

A FEASIBILITY STUDY FOR SELF-POWERED
ARTIFICIAL HEART SYSTEMS

by 1264

MARK EDWARD SHARP

B. S., Kansas State University, 1968

A THESIS

submitted in partial fulfillment of the
requirements for the degree

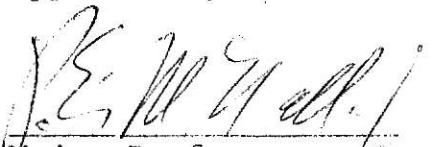
MASTER OF SCIENCE

Department of Mechanical Engineering

KANSAS STATE UNIVERSITY
Manhattan, Kansas

1970

Approved by:


Major Professor

LD
2668
T4
1970
55

TABLE OF CONTENTS

Chapter	Page
I. INTRODUCTION	1
1. Introduction	1
2. Objectives	4
II. LITERATURE REVIEW AND ANALYSIS	6
III. METHODS AND APPARATUS.	18
1. Apparatus.	18
2. Subjects	27
3. Procedure.	28
IV. RESULTS.	32
V. DISCUSSION	46
VI. SUMMARY AND CONCLUSIONS.	57
1. Conclusions.	57
2. Future Research.	59
LIST OF REFERENCES.	61
APPENDICES.	63
ACKNOWLEDGEMENTS.	68

LIST OF TABLES

Table	Page
1. Calculated Cardiac Pumping Power Output and Corresponding Measured Respiratory Power Output at Two Levels of Activity	33
2. Oxygen Uptake of Test Subjects at Different Threshold Pressure Resistances.	36
3. Mean Minute Volume of Four Test Subjects at Four Resistance Pressure Levels	42
4. Rank Correlation of Physical Condition and the Ratio <u>Respiratory Power</u> of Walking Test Subjects. . . Cardiac Power	44
5. Rank Correlation of Athletic History and the Ratio <u>Respiratory Power</u> of Walking Test Subjects. . . Cardiac Power	45
6. Physical Characteristics of Test Subjects	64

LIST OF FIGURES

Figure	Page
1. Diagram of Experimental Apparatus.	20
2. Sample Data Sheet.	23
3. Test Subject Performing Treadmill Walking Tests.	25
4. Test Subject Performing Sedentary Tests.	26
5. Graphical Comparison of Respiratory Power Output and Cardiac Power Output.	34
6. Oxygen Uptake vs. Pressure at the Sedentary Activity Level	38
7. Oxygen Uptake vs. Pressure at the Walking Activity Level	39
8. Oxygen Uptake vs. Respiratory Power Output at the Sedentary Activity Level.	40
9. Oxygen Uptake vs. Respiratory Power Output at the Walking Activity Level.	41
10. Schematic Wiring Diagram of Experimental Apparatus.	63
11. Cardiac Output vs. Oxygen Uptake from the Data of Reeves, et al., 1961	65

CHAPTER I

INTRODUCTION

One of the primary considerations in the development of an artificial heart system is the energy source to power the artificial heart pump. Considerable research is currently being carried out to develop a suitable, implantable energy source. The limitations of size, weight, and performance are dictated by the physiological requirements of the artificial heart patient. The complete artificial heart system should fit into the chest cavity which is occupied by the patient's natural heart.

Much of the current research in this area is with electrical power sources. An electric battery is a convenient means of storing electrical energy, but it does require frequent recharging or replacement. Size and weight limitations of an in vivo energy source also impose severe restrictions on the use of a storage battery for powering an artificial heart. Another energy source currently being studied by various researchers is a miniature atomic reactor. Heat developed from the nuclear energy operates a heat engine which, in turn, powers the artificial heart. This small reactor could provide energy for several years and greatly reduce the problem of frequent rechargings. However, a heat engine must receive heat energy at a high temperature and dissipate heat at the temperature of its environment. If an

engine were operating internally it would give off considerable heat to the body. The effect of this additional endogenous heat on the body is not fully understood and is the subject of considerable research interest at this time.

A problem basic to the above mentioned energy sources is their inherent inefficiency in converting energy from one form to another. The efficiency of an internal device should be as high as possible to reduce the amount of heat dissipated to the body tissue, and to reduce the power drain on the prime energy storage device and thus increase the life of each energy charge. The electrical, thermal, and nuclear energy systems also have another common problem. All require a periodic transfer of energy into the body to replenish the supply of internally stored energy. It would be desirable to provide an energy system which, once implanted in the body, required neither periodic maintenance nor an external energy source. This would increase the patient's mobility and make such a transplant more desirable from the patient's point of view.

One energy source which has received little attention is the body's own muscles. If an existing body muscle, or muscle system, could be utilized to power an artificial heart the problems of transmitting energy into the body would be eliminated. Preliminary studies indicated that it may be feasible to utilize the respiratory muscles to perform work of both breathing and pumping blood. Since the respiratory function is rhythmic and involuntary, it is conceivable that these motions could be utilized to

produce power for an artificial heart. An additional advantage could be realized by this system if the power supplied to an artificial heart were controlled directly by the rate and depth of the respiratory movements. Variations in the cardiac power output and in the breathing rate and depth are related, i.e., as one's activity level increases both respiratory activity and blood circulation are increased, and vice versa. Thus, such a system might provide a practical means of controlling the power delivered to an artificial heart.

The feasibility of such a system was examined in experimental research reported in this thesis. The scope is limited to feasibility only from the standpoint of work output capability of the respiratory muscles--the actual design of an implantable device was not considered in this research. Tests were conducted to determine if the respiratory muscles could produce sufficient power to operate an artificial heart and still maintain a normal ventilation. Margaria, et al. (1960) found that even during strenuous exercise the mechanical work of normal breathing is about 8% of the maximum capability as calculated from the static pressure-volume prediction of maximum work capability which was developed earlier by Rahn, et al. (1946). Craig, (1960) showed that the maximum work capability of a dynamic breathing cycle was about 70% of the predicted maximum, all of which indicates a reserve capacity for respiratory work even during maximum ventilation. The purpose of the present research was to examine this reserve power potential at two activity levels to assess the feasibility of providing power to an artificial heart.

Objectives

This thesis proposes that an artificial heart could be powered by energy derived from the continual movements of the respiratory muscles. The feasibility of such a system is examined solely on the basis of the power output capabilities of the respiratory muscles and the relationship between that power capability and the amount of pumping power exerted by a normal heart as one's activity level changes. The following specific questions are examined and answered according to the results of experiments conducted.

It is desired to first learn the continuous power output capability of the respiratory muscles at two normal ventilation rates for several individual subjects. The two ventilation rates are established by each individual's normal response to two activity levels. Secondly, the relationship between respiratory power output capability and blood pumping power output of the heart at each of two activity levels is to be determined.

Two secondary questions were also considered in this research. First, the oxygen cost of performing additional work with the respiratory muscles was determined. This has also been considered in previous work (McGregor and Becklake, 1961) but not at the magnitude of increased respiratory workload of the present research. Finally, it was expected that the ability of an individual to perform at the high respiratory workloads would be related to his degree of aerobic physical fitness. Therefore, test subjects were chosen with widely varying athletic histories

and degrees of physical fitness to facilitate determining this correlation.

CHAPTER II

LITERATURE REVIEW AND ANALYSIS

The first step in this research was the quantification of the power output requirements of an artificial heart. The pumping power output of the normal human heart was determined from the work of Robinson, et al. (1967). They calculated the blood pumping power output by analyzing the pressure-volume relationships of the ventricles of a normal heart during systole. They also developed a useful family of curves which shows the relationship between the cardiac index and the average cardiac power index (watts of pumping power/square meter of body surface area) for various ages of healthy men. The equation for calculating the cardiac power is given by Robinson, et al. (1967).

$$\text{Power} = 0.68V_c (P_L + P_R - 4) \times 2.22 \times 10^{-3} \text{watts} \quad (1)$$

where:

V_c = Cardiac Output (liters/minute)

P_L = Peak left ventricle pressure (mmHg)

P_R = Peak right ventricle pressure (mmHg)

There are several published reports in which researchers developed a linear regression equation to relate the cardiac output of test subjects to the subjects' rates of oxygen consumption:

(Granath, et al., 1964; Tabakin, et al., 1964). The accurate measurement of cardiac output calls for facilities and equipment not accessible to the author. It was judged that the best way to determine cardiac output of the test subjects was to combine the subjects' oxygen uptake with the results of previous research to calculate the cardiac output. The data provided by Reeves, et al. (1961) was analyzed by linear regression analysis (see Appendix C). The resulting linear equation (2) relates the cardiac output to the oxygen uptake of test subjects reported by Reeves, et al. (1961).

$$\text{Cardiac Output} = 3.4 + 6.1 (\text{Vo}_2) \quad (2)$$

where:

Vo_2 = Oxygen uptake in liters per minute

Cardiac output measured in liters per minute

Several investigators (Granath, et al. 1964; Bevegard, 1960; McGregor and Becklake, 1961) have shown that the cardiac output varies according to body position. The work by Reeves, et al. (1961) concerned test subjects in the same body position and activity as those of the present study (treadmill walking) and their results are probably more applicable to the present study than those of the other researchers mentioned above.

The cardiac output of test subjects, as determined by use of equation (2), was used in conjunction with the above-mentioned

work of Robinson, et al. (1967). After converting the cardiac output of each test subject to the units of the cardiac index (liters/minute/meter²), the corresponding cardiac pumping power index was read directly from the graph given by Robinson, et al. (1967). Although this method may give significant errors for the cardiac power output of an individual test subject, it is based on the mean values of data taken from a large number of test subjects and should provide the most probable values for normal humans chosen at random.

The second aspect of this research was the study of published literature to determine the reserve power output capability of the respiratory muscles of normal human subjects and to compare it with the heart power output. The work output of the respiratory muscles has been examined by several previous investigators and some important findings are described below.

The amount of work performed by the respiratory muscles divided by the time elapsed in its performance is the work rate, or power output of the respiratory system. The term "Respiratory power output" will be used as the total power output of the respiratory system reduced by that required for normal breathing. Thus, respiratory power output represents a reserve power capability which is available to do work other than performing the normal breathing function.

Previous investigators have used various techniques to cause the respiratory muscles to expend more power than is required for normal breathing. A very effective method is that of adding an air flow restriction to the airway. When the subject inhales

through such an air flow restriction, he must reduce the pressure in his airway by enough to cause sufficient air flow through the restriction and likewise, to exhale, he must increase his airway pressure above that normally required. The resulting pressure fluctuations are greater than those incurred during normal breathing, and the respiratory muscles must expend additional power to maintain an adequate ventilation.

Perhaps the earliest significant contribution to the study of respiratory work is by Rahn, et al. (1946). They noted that the amount of work performed by the respiratory muscles during a single breath may be determined by a graphical integration procedure as follows. If the instantaneous pressure and volume of the lungs are recorded simultaneously on separate axes of a single graph, the resulting points form a closed curve for each breathing cycle. The graphical area enclosed by that curve is equal to the total amount of work performed by the respiratory muscles. The theory of this method is fundamental in classical thermodynamics and its application to respiratory work is more rigorously developed in a following portion of this chapter. Several investigators have measured respiratory work output by the method outlined by Rahn, et al. (1946).

McGregor and Becklake (1961) determined the oxygen cost of performing respiratory work. They measured the oxygen consumption at rest for normal breathing and during resistance breathing of varying degrees of severity. The oxygen cost of performing respiratory work was estimated by relating the increased oxygen consumption to the increased respiratory work output of resistance

breathing. They showed that the change in oxygen consumption with increased respiratory work due to resistance breathing was given by the following regression equation:

$$V_{O_2} = 26.52(\text{work}) + 27.5 \quad (3)$$

where:

Work = change in respiratory work in kilogram-meters/minute

V_{O_2} = change in oxygen uptake in cubic centimeters/minute

McGregor and Becklake (1961) observed values of respiratory work output up to only 1.3 watts during resistance breathing. The pumping power output of an average subject's heart is greater than 1.3 watts for nearly any activity level other than relaxing or sleeping: (Robinson, et al. 1967). The present research was designed to investigate the respiratory work output at values up to 4 or 5 watts, the approximate heart pumping power output during moderate activity such as walking. No values for the oxygen cost of respiratory work during near maximum breathing efforts were found in the literature. Craig (1960) examined the maximum respiratory work output of one breathing cycle in humans. Ten healthy male adults breathed through a small tube with maximum breathing effort. The pressure-volume diagram of the respiratory system was recorded during these maximal efforts and the respiratory work output of each subject was determined by graphically integrating the pressure-volume record. Craig (1960)

reports that the average work output of a maximal effort, dynamic breathing cycle is about 69% during expiration, and 77% during inspiration of the maximum respiratory work capability as predicted by a method presented by Rahn, et al. (1946). The mean respiratory power output of maximal effort breathing cycle (as calculated from Craig's data) is 2.8 watts.

Since the work of Craig (1960) was intended to be a study of maximum respiratory work output of one breathing cycle, the rate of muscular movement was necessarily quite slow to permit the respiratory muscles time to develop maximum force. Craig reports that the mean time for completion of one breathing cycle (inspiration + expiration) is 25.7 seconds, a period much longer than that of most persons' normal breathing. Agostini and Fenn (1960), using subjects trained in respiratory tests, have examined the maximal respiratory efforts in terms of the amount of work done by the respiratory muscles and of the time required in performing the work. They found that the maximum work done by the respiratory muscles in one complete inspiration and expiration increases as the air flow resistance to breathing is increased and the velocity of muscle shortening is decreased. According to Agostini and Fenn (1960), the respiratory work output of a single maximal effort breathing cycle decreases linearly with the increasing inverse of the time required for the maneuver. A complete graph or table of their results was not included in the published paper. They did include a graph which shows the work done by respiratory muscles during forced maximum inspiration against different resistances plotted as a function of the

inverse of inspiration time. The graph shows that the maximum respiratory work output during inspiration decreases linearly from about 4.2 kilogram-meters at $(1/\text{time}) = 0$ to 1.0 kilogram-meters at $(1/\text{time}) = 1.1$. If an individual were breathing deeply at a normal rate, about 16 breaths per minute, his approximate, maximum respiratory power output capability during inspiration alone is about $6\frac{1}{2}$ watts. Craig (1960), and Agostini and Fenn (1960) indicated that the respiratory work capability was characteristically greater during expiration than during inspiration. On that basis the maximum respiratory power output during expiration is at least $6\frac{1}{2}$ watts for a total respiratory power output capability of at least 13 watts. One should bear in mind that this maximum power output represents a maximum muscular effort and as such, could be maintained for only a short time. Since 13 watts is considerably more than the desired continuous power output of 4-5 watts, it is conceivable that the human respiratory muscles possess reserve power potential of sufficient magnitude to power an artificial heart.

Gee, et al. (1968) studied the effects of viscous resistance to breathing on the work capacity of university age males. They tested subjects at high workloads and placed a viscous resistance in the inspiratory and expiratory airways both separately and simultaneously. Exercise was performed in a seated position on a bicycle ergometer up to workloads approaching maximal oxygen consumption. The degree of airway obstruction was evaluated in terms of the amount by which each obstruction reduced the

subjects' maximum breathing capacity. Gee, et al. (1968) showed that "healthy young men can perform severe work requiring more than 80% of maximum oxygen consumption in the presence of an external airway obstruction that causes a 30% reduction in maximum breathing capacity." This work level was performed without complaints of dyspnea. The maximum mouth pressure for the airway obstruction causing a 30% reduction of maximum breathing capacity was about ± 30 cm H₂O (.43psi) relative to atmospheric pressure. In addition Gee, et al. (1968) reports no significant change in Vco₂, Vo₂, or heart rate due to the pressure breathing. He did note that the most severe airway obstruction caused slight hypoventilation at the highest workloads. Although the present research utilized a maximum pressure greater than 30 cm H₂O, the maximum workload was considerably less (about 50% of maximum oxygen consumption) than that employed by Gee, et al. (1968). Since they indicated there were no feelings of respiratory distress at the higher workloads, it seemed probable that at lower workloads the pressure could be increased to greater maximums without dyspnea.

The above-mentioned studies (Gee, et al., 1968; Craig, 1960; Agostini and Fenn, 1960) utilized a viscous resistance to obtain an airway obstruction. A viscous resistance provides a pressure gradient which is decreasing in the direction of the air flow and which changes as the air flow rate is changed. Another type of resistance, a threshold resistance, is sometimes used in pulmonary research. Threshold resistance to air flow is characterized by a constant pressure difference regardless of the air flow

rate. The theory used in calculating the respiratory work output is the same regardless of the type of air flow resistance used. Continuous instantaneous pressure and flow rate recordings of the air in the mouth or mouthpiece were used to quantify the respiratory work output. Work per breath was calculated by integrating the product of incremental pressure and incremental change in volume during an inspiration and during an expiration, the duration of one breath. This principle is basic to classical thermodynamic theory. Its application in this study of respiratory work is explained in the following paragraphs.

Van Wylen and Sonntag (1965), page 62, provide a concise explanation of the method for finding the work done on a system of air confined within a variable volume container, in this instance the lungs. Their simplified system consisted of a cylinder and piston. The amount of work done by the piston as it moved was determined for the theoretical case with various pressure conditions imposed on the air in the cylinder. They showed that the amount of work done during a single movement of the piston is given by the following equation.

$$\text{Work} = \int P dV \quad (4)$$

The piston and cylinder arrangement is a simple mechanical analogy for the human respiratory system. During normal inspiration, the pressure within the lungs is less than atmospheric pressure and the lung volume is increasing. During expiration, the pressure is greater than atmospheric and the volume is

decreasing. The work is negative in each case since (P) and (dV) are of opposite sign. Following the sign convention established for equation (4) this is interpreted to mean that work is being done on the air in the lungs and airways during normal inspiration and expiration. The work per breath is calculated by summing the work done during inspiration and that done during expiration.

In normal breathing both the lung air pressure and the flow rate of air varies over a range of values making an exact mathematical integration almost impossible. Three techniques have been used to evaluate the integral. Fletcher and Bellville (1966) developed a system in which the integration is performed electronically. The pulmonary pressure and the air flow rate were sensed electronically and recorded on magnetic tape. Later, the taped data was fed into an analog computer which provided the desired mathematical functions and printed the calculated results.

Another technique uses simpler and more readily available equipment. Craig (1960) described a method of simultaneously recording the pressure of the lungs and the volume of the lungs on the same graph. The resulting curve was integrated graphically to provide the desired work calculation. When many subjects are studied considerable data must be evaluated and this method becomes quite tedious.

Although it limits the type of resistance that may be used to increase the respiratory work, a threshold resistance apparatus

described by Campbell, et al. (1957) was used in the present research. Campbell showed that a threshold resistance has two distinct advantages over a viscous resistance. First, the pattern of breathing through the resistance is the same as that at an equivalent ventilation without resistance. The only change in mechanical work done by the respiratory muscles during resistance breathing was due directly to the added resistance. In addition, the pressure drop across the resistance is nearly independent of the flow rate, which simplifies the calculation of the respiratory work output as is shown below. The variable P of equation (4) becomes a constant and may be placed outside the integral.

$$\text{work} = P \int_{V_1}^{V_2} dV = P \int_{t_1}^{t_2} \frac{dV(t)}{dt} (dt) \quad (5)$$

Since only the quantity dV is to be integrated, the average $\frac{dV(t)}{dt}$, the minute volume of air breathed, may be used in place of the instantaneous value $dV(t)$. The minute volume is also a constant for a given activity level so it too may be removed from within the integral.

$$\begin{aligned} \text{work} &= P \left[\frac{dV(t)}{dt} \right] \int_{t_1}^{t_2} (dt) \\ &= P \left[\frac{dV(t)}{dt} \right] (t_2 - t_1) \end{aligned} \quad (6)$$

The amount of respiratory work performed against a constant

threshold resistance is simply the product of minute volume, pressure, and the elapsed time. At this point it should be noted that power is simply the time rate of doing work. The above expression for respiratory work output, equation (6), may be modified to determine respiratory power output as follows:

$$\text{Power} = \frac{\text{work}}{t_2 - t_1} = P \left(\frac{dV(t)}{dt} \right) \quad (7)$$

The respiratory power output is simply the product of threshold pressure and minute volume. The method presented is a convenient and simple means of measuring one's respiratory power output.

CHAPTER III

METHODS AND APPARATUS

As was stated earlier, the results desired of the present research were the maximum work output capability of the human respiratory muscles and the oxygen cost of various levels of respiratory work output. Also to be determined were the cardiac power output and the relationship between cardiac power output and maximum respiratory power output at two different physical exercise levels. Cardiac pumping power output was estimated by the method explained in Chapter II, relating cardiac power output to oxygen consumption rate. Respiratory power output was determined by multiplying the minute volume and the threshold resistance pressure, a procedure also explained in depth in Chapter II. The experimental apparatus used in the present research permitted measurement of three physiological parameters; oxygen consumption rate, respiratory minute volume, and air pressure in the mouth during resistance breathing.

An experimental method of measuring respiratory work output and the oxygen cost of respiratory work is described by Campbell, et al. (1957). They presented a description of the apparatus used to obtain a threshold pressure resistance to breathing. The threshold resistance has two inherent advantages over viscous resistance to breathing for respiratory work output studies. The Calculations for determining work output are greatly simplified

by use of a constant threshold pressure restriction, and also the pattern of breathing with a threshold resistance is similar to the pattern of normal breathing. An apparatus similar to that described by Campbell, et al. (1957) was constructed for use in the present research. The threshold pressure resistance was developed in each of two large, steel barrels, 14 inches in diameter and 27 inches high. Figure 1 is a schematic diagram of the complete resistance breathing system. Brief specifications of the experimental apparatus components are included in Appendix E. As shown in Figure 1, the inspiratory resistance and expiratory resistance were each provided by a separate barrel. Two rigid metal tubes, $1\frac{1}{2}$ inches in diameter, were fitted through the top of each barrel and the joints were brazed to form an air-tight seal. One long tube in each barrel extended nearly to the bottom of the barrel and the short tube extended only about 1 inch through the top of the barrel. Both barrels were partially filled with water so that one tube opening was underwater and one tube opened above the water in each barrel. The expiratory and inspiratory airway hoses were connected to the barrels in such a way that air entered each barrel through the long tube, bubbled up through the water, and left the barrel via the short tube. Before air can flow through the underwater tube and bubble to the water surface, a sufficiently large pressure difference must be established between the air in the submerged tube and the air above the water. This pressure difference is determined largely by the height of water above the submerged tube opening and was relatively constant for air flow

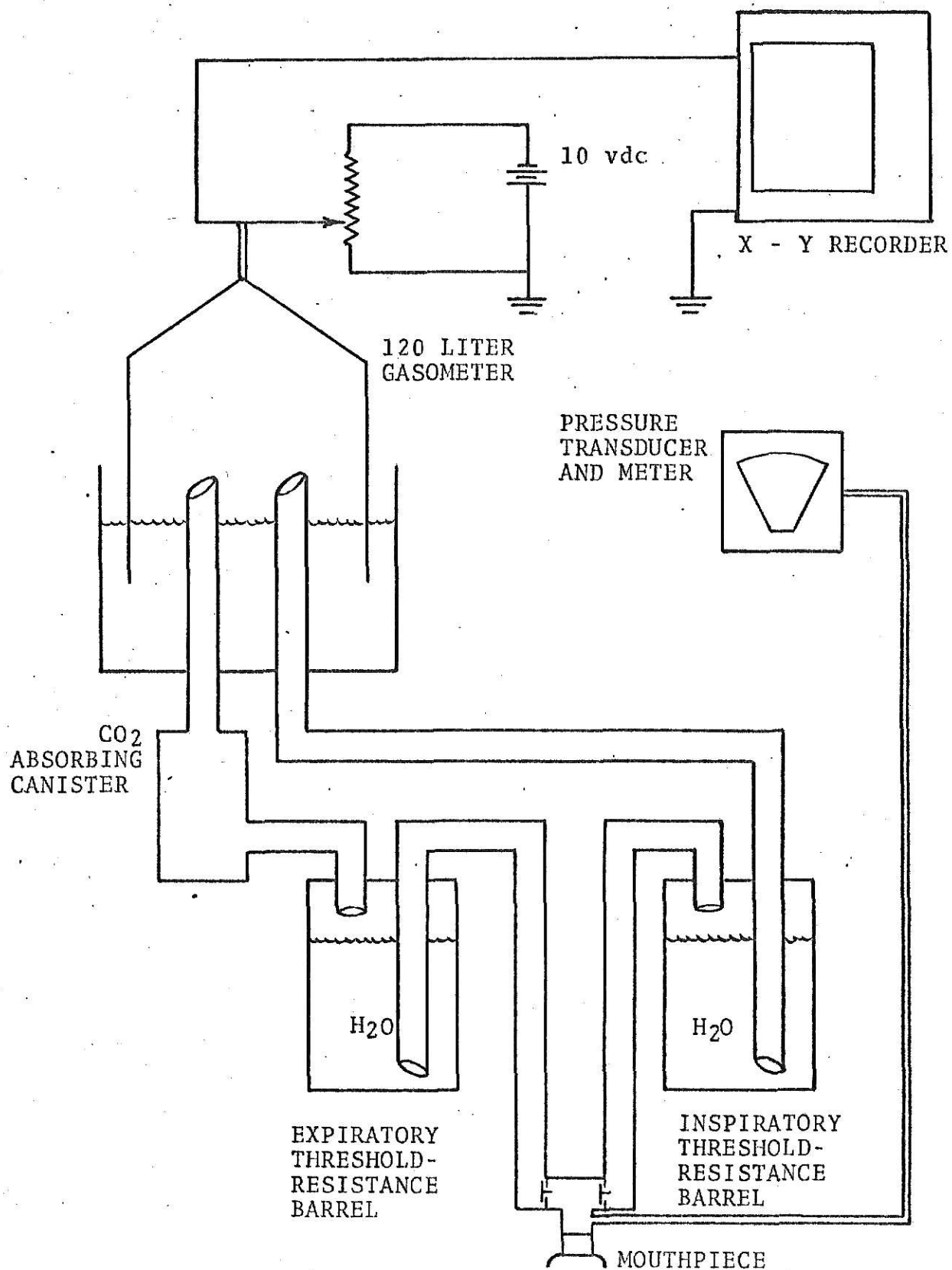


FIGURE 1
DIAGRAM OF EXPERIMENTAL APPARATUS

rates developed in breathing. By varying the amount of water in the barrels, differential threshold pressures up to 50 cm H₂O (.71 psi) were available to the researchers.

The spirometer indicated in Figure I was a chain compensated 120 liter gasometer. It was used to measure the changes in air volumes during minute volume measurements and during oxygen consumption tests. The 120 liter capacity was large enough to permit the use of room air in closed circuit spirometry oxygen consumption tests and still maintain the oxygen concentration above 19% in the air being breathed. (Best and Taylor, 1961)

A two-way breathing valve was connected to the mouthpiece to separate the inspiratory and expiratory air flows. With proper connection of the hoses and valves, the subject was obliged to breathe through any one of four air circuits--a closed circuit, with or without the resistances, for oxygen consumption tests; or an open circuit, with or without resistance, in which the subject inspired air from the gasometer and expired into the room. The open circuit test was used to measure the subjects respiratory minute volume by determining the average time rate at which air was removed from the gasometer.

The volume of the gasometer was recorded by using a constant voltage DC power supply, a linear output potentiometer, and an X-Y electronic recorder. A schematic wiring diagram is provided in Appendix A to show how the components were connected. The potentiometer was attached to the gasometer in such a way that vertical movements of the gasometer bell rotated the potentiometer

in a proportionate amount. With a constant voltage of 10 volts placed across the potentiometer resistance, rotation of the potentiometer produced a varying voltage at the "wiper" terminal. The varying wiper voltage was fed into the vertical axis input of the X-Y plotter pen. By varying the sensitivity of the recorder, it was calibrated to read either in (ten liters)/inch or in (liters)/inch scale factors. The vertical axis was used to record the gasometer volume changes while the horizontal axis of the X-Y recorder was driven by a time base internal to the X-Y recorder. The rate of horizontal recorder pen movement was set at a constant 0.02 inches/second for all tests. With the vertical axis sensitivity set at one liter per inch the slope of the closed-circuit spirometry graph was well defined for the activity levels studied in this research, i.e., oxygen consumption rates of from 0.4 liters/minute up to about $2\frac{1}{2}$ liters/minute. During respiratory minute volume measurements, the vertical pen movement was calibrated at 10 liters per inch. As the subject inspired air from the gasometer the recorder graph moved downward in a "stairstep" fashion. The average slope of the "stairstep" graph was the average rate at which air was removed from the gasometer, or the respiratory minute volume when expressed in liters per second. A typical data sheet is shown in Figure 2. All of the data of a single test was recorded on one graph as shown. The X-Y plotter was equipped with a detented calibration knob to facilitate rapid switching from one scale factor to another and the starting points were manually reset for the various phases of each test.

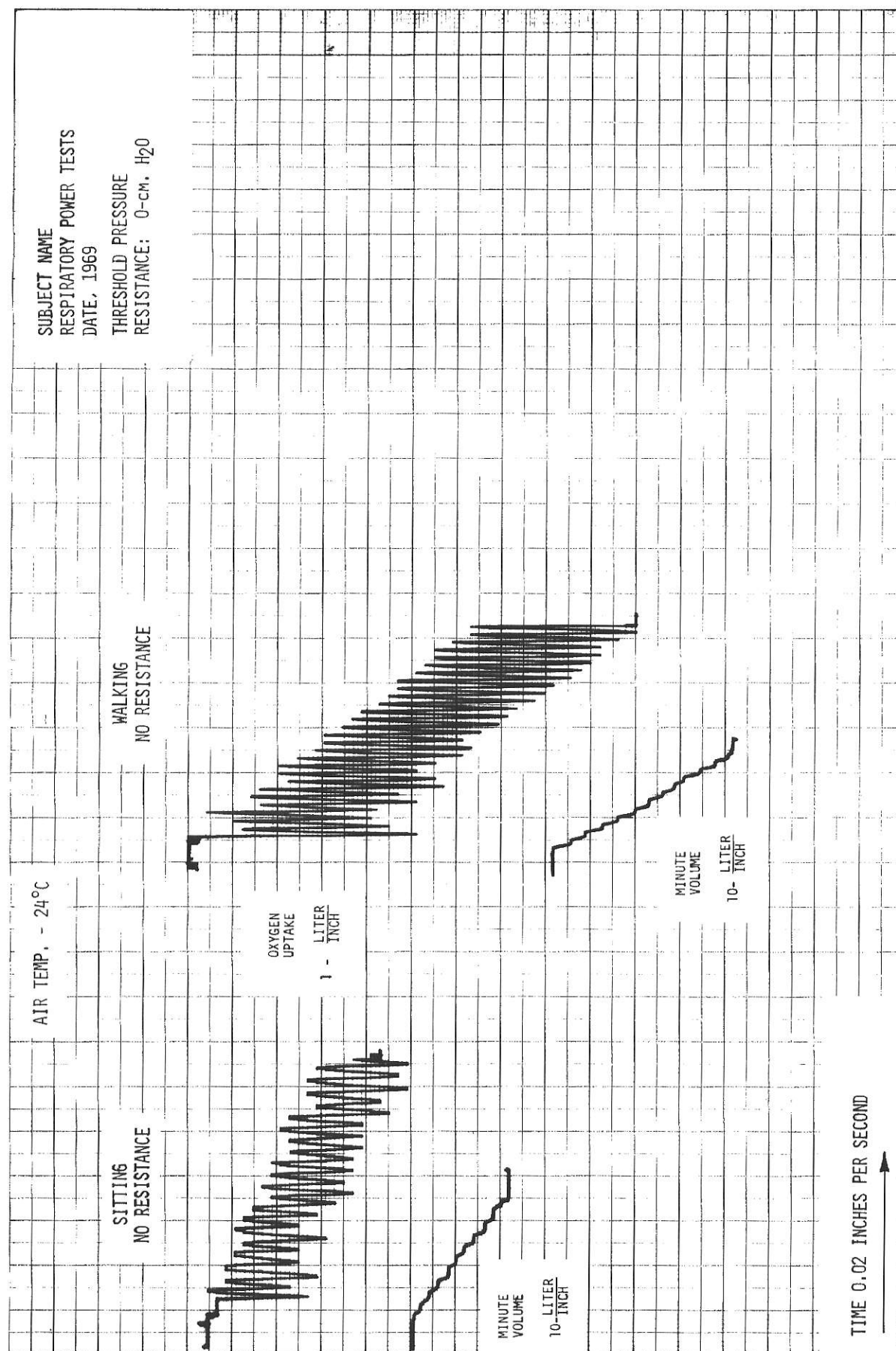


FIGURE 2
SAMPLE DATA SHEET

The pressure of the air being breathed was sensed with an electronic pressure transducer. A small tube connected the mouthpiece opening to the pressure transducer, which was electrically connected to a pressure indicating meter and signal amplifier. The X-Y recorder mentioned earlier was then used to record the mouth pressure as a function of time while the test subject breathed. It was noted that small rapid pressure fluctuations were present and were probably a result of the method used to develop the threshold pressure, bubbling air through water. The peak-to-peak pressure fluctuations were less than 10% of the approximate mean threshold pressure and were not believed to have a significant detrimental effect on the validity of the data. The threshold pressure used in calculating the respiratory work output was evaluated for each pressure level by visually inspecting this graph of mouth pressures as a function of time, and establishing a mean value by eye.

The complete system described above is shown in use by a test subject in Figures 3 and 4. The apparatus for treadmill walking is shown in Figure 3. In the foreground are the threshold resistance barrels which produced the expiratory and inspiratory resistance pressures. Reinforced plastic tubes connect the barrels to the mouthpiece and the gasometer. The carbon dioxide absorbing canister is shown hanging on the gasometer. Baralyme was used to absorb the exhaled carbon dioxide and its indicating change of color was easily visible through the clear plastic canister. The X-Y recorder and other electrical equipment was placed on a table within convenient access for the equipment

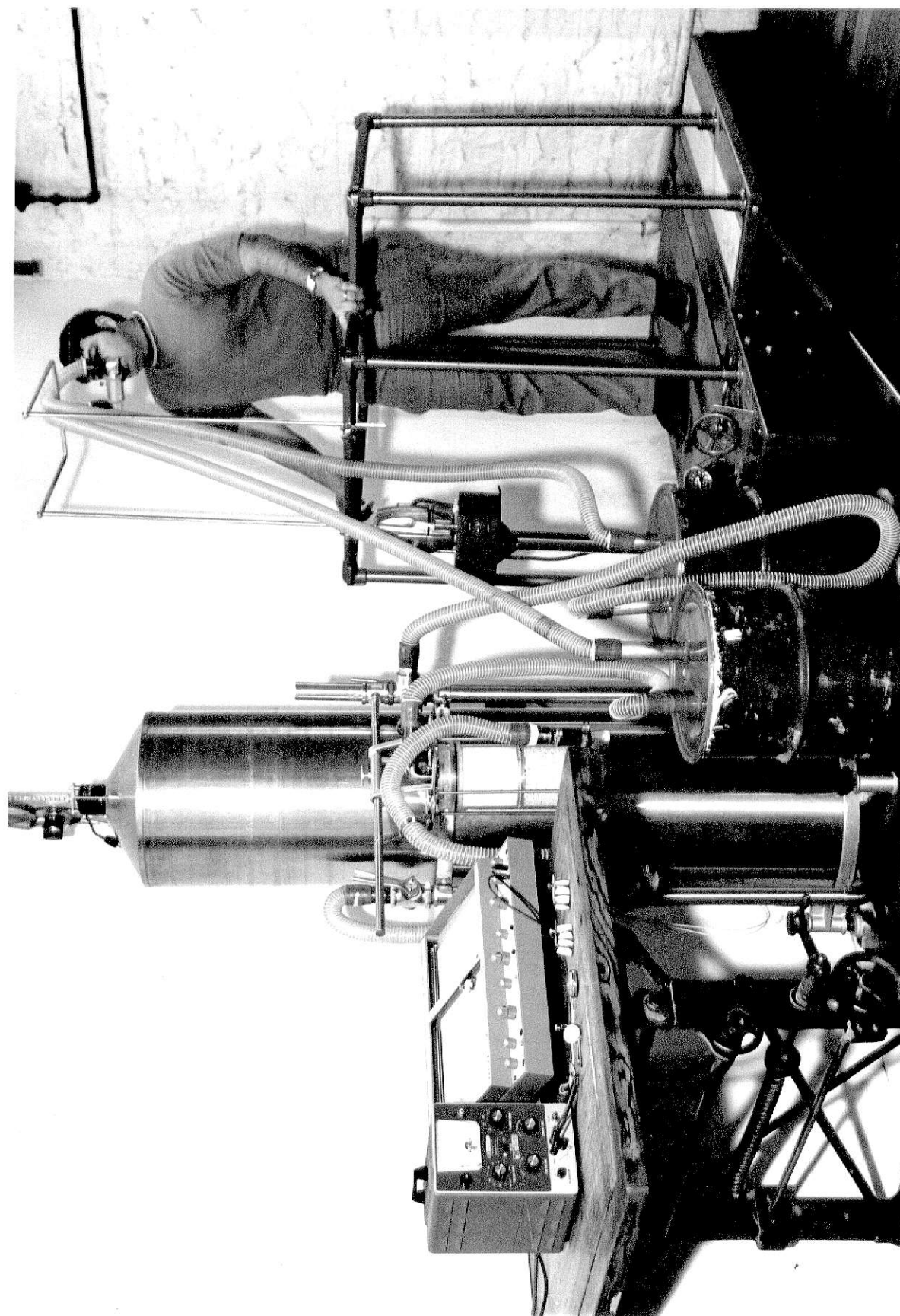


FIGURE 3
Test Subject Performing Treadmill-Walking Tests



FIGURE 4
Test Subject Performing Sedentary Tests

operator. Test subjects entered the treadmill from the platform shown on the right side of Figure 3. Once on the platform they began breathing through the mouthpiece, stepped onto the moving treadmill belt, and proceeded with the test.

A typical test subject in the sedentary test position is shown in Figure 4. The seated tests were conducted in the same general manner as the treadmill walking tests except that the test subjects were seated for a few minutes before beginning to breathe in the mouthpiece to help assure that their body systems were in steady state for the sedentary activity level.

Subjects

Nine healthy young males, ages 20 to 40, (mean 25.6) served as test subjects in the present research (see Appendix B). Two subjects were trained distance runners, one of which was an Olympic team member. The other subjects were not in this high state of physical condition. Most were students and office workers, some of whom were doing daily jogging exercises. It is admitted that these subjects are probably not typical of that portion of the human population which might require the use of an artificial heart. They were younger and physically more active than most heart patients.

Before each test subject participated in the test series, he was familiarized with the testing apparatus, methods, purpose of the research, and his individual contribution to the findings as a whole. Each was assured that he was free to discontinue or

postpone a portion of the test at any time he so desired.

Procedure

In keeping with the intent of this present research, to determine the maximum respiratory work capability of human test subjects, all test subjects began the tests at low respiratory work levels and progressed to higher work levels in subsequent tests. This procedure permitted the effects of learning or conditioning to tend to increase the ability of the test subjects to perform respiratory work. The time interval between successive tests on each test subject was at least one week. Each test subject was scheduled for testing at a time convenient to the subject. Because of conflicting time schedules, no attempt was made to conduct all tests at the same time of day.

Each test subject reported to the experimental facility at his appointed time dressed in casual street clothes. At that time he was furnished a sterilized rubber mouthpiece and instructed on the use of the mouthpiece. The first tests were conducted with the subjects walking on a treadmill at 3 miles per hour and on a 6% grade. The data was taken to provide respiratory minute volume and oxygen consumption rate determinations for breathing both normally and then with a threshold pressure resistance of 25 cm H₂O (.35 psig) added to both expiration and inspiration. This test required about 25 minutes to complete and was easily accomplished by most of the test subjects. Immediately following completion of the test, each test subject expressed comments and

answered a few oral subjective questions to aid the researchers in evaluating the test subjects' reactions to the respiratory work tests. Each test subject also completed the questionnaire shown in Appendix D which provided information concerning his athletic history. Finally each test subject was asked if he thought that he could breathe against a greater threshold pressure; and, if so, he was asked to return for further tests during the following week. Those subjects expressing a desire to participate in tests at higher respiratory work output levels were scheduled to perform the second test at least one week after the first test.

Before continuing in the test series, the test subjects were ranked, according to information they provided on the questionnaires, into two numerical rankings--one which ordered their total years of participation in varsity athletics as high school and university students, and another which ordered their present degree of physical conditioning. These rankings were used at the completion of all the tests to show the correlation between the subjects' rankings and their maximum respiratory work output capability.

The second phase of the testing included two different exercise levels; a relaxed seated position, and again treadmill walking at 3 miles per hour on a 6% grade. The threshold pressure resistance was increased to 35 cm H₂O ($\frac{1}{2}$ psig) on both the inspiration and expiration airways. The subjects again reported for testing at their individually appointed times dressed in casual street clothes. The first portion of this test phase consisted

of minute volume and oxygen consumption determinations for the subject seated and breathing against the threshold pressure resistance. On completion of the sedentary test the subject began walking on the treadmill while continuing to breathe against the threshold pressure during both inspiration and expiration. After allowing 3 minutes for the subject's body to reach equilibrium at the walking exercise level, respiratory minute volume and oxygen consumption data were taken to complete the second test phase. After the subject had relaxed for a few minutes following the tests, he was again questioned to evaluate his subjective reactions to the test situation. Those test subjects who successfully completed the second phase of this test were requested to return for a third time during the following week for tests at still higher respiratory workloads. The appointments were set for one week after the second phase for each individual test subject.

The third phase of testing was very similar to the second phase but with two changes. The minute volume and oxygen consumption were measured in the sitting position both with and without the threshold resistance in the airway. The other change was an increase in the threshold pressure resistance to 46 cm H₂O (0.66psig) in both the inspiratory and expiratory airways. When the test subjects reported for testing, they were furnished a sterilized mouthpiece and were given a chance to casually breathe through the threshold resistance to "get the feel" of it prior to the actual test. This step was included because most

subjects were quite surprised by the magnitude of the threshold pressure at the level of 46 cm H₂O (0.66psig). It was anticipated that without prior exposure to the high resistance some subjects would have to interrupt the test procedure when the threshold pressure was connected to the airways.

Minute volume and oxygen consumption were determined for the sitting position both with and without the threshold resistances connected in the airways. As in the second phase the subjects began walking on the treadmill and continued to breathe through the threshold resistance when they had completed the sedentary portion of the experiment. After allowing the subjects to walk for three minutes, their respective respiratory minute volumes and oxygen consumption rates were recorded to complete the tests. After this final phase of the tests, the subjects were questioned concerning their reactions to the high-resistance breathing and their ability to withstand the threshold pressure and ventilate sufficiently.

CHAPTER IV

RESULTS

The results of the present research include the development of several relationships between physiological parameters. The primary information is the relationship between maximum respiratory power output and the cardiac pumping power output of test subjects at two levels of activity. This relationship shows how well the respiratory muscles can produce power of magnitude comparable to the power requirements of an artificial heart. Cardiac power output was calculated for each subject at both activity levels, based on the method developed in Chapter II, which utilized the oxygen uptake rate in conjunction with the works of Robinson, et al. (1967) and Reeves, et al. (1961). The results of this comparison of respiratory and cardiac power output are shown in Table 1 for both the sedentary and the treadmill walking tests. Figure 5 presents the data of Table 1 graphically.

TABLE 1

CALCULATED CARDIAC PUMPING POWER OUTPUT AND CORRESPONDING
MEASURED RESPIRATORY POWER OUTPUT AT TWO LEVELS OF ACTIVITY

Sub- ject	Respiratory Power (watts)		Cardiac Power (watts)		Ratio - <u>Respiratory Power</u> <u>Cardiac Power</u>	
	Sitting	Walking	Sitting	Walking	Sitting	Walking
1	1.8	4.6	1.3	3.7	1.37	1.24
2	4.2	4.1	1.5	4.0	2.80	1.02
*3	-	4.0	-	4.4	-	.91
4	3.5	4.1	2.1	6.2	1.67	.66
5	1.6	3.4	2.1	4.9	.76	.69
6	3.5	3.8	1.4	3.3	2.50	1.15
*7	-	3.2	-	4.2	-	.76
8	1.1	2.9	1.3	4.3	.85	.67
9	1.6	6.1	1.4	5.1	1.14	1.20
mean	2.5	4.0	1.6	4.5	1.58	.92

*Subject did not perform a sedentary activity respiratory power test.

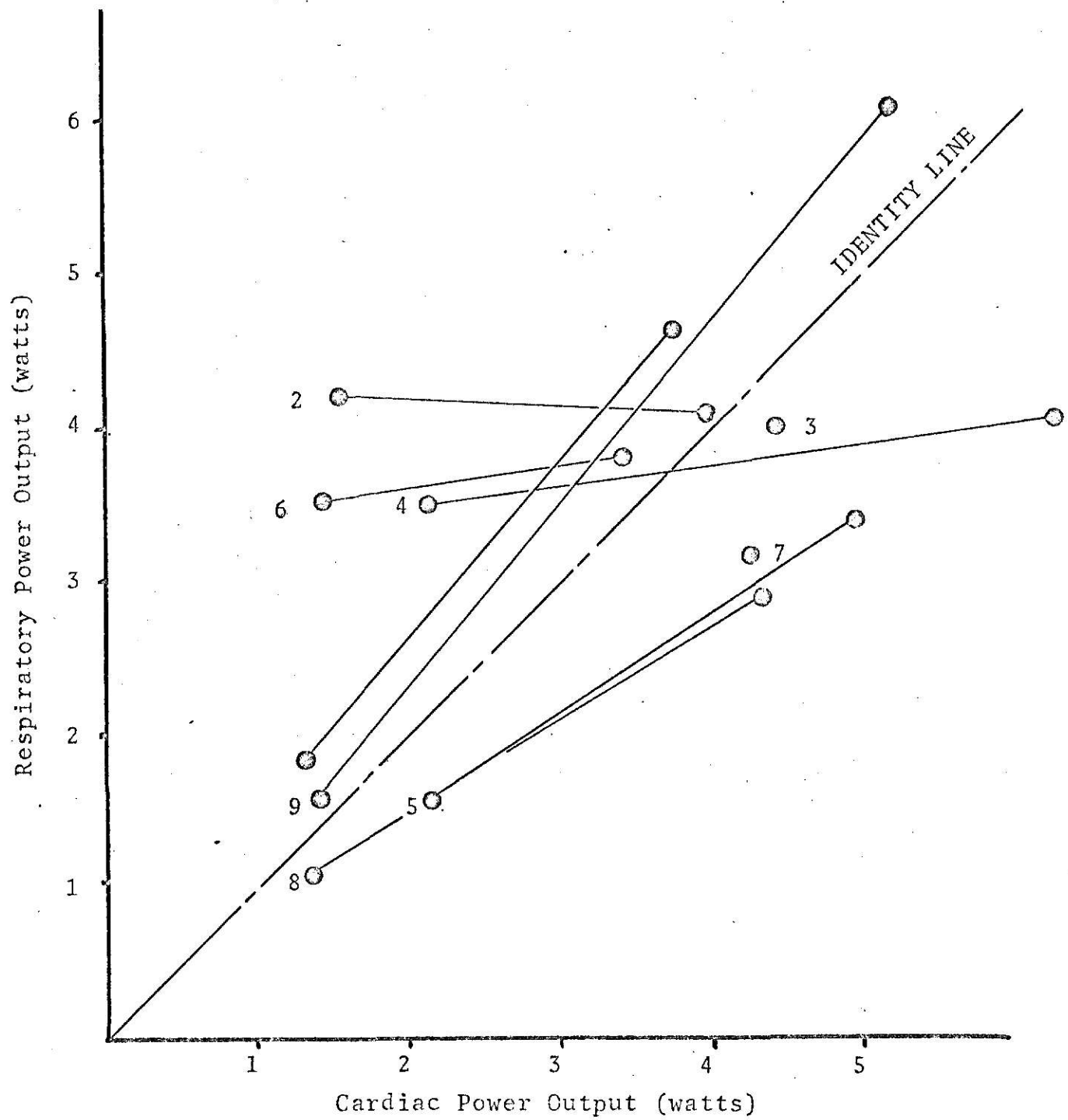


FIGURE 5

GRAPHICAL COMPARISON OF RESPIRATORY POWER OUTPUT
AND CARDIAC POWER OUTPUT

The oxygen consumption rate of test subjects during continuous threshold pressure resistance breathing was evaluated at four different pressure resistance levels; 0, 25, 35, and 46 cm H₂O (0, .36, .50, and .66 psig). Table 2 shows the oxygen uptake and respiratory power output of each test subject at the four levels of pressure resistance. Figures 6, 7, 8, and 9 show graphically the data of Table 2. The linear regression lines indicate that oxygen uptake increases as respiratory work is increased or as the resistance pressure is increased. In Figure 7 the points are widely scattered and the corresponding correlation coefficient, $r = .32$, is not statistically significant. The correlation coefficients of Figures 6, 8, and 9 are, however, significantly different from zero at the 95% confidence level. All of the values of oxygen volume are corrected to STPD (Standard Temperature, Pressure, Dry) conditions.

TABLE 2

OXYGEN UPTAKE OF TEST SUBJECTS AT DIFFERENT
THRESHOLD PRESSURE RESISTANCES

Pressure (cm H ₂ O)	Subject	Respiratory Power Output (watts)		Oxygen Uptake (liters/minute)	
		Sitting	Walking	Sitting	Walking
0	1	0	0	.34	1.27
0	2	0	0	A	1.54
0	3	0	0	A	2.24
0	4	0	0	A	1.64
0	5	0	0	.34	1.20
0	6	0	0	.33	1.47
0	7	0	0	A	1.52
0	8	0	0	A	1.49
0	9	0	0	.36	1.98
25	1	A	A	A	A
25	2	A	4.10	A	1.68
25	3	A	4.00	A	2.00*
25	4	A	4.10	A	2.92
25	5	A	2.51	A	1.68
25	6	A	2.50	A	1.64
25	7	A	3.18	A	1.91*
25	8	A	2.90	A	1.75
25	9	A	4.64	A	1.85
35	1	1.78	3.23	.53	1.54
35	2	3.70	6.10	B	B
35	3	C	C	C	C
35	4	3.47	4.50	.80	B

TABLE 2
(continued)

Pressure (cm H ₂ O)	Subject	Respiratory Power Output (watts)		Oxygen Uptake (liters/minute)	
		<u>Sitting</u>	<u>Walking</u>	<u>Sitting</u>	<u>Walking</u>
35	5	A	A	A	A
35	6	3.47	3.23	.58	1.69
35	7	C	C	C	C
35	8	1.10	C	.47	C
35	9	1.50	5.29	.76	2.69
46	1	1.52	4.60	.50	1.53
46	2	4.20	C	.69	C
46	3	C	C	C	C
46	4	C	C	C	C
46	5	1.58	3.40	.88	2.21
46	6	3.00	3.80	1.33	1.38
46	7	C	C	C	C
46	8	C	C	C	C
46	9	1.59	6.10	.69	2.28

*The indicated values of oxygen uptake were estimated for use in Table 1 and Figure 5. The estimated values were determined from the regression equation of Figure 9 (Oxygen Uptake vs. Respiratory Power). The test subjects in these cases performed the resistance breathing satisfactorily but due to equipment malfunctions their original test results were invalid.

A--Positions marked 'A' indicate no test was conducted.

B--Positions marked 'B' indicate the test data was invalid

C--Positions marked 'C' indicate the subject was unable to complete the test.

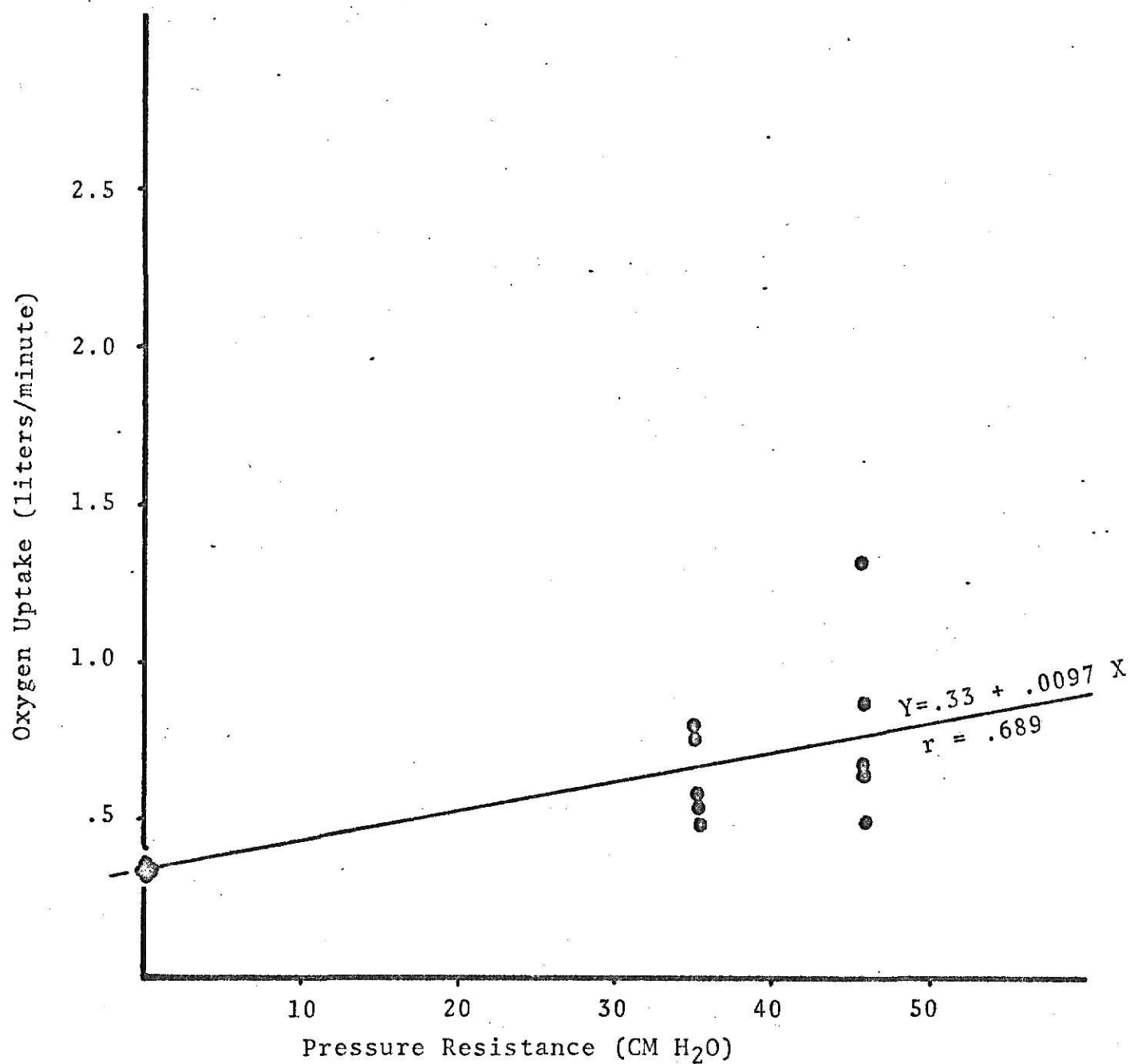


FIGURE 6
OXYGEN UPTAKE vs. PRESSURE AT THE
SEDENTARY ACTIVITY LEVEL

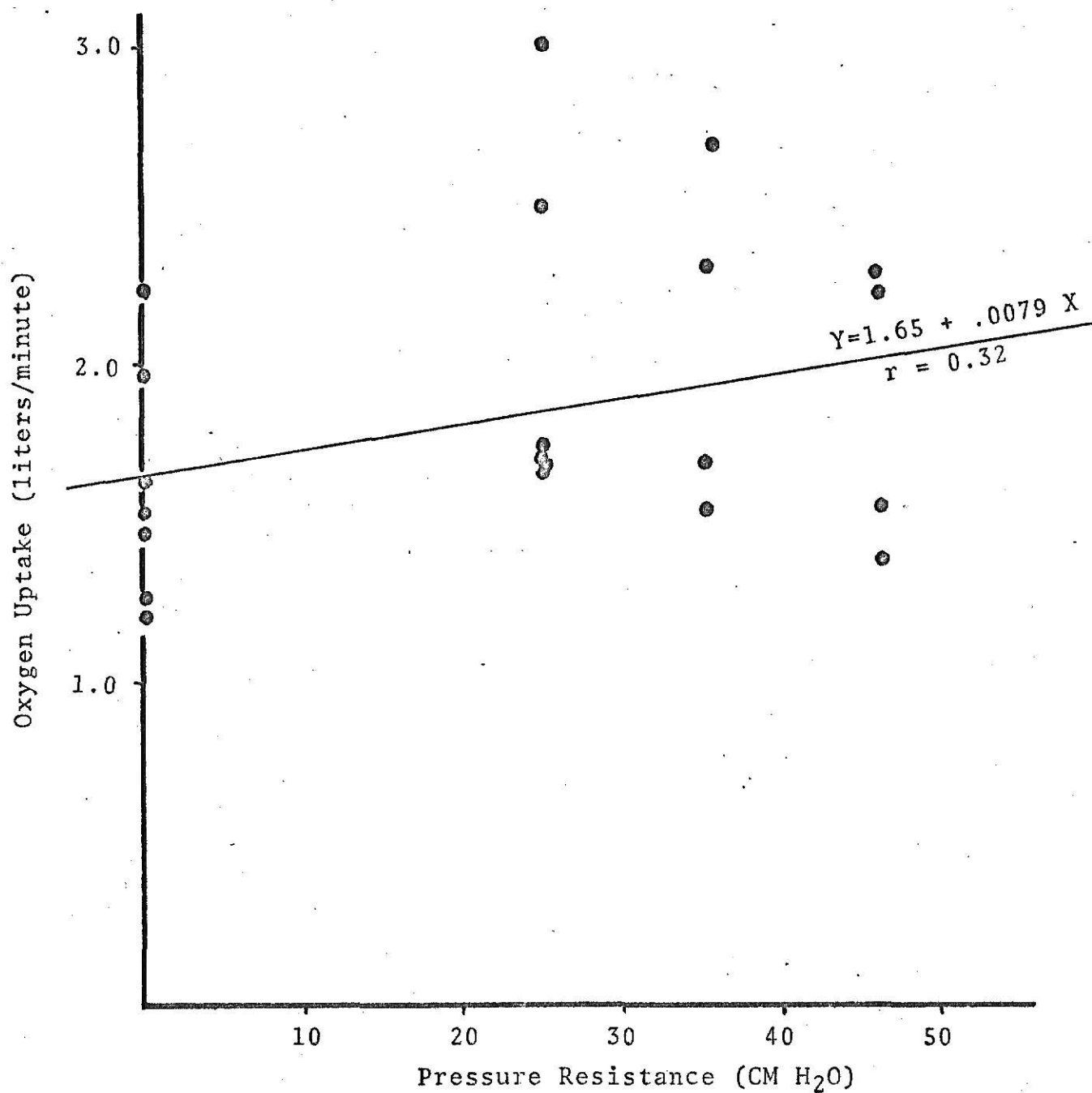


FIGURE 7
OXYGEN UPTAKE vs. PRESSURE AT THE
WALKING ACTIVITY LEVEL

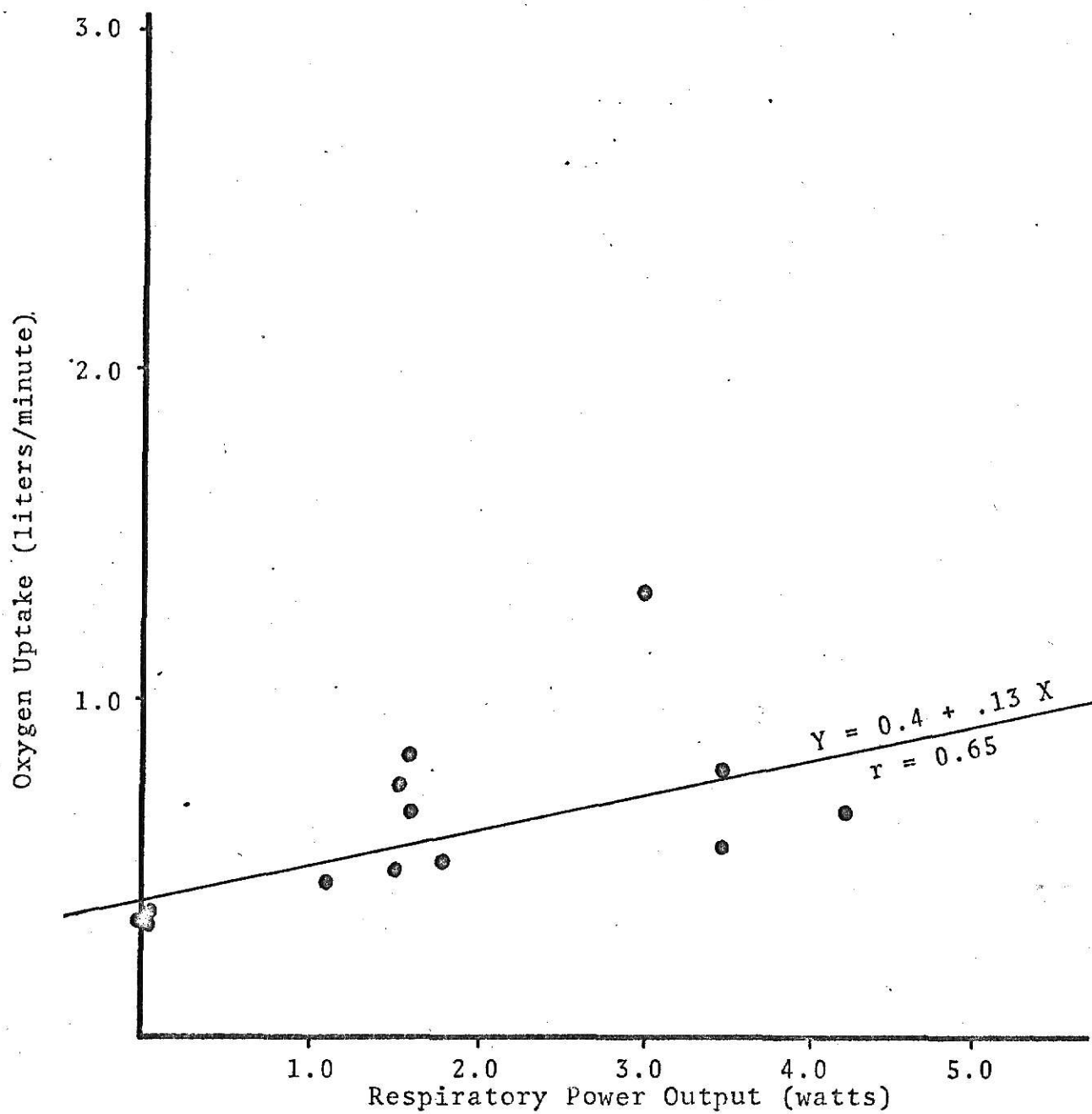


FIGURE 8
OXYGEN UPTAKE vs. RESPIRATORY POWER
OUTPUT AT THE SEDENTARY ACTIVITY LEVEL

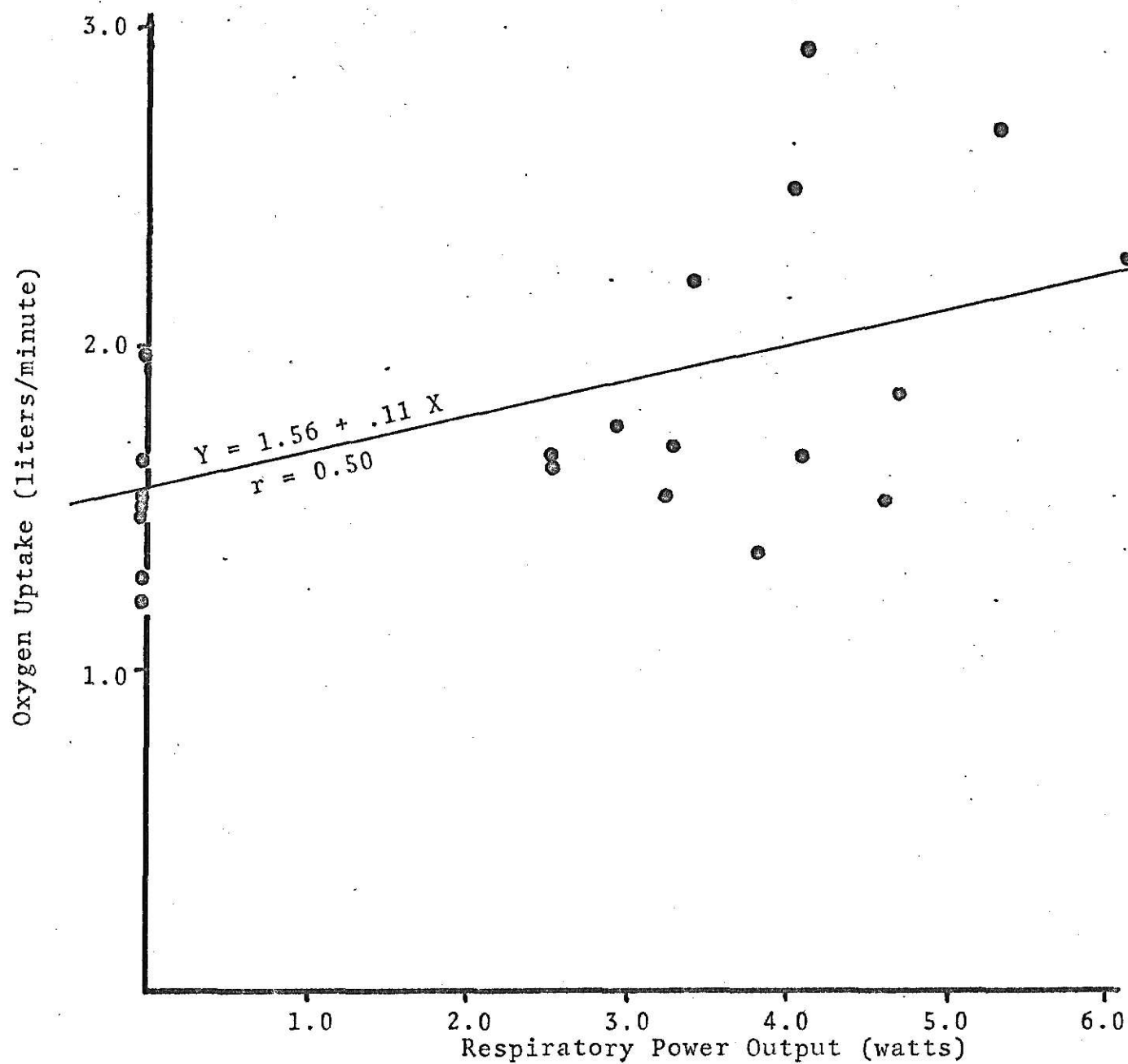


FIGURE 9

OXYGEN UPTAKE vs. RESPIRATORY POWER
OUTPUT AT THE WALKING ACTIVITY LEVEL

Another area of interest in the present research was a determination of the presence or absence of hypoventilation as the test subjects breathed against the high threshold pressure resistance while exercising. Gee, et al. (1968) found that under high exercise workloads and with a moderate viscous resistance to breathing (maximum pressure up to 30 cm H₂O), some test subjects were unable to maintain a normal ventilation.

The present research also showed that test subjects tended to hypo-ventilate when the pressure resistance to breathing was increased. Only four test subjects completed all four test phases at the walking activity level. The combined mean minute volume for these four test subjects is shown below in Table 3 with the corresponding pressure resistances to breathing.

TABLE 3

MEAN MINUTE VOLUME OF FOUR TEST
SUBJECTS AT FOUR RESISTANCE PRESSURE LEVELS

Pressure (cm H ₂ O)	Mean Minute Volume (liters/minute)
0	40.6
25	39.0
35	34.1
46	29.6

Test subjects were intentionally selected with widely varying degrees of physical conditioning and with different athletic histories. A ranking was established which ranked the test subjects according to their accomplishments in each category. Table 4 indicates the ranking of the test subjects according to their present state of physical conditioning and shows their respective ranking of respiratory power output while walking on the treadmill. The test subjects' rankings according to their athletic history is shown in Table 5. The ranking is based on the number of years of varsity athletics in which each test subject participated during his high school and university years. Table 5 also compares the test subjects' rankings in athletic history to their respective rankings of respiratory power outputs. The rankings of athletic history and of present physical condition were determined from an analysis of the answers to questionnaires which were answered by the test subjects. Rank of respiratory power output was based on the ratio of (respiratory power output)/(cardiac power output) for each test subject while walking on the treadmill. This ratio is a simple measure of each test subject's ability to perform respiratory work of magnitude equal to his own cardiac power output.

TABLE 4
RANK CORRELATION OF PHYSICAL CONDITION
AND
THE RATIO $\frac{\text{RESPIRATORY POWER}}{\text{CARDIAC POWER}}$ OF WALKING TEST SUBJECTS

Test Subject	Ranking of Present Physical Condition*	Ranking of the Ratio $\frac{\text{Respiratory Power}}{\text{Cardiac Power}}$
6	1	3
5	2	7
4	3	9
2	4	4
7	5	6
1	6	1
9	7	2
8	8	8
3	9	5

R = -0.10 by Spearman's Rank Correlation Method

*Ranking based on information provided on questionnaires filled in by the test subjects

TABLE 5
RANK CORRELATION OF ATHLETIC HISTORY
AND
THE RATIO $\frac{\text{RESPIRATORY POWER}}{\text{CARDIAC POWER}}$ OF WALKING TEST SUBJECTS

Test Subject	Ranking of Athletic History*	Ranking of the Ratio $\frac{\text{Respiratory Power}}{\text{Cardiac Power}}$
6	1	3
5	2	7
1	3	1
2	4	4
9	5	2
3	6	5
8	8	8
7	8	6
4	8	9

R = 0.60 by Spearman's Rank Correlation Method

*Ranking based on the sum of years that the subject participated in intercollegiate and high school varsity athletics, with intramural participation considered to rank those of equal sums.

CHAPTER V

DISCUSSION

The experimental results reported in Chapter IV show that in some cases the power output capability of the respiratory muscles is greater than the pumping power output of the heart at two levels of activity--sitting, and walking on a 6% grade. By placing an airflow resistance in the inspiratory and expiratory airways of test subjects an additional workload was imposed upon the respiratory muscles. The work done by the respiratory system in breathing against the imposed resistance was determined from the pressure and flow rate measurements of the air being breathed by the test subjects. It is not likely that a self-powered, implanted artificial heart system would present a threshold pressure resistance workload to the entire respiratory system as was the case in the present research. Rather, it would be more desirable to utilize the movements of the breathing muscles to produce power without affecting the pressure balance of the lungs. However, the use of a resistance pressure to effect a respiratory muscle workload is a simple, easily quantified means of measuring the work output of the respiratory muscles. The applicability of the results of the present research to the concept of a self-powered artificial heart system is dependent on several factors.

The normal activity of the respiratory muscles is both continuous and involuntary. If an extra workload were placed on the respiratory muscles, it is conceivable that they would continue to function normally. Campbell, Howell, and Peckett (1957) placed a threshold resistance of 20 cm H₂O (.29psig) in the expiratory airway of unconscious human subjects apparently without greatly affecting the normal ventilation. It is, of course, essential that the self-powered artificial heart system be operated entirely by involuntary muscular activity.

Table 1 and Figure 5 provide a graphical and tabular comparison of the cardiac power output and the maximum respiratory power output for each test subject at two activity levels; seated and walking on a treadmill at 3 miles per hour on a 6% grade. The "identity line" in Figure 5 divides the graph along the locus of points of equal respiratory and cardiac power outputs. The plotted points show that the test subjects varied widely in their ability to perform respiratory work. Five of the walking subjects and two seated subjects were unable to produce respiratory power equal to their own cardiac power outputs. The reason that those five were unable to supply sufficient respiratory power are worthy of further consideration.

Following each test, the test subject was questioned to determine his reaction to the imposed respiratory workload. Surprisingly, most test subjects found the resistance breathing more difficult in the sitting position than while walking, even though considerably more respiratory work was required at the greater

minute volumes while walking. This reaction was predominantly expressed by those test subjects who developed more respiratory power than cardiac pumping power. Several subjects who did not have high respiratory power outputs were able to perform resistance breathing at a higher pressure while sitting than while walking on the treadmill, the opposite reaction to that reported by the subjects with the greater respiratory power output.

Some subjects, when they were unable to complete a phase of the testings, stated that their respiratory muscles were not capable of maintaining an adequate ventilation during the resistance breathing. They believed their muscles were not physically strong enough to inhale and exhale against the pressures. A reflex inhibition of maximal effort may have prevented their respiratory muscles from exerting a maximum effort against the pressure resistance. The actual reason for the wide variation in the test subjects' abilities to perform respiratory work should be examined in future research. If the respiratory muscles of some individuals are actually only strong enough to produce the low work outputs measured in the present research, then it is not likely that the respiratory muscles of those persons could fulfill the power requirements of an artificial heart. On the other hand, if the respiratory muscles are limited by a pressure sensitive inhibitory reflex, then they may possess sufficient strength to power an artificial heart. The proposed implanted energy system would not need to interfere with the normal air pressures of the respiratory tract, and would thus

permit the respiratory muscles to develop their full power capability.

The other reasons that test subjects were unable to complete a part of the test were directly related to the presence of abnormal airway pressures during resistance breathing. Three primary problems which occurred were pain in the ears, inability to maintain an airtight seal between the lips and the mouth-piece, and general discomfort and irritation in the nose and throat. These problems were caused by the air pressure developed in the airways and would not need to be present in an implanted energy system.

The oxygen consumption rate of each test subject was measured at each respiratory workload by closed circuit spirometry. Oxygen uptake was determined by visually evaluating the average rate of change of the gasometer volume from the X-Y plotter record. Room air was used in the closed circuit spirometry and a large carbon dioxide absorbing canister was used in all oxygen consumption tests. Figures 6, 7, 8, and 9 show that the oxygen consumption varied widely, particularly at the increased resistance pressure levels. The general trend, however, was an increase in oxygen uptake as either the pressure or the respiratory power output was increased.

One of the requirements of a self-powered artificial heart system is that the respiratory muscles should be able to maintain the desired power output to the artificial heart without reducing the minute volume of air to the lungs. The presence of hypoventilation is indicated by the data shown in Table 3.

In three of the four cases the respiratory minute volume while walking was decreased as the pressure was increased. Gee, et al. (1968) also noted that test subjects tended to hypoventilate when an air flow resistance was added to their external airways. The tests in the present research were of sufficient duration that the respiratory muscles should have adjusted their breathing patterns to compensate for the increased pressure of resistance breathing. Therefore, it is possible that the hypoventilation indicated in Table 3 is caused directly by the increased workload on the respiratory muscles. If so, then the power output of the respiratory muscles should be restricted to values less than the power level which would induce hypoventilation.

The athletic ability of a person depends to a great extent on the ability of his respiratory system to provide sufficient oxygen to his body. Table 4 and 5 show a rank correlation between the test subjects' respiratory power output and two different aspects of the test subjects' athletic abilities. A ranking of each test subject's present physical condition was established from information provided on the questionnaires completed by each test subject. In view of the widely varying levels of athletic ability of these nine test subjects this was judged to be an acceptable method of ranking. According to Table 4 there is little correlation between one's present physical condition and his ability to perform respiratory work ($R = -0.10$). When only the number of years of athletic activities was considered in ranking the test subjects (Table 5), Spearman's rank correlation coefficient was $R = 0.60$. Neither $R = -.10$ nor

$R = 0.60$ is statistically different from zero at the 95% confidence level. The wide difference between the two correlation coefficients may, however, indicate that one's ability to perform respiratory work is related more to the extent of his athletic participation in high school and university years than to his present degree of physical condition.

In the present research, the whole of which was made up of several interrelated phases, an erroneous measurement or calculation in one phase would affect the accuracy of the entire experiment. Perhaps the most difficult measurement was the oxygen consumption rate as determined by average slope of the curve describing the volume of the gasometer vs. time. During the sedentary oxygen consumption tests, the addition of the threshold pressure resistance caused the breathing to be quite erratic, which made the slope of the oxygen consumption line difficult to determine. It was noted that during resistance breathing the inspiratory end points more clearly defined the slope than did the expiratory end points. This was also noted by other researchers (Agostini and Fenn, 1960) and is apparently a characteristic breathing pattern when there is a pressure resistance on the expiratory airway.

Air leaks are probably the largest potential source of errors in closed circuit spirometry oxygen consumption tests. Every effort was made to carefully construct and seal the apparatus to prevent air leaks. In the present research, portions of the airway tubes and the threshold resistance barrels were continuously under positive or negative gauge pressure which

intensified the need of eliminating air leaks. The system was checked periodically for leaks by pressurizing the entire closed circuit spirometry system and observing the rate of pressure loss in the system. The test program was continued only when the airway system was shown to be virtually free from the air leaks.

Although the mechanical system had been proven leak free, it was possible to lose air at the mouthpiece when the test subject was expiring against a high threshold pressure resistance. The mouthpiece used for these tests was originally designed for use in normal metabolic rate tests in which the air flow resistance is kept at a minimum. The mouthpiece was placed inside the test subject's lips and when there was little pressure resistance, the natural compliance of the lips in conforming to the shape of the mouthpiece maintained the air seal at the mouthpiece. However, when the resistance pressure was introduced a considerable effort on the part of the test subject was required to press the lips tightly around the mouthpiece to form an airtight seal. The problem became intensified when the pressure was increased. The facial muscles began to fatigue after a few minutes of the test and in some cases the test subjects were unable to maintain an airtight pressure seal between the lips and the mouthpiece. It would be desirable to correct this condition in any research conducted in the future. Cook, et al. (1964) also encountered this leakage problem and they designed another type of mouthpiece for pressure breathing which apparently eliminated air leakage at the mouthpiece.

The respiratory minute volume of each test subject was determined during both the sedentary and treadmill walking tests and at the same pressure resistances which were employed in the oxygen uptake tests. As in the oxygen uptake tests, leaks in the airways were potentially significant sources of error. Since the respiratory minute volume was measured either immediately after or immediately before the oxygen uptake tests, the complete system was checked for leaks before both parts of the tests. The minute volume was determined by the amount of air inspired and so the leak at the mouthpiece as described above did not directly affect the minute volume values. The graph of the gasometer volume vs. time was used to determine minute volume. The gasometer volume decreased as air was inspired from it and remained constant during expiration as the test subjects expired to the atmosphere. This gasometer movement pattern described a "stairstep" shaped graph on the X-Y plotter paper and the average slope was easily determined. The test subject's minute volume was equal to the graph slope measured in liters per minute.

There are many factors to be considered in applying the results of the present research to the concept of a self-powered artificial heart system. As with a maximal effort test of a skeletal muscle system, it is difficult to specify the maximum capability of the respiratory muscles because of the subjective nature of such a test. Also the only convenient means of measuring the respiratory muscles' work output is by measuring both the pressures that are developed in the lungs and the volume of air pumped by the lungs, from which a calculation of respiratory

work output may be made. Except in extremely rapid breathing such as a maximum breathing capacity test the lung pressures developed during maximal respiratory breathing efforts are greater than the pressures experienced in any kind of normal breathing. The high pressure changes may excite a reflex which tends to restrict the force being exerted by the respiratory muscles. If this is the case, then the respiratory muscles may be capable of producing more power when coupled to an internal load which does not present a pressure resistance to the lungs. It is known that during certain involuntary responses, such as sneezing, the diaphragm exerts far greater force than is normally possible in a voluntary maximum exertion, (Campbell, 1958). Perhaps the respiratory muscles could develop their full potential power capability only when coupled to an internal workload which does not alter the pressure of the air being breathed.

During maximally rapid breathing such as a maximum breathing capacity (MBC) test, the power output of the respiratory muscles of normal, male test subjects was reported by Milic-Emili, et al. (1964). They found that the average respiratory power output was 38.1 watts for both expiration and inspiration, with 12.6 watts developed during the inspiration only. These MBC tests were conducted under minimum restrictive pressure conditions and were of very short duration, about 15 seconds. Thirty-eight watts is much greater than the 5 or 6 watts present research indicates as a maximum continuous, long-term, power capability of the respiratory muscles. This indicates that the respiratory muscles possess considerable reserve power capability

in excess of the power required to operate an artificial heart.

The normal contraction of the human diaphragm muscle causes inspiration of air into the respiratory system. During expiration the diaphragm is either relaxed or producing an antagonistic effort against expiration. When a pressure resistance to breathing is placed in the expiratory airway, the diaphragm is able to indirectly perform the additional, apparently expiratory work, even though it is basically an inspiratory muscle. The diaphragm performs this "expiratory work" by maintaining a greater residual volume of the chest. When the residual volume of the chest is increased, the passive expiratory force of the chest wall is also increased to overcome the expiratory resistance. Campbell, Howell, and Peckett (1957) also noted this phenomenon in both conscious and unconscious human subjects. They showed that unconscious humans could expire against a threshold pressure of up to 15-20 cm H₂O without the use of the expiratory (abdominal) muscles by using the inspiratory muscles to maintain an increased lung residual volume. When a test subject is breathing against a threshold resistance on both his expiration and inspiration, his inspiratory muscles probably perform most of the work of breathing. It may be possible that if the actual power capability of the respiratory muscles exceeds that shown in the present research, then an artificial heart system could be powered by the inspiratory muscles alone, which would greatly simplify the energy transmission mechanism. This unknown can best be resolved through additional research.

Previous researchers (Cook et al., 1964) report that one's ability to produce high respiratory airway pressures improves with long term practice. Most test subjects in the present research were able to perform respiratory work equivalent to their corresponding cardiac work rate while sitting. It is conceivable that, with additional practice, all subjects could develop sufficient respiratory power for an artificial heart. A potential artificial heart recipient could be subjected to some form of respiratory exercise designed to increase his respiratory power output capability before he actually received a self-powered artificial heart system. After he received an artificial heart, he would work at gradually increasing levels of activity until his respiratory power capability increased to the extent that he could lead a normal life. The fact that several subjects were able to produce respiratory power in excess of their own cardiac power outputs at both activity levels indicates that the self-powered artificial heart system may be feasible and that the concept is worthy of further research.

In addition, the possibility of reducing the power requirements of the artificial heart system should be considered. Perhaps a small, constant power electrical battery or similar energy storage system could be used to supply the basal cardiac power requirements with the respiratory muscles providing the additional power input to the artificial heart. Such a compromise may be necessary to provide an artificial heart system which is adaptable to the patient's changing activity levels.

CHAPTER VI

SUMMARY AND CONCLUSIONS

Four specific objectives, stated in Chapter I, were examined in this thesis. The feasibility of powering an artificial heart by power obtained from the movements of the respiratory muscles was considered. The methods of calculation respiratory power output capability and cardiac power output are discussed in Chapter II. Nine caucasian males, age 20 to 40, served as test subjects in various pulmonary function tests designed to evaluate the feasibility of the self-powered artificial heart system. Two activity levels were examined; sedentary, and treadmill walking at three miles per hour on a 6% grade.

Conclusions

1. The maximum respiratory power output of nine test subjects was determined at two activity levels. This maximum power ranged from 2.9 to 6.1 watts (mean 4.0 watts) while walking. The range was from 1.1 to 4.2 watts (mean 2.5 watts) in the sedentary activity level.
2. The ratio of respiratory power output capability to cardiac power output was evaluated at two activity levels. The ratio ranged from 0.66 to 1.24 (mean 0.92) while walking and from 0.76 to 2.80 (mean 1.58)

while sitting.

3. The oxygen cost of performing respiratory work was determined and is given by the following regression equations:

$$\text{Oxygen uptake while seated} = 0.4 + 0.13 (\text{Respiratory Power Output})$$

$$\text{Oxygen uptake while walking} = 1.56 + 0.11 (\text{Respiratory Power Output})$$

Oxygen uptake in liters/minute

Respiratory power in watts

4. The ratio of respiratory power to cardiac power output was compared to the test subjects' athletic abilities. Two aspects of the subjects' athletic background were considered--their present degree of physical conditioning; their total number of years of varsity athletics participation. A rank correlation analysis showed $R = 0.60$ for the athletics history of test subjects and $R = -0.10$ for their present degree of physical condition, neither of which is statistically significant at the 95% confidence level.
5. Four of the nine test subjects were able to produce respiratory power equal to their corresponding cardiac power outputs at both activity levels. The self-powered artificial heart system is feasible if the respiratory power capability can be significantly

improved through exercise or if the power requirements of the heart system may be reduced by use of an auxiliary power source.

Future Research

The present research was, to the author's knowledge, the only study made to date to examine the proposition stated in Chapter I. The present research was somewhat limited in terms of facilities and number of test subjects. However, it is a start into an area which could ultimately result in the release of artificial heart patients from external power requirements. The results of this study indicate that the respiratory muscles are capable of producing power of magnitude comparable to the pumping power output of the normal heart. Future research should be directed to two aspects of the self-powered artificial heart system.

First, the results of the present study should be verified and expanded--perhaps by using more test subjects and at different activity levels. The long term effects of causing the respiratory muscles to produce additional power should be examined. Hypoventilation, if present whenever the respiratory muscles must produce additional power, may limit the power output of the respiratory muscles to levels which do not cause hypoventilation. These are questions which were not adequately answered in the present research and which must be considered as the concept of a self-powered artificial heart is developed.

The other area of future research was not considered in the present study. That is the development of a method of loading the respiratory muscles with a device to convert respiratory power to a form of power input to an artificial heart. The present research only indicated that the power capability was enough to consider further work, but suggested no way to harness the power and use it to drive an artificial heart. The design of a device could be perfected by surgically implanting experimental devices in animals. By attaching a power-dissipating mechanism to animals' respiratory muscles, the researcher avoids encountering the undesirable effects of changing the pressure within the test subjects' airways. By this method the power output capability of the respiratory muscles could be examined without concern for the possible erroneous results resulting from the abnormal airway pressures.

LIST OF REFERENCES

- Agostini, Emilio and W. O. Fenn. "Velocity of Muscle Shortening As a Limiting Factor In Respiratory Air Flow." J. Appl. Physiol., May 1960, 15(3): 349-353.
- Best, C. H.; and N. B. Taylor. The Physiological Basis of Medical Practice, Chapter 32. Williams and Wilkins Co., Baltimore, 1961
- Bevegard, S; A. Holmgren, and B. Jonsson. "The Effect of Body Posture on the Circulation at Rest and During Exercise with Special Reference to the Influence on the Stroke Volume." Acta Physiol Scand, 1960, 49: 279-298.
- Campbell, E. J. M. The Respiratory Muscles. London: Lloyd-Luke Ltd., 1958
- Campbell, E. J. M.; E. K. Westlake, R. M. Cherniack. "Simple Methods of Estimating Oxygen Consumption and Efficiency of the Muscles of Breathing." J. Appl. Physiol. Sept, 1957, 11(2): 303-308.
- Campbell, E. J. M.; J. Howell, B. Peckett. The Pressure-Volume Relationships of the Thorax of Anesthetized Human Subjects; A Comparison of the Effects of Expiratory Resistance and Positive Pressure Inflation." J. Physiol. May 1957, 136(3): 563-568.
- Cook, C. D.; J. Mead, M. M. Orzalesi. "Static Volume-Pressure Characteristics of the Respiratory System During Maximal Efforts." J. Appl. Physiol., Sept 1964, 19(5):1016-1022.
- Craig, A. B., Jr. "Maximal Work of One Breathing Cycle." J. Appl. Physiol., Nov 1960, 15(3):1098-1100.
- Fletcher, G. and J. W. Bellville. "On Line Computation of Pulmonary Compliance and Work of Breathing." J. Appl. Physiol., 1966, 21(4):1321-1327.
- Gee, J. B. L.; G. Burton, C. Vassallo, J. Gregg. "Effects of External Airway Obstruction on Work Capacity and Pulmonary Gas Exchange." American Review of Respiratory Diseases., 1968, v. 98:1003-1012.

- Granath, A.; B. Jonsson, T. Strandell. "Circulation in Healthy Old Men, Studied by Right Heart Catheterization at Rest and During Exercise in Supine and Sitting Position." Acta Medica Scandinavica, Oct 1964, 176(4):425-445.
- McGregor, M., and M. Becklake, "The Relationship of Oxygen Cost of Breathing to Respiratory Mechanical Work and Respiratory Force." J. Clin. Invest., June 1961, 40(4):971-980.
- Margaria, R., G. Milic-Emili, J. M. Petit, and G. Cauagna. "Mechanical Work of Breathing During Muscular Exercise." J. Appl. Physiol., May 1960, 15(3):354-358.
- Milic-Emili, J.; M. M. Orzalesi, C. D. Cook, J. M. Turner. "Respiratory Thoraco-Abdominal Mechanics in Man." J. Appl. Physiol., March 1964, 19(2):212-223.
- Rahn, H.; A. B. Otis, and L. E. Chadwick and W. O. Fenn. "The Pressure-Volume Diagram of the Thorax and Lung." Am. J. Physiol., May 1946, 146(2):161-178.
- Reeves, J. T.; R. F. Grover, S. G. Blount, and G. F. Filley. "Cardiac Output Response to Standing and Treadmill Walking." J. Appl. Physiol., March 1961, 16(2):283-228.
- Robinson, T. C.; R. J. Harvey, L. Van Someren, C. G. LaFarge, W. F. Bernhard, J. C. Norman. "Determination of Blood Pumping Power Requirements for an Artificial Heart." Progress in Biomedical Engineering. Edited by Fogel and George; Spartan Books, Washington, D. C. 1967.
- Tabakin, B. S., J. S. Hanson, T. W. Merriam, Jr., and E. J. Caldwell. "Hemodynamic Response of Normal Men to Graded Treadmill Exercise." J. Appl. Physiol., May 1964, 19(3):457-464.
- Van Wylen, G. J., and R. E. Sonntag. Fundamentals of Classical Thermodynamics, New York: John Wiley and Sons, Inc., 1965

APPENDIX A

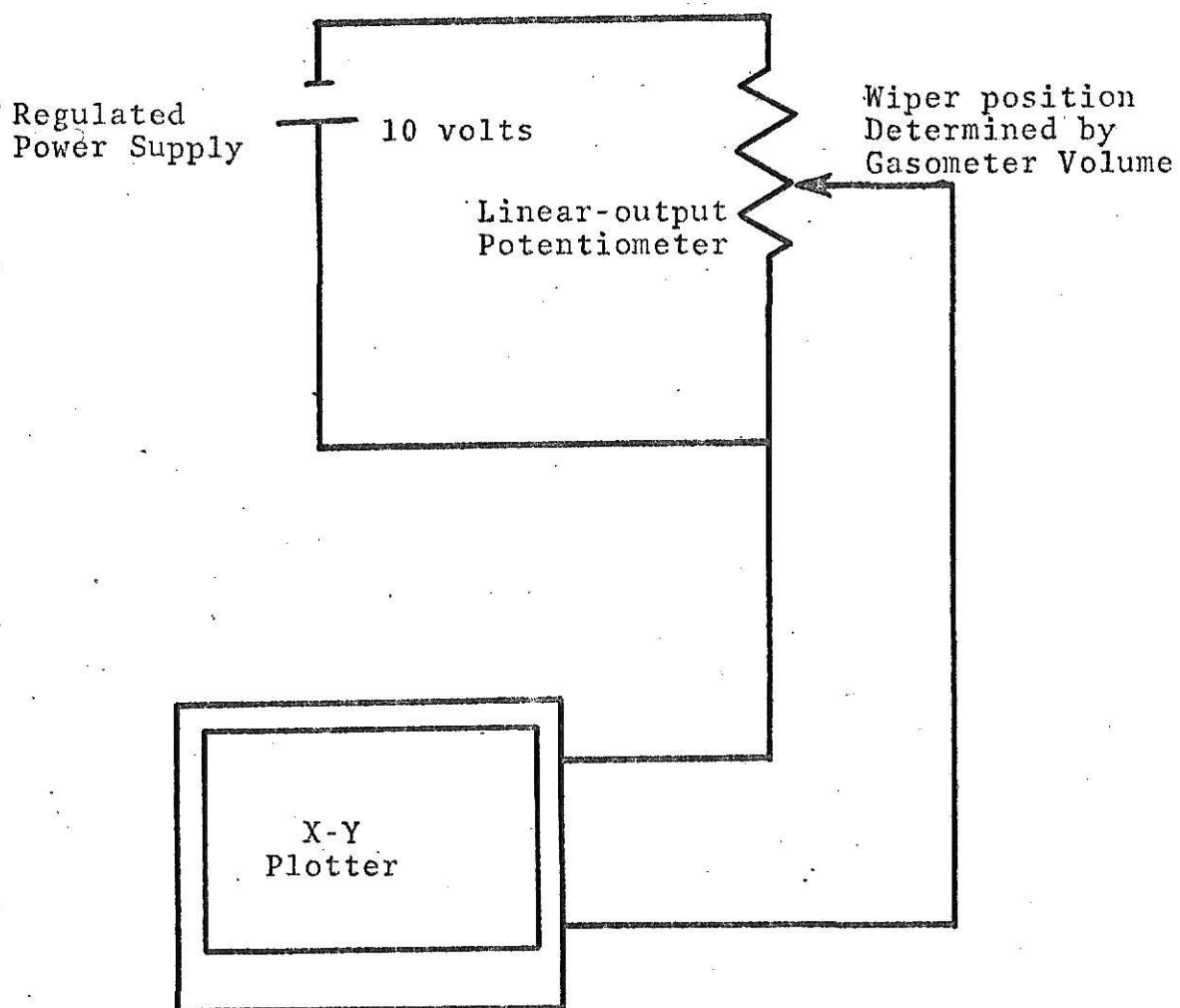


FIGURE 10

SCHEMATIC WIRING DIAGRAM OF EXPERIMENTAL APPARATUS

APPENDIX B

TABLE 6

PHYSICAL CHARACTERISTICS OF TEST SUBJECTS

Subject	Age	Weight (lb)	Height (in)	Vital Capacity (liters)	Years of Athletics
1	23	150	70	5.6	4
2	31	193	72	5.1	3
3	20	180	70	4.1	3
4	40	150	66	4.4	0
5	22	155	71	5.1	7
6	23	157	73	4.9	7
7	25	170	72	5.4	0
8	26	129	66	3.7	0
9	22	245	72	5.4	3
mean	26	170	70	4.9	3

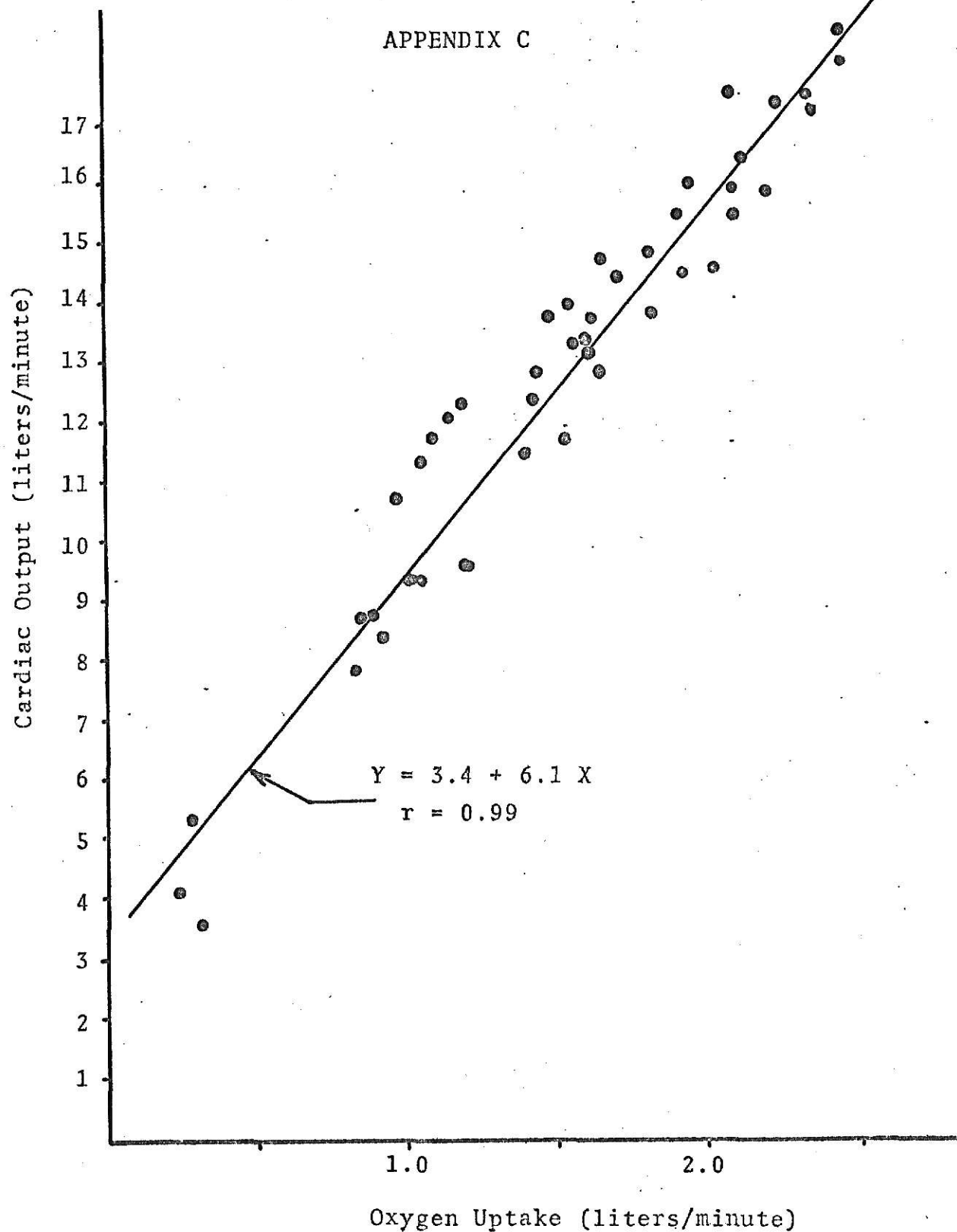


FIGURE 11
CARDIAC OUTPUT vs. OXYGEN UPTAKE FROM
THE DATA OF REEVES, et al., 1961

APPENDIX D

PERSONAL VITAL DATA AND QUESTIONNAIRE
Respiratory Power Performance Tests

NAME _____ DATE _____

AGE _____ WEIGHT _____ (lbs) HEIGHT _____ SEX: m f

VITAL CAPACITY _____ (liters)

ATHLETIC HISTORY

High school: (check one)

- a. participated every season
- b. participated few sports
- c. participated in no sports

University:

- a. varsity (____) years
- b. intramurals (____) years
- c. personal fitness program
(specify) _____

d. none

Other Post-collegiate fitness (if applicable):

- a. personal fitness program

- b. manual labor occupation

- c. presently involved in little
or no physical exercise

APPENDIX E

EQUIPMENT USED FOR TAKING DATA

1. Regulated Power Supply
Heathkit Model 1P-20
2. X-Y Recorder
Honeywell 550 X-Y Recorder
3. Gasometer
Warren E. Collins, Inc. Model P-1700-F
Chain Compensated Gasometer
120 liter model
4. Treadmill
Warren E. Collins, Inc. Model P-2000-E
1-8 mph Standard Model
18" wide belt
28" wide walking platform
54" long walking belt surface
5. Two-way Breathing Valve
Warren E. Collins, Inc. Model P-306
High Velocity, Low Resistance Breathing Valve
1½" Inside Diameter Tubes
6. Pressure Transducer
Pace-Wiancko Pressure Transducer Kit, Model KP15
Pace-Wiancko Transducer Indicator, Model CD25

ACKNOWLEDGEMENTS

Indebtedness is acknowledged to Drs. Preston E. McNall, Arlin M. Feyerherm, Stephan A. Konz, and Emerson L. Besch for their guidance in conducting the experimental research; to Ralph Gwinn for assistance with the pulmonary function equipment and auxiliary experimental apparatus; to Kay Sharp and Alice Neese for assistance in preparing the manuscript; and to the test subjects for their interest in this project and for their respiratory efforts.

A FEASIBILITY STUDY FOR SELF-POWERED
ARTIFICIAL HEART SYSTEMS

by

MARK EDWARD SHARP

B. S., Kansas State University, 1968

AN ABSTRACT OF A THESIS

submitted in partial fulfillment of the
requirements for the degree

MASTER OF SCIENCE

Department of Mechanical Engineering

KANSAS STATE UNIVERSITY
Manhattan, Kansas

1970

Name: Mark E. Sharp Date of Degree: January 30, 1970

Institution: Kansas State University Location: Manhattan, Kansas

Title of Study: A FEASIBILITY STUDY OF SELF POWERED
ARTIFICIAL HEART SYSTEMS

Pages in Study: 68 Candidate for Degree of Master of Science

Major Field: Mechanical Engineering

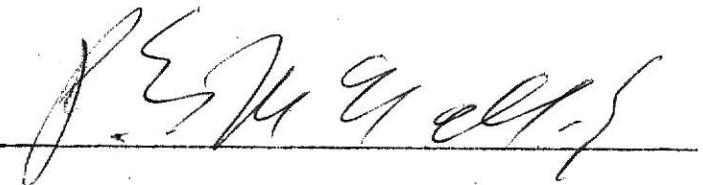
This thesis examines the feasibility of an artificial heart system in which the respiratory muscles, while performing the normal breathing function, supply the power to drive the artificial heart pump. Nine male test subjects were subjected to a series of tests to evaluate the potential power output capability of the human respiratory muscles. The subjects breathed continuously against a threshold pressure resistance and the power output of the respiratory muscles was calculated from the pressure and the air flow rates. Tests were conducted with the subjects at two different activity levels; a relaxed seated position, and walking on a treadmill at 3 mph on a 6% grade.

An analysis of the data of previous researchers provided a means of calculating the cardiac pumping power output of the test subjects when the type of activity and the oxygen uptake rate were known. Each subject's cardiac power output was compared to his respiratory power output at each activity level. The ratio of each subject's maximum respiratory power output to his cardiac power output was evaluated at both activity levels.

The ratio ranged from .66 to 1.24 (mean 0.92) while walking and from .76 to 2.80 (mean 1.58) while sitting.

The oxygen cost of performing respiratory work was determined. Also, the correlation between one's athletic background and his ability to perform respiratory work is discussed.

MAJOR PROFESSOR'S APPROVAL

A handwritten signature in cursive script, appearing to read "J. E. M. G. et al.", is written over a horizontal line.

VITA

Mark E. Sharp

Candidate for the Degree of
Master of Science

Thesis: A FEASIBILITY STUDY OF SELF POWERED ARTIFICIAL
HEART SYSTEMS

Major Field: Mechanical Engineering

Biographical:

Personal Data: Born in Clay Center, Kansas, October 1,
1945, the son of Wayne E. and Rosalie M. Sharp.

Education: Attended grade school near Wakefield, Kansas;
graduated from Manhattan High School in 1963;
received the Bachelor of Science degree from Kansas
State University, with a major in Mechanical
Engineering, in June, 1968; completed requirements
for the Master of Science Degree in January, 1970.

Professional Experience: Employed by Texas Instruments,
Incorporated in Dallas, Texas, from June 1968 to
the present; an associate member of the American
Society of Mechanical Engineers.