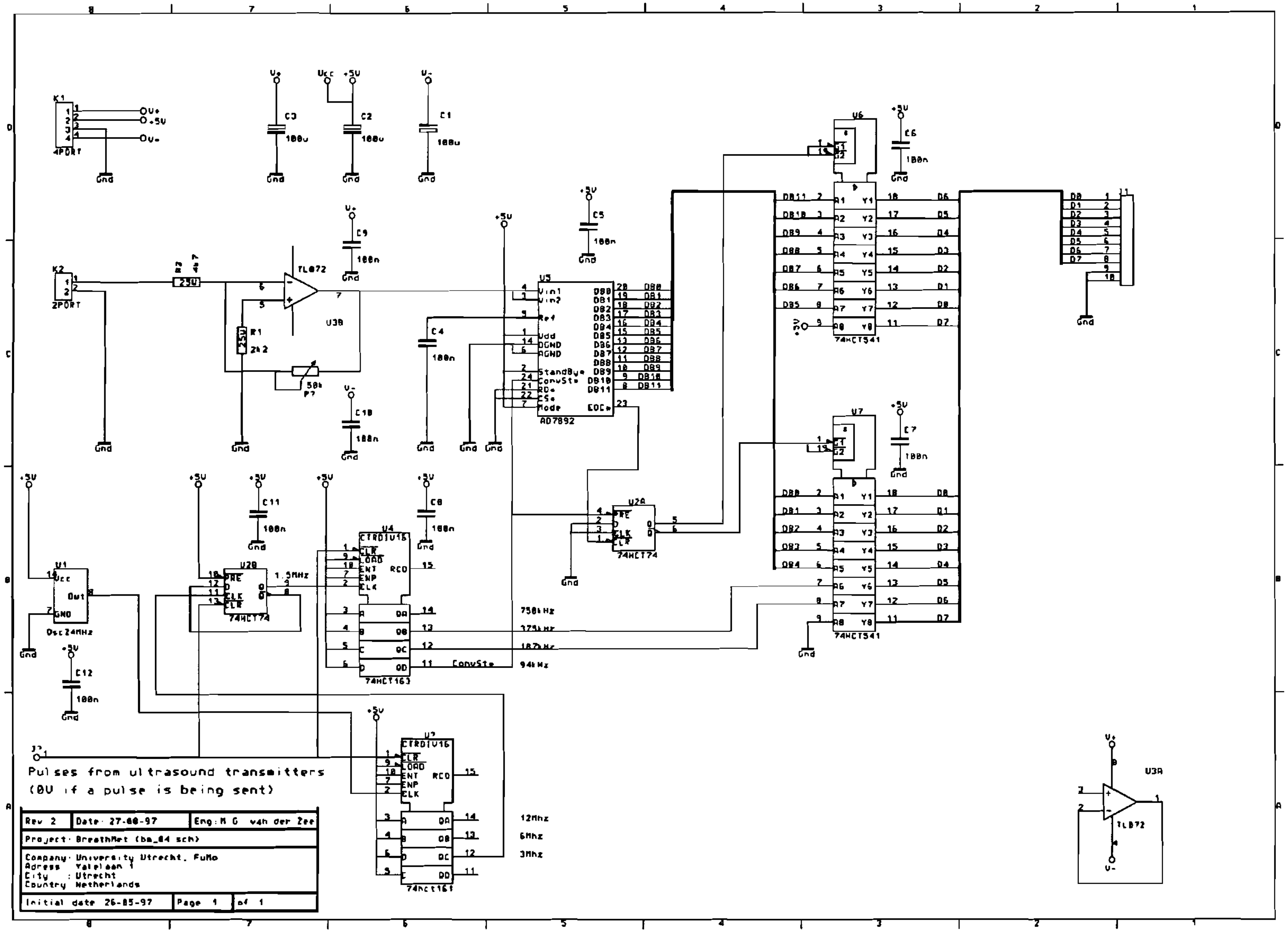


Rev. 1	Date 14-04-1997	Eng M G v/d Zee
Project. BreathMet (bm_07a? sch)		
Company: FuMo Adress Yalelaan 1 City Utrecht Country: Netherlands		
Initial date: 14-04-1997	Page: 2	of: 2

I/O block


Rev: 1.0
Date: 17-04-97
Eng: A.G. van der Zee
Project: BreathNet (bn_03 sch)
Company: University Utrecht, DGK, FuMo
Adress: Yalelaan 1
City : Utrecht
Country: Netherlands
Initial date 16-04-97
Page 1 of 1
74HCT138
74HCT574
74HCT540
74hct02

A/D converter block



Appendix B: Circuit board layouts

Input block

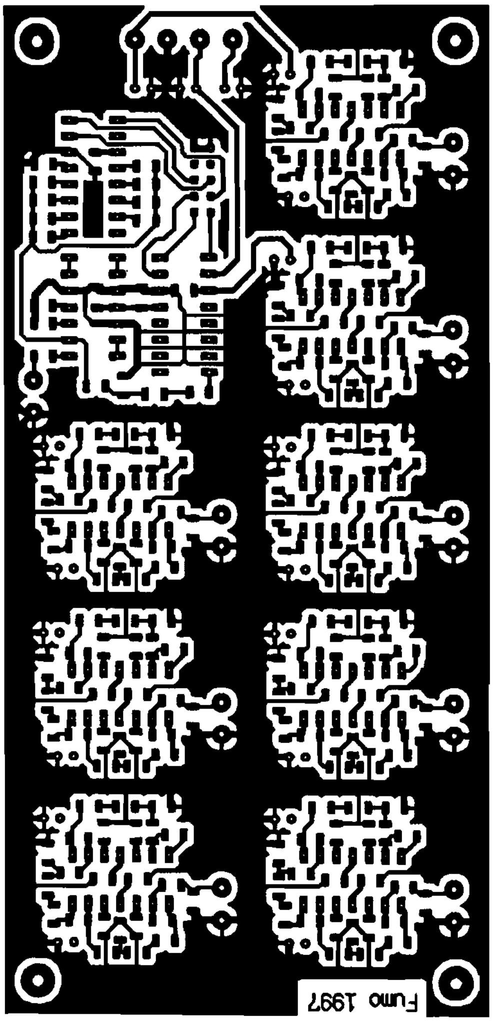


Figure 37: Input board, bottom layer

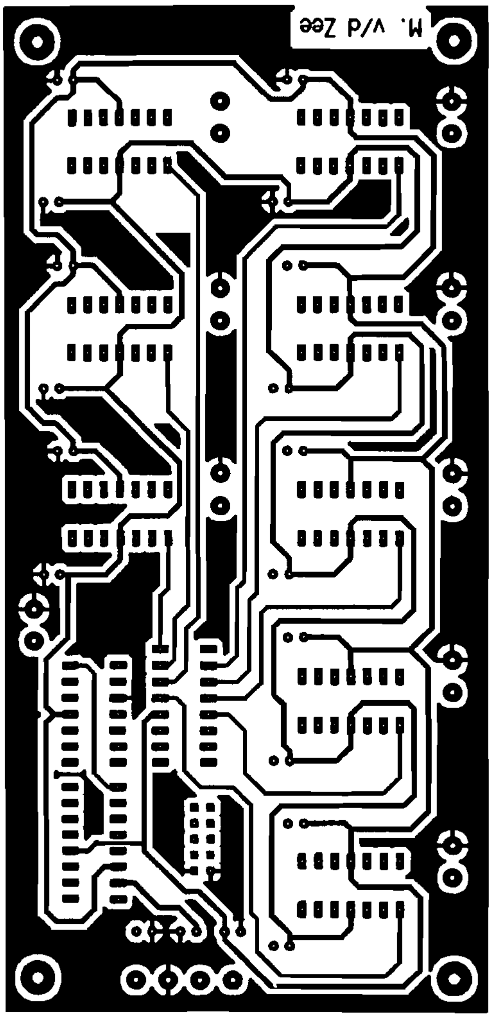


Figure 36: Input board, top layer

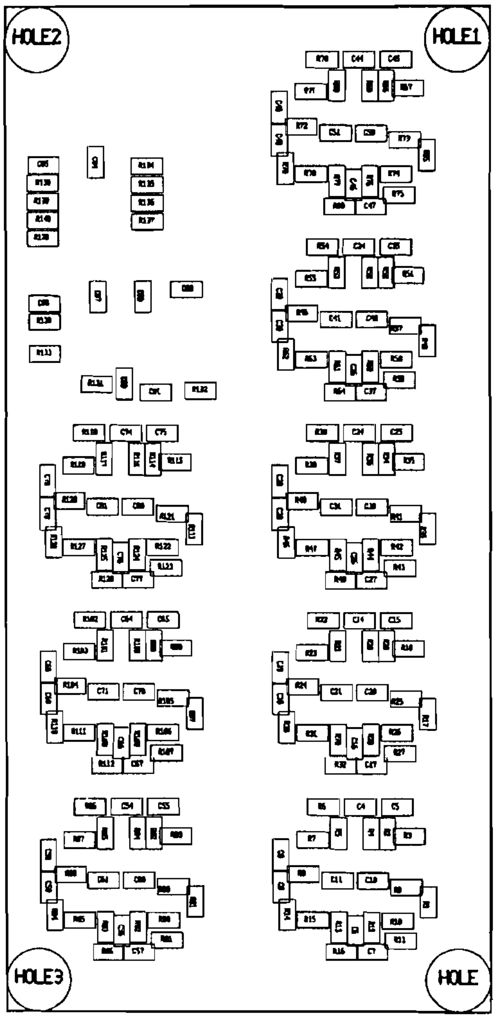


Figure 39: Input board components, bottom side

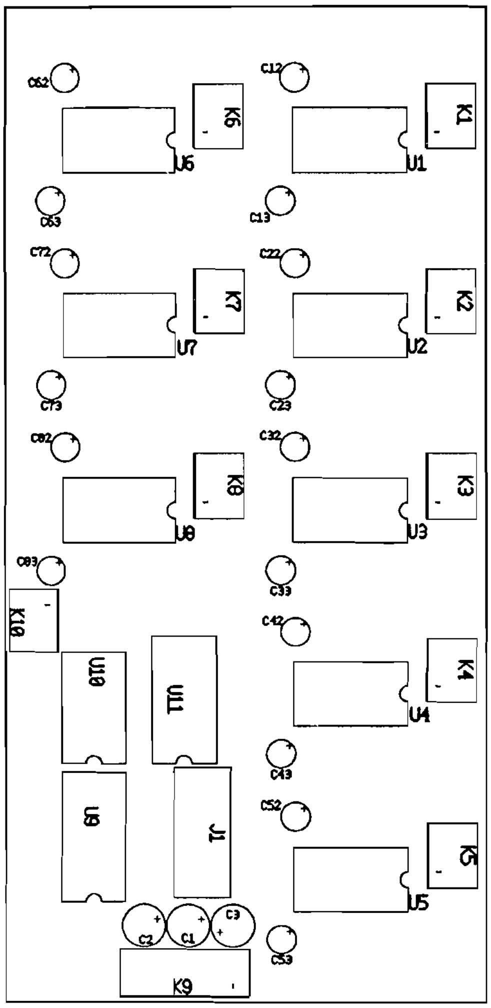


Figure 38: Input board components, top side

Output block

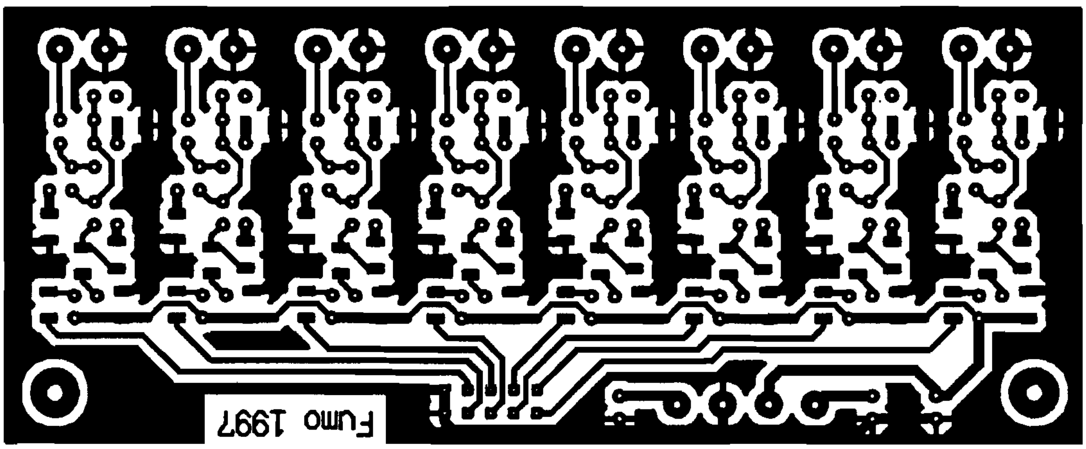


Figure 40: Output board, bottom layer

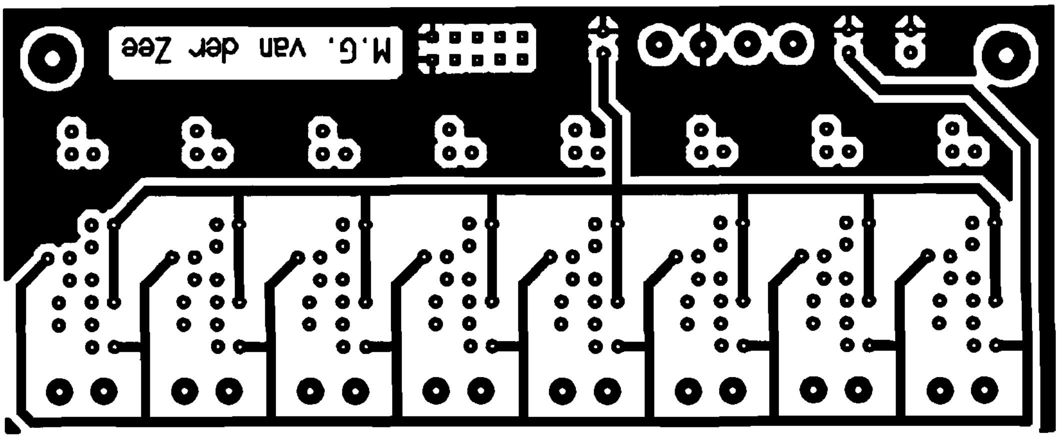


Figure 41: Output board, top layer

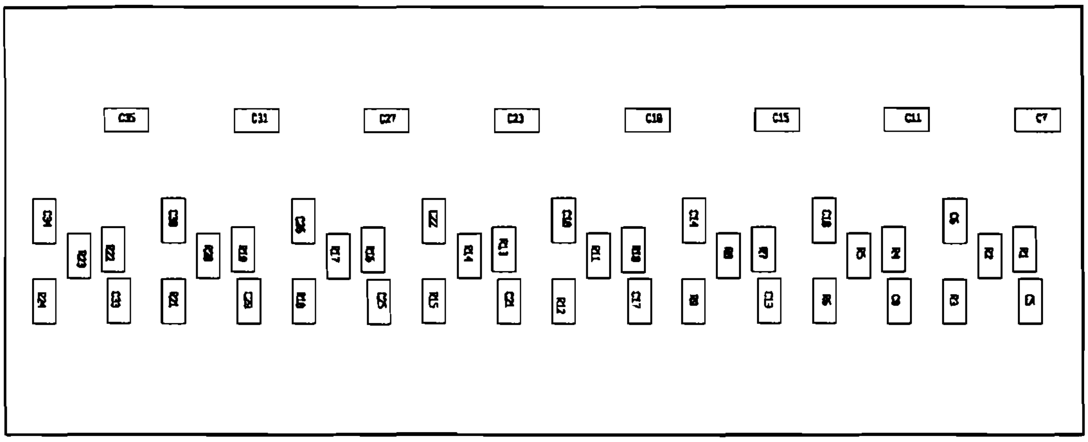


Figure 43: Output board, components bottom side

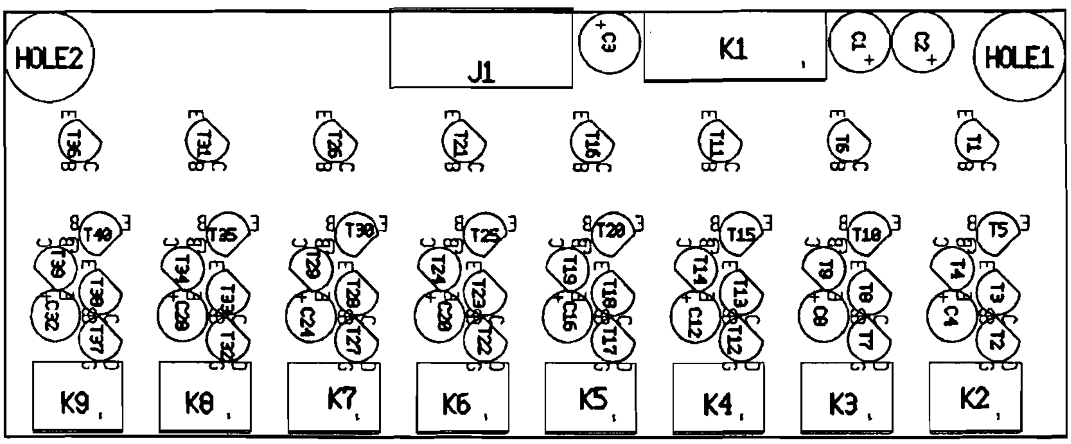


Figure 42: Output board, components top side

I/O block

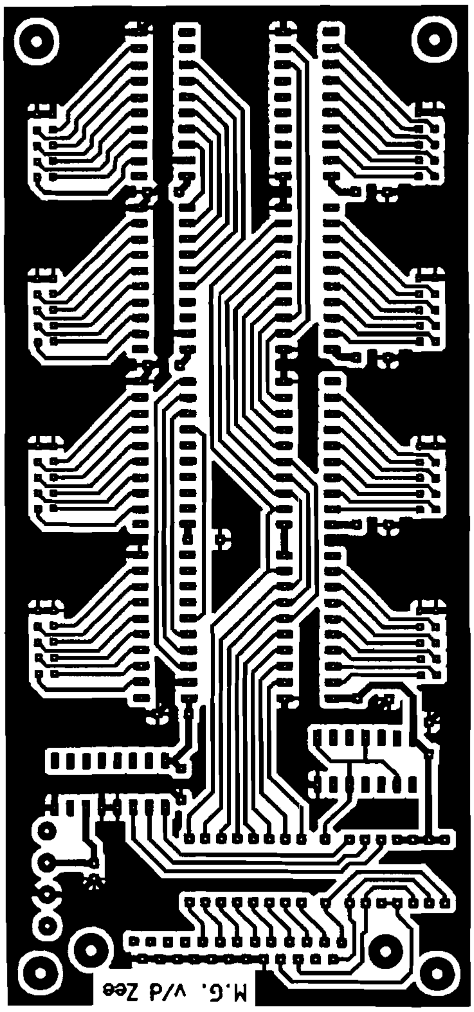


Figure 44: I/O board, bottom layer

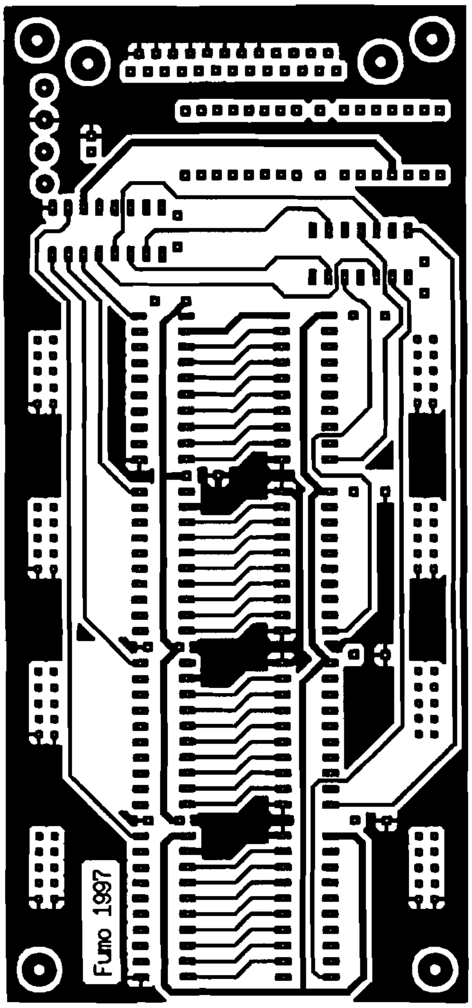


Figure 45: I/O board, top layer

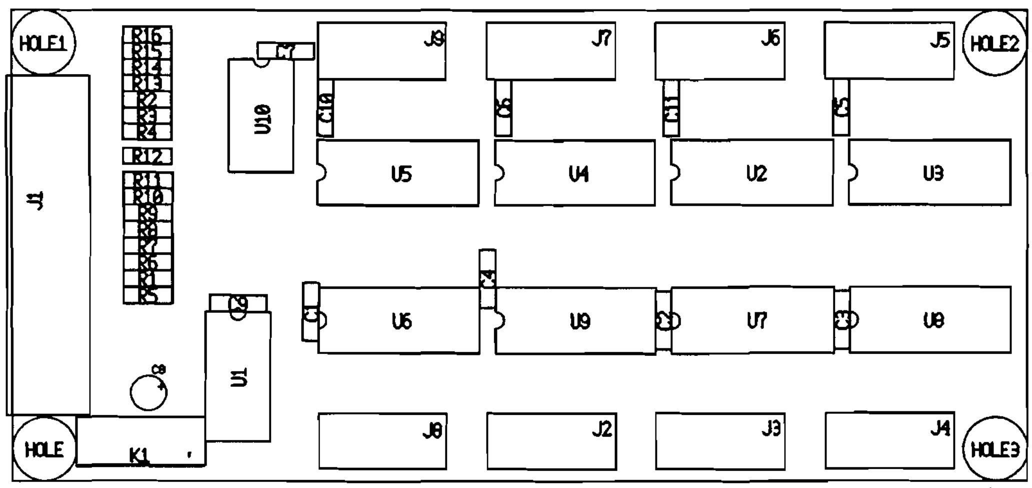


Figure 46: I/O board, components top side

A/D converter block

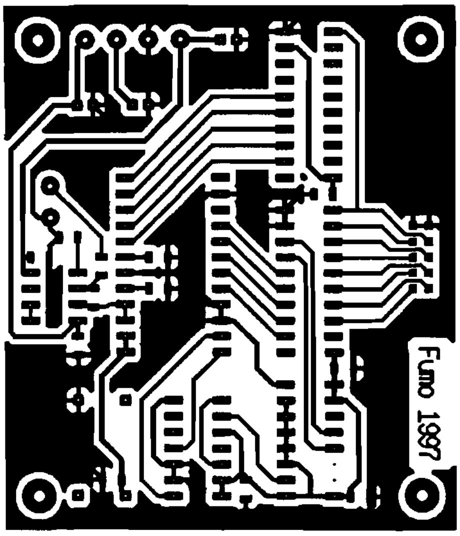


Figure 50: A/D board, bottom layer

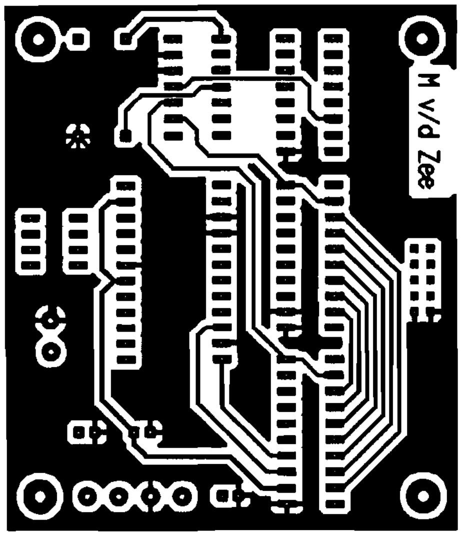


Figure 49: A/D board, top layer

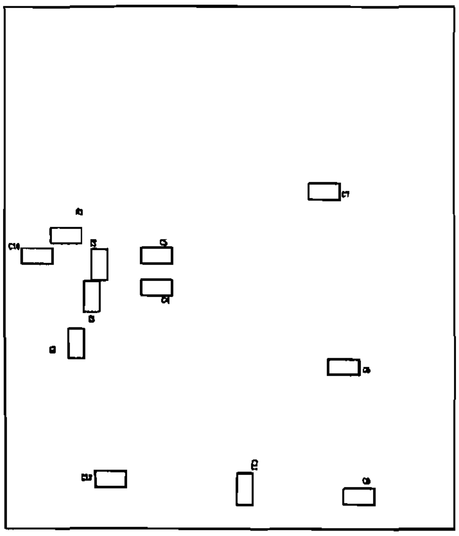


Figure 48: A/D board, components bottom side

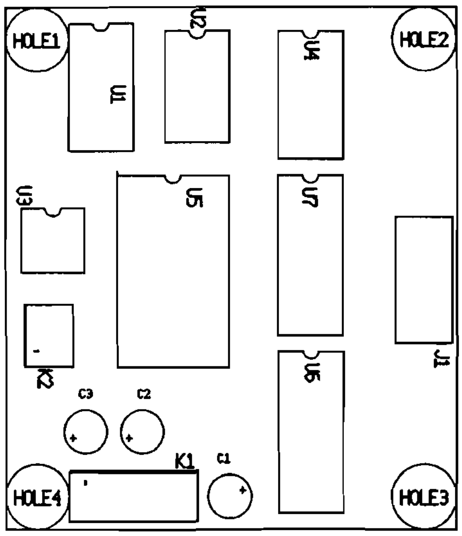


Figure 47: A/D board, components top side

Appendix C: Literature

List type 1

Ref.:	*Titel:*	*Author:*	*Publisher:*	*Year:*
ana1	Atlas of topographical anatomy of domestic animals	Peter Popesko	Saunders, Philadelphia	1970
ana2	Anatomie van het paard	Dr. K.M. Dyce / Dr. C.J.G. Wensing	Scheltema & Holkema, Utrecht (Netherlands)	1980
dia1	Clinical Diagnosis of diseases of large animals	W.J. Gibbons	Lea & Febiger, Philadelphia (USA)	1966
imp1	Equine veterinary journal (1990,22:2), Impedance plethysmography,	D.P. Attenburrow / F.C. Flack / M.J. Portergill	London	1990
imp2	Principles of applied biomedical instrumentation	L.A. Geddes / L.E. Baker	John Wiley & Sons inc., London (UK)	1968
met1	The veterinary clinics of North America. Large animal practice ;vol. 1, no. 1	R.A. Willoghby / W.N. McDonell	Saunders , Philadelphia	1979
met2	Scintigraphical analyses of pulmonary function in dogs	Cécile Clercx	Universiteit Utrecht, Utrecht (Netherlands)	1988
met3	Pulmonary function testing	Reuben M. Cherniack	Saunders, Philadelphia	1977
met4	New methods of studying gaseous exchange and pulmonary function	Alfred Fleisch	Charles C Thomas, Springfield, Illinois (USA)	1954
met5	Applied respiratory physiology	J.F. Nunn	Butterwoths, London (UK)	1969
met6	Het capnogram; een diagnostisch hulpmiddel by COPD?	F. van Dixhoorn	University Utrecht, Verterinary	1978
met7	Effect of a mask and pneumotachograph on tracheal and nasopharyngeal pressures, respiratory frequency, and ventilation in horses	S.J. Holcombe, W.L. Beard, K.W. Hinchcliff	Am J Vet Res 57: 3, 250-3, Mar	1996
phy1	Veterinary Physiology	J.W. Phillis	Wricht scientechnica, Bristol (UK)	1976
phy2	Duke's physiology of domestic animals	Melvin J. Swenson / William O. Reece	Comstock, Ithaca	1993
phy3	Textbook of veterinary physiology	James E. Breazile	Lea & Febiger, Philadelphia	1971
phy4	The lung : clinical physiology and pulmonary function tests	Julius H. Comroe jr. & Robert E. Forster II & Arthur B. Dubois & William A. Briscoe & Elizabeth Carlsen	Year book medical publishers, Chicago (USA)	1971
res1	Ventilation and gas exchange in each lung of the anaesthetised horse	Yves P.S. Moens	Universiteit Utrecht, Utrecht (Netherlands)	1968
res2	Equine anesthesia	William W. Muir / John A.E. Hubbell	Mosby Year book, St. Louis	1991
tst1	Organ Function Tests in Toxicity Evaluation	Charles A. Tyson / Daljit S. Sawhney	Noyes Publications, New Jersey (USA)	1985
tec1	Techniques in the life sciences. Physiology. P4, techniques in respiratory physiology		Shannon, County Clare	1984
ult1	Veterinary diagnostic ultrasound	Thomas G. Nyland / John S. Mattoon	Saunders, Philadelphia	1995
ult2	Veterinary diagnostic ultrasound	William E. Blevins / William R. Widmer	Purdue University, West Lafayette	1990
ult3	Veterinary Ultrasonography	P.J. Goddard	CAB International, Wallingford (UK)	1995
ult4	A thinkers guide to ultrasonic imaging	Raymond L. Powis / Wendy J. Powis	Urban & Schwarzenberg, Munchen (Germany)	1984

List type 2

Attenburrow, D.P. & Flack, F.C. & Portergill, M.J.
Impedance plethysmography
Equine veterinary journal, Vol. 22 (1990), No. 2
London, 1990

Blevins, William E. & Widmer, William R.
Veterinary diagnostic ultrasound
West Lafayette, Purdue University, 1990

Breazile, James E.
Textbook of veterinary physiology
Philadelphia, Lea & Febiger, 1971

Cherniack, Reuben M.
Pulmonary function testing
Philadelphia, Saunders, 1977

Clercx, Cécile
Scintigraphical analyses of pulmonary function in dogs
Diss. Veterinary University Utrecht (Netherlands), 1988

Comroe jr., Julius H. & Forster II, Robert E. & Dubois, Arthur B. & Briscoe, William A. & Carlsen, Elizabeth
The lung : clinical physiology and pulmonary function tests
Chicago (USA), Year book medical publishers, 1971

Diverse writers
Techniques in the life sciences. Physiology. P4, techniques in respiratory physiology
County Clare, Shannon, 1984

Dixhoorn, F. van
Het capnogram; een diagnostisch hulpmiddel by COPD?
Diss. Veterinary University Utrecht (Netherlands), 1978

Dyce, K.M. & Wensing, C.J.G.
Anatomie van het paard
Utrecht (Netherlands), Scheltema & Holkema, 1980

Fleisch, Alfred
New methods of studying gaseous exchange and pulmonary function
Springfield, Illinois (USA), Charles C Thomas, 1954

Geddes, L.A. & Baker, L.E.
Principles of applied biomedical instrumentation
London (UK), John Wiley & Sons inc., 1968

Gibbons, W.J.
Clinical Diagnosis of diseases of large animals
Philadelphia (USA), Lea & Febiger, 1966

Goddard, P.J.
Veterinary Ultrasonography

Wallingford (UK), CAB International, 1995

Holcombe, S.J. & Beard, W.L. & Hinchcliff, K.W.
Effect of a mask and pneumotachograph on tracheal and nasopharyngeal pressures, respiratory frequency, and ventilation in horses
Am J Vet Res, Vol. 57 (Mar 1996), No. 3, p. 250-3

Moens, Yves P.S.
Ventilation and gas exchange in each lung of the anaesthetised horse
Diss. Veterinary University Utrecht (Netherlands), 1968

Muir, William W. & Hubbell, John A.E.
Equine anesthesia
St. Louis, Mosby Year book, 1991

Nunn, J.F.
Applied respiratory physiology
London (UK), Butterworths, 1969

Nyland, Thomas G. & Mattoon, John S.
Veterinary diagnostic ultrasound
Philadelphia, Saunders, 1995

Phillis, J.W.
Veterinary Physiology
Bristol (UK), Wricht scientechnica, 1976

Popesko, Peter
Atlas of topographical anatomy of domestic animals
Philadelphia, Saunders, 1970

Powis, Raymond L. & Powis, Wendy J.
A thinkers guide to ultrasonic imaging
Munchen (Germany), Urban & Schwarzenberg, 1984

Swenson, Melvin J. / Reece, William O.
Duke's physiology of domestic animals
Ithaca, Comstock, 1993

Tyson, Charles A. &Sawhney, Daljit S.
Organ Function Tests in Toxicity Evaluation
New Jersey (USA), Noyes Publications, 1985

Willoghby, R.A. &McDonell, W.N.
The veterinary clinics of North America. Large animal practice ;vol. 1, no. 1
Philadelphia, Saunders, 1979