Ultrasonic apnea monitor

by

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INTRODUCTION

Apnea monitors are devices most commonly used to detect life threatening episodes in both adults and infants with respiratory problems. The physician's decision for such patients heavily depends on the data he gets from the monitors, so accurate and reliable information is expected from the apnea monitors.

In the case of infants, apnea monitors are primarily used for detection of apnoeic attack, the absence of respiration for a period of time. Obviously, arterial oxygen tension (PaO_2) monitoring would be of great importance because the results are direct, accurate, and reliable. But, since incision and catheterization are involved, other threats to the life of the infant, like infection and hemorrhage, may appear. It is logical, therefore, to look for other techniques which provide data reliable as PaO_2 monitoring, and yet which require no invasion of the body.

Plethysmographs and monitor pads are now commonly used in hospitals. They detect changes in the volume of the chest or the changes in the resistance of the tissue between a pair of electrodes placed on the chest. These techniques lack the invasive disadvantages of PaO_2 monitoring, but they have their own disadvantages, such as electrode interference, skin irritation, and fall in PaO_2 due to handling of the infant for routine care.

Infants' PaO_2 levels usually fall considerably when they are disturbed. These PaO_2 decreases are observed even during routine handling of the baby such as changing the under-sheet in the incubator, feeding the baby or attaching an electrode to the baby for observation of respiratory or heart rate. Sometimes these decreases in PaO_2 necessitate respiratory support.

In order not to disturb the infant, contactless apnea monitors, such as air-filled mattresses or capacitance change detectors were devised. These monitors and other similar ones detect movements due to respiration rather than respiration itself. Thus, in cases where the upper airway is blocked, according to Hering-Breuer Reflex theory, the infant would make an effort for respiration even though there is not true respiration. These cases are not detected by contactless monitors. Furthermore, some of these techniques like air-filled mattresses, can be used only for a short time period; and some others such as capacitance-change detectors are particularly sensitive to moving objects around the baby also.

In this thesis, a novel contactless technique for monitoring apnea is described which lacks most of the disadvantages of other contactless monitors, and, with the help of a minicomputer, detects upper airway occlusion.

This technique utilizes a pair of ultrasonic air

pressure transducers functioning as a transmitter and a receiver. The transmitter sends pressure waves (40 KHz) toward chest of the infant, and the second transducer picks up the reflected waves. An electronic circuit connected to the transducers detects the phase shift of the reflected waves with respect to the transmitted waves. When the chest wall is moving, the phase shift changes, and when there is no chest wall movement the phase shift remains constant, a condition which is detected and used to trigger an alarm.

The computer section of this technique is for detection of upper airway occlusion (U.A.O.). The waveform of the chest wall movement during U.A.O. is considerably different from the normal waveform. The computer continuously processes the waveform and turns on the alarm when the specific changes of U.A.O. in the waveform appear.

For primary testing of this monitor, a respiration pump was used. A plate attached to the piston of the pump was used as a reflector. When the piston was stopped, the alarm sounded.

For further evaluation of this technique, animal trials were instituted. The subjects were young mongrel dogs ranging from 5 to 10 lbs. The dogs were anesthetized to prevent body movements. A thermistor was placed in the lumen of an endotracheal tube and then the dog was intubated. Conventional respiration waveforms from a thermistor bridge

were then taken. Though the waveforms from the two methods were not exactly the same, both indicated respiration very well. During these trials, the buzzer of the contactless monitor never sounded, i.e., no false alarm was observed.

To verify the validity of the computer program, the endotracheal tube was clamped (artificial U.A.O.) several times. The computer detected the occlusion in most of the trials.

LITERATURE REVIEW

Apnea and Causes

Apnoeic attack, defined as the absence of the breathing for a period of more than 20 seconds (1), is a major threat to the lives of infants of very short gestational (life in the uterus) age (2). The condition has a high mortality rate and is likely to be a possible cause of hypoxic brain damage among survivors (3).

Apnoeic attacks should not be mistaken with Sudden Infant Death Syndrome (SIDS) which has been established as a distinct disease entity (4, 5). In a significant proportion of SIDS cases viral infection is present. In addition, most cases of SIDS occur between 4 and 16 weeks, suggesting some physiological vulnerability during this age period (6), while apnoeic attack is usually seen during the first days or even during the first hours after birth.

Apnoeic attacks may occur in association with a variety of disorders, but the most convincing explanation is physical immaturity (1). In order to expand the chest and thereby increase the negative intrapleural pressure so that air can enter the lungs, a certain degree of rigidity of the chest wall is necessary and is normally provided by the ribs. Since newborn infants, particularly prematures, do not have a rigid bony thorax, some difficulty in expanding the lungs

occurs (7).

Slow and late appearance of alveoli may be another reason confirming the explanation of immaturity for apnea in infants. A mature newborn has about 20 million alveoli. By the time the individual reaches maturity, the number of alveoli reaches about 300 million; thus, much of the development of the alveoli occurs after birth (7).

Some premature infants simply forget to breathe while they are in deep sleep. These cases are believed to be due to immaturity of autonomic nervous system control mechanism. Thus it may not be far from fact if one accepts the general explanation of immaturity as the cause of apnea in infants.

Techniques for Monitoring

Because of the significant threat of apnoeic attack to the lives of infants, monitoring respiration of newborns, primarily in cases of premature birth, is of considerable importance. The most reliable and accurate method for monitoring respiration is monitoring the PaO_2 level in the arterial blood. In this method a catheter is inserted, usually into an umbilical artery, until the oxygen electrode tip is in the lower aorta. Although this method directly monitors oxygen tension level and hence accurate and reliable results are obtained, incision, asepsis and catheterization are involved; therefore, some side threats to the

patient, like hemorrhage and infection, are created. Because of these threats and the delicate and time-consuming procedure of setting up the system, many physicians and researchers have been searching for noninvasive and safe techniques to monitor respiration.

Several of these noninvasive methods described here utilize thermistors in respiration detection (8). Thermistors are electrical devices which exhibit an inverse relation between temperature and resistance. They are placed in the oral or nasal airflow. The exhaled gas is at a different temperature compared to the inhaled gas. As a result, the resistance of a thermistor element will change with inspiration and expiration if it is placed in the airflow.

Thermistors are placed in the lumen of an endotracheal tube and then the patient is intubated. The output of the thermistor is fed into an electronic circuit for monitoring and recording. This technique necessitates anesthetization of the patient, an undesirable situation because anesthetization relaxes the control mechanism of the nervous system and causes the patient to make less of an attempt for respiration, thus increasing the chances of apnea.

Pharyngeal probes, using thermistors, have also been developed (8). One which is commonly used, utilizes a 7F (for adults) of 5F (for children) nasogastric tube. The

tube with the thermistor inside is positioned in the posterior pharynx via a nontoxic flexible catheter. The posterior pharynx receives nasal and oral airflow and can be catheterized comfortably and safely. An electronic circuit is also needed to be added to the system to condition the signal and to detect respiration. The disadvantage with this technique is that the pharyngeal placement of the probe is not well tolerated in children. Also, after many hours of continuous use the signal may become attenuated as a result of mucous accumulation.

Other, noninvasive, techniques of respiration monitoring have been developed. These techniques can be divided into two groups, plethysmography and motion detecting techniques.

Plethysmography generally measures change in volume. A body plethysmograph was the first apparatus used for measuring ventilation in the newborn (7). In this technique the infant would be placed in a closed chamber, while the infant's mouth is kept outside the chamber. With respiration, the volume of the chest changes and consequently the pressure inside the chamber changes. The fluctuation in the pressure is picked up by a pressure transducer attached to the chamber.

Using this technique, early workers were bothered with the air leak around the infant's neck. To overcome this

problem Karlberg and Koch (9) used a reverse plethysmograph. In this adaptation the infant breathed through a mask attached to a wide tube connected to a plethysmograph.

Major disadvantages of the masks are that they disturb the infant's normal respiratory pattern and produce an intolerable increase in the dead space. Low dead space, low resistance nasovalves have been constructed (10), but the general problem with such devices as well as with the body plethysmograph still persists; i.e., the system, being a rebreathing one, must be used only for a short period of time to avoid significant CO_2 accumulation.

Simpler plethysmographs such as: i) capacitance plethysmograph, ii) elastic-resistance strain-gage plethysmograph, and iii) impedance plethysmograph, have none of the problems associated with the body plethysmograph and yet are less expensive and easier to handle.

The first of these simpler methods measures the change in capacitance between the skin and an outer rigid electrode. The second method measures the change in the resistance of a conducting liquid in an extendable rubber tube placed around the body. These two methods are usually used to measure blood flow in extremities (11). An increase in volume of blood in a limb results in a decrease in separation

of the plates and an increase in capacitance, or results in an increase in the length of rubber tube of the related plethysmograph. In the first technique the electrical stability is a problem, and in the second method the electrical resistance is very low, and also the relation between resistance of the conductive liquid and the length of the tube is not linear.

A related method, impedance plethysmography, is based on the fact that changes in a tissue volume are accompanied by changes in the electrical resistance of the tissue. The electrical resistance can be measured by attaching electrodes to the tissue and requiring a "low-level" current to flow through the tissue. The voltage seen across the electrodes is proportional to the electrical impedance (11, 12).

The volume of the lungs changes with respiration and therefore the resistance between a pair of electrode attached to an infant's chest will change. This method requires a high frequency oscillator (50-100 KHz) which sends a current (less than 0.3 ma) through electrodes placed on the chest wall of the infant (12).

The general disadvantages of plethysmographs are skin irritation, interference with patient care, and slow application in emergencies (13). Also, there might be false alarms due to lead or electrode failure.

Instead of plethysmographs, monitor pads can be used. Monitor pads, usually made of disposable foam rubber, have four conductive vinyl surfaces, each representing standard electrocardiography (ECG) limb-lead points (RA,LA,RL,LL), each vinyl surface is connected to a snap to be attached to external wires. Monitor pads are worn on the back or around the chest of the infant. They are often used to obtain the ECG, but can also be used as an apnea monitor by measuring the impedance changes between the transthoracic RA-LA leads (13). By experiment (13), it has been shown that monitor pads do not have most of the disadvantages of the plethysmographs described.

Both monitor pads and plethysmographs do share a common disadvantage. They are extremely sensitive. So much so that they usually pick up the heart pulse as well as respiration.

A rather different technique which lacks the above disadvantage is the abdominal respiration detector. Increase in volume of the abdomen on respiration increases the radius of curvature of the anterior abdominal wall (14). The detector consists of a flat cylindrical capsule about 20 mm in diameter and 3 mm deep. The open end may be covered with a thin rubber diaphragm, or it may be applied directly to the abdominal wall to which the capsule is secured with adhesive tape. A nozzle in the side of the capsule is

connected by a length of flexible tubing to a pneumatic transducer. This may be either a sensitive capacitance manometer (14) or an electro-magnetic movement transducer (14) attached to a similar diaphragm capsule.

An increase in the volume of the abdomen causes a decrease in the volume and consequently an increase in the pressure of the cylinder. Volume changes of less than 0.1% can be detected (14). In this technique respiratory movements are reproduced with remarkable faithfulness provided the subject does not change position.

All the techniques described so far regardless of their advantages and disadvantages are accompanied by a potential danger. They all have something in contact with the patient. Observation of infants during routine care has shown handling for any reason, such as X-ray, heart rate, and respiration observations, often causes a fall in the arterial oxygen tension (15). Although the infant usually recovers spontaneously, a series of procedures, one following closely upon the other, may cause a prolonged fall in PaO_2 necessitating respiratory support. In cases of very ill patients even minor disturbances like attaching an electrode to the infant or changing the under sheet in the incubator causes a fall in PaO_2 of as much as 30-40 mm Hg from a normal value of 80-90 mm Hg (15).

To avoid disturbing the infant, contactless devices

for monitoring respiration in infants are desirable. The air mattress, used in some hospitals, is listed under the category of contactless devices for monitoring respiration. A heated thermistor is placed inside an air filled mattress. Breathing movements cause air to flow from one section of the mattress to another, cooling the thermistor and thus providing a measure of respiratory movements.

The major disadvantage of the air mattress for recording respiration patterns is that over the period of a few breaths, the sensitivity changes. A second disadvantage is the double peaking effect, which means that the thermistor becomes cool on both inspiration and expiration, thus providing two signal peaks per respiration cycle.

A different contactless respiration monitor, the under-mattress pressure sensor, utilizes a thin 150 mm square pad in which electrode plates are incorporated. This pad is placed under or slightly above the chest. Respiratory movements change the capacitance between the electrode plates. Comparison of this device with impedance respirometer has shown that the changes in sensitivity as the infant moves about are quite considerable; e.g., the sensitivity changes by a factor of 10-20 for different positions to which the infant moves. In some cases, the sensitivity changes over the interval of a few breaths without any apparent change in

the body position (16).

A capacitance sensor can also be used to monitor the respiration without contact. A pair of electrodes is placed near the body. Respiratory movements change the capacitance between electrodes. The measurement of capacitance changes is achieved by applying a 200 KH_z signal across the electrodes and detecting the current flow with a phase sensitive amplifier. The disadvantages of this system are: i) it is too sensitive to people moving nearby, and ii) the incubator should be electrically shielded. This type of system cannot readily be made portable (16).

Another technique which lacks the disadvantages of the two mentioned contactless monitors is the Radar Respiration Monitor. This system senses chest wall movements by detecting changes in a reflected electromagnetic micro-wave signal. A microwave source sends signals in form of radio frequency energy to the chest. The detector output is a periodic wave as the infant's chest moves back and forth. The period of this wave corresponds to the time the chest covers half the microwave wavelength (30 mm if a 10 GHz source is used).

This system has proved to be very satisfactory where electrodes and wires are hazardous. Nevertheless, the 10 GHz microwave frequency of the most readily available miniature microwave components is not ideal for chest wall movement. With very large inspired volumes, as in gasps,

the chest wall movement may be more than half the microwave wavelength, and this results in a distorted signal. Besides, the output from the detector may give a very distorted signal because it depends on the whereabouts on the standing wave pattern as chest movements occur.

All the techniques described in this chapter, except for the direct measurement of PaO_2 and the pharyngeal probe, detect either changes in the volume or movement of the chest. The characteristics of these techniques bring about a problem for monitoring apnea. Upper airway occlusion may not be detected because obstruction of the upper airway in infants may be associated with continued respiratory effects without air entering the lungs (17). These efforts cause changes in volume, electrical resistance, or position of the chest; thus, these changes are detected by monitors as respiration while there has not been true respiration.

Treatment of Apnea

Thus far, no effective treatments for apnoeic attacks have been found, primarily because the causes of apnea are not known. If the explanation of immaturity for apnoeic attacks is correct, then the best treatment is to wait for the infant to grow, providing apnoeic attacks are detected and treated. Nevertheless, there have been some ways

suggested or tested to prevent apnea. One method advocated for treatment of apnea is raising the ambient oxygen concentration, but it is often ineffective and may expose the infant to the risk of hypoxia between attacks (1).

Another way of treatment of apnea is the use of continuous positive airway pressure (C.P.A.P.). Applying C.P.A.P. to infants with severe recurrent apnea appears to provide respiratory drive for the infant, perhaps due to stimulation of the Hering-Breuer inflation reflex. This technique has proved to be effective in decreasing the rate of apnoeic attacks, but can not remove the attacks totally. Thus, the use of apnea monitors which provide reliable data and expose the infant to minimum side effects of monitoring is of great importance.

NEW TECHNIQUE AND DESIGN

The literature review indicates that monitoring of respiration is of great importance. However, we also learned that some of those techniques pose threats or are of low reliability. Among those techniques, contactless methods were found to be least harmful.

The contactless technique described here eliminates both the cost and accompanying skin problems of electrodes. In addition, with the help of a computer, upper airway occlusion (U.A.O.), an important concern of physicians, can be detected.

Basic Criterion

This ultrasonic technique is based on the standing wave phenomenon. A low level ultrasonic (40 KHz) air-pressure transducer[1], placed in front of the infant's chest or abdomen wall, sends pressure waves to the infant. The waves are reflected by the infant's chest wall with the same frequency, but with a different phase angle. A transducer similar to the transmitter, picks up the reflected waves and produces 40 KHz AC voltage at its output. If the transducers are side by side (see Figure 1), the phase angle of the reflected wave at the pick-up transducer is

[1]Model 70100 40 KHz, Linden Laboratories, Inc., Box 920, State College, PA, 16801 (814-355-5491).

$$\phi = 2\pi \cdot 2 \cdot l \cdot \frac{F}{V} + K \qquad (1)$$

where

l = distance between electrodes and the chest wall

F = frequency of oscillation

V = velocity of sound in air = 330 m/sec

K = constant

Since V, F, K are constant, changes in ϕ are proportional to changes in l,

$$d\phi = \frac{4\pi F}{V} \cdot dl \qquad (2)$$

According to Equation (2), if l is constant, then there is no change in the phase angle. In other words, continuous change in ϕ indicates continuous change in l, the distance between the electrodes and the infant, and thus indicates that the chest wall is moving.

The change in the phase angle difference between the transmitted and the reflected waves is proportional to the displacement of the infant's chest wall. In order to find the proportionality between the displacement and the phase angle difference, we must go back to Equation (1).

$$\phi = \frac{4\pi Fl}{V} + K$$

$$\frac{F}{V} = \frac{1}{\lambda} \quad (\lambda = \text{wavelength of the waves}) \qquad (3)$$

$$\phi = \frac{4\pi l}{\lambda} + K$$

The maximum change in phase angle difference can be 2π radians; thus

$$\Delta\phi = 2\pi$$

$$2\pi = \frac{4\pi}{\lambda} \cdot \Delta l$$

$$\Delta l = \frac{2\pi \cdot \lambda}{4\pi}$$

$$\Delta l = \frac{\lambda}{2} \qquad (4)$$

on the other hand,

$$\lambda = \frac{V}{F} = \frac{330 \text{ m/sec}}{40 \text{ KHz}}$$

$$\lambda = \frac{330 \times 10^3 \text{ m/sec}}{40 \times 10^3 \text{ rad/sec}} \times 10^3 \frac{\text{mm}}{\text{m}} = 8.25 \text{ mm}$$

$$\frac{\lambda}{2} = 4.12 \text{ mm}$$

Thus for every 4.12 mm of chest wall displacement, there would be 2π radian change in the phase angle.

Circuit Design

In order to find the phase angle difference between the transmitted and the reflected waves, one should know a little about the signals going in and out of the transducers. The input and output signals from the transducers are sinusoids with different amplitudes; but with the same frequency of 40 KHz. The amplitude of the signal from the pickup transducer, however, varies with changes in the

distance between the infant's chest wall and the transducers.

These fluctuations in the amplitude cause some problem in measuring the phase angle difference between the two waves, because in the technique we have chosen for measurement of phase angle difference, the amplitudes of the waves should be constant. To better understand how the measurement of phase angle difference is made, consider Figure 2. Figure 2 (a and b) shows a pair of square waves with the same amplitude, but with a lag time between the two waves. Figure 2 (part c) shows the summation of the two waves. If this waveform is rectified, a series of pulses with equal widths will remain. The widths of all pulses are equal and proportional to the phase angle difference. As the phase angle difference increases, the width of the pulses will decrease and vice versa.

A block diagram of the apnea monitor is shown in Figure 3. In Figure 3, block 1 is a sine wave generator. The oscillator drives the transmitter transducer, and the reflected wave drives the receiver transducer. The output of the receiver transducer is a 20 mV peak-to-peak sinusoid. To reduce the noise interference with this low amplitude signal a 10 KΩ resistance is placed across the output of the receiver transducer. The output is then fed into 2 stages of linear amplification in order to have a signal comparable in amplitude to the output of the

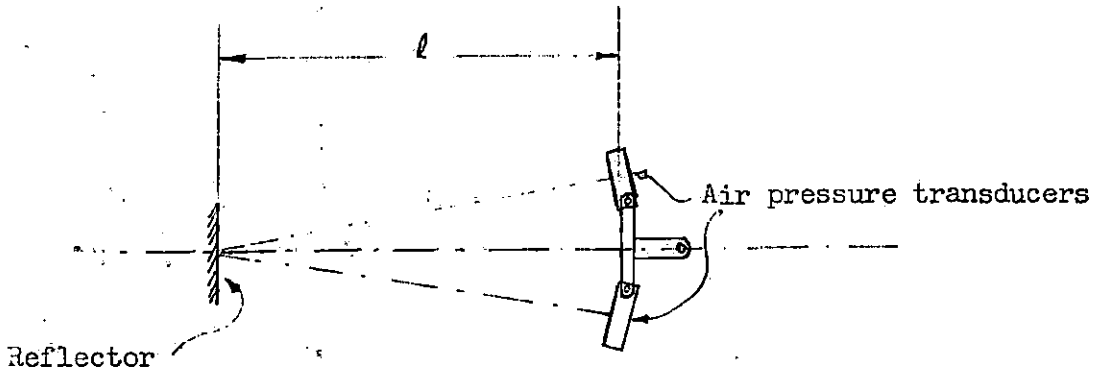

Figure 1. Illustration of relative positioning of the transducers and the chest

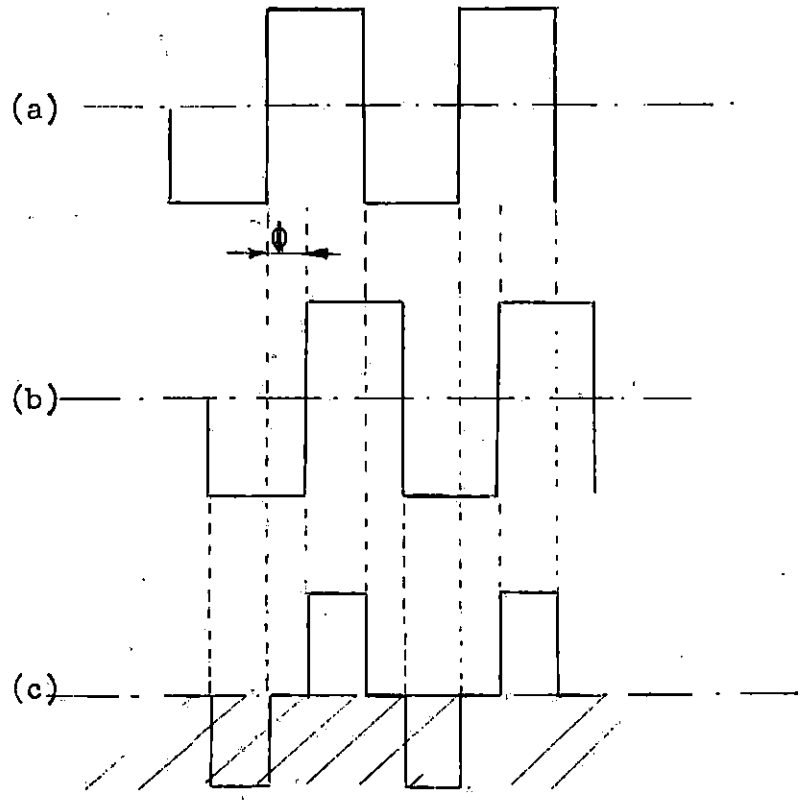

Figure 2. A geometrical demonstration of why the summation of two equal amplitude square waves represent the time lag between the two waves. ϕ is the time lag between (a) and (b). Part (c) is the sum of (a) and (b)

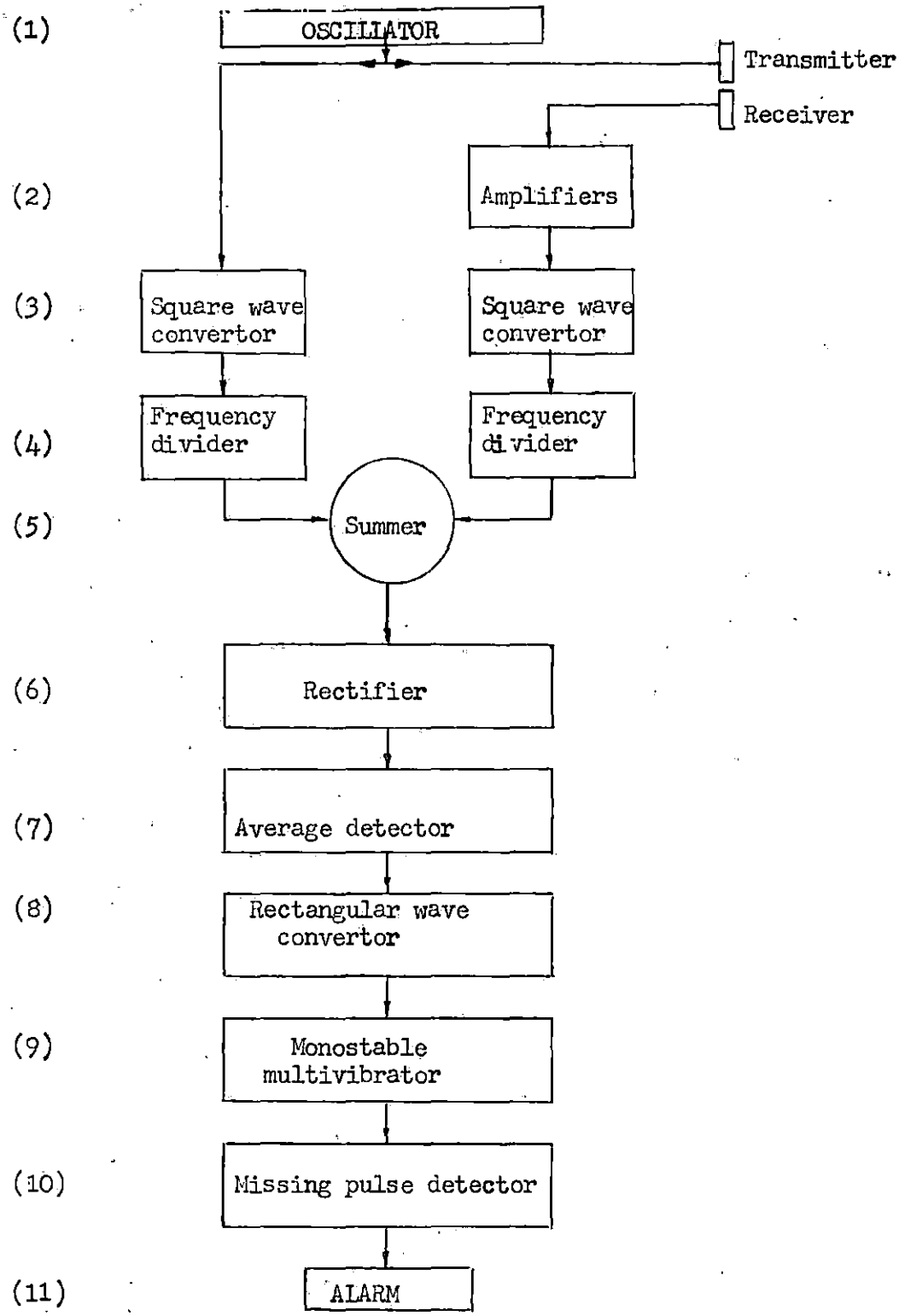

Figure 3. Block diagram of the ultrasonic apnea monitor

oscillator. Two stages of amplification rather than one are used to avoid reduction of the bandwidth.

Block 3 in the diagram produces a pair of square waves at the same frequency of the oscillator. One of these square waves is in phase with the transmitted signal and the other square wave is in phase with the received signal.

In block 4 two simple monostable multivibrators halve the frequencies of the square waves and increase the apparent wavelengths to overcome the double peaking form of the respiration pattern due to large displacement of the chest wall in deep respiration such as gasps or sighs. In deep respiration the displacement of the chest wall may be larger than half of the wavelength (4 mm). In such cases, the phase angle will repeat itself and causes the respiration pattern to have two peaks for each cycle of breathing. This is considered a distorted signal and should be eliminated.

At block 5 a CMOS fast operational amplifier is used to add the outputs of the monostable multivibrators of block 4.

Blocks 6 and 7 are simply a half-wave rectifier and an average detector, respectively. The output of the average detector is a low-frequency voltage proportional to the phase angle difference between sinusoids at the transducers. The magnitude of this voltage fluctuates with respiration.

The waveform of this fluctuation is the reproduction of the chest wall movement, and is an alternating voltage added to a constant voltage.

Blocks 8 through 11 are the alarm section. Block 8 is a comparator which converts the fluctuating voltage, proportional to the phase angle difference, to a rectangular waveform. At the negative going edge of each of the rectangular cycles, a pulse is generated by a monostable multivibrator. Each pulse represents a complete cycle of respiration.

Block 10 is a missing pulse detector. This block triggers the alarm whenever no pulse is generated during a specified period of time.

Detailed circuitry and the waveforms of signals at several points along the path are given in Appendix A.

Computer Programming for Detection of Upper Airway Occlusion

The computer portion of this monitor, designed to detect Upper Airway Occlusion (U.A.O.), is based on the following criterion. During U.A.O. the respiratory waveform is both decreased in rate and prolonged in width. Experiments on puppies used for evaluation of this technique have showed that the respiration rate decreases to two-

thirds of the original rate, and the duration or width of the waveform increases by a factor of 1.5. These characteristics are used to detect U.A.O. A generalized flow diagram of the computer program is in Figure 4.

Block 1 of the flow chart in Figure 4 measures the width and the period of the respiratory cycles and stores them in designated memory locations. This is accomplished by repeatedly taking samples and comparing them with a threshold. Whenever the sample is greater than the threshold, it indicates the onset of a respiration cycle. The computer then waits for 10 milliseconds, increments the time count index, and then takes another sample. This loop is repeated until a sample is smaller than the threshold. The time count index represents the width of the cycle. The action of taking a sample and waiting for 10 milliseconds continues until a new cycle is detected. Now the time count index indicates the period of the cycle. The whole procedure will be repeated six times.

The second block simply finds the average values of the widths and the periods of the first five cycles.

In the third block, the sixth cycle is compared to the results of block 2. At this time only the widths of the cycles are compared, because changes in the width exceed other changes in the waveform of the cycle during

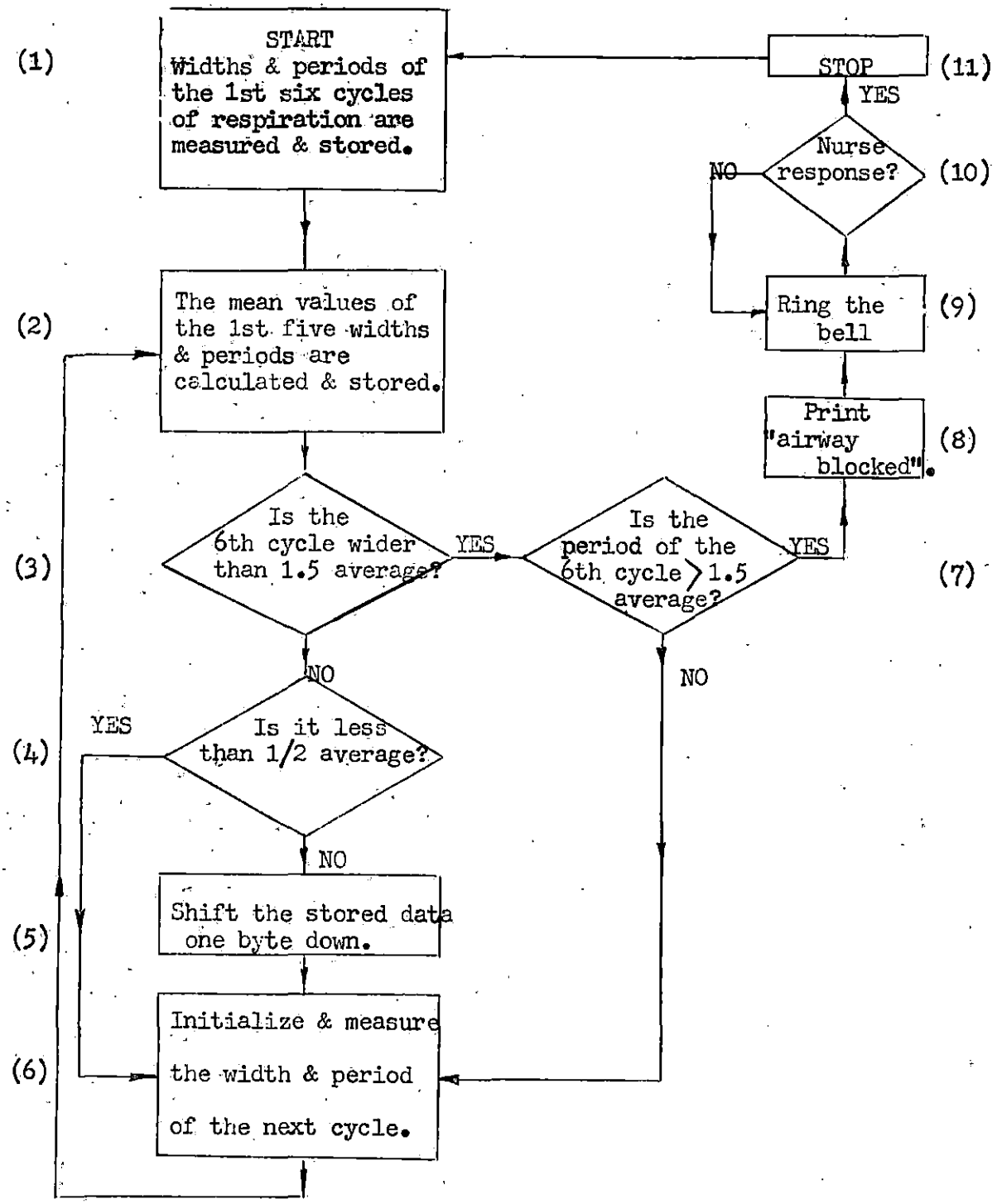

Figure 4. Generalized flow chart of the U.A.O. detecting program.

U.A.O. If the width of the 6th cycle is greater than 1.5 times the average value, there might be an U.A.O. occurring. If not, then the computer checks if there has been any body movement. This is done by block 4. Experiments on dogs have shown that when the body is moving, the widths of the cycles are very small, often less than 1/3 of the mean value. In block 4 the width of the sixth cycle is compared to 1/2 of the average value. If the width is more than 1/2 the average value, the cycle is considered normal and the computer proceeds to the fifth block. If the width is less than 1/2 the average value, block 5 is by-passed and the flow of the execution of the program shifts to block 6.

In block 5 the data regarding the first cycle are erased and the next five cycles are considered as the cycles for calculation of the average values of width and periods.

Block 6 initializes the program and measures the width and period of the next coming cycle, and then shifts the flow of the program to block 2.

The computer will go through this loop from block 6 to block 2 so long as respiration is normal. If the respiration becomes abnormal then the computer branches out of the loop and goes to block 7.

In block 7 the period of the sixth cycle is compared to

1.5 times the average period. If the period has not increased, that cycle is considered as a deep breath or sigh and ignored. If the period is more than 1.5 times the average value, then an U.A.O. has occurred. Block 8 prints the message "Airway blocked" on the screen of the terminal.

Blocks 9 and 10 ring the bell, until the attendant responds and stops execution of the program. The detailed program is in Appendix B. A digital PDP 8/e minicomputer was used.[1]

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[1]Digital Equipment Corporation, Maynard, Mass.

EXPERIMENTAL EVALUATION

Method

In this section evaluation of the general operation as well as the reliability and consistency of the ultrasonic respiration monitor are described. The general operation was tested by checking the detection of motion, and the reliability was verified by comparing the outputs of this device and that of a reliable monitor.

In the first stage of this experiment the general operation of the system was tested. For this purpose a plate was attached to a rod secured to the piston of a respiration pump[1] as shown graphically in Figure 5. When the pump was turned on, the piston would slide back and forth; thus, the movement of this plate would be a simulation of a chest wall respiration pattern. The apnea monitor should have turned the alarm on whenever the pump was off for a predetermined length of time.

In the second stage of this experiment, using a live subject, the consistency and similarity between the waveforms of the ultrasonic monitor and another reliable monitor were verified. As a live subject, a mongrel young dog was chosen for two reasons:

1) there was no safe and easy access to human infants,

2) a puppy dog would breathe shallower than a grown dog, and hence there would be more similarity between this case and the case of a newborn.

[1]Model #507, respiration pump, Harvard Apparatus, Hillis, Mass., 02054.

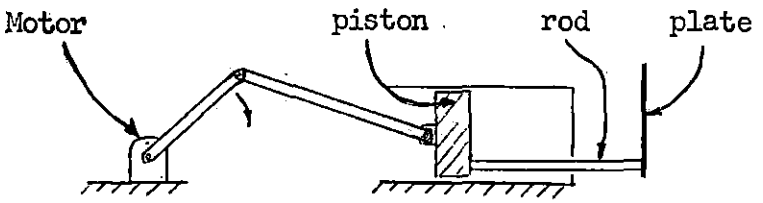


Figure 5. A plate attached to the piston with a rod to simulate the chest wall respiration pattern

For evaluation and verification of the accuracy of the ultrasonic monitor, a thermistor was used. As mentioned earlier in literature review, thermistors detect changes in the temperature of the air surrounding the device. When a thermistor is placed in the oral or nasal airflow, it will detect the temperature changes during expiration and inspiration; therefore, it reliably detects respiration. Furthermore, if the airway is blocked, since there is no air flow around the thermistor element, the blockage is reflected in the temperature of the thermistor by a constant decrease, a case which very well indicates upper airway blockage.

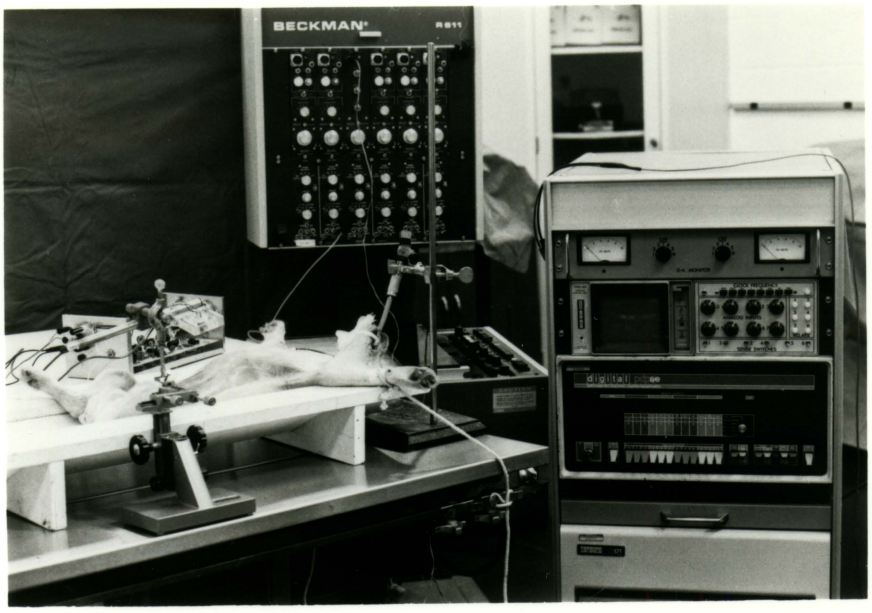
In this experiment, a thermistor was placed in the lumen of an endotracheal tube and the dog was anesthetized with pentobarbital (0.25 ml/kg) and then intubated. Anesthetization makes the dog more cooperative during intubation and prevents the dog from changing position during the experiment. Fortunately, this situation is similar to that of a premature infant which rarely moves about.

After intubation, the dog was shaved on the chest and abdomen and placed on a V-board in the supine position. The thermistor output was fed into the thermistor coupler of a Beckmann recorder. Then the transducers were placed in front of the abdomen of the dog and the distance between the

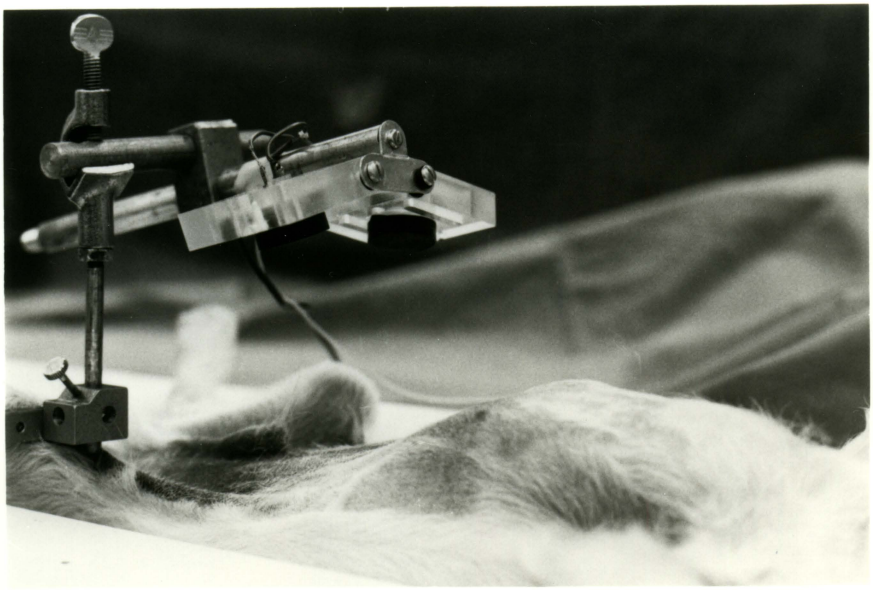
transducers and the abdomen was adjusted in such a way that normal lines protruding from the transducers intersected at a point near the skin surface (see Figure 6). The transducers were then connected to the monitor and the output of the monitor was fed into both the minicomputer and the Beckmann recorder.

First, normal patterns of respiration were recorded. Then the dog was moved to different positions and recordings were obtained. Finally, using a hemostat, the endotracheal tube was clamped while the waveform pattern was being recorded. An assistant marked the recordings whenever upper airway occlusion was applied or removed, and whenever occlusion was detected by either the computer or the monitor.

Figure 6. Illustration of the set-up of the experiments: (a) the anesthetized dog is intubated and placed in supine position on a V-board. The transducers are placed in front of the abdomen and connected to the ultrasonic apnea monitor. The output of the monitor is fed into both the Beckman recorder and the minicomputer, (b) the transducers are placed in front of the area immediately below the thorax in such a way that the normal lines protruding from the transducers intersect near the skin surface



(a)



(b)

RESULTS

The experiment was run several times. In the first experiment, recordings of normal respiration patterns were made. Figure 7(a) shows a sample of that recording. Figures 7(b) and 7(c) show the respiration pattern while upper airway occlusion was applied at the end of inspiration, and expiration, respectively. Note that there are two peaks in the monitor trace for every cycle of respiration.

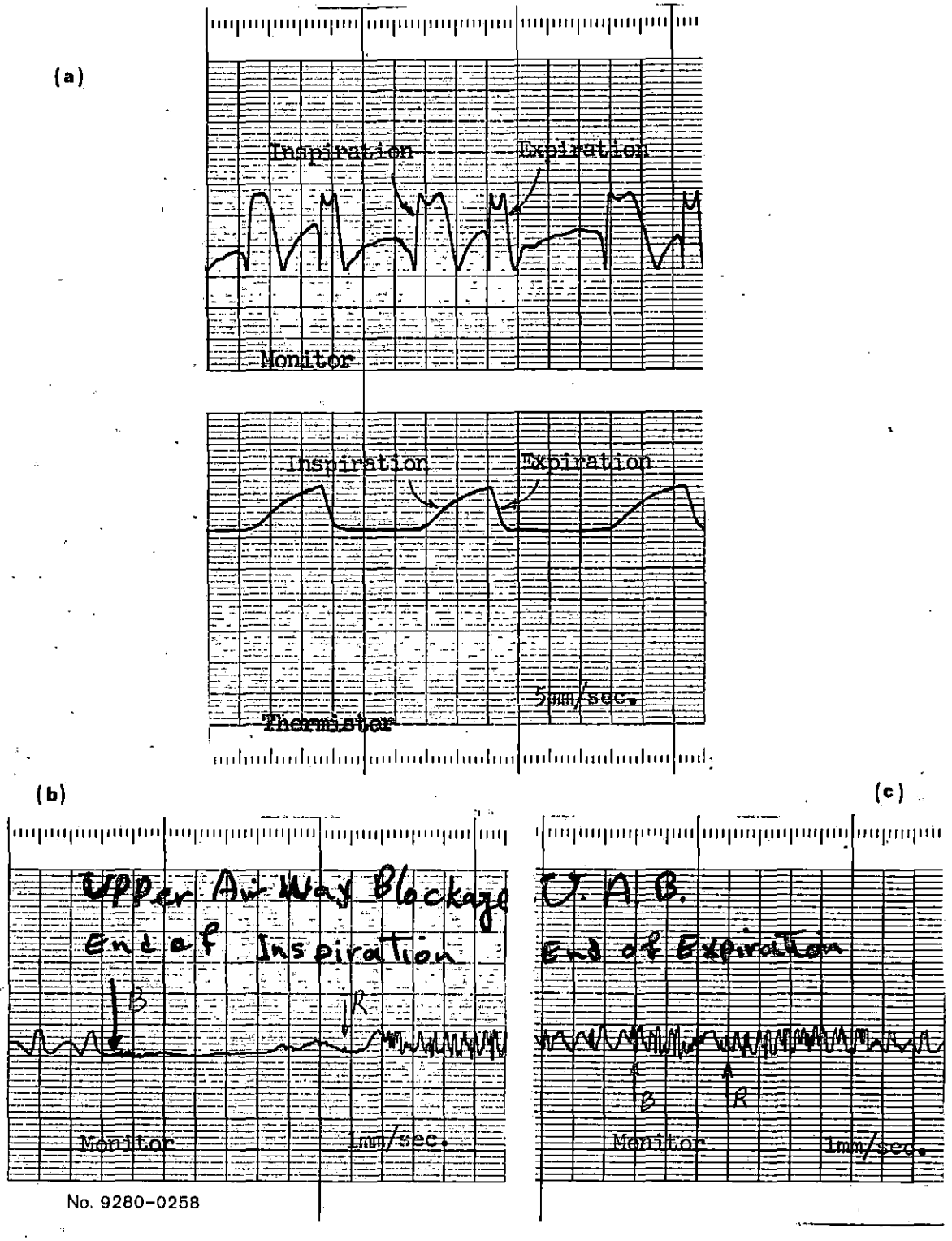
In the second experiment, a pair of frequency dividers were added to the monitor to eliminate the double peaking effect. The recordings of respiration while the dog was in different positions were then taken. Figure 8(a) is the recording taken while the dog was lying on its side. Figures 8(b) and 8(c) show the recordings while the dog was in the supine position and the transducers were facing the dog's chest and abdomen, respectively. Note that in Figures 8(b) and 8(c), heart beats are picked up and recorded by the monitor.

During different phases of respiration (in the middle or at the end of inspiration or expiration) the upper airway was occluded and the waveforms were recorded. Figure 9 shows these recordings. In all cases except when U.A.O. was applied at the end of expiration, the monitor alarm was

Figure 7. Recordings of respiration pattern:

(a) normal respiration pattern,
(b) respiration pattern with U.A.O. applied at the end of inspiration,
(c) respiration pattern with U.A.O. applied at the end of expiration

(B indicates application of occlusion and R indicates its removal)


(a)
Inspiration
Expiration
Monitor
Inspiration
Expiration
5mm/sec.
Thermistor
(b)
Upper Air Way Blockage
End of Inspiration
B
R
Monitor
1mm/sec.
(c)
U. A. B.
End of Expiration
B
R
Monitor
1mm/sec.

No. 9280-0258

Figure 8. Recordings of respiration with the dog in different positions: (a) the dog laid on its side, (b) the dog in supine position and the transducers facing the chest, (c) the dog in supine position and the transducers facing the abdomen

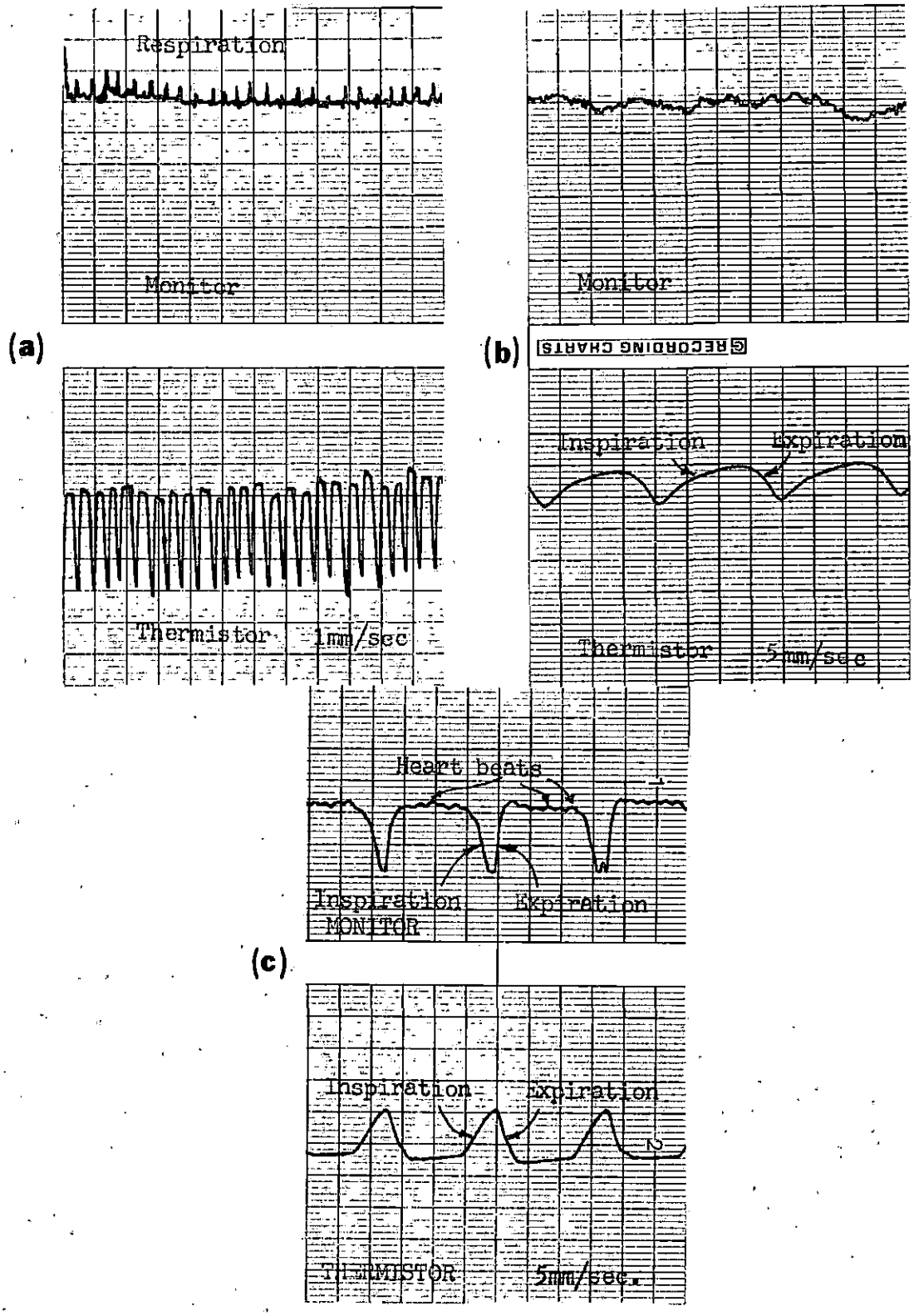
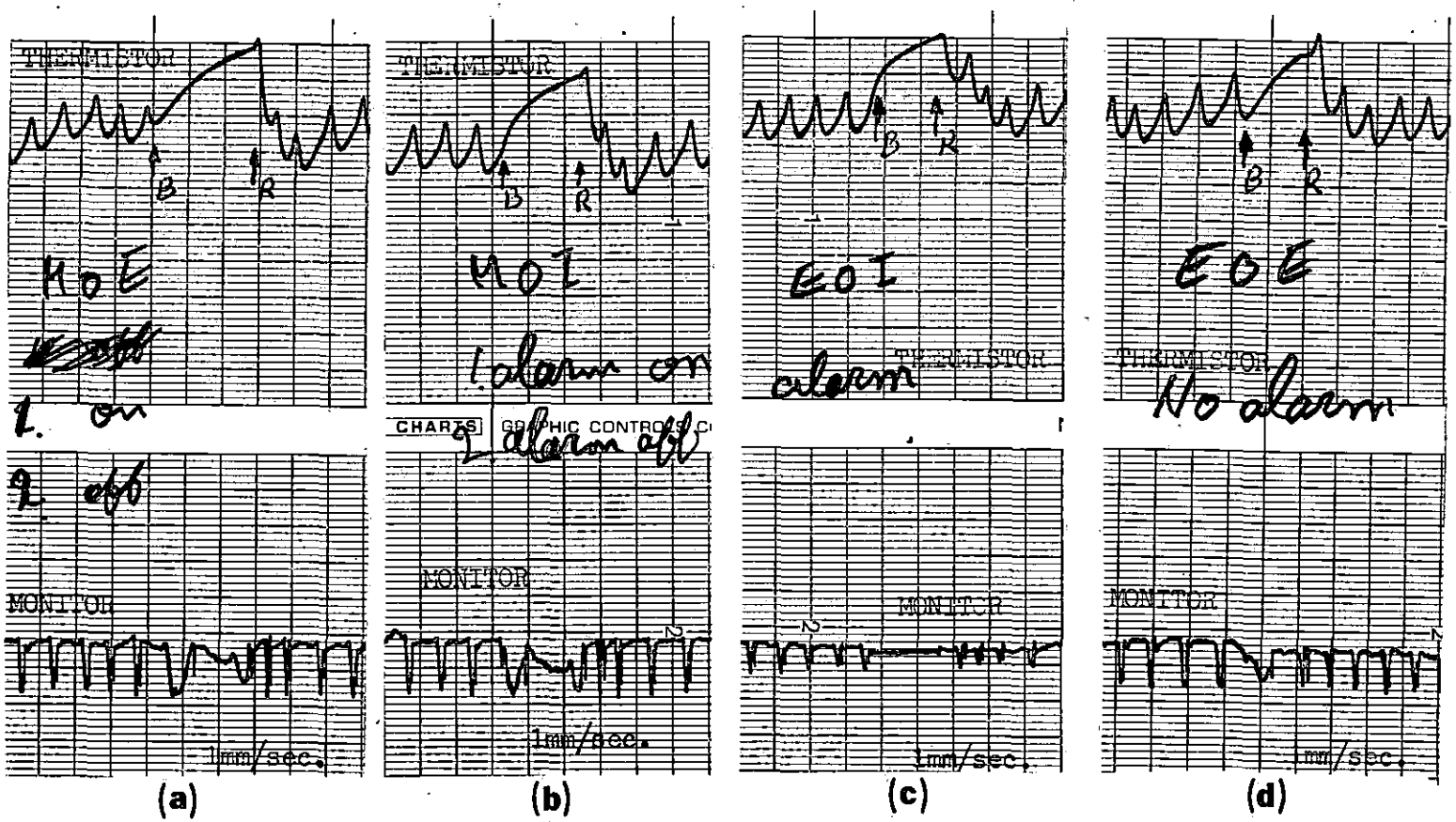

Respiration
Monitor
(a)
Monitor
(b)
RECORDING CHARTS
Thermistor 1mm/sec
Inspiration
Expiration
Thermistor 5mm/sec
Heart beats
Inspiration
MONITOR
Expiration
(c)
Inspiration
Expiration
THERMISTOR 5mm/sec.

Figure 9. Changes in the pattern of respiration when U.A.O. is applied at the (a) middle of expiration, (b) middle of inspiration, (c) end of inspiration, (d) end of expiration. Note that only in case (d) U.A.O. was not detected by the monitor


THERMISTOR
B
R
EOE
1. on
2. off
MONITOR
1mm/sec.
(a)
THERMISTOR
B
R
NOE
1. alarm on
2. alarm off
CHARTS
MONITOR
1mm/sec.
(b)
B
R
EOE
alarm
THERMISTOR
MONITOR
1mm/sec.
(c)
THERMISTOR
B
R
EOE
No alarm
MONITOR
mm/sec.
(d)

activated. It should be mentioned that the monitor time threshold was set at 7 seconds and no computer was used for detecting U.A.O.

In the next experiment, a minocomputer was used for detection of U.A.O. The results are as follows. First, normal respiration was recorded; it was noticed that occasionally, because of external noise, the phase of the monitor waveform changed 180°, as shown in Figure 10. Denote the resultant phases "phase a" and "phase b". Also, note the presence of heart beats in the monitor waveform. Then the upper airway was occluded several times. The results showed that when the waveform was in "phase a" and there was no phase shift during the occlusion, the computer detected U.A.O. and triggered the computer buzzer (Figure 11a). When the waveform was in "phase b" and there was no shift to "phase a", the computer did not detect the occlusion (Figure 11b). When there was a shift from "phase a" to "phase b", two out of five applications of U.A.O. were detected by the computer, while only one U.A.O., was detected by the monitor alarm (Figure 12). When there was a shift from "phase b" to "phase a", out of four applications of U.A.O., one was detected by the computer and one was detected by the monitor alarm (Figure 13). Note that in all figures (Figures 8-13), heart beats are recorded.

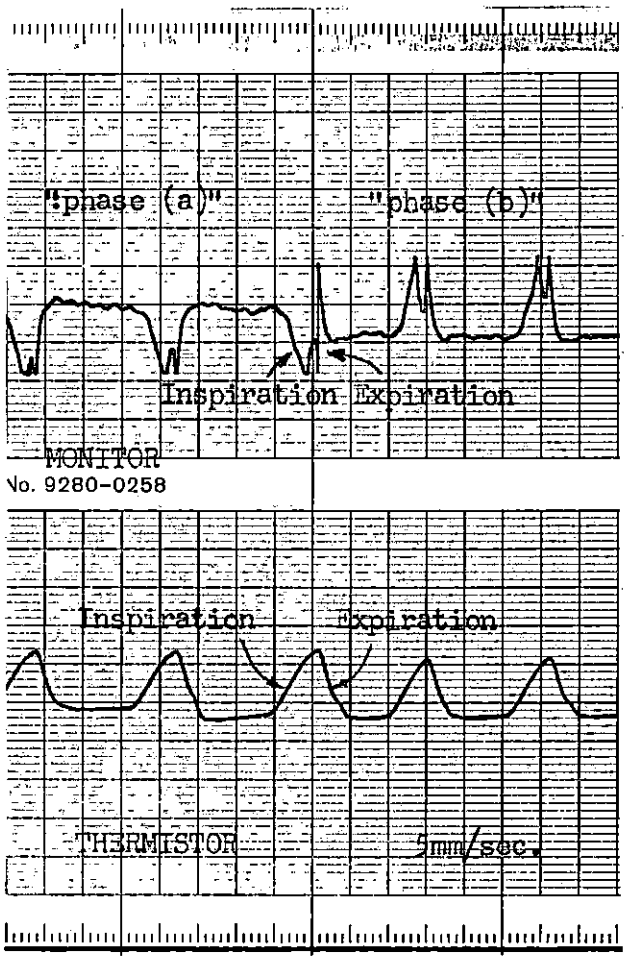


Figure 10. Occasional change in the monitor waveform phase

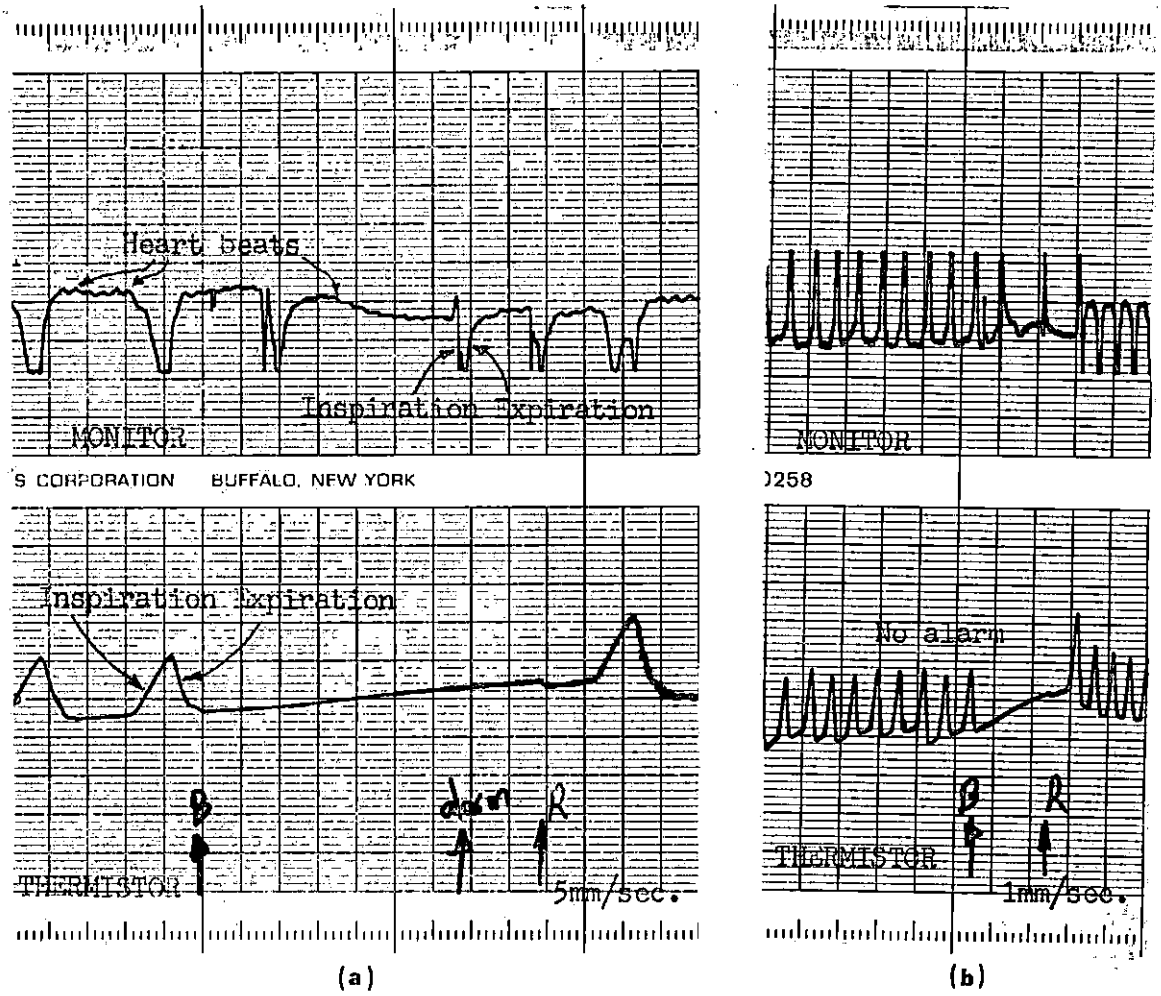


Figure 11. Detection of U.A.O. when no phase shift is present: (a) waveform in "phase a", (b) waveform in "phase b". The computer detects U.A.O. if the monitor waveform is in "phase a" and the phase does not shift. Blockage was applied at point B and removed at point R

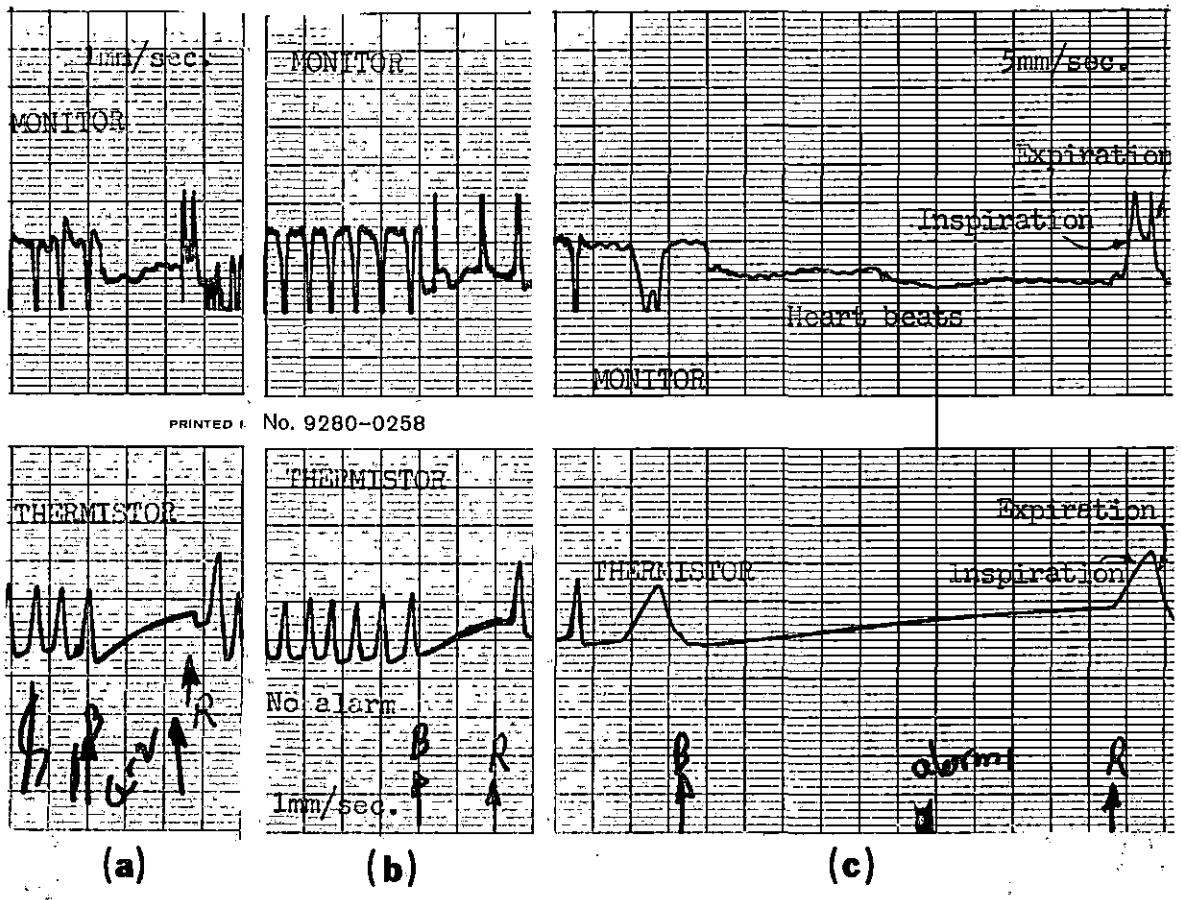


Figure 12. Detection of U.A.O. when a shift from "phase a" to "phase b" occurs. Case (a) was detected by the computer. Case (b) was not detected. Case (c) was detected by the monitor alarm. Blockage was applied at point B and removed at point R

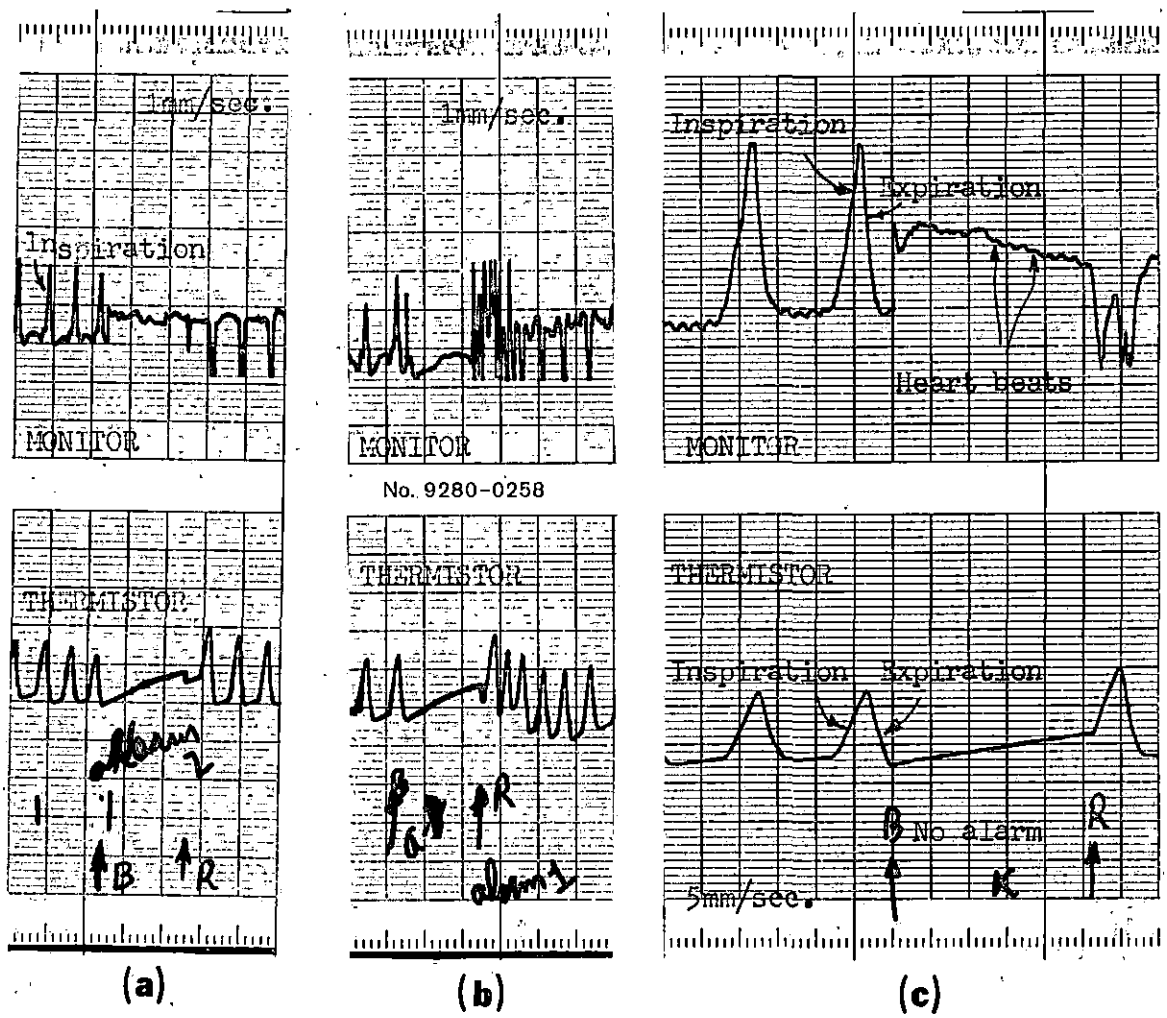


Figure 13. Detection of U.A.O. when a shift from "phase b" to "phase a" is present. Case (a) was detected by the computer. Case (b) was detected by the monitor alarm. Case (c) was not detected. Blockage was applied at point B and removed at point R

DISCUSSION

Implications of Results

The performance of the ultrasonic apnea monitor indicates that under certain conditions it detects respiration motions very well and reproduces the respiration patterns acceptably. In all experiments on dogs there was not a single incident of false alarm, even in cases where respiration was shallow or the dog was lain in a position such that the abdomen wall displacement was very small. Figure 8(a) supports such a statement.

The monitor is sensitive enough to pick up the heart beats, as shown in Figures 8 through 13, but the alarm section ignores such motions. Figure 12(c) and 12(b) are good examples of such cases that during a period of apnea even though the heart beat was detected and recorded, the monitor alarm was activated. It is the presence of respiration, not the heart beat, which turns off the activation of the alarm.

The faithfulness of reproduction of the respiration pattern is demonstrated in Figures 8-13. The slight difference between the widths of the monitor waveform and the thermistor pattern is due to the slow response of the thermistor. The different waveforms (double peaking waveform) of Figure 7 (and others) are also due to the small wavelength of the ultrasonic signal. As mentioned in the

previous chapter, if the abdomen wall displacement is larger than half the ultrasonic wavelength, the phase angle difference exceeds 2π radians and repeats itself; thus, a double peaking waveform is generated. This problem was solved by adding a pair of frequency dividers to the system. The frequency dividers increase the apparent wavelength and thus eliminate the problem. (There was no frequency divider in the circuit when the recording shown in Figure 7 was made.)

In relation to faithfulness of the reproduction of the respiration waveforms, one may point to Figures 8(a) and 8(b) as counter examples. Figure 8(a) shows that when the transducers are facing the side of the dog respiration is detected but the waveforms are not similar to the waveforms of the thermistor.

Figure 8(b) shows that if the transducers are facing the chest of the dog, the monitor can not detect respiration, mainly because of the shape of the thoracic bone of the dog which reflects the transmitted signals to a direction different from the position of the receiver transducer.

Thus we come to the conclusion as suggested by Figure 8(c) that the best position for monitoring respiration of a dog, using this monitor, is the supine position while the transducers are facing that part of abdomen immediately

below the thorax. It is this position which is used for detection of upper airway occlusion.

Upper airway occlusion can be detected under certain conditions. The monitor, without the help of a computer, can detect U.A.O. if the occlusion occurs at any phase of respiration except at the end of expiration and the monitor alarm is set for a 10 second or shorter apnoeic attack. Figures 7(b) and 7(c) and Figure 9 can be used to show the reasons, in this effect. Figure 7(b) shows that when the airway is occluded at the end of inspiration no immediate respiratory effort is made by the dog, a condition which triggers the alarm. The lack of respiration is due to the stimulation of inflation receptors in the lungs which inhibit respiration.

Figure 7(c) shows a recording of respiration when the upper airway is blocked at the end of expiration. Respiratory efforts are still present, because respiratory center is not inhibited by the inflation receptors in the lungs. Yet the pattern is considerably different from that of Figure 7(b). It is this case which is not detected by most motion detecting monitors.

Figure 9 shows recordings of U.A.O. when applied at different phases of respiration. In Figure 9(a) the airway is blocked in the middle of expiration. The monitor alarm was triggered, but was deactivated when the next effort for

respiration was made. The case of Figure 9(b) had the same kind of response. In Figure 9(c) the alarm was triggered and was not deactivated until the blockage was removed. In the case of Figure 9(d), the alarm remained silent during U.A.O.

For the U.A.O. applied at the end of expiration to be detected, a minicomputer should be used, but first the specific changes in the patterns of respiration during U.A.O. should be studied to permit an appropriate computer program to be written. Figure 14 shows the typical changes in the respiration waveform, when U.A.O. is applied.

It should be noted that the pattern of respiration during U.A.O. changes in respect to both width and rate. When a respiratory effort is initiated, the chest begins to expand, but the upper airway is blocked and the chest cannot expand easily. Since the inflation receptors are not stimulated to inhibit respiration, the chest continues to expand, and thus the width of the respiratory pattern increases. Tests have shown that the increase is at least 50% of the original width.

For the same reasons, the period of the waveform increases. But the second respiratory effort after application of U.A.O. is initiated sooner and stronger compared to normal respiratory efforts. Therefore, only the first cycle after application of U.A.O. undergo significant changes.

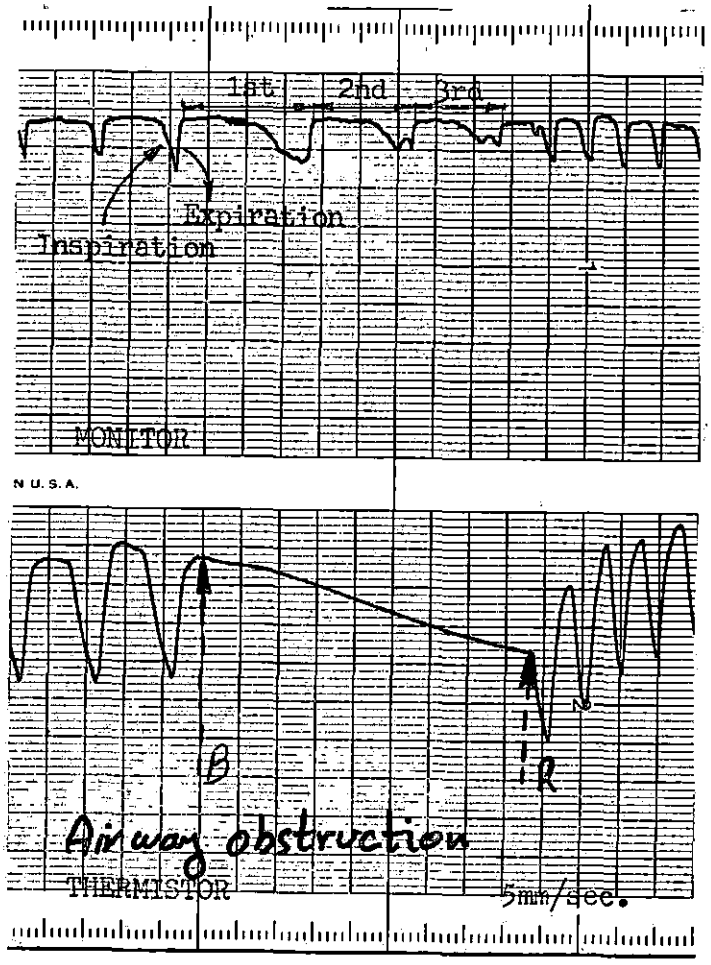


Figure 14. Changes in the respiration pattern when upper airway is occluded at the end of expiration. The width increases and the rate decreases. Note that the 2nd cycle of respiration after blockage was applied is smaller in period than the 1st cycle immediately after U.A.O. Blockage was applied at point B and removed at point R

These changes in the respiratory pattern cannot be mistaken for changes due to gasps or sighs, because in gasps and sighs only the amplitude of the waveform increase, not the width.

The minicomputer has shown to be effective in detecting U.A.O. However, there are cases which may not be detected by the computer. Phase shift is involved in such situations. The phase shift is believed to be due to external electrical noise which makes the monostable multimultivibrator output (described on page 23) to go high on the edge of the second cycle of the input signal, and thus changing the phase angle. As shown in Figure 10, when phase angle changes, a different threshold is required for detection of the onset and the end of each cycle of respiration pattern, and thus the computer program used is unable to detect U.A.O.

However, in cases where phase shift is present, U.A.O. may still be detected, e.g., Figures 12(a), 12(c), 13(a) and 13(b). Among these cases, 12(c) and 13(b) were detected by the monitor, and the rest were detected by the computer. The computer detected these cases because the threshold needed for the detection of the onset and the end of each cycle of respiration had changed very little, a condition in which the computer could distinguish between different cycles of respiration from each other (see Figures 12(c) and

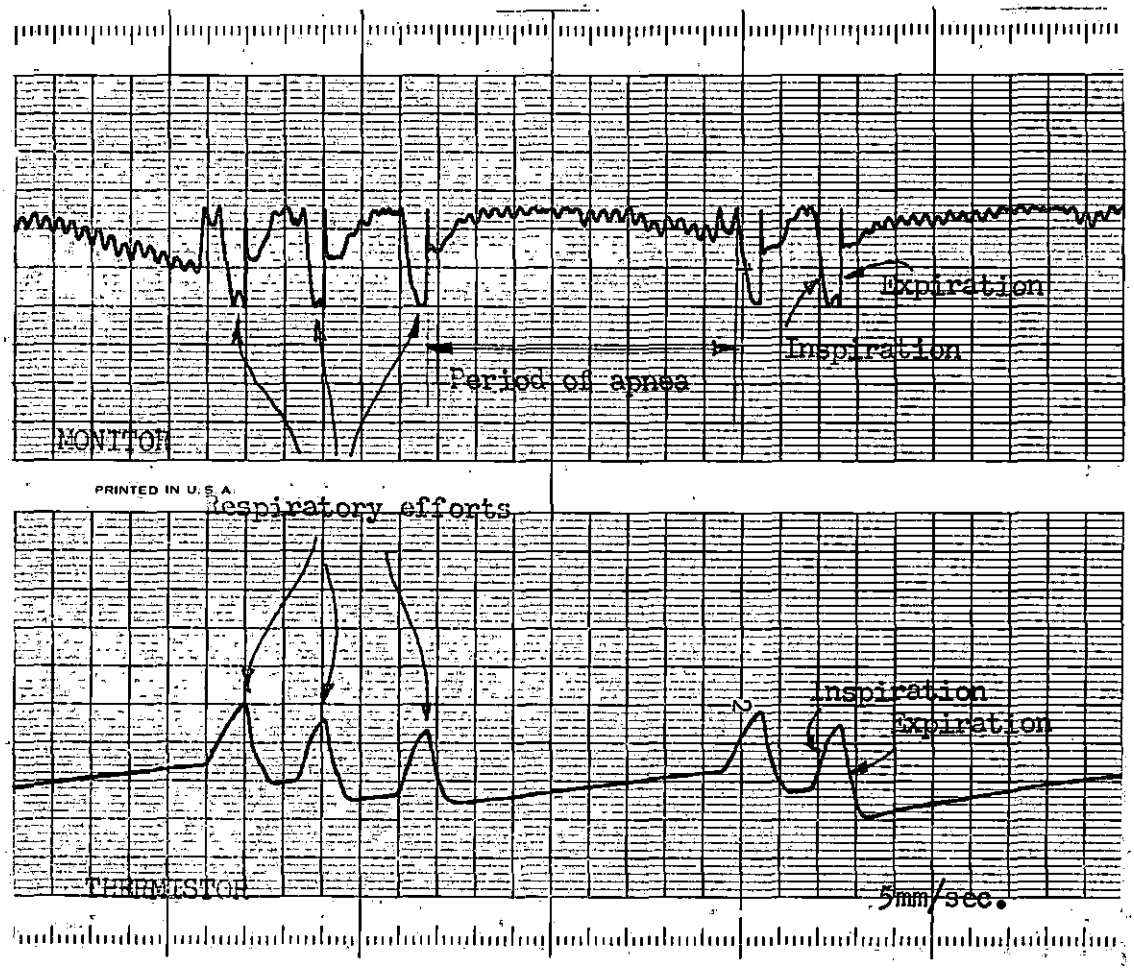


Figure 15. Biot breathing pattern. In this pattern, several respiratory efforts are made followed by a period of apnea

13(a)).

Figure 12(b) shows a case in which, even though the threshold change was small, U.A.O. was not detected, because what is considered as the width of the signal in "phase a", is considered the period of relaxation of respiration in "phase b". Thus, when a phase shift from "phase a" to "phase b" occurs, since there is an apparent decrease in the signal width, the computer does not consider such a change a blockage. The computer could be programmed to detect an increase in the period of relaxation, as a blockage, but then the Biot breathing pattern cannot be detected. Figure 15 shows a sample of Biot breathing pattern which was recorded during one of the tests. In this pattern there is a burst of respiration and then a period of apnea. The computer has been programmed to detect Biot breathing pattern and disregard it. In Figure 12(b), the case appears to the computer to be a Biot breathing pattern and is disregarded.

Conclusion and Suggestions for Improvements

Apnoeic attack, the absence of breathing for a period of time, is usually seen in premature infants during the first days or even during the first hours after birth. It is a condition which in most cases results in the death of

the infant or may cause hypoxic brain damage among survivors. In orders to prevent such disasters, apnea monitors are instituted to detect apnoeic attacks and to provide the physician with essential information for treating the patient.

Among different monitors, invasive techniques are the most reliable methods, but they pose additional threats, such as infection or hemorrhage, to the infant. Therefore, contactless monitors are more desirable. The ultrasonic apnea monitor described in this thesis carries neither the threat of the invasive methods nor many of the disadvantages of other contactless techniques.

The ultrasonic apnea monitor utilizes a pair of air-pressure transducers. One of the transducers sends pressure waves to the infant's chest or abdomen. The reflected waves are picked up by the second transducer. The monitor continuously monitors the phase difference between the transmitted and the reflected waves. Whenever the phase difference remained constant, a condition indicating no respiration, the monitor activates the alarm. In addition to this monitor, a minicomputer is utilized to detect the U.A.O. which are not detected by the monitor.

As indicated in the discussion of the results, the ultrasonic apnea monitor is a reliable device which can detect respiration, reproduce the respiration pattern

faithfully, and trigger an alarm whenever apnoeic attack occurs. It functions best when the transducers are facing the area just under the thorax when small dogs are used as the test animal.

The monitor can detect U.A.O. if U.A.O. occurs at any phase of respiration except at the end of expiration, and the monitor alarm is set for a 10 second or shorter apnoeic attack. The computer program can detect U.A.O. occurred at the end of expiration if certain conditions are met. These are: 1) the waveforms should be in "phase a"; and 2) the phase should not shift. The first condition can be met simply by adjusting the distance between the transducers and the abdomen. The second condition can be met by either removing the frequency dividers, which causes the double peaking effect to reappear, or by filtering out the external noise.

A better suggestion for eliminating the phase shift phenomenon is to replace the transducers with lower frequency operating transducers, e.g., 25 KHz, which increases the wavelength, and thus eliminating the phase shift problem. It should also be noted that decreasing the frequency may cause some hearing problems for the patient when the upper limit of hearing sensitivity is approached.

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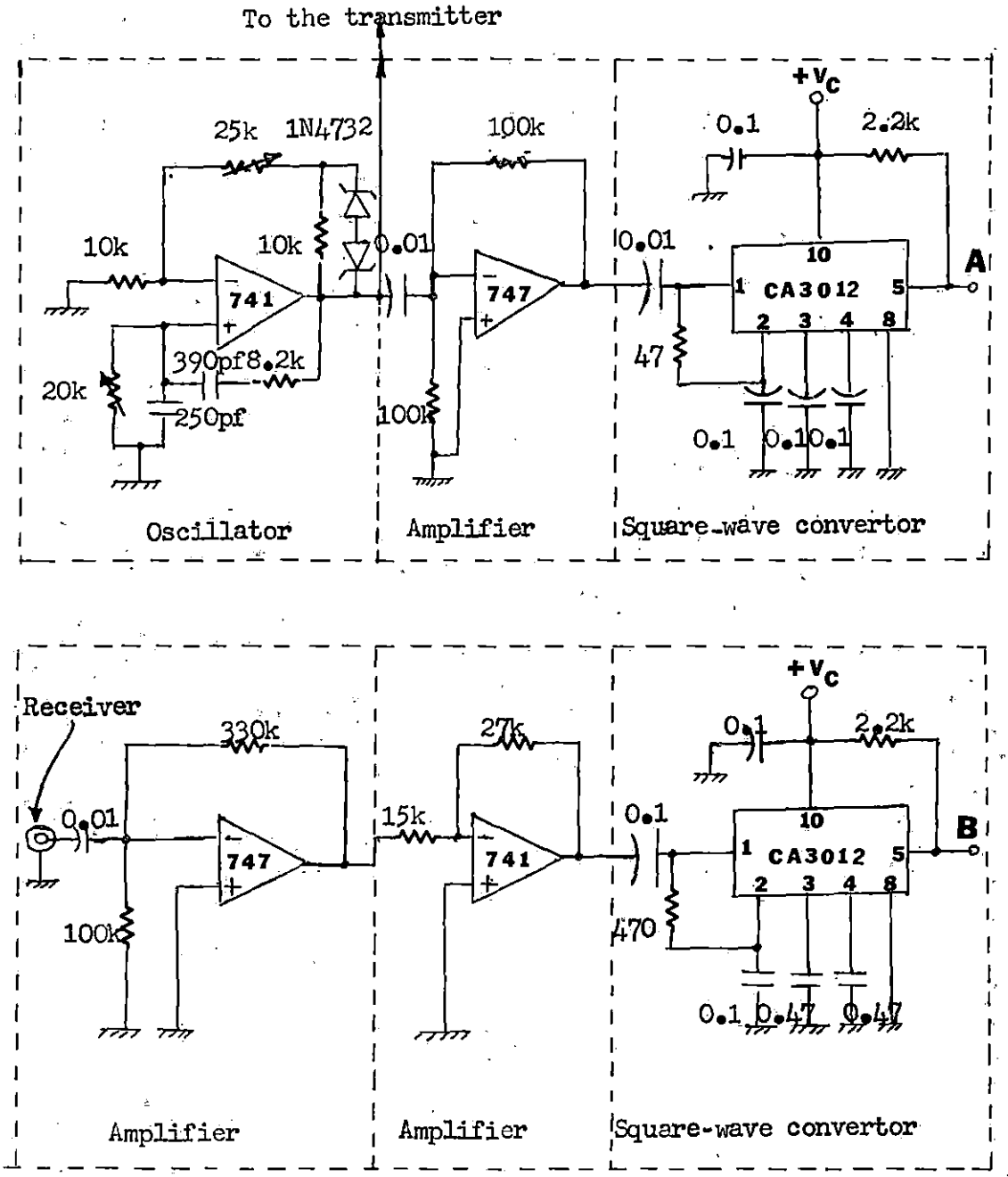
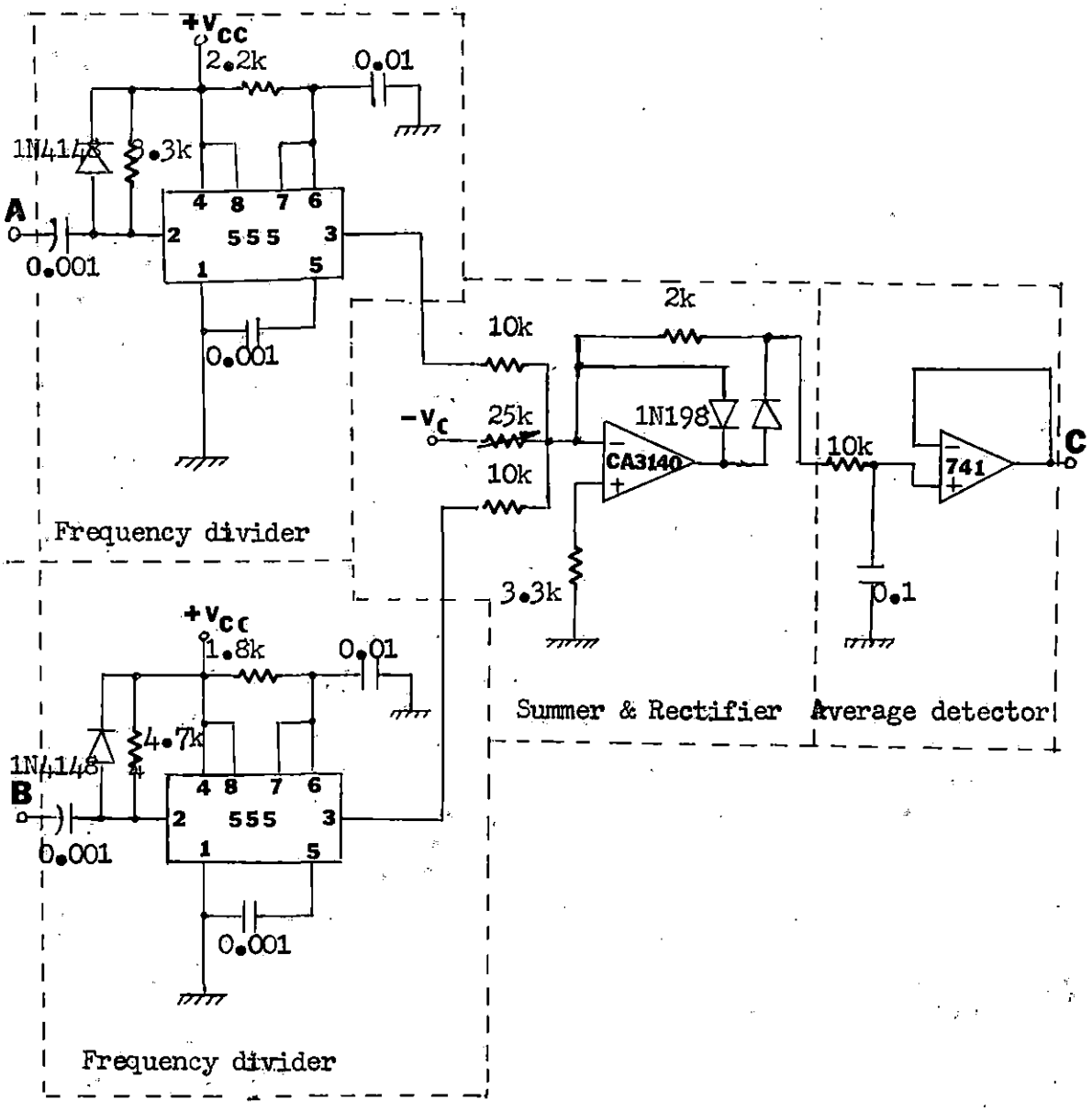
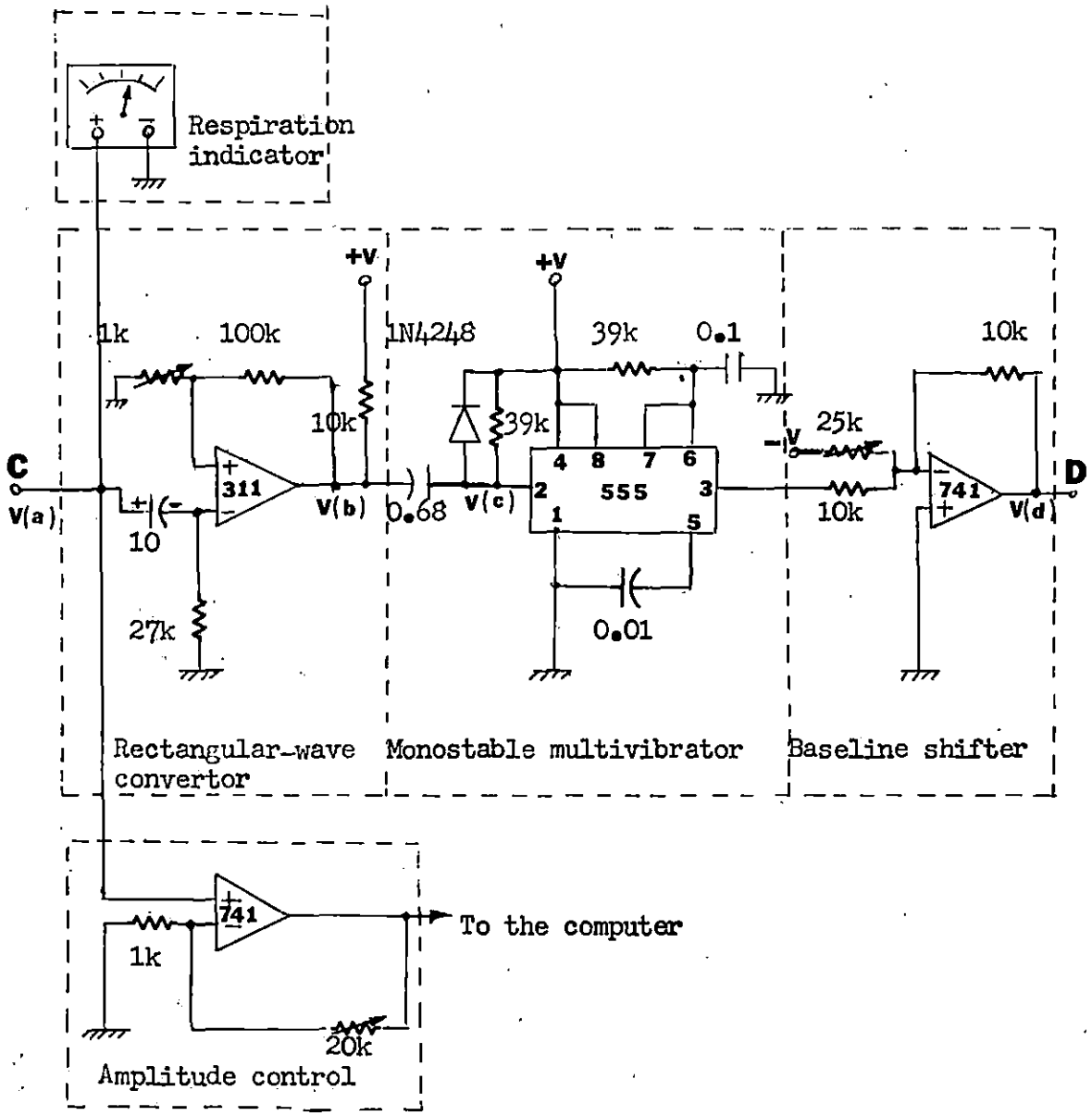
APPENDIX A: ELECTRONIC CIRCUITRY



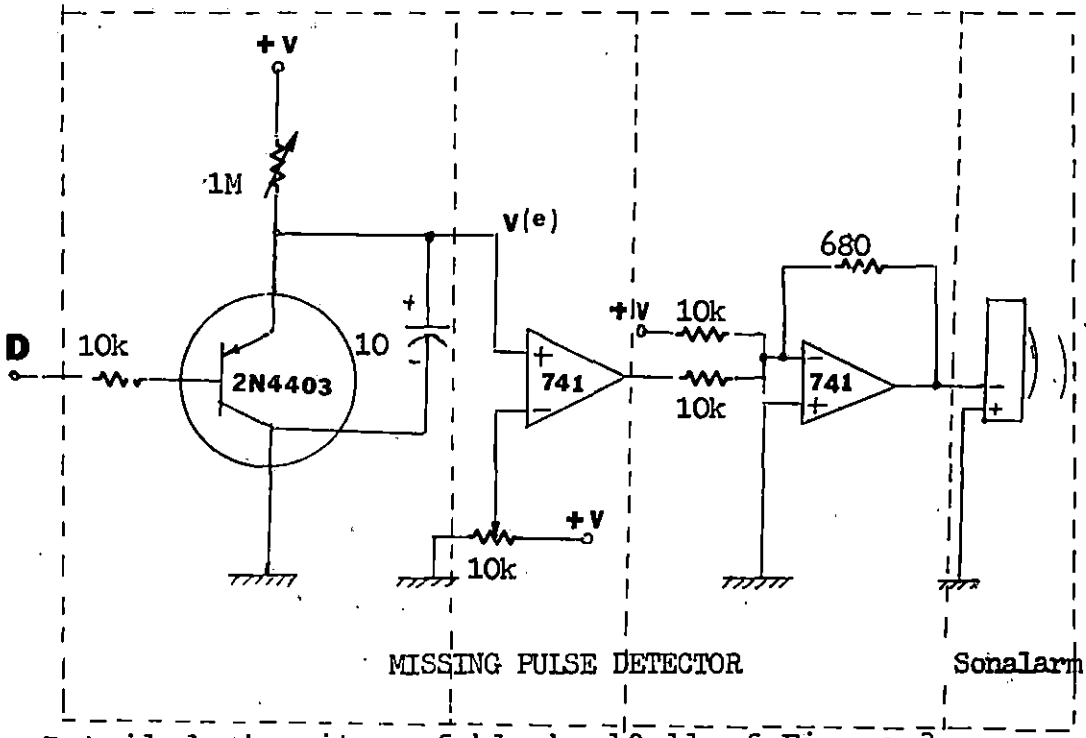
Detailed circuitry of blocks 1-3 of Figure (all resistors are OHMS and all capacitors in microFARADS unless otherwise mentioned)



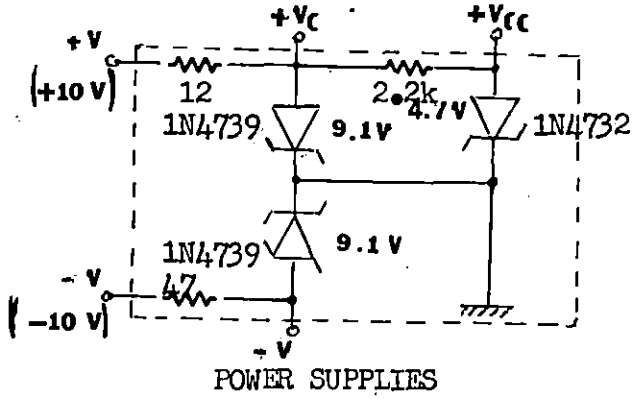
Detailed circuitry of blocks 4-7 of Figure 3



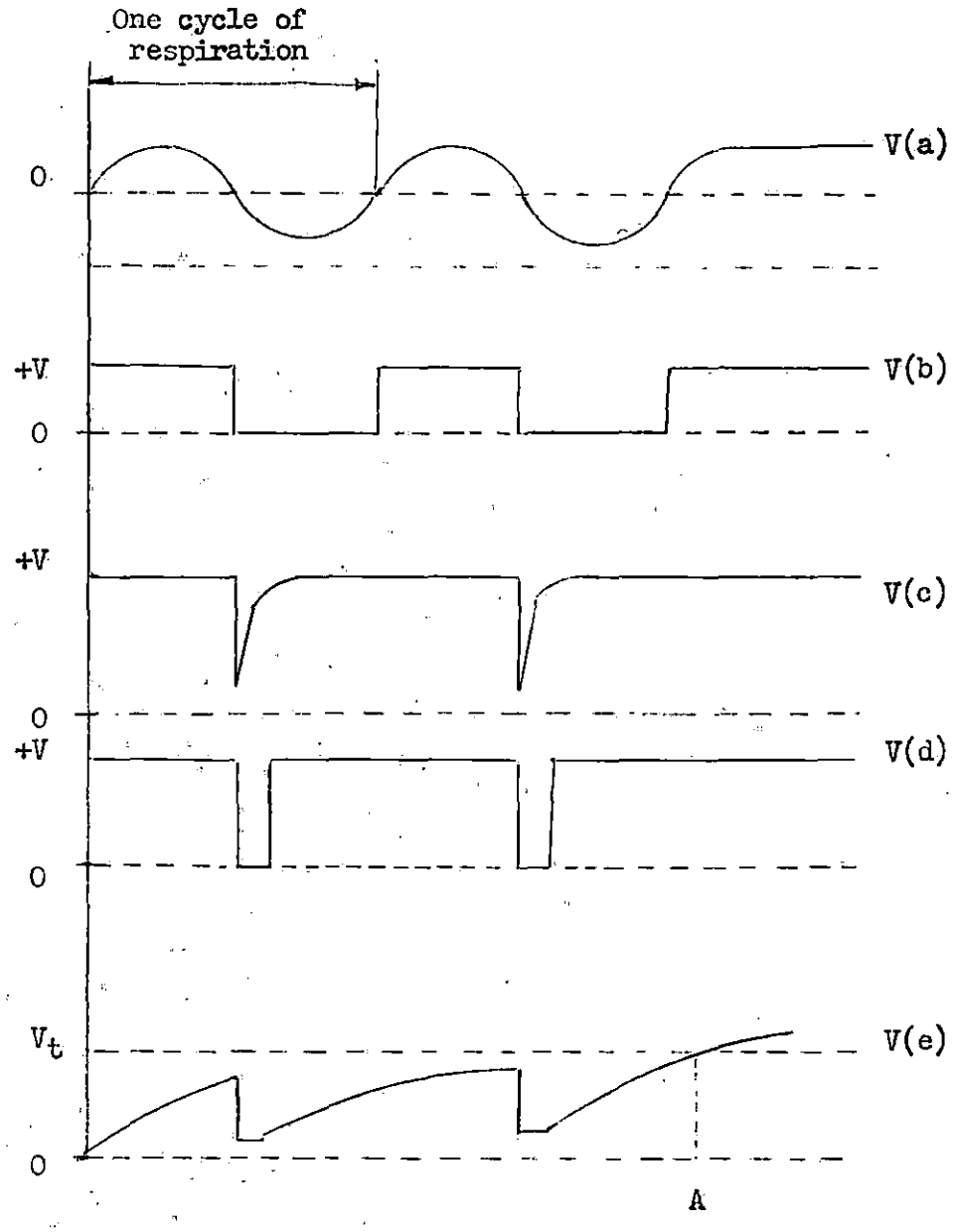
Detailed circuitry of blocks 8-9 of Figure 3



Detailed circuitry of blocks 10-11 of Figure 3



+V and -V are used in the alarm section, $+V_c$ and $-V_c$ are used in the rest of the circuit



Waveforms of signals at several points of the alarm circuitry. The alarm is activated at point A when V(e) exceeds the threshold voltage (V_t)

The ultrasonic apnea monitor.

APPENDIX B: COMPUTER PROGRAMS

```
      0020  *20
00020 0000  WAIT,   000          / SUBROUTINE TO WAIT 10 MSEC.
00021 7200          CLA
00022  7000         NOP
00023 2125          ISZ W
00024 5022          JMP .-2
00025 1126          TAD MW
00026 3125          DCA W
00027 2127          ISZ Y
00030 5022          JMP .-6
00031 1130          TAD MY
00032 3127          DCA Y
00033 5420          JMP I WAIT
00034 0000  TYPE,   000          / SUBROUTINE TO PRINT THE CONTENTS
00035 6041          TSF          / OF THE ACCUMMULATOR.
00036 5035          JMP .-1
00037 6046          TLS
00040 5434          JMP I TYPE
00041 0000  BELL,   000          / SUBROUTINE TO ACTIVATE THE BUZZER.
00042 7200          CLA
00043 1123          TAD K207
00044 4034          JMS TYPE
00045 6501          6501
00046 5042          JMP .-4
00047 7402          HLT
00050 7200  START,  CLA          / THIS PORTION INITIALIZES THE
00051 3074          DCA MAG      / PROGRAM.
00052 3075          DCA T        / T IS THE WIDTH OR THE PERIOD.
00053 3110          DCA QUOT1    / QUOT1 IS THE AVG. OF THE WIDTHS.
00054 3111          DCA QUOT2    / QUOT2 IS THE AVG. OF THE PERIODS.
00055 3106          DCA SUMA
00056 3107          DCA SUMT
00057 1077          TAD MTS      / MTS & TS ARE THE LOCATIONS THE
00060 3076          DCA TS       /PERIODS ARE STORED.
00061 1101          TAD MA       / MA &AS ARE THE LOCATIONSTHE WIDTHS
00062 3100          DCA AS       / ARE STORED.
00063 1103          TAD MCOUNT
00064 3102          DCA COUNT
00065 1112          TAD MB
00066 3113          DCA BS       / BS IS THE LOCATION OF THE 2nd WIDTH.
00067 1115          TAD MC
00070 3114          DCA C        / C IS THE LOCATION OF THE 2nd PERIOD.
00071 1122          TAD MCODE
00072 3121          DCA CODE
00073 5777          JMP BEGIN
00074 0000  MAG,0
00075 0000  T,0
00076 0170  TS,170              / STARTING LOCATION OF THE PERIODS.
00077 0170  MTS,170
00100 0160  AS,160              / STARTING LOCATION OF THE WIDTHS.
```

```
00101  0160  MA,160
00102  7772  COUNT,-6
00103  7772  MCOUNT,-6
00104  7773  M5,-5
00105  0005  MQ,5
00106  0000  SUMA,0
00107  0000  SUMT,0
00110  0000  QUOT1,0          / AVG. OF THE WIDTHS.
00111  0000  QUOT2,0          / AVG. OF THE PERIODS.
00112  0161  MB,161           / LOCATION OF THE 2ND WIDTH.
00113  0161  BS,161
00114  0171  C,171            / LOCATION OF THE 2ND PERIOD.
00115  0171  MC,171
00116  7777  M1,-1
00117  0165  M165,165         / LOCATION OF THE 6TH WIDTH.
00120  0175  M175,175         / LOCATION OF THE 6TH PERIOD.
00121  0140  CODE,140
00122  0140  MCODE,140
00123  0207  K207,207
00124  7773  NUM5,-5
             DECIMAL
00125  7406  W,-250
00126  7406  MW,-250
00127  7770  Y,-8
00130  7770  MY,-8
             OCTAL
       0140  *140
00140  0301                301
00141  0311                311
00142  0322                322
00143  0327                327
00144  0301                301
00145  0331                331
00146  0240                240
00147  0302                302
00150  0314                314
00151  0317                317
00152  0303                303
00153  0313                313
00154  0305                305
00155  0304                304
00156  0215                215
00157  0212                212
00160  0000                000
```

```
        0200  *200
00200   7200  BEGIN,   CLA          / BEGIN FINDS THE ONSET OF INSPIRATION.
00201   6560            6560
00202   3074            DCA MAG
00203   6561            6561
00204   1074            TAD MAG
00205   7510            SPA
00206   5200            JMP BEGIN
00207   7200  SURFACE,  CLA          / INTHIS SECTION THE WIDTH IS MEASURED.
00210   4020            JMS WAIT
00211   2075            ISZ T
00212   6560            6560
00213   3074            DCA MAG
00214   6561            6561
00215   1074            TAD MAG
00216   7500            SMA
00217   5207            JMP SURFACE
00220   7200            CLA
00221   1075            TAD T
00222   2100            ISZ AS
00223   3500            DCA I AS
00224   1104            TAD M5
00225   3124            DCA NUM5
00226   4020  PERIOD,   JMS WAIT     / IN THIS SECTION THE PERIOD IS MEASURED
00227   2075            ISZ T
00230   6560            6560
00231   3074            DCA MAG
00232   6561            6561
00233   1074            TAD MAG
00234   7510            SPA
00235   5226            JMP PERIOD
00236   7200            CLA
00237   1075            TAD T
00240   2076            ISZ TS
00241   3476            DCA I TS
00242   3075            DCA T
00243   2102            ISZ COUNT
00244   5207            JMP SURFACE
00245   7200            CLA
00246   1104            TAD M5
00247   3102            DCA COUNT
00250   1077            TAD MTS
00251   3076            DCA TS
00252   1101            TAD MA
00253   3100            DCA AS
00254   2100  ADD,      ISZ AS       / THIS PORTION FINDS THE SUM OF THE 1st
00255   2076            ISZ TS       / FIVE WIDTHSAND THE SUM OF THE 1st FIVE
00256   1500            TAD I AS     / PERIODS.
00257   1106            TAD SUMA
00260   3106            DCA SUMA
```

```
00261  1476          TAD ITS
00262  1107          TAD SUMT
00263  3107          DCA SUMT
00264  2102          ISZ COUNT
00265  5254          JMP ADD
00266  1106          TAD SUMA      /--------------------------------
00267  7041          CIA
00270  1105  DVIDE1, TAD MQ        / THIS PORTION FINDS THE AVG.OF
00271  2110          ISZ QUOT1     / THE WIDTHS.
00272  7510          SPA
00273  5270          JMP DVIDE1    /--------------------------------
00274  7200          CLA
00275  1107          TAD SUMT
00276  7041          CIA
00277  1105  DVIDE2, TAD MQ        / THIS PORTION FINDS THE AVG. OF
00300  2111          ISZ QUOT2     / THE PERIODS.
00301  7510          SPA
00302  5277          JMP DVIDE2
00303  7300          CLA CLL       / THIS PORTION CHECKS THE WIDTH FOR
00304  1110          TAD QUOT1     / 50% INCREASE.IF YES CHECKS THE PERIOD.
00305  7010          RAR
00306  1110          TAD QUOT1
00307  7041          CIA
00310  2100          ISZ AS
00311  1500          TAD I AS
00312  7500          SMA           /
00313  5374          JMP TEST      /--------------------------------
00314  1110          TAD QUOT1     / THIS PORTION CHECKS THE WIDTH
00315  7510          SPA           / FOR 50% DECREASE.IF YES IGNORES THE
00316  5351          JMP CONT      / CYCLE.-------------------------
00317  7300          CLA CLL
00320  1111          TAD QUOT2     / THIS PORTION CHECKS THE PERIOD
00321  7010          RAR           / FOR 50% INCREASE. IF YES IGNORES
00322  1111          TAD QUOT2     / THIS CYCLE.
00323  7041          CIA
00324  2076          ISZ TS
00325  1476          TAD I TS
00326  7500          SMA
00327  5351          JMP CONT      /--------------------------------
00330  7200  INITIAL, CLA          / THIS PORTION INITIALIZES THE PROGRAM
00331  1104          TAD M5        / FOR GOING THROUGH THE LOOP,AND SHIFTS
00332  3102          DCA COUNT     / THE STORED DATA 1 BYTE.
00333  1101          TAD MA
00334  3100          DCA AS
00335  1077          TAD MTS
00336  3076          DCA TS
00337  2076          ISZ TS
00340  2100          ISZ AS
```

```
00341  2114           ISZ C
00342  2113           ISZ BS
00343  1513           TAD I BS
00344  3500           DCA IAS
00345  1514           TAD IC
00346  3476           DCA ITS
00347  2102           ISZ COUNT
00350  5337           JMP .-11
00351  7200  CONT,     CLA          / THIS PORTION INITIALIZES THE PROGRAM
00352  1116           TAD M1       / WHEN THE LAST CYCLE OF RES PIRATION
00353  3102           DCA COUNT    / IS IGNORED.
00354  1115           TAD MC
00355  3114           DCA C
00356  1112           TAD MB
00357  3113           DCA BS
00360  1117           TAD M165
00361  3100           DCA AS
00362  1120           TAD M175
00363  3076           DCA TS
00364  3110           DCA QUOT1
00365  3111           DCA QUOT2
00366  3106           DCA SUMA
00367  3107           DCA SUMT
00370  6502           6502
00371  5207           JMP SURFACE
00372  7402           HLT
00373  5050           JMP START
00374  7300  TEST,     CLA CLL      / THIS PORTION CHECKS FOR 50%
00375  1111           TAD QUOT2    / INCREASE IN THE PERIOD.
00376  7010           RAR
00377  1111           TAD QUOT2
00400  7041           CIA
00401  2076           ISZ TS
00402  1476           TAD I TS
00403  7500           SMA
00404  5206           JMP PRINT
00405  5777'          JMP CONT
00406  7200  PRINT,    CLA          / THIS PORTION PRINTS "AIRWAY BLOCKED".
00407  6046           TLS
00410  7200           CLA
00411  1521           TAD I CODE
00412  7450           SNA
00413  4041           JMS BELL
00414  4043           JMS TYPE
00415  2121           ISZ CODE
00416  5210           JMP .-6
00577  0350
       $
00177  0200
```

ADD	0254
AS	0100
BEGIN	0200
BELL	0041
BS	0113
C	0114
CODE	0121
COT	0351
COUNT	0102
DVIDE1	0270
DVIDE2	0277
INITIA	0330
K207	0123
MA	0101
MAG	0074
MB	0112
MC	0115
MCODE	0122
MCOUNT	0103
MQ	0105
MTS	0077
MW	0126
MY	0130
M1	0116
M165	0117
M175	0120
M5	0104
NUM5	0124
PERIOD	0226
PRINT	0406
QUOT1	0110
QUOT2	0111
START	0050
SUMA	0106
SUMT	0107
SURFAC	0207
T	0075
TEST	0374
TS	0076
TYPE	0034
W	0125
WAIT	0020
Y	0127

In periphery to the computer 2 channels of the A/D convertor and two sense switches are used. In channel 0 of the A/D convertor, the respiration waveform from the monitor is plugged. The potentiometer connected to Channel 1 is used for setting the threshold. The settings are:

	Potentiometer Setting											
	1.92	2	2.1	2.2	2.3	2.4	3.38	3.6	3.8	4.04	4.4	4.5
Volts	-7.52	-6.75	-6.51	-6.38	-6.15	-5.42	-4	-3.5	-3	-2.5	-2	-1.5

The first sense switch, if down, discontinues the computer buzzer. The second switch, if down, halts the execution of the program. Then, if the continue switch is pressed, the program would be run from the start.

In order to run the program, the same switches should be up and the switch to the left of the register switches on the minicomputer should be down. The program starts at location 50.