University of Southern Queensland

Faculty of Engineering and Surveying

Respiration Monitor and Recording System

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ABSTRACT

The correct operation of the body's respiration system is of paramount importance for the maintenance of human life. Respiration failure or interruption for periods exceeding a few minutes can be harmful and in some cases, fatal.

The metal foil strain gauge has been used for decades within the engineering and construction fields. This dissertation has researched a method of attaching a strain gauge to a flexible material to act as a sensing element for a respiration monitoring system.

The project has succeeded in showing that the metal foil strain gauge can effectively register the respiration rate of patients, in a cost effective manner. By combining this sensor assembly with a low cost processor, the project outcome of producing a respiration monitor and recording system has been realised.

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Andrew Allan MacMillan

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Signature

Dated

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I would be remiss in failing to mention the support of my wife and family who have supported me throughout the year and provided me with the time and space needed to complete this project.

I dedicate this dissertation to my daughter Rayna, my Muse, who provided the initial inspiration and whose example inspires me still.

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1.0 INTRODUCTION

The physiology and operation of the human body is one of the most amazing feats of biological engineering in nature. Arguably the respiration system is the most important part of the body, with the average human adult taking over 20,000 breaths in a 24 hour period.

The correct operation and function of the human respiratory system is one that the medical sciences have grappled with for centuries. Failure or interruption of this systems correct operation can lead to injury, brain damage or death if left undiagnosed and untreated. This is particularly true in neo-natal children where their lung function does not become fully developed for several months after birth.

Being able to easily and effectively monitor the respiration system is considered to be of paramount importance by medical science. Attendance at any intensive care unit in a hospital easily confirms this to be true. Unfortunately this monitoring equipment is expensive and difficult to operate, and hence is not available to the general public.

Parents of new born children can attest to the fact that one of their greatest fears is that the child may stop breathing. Many parents suffer from broken sleep because they are constantly checking that the child is in fact still breathing whilst asleep. While not a common occurrence, a general cessation of breathing, also known as apnoea (and sleep apnoea while asleep) is common enough for the event to be a real concern to people.

However, because the adult respiratory system is fully developed it rarely leads to death. Current diagnostic techniques are expensive and usually require attendance at a sleep centre, where confirmation of the condition can occur.

The use of strain gauges within industry in general and engineering in particular is wide spread. The strain gauge is a simple device that changes its resistance depending on the amount of physical strain placed on the gauge. The advantages of using strain gauges are that they are an inexpensive sensing device and the physical parameters of their operation are well know, and hence their output is predictable under known conditions.

1.1 OBJECTIVE

The objective of this project is to design, construct and test a Respiration Monitor and Recording System utilising strain gauges as the sensing element. Once the prototype is developed it will undergo experimental testing to determine the accuracy of data that is gathered. The initial design will provide a record of the patients' respiration over a period of time. The data is recorded and stored into flash memory, where it is then able to be downloaded, converted into a decimal equivalent and used to provide a graphical display of the patient's breathing pattern.

1.1.2 Specification

The complete list of specification objectives is supplied in the project specification (Appendix A). The main design objectives for this project are listed below:

- Design a digital data logging system to accept inputs from analogue strain gauges in a single input system.
- Determine a safe and effective method of attaching the sensing elements to the human body.
- Construct a working prototype of the design.
- Carry out experiments on individuals of varying age groups to determine the validity of the design.
- Produce a graphical display from the raw data.
- Incorporate an apnoea monitor and alarm system (if time permits).

1.2 DESIGN PARAMETERS

The following parameters or boundaries were used when designing the respiration monitor device:

Cost - The objective of the respiration monitor was to design a working functional device that could be delivered for under $100.00. The reason for this limit was that personal experience had shown that effective monitors, especially for neo-natal children can be very expensive, and are often out of the price range that parents can afford.

Simplicity - In keeping with producing an inexpensive device, the associated circuitry was to be kept as simple and functional as possible. To achieve this all components can be purchased “off the shelf”.

Comfort - The final design had to be such that the patient wearing the device would not suffer any undue discomfort. This is especially true with neo-natal children, where small discomforts can prevent sleep from occurring.

Safety - The design had to be safe and present no risk to the patient. For this reason the system has been designed to operate using a battery power supply. The sensor assembly has a single lead between the patient and the remainder of the circuit. This lead must be inaccessible to neo-natal children (preferably by be kept under clothing) to prevent a choking hazard.

Strain Gauge – The sensing device for this project was to be based on a standard metal foil strain gauge.

2.0 RESPIRATION

2.1 DEFENITION

Respiration is the process of gaseous exchange between an organism and its environment. In humans it is the specialised organs (lungs) that provide for the efficient inhalation of oxygen and expulsion of carbon dioxide from the body's tissues.

(Oxford 1997, p 567).

2.1.2 How does respiration occur?

The respiration system consists of two distinct, yet complimentary parts; the upper and lower respiratory tracts. The upper tract is responsible for warming and filtering the inhaled air, while the lower tract is responsible for the exchange of gasses; being inhaled oxygen and exhaled carbon dioxide (Ferrer 2007).

2.1.3 External physical changes due to respiration

When inhalation occurs, the chest (thoracic cavity) expands and the diaphragm contracts. The expansion of the chest occurs as the lungs become filled with air. When exhalation occurs, the size of the chest decreases and the diaphragm relaxes (Ferrer 2007).

The physical movement of the chest occurs at both the front and back of the thoracic cavity. This project aims to sense this chest movement with strain gauge(s) in order to detect and record a person's breathing.

The importance of the human body to be able to effectively and efficiently breathe cannot be understated. When a baby is first born their respiration rate can vary from 30 to 60 inhalations per minute (Lewer + Robertson 1989, p 45). Any prolonged interruption or disturbance in this breathing can have fatal consequences.

2.2 TYPES OF BREATHING PATTERNS

The following types of breathing are the main types found in adults and are described fully in (Farrell, 2006, p 482) and are detailed in the following chapters:

2.2.1 Eupneoa

2.2.2 Bradypnoea

2.2.3 Tachypnoea

2.2.4 Hypoventilation

2.2.5 Hyperventilation

2.2.6 Apnoea

This list does not include Cheyne-Stokes or Biot's respiration, which can be considered outside the scope of this project.

2.2.1 Eupnoea

Eupnoea is normal breathing whilst at rest; it will vary between 12 to 18 breaths per minute for an adult (Farrell, 2006, p 482). A simulation of this breathing pattern is shown as Figure 2.1. A point to note, is that this breathing will be interrupted approximately every 100 breaths by a sigh, however the actual rate will vary from person to person. This sigh is dependent on posture, weight, age and several other factors. This sigh is used by the body to correct any imbalances between current oxygen and carbon dioxide levels in the body (Ferrer, 2007).

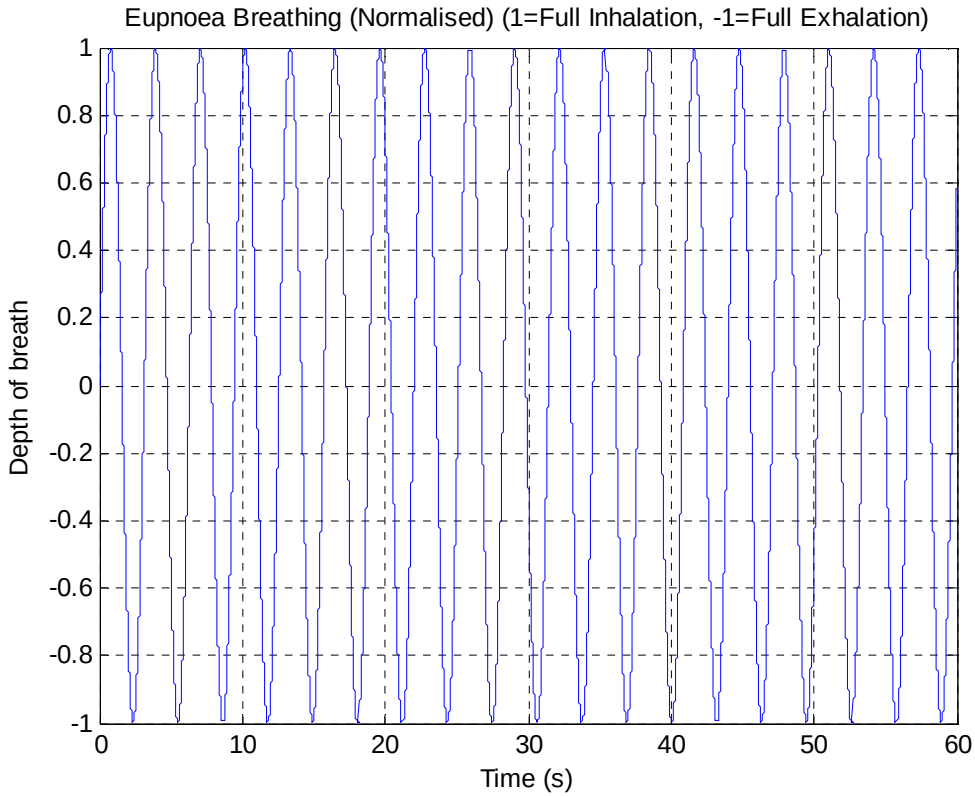

Figure 2.1 Normalised Eupnoea Breathing Pattern Simulation

2.2.2 Bradypnoea

This type of breathing is slower than normal breathing and is generally less than 10 breaths per minute as the simulation in Figure 2.2 shows. Bradypnoea indicates that there is a problem with the patient, which may include drug overdose, brain injury or intracranial pressure (Farrell, 2006, p 482).

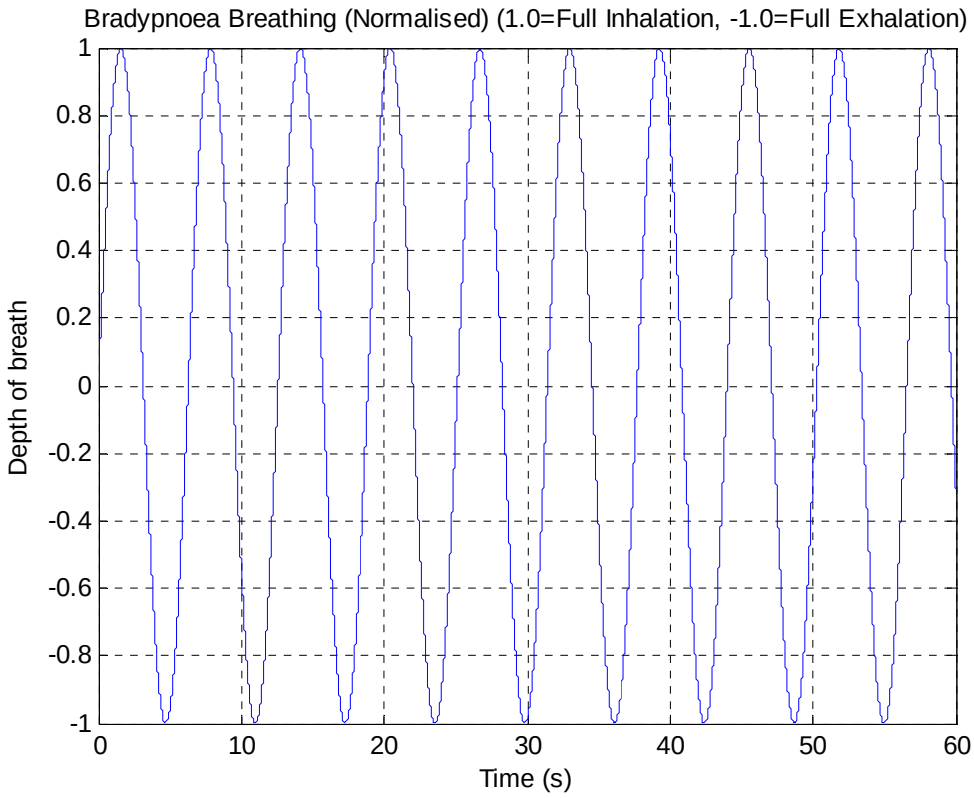

Figure 2.2 Normalised Bradypnoea Breathing Pattern Simulation

2.2.3 Tachypnoea

This type of breathing is rapid and shallow and is associated with pneumonia and rib fractures. If the respiration rate is greater than 24 breaths per minute then it is classed as tachypnoea (Farrell, 2006,). Figure 2.3 shows a simulated normalised breathing pattern for people suffering tachypnoea. As can be seen in Figure 2.3, the depth of breath has decreased with respect to the graph for eupnoea (Figure 2.1).

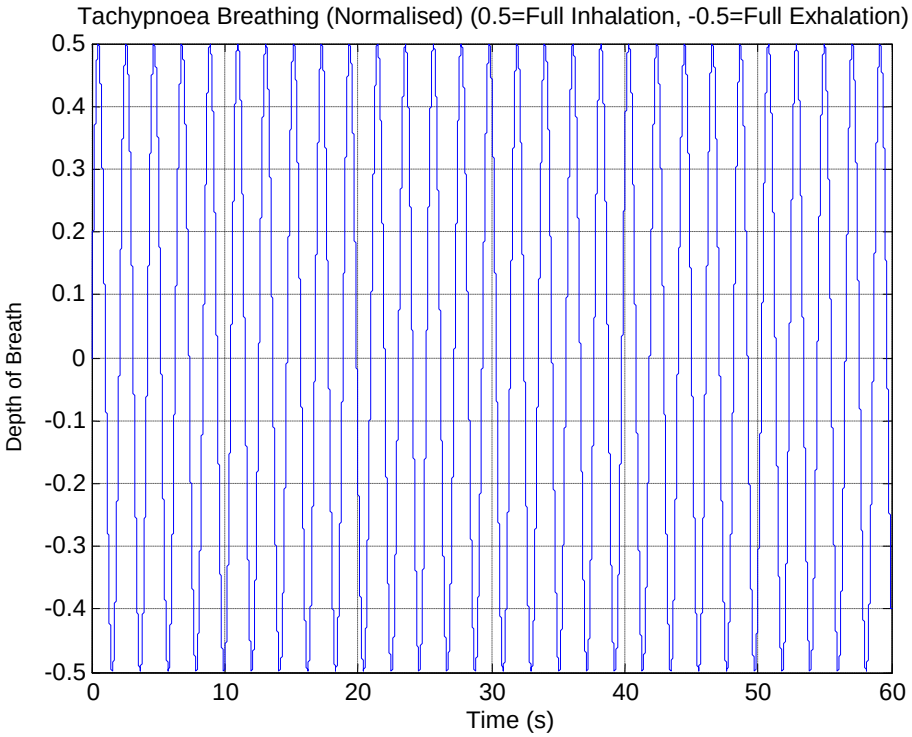

Figure 2.3 Normalised Tachypnoea Breathing Pattern Simulation

2.2.4 Hyperventilation

Hyperventilation is an increased depth and rate of breathing above the 12-18 breaths per minute for a normal adult human. This causes an increased amount of oxygen in the blood; which can lead to respiratory alkalosis. Respiratory alkalosis may lead to a decrease of carbonic acid in the blood and can be caused by either extreme anxiety, hypoxemia or several other factors (Farrell 2006, p 482) . The simulated hyperventilation normalised breathing pattern is shown in Figure 2.4. As can be seen from the graph below, the depth of breath has increased in comparison to the eupnoea graph (Figure 2.1).

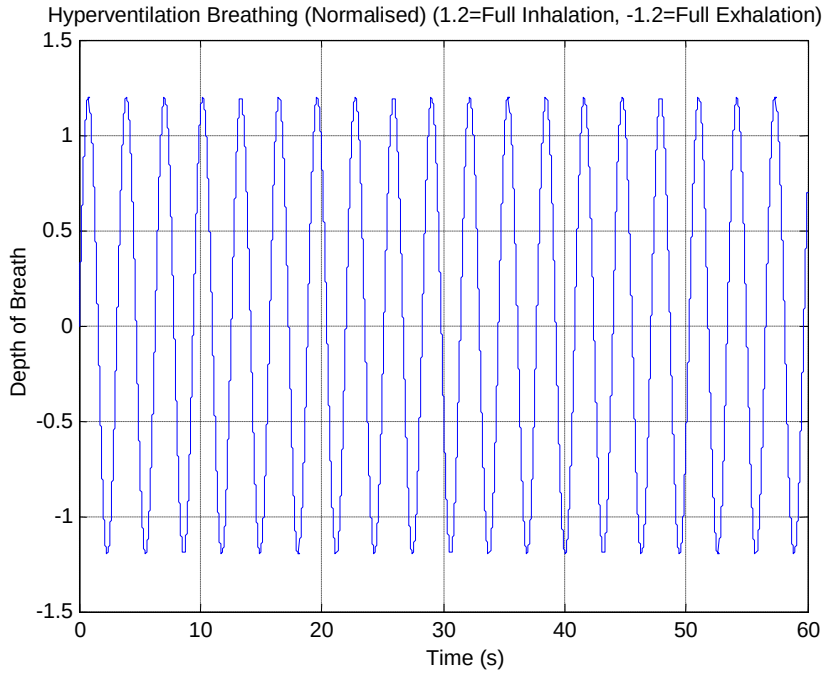

Figure 2.4 Normalised Hyperventilation Breathing Pattern Simulation

2.2.5 Hypoventilation

Hypoventilation is defined as shallow and irregular breathing. This Type of breathing pattern can result in limited impulses from the brain to the respiratory muscles, depressed respiratory centres and limited thoracic movement (Farrell 2006, p 280). Hypoventilation is often confused with hyperventilation due to the similarities in their spelling. As a result of suffering hypoventilation, a person will not be receiving enough oxygen or expelling the resultant carbon dioxide from their body. This causes the person to become lethargic and slow to respond to outside stimulus.

2.2.6 Apnoea

Apnoea is where there is a cessation of breathing. When the person is asleep it is known as sleep apnoea. If left untreated this condition is life threatening (Farrell 2006, p 482). A simulated apnoea breathing is shown in Figure 2.5, and is indicated by the flat red line on the graph.

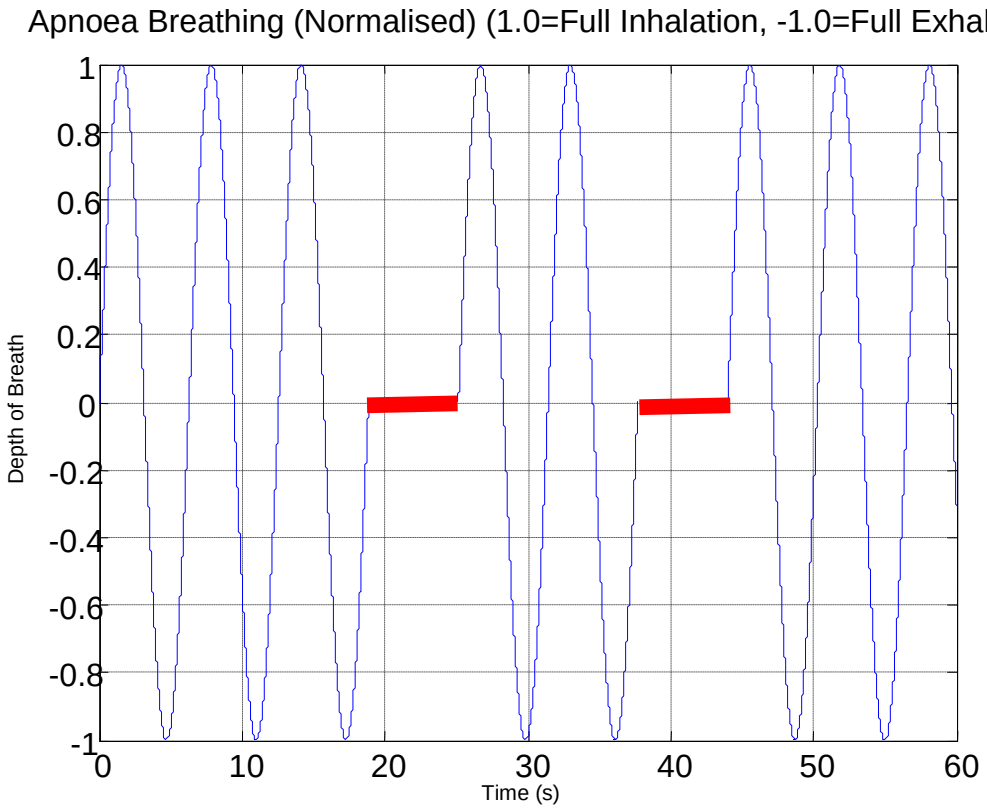

Figure 2.5 Normalised Apnoea Breathing Pattern Simulation

3.0 CURRENT RESPIRATION MONITORING EQUIPMENT

There are many different types of respiration monitoring equipment that is commercially available on the market today. Some of this equipment is suited for use in a medical facility and some of it is suitable for personal use.

The different types of respiration monitoring can be broken down into two broad categories, invasive and non-invasive. The invasive method of measuring a person's lung capacity and function is generally performed in a clinical or hospital situation. For this reason it is unlikely to be suitable for home use.

3.1 INVASIVE MONITORING

3.1.1 Pneumotacography

Pneumotacography utilises differential flow meters that are placed at each end of a restriction tube. Flow is determined by the ratio of the pressure difference between each end of the tube and the known resistance to flow as shown in equation 3.1.

$$\phi = \frac{P_1 - P_2}{R} \qquad \textbf{Equation (3.1)}$$

ϕ = Flow (litres/min)

P = Pressure (Pascals)

R = Flow Resistance ()

The patient breathes into the tube and the flow rate is recorded, from this the lung function and capacity can be determined to a reasonable degree of accuracy (Neuman, 2003c, p 3).

3.1.2 Spirometry.

Spirometry is used to measure the volume of exhaled air from a patient. This test is used to determine the amount of respiratory impairment that the test subject is suffering from. For this test to be effective it requires the cooperation of the patient (Springhome, p 26) and consequently it cannot be used on neo-natal or very young children.

This test is able to distinguish between obstructive and restrictive disorders and detect for early signs of respiratory impairment (Springhome, p 27). The test is undertaken by placing a nose clip on the patient and then having them breathe through a mouthpiece. As can be seen in Figure 3.1 the volume of air flow is recorded and the capacity of the lungs can be determined.

Although used in clinical situations, the Spirometry test is not useful for either home use, or the general monitoring of a patient's respiration rate.

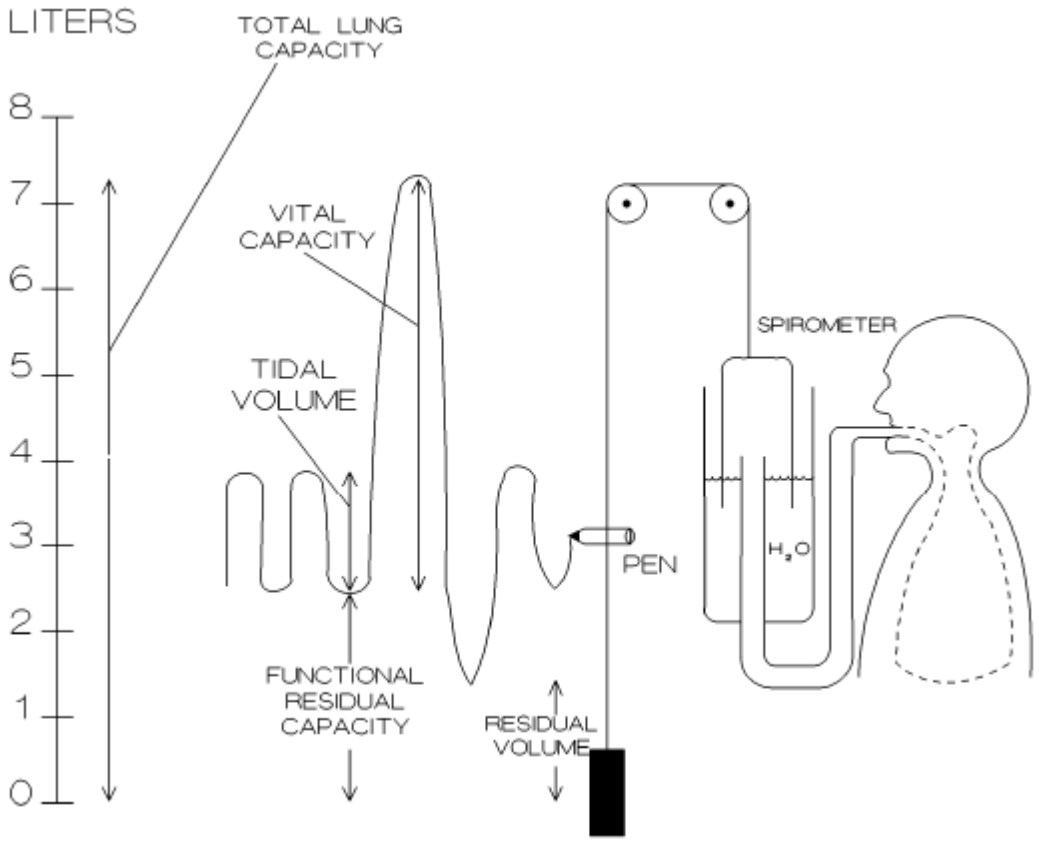

Figure 3.1 Spirometry Test (Neuman, 2003c, p 2)

3.1.3 Nasal Thermistry.

Nasal thermistry is another method of measuring the flow of air out of a patient's lungs. This method involves using a thermistor placed inside the nasal passage to detect increases in ambient air temperature caused by the exhalation of a breath (Neuman, 2003c, p 2). Figure 3.2 shows the attachment of a thermistry sensor to a neo-natal child (Neuman, 2003a, p 28). It can be clearly seen that this type of sensing is only practical on sedentary patients.

Figure 3.2 Nasal Thermistry on a Neo-Natal Child (Neuman, 2003a, p 27)

The nasal thermistor is very sensitive to changes in temperature, however it does not provide for a linear output. To achieve a linear output, a linearising circuit can be employed. This circuit consists of a series of resistors placed in series and parallel with the thermistor. The result is a more linear response curve to changes in detected temperature (Neuman, 2003b, p 22).

3.1.4 Flow Metering

Rotating vane and ultrasonic flow meters can be used to detect the flow of air from an exhaled breath. The rotating vane is attached to a small turbine whose velocity and output is determined by the rate of exhalation. The ultrasonic flow meter uses changes in frequency of the received signals to estimate flow velocity (Neuman, 2003c, p 2).

3.1.5 Analysis of Invasive Monitoring

When in a controlled or clinical environment invasive monitoring can be easily controlled by the appropriate medical personnel. This type of monitoring is not practical in a standard home environment due to the complexity of some of the equipment and the requirement for trained medical people to be present. The spirometry test in particular is not suited for the long term monitoring of a patient's respiration.

For a monitoring system to be effective within a home environment it must be simple to operate and cause a minimum of discomfort to the patient wearing the device. Observations by the author have revealed that simple and common occurrences can cause major discomfort for the patient. An example of this can be seen in Figure 3.2, the sensing device is held in position by the use of surgical tape. This tape if left in prolonged contact with the patient can cause allergic reactions, resulting in blistering and rashes forming on the skin.

If any of these devices were to be used in a home environment it can be safely assumed that issues of safety, equipment calibration, training and availability would be the responsibility of the issuing medical authority.

For these reasons it is assumed that this type of monitoring is not suitable for use in a home environment unless directed to by a competent medical practitioner.

3.2 NON-INVASIVE MONITORING

3.2.1 Static Charge Sensitive Bed

The diaphragmatic EMG uses a sensor that is embedded in a patient's sleeping mattress. This sensor (transducer) is used to detect any movement by the patient during sleep. As the transducers are extremely sensitive they can detect the respiration of the patient, the amplitude of the detected signal varies with the position of the body (IST, p 29).

3.2.2 Respiratory Inductance Plethysmographs (RIP)

Respiratory Inductance Plethysmographs **(**RIP) uses two elastic bands that are worn by the patient; one is worn on the chest and the other is worn on the abdomen. These bands have an inductor embedded within each of them. Each inductor is energised by a sinusoidal waveform. As the patient breathes the cross sectional area of the inductors also changes in response to the breath being inhaled and exhaled. This change in inductance is then used to calculate the thoracic and abdominal cross sectional areas (Neuman, 2003c, p 10), thus the depth of respiration of the patient wearing the sensor can be determined. Figure 3.3 shows the position of the elastic bands on a patient.

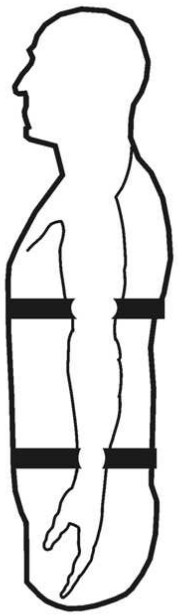

Figure 3.3 RIP Sensor Positions (Gandis, p 6)

3.2.3 Dynamic Air Pressure

The dynamic air pressure method of respiration monitoring uses a bellows arrangement that is attached via a pneumatic hose to a pressure detector. Changes in pressure due to patient respiration are felt by the air bag. The air pressure detector converts the detected changes in air pressure, with the converter sampling at a rate of 1 kHz and amplifying the signal for processing by the personal computer (Takarada, p 3). This particular arrangement was designed for use in a dental chair; however the principal of operation is very similar for monitoring the respiration of a sleeping patient.

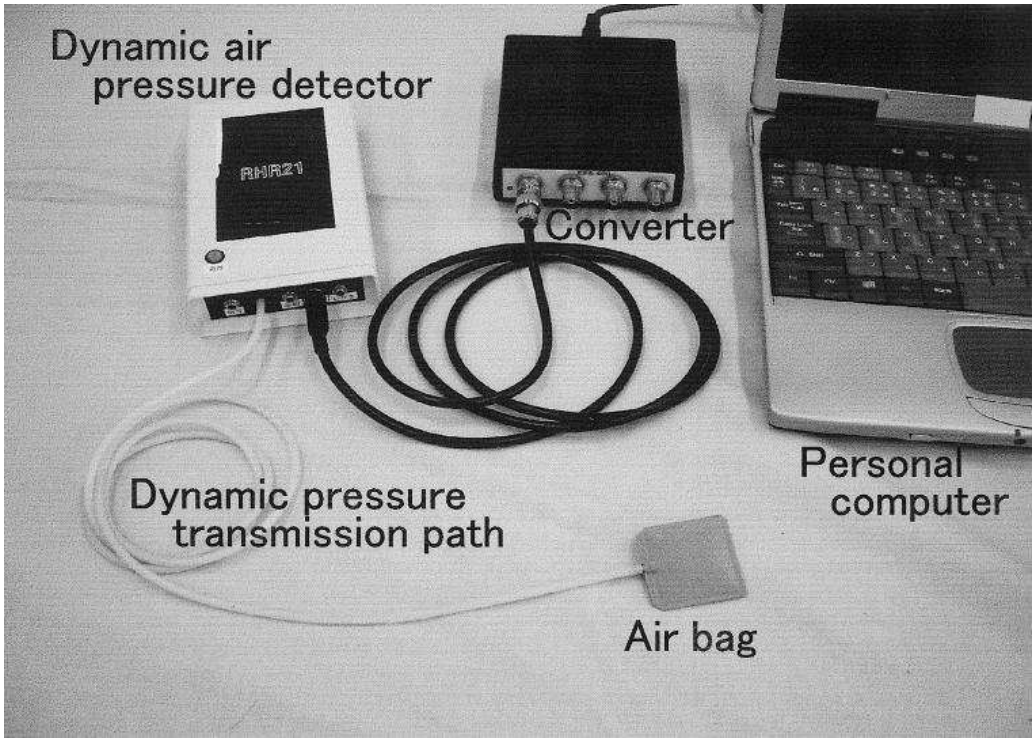

Figure 3.4 Dynamic Air Pressure Monitoring System (Takarada, p 3)

3.2.4 Analysis of Non Invasive Respiration Monitoring

The use of RIP has become a common form of respiration monitoring since the 1990's. The reason for this is the expiry of patents and the reduction in associated costs (Gandis, 2007, p5). This type of monitoring appears to offer an effective method of recording a patients breathing. No cost analysis was produced for this report as it was not possible to find this information. The technical details and cost of the static charge sensitive bed were not investigated for this project as they appear to be very expensive and offered no useful information to the development of the project's

hardware. Further development of the project's device may warrant further investigation into the signal processing used in the static charge sensitive bed in the future.

The author of this paper has utilised a dynamic air pressure monitoring system for maintaining a vigil on a persons breathing. It was found that the device had to be taped into place, as any variation in its position caused the alarm to sound. While this was not a daily occurrence, it was inconvenient and did cause many sleepless nights. The use of a pneumatic tube is also problematic. If the tube becomes kinked or blocked in any way, then it causes the alarm to trigger. This may seem trivial; however parents operating this device in the middle of the night will often be concentrating on more pressing matters than the lay of a tube.

The advantage of these devices is that the delay in activation of the alarm can be programmed into the device. Consequently, a cessation of breathing for example of five seconds will not trigger the alarm, but a cessation of ten seconds will. The device does not have the capability to record the respiration rate, only to monitor the breathing event and activate the alarm if it ceases. It was also found that these devices had a price that started at $200.00. While this is not a large amount of money, the cost can be prohibitive to some people.

4.0 STRAIN GAUGES

4.1 DEFINITION

Strain is defined as the ratio of change in length (ΔL) to the original length (L). The formula for strain (Kyowa, 2005, p 4) is expressed below as equation 4.1.

$$\mathrm{E} = \frac{L + \Delta L}{L} \qquad \textbf{Equation (4.1)}$$

E = Strain

L = Original Specimen Length (m)

ΔL = Change in Specimen Length (m)

Strain is a dimensionless quantity as the units cancel each other out.

4.2 HISTORY

The strain gauge is one of those items where the principle was discovered, then had to wait the better part of century to be put into practical use. Lord Kelvin (b.1824 – d.1907) was plagued with variations in resistance due to changes in the physical strain that was placed upon a piece of wire. He noted this discovery in 1856. However, not realising its significance the concept lay dormant until the 1930s (Windrow + Hollister, 1982, p 8).

In 1938 a method of bonding resistance wire, such as a copper/nickel alloy, to a structure that allowed for the measurement of surface strain was discovered. The advent of World War II and the requirements of the aircraft industry led to its use in structural testing of various aircraft designs (Windrow + Hollister, 1982, p 15).

The invention of the foil strain gauge in 1952, by The Saunders-Roe Company (Windrow + Hollister 1982, p 17) in the United Kingdom, meant that the use of these gauges expanded beyond the boundaries of the aircraft industry. Since then, the use of strain gauges can be found in all types of industries, ranging from aircraft to building construction.

Today there are many different types of strain gauges in use, including but not limited to:

- Electrical resistance strain gauge
- Capacitance strain gauge.
- Pizo-electric strain gauge.

Each of these gauges will be discussed in detail.

4.3 ELECTRICAL RESISTANCE STRAIN GAUGE

4.3.1 Concept of Operations

When a piece of metal that is capable of conducting an electrical current is placed under strain, the metal becomes elongated and thinner. This elongation causes the electrical resistance of the wire to increase. If the same piece of wire is compressed, the metal becomes shorter and broader. This shortening causes the electrical resistance of the wire to decrease. It is the ratio between the change of the physical properties and the applied force to the wire that the strain gauge measures. This indirect measurement of applied force via resistive changes means that very accurate and sensitive changes in the physical properties of the gauge can be used to determine the relative motion of the object under test. To determine the sensitivity of a gauge the gauge factor, which is a constant, must be calculated.

4.3.2 Gauge Factor

The sensitivity of a resistance strain gauge is known as its Gauge Factor (*k*). The Gauge Factor is a dimensionless quantity and is used to illustrate the suitability of a material for use in a strain gauge. The value of the constant *(k*) is determined as follows in equation 4.2:

$$k = 1 + 2v + \frac{\Delta\rho / \rho}{\Delta l / l} \qquad \textbf{Equation (4.2)}$$

k = Gauge Factor

v = Poisson's Ratio

ρ = Resistivity (Ohms)

l = Length (m)

The value of Poisson's Ratio (v) for most conductive metals is 0.3. This means that the sensitivity for the majority of materials used in the construction of strain gauges has a gauge factor of approximately $k = 2$.

As the gauge is placed under physical strain throughout its operating life, care must be taken to ensure that the physical limits of the material are not exceeded. This means that the elastic limits of the material must never be breached. If the gauge's elastic limit is exceeded then the gauge is rendered useless. Exceeding the elastic limit causes changes in internal stresses of the material to approach zero. Consequently the Poisson's Ratio approaches a value of 0.5, which when combined with the variations in resistivity and length causes the subsequent readings to become inaccurate. Table 4.1 shows some of the materials that are used in strain gauge construction, their gauge factors and alloy composition.

Table 4.1 Gauge Factors (Windrow + Hollister, 1982, p 8)

Material (%)	Gauge Factor (Typical)
Copper – Nickel (55-45)	+2.1
Nickel – Chromium (80-20)	+2.2
Iron – Chromium – Aluminium (70-20-10)	+2.2
Nickel – Chromium (75-20)+Fe+Al	+2.1
Iron	+4
Nickel	-12

4.3.3 Features of Resistive Strain Gauges

Like all branches of engineering and the physical sciences, a desirable set of parameters for repeatability and reliability are required. The following list shows the characteristics that are considered when gauge design and construction is undertaken:

- Linear sensitivity in the elastic range
- High resistivity
- Low hysteresis
- High strain sensitivity
- Low temperature coefficient
- Wide temperature range
- Good fatigue life.

4.3.4 Construction and Operation

The modern resistive strain gauge is constructed using a metal foil that is placed onto a backing material. The physical size of the gauge is quite small, with the overall length ranging from 5 to 20 mm. Figure 4.1 shows the length of the gauge, and the resistive foil pattern that is located on the backing material.

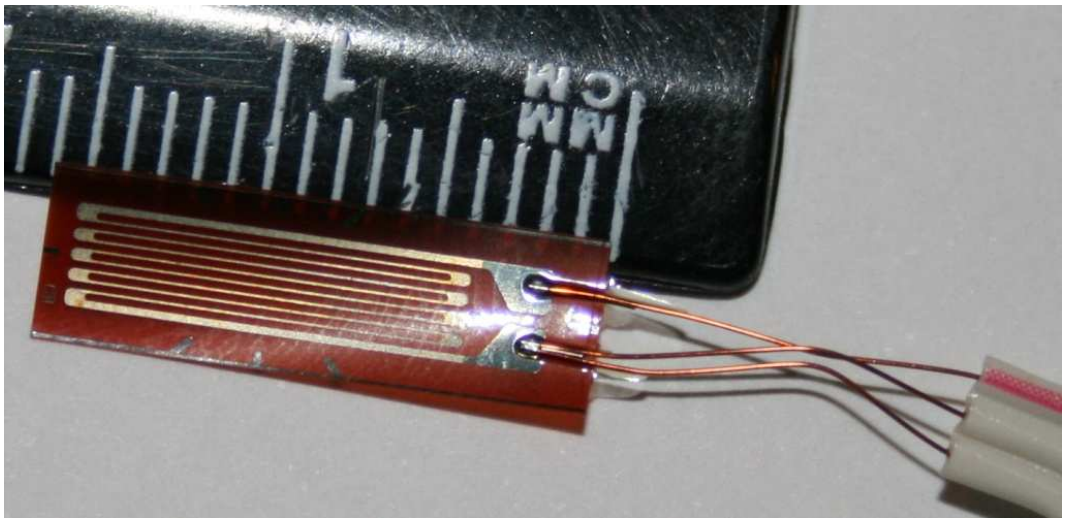

Figure 4.1 Three Wire Strain Gauge (Photograph by Author)

When the gauge is placed onto the object under test it is bonded into place using a resin or glue. As can be seen in Figure 4.2 the strain gauge is constructed from three materials that are bonded together to form a complete gauge. The metallic foil is sandwiched between a base and plastic laminate that protects the foil from damage.

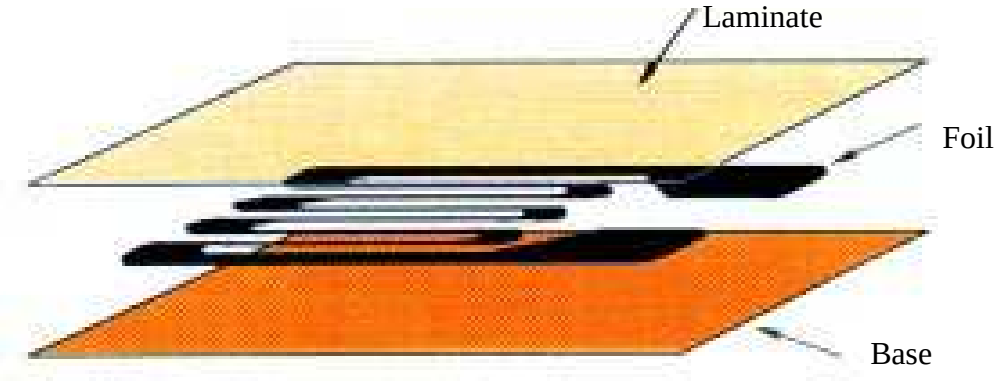

Figure 4.2 Metallic Foil Gauge Construction (Kyowa, 2005, p6)

Depending upon the orientation of the gauge it is possible to measure applied forces in any direction. Figure 4.3 shows how the metal foil is arranged in a hairpin arrangement. As the gauge is placed under stress along the Y-axis (top to bottom of Figure 4.3) its resistance is increased. This is because the overall cross sectional area of the gauge is decreased. Conversely, compression along the Y-axis increases the cross sectional area, causing the resistance to decrease. Strain that is applied across the X-axis cause no changes in the resistance of the gauge, and hence no change in output voltage to the sensing equipment that is used to measure the strain applied to the gauge.

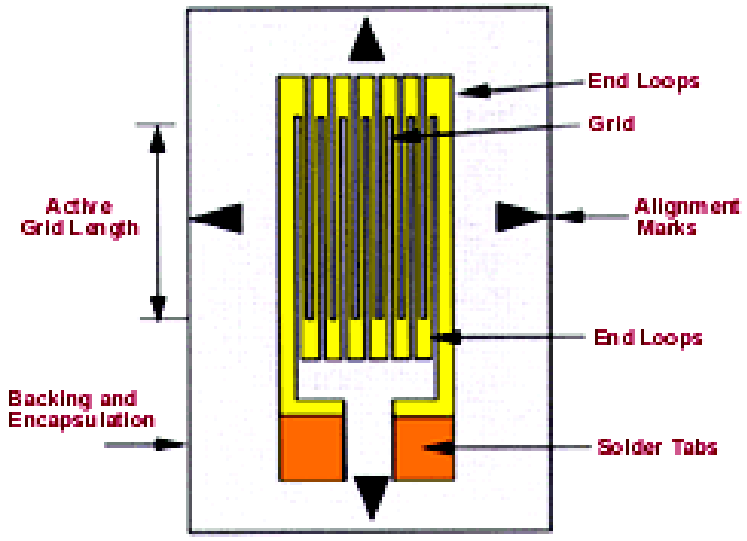

Figure 4.3 Typical Metal Foil Strain Gauge (Sensorland, p 1)

The amount of change in resistance due to the amount of applied strain can be calculated from equation 4.3 (Kyowa, p 8).

$$\frac{\Delta R}{R} = k \times \varepsilon \qquad \textbf{Equation (4.3)}$$

R = Original Resistance (Ohms)

ΔR = Change in Resistance (Ohms)

k = Gauge Factor

ε = Strain applied

Once the change in resistance is known, it is then possible to determine the change in voltage using a whetstone bridge. A whetstone bridge is used to indirectly determine the change in resistance of the gauge via the change in potential across the bridge. This is done because the changes in resistance are so minute, that a standard ohmmeter cannot measure the change (Kyowa, 2005, p 8).

4.3.5 Whetstone Configurations

Depending on the application, the whetstone bridge can be configured several ways. The bridge can be configured to use one, two or four gauges depending on the desired application (Kyowa, 2005, p 9). Figure 4.4 shows a one gauge configuration, sometimes called a quarter wave-bridge.

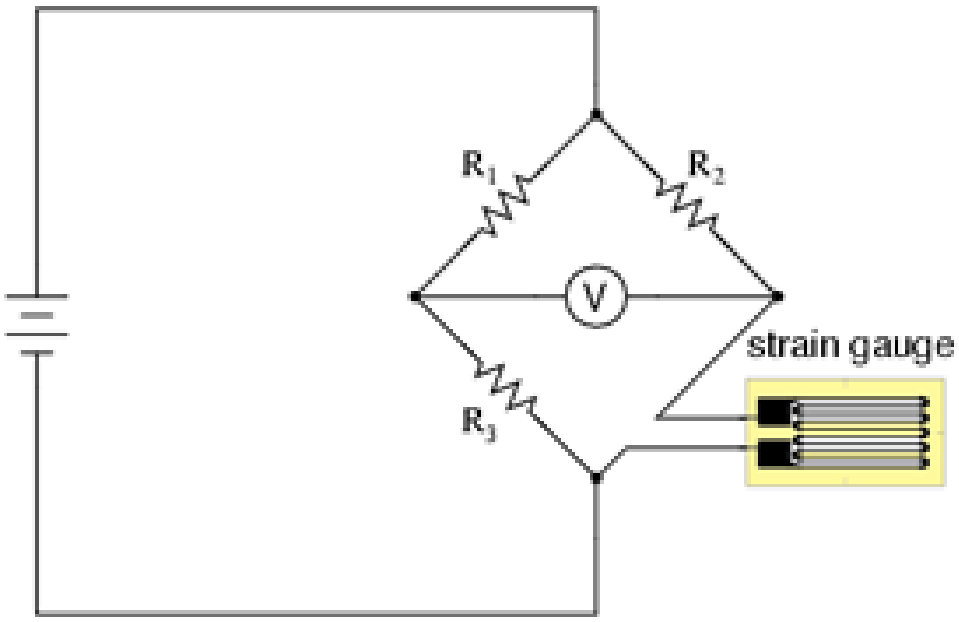

Figure 4.4 One Strain Gauge Circuit Configuration (Sensorland, 2007, p 5)

Equation 4.4 (Kyowa, 2005, p 9) is used for calculating the output voltage of the one gauge circuit.

$$V = \frac{\Delta R}{R} \times E \times \frac{1}{4} \qquad \textbf{Equation (4.4)}$$

V = Output Voltage (V)

E = Applied Voltage (V)

ΔR = Resistance Change (Ohms)

Figure 4.5 shows a two gauge configuration, also called a quarter wave bridge with temperature compensation.

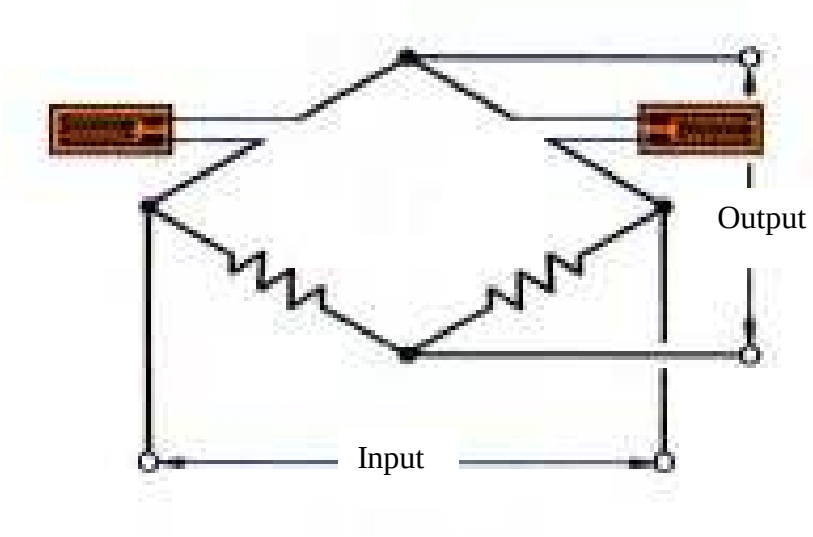

Figure 4.5 Two Gauge Configuration with Compensation (Kyowa, 2005, p 11)

The use of a second strain gauge in place of a fixed resistor allows for fluctuations in temperature to be neutralised. This occurs because the second gauge is placed in the same physical location; however, no strain is applied to the compensation gauge. Therefore any temperature variations occur to both sides of the bridge circuit. Equation 4.4 is also used to calculate the output voltage of this circuit when the second gauge is not placed under strain.

If the second gauge is placed under strain then the temperature compensation still occurs, however the sensitivity of the circuit is decreased. Equation 4.5 is used to calculate the output of this circuit (Kyowa, 2005, p 10).

$$V = \left(\frac{\Delta R1}{R1} - \frac{\Delta R2}{R2} \right) \times E \times \frac{1}{4} \qquad \textbf{Equation (4.5)}$$

V = Output Voltage (V)

E = Applied Voltage (V)

ΔR = Change in Resistance (Ohms)

R = Resistance (Ohms)

Figure 4.6 shows the two gauge configuration with temperature compensation (Kyowa, 2005, p 10). When this configuration is used, each of the gauges is placed under strain. This means that the overall sensitivity of the bridge is increased.

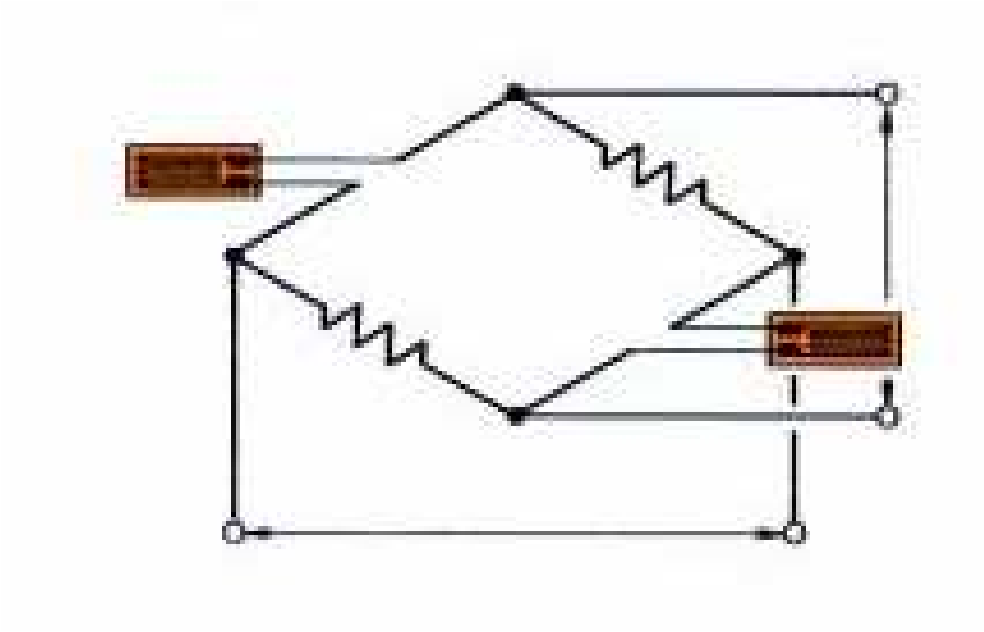

Figure 4.6 Two Strain Gauge Circuit Configuration (Kyowa, 2005, p 11)

Equation 4.6 is used to calculate the output when the circuit is configured this way (Kyowa, 2005, p 10).

$$V = \left(\frac{\Delta R1}{R1} + \frac{\Delta R2}{R2} \right) \times E \times \frac{1}{4} \qquad \textbf{Equation (4.6)}$$

V = Output Voltage (V)

E = Applied Voltage (V)

ΔR = Change in Resistance (Ohms)

R = Resistance (Ohms)

When it is desirable to have extremely sensitive strains measured a four gauge configuration can be used. This system is rarely used for strain measurement and is more often found in strain gauge transducers (Kyowa, 2005, p 10).

4.3.6 Compensation Requirements

Due to their sensitivity, compensation for temperature induced variations is required with most strain gauges. The use of a second strain gauge in the bridge configuration was discussed in section 4.2.6.

The other part of the strain gauge that must be compensated for is the leads attaching the gauge to the bridge circuit. The use of two wires as shown in the bridge circuits does not allow for thermal changes in resistance of the wires. This is overcome by placing a third wire onto the strain gauge and modifying their input in the bridge circuit. Figure 4.7 shows the circuit configuration for this compensation.

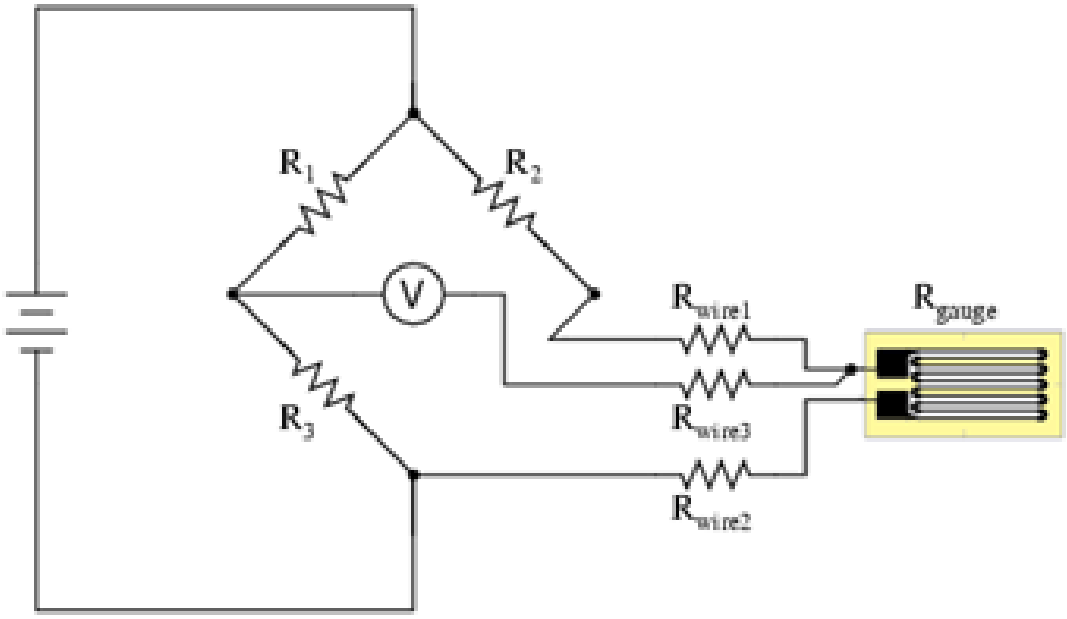

Figure 4.7 Three Wire Circuit Configuration (Sensorland, 2007, p 7)

By configuring the circuit with three wires, the lead resistance is distributed to adjacent sides of the bridge. With the third wire being positioned on the outside of the bridge circuit, it has a negligible effect on circuit operation or sensitivity (Kyowa, 2007, p 14).

4.4 CAPACITANCE STRAIN GAUGES

The development of the capacitance strain gauge was undertaken by Hughes Aircraft Company and Boeing in separate projects in the 1960s. The main reason for this development was the need to measure stress and strain loads at temperatures exceeding 800 degrees Celsius (Windrow + Hollister, 1982, p 292).

4.4.1 Concept of Operation

Equation 4.7 provides the basic operating principle for a capacitance strain gauge (Windrow + Hollister 1982, p 291):

$$C = \frac{ak}{d} \qquad \textbf{Equation (4.7)}$$

C = Capacitance (Farads)

a = Cross Sectional Area (m^2)

k = Dielectric Constant

d = Distance between plates (m)

By varying the distance between the plates in relation to the strain that is applied to them, it is then possible to determine the strain that is applied to the object under test.

If a dielectric of air is used, and the cross sectional area is known, then it stands to reason that it is only the variations in distance that will cause the capacitance to vary. Equation 4.7 tells us that the physical properties or geometrical features of the gauge can be used to determine the design of the capacitive gauge. The main area of concern in this design must be the dielectric used in the gauge. For the gauge to be useful it must have a dielectric that provides a linear change with respect to temperature. (Windrow + Hollister, 1982, p 292).

4.4.2 Operation

The operation of this type of strain gauge requires the use of an alternating current power supply. The gauge is placed into a bridge configuration with variations in the capacitance causing variations in current flow, which is then detected. The principle of operation of the bridge configuration is similar to the metal foil gauge described in section 4.2.5. For the successful operation of such a device the capacitance of the cables must also be taken into account. These cable capacitances will also vary with

temperature and humidity, which can lead to spurious signals being introduced into the system (Neubert, 1967, p 19).

The European Organization for Nuclear Research (CERN) developed a capacitance strain gauge (Ozelis, p 1) that used Kapton as the dielectric material. This material has a thickness of 0.0001” to 0.0005”. It was discovered that thinner material for the dielectric was not suitable or available (Ozelis, p 7). This particular gauge was developed in an attempt to measure the stress applied to superconducting magnetic coils.

Whilst these gauges have uses within the aerospace industry and certain specialist applications, they do not appear suitable for simple strain measurement tests.

4.5 PIEZOELECTRIC OR SEMICONDUCTOR STRAIN GAUGES

Nature has provided engineers with a natural strain gauge, the quartz crystal. When a crystal is subject to stress an electrical charge is generated on its surface. The man made version of this substance is Polarised Polycrystalline Ceramic (Neubert, 1967, p 20).

4.5.1 Concept of Operation

Equation 4.8 provides the basic operating principle for the piezoelectric gauge:

$$V = hEt \qquad \textbf{Equation (4.8)}$$

V = Voltage generated (V)

h = Piezoelectric Strain coefficient (V/m)

E = Strain applied

t = Slab thickness of crystal (m)

As a strain is applied to the crystal a surface charge is generated. A piezoelectric crystal of 1mm in thickness with an applied strain of $E = 10^{-6}$ produces an output of approximately 4.5 Volts (Neubert, 1967, p 20). Electrodes are attached to the surface of the crystal and relay the voltage to the required detector circuitry. Because a surface charge is only produced when strain is applied, this type of gauge is utilised in dynamic systems such as turbine blade measurement (Neubert, 1967, p 20).

One advantage of this type of strain gauge is that the measurement voltages are generated by the application of strain to the crystal, thus requiring no external power supply.

4.6 OTHER STRAIN GAUGES

4.6.1 Inductive - The inductive strain gauge is currently used within the medical fraternity. This type of sensing operation is known as Respiratory Inductance Plethysmography. This type of gauge uses the variation in mutual inductance between the coils when they are moved to determine the amount of strain that has been applied by the patient's breathing.

4.6.2 Vibrating Wire - The vibrating wire gauge uses a piece of steel wire that is held in tension between two points. A vibration is then set up in the wire by use of a magnetic coil. The coil acts as the detector unit for the natural vibrating frequency of the wire. A comparison of the current natural frequency of the wire and the natural frequency when it was first installed is then made. Any difference in the two frequencies shows that there has been movement between the two points the wire is attached to. While suitable for large sensing operations such as buildings, this method of measurement is impractical for use on humans due to its size.

4.6.3 Fibre Optic - Fibre optic strain gauges operate by varying the wavelength of received light passing through the fibre optic cable as the applied strain varies at the detector. This is achieved by altering the fibre Bragg grating that is embedded within the cable. The Bragg grating controls the refractive index of the cable, allowing certain wavelengths to pass while attenuating others (Globalspec,2007, p 1). This type of strain measurement could prove suitable for this project. The main difficulty being that the use of fibre optics is a specialised field requiring expensive equipment and a high degree of technical proficiency when manufacturing fibre optic products.

4.6.4 Pneumatic - The pneumatic strain gauge operates by measuring the difference in pressure caused by the restriction in flow of a gas stream. Variations in the flow of gas into a sealed chamber alter the pressure felt within the chamber. This difference is then detected and recorded (Neubert, 1967, p 8).

4.6.5 Acoustic - The acoustic strain gauge operates by 'plucking' a taught wire which vibrates at a known frequency. If the strain on the wire is altered, then the frequency of the wire is also altered (Neubert, 1967, p 8). This strain gauge system is

complicated in comparison to other methods that are currently available and was not considered for this project.

4.7 PROJECT GAUGE

For the purposes of this project, it was decided to utilise the electrical resistance gauge. The main criteria that were used to determine which gauge to use are as follows:

Reliability - The electrical resistance strain gauge has been used in various industries for decades. Over that period of time it has proven itself to be an extremely reliable method for measuring relative strain and the movement of the material to cause that strain.

Low Cost - One of the objectives of this project was to design a low cost alternative to the current monitoring devices that are currently available. The strain gauge used in this project cost $10.00(AU).

Ruggedness - The metal foil gauge can be easily damaged if it is bent or the elastic properties of the gauge are exceeded. However, simple handling precautions will minimise the risk of this occurring. As the elastic strap is adjusted to a neutral position (no strain) when first fitted to a patient, the material limits of the gauge are not likely to be breached.

Error Compensation - The error compensation requirements for the chosen gauge are described in chapter 5.

Simplicity - This gauge and the associated electronics utilise a direct current power supply. This means that the manufacturing resources required are minimal with respect to complicated circuitry. The attachment of the gauge to the backing material is described in Appendix B2.

The gauge that is used in this project is KYOWA Strain Gauge: Type KFG-10-120-C1-11L3MSR. This gauge appears to be well suited to the task and should provide a low cost and reliable sensing element for the monitor system. The method of bonding and circuit operation is described in chapter 6.

5.0 PROJECT DESIGN OVERVIEW

5.1 INTRODUCTION

The design of the respiration monitor circuitry was completed using the Altium 6 Designer software package. This software package was chosen for the task as it contained an excellent reference library of components that were able to be edited to create new components as required.

The circuit design underwent several revisions and modifications as the research project progressed, with the complete circuit design for the project attached as Appendix B1.

5.2 CIRCUIT ASSEMBLIES AND TESTING

To facilitate in designing a simple yet effective circuit the design was broken down into three distinct and readily identifiable assemblies. The three assemblies are the sensor assembly, the amplifier assembly and the processor assembly.

The computer is used for downloading the code and uploading the sensed data. As the software component for this project is small, it was decided to incorporate discussion of any software requirements with the hardware assembly that utilises it. The block diagram shown as Figure 5.1 and gives a pictorial representation of the operation of the project.

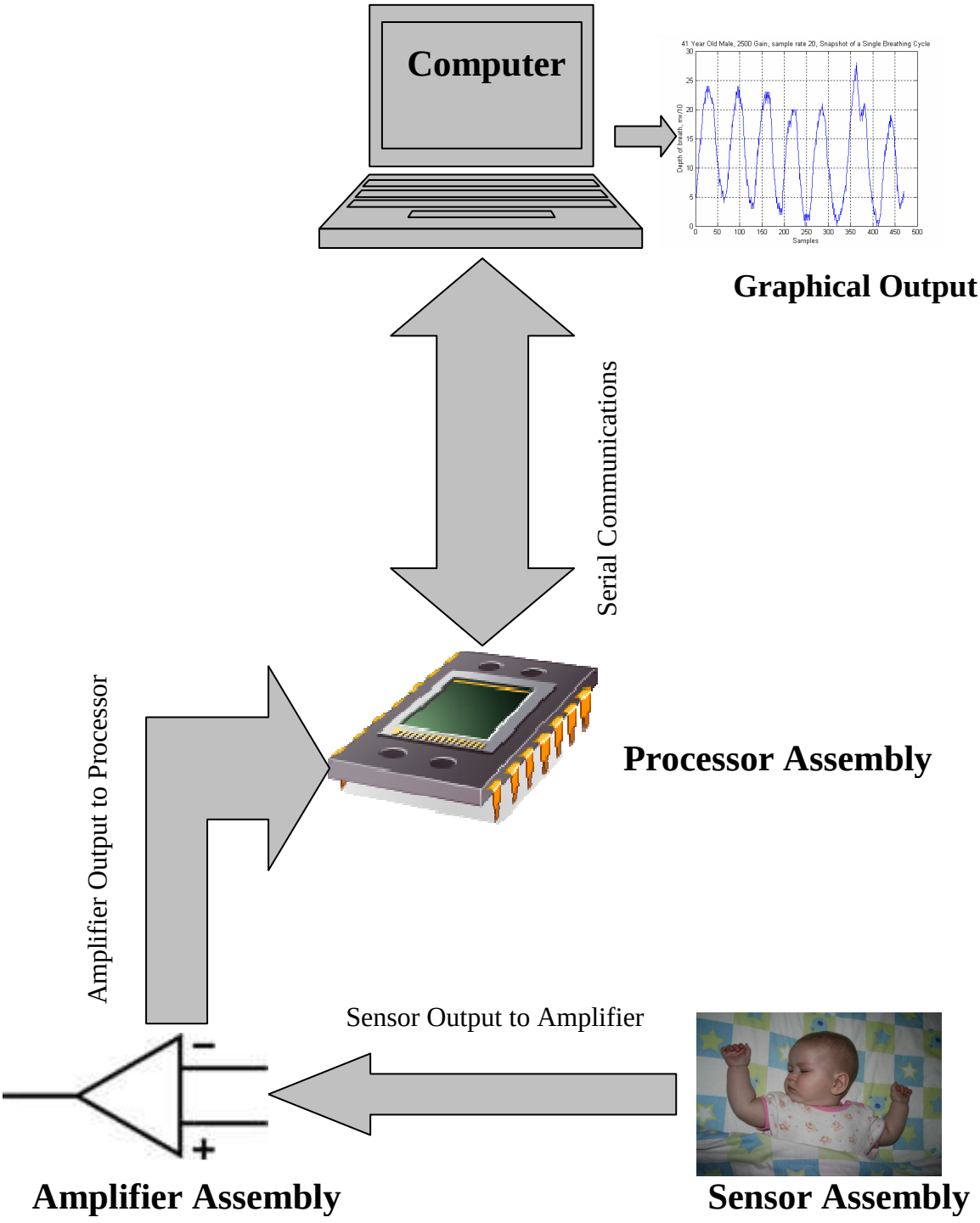

Figure 5.1 Project Block Diagram

Each of the separate assemblies is covered in the following chapters:

6.0 Sensor Assembly - The sensor assembly includes the design and construction details for using a metal foil strain gauge within the sensing unit. The operation and outputs from the whetstone bridge are discussed, with some of the typical outputs discussed.

7.0 Amplifier Assembly - The amplifier assembly includes the type of amplifier used and the justification for its use. Typical outputs from the amplifier are displayed and the problems that occurred during construction are presented.

8.0 Processor Assembly - The processor assembly includes the microprocessor and memory used and the justification for its use. The source code for the processor is also discussed and examples of the data that was stored in memory are displayed and discussed.

6.0 SENSOR ASSEMBLY

The sensor assembly is the most important part of this device. The basis for this design is based upon a similar assembly that has been developed by The Flinders Medical Centre (FMC). The sensor unit developed by FMC is designed for testing the strain placed upon the gut area of the human body; and is called “BME 1432, Strain Gauge Assembly”.

A copy of the procedure to manufacture the BME 1432 is available from the FMC upon request.

6.1 DESCRIPTION

The sensor assembly for this project was constructed using the following metallic foil strain gauge:

KYOWA Strain Gauge: Type KFG-10-120-C1-11L3MSR

Temperature Compensated for Steel

Gauge Factor 2.1 +/-1.0%

Resistance 120.4Ω +/-0.4Ω

This type of gauge was chosen because of its relative cheapness at only ten dollars per gauge. As can be seen in Figure 6.1, the small size of the gauge makes it ideal for this type of sensing operation.

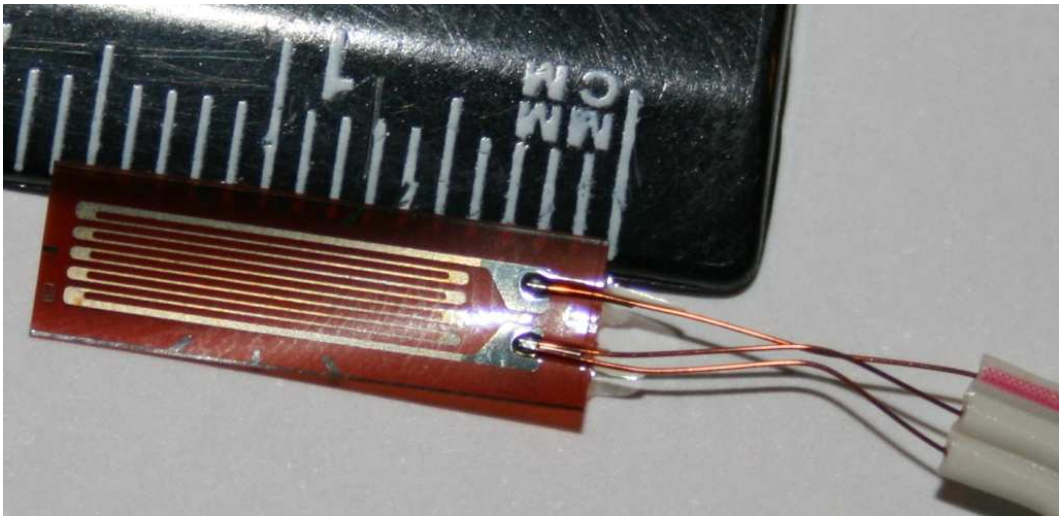

Figure 6.1 KYOWA Strain Gauge (Photograph by Author).

6.2 CONSTRUCTION

A review of the procedure supplied by the FMC showed that the strain gauge was bonded to a carbon weave piece of material. As no specific type of material was designated, the sensor material used in the project is made from silicon rubber. The complete details of which can be found in Appendix B2. Initial examination of this material showed that there was very little deformation or movement of the material when strain is applied in parallel with the weave of the material. To allow for adequate movement, the material was cut on the diagonal at an angle of 45 degrees. This allows for enough movement to be sensed by the gauge, without the danger of excessive movement causing the gauge to go beyond its elastic limits. A complete procedure for producing this type of sensor unit is contained in Appendix B2.

The bonding is done by the application of Locktite to the gauge which is then press fitted into place. After curing, the gauge, exposed wires and 20mm of cable are then covered in a protective layer of non corrosive silicon gel.

To allow the sensor assembly to be attached to the person being monitored, the silicon material is sewn onto a piece of elastic that is the same width as the backing material. This allows for a comfortable fit to the patient, whilst holding the sensor firmly against the chest. Figure 6.2 shows the sensor attached to the backing material.

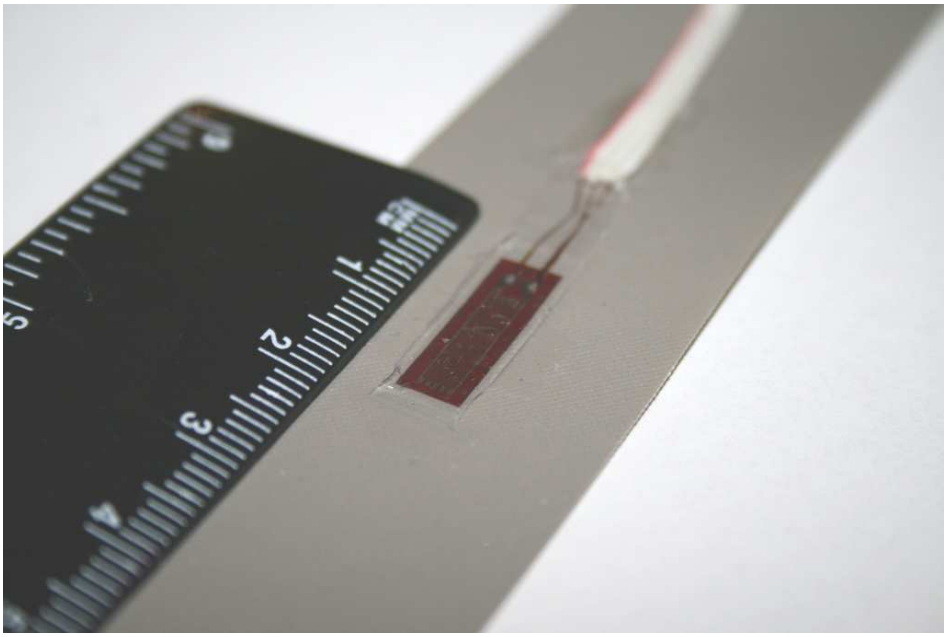

Figure 6.2 Strain Gauge and Backing

6.3 BACKING MATERIAL TESTING

The FMC did not provide for a specific material that should be used to attach the strain gauge to. The silicon material was chosen because it appeared to have a suitable amount of flexibility, while retaining an inherent strength to prevent accidental damage.

In order to determine the suitability of the material used for the backing, it was placed under a sheer test. The test was conducted in the Faculty of Engineering and Surveying. The material was cut into dog-bone shapes, as can be seen in Figure 6.3.

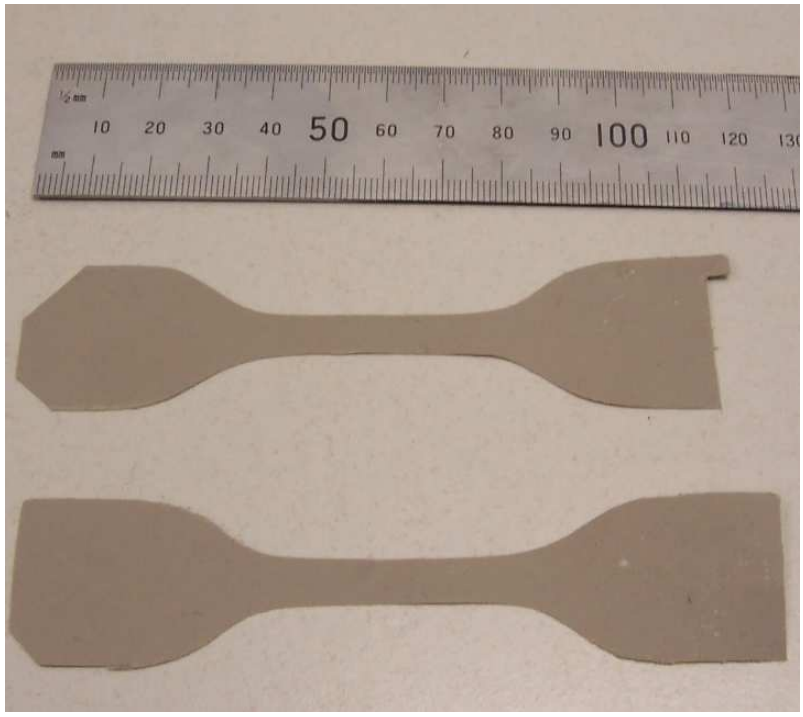

Figure 6.3 Tensile Test Shape for Silicon Material

The results of the test show that the material has a reasonable amount of linear movement when it is first placed under strain. The material was tested until destruction occurred. Figure 6.4 shows the results of the strain test that was conducted on the material. No external sources of information were consulted prior to this test being conducted; hence there are no external references available to compare this test data with.

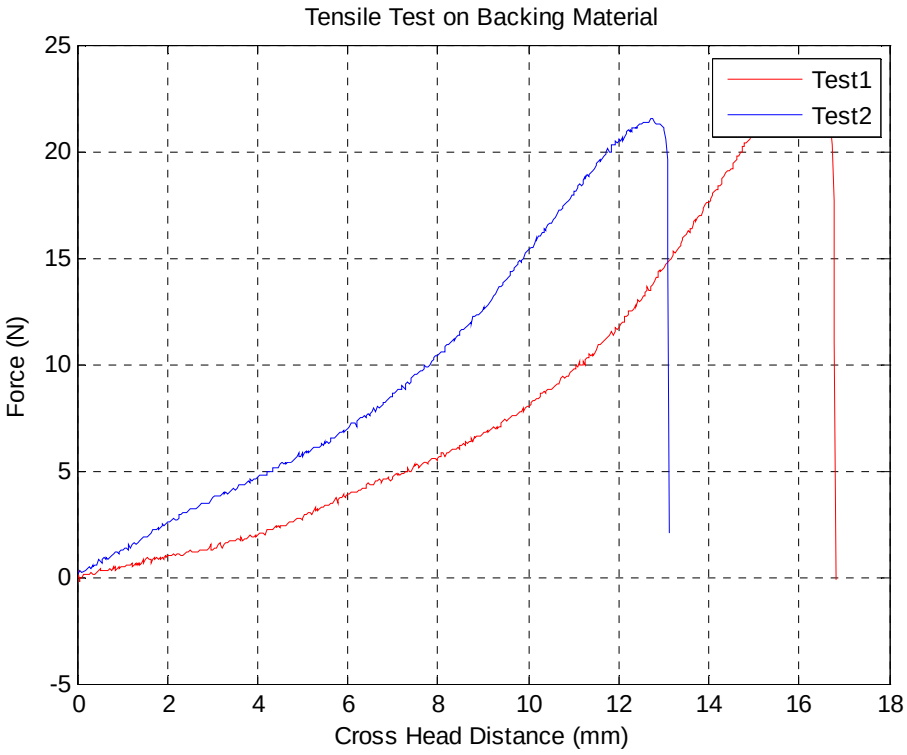

Figure 6.4 Strain Test of Silicon Backing Material

6.3.1 Material Analysis

As can be seen from Figure 6.4 two tests on the material were conducted. The MATLAB code and data for these tests is provided in Appendix C1. Analysis of the results reveals that the deformation when placed under strain is not perfectly linear for this material. However, this material does appear to have enough linear characteristics at the low end of the scale to suit the purpose for which it is being applied. The amount of movement experienced by the material is approximately 3 to 4mm for an adult and 1 to 2mm for a neo-natal child when used for respiration monitoring.

The actual movement experienced by the portion of material under the strain gauge is considerably smaller than the full material movement. No data has been collected as to the amount of actual movement of either the material under the gauge, or the gauge itself.

6.4 SENSOR ASSEMBLY CIRCUIT

The sensor assembly of the project is shown in Figure 6.5 and consists of an adjustable voltage regulator and whetstone bridge configuration. The purpose of this design is to provide a low voltage that exhibits a stable characteristic for the operation of the strain gauge sensing element. The output of the whetstone bridge is then used to supply the amplifier section of the circuit (Chapter 7).

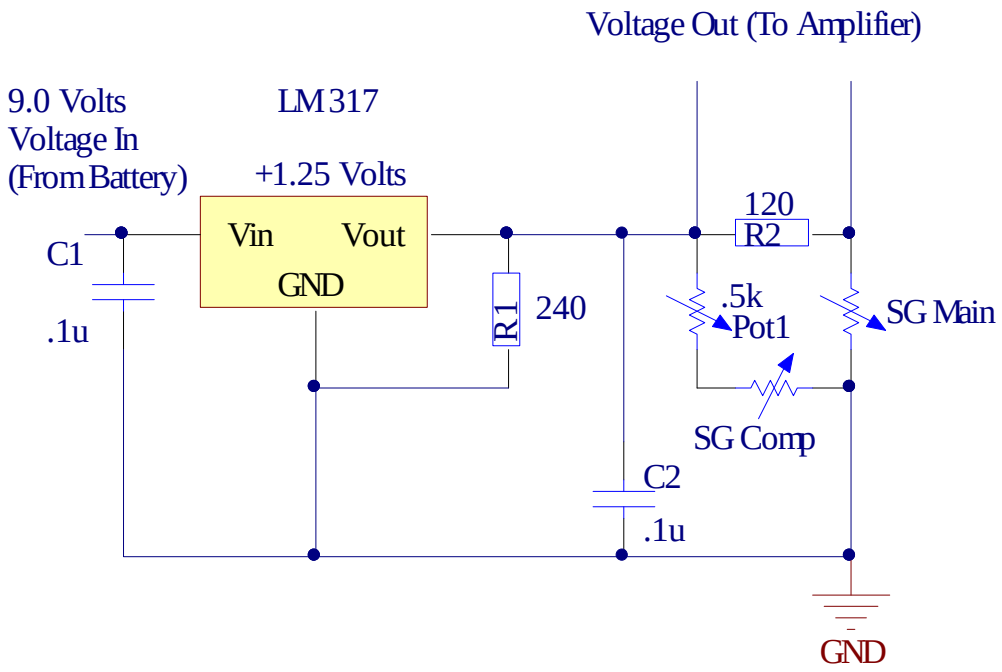

Figure 6.5 Sensing Assembly Circuit

6.4.1 Power Supply

The main power supply for the entire circuit is supplied by a 9.0 volt battery. Each section or sub-assembly of the circuit (sensor, amplifier and processor) has its own regulated voltage supply. The separation of the voltage supplies was deemed necessary due to electromagnetic interference (EMI) effects between each of the sub-assemblies.

The sensing circuit utilises a LM 317 voltage regulator for an independent power supply. The front page of the data sheet for this device can be found in Appendix D1. The regulator has an adjustable voltage range from 1.25 volts to 5.0 volts. The initial voltage supplied from the voltage regulator was set at 5.0 volts. This voltage was selected as it seemed a reasonable assumption that the minute changes in resistance of

the gauge would supply a correspondingly minute output voltage from the whetstone bridge.

However, during one experiment of the sensor assembly the gauge became warm and the author suffered some minor amount of discomfort to the chest. This occurred on only one occasion; however a further review of Kyowa literature indicated that 1.0V was a suitable supply voltage for a strain gauge assembly. Consequently it was decided to reduce the voltage supply to the sensor assembly. Reduction in the voltage to the sensor assembly has achieved two things; the risk of harm from burning to the person wearing the sensor has been reduced and the current draw from the main battery has also been reduced.

Due to supply difficulties a 1.0V regulator was not available; this is why the LM 317 voltage regulator was chosen. The measured value from the voltage regulator is 1.259 volts.

6.4.2 Whetstone Bridge

The principal of operation for the whetstone bridge used in the sensor assembly is described in chapter 4.4. Using equation 4.4 (reproduced below), it is possible to determine the output voltage of the whetstone bridge.

$$V = \frac{\Delta R}{R} \times E \times \frac{1}{4} \qquad \textbf{Equation (4.4)}$$

V= Output Voltage (V)

ΔR = Resistance Change (Ohms)

E = Applied Voltage (V)

However, the change in resistance for this design when the patient is breathing is not known. Consequently equation 4.4 must be rearranged so that the resistance variation (ΔR) becomes the subject of the equation. Equation 6.1 is used to calculate the variations in the strain gauge's resistance whilst in operation.

$$\Delta R = \frac{V \times R \times 4}{E}$$ **Equation (6.1)**

ΔR = Resistance Change (Ohms)

V = Output Voltage (V)

E = Applied Voltage (V)

Using Equation 6.1 the power dissipation from the strain gauge was calculated to be

To determine the strain that is applied to the sensor assembly, equation 4.3 (reproduced below) is substituted into equation 4.4. Rearranging the equation to make the strain (ε) the subject of the equation, results in equation 6.2.

$$\frac{\Delta R}{R} = k \times \varepsilon$$ **Equation (4.3)**

$$\varepsilon = \frac{V}{E \times k} \times 4$$ **Equation (6.2)**

ε = Applied Strain

k = Gauge Factor (From Manufacturer)

V = Output Voltage (V)

E = Applied Voltage (V)

By calculating the strain that is applied to the sensor assembly it is then possible to compare the different depths of respiration between patients and also to make comparisons between various age groups.

As there are variations in manufacturing tolerances, the components used in the construction of the whetstone bridge are not exactly the same as specified by their manufacturer. The components chosen for the prototype have a tolerance of 5%. This tolerance is sufficient for the system to operate correctly. To overcome this problem a potentiometer was installed in one of the legs of the bridge. This allows for the bridge to be balanced before any readings are taken by the processor assembly. The potentiometer used is $500\,\Omega$'s and is adjusted as required after the patient is wearing the sensor assembly. This size potentiometer proved suitable for all tests that were conducted on the prototype. After each test the potentiometer was removed from the circuit and the resistance value measured. The measured values were within +/- 5% of

the actual strain gauge resistance. Because each person will wear the sensor differently and no two people will have the same amount of tension on the elastic (hence different initial applied strain), the adjustment must occur after fitment of the device, but before sensing operations commence. The main purpose of this adjustment is to ensure that the output from the whetstone bridge is greater than zero volts.

Due to the analogue to digital conversion that must take place for the strain readings to be recorded the initial output of the sensor assembly cannot be either zero or a negative voltage. For this reason the sensor assembly always starts out in an unbalanced condition. It was decided to have an unbalanced bridge instead of a dc offset in the amplifier assembly. This method was chosen as the starting strain and hence sensor output voltage can never be known due to the sensitivity of the gauge. This sensitivity means that a simple adjustment of the sensor assembly can seriously affect the input into the amplifier. If the amplifier becomes saturated then the processor assembly cannot differentiate between samples and returns a maximum value for each sample.

The resultant output of the sensor assembly therefore does not resolve around a zero value. To overcome this problem, a MATLAB script has been developed to produce a normalised output regardless of the apparent strain being recorded. The methodology for this code is described in chapter 10.

6.4.3 Sensor Assembly Outputs

The output from the sensor assembly has a different starting value each time the respiration monitor is first switched on, the reason behind this was explained in 6.5.2. For this reason the values given below and the calculations that are conducted are of results obtained during a typical monitoring procedure.

Prior to testing the sensor assembly on a patient, a series of readings were taken to measure the output of the strain gauge when it is placed in a neutral position. To obtain a neutral position, the sensor assembly is laid down on a bench and the strap bought under enough tension so as to remove any folds or buckles in the silicon backing material. Each end of the strap is then made secure and power is supplied to the whetstone bridge.

When tested in the neutral position, the output from the whetstone bridge was found to be of a constant value, with no variation in output detected over a 30 minute period.

This test was conducted on four separate occasions and returned the same stable result each time.

To simulate a real world environment where the sensor assembly would be attached to a patient prior to each sleep period, the sensor was attached as per the testing procedure on four different occasions and the voltages recorded in Table 6.1. Each of these readings was taken after the individual had fully exhaled and held their breath for several seconds. It was not feasible to test this on a neo-natal child. After the sensor assembly had been strapped to the patient, the following voltages were measured at the output of the whetstone bridge.

Table 6.1 Whetstone Bridge Outputs (Fully Exhaled – Pre Amplification)

Subject	Age	Voltage Readings (mV)
Male	41 years	1.2mV
Male	41 years	1.8mV
Male	5 years	0.85mV
Male	5 years	0.97mV

As can be seen from table 6.1 no two readings are identical and despite the test subject holding their breath some minor fluctuations were noted. These variations were +/- 0.05mV in magnitude. This equates to a variation in stability of 4.12 to 5.15%. This variation in stability was deemed to be of a low enough magnitude to have virtually no effect on the subsequent tests.

This phenomenon was also noted when the sensor assembly was tested on the System 5000. When the zeroing component of the System 5000 test was conducted, variations in strain were noted when the sensor assembly was placed on a bench prior to attachment to the test subject.

Using equation 6.2 this information is now used to calculate the strain applied to the gauge prior to recording the respiration rate. The results of these calculations are displayed in table 6.2.

$$\varepsilon = \frac{V}{E \times k} \times 4 \qquad \textbf{Equation (6.2)}$$

ε = Applied Strain

k = Gauge Factor (From Manufacturer)

V = Output Voltage (V)

E = Applied Voltage (V)

Table 6.2 Gauge Strain (Fully Exhaled – Pre Amplification)

Subject	Age	k	E (V)	V (mV)	ε
Male	41 years	2.1	1.2596	1.2mV	.001815
Male	41 years	2.1	1.2596	1.8mV	.002722
Male	5 years	2.1	1.2596	0.85mV	.0012854
Male	5 years	2.1	1.2596	0.97mV	.0014668

This series of calculations shows the type of strain that the sensor package is placed under prior to any recordings being taken. This data is not useful at this stage of development because there is no comparable data available for other materials or strain gauges in this configuration. However, if the design is to proceed beyond the prototype stage then further investigation will be required to find the optimal strain gauge and backing material.

6.4.4 Sensor Assembly Output Whilst Monitoring

The only recordings taken from the whetstone bridge during monitoring operations were done using a multimeter. It was not deemed feasible to take qualitative readings due to the constantly varying voltage as the strain on the sensor assembly was varying.

A multimeter was used to confirm that there was a varying output voltage from the whetstone bridge as strain was applied to the sensor assembly.

7.0 AMPLIFIER ASSEMBLY

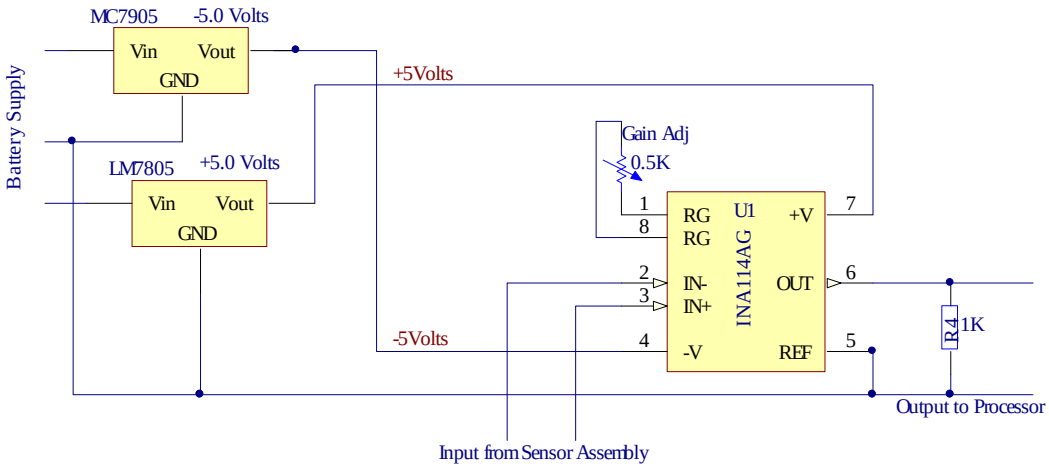

Figure 7.1 Amplifier Assembly Circuit

The amplifier assembly consists of a separate power supply and a Burr-Brown instrumentation amplifier.

7.1 POWER SUPPLY

The power supply for the amplifier assembly is provided by two volt regulators. The amplifier requires a positive and negative 5.0 volt supply; the regulators used are the LM7805 and MC7905 respectively. The front page of the data sheets for these devices can be found in appendices D2 and D3. A separate power supply was incorporated into the amplifier in an attempt to overcome a voltage drift problem that occurred when experiments on the circuit were conducted.

7.2 AMPLIFIER

The amplifier assembly is a Burr-Brown instrument amplifier. The front page of this data sheet can be found in Appendix D4. The Burr-Brown was chosen for the project because it offers a simple method for adjusting the gain of the amplifier. The characteristics of this amplifier included a low offset voltage of $50\mu V$ and a low drift rate for temperature variation of $0.25\mu V/^{o}C$.

Due to the small variations in input voltage from the sensor assembly, the low drift rate is of particular importance in this design. If the drift rate is too prominent then it can have a serious affect on the output of the amplifier over periods exceeding five minutes.

7.3 AMPLIFIER GAIN

The gain for the amplifier assembly is controlled by a single potentiometer; this offers the ability to adjust the gain required depending on the patient wearing the device. It was noted during the operational tests that the gain required for a neo-natal child was considerably higher than was required for an adult. A comparative analysis of the gain used for this project is provided in chapter 10. The main advantage of the easy adjustable gain of the amplifier is that it can allow for both qualitative and quantitative monitoring to be conducted for the device.

To test that the amplifiers gain was of sufficient magnitude for the processor assembly to discriminate between variations in strain, the voltages from sensor assembly the theoretical output for the amplifier were calculated and compared with the actual outputs. Table 7.3 shows the calculation and actual output for the amplifier assembly.

Table 7.1 Amplifier Output, Theoretical and Actual

Subject	Age	Input	Gain	Theory	Actual	Difference
Male	41 years	1.2mV	100	120mV	122mV	+1.01%
Male	41 years	1.8mV	625	1.125	1.115V	-0.89%
Male	5 years	0.85mV	1000	0.85V	0.855V	+0.58%
Male	5 years	0.97mV	2500	2.425V	2.490V	+2.61%

To confirm that the output of the amplifier would reflect a change in voltage as the strain on the sensor assembly was varied, the sensor assembly was affixed to a work bench at one end. The other end was left loose, then while holding the loose end of the sensor a small amount of strain was applied by hand in an attempt to simulate a breathing pattern. Variations in the amplifier voltage were noted on the voltmeter connected across the amplifier output.

7.3.1 Qualitative Monitoring

For this project qualitative testing is defined as using the supplied data to produce an output that can be used to show a sinusoidal breathing pattern of the patient. The display of this breathing pattern is distinctly sinusoidal in shape and can be used to show a finer detail of a patient's breathing pattern. During testing of the project the

gain was set to 2500 and the sampling rate was set to 20Hz (20 samples per second) to achieve a qualitative monitoring result. After conducting multiple experiments where the gain and sampling rates were varied, it was found that it is the sampling rate that determines the accuracy of the output. This result is in line with Nyquist's Sampling Rate (Nise, 2000, p 781) which states:

> *"The sampling rate must be at least twice the bandwidth of the signal, or else there will be distortion."*

Figure 7.2 is an example of the output from a qualitative test that was conducted on the author. Figure 7.3 is a snapshot of a single breathing cycle from the same test.

Further monitoring was performed in an effort to find the optimal gain and sampling rates. It was found that when the gain was increased beyond approximately 2,500 there was no appreciable increase in either quality or definition of the output.

Due to the sensitivity of the sensor assembly in registering changes in chest movement, the graphical output becomes cluttered with interference when the gain was increased beyond 2,500.

As this test was conducted under controlled conditions, the test subject was induced to take a deeper breath than normal (the sigh). This event can be seen on Figure 7.2 after approximately 350 samples.

As can be seen from viewing Figures 7.2 and 7.3 the design is capable of recording a definite sinusoidal output in response to a patient's breathing pattern. Figure 7.2 appears to be an uneven sinusoidal output on first viewing, however this appearance is misleading due to the sheer quantity of samples being taken and recorded over the period of testing. Analysis of the data stream used in the MATLAB code also shows the varying output from the amplifier. The code for this output is provided in Appendix C6.

The difference in each recorded depth of breath in this example is due to the different depths of breath taken by the person being monitored. The drift problem that was mentioned in Chapter 7.2 was not apparent in this test as it was conducted over a short time frame of only 23 seconds.

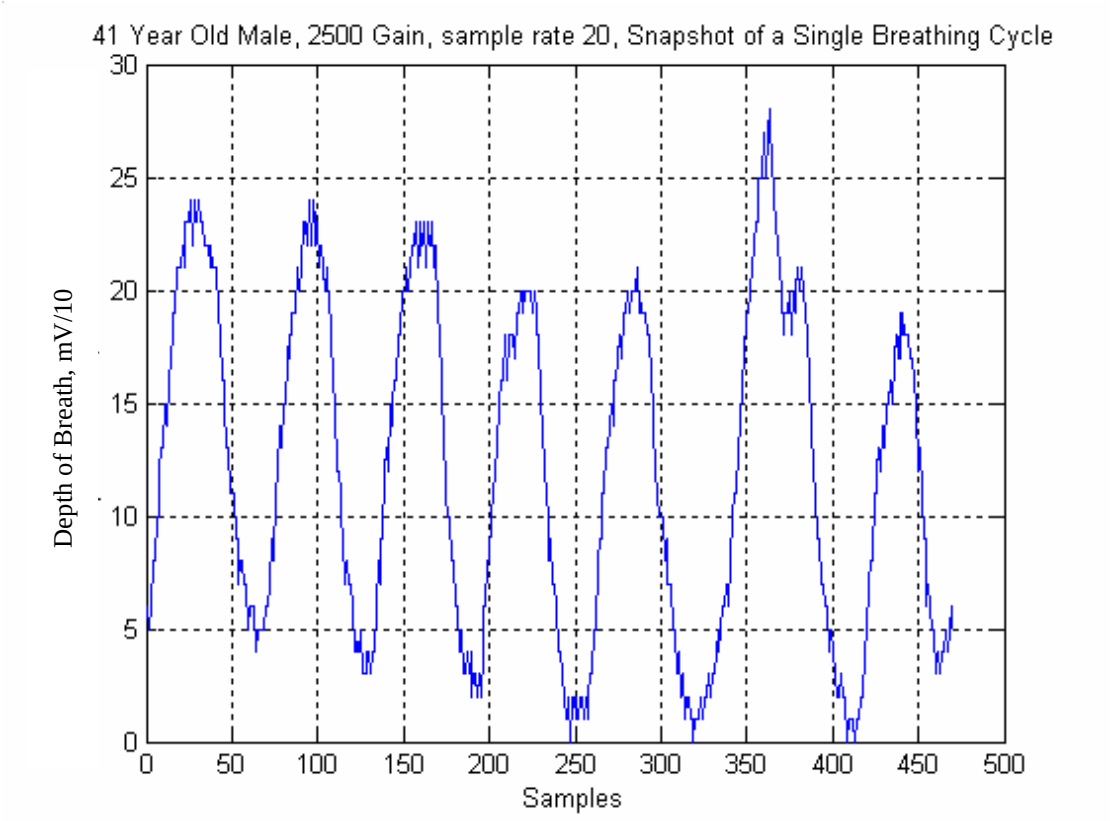

Figure 7.2 Qualitative Monitoring on Author (2500 Gain)

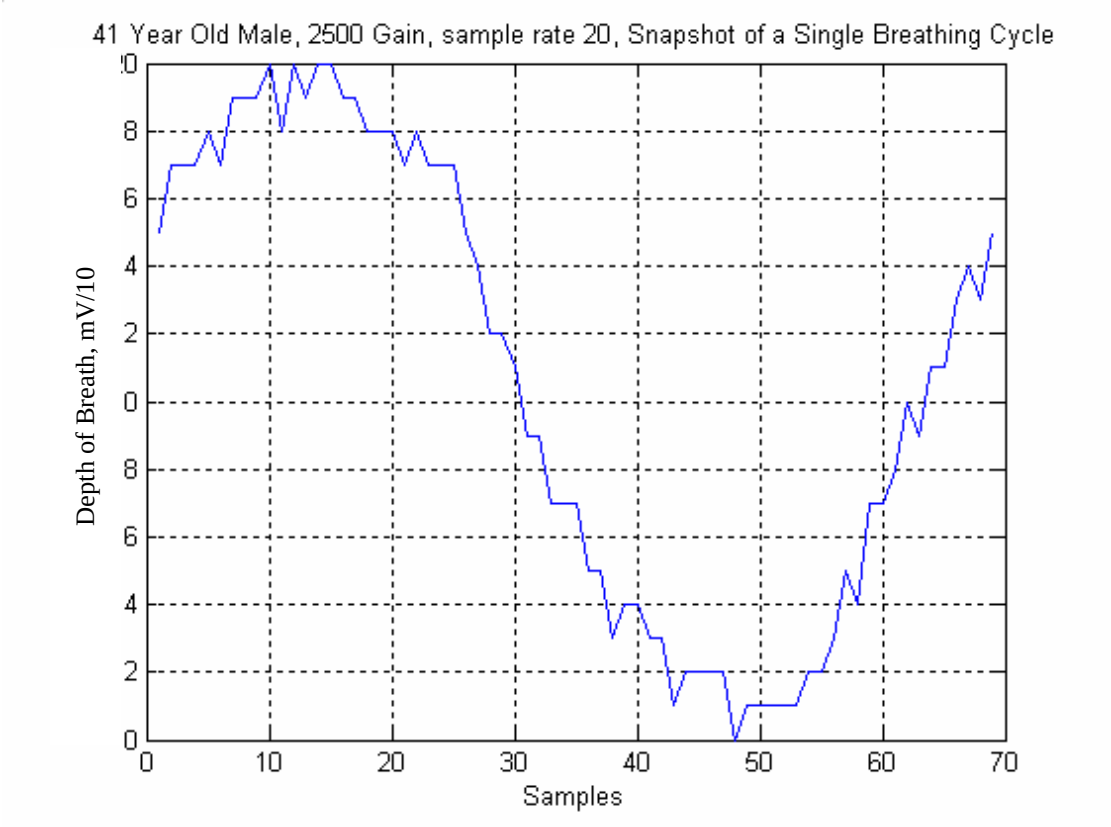

Figure 7.3 Qualitative Monitoring Snapshot for Single Breath of Author

7.3.2 Quantitative Monitoring

For this project quantitative testing is defined as using the supplied data to produce an output that can be used to show an event occurrence. An event occurrence is defined in this paper as the peak or trough of a graph during the inhalation or exhalation of a breath. This information can be used to show the breathing events of a patient over a longer period of time. Figures 7.4 and 7.5 show two examples of the output from quantitative tests that were conducted on the amplifier assembly, during the testing of the amplifier the gain was set to 278 and 139, with a sampling rate set to 20Hz (20 samples per second) for both tests.

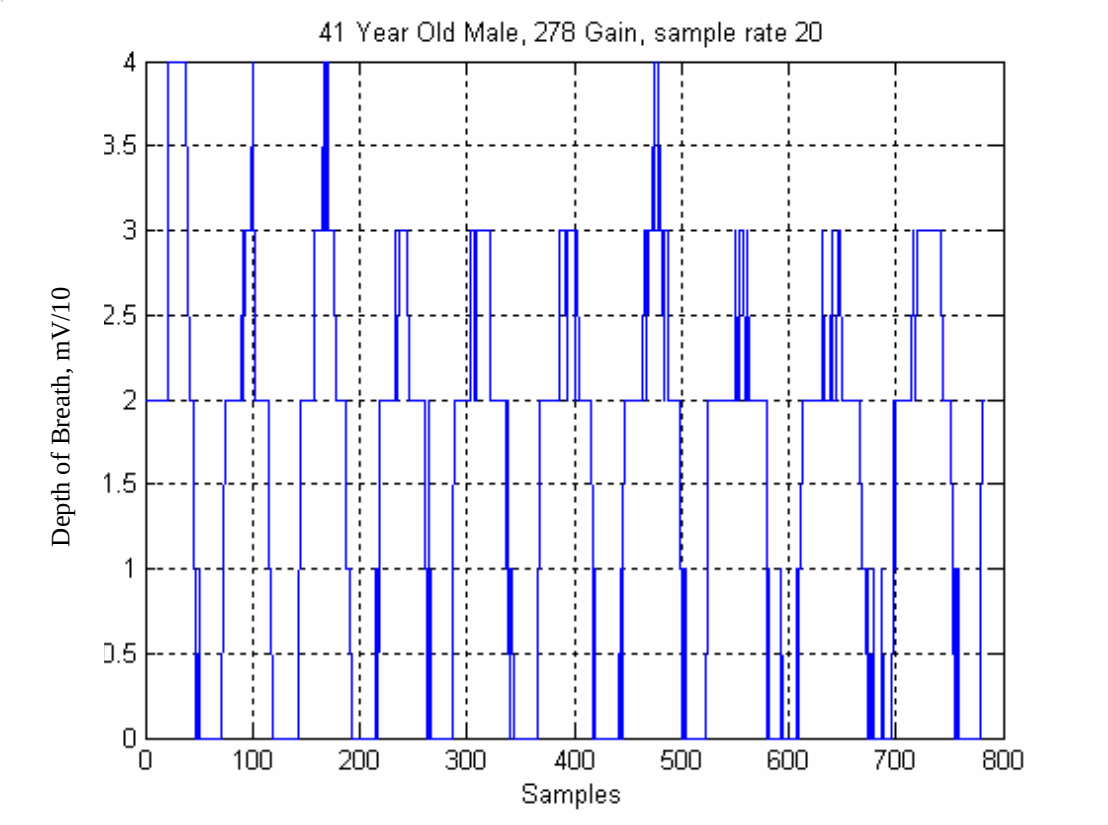

Figure 7.4 Quantitative Monitoring Snapshot on Author (278 Gain)

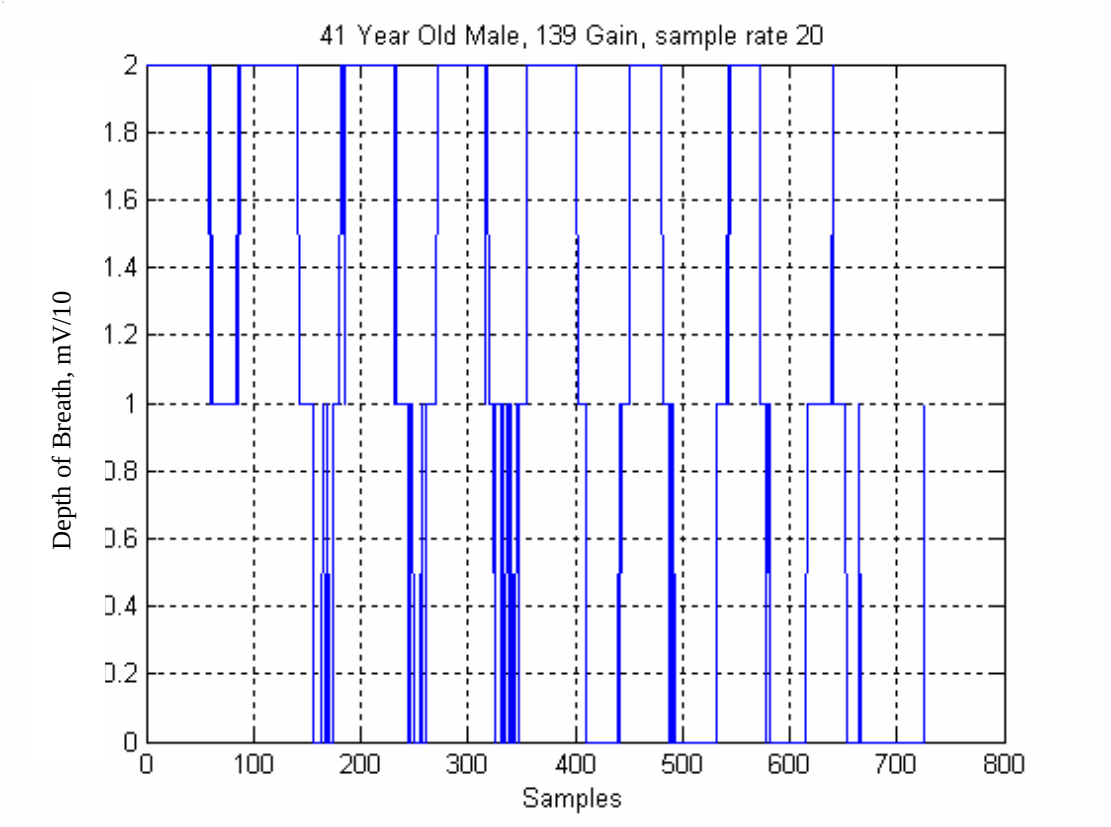

Figure 7.5 Quantitative Monitoring Snapshot on Author (139 Gain)

From Figures 7.4 and 7.5 it is possible to deduce that the change in gain is responsible for the changes in the quality of the displayed data. As the gain is increased, so the quality of the data is increased, with the output starting to take on a qualitative appearance when the gain is increased above 500. Snapshots of single breaths for low gain monitoring reveal that the output is square wave in shape, and provide no new information for analysis.

Figure 7.6 shows how a more sinusoidal output is achieved when the gain is raised to 625. Figure 7.7 is a snapshot of a single breathing cycle from the same test. This monitoring was conducted to confirm that it is the gain of the amplifier, and not the sampling rate of the processor that has been used to determined the result.

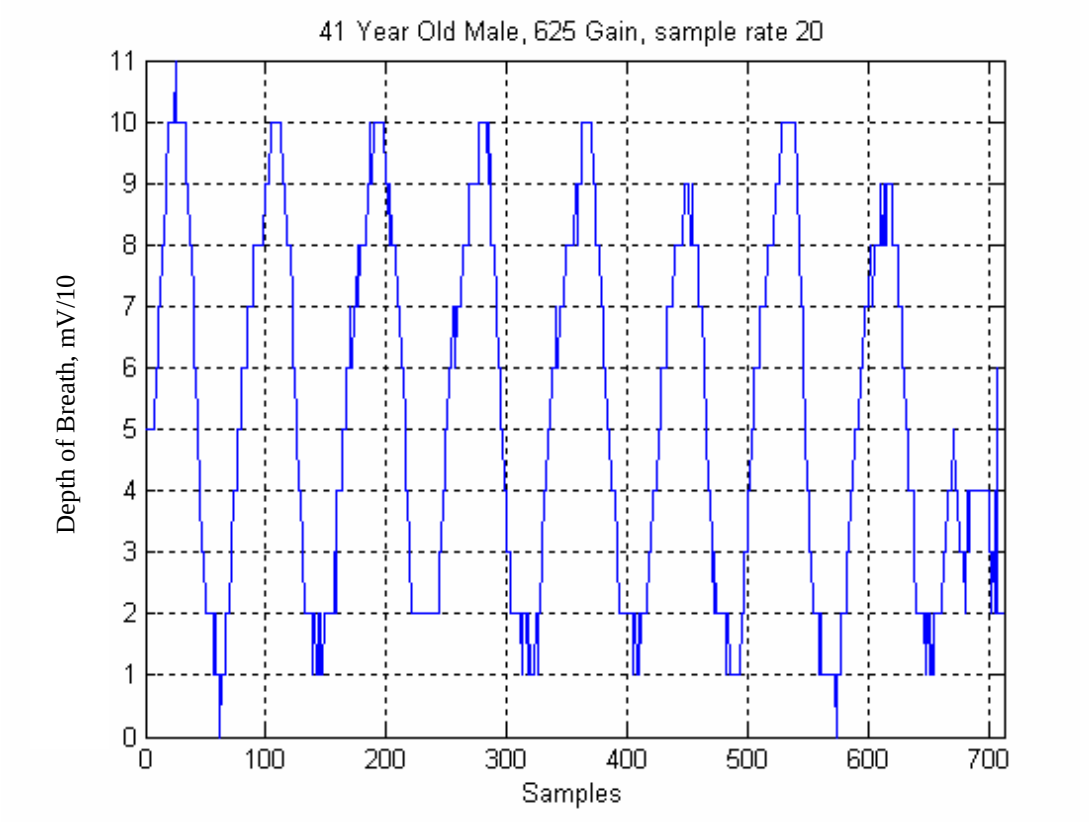

Figure 7.6 Quantitative Monitoring on Author (625 Gain)

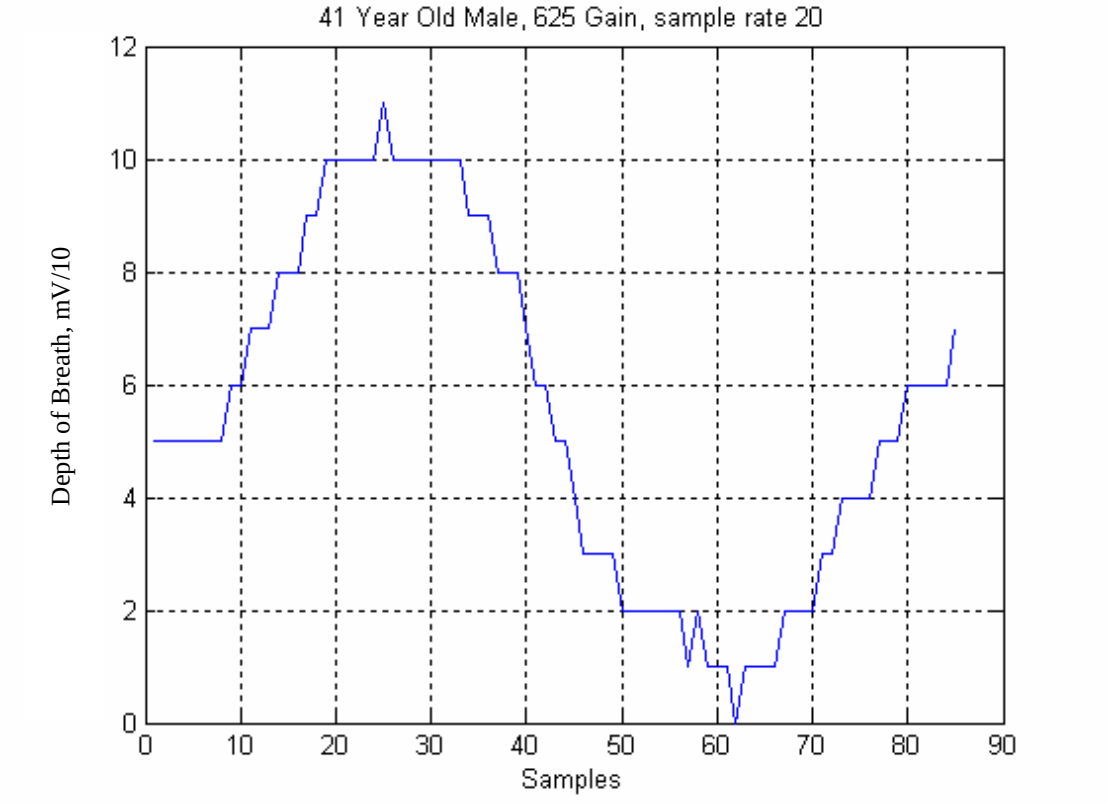

Figure 7.7 Quantitative Monitoring Snapshot on Author (625 Gain)

7.3.3 Qualitative Versus Quantitative Monitoring Requirements

As was demonstrated by the results from testing the outputs from the amplifier assembly, this design is capable of monitoring not only the physical changes that occur whilst a person is breathing (quantitative), but can be used to look closely at a person's actual breathing pattern (qualitative).

It is envisioned that there is an application for both types of monitoring to be conducted in the future with this design. The qualitative monitoring has the potential to be of use when the data is used to look at a patient's breathing in finer detail. The quantitative monitoring results from the testing are the mainstay of this project. If the respiration monitor is used purely as a device to monitor a patient's breathing then the quantitative setting of low gain is adequate for the task. The qualitative monitoring results from the testing have shown themselves to be an extension of the quantitative results. For this reason variations in the gain resistor will also determine the ability to select the correct qualitative recording parameters. In hardware, the installation of a multi-pole switch can be incorporated to change the value of the gain resistor. The final values of optimal gain resistors would be determined after further testing and in consultation with medical personnel regarding the exact parameters that would need to be monitored and recorded.

7.4 OUTSTANDING AMPLIFIER ISSUES

The amplifier in its present configuration is plagued by a drift in the quiescent output. This drift is intermittent in nature and irregular in amplitude. As stated in chapter 6.5.3 the output of the whetstone bridge can be considered constant in value for testing purposes. Measurements taken concurrently at the input and outputs of the amplifier assembly show that the input was stable and the output was varying. In an attempt to remove the source of drift, electrolytic capacitors were installed at the inputs to the amplifier and filtering capacitors on the output of the voltage regulators. These components failed to remove the drift, leaving it as an outstanding issue that must be resolved for the device to operate correctly.

8.0 PROCESSOR ASSEMBLY

The processor assembly consists of a separate power supply, PICAXE-18X microprocessor and 24FC256k EEPROM memory chip.

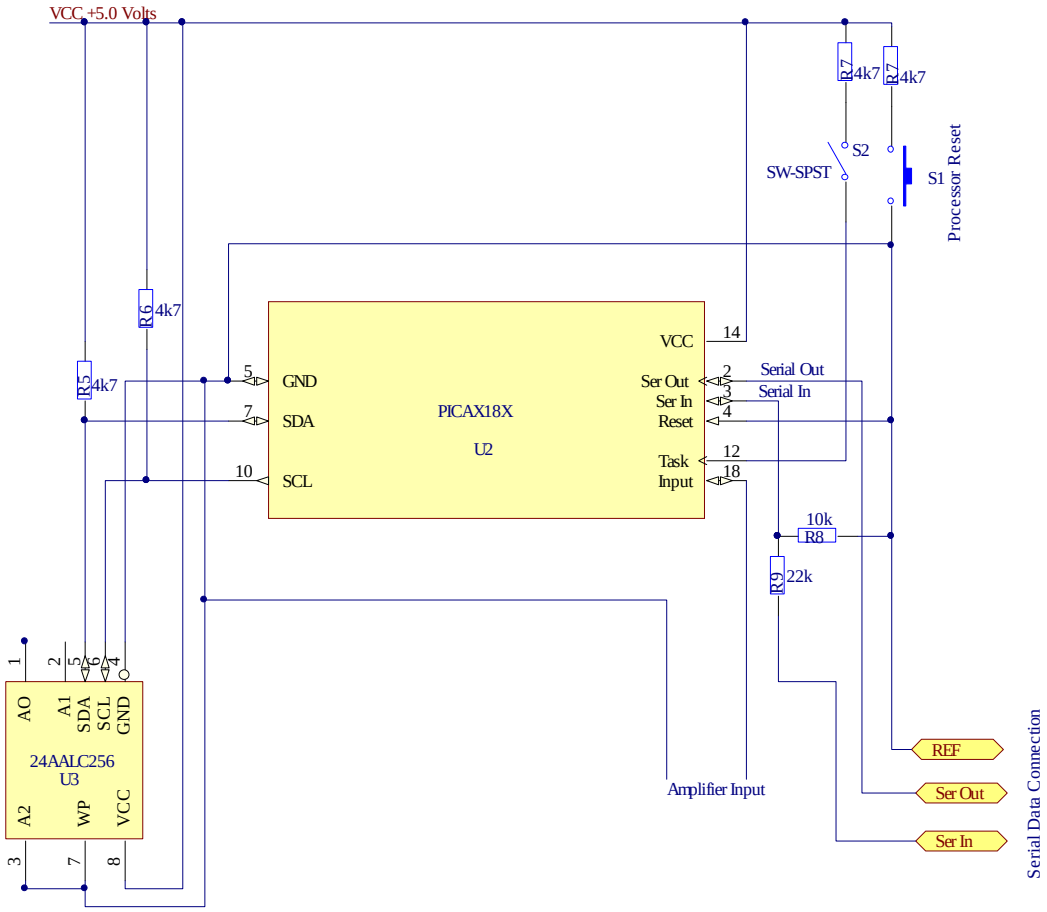

Figure 8.1 Processor Assembly Circuit

8.1 POWER SUPPLY

The power supply for the processor assembly is provided by a single volt regulator; the regulator used is the LM7805. The front page of the data sheet for this device can be found in Appendix D2. A separate power supply was incorporated into the amplifier in an attempt to overcome a voltage drift problem that occurred when experiments on the circuit were conducted. Unfortunately this did not solve the drift problem. However, when power was removed from the processor assembly the drift in the amplifier remained. This indicated that the problem is isolated to the amplifier assembly and is not caused or affected by the processor.

8.2 PROCESSOR ASSEMBLY AND MEMORY

8.2.1 Processor

The processor chosen for this project is the PICAXE-18X. This processor was chosen due to the following factors:

Programming - The PICAXE-18X is very simple to program. It utilises a form of BASIC that has been developed for this family of processors. This type of programming proved to be very useful in the development of the code required for the processor.

Cost - As one of the objectives of the project was to develop a low cost monitoring device, the low cost of this processor at less than $20.00(AU) for a single unit makes it ideal for this project.

Accessibility - The processor is programmed via the serial port on a Windows based machine. This means that changes in the code can be implemented by simply connecting a three pin plug to the processor and downloading the code from the complier.

Processor Speed - The PICAXE-18X processor runs at 4 MHz. This clock speed has proved to be more than acceptable for the project. As the Nyquist rate for a neo-natal child is 2 Hz in a worst case scenario, as discussed in chapter 2.1.3, a faster processor was deemed to be unnecessary.

Analogue to Digital Conversion – This processor uses a 10 bit analogue to digital converter (ADC). Using a 10 bit ADC allows for 1012 different voltage levels to be utilised when recording the sensed data into memory. This level of discrimination proved to be sufficient to accurately record the output from the amplifier assembly.

8.2.2 Memory

The memory chosen for this project is a 24FC256k EEPROM. This memory size is only 256kbytes. It was deemed prudent to utilise a smaller memory chip during the design of the respiration monitor as a larger memory size would not enhance the development process. The front page of the data sheet for this device can be found in Appendix D5.

The write cycle for this device is 5ms. As was discussed in chapter 7 a sampling rate of 50ms (20 samples per second) was found to provide a suitable signal for the output from the amplifier to the processor. This means that there is ample time for the current write operation to be completed before the next write cycle is required.

8.3 PROCESSOR ASSEMBLY OPERATION

Programming is done on the complier that is supplied from the manufacturer of this microchip. The complete code listing for the processor assembly is listed in Appendix B3. To achieve the aim of keeping the project as simple as possible, the processor assembly acts as a data logger, and as previously stated the processor is programmed in BASIC. All processing of the data gained from the processor assembly is conducted external to the microprocessor. The flow chart for the operation of the processor program is shown below in Figure 8.2.

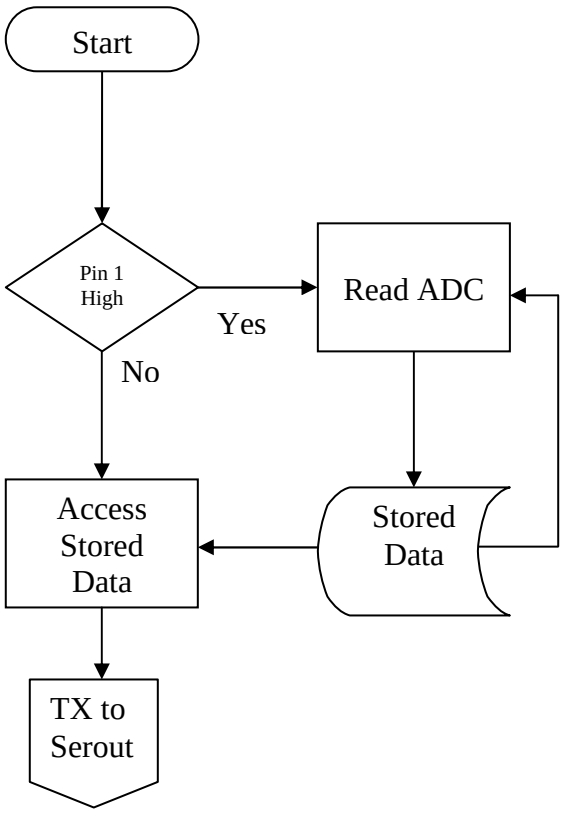

Figure 8.2 Processor Assembly Flowchart

The processor hardware is programmed via the serial input pin, which can be seen in Figure 8.1. Each time the parameters within the program are changed, the new code is downloaded into the processor. Once the code is installed, the reset button is activated.

This allows the processor to decide which protocol to follow, to start sensing and storing data operations, or to start reading the memory and transmitting the stored data back to the computer via the serial out pin and the computer serial communications port.

The processor assembly has two modes of operation, sensing and data recording or data download. Each of these operations will be discussed below. The procedure for downloading new code, conducting sensing operations and data retrieval and graphing are detailed in Appendix B4. Figure 8.3 is a display of the programming editor used to code the microprocessor.

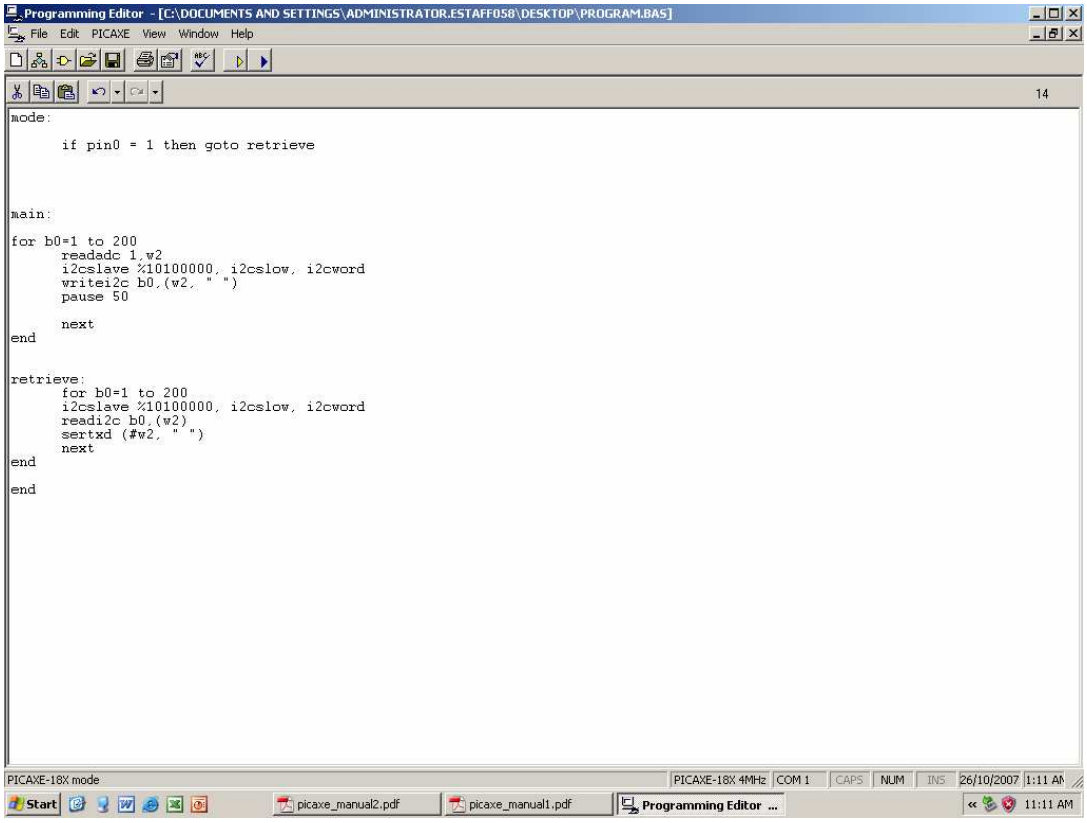

Figure 8.3 Programming Editor

8.3.1 Sensing and Data Recording Code

The default setting for the processor assembly is the sensing operation. Thus, when the device is first activated, it starts recording the output from the sensing assembly via the amplifier assembly immediately. The data is stored in the EEPROM in consecutive memory locations, with write operations continuing until power is removed from the processor. The complete data download procedure is detailed in Appendix B4.

The data is stored as a ’10 bit’ binary word, with each word requiring two memory locations. Thus a 256kbyte memory chip has 128k memory locations for storage. This allows for 106 minutes of sensing operations to be stored (at a sample frequency of 50Hz). As neo-natal children can sleep for up to 10 hours (in the author’s experience) this amount of storage is patently insufficient for extended sensing. Requirements for future devices are discussed in chapter 10. However, for prototype testing this size memory proved adequate.

The reason a’10 bit’ word in memory is used is because the ADC is a 10 bit device. The advantage of using a 10 bit ADC is that the converted analogue signal maintains a high degree of resolution after the conversion.

The following code is used by the processor to place the sensed data into its memory location.

```
for b0=1 to 200
        readadc 1,w2
        i2cslave %10100000, i2cslow, i2cword
        writei2c b0,(w2, " ")
        pause 50
        next
end
```

As can be seen from the above code, the data is saved to an external source; in this case the “i2cslave” denotes the memory device the data is stored in.

The sensing operation will continue for the duration of the loop that is defined by the user. Using a sample time of 50ms, for each hour that sensing is required the loop must be 72,000 iterations long.

8.3.2 Data Retrieval and Code

The data is retrieved by activating S2, a single pole single throw switch. This applies an active high to pin 12 of the microprocessor. The complete data retrieval procedure is detailed in Appendix B4.

The data is transmitted via the serial out pin on the processor to the computer serial communications port. The data is read by the computer’s hyperlink terminal, where it is displayed in numerical format. The data from the hyperlink terminal can then be saved into either a text document or placed directly into the MATLAB code or similar

software for graphing purposes. The data is transmitted to the computer at a rate of 4,800 baud. As this is a slow transfer rate, it is impractical at this stage to download large volumes of data. Future enhancements to improve this data transmission are discussed in chapter 9.0.

Examples of the MATLAB code and the raw data used to produce the graphical display are in appendices C5 and C6. The MATLAB code is used to remove any imbalance caused by the whetstone bridge. This is achieved by searching for the smallest reading stored in the sensed data and subtracting it from all other readings. When graphed this way the lowest reading will be zero for every graph produced, regardless of the data input. This allows for a meaningful comparison between graphed data sets, as all use zero as the starting reference on their respective graphs.

Once the data has been retrieved, S2 should then be opened in preparation for the next sensing operation. If S2 remains closed, then restarting the device automatically initiates the download procedure.

The data will remain in memory until it is overwritten by the next sensing operation.

The following code is used by the processor to remove the data from memory for transmission onto the serial out line:

```
retrieve:
        for b0=1 to 200
        i2cslave %10100000, i2cslow, i2cword
        readi2c b0,(w2)
        sertxd (#w2, " ")
        next
end
```

9.0 DISCUSSION AND CONCLUSIONS

9.1 INTRODUCTION

The use of strain gauges to measure the relative motion of a body has been in use for decades as discussed in chapter 2. The monitoring of a person's respiration is considered to be of extreme importance within the medical fraternity as the examples in chapter 3 show. The aim of the project in marrying these two principles together into a workable, yet affordable respiration monitor have been shown to be possible from this project.

9.2 DISCUSSION

The sensor assembly for this project is the most important of the assemblies within this project. It was quickly realised that failure to manufacture a useable device would render the remainder of the project untenable. With the procedure for bonding of the strain gauge to the silicon material proven to be useable, the manufacture of the sensing device was a success.

Incorporation of the sensor assembly into a whetstone bridge is the only viable method for producing an output that is useable for the amplifier assembly. The whetstone bridge has proven to be stable over many hours. The ability to provide the voltage offset that is needed by the ADC to prevent false zero readings has helped keep the hardware requirements for the amplifier assembly to a minimum.

The amplifier assembly has proven to be problematic in this project. Despite repeated attempts to rectify the voltage drift at the output, the problem remains. The actual amplifying process and the ability to change the gain as desired for a particular sensing operation was of great benefit during the design process and the subsequent monitoring tests of different age groups.

The use of a PICAX-18X processor and associated memory greatly reduced the amount of time required for programming. The use of a BASIC programming language and documentation supplied from the manufacturer allowed for easy problem rectification and modification. Incorporation of the reset switch (S1) removed the need to download new code each time a monitoring session was started. The use of a data retrieval switch (S2) allows for easy data retrieval and uploads to the computer. However, this is mitigated by having to use the hyper terminal as the

upload mechanism. This method is suitable for development of the prototype, but a different method of data transfer from the processor assembly will be required if the device is to be used for supplying data for qualitative graphing in a real world environment. If the device is only used as a sensor and alarm, then the requirement for faster data uploads will not be required.

MATLAB was used in this project as the software for producing the graphical displays seen throughout this dissertation. As this is a specialist software package, it is unreasonable to assume that it is available to all users. The data that is displayed on the hyper link terminal is directly compatible with windows based software such as Excel, making it easy for any user to produce graphs if required.

Although the majority of the test of this device was conducted on an adult subject, enough testing was conducted on a neo-natal child to indicate that the concept holds true across various age groups. Future development would require a more thorough testing regime if the monitor is to proceed beyond the prototype stage.

9.3 RECOMMENDATIONS

For this device to proceed beyond the proof of concept and prototype stage it is recommended that research into the following areas be considered:

Sensor Backing Material – The current backing material used was the first one that was readily available. As the graphs in Figure 6.4 show, the material is not perfectly linear at the low range of applied strain. Investigations into a more suitable material may produce a more accurate and linear response to a patients breathing.

Strain Gauge – The strain gauge used in this project proved to be reliable and produce repeatable results for the duration of the project. To improve the sensitivity of the output, a second gauge placed under strain can be used. Improvement of the sensitivity of the sensor assembly will allow for a more accurate qualitative result to be viewed from the supplied data. Tests, or manufacturer's data, to determine the long term viability of a metal foil strain gauge in this type of sensing device would also need to be gathered and analysed.

Amplifier – The voltage drift problem within the amplifier assembly needs to be addressed. If this problem can be overcome, then this type of instrument amplifier should prove suitable for future devices.

Processor – The PICAX-18X was an ideal processor for development of this device. The microprocessor required for a functioning monitor should be as simple as possible. If the device is continue beyond the prototype stage, then the processor can be downgraded to a less powerful device. This would help reduce the overall cost, as a smaller processor in the same family is less than half the price.

Data Uploads – As stated previously, if the device is used as a monitoring and alarm device, then the use of serial data transmission is suitable. If the device is required to upload large amounts of data then investigations into parallel data transmission should be conducted.

9.4 CONCLUSION

The concept for this project came from personal loss and was driven by a desire to help others avoid a similar loss. Although the respiration and monitoring device described within these pages has not progressed beyond the prototype stage, it is the author's earnest desire to see it do so.

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APPENDIX A

ENG 4111/2 Research Project

Project Specification

For: Dr John Leis

Topic: Respiration Monitor and Recording System

Supervisor: Ms Lyn Brodie

Project Aim: This project aims to design, construct and validate a respiration monitor and recording system; utilising strain gauges as the sensing element. Incorporated into the system will be an apnoea monitor and alarm system.

PROGRAMME: Issue A, 15th March 2007

1. Research information on the design and operation of strain gauges.

2. Research the basic human respiration system; including the main types of respiration and their symptoms.

3. Design a digital data logging system to accept inputs from analogue strain gauges in a single input system.

4. Construct a working prototype of the design.

5. Carry out experiments on individuals of varying age groups to determine the validity of the design.

6. Produce a graphical display from the raw data.

As Time permits:

7. Incorporate an apnoea monitor and alarm system.

8. Investigate the use of dual strain gauge inputs for comparing respiration function for both left and right lungs.

9. Incorporate an FM transmitter into the design for remote monitoring for occurrences of apnoea events.

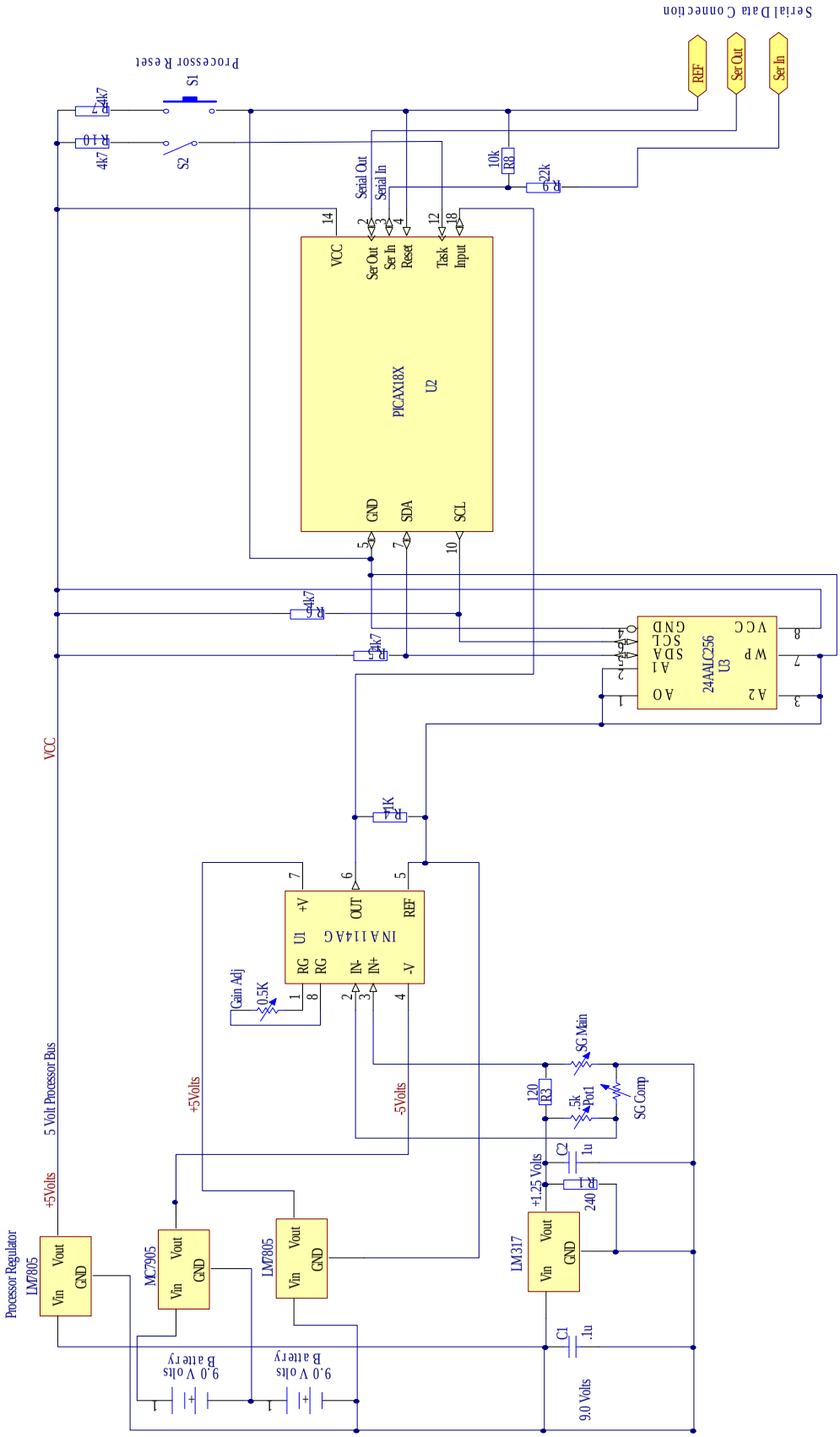
AGREED:

_____ (Student) _____/_____/_____

_____ (Supervisor) _____/_____/_____

APPENDIX B1

Project Circuit Diagram



APPENDIX B2

Design Procedure for Sensor Assembly

Materials Required

Silicon Rubber Sheet – 403-279 (Laird Technologies - K177-NA-30X30)

Loctite 5145 (Neutral Cure)

Pacer ZAP glue (Super Glue or similar)

White Spirits

Lint Free Cloth

Clear Tape

Elastic 1m long, 3.5cm wide

1. Cut a piece of silicon rubber sheet 20 cm long and 3.5cm wide. Ensure that the weave of the material is at 45 degrees to the cut, as shown below in Figure B1.

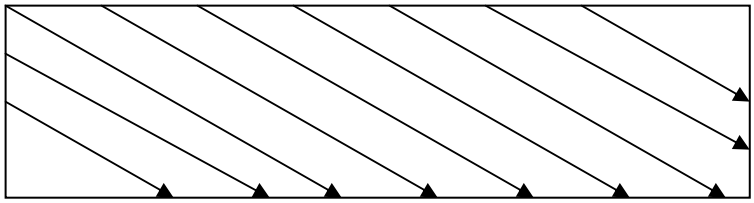


Figure B1. Rubber Sheet Orientation

2. Clean the material with white spirits or similar cleaner using the lint free cloth, ensuring no residue is left on the surface. Affix the material to the work bench using a piece of clear tape at each end.

3. Coat the bottom of the strain gauge with Pacer ZAP glue.

4. Place the gauge in the centre of the material, with the gauge parallel to the longest edges.

5. Using a piece of clear tape to cover the gauge, press the gauge down firmly into place and hold for approximately 20 seconds or until the glue has cured.

6. Remove tape from the gauge carefully.

7. Apply glue to the first 30mm of insulated cable and repeat steps 5 and 6.

8. Apply enough silicon gel to cover the gauge, bare wires and 30mm of insulated cable.

Caution

Ensure the cable runs parallel to the longest edge before the silicon is applied

9. Using a piece of hard plastic card (such as a credit card) smooth the silicon gel into place and remove any excess.

10. Allow to cure for 24 hours. The gauge when attached correctly should look similar to that shown in Figure B2.

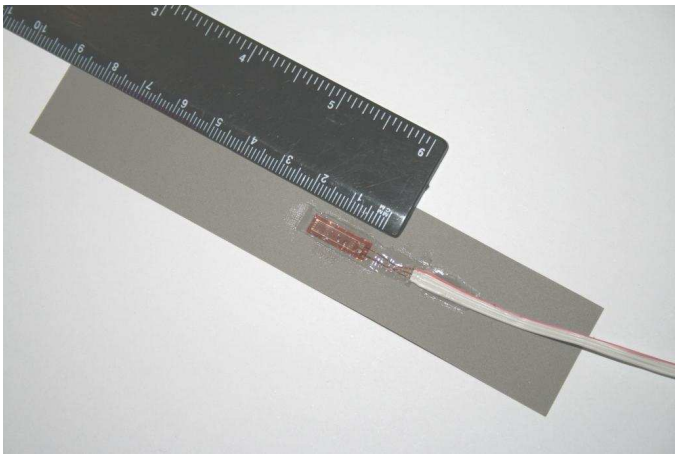


Figure B2. Gauge Attached to Backing

11. Sew the material onto the centre of the piece of elastic as shown in Figure B3.

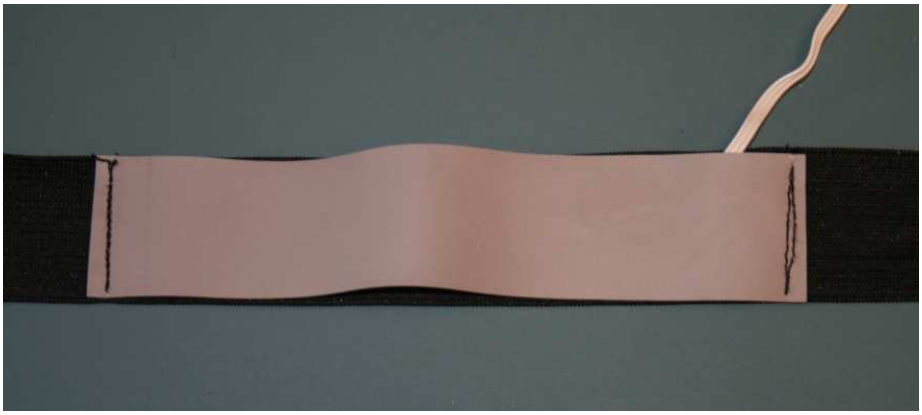


Figure B3. Backing Attached to Elastic Strap

APPENDIX B3

Processor Code

```
mode:

        if pin0 = 1 then goto retrieve

main:

for b0=1 to 200
        readadc 1,w2
        i2cslave %10100000, i2cslow, i2cword
        writei2c b0,(w2, " ")
        pause 50
        next
end

retrieve:
        for b0=1 to 200
        i2cslave %10100000, i2cslow, i2cword
        readi2c b0,(w2)
        sertxd (#w2, " ")
        next
end

end
```

APPENDIX B4

Monitoring Procedure

The following details the complete procedure for operating this device, from loading the code to producing a graphical output.

a. Connect the monitoring device to the computer using the supplied serial data cable to the serial communications port 2

b. Switch on the computer, start the editor software and load the code file into the editor.

c. Choose the number of hours you wish to monitor for. For every hour of monitoring add 72000 to the loop.

d. Ensure that the monitoring device is switched on and S2 is open (otherwise the processor will attempt to download data from the memory). Press F5 on the keyboard.

e. Attach the sensing assembly to the person being monitored in the same position as the chest band in Figure 3.3.

f. Press and release switch S1 (reset). The processor will now record the sensed signals.

g. Close the editor software and open a Hyperlink terminal, choosing the same serial communications port (2) that the monitoring device is connected to.

h. Close switch S2 and press reset. The processor output will now be displayed on the hyperlink terminal.

i. Save displayed data into a text file.

j. Import the data into the MATLAB code as variable ‘a’ found in appendix C6.

k. Save and plot as required.

Appendix C1

Carbon Weave Tensile Test MATLAB Code

```
clear; clc; close all
Test1=[0.0,0.00
0.3,0.03
0.0,0.08
0.0,0.11
0.1,0.13
0.1,0.16
.....
21.3,16.71
20.3,16.73
17.7,16.76
11.1,16.79
-0.1,16.80];

crosshead=Test1(1:639,2);
b=length(crosshead);

force=Test1(1:639,1);
c=length(force);

plot(crosshead,force,'r');

title('Tensile Test on Backing Material', 'FontSize',24);
xlabel('Cross Head Distance (mm)', 'FontSize',24);
ylabel('Force (N)', 'FontSize',24);
hold


Test2=[0.2,0.00
0.1,0.01
0.1,0.03
0.3,0.06
0.2,0.09
.....
21.0,13.02
20.6,13.04
19.6,13.07
15.1,13.10
2.1,13.12];

crosshead2=Test2(1:501,2);
force2=Test2(1:501,1);

plot(crosshead2,force2)

grid
legend('Test1','Test2')
```

Appendix C2

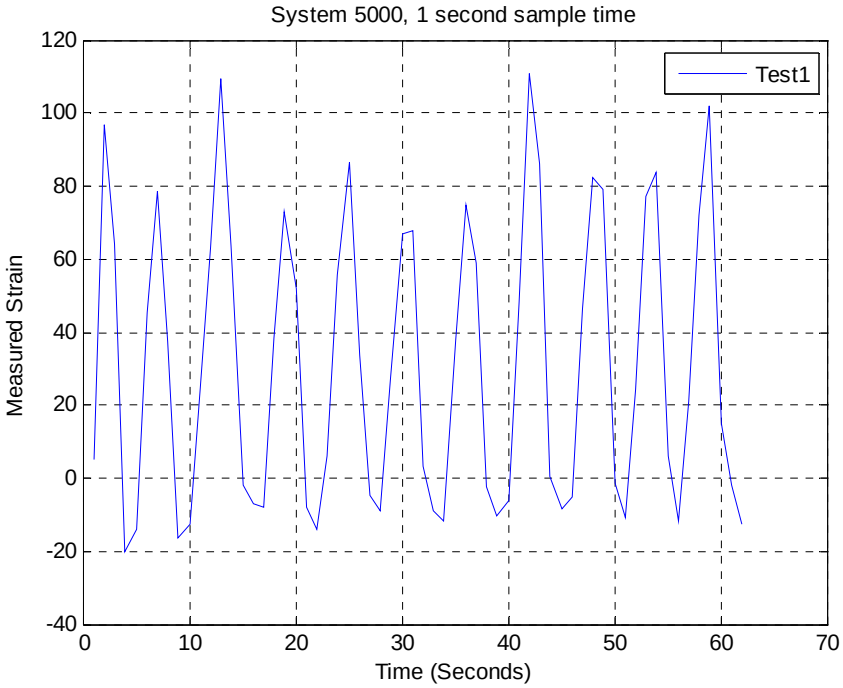
System 5000 Test 1

```
clear; clc;
close all;
% Test Data
Test1=[1    15.9000 4.94
2   16.9000 96.91
3   17.9000 64.27
4   18.9000 -20.27
5   19.9000 -13.84
.....

59  73.9000 101.85
60  74.9000 14.83
61  75.9000 -1.98
62  76.9000 -12.852];

% Define the Data
crosshead2=Test1(1:62,3);
force2=Test1(1:62,1);

% Plot Data
plot(force2,crosshead2)
grid
xlabel('Time (Seconds)')
ylabel('Measured Strain')
title('System 5000, 1 second sample rate')
legend('Test1')
```



Appendix C3

System 5000 Test 2

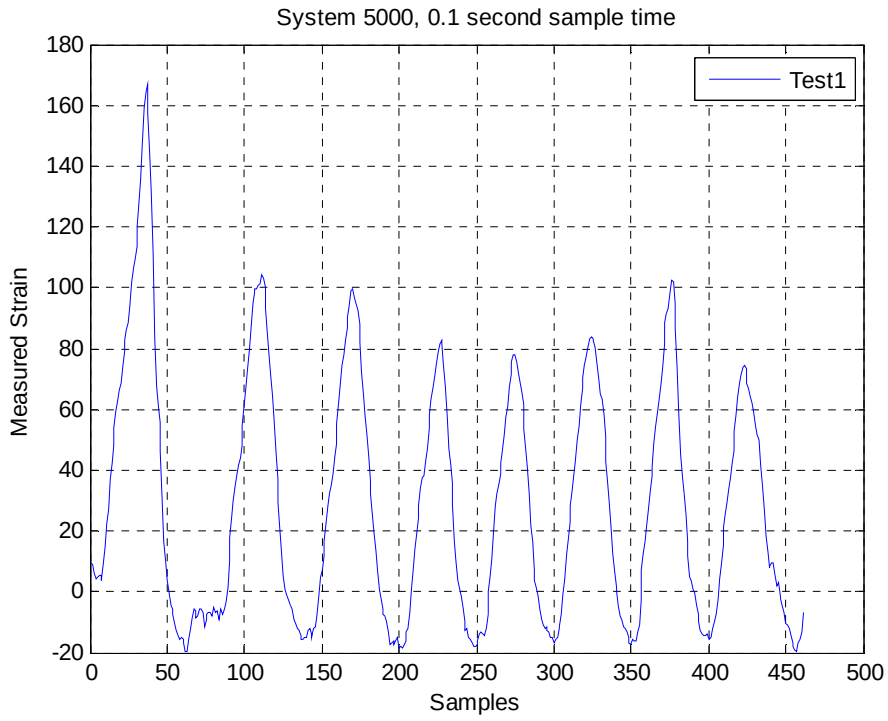
```
clear; clc;
close all;
% Test Data
Test1=[1    3.6000  9.39
2   3.7000  8.89
3   3.8000  5.93
4   3.9000  4.45
5   4.0000  4.94
.....

458 49.3000 -16.79
459 49.4000 -15.81
460 49.5000 -13.34
461 49.6000 -6.92];

% Define the Data

crosshead2=Test1(1:461,3);
force2=Test1(1:461,1);

% Plot Data
plot(force2,crosshead2)
grid
xlabel('Time (Seconds)')
ylabel('Measured Strain')
title('System 5000, 0.1 second sample time')
legend('Test1')
```



Appendix C4

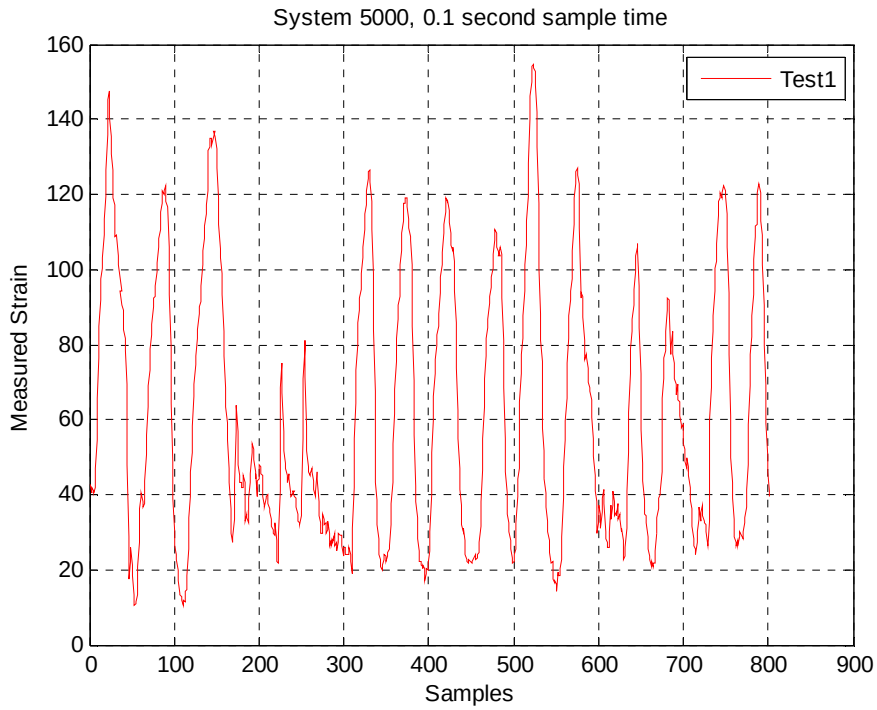
System 5000 Test 3

```
clear; clc;
close all;
% Test Data
Test1=[1    2.0000  42.94
2   2.1000  42.94
3   2.2000  40.47
4   2.3000  41.96
5   2.4000  40.97
.....

798 81.7000 59.23
799 81.8000 50.35
800 81.9000 45.90
801 82.0000 40.97
802 82.1000 39.49];

% Define Data
crosshead2=Test1(1:802,3);
force2=Test1(1:802,1);

% Plot Data
plot(force2,crosshead2,'r')
grid
xlabel('Time (Seconds)')
ylabel('Measured Strain')
title('System 5000, 0.1 second sample time')
legend('Test1')
```



Appendix C5

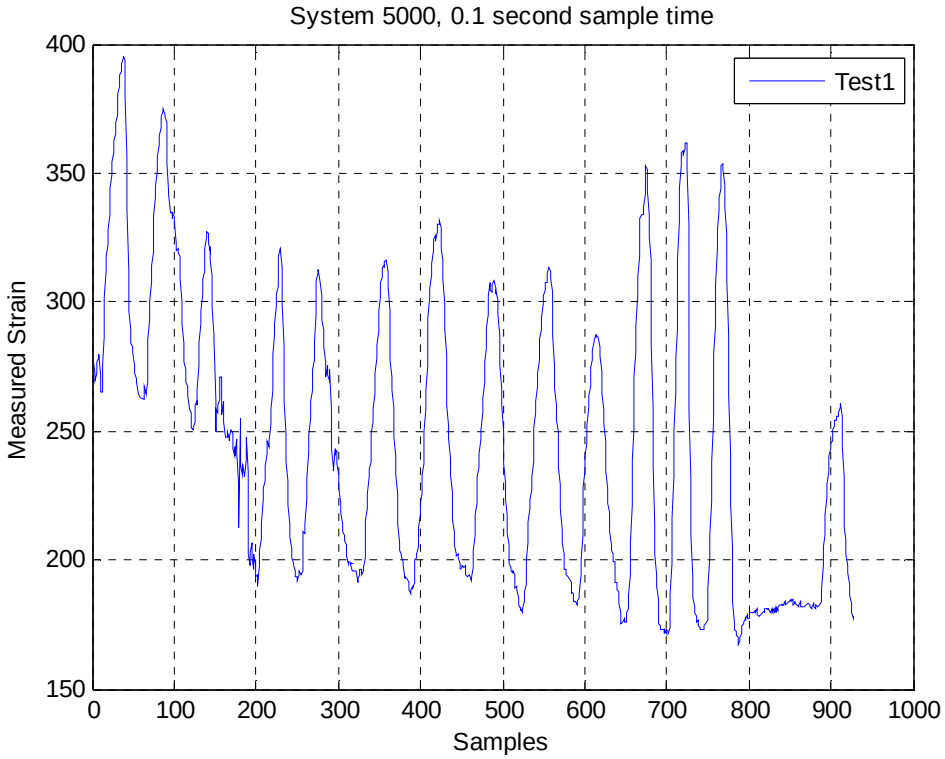
System 5000 Test 4

```
clear; clc;
close all;
% Test Data
Test1=[1    76.0000 279.51
2   76.1000 278.53
3   76.2000 272.10
4   76.3000 269.63
5   76.4000 272.60
.....

924 168.3000    183.20
925 168.4000    182.21
926 168.5000    179.25
927 168.6000    177.77
928 168.7000    178.26];

% Define Data
crosshead2=Test1(1:928,3);
force2=Test1(1:928,1);

% Plot Data
plot(force2,crosshead2,'r')
grid
xlabel('Samples')
ylabel('Measured Strain')
title('System 5000, 0.1 second sample time')
legend('Test1')
```



Appendix C6

MATLAB Code Example

```
% a=[ PLACE YOUR DATA FROM YOUR SENSING OPERATION IN HERE]
c=min(a)
d=max(a)
e=d-c;
a=a-c;
b=length(a);

% Plot Data
plot(a);
axis([0 b 0 e]);
xlabel('Samples');
ylabel('Depth of breath')
title('NAME, XXXX Gain, sample rate 20');
grid
```

Appendix C7

MATLAB Code Example (complete)

```
% Test Data (Figure 7.2)

a=[ 142 141 142 144 144 …… 141 140 141 142 141]

% Test Data (Figure 7.3)

% a=[155 157 157 157 158 157 159 159 159 160 158 160 159 160 160 159
159 158 158 158 157 158 157 157 157 155 154 152 152 151 149 149 147
147 147 145 145 143 144 144 143 143 141 142 142 142 142 140 141 141
141 141 141 142 142 143 145 144 147 147 148 150 149 151 151 153 154
153 155 ]

c=min(a)

d=max(a)

e=d-c;

a=a-c;

b=length(a);


% Plot Data

plot(a);

axis([0 b 0 e]);

xlabel('Samples');

ylabel('Depth of breath, mV/10')

title('41 Year Old Male, 2500 Gain, sample rate 20, Snapshot of a
Single Breathing Cycle');

grid
```

Appendix D1

Adjustable Voltage Regulator

National Semiconductor

May 1996

LM117/LM317A/LM317
3-Terminal Adjustable Regulator

General Description

The LM117 series of adjustable 3-terminal positive voltage regulators is capable of supplying in excess of 1.5A over a 1.2V to 37V output range. They are exceptionally easy to use and require only two external resistors to set the output voltage. Further, both line and load regulation are better than standard fixed regulators. Also, the LM117 is packaged in standard transistor packages which are easily mounted and handled.

In addition to higher performance than fixed regulators, the LM117 series offers full overload protection available only in IC's. Included on the chip are current limit, thermal overload protection and safe area protection. All overload protection circuitry remains fully functional even if the adjustment terminal is disconnected.

Normally, no capacitors are needed unless the device is situated more than 6 inches from the input filter capacitors in which case an input bypass is needed. An optional output capacitor can be added to improve transient response. The adjustment terminal can be bypassed to achieve very high ripple rejection ratios which are difficult to achieve with standard 3-terminal regulators.

Besides replacing fixed regulators, the LM117 is useful in a wide variety of other applications. Since the regulator is "floating" and sees only the input-to-output differential voltage, supplies of several hundred volts can be regulated as long as the maximum input to output differential is not exceeded, i.e., avoid short-circuiting the output.

Also, it makes an especially simple adjustable switching regulator, a programmable output regulator, or by connecting a fixed resistor between the adjustment pin and output, the LM117 can be used as a precision current regulator. Supplies with electronic shutdown can be achieved by clamping the adjustment terminal to ground which programs the output to 1.2V where most loads draw little current.

For applications requiring greater output current, see LM150 series (3A) and LM138 series (5A) data sheets. For the negative complement, see LM137 series data sheet.

LM117 Series Packages and Power Capability

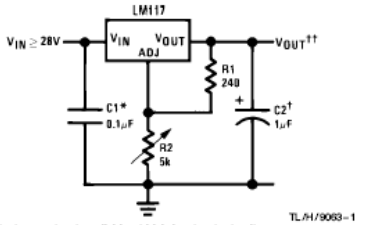
Part Number Suffix	Package	Rated Power Dissipation	Design Load Current
K	TO-3	20W	1.5A
H	TO-39	2W	0.5A
T	TO-220	20W	1.5A
E	LCC	2W	0.5A
S	TO-263	4W	1.5A

Features

- Guaranteed 1% output voltage tolerance (LM317A)
- Guaranteed max. 0.01%/V line regulation (LM317A)
- Guaranteed max. 0.3% load regulation (LM117)
- Guaranteed 1.5A output current
- Adjustable output down to 1.2V
- Current limit constant with temperature
- P+ Product Enhancement tested
- 80 dB ripple rejection
- Output is short-circuit protected

Typical Applications

1.2V–25V Adjustable Regulator



TL/H/9063–1

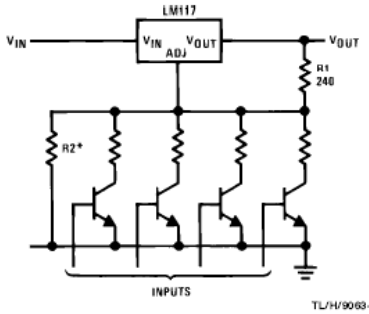
Full output current not available at high input-output voltages

*Needed if device is more than 6 inches from filter capacitors.

†Optional—improves transient response. Output capacitors in the range of 1 μF to 1000 μF of aluminum or tantalum electrolytic are commonly used to provide improved output impedance and rejection of transients.

$$\dagger\dagger V_{OUT} = 1.25V\left(1 + \frac{R2}{R1}\right) + I_{ADJ}(R_2)$$

Digitally Selected Outputs



TL/H/9063–2

*Sets maximum V_{OUT}

RRD-B30M56/Printed in U. S. A.

http://www.national.com

LM117/LM317A/LM317 3-Terminal Adjustable Regulator

Full details for this component can be obtained from the following location:

http://www.datasheetcatalog.com/datasheets_pdf/L/M/3/1/LM317.shtml

Appendix D2

5.0 Volt Regulator

National Semiconductor

July 1999

LM78XX
Series Voltage Regulators

General Description

The LM78XX series of three terminal regulators is available with several fixed output voltages making them useful in a wide range of applications. One of these is local on card regulation, eliminating the distribution problems associated with single point regulation. The voltages available allow these regulators to be used in logic systems, instrumentation, HiFi, and other solid state electronic equipment. Although designed primarily as fixed voltage regulators these devices can be used with external components to obtain adjustable voltages and currents.

The LM78XX series is available in an aluminum TO-3 package which will allow over 1.0A load current if adequate heat sinking is provided. Current limiting is included to limit the peak output current to a safe value. Safe area protection for the output transistor is provided to limit internal power dissipation. If internal power dissipation becomes too high for the heat sinking provided, the thermal shutdown circuit takes over preventing the IC from overheating.

Considerable effort was expanded to make the LM78XX series of regulators easy to use and mininize the number of external components. It is not necessary to bypass the output, although this does improve transient response. Input bypassing is needed only if the regulator is located far from the filter capacitor of the power supply.

For output voltage other than 5V, 12V and 15V the LM117 series provides an output voltage range from 1.2V to 57V.

Features

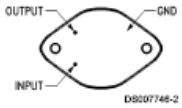
- Output current in excess of 1A
- Internal thermal overload protection
- No external components required
- Output transistor safe area protection
- Internal short circuit current limit
- Available in the aluminum TO-3 package

Voltage Range

LM7805C	5V
LM7812C	12V
LM7815C	15V

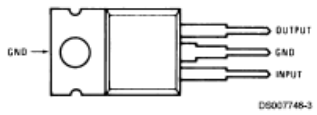
Connection Diagrams

Metal Can Package
TO-3 (K)
Aluminum



Bottom View
Order Number LM7805CK,
LM7812CK or LM7815CK
See NS Package Number KC02A

Plastic Package
TO-220 (T)



Top View
Order Number LM7805CT,
LM7812CT or LM7815CT
See NS Package Number T03B

Full details for this component can be obtained from the following location:

http://www.it.lth.se/datablad/Analog/voltage/lm78xx.pdf

Appendix D3

Negative 5.0 Volt Regulator

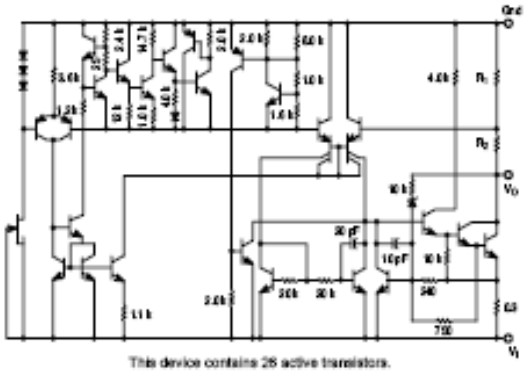
MC7900 Series

1.0 A Negative Voltage Regulators

The MC7900 series of fixed output negative voltage regulators are intended as complements to the popular MC7800 series devices. These negative regulators are available in the same seven−voltage options as the MC7800 devices. In addition, one extra voltage option commonly employed in MECL systems is also available in the negative MC7900 series.

Available in fixed output voltage options from −5.0 V to −24 V, these regulators employ current limiting, thermal shutdown, and safe−area compensation − making them remarkably rugged under most operating conditions. With adequate heatsinking they can deliver output currents in excess of 1.0 A.

- No External Components Required
- Internal Thermal Overload Protection
- Internal Short Circuit Current Limiting
- Output Transistor Safe−Area Compensation
- Available in 2% Voltage Tolerance (See Ordering Information)
- Pb−Free Packages May be Available. The G−Suffix Denotes a Pb−Free Lead Finish

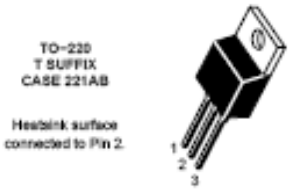


This device contains 26 active transistors.

Figure 1. Representative Schematic Diagram

ON Semiconductor®

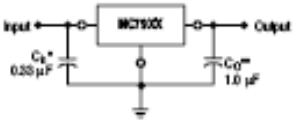
http://onsemi.com



Pin 1. Ground
2. Input
3. Output

Heatsink surface (shown as terminal 4 in case outline drawing) is connected to Pin 2.

STANDARD APPLICATION



A common ground is required between the input and the output voltages. The input voltage must remain typically 2.0 V above more negative even during the high point of the input ripple voltage.

XX, These two digits of the type number indicate nominal voltage.
* C_{in} is required if regulator is located an appreciable distance from power supply filter.
** C_O improve stability and transient response.

ORDERING INFORMATION

See detailed ordering and shipping information in the package dimensions section on page 11 of this data sheet.

DEVICE MARKING INFORMATION

See general marking information in the device marking section on page 14 of this data sheet.

Full details for this component can be obtained from the following location:

http://www.onsemi.com/pub/Collateral/MC7900-D.PDF

Appendix D4

Instrumentation Amplifier

BURR-BROWN®

INA114

Precision INSTRUMENTATION AMPLIFIER

FEATURES

- LOW OFFSET VOLTAGE: 50µV max
- LOW DRIFT: 0.25µV/°C max
- LOW INPUT BIAS CURRENT: 2nA max
- HIGH COMMON-MODE REJECTION: 115dB min
- INPUT OVER-VOLTAGE PROTECTION: ±40V
- WIDE SUPPLY RANGE: ±2.25 to ±18V
- LOW QUIESCENT CURRENT: 3mA max
- 8-PIN PLASTIC AND SOL-16

APPLICATIONS

- BRIDGE AMPLIFIER
- THERMOCOUPLE AMPLIFIER
- RTD SENSOR AMPLIFIER
- MEDICAL INSTRUMENTATION
- DATA ACQUISITION

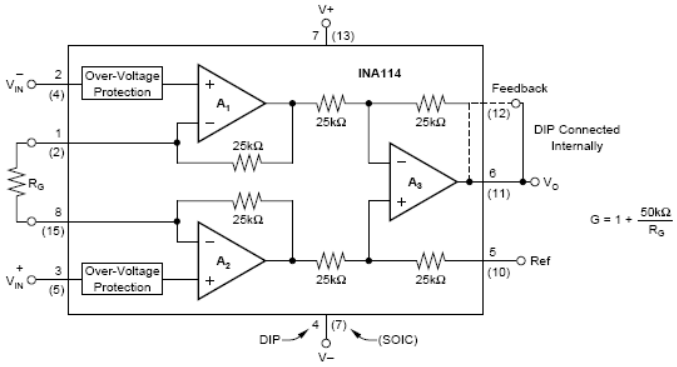
DESCRIPTION

The INA114 is a low cost, general purpose instrumentation amplifier offering excellent accuracy. Its versatile 3-op amp design and small size make it ideal for a wide range of applications.

A single external resistor sets any gain from 1 to 10,000. Internal input protection can withstand up to ±40V without damage.

The INA114 is laser trimmed for very low offset voltage (50µV), drift (0.25µV/°C) and high common-mode rejection (115dB at G = 1000). It operates with power supplies as low as ±2.25V, allowing use in battery operated and single 5V supply systems. Quiescent current is 3mA maximum.

The INA114 is available in 8-pin plastic and SOL-16 surface-mount packages. Both are specified for the –40°C to +85°C temperature range.



International Airport Industrial Park • Mailing Address: PO Box 11400, Tucson, AZ 85734 • Street Address: 6730 S. Tucson Blvd., Tucson, AZ 85706 • Tel: (520) 746-1111 • Twx: 910-952-1111
Internet: http://www.burr-brown.com/ • FAXLine: (800) 548-6133 (US/Canada Only) • Cable: BBRCORP • Telex: 066-6491 • FAX: (520) 889-1510 • Immediate Product Info: (800) 548-6132

 PDS-1142D Printed in U.S.A. March, 1998

Full details for this component can be obtained from the following location:

http://www.chipcatalog.com/Datasheet/1B07243D35E968BC30F56580F0DC8B03.htm

Appendix D5

256K Serial EEPROM

MICROCHIP 24AA256/24LC256/24FC256

256K I²C™ CMOS Serial EEPROM

Device Selection Table

Part Number	Vcc Range	Max. Clock Frequency	Temp. Ranges
24AA256	1.8-5.5V	400 kHz(1)	I
24LC256	2.5-5.5V	400 kHz	I, E
24FC256	1.8-5.5V	1 MHz(2)	I

Note 1: 100 kHz for Vcc < 2.5V.
2: 400 kHz for Vcc < 2.5V.

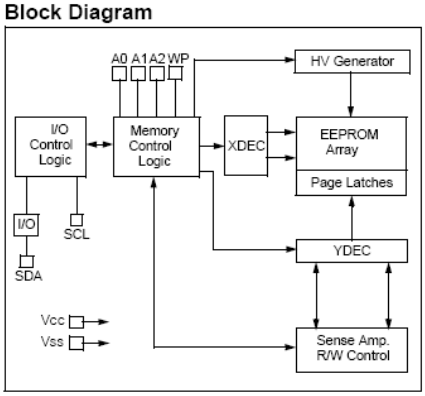
Features

- Low-power CMOS technology:
  - Maximum write current 3 mA at 5.5V
  - Maximum read current 400 µA at 5.5V
  - Standby current 100 nA typical at 5.5V
- 2-wire serial interface bus, I²C™ compatible
- Cascadable for up to eight devices
- Self-timed erase/write cycle
- 64-byte Page Write mode available
- 5 ms max. write cycle time
- Hardware write-protect for entire array
- Output slope control to eliminate ground bounce
- Schmitt Trigger inputs for noise suppression
- 1,000,000 erase/write cycles
- Electrostatic discharge protection > 4000V
- Data retention > 200 years
- 8-pin PDIP, SOIC, TSSOP, MSOP and DFN packages, 14-lead TSSOP package
- Standard and Pb-free finishes available
- Temperature ranges:
  - Industrial (I): -40°C to +85°C
  - Automotive (E): -40°C to +125°C

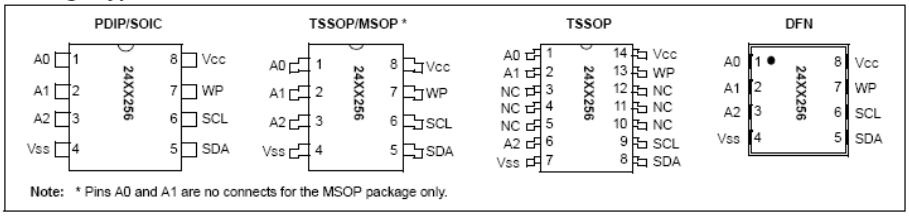
Description

The Microchip Technology Inc. 24AA256/24LC256/24FC256 (24XX256*) is a 32K x 8 (256 Kbit) Serial Electrically Erasable PROM, capable of operation across a broad voltage range (1.8V to 5.5V). It has been developed for advanced, low-power applications such as personal communications or data acquisition. This device also has a page write capability of up to 64 bytes of data. This device is capable of both random and sequential reads up to the 256K boundary. Functional address lines allow up to eight devices on the same bus, for up to 2 Mbit address space. This device is available in the standard 8-pin plastic DIP, SOIC, TSSOP, MSOP, DFN and 14-lead TSSOP packages.

Block Diagram



Package Types



Note: * Pins A0 and A1 are no connects for the MSOP package only.

*24XX256 is used in this document as a generic part number for the 24AA256/24LC256/24FC256 devices.

Full details for this component can be obtained from the following location:

http://ww1.microchip.com/downloads/en/devicedoc/21203M.pdf