THEORETICAL ANALYSIS
AND
PRELIMINARY DEVELOPMENT
OF AN
INDIRECT BLOOD PRESSURE RECORDING SYSTEM

by
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B.S., Case Institute of Technology
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SUBMITTED IN PARTIAL FULFILLMENT
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Signature of Author.
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Certified by.......
Thesis Supervisor

Accepted by....
Chairman, Departmental Committee
on Graduate Students

PREFACE

The Research Program Represented by the Data Here Assembled and Reported Was Done Under the Medical Preceptorship of Dr. Ralph Adams[1] and the Technical Supervision of Professor Kenneth Wadleigh[2].

These studies were supported in part by grant-in-aid from the New Hampshire Heart Association.

The medical work was all done at Huggins Hospital, Wolfeboro, New Hampshire.

The academic work was all done at Massachusetts Institute of Technology, Cambridge, Massachusetts.

A portion of the medical appraisal and evaluation was done at Boston University School of Medicine, Boston, Massachusetts.

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Author's Abstract:

The basic problem is to determine whether or not a satisfactory method can be developed for the continuous measurement of blood pressure by indirect techniques. Upon the establishment of some overall criteria, and with a survey into the field, various possible approaches to the problem are considered. Each of the possible approaches is preliminarily analyzed, and the most realistic possibility is analyzed in detail.

The technique which is studied in detail is based on a pressure variable parameter of the cardiovascular system. This parameter is arterial volume. It is shown that the arterial volume is quasi-linearly related to intra-arterial pressure for a theoretical model, and that experimental data substantiates this analysis. The logic behind the pressure variable parameter system is simply; since volume can be measured externally with greater ease that pressure, then with a relatively simple system blood pressure can be monitored.

A theoretical analysis and experimental documentation is made for each assumption in the proposed theory. Based on the proposed theory, a developmental transducer system is proposed and built. With various analyses, the transducer is modified and ultimately, a linearized transducer is presented.

With a linearized capacitance transducer, the system is subjected to clinical conditions, and various effects are measured and analyzed. The results of a preliminary program of analysis indicate that under controlled conditions, the proposed system has some significant potentialities. Over short periods of time, the system is able to record human blood pressure continuously to an accuracy of a few percent.

It should be noted that under certain clinical conditions, the proposed theory appears to be completely invalid. These conditions limit the effectiveness of the proposed device.

With further development, the author feels that some of the present limitations can be overcome. A program for further development work is suggested, along with a new transducer the design of which is based on the information gained by the present analysis.

Thesis Supervisor: Kenneth R. Wadleigh
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I. INTRODUCTION

A basic concern of this thesis is to analyze methodologies for the measurement of human blood pressure by indirect means. The background out of which this concern arises is conceptually rich in potentialities. An approach to a medio-technological problem can obviously take many forms. In recent years there has been a trend toward the integration of disciplines, particularly medio-technological, in order that medically based concerns may use the knowledge and guidance of specialized fields of technology. This approach is not new, for in the era of Newtonian Physics, considerable work was done between men in "specialized" fields for the advancement of medical science. Poiseuille is a good example of this type of discipline integration. What appears to be new, in practice, is the application of specialized technological knowledge to medical problems, part of which are engineering in nature. This "cross pollination", as it were, appears to be somewhat different, in degree, from the historical approach, particularly in the sciences. In Technology, the objective is a particular solution to a particular problem which meets the particular needs of society at this point in history. As we know, pure scientific investigation is that approach which seeks to understand and to broaden man's knowledge of the universe in which he lives. In Latin, Scire means to know, hence the word Science. However, Ingeniare in Latin means to plan, to contrive, to build, and hence the origin of the concept of engineering.

There has been, within this program for a thesorial dissertation, a sincere attempt to study objectively a medio-technological problem, giving as full consideration as possible to both disciplines. Under the guidance of preceptors in both Medicine and Engineering, this thesis has been undertaken.

1.1 The Basic Concern from a Medical Point of View

The basic concern out of which this particular study arises, is a medical concern for adequate knowledge of the clinical status of patients under various conditions of duress. Clinical medicine is that branch of the medical profession which is concerned with the on-going treatment of patients. It can be research oriented but is directed toward the actual treatment of individuals.

It is obvious that clinical medicine is broader in scope than the field of clinical surgery, however, the later was the primary impetus for this project.

It is essential for the clinical surgeon to have adequate knowledge of the condition of his patient if optimum medical results are to be obtained. The advent of electronics has provided another avenue of approach to enable the surgeon to assay somewhat more objectively, the condition of his patient. Instrumentative methods can only extend the clinician's awareness into the basic problems of homeostasis. It is from this general background that a new approach to the continuous monitoring of blood pressure was deemed desirable and necessary.

The continuous measurement of human blood pressure by indirect means has been a concern for which a number of generations of medical clinicians, research scientists, physiologists, and others, have sought realization. Continuous, accurate recording of blood pressure by indirect means would be an important addition to the field of medicine. It is the intent of this dissertation to discuss conceptually a new method which appears to have some promise and potentualities.

1.2 Technological Problems and Implications

From a technological point of view, the recording of blood pressure by a pitot tube cannula is a most desirable method, since one measures pressure directly. This technique, however, is attended by such problems as coagulation and damping. The conventional method of indirect recording of blood pressure by the sphygmomanometer is, again speaking from a technological point of view, quite insufficient because of its inaccuracies, and also because of the dependence upon interpretation by the practicing physician. Technological difficulties arise from the fact that many features which are constant in a physical sense, definitely are not constant when combined within the body. One can sight numerous examples to indicate that uniformity in isolation does not necessarily remain as uniformity in combination. However, when one draws an experimental or theoretical model about the body, one can clarify thought processes and come to a reasonable approach to this concern.

1.3 Medical Problems and Implications

The problems of recording human blood pressure by indirect means are not simple in nature. The human body is extremely complex in a medical sense as well as in a technological sense, and specifically in reference to the recording of blood pressure by indirect means. For example, it is known that the sphygmomanometer is an instrument which will yield blood pressure by indirect means to an accuracy of between 10% and 15%. This is a convenient way of recording blood pressure but it does not record it on a continuous basis. The implications in the direct method, using intra-arterial cannulas, are easily followed. The introduction of any foreign body into an artery is traumatic and may cause ischemia, thrombosis, or even more serious complications. If equally valid information could be obtained painlessly, accurately, and without risk, expanded application and usefulness could be anticipated. As attested by extensive literature which has been scanned in the course of this investigation, there have been literally hundreds of attempts at a solution to this problem from the inception of the sphygmomanometer (1) to the present day, intra-arterial catheter transducer system. The current state of the art is highly developed, but the measurement of blood pressure by a convenient, indirect and accurate means has awaited a method for fulfillment. Personal communications from Dr. Julius Comroe, of the University of California Medical School, and from Dr. Gerald Meade of the Harvard University School of Public Health, have indicated their opinion that a completely satisfactory solution to the problem of indirect recording of blood pressure is as yet unannounced.

Further survey of the literature (2-21), with regard to types of investigations and developments which have been made, shows that all of the numerous systems which have evolved may be reduced to only two basic ideas. The first is the direct measurement of blood pressure by insertion of a cannula or needle, and a recording of the dynamic response by the various techniques. The second is the sound system which was discovered by Korotkow (22) in 1907, and which is better known to the medical profession as sphygmomanometer. The system involves the analysis of the sound characteristics. As reported by others (23-32), we have also found considerable deviation between intra-arterial pressure readings and sphygmomanometer readings. There has been a great deal of work done to "educate" the clinician to these facts, and also to the fact that individuals will read the same blood pressure differently by cuff technique in accordance with their ability to analyze the sound characteristic. The American Heart Association has published a booklet (33) to this effect, and has attempted to make it available to practicing physicians throughout the country. The accurate recording of human blood pressure has achieved new importance with developments in modern surgery, the advent of space medicine, and the recognition of close inter-relationships between blood pressure, medication, and psychosomatic or psychiatric mental status. An accuracy is needed which is better than that obtainable with a stethoscope and cuff technique and on a continuous time basis.

1.4 Historical Review of Indirect Blood Pressure Recording System

Historically it appears that the first measurements of blood pressure were made by Stephen Hales (34). He connected a flexible trachea to the blood vessels of a goose and measured the rise of blood in a vertical manometer. Poiseuille (35) in 1828 made significant improvements on Hales' concept by using a piezometer tube to record blood pressure continuously. This was the beginning of blood pressure measurement. The direct method by definition, requires some type of cannulation procedure.

Quite often, it is undesirable to cannulate an artery for the evaluation of blood pressure. To eliminate cannulation procedures, instruments, which have become known as sphygmomanometers, were developed around the beginning of the twentieth century. In their simpliest form, they consist of an inflatable annular torus which completely surrounds the brachial artery. To this torus is attached a U-tube, or well manometer, an inflating pump or bulb, and a variable control exhaust valve. One of the first applications of this device was by Riva Rocci (1) which became known as the "Palpation Method". The technique (over simplified for descriptive reasons) used by many clinicians today, was first introduced by Korotkow in 1907 and is known as the "Auscultatory Method". After the cuff is applied, the bell of a stethoscope is placed over the brachial artery.

Then as the pressure in the cuff is increased, audible sounds (as heard in the stethoscope) appear with each pulsatile variation in blood pressure. The pressure is then increased until these sounds become inaudible. Thereupon, the system is slowly exhausted until an audible sound reappears. The reappearance of a sound is believed to be due to the formation of an orifice and a pulsating high velocity jet stream. The cuff pressure, concurrent with the sound is taken to be the systolic or maximum pressure within the arterial system. Upon further reduction of the cuff pressure, the audible sounds will diminish and ultimately disappear. The cuff pressure at the diminishing or disappearance of sound is recorded as the diastolic or minimum pressure level. Various techniques for the improvement of this "Auscultatory Method" technique have been tried over the years. As previously indicated, a great deal of work has been done to improve the practice of this useful technique.

The sphygmomanometric technique for blood pressure measurement is almost universally accepted as a satisfactory method of making single observations of the maximal and minimal pressure levels. This technique is not readily adapted to continuous recording because of its very nature. There have been numerous ideas and developments for the adaptation of the sphygmomanometer to automatic registration. The literature describes many instruments for automatic recording sphygmomanometers of two general types. One type records the arterial manifestation (such as sound, radial pulsations, or volumetric change) simultaneously with a record of the occluding cuff pressure. This type of device leave the interpretation to the physician. The second type is similar except that the instrument does the interpretation mechanically or electronically and indicates the corresponding pressures directly.

Instruments of the first type have the advantage in that errors and artifacts can be readily noticed and pressure levels can be more intelligently obtained. On the other hand, the data is not immediately available and is subject to the personal bias and interpretation of the physician.

Instruments of the second kind have the advantage of being relatively consistent, but are complex, and subject to mechanical disarrangement and maladjustment.

Yet there are many occasions in medicine when a recording sphygmomanometer would be extremely valuable. This is borne out by the extensive literature over many years describing various methods and equipment designed for this purpose. The problem of continuous monitoring of blood pressure is a very difficult one at best. The fact that none of these devices thus far developed have ever been widely accepted by the medical profession is proof of a need for further research and development.

One could discuss numerous aspects of some of these devices, but it should suffice to mention a few.

Various instruments have been developed over the years, but these are regarded to have disadvantages of one sort or another which has usually been recognized (36) by the inventor. The Taylor Instrument Company (37) no longer manufactures commercially the instrument which it had developed earlier, because of impracticality. The Ferrand Optical Company (38) recently brought out a version of a sound system instrument which uses a sensitive microphone and records blood pressure changes and sound variations on a recording chart. This mechanism is similar to the Gilson Accoustical Pneumatic System (39). An ingenious device was originated by Lanze (40) which was similar in principle. Ornberg (41), Stokis (42), and Doupe (43) and others(44-47) have employed other methods for attempted solution to the problem of recording blood pressure indirectly. A major objection to almost all of these systems is that a cuff is required. and the interpretation of the arterial manifestation is not accurate enough.

Furthermore, a cuff which is required for the occlusion of an artery, constantly or intermittently, over a long period of time, is uncomfortable and may be injurious to the extremity. While a system can be electronically programmed to overcome some of the inherent difficulties, a system of this type gives only an intermittent recording of blood pressure which is not always medically desirable. The National Bureau of Standards (38) dissatisfaction with the state of the art was revealed in 1954. with the development of a unit which records intermittently by an electronically programmed unit, using a cuff and a pickup microphone, which is fairly well engineered.

An interesting variation on the indirect method of blood pressure measurement was recently described by Wood, et al (48) in which a pressure capsule is used to occlude the small vessels in the pinna of the ear. A photocell incorporated in the capsule registers the dynamic variations in the optical density of the pinna as the blood flows through the small arteries. It is possible to interpret these variations in order to derive the two pressure levels. This system has been used successfully in place of the sphygmomanometer by the anesthesia staff of one well known hospital (49).

1.5 Summary of the Results of a Literature Search

After intensive investigation of the subject, it was found that all previously developed systems for recording human blood pressure involved two basic ideas. First, all indirect methods are based on arterial constriction and the creation of a jet stream; and in the final analysis are nothing more or less than what the clinician has been doing by using his cuff and stethoscope for many years. Second, the cannulation methods involve the introduction of cannulas into the arterial section and then the recording of the associated pressure change on various kinds of recording systems - the most recent of which is electromechanical in nature. Years of research have been spent to develop new systems which have all, in the final analysis, been reduced to one of these two basic areas.

All of this work, including investigations by the author, suggested that a new method should be sought and investigated.

II. PRELIMINARY INVESTIGATION AND ANALYSIS

2.1 Definition of the Problem and Criteria for Solution.

The Problem: The problem is to determine whether or not a satisfactory method can be developed for the continuous measurement of blood pressure by indirect techniques.

The Criteria: This project is of a medio-technological nature and as such there are certain criteria which must be understood before any analysis can be undertaken. The proposed criteria will be to develop a monitoring system:

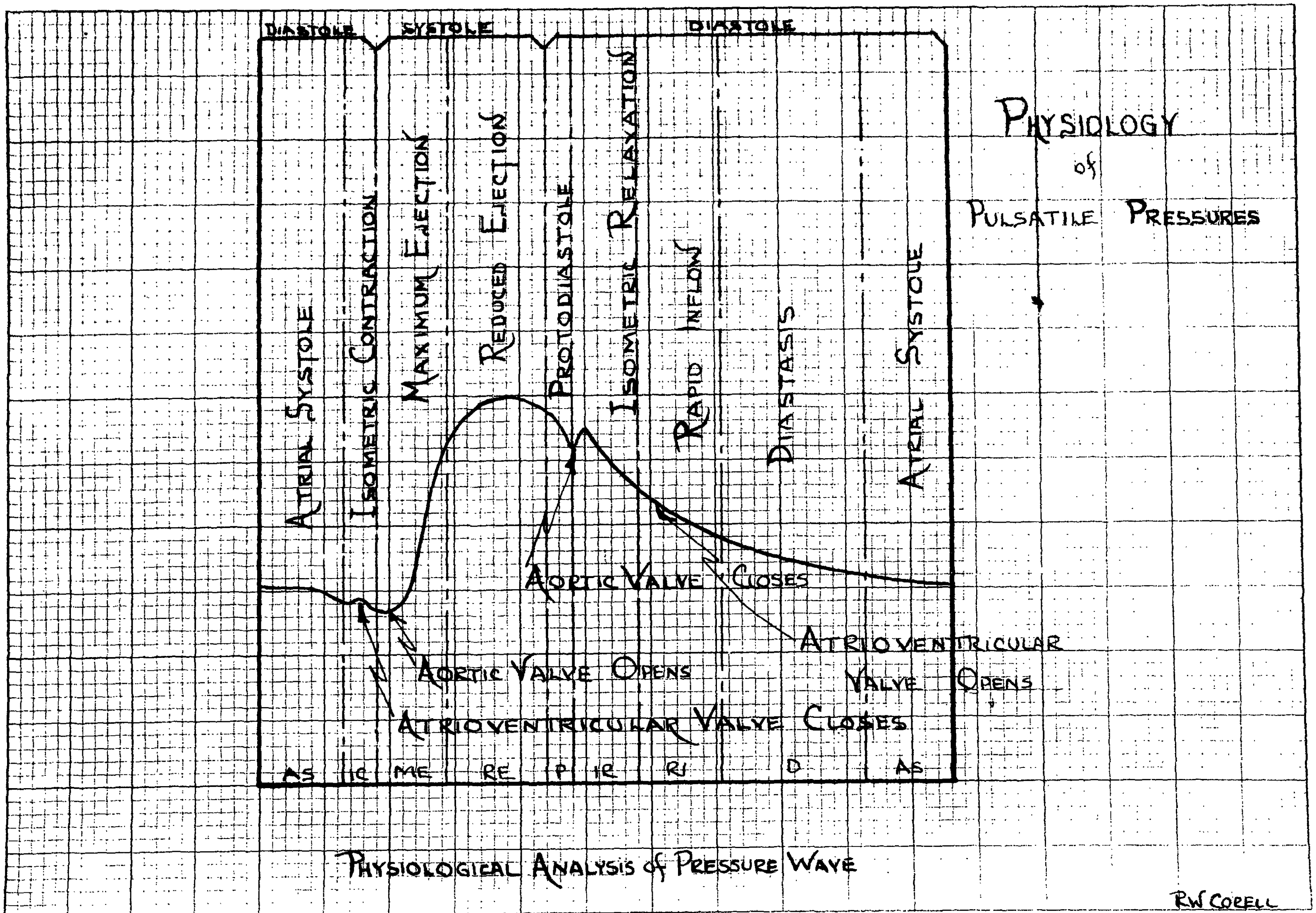
1.) To record blood pressures on a continuous basis. This means to register blood pressures graphically as a function of time.

2.) To obtain blood pressure by non-traumatic techniques. This is the underlying assumption of the indirect method. Non-traumatic techniques will be understood as methods which do not involve injury to, or the penetration of, skin tissues.

3.) To record blood pressures that will compare in accuracy to the selected standard for comparison.

4.) To record blood pressure regardless of the patient's status or condition.

5.) To monitor blood pressure so that characteristic cardiac functions are observable. It is hoped that any new device will be sensitive to changes in cardiac function, such as extra-systole, ventricular fibrillation, and the closure of the aortic valve. (As indicated in the Physiological Analysis of the Pressure Wave diagram on the following page.)


Physiology of Pulsatile Pressures
Diastole
Systole
Diastole
Atrial Systole
Isometric Contraction
Maximum Ejection
Reduced Ejection
Protodiastole
Isometric Relaxation
Rapid Inflow
Diastasis
Atrial Systole
Aortic Valve Closes
Atrioventricular Valve Opens
Aortic Valve Opens
Atrioventricular Valve Closes
AS
IC
ME
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Physiological Analysis of Pressure Wave
RW Corell

Figure 2.0

6.) To Monitor blood pressure under the same conditions that are imposed on the intra-arterial blood pressure method.

7.) To be simple in application and practice.

2.2 Method of Investigation

The method of investigation to be used in the dissertation is outlined graphically on the following page. Upon the establishment of some overall criteria, and with a survey into the field and its' literature, one would then be in a position to consider some possible methods for solving the defined problem. Once the various methods have been isolated, a preliminary analysis of each method is necessary in order to evaluate its potentialities. Based on that preliminary analysis, one method is to be singled out as being the most realistic possibility for potential solution to the problem.

Once a method of approach has been chosen it is obviously necessary to develop the basic theory and analyze the nature of the technique. From the knowledge gained by a theoretical analysis, and in light of the assumptions of the theoretical model, an operational model is to be designed and a program of preliminary development is to be undertaken. Once an operational model is a practicality, a program of evaluation is undertaken to analyze the potentialities and limitations.

2.3 Results of Literature Review and Survey of the Field

To enable the author to become better versed in the field of blood pressure measurement, a program of study was undertaken. During this study, the resources of the medical libraries in the Boston area were consulted. These libraries included Harvard University Medical Library, Boston Medical Library, Boston University Medical Library, Boston Public Library, MIT Science Library, Huggins Hospital Library, the Library of

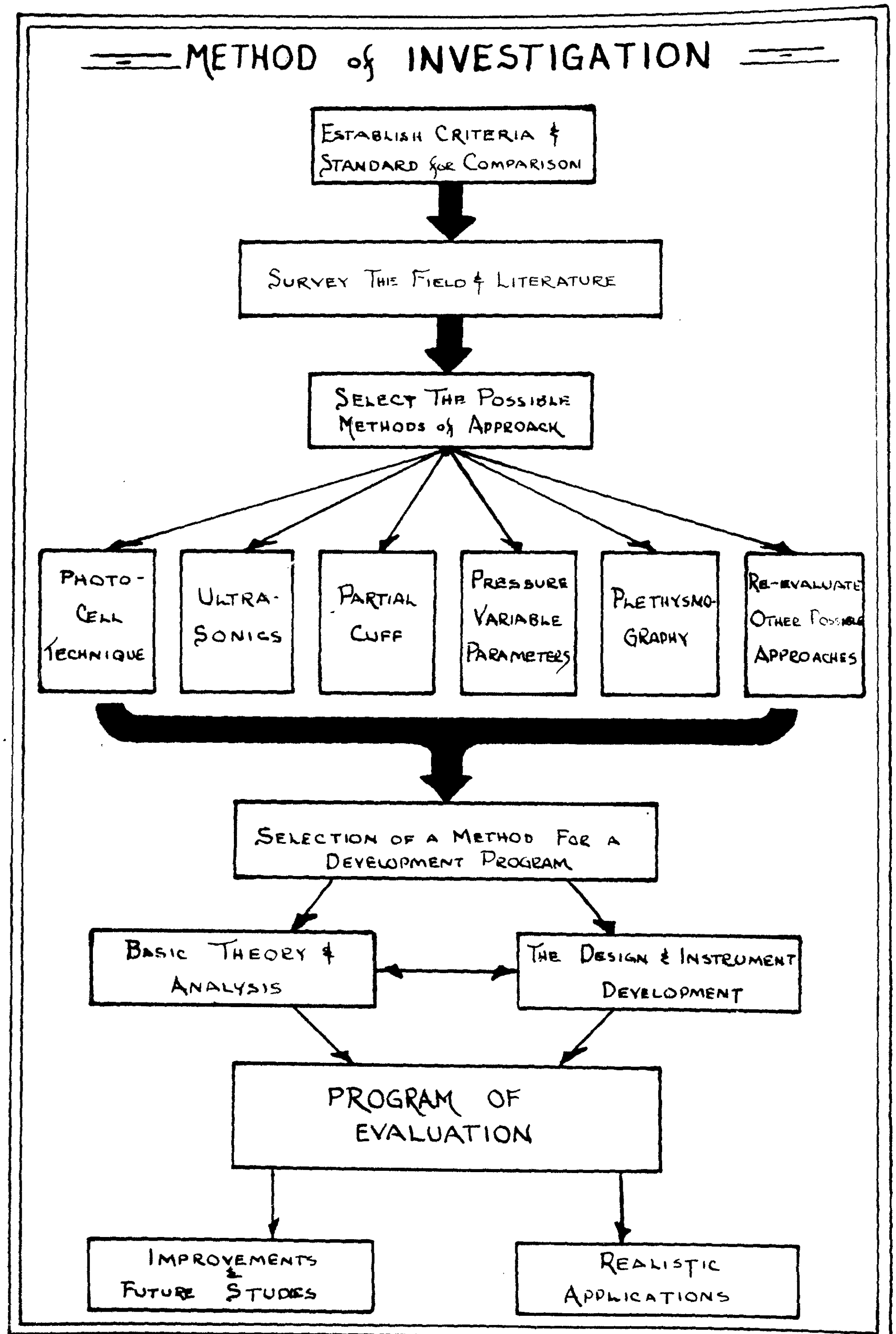

METHOD of INVESTIGATION
Establish Criteria & Standard for Comparison
Survey The Field & Literature
Select The Possible Methods of Approach
Photo-Cell Technique
Ultra-Sonics
Partial Cuff
Pressure Variable Parameters
Plethysmography
Re-evaluate Other Possible Approaches
Selection of a Method For a Development Program
Basic Theory & Analysis
The Design & Instrument Development
PROGRAM OF EVALUATION
Improvements & Future Studies
Realistic Applications

Fig. 2.1

Congress, and some translations from the United States Joint Public Research Service as well as many other governmental and commercial sources. In addition to the preceding sources, personal conversations were held with some of the leaders in physiological research such as Dr. Gerald Mead of Harvard University School of Public Health and Dr. Kurt Lion of the Biophysics Department at MIT. Furthermore, I had the privilege of personal correspondence with some leaders in the field, such as Dr. C. J. Wiggers of Western Reserve University and Dr. Julius Comroe of the University of California.

The benefits gained from this type of search are extremely important, but can never be fully communicated in a dissertation. One can only see some of the minor results in terms of a bibliography, whereas the major results lead one to very important conclusions concerning the state of the art.

Searches through the literature along with the numerous studies in our laboratories, led to the following four conclusions: First, all of the presently available indirect methods for recording of blood pressure are based on arterial constriction in the creation of a jet stream; and they are, at end-point, identical to the clinically applied method of sphygmomanometry. Second, cannulation methods are capable of yielding highly accurate readings, but are subject to the hazards of arterial injury or thrombosis. They require skillful management, often meaning the full time attention of a trained individual. Third, some of the finest, most eloquent applications of scientific knowledge known to me have combed these two basic methods for improvement, until there are no currently visible areas for further development. Fourth, some new approach to the problem must be found.

Serious consideration of the problem, leads one to look realistically into six major areas of possibility. A preliminary evaluation program was undertaken in each of these six areas to ascertain which one or more might yield a realistic solution.

These six areas are:

1.) Photoelectric techniques.

2.) Ultra-sonic measurements.

3.) Partial cuff techniques, based on low frequency pressure waves.

4.) Chemical or physical pressure variable parameters.

5.) Plethysmographic techniques.

6.) Re-evaluation of some of the other previously developed approaches, similar to some of the devices mentioned in the historical review.

Each of these areas will be discussed individually.

2.4 Photoelectric Technique

Another research program with which the author has been associated, is a program to assay the clinical condition of patients, particularly the effects of arterial oxygen content. One method of analysis in the problem of oxygen sufficiency is the use of photoelectric oximetric procedures which yield very prompt and highly reliable information about arterial oxygen saturation. Oximetry (50), the name given to photoelectric techniques, is a reliable means for detecting immediate arterial desaturation. The response characteristics of the photoelectric transmission oximeter (51), are remarkably sensitive to minute changes in oxyhemoglobin.

The basic photoelectric oximeter, (photograph P.2) - after Milliken and Wood (52), consists of a constant intensity light source and two receiving photoelectric cells. One of these cells is sensitive to changes in light transmission above 600 millimicrons (red light and up), whereas the other is sensitive to changes in optical density due to transmission in the region of 800 millimicrons (infra-red). The characteristics of these two cells are quite different. In the broad sense, the infra-red cell "sees" only volumetric changes in the ear, whereas the red cell "sees" a change in the volume as well as change in "color" of the oxyhemoglobin; and thus, the electrical potential difference between these two outputs is a measure of the oxyhemoglobin content.

The actual fundamental characteristics of the photoelectric cell are well known, but the physiologic function portrayed by the photoelectric cells are not yet fully understood. It is clearly evident from our studies, as well as those of Schotz and Birkmire (49), that the photoelectric cell output is much more indicative of physiologic status than just the recording of blood oxygen saturation.

The output of the infra-red photoelectric cell alone, is frequently referred to as the IR pulse. It has a strange similarity to the tracing registered by an intra-arterial cannula.

The infra-red pulse, by virtue of its physical similarity to an intra-arterial tracing, suggests a method for recording of blood pressure. Its output signal originates in the infra-red cell of the Millikan-Wood earpiece of the oximeter.

During our oxygen saturation studies, it was noted that not only did we obtain a measure of the oxygen content, but also we noted a minute variation of the oxygen saturation output signal. This variation was found to be in phase with the intra-arterial blood pressure. This effect can be seen in Figure No. 2.2. From Figure No. 2.2 it would appear that there

P.2

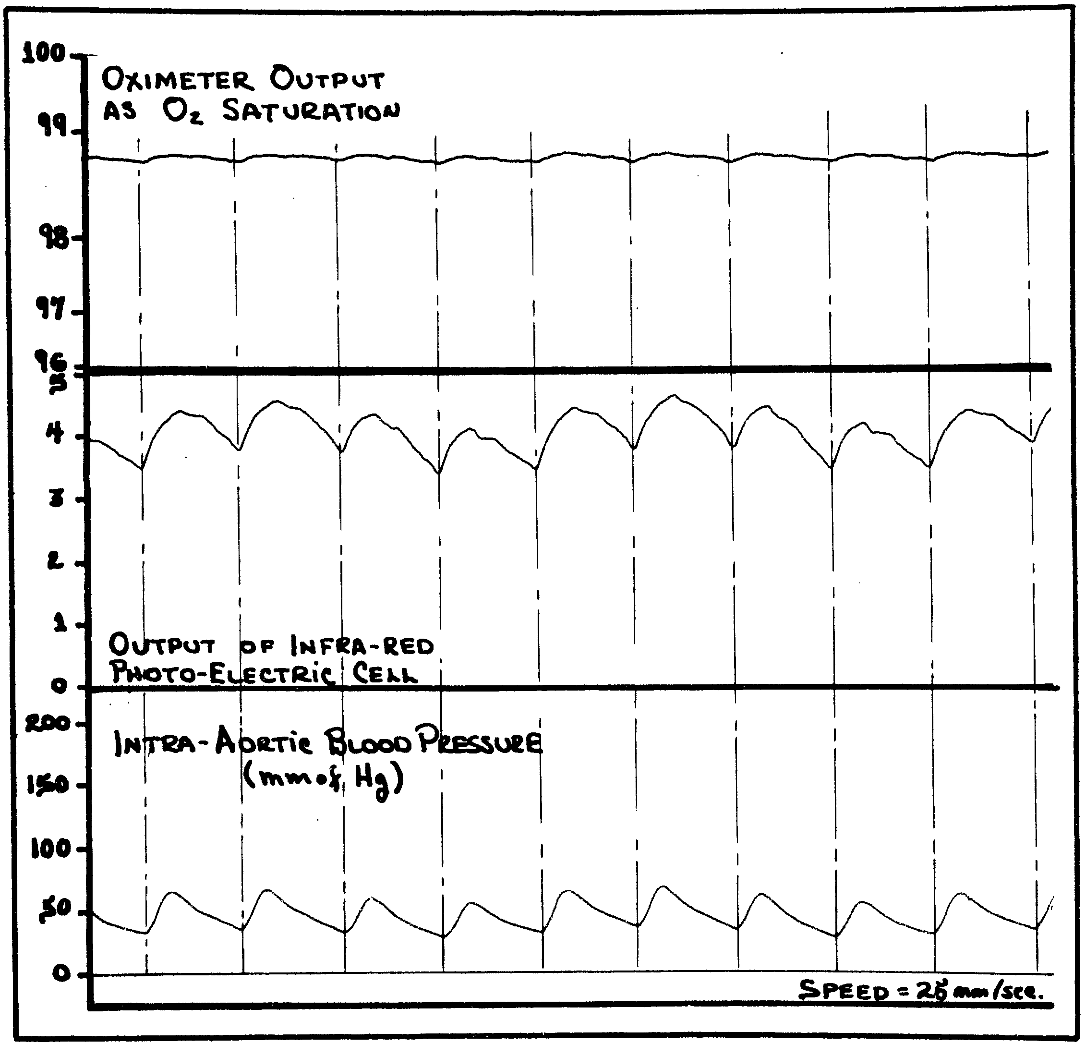


Figure No. 2.2 - This tracing, taken from Case No. 5796, shows the phase relationship between the output of the Oximeter in total and the intra-aortic blood pressure.

exists some direct relationship to blood pressure as measured within the human aorta. This fact led to the program of study to find the nature of this relationship (See Appendix C - Part 1). The infra-red cell measures the optical opacity of the pinna of the ear. The question as to what affects the optical opacity is not clear. The opacity could be affected by both chemical and physical pheononmena. Since the physiological nature of the cardiovascular system is extremely complex, it was decided that an experimental approach would yield more realistic answers. It was believed that the major effect upon optical opacity was due to internal blood pressure. The question that remained to be answered was, what is the relationship between blood pressure and the I.R. Pulse. From the initial observation, it appeared that a direct relationship existed between I.R. Pulse and blood pressure. The possibility was so provocative that a program of statistical comparison between the infra-red pulse and the actual intra-arterial pressure was undertaken. Recordings from many operative cases were subjected to detailed analysis. Some of the data from these analyses are included in Appendix C. The first series of analyses yielded very provocative information. In these cases, the patients were under operative conditions which were not too serious. Preliminary data suggested a very interesting correlation between the signals, the I.R. Pulse and blood pressure.

Upon more extended investigation, however, it was noted that there were marked discrepancies under variable clinical conditions; this required collateral study into the nature of the infra-red pulse. One noted that the infra-red pulse was always in phase with the blood pressure, however, the amplitude of oscillations varied from case to case. It sometimes varied for an individual to such an extent that the direct correlation was proved to be invalid, and not reasonable as a method for recording of blood pressure.

Further study (A synopsis of a preliminary study is contained in Appendix C, Part 2) of the I.R. Pulse as a monitor yielded information which suggests that the infra-red pulse is a method by which one can evaluate circulatory function in terms of peripheral diffusion, and possibly in terms of cardiac output. The I.R. Pulse has proved to be an informative prognosticator, as well as an astonishingly accurate physiological monitor of cardiac and respiratory function. The potentialities of any instrument which indicates the effectiveness of cardiac function is obviously great. Others (49) have drawn similar conclusions about the nature of the I.R. Pulse, and strong evidence has accumulated to show the direct correlation between the I.R. Pulse and the effectiveness of the cardiac function. Even though the I.R. Pulse proved to be invalid as a method for recording of blood pressure in itself, the output signal has provided a new avenue for study of the fundamental problem of monitoring circulatory function with high sensitivity, considerable accuracy, and a promise of being eventually more revealing in its methods than any monitoring signal yet known.

2.5 Ultra-Sonic Measurements for Blood Pressure Indication

The "speed of sound" within a fluid is directly related to the bulk modulus of compression for that particular fluid. The bulk modulus of compression is defined by:

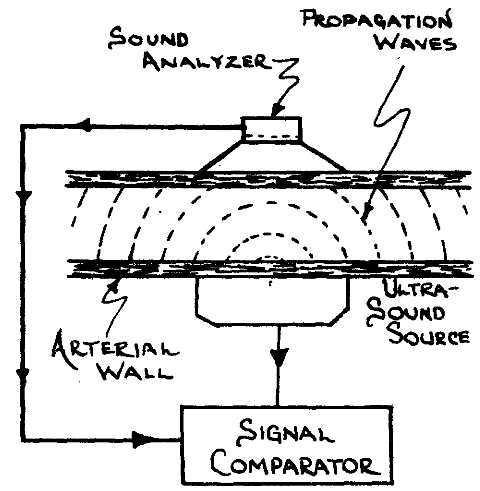
$$\beta = \rho \frac{dp}{d\rho}$$

p = PRESSURE
ρ = DENSITY
β = BULK MODULUS

and in turn, the speed of sound or the velocity of pressure wave propagation is defined by:

$$C = \sqrt{\beta/\rho}$$

Since the propagation speed is related to the $\left[\frac{dp}{d\rho}\right]^{1/2}$ (for a rigid system), one wonders if this fact might provide an avenue of approach for the indirect measurement of blood pressure. Figure 2.3 schematically shows the basic concept behind this idea. An ultra-sonic sound source would emit small pressure perturbations which would be received by the sound analyzer at a fixed distance from the source. The time required for this propagation wave to reach the analyzer would be a direct measure of the speed of sound (c). The time lapse would be obtained by the signal comparator. Based on this type of configuration, one might very well be able to correlate the speed of sound to the blood pressure.



Blood Pressure Measurement Based on Ultra-Sonics

Figure 2.3

Considering blood to be similar to water, one is led to a false conclusion about the potentialities of this idea. Water has a bulk modulus of approximately 300,000 psi. and the density varies by one part in a million under a change in pressure of 100 mm of Hg. and at a temperature of 100° F (98.6). Obviously one is led to believe that the associated variation in (c) is almost immeasureable. This type of order of magnitude analysis is not complete since in the cases of blood, $\left[\frac{dp}{d\rho}\right]^{1/2}$ is not a measure of the speed of sound. In water, the speed of wave propagation is of the order of 5000 ft/sec. For blood, in a non-elastic system, (53), the propagation velocity is approximately 1000 ft/sec. The 20 ft/sec velocity of propagation within the vascular system is a complex function of numerous variables. First the blood is a heterogeneous mixture of fluid, solids and gases. This complexity of phase, greatly effects the nature of the fluid. Furthermore, the velocity of the arterial pulse is largely effected by the elastomeric nature of the arterial walls. (54)

The following considerations are based on data for the velocity of wave propagation for the arterial blood pressure, particularly with reference to aortic studies. King (55) found from a theorectical analysis of the arterial system that the velocity of wave propagation increases with the mean blood pressure within the artery. His theory is based on the assumption of thin-walled elastomeric tube containing a non-viscous homogeneous fluid. His theory leads to conclusions that are consistent with the qualitative results from experiments.

Experimental results by Hallock and Benson (56) not only indicate that the propagation velocity varies with pressure, but that it does so in a nearly linear manner. The slope of the wave velocity-pressure curves is of the order of 3 meters/sec/50 mm of Hg. For the pulse-wave velocity in an elastic system, this is a variation of 25%; however, for the case of the non-elastic system it is much different. Based on the studies of Parnell (53), the non-elastic system would be of the order of 0.3%.

Furthermore, the effects of chemistry can appreciatively alter the velocity of propagation. No real data is available as to the effects on blood, but from the effects of water one can draw some conclusions. Water @ 100° F and 14.7 psi, and air-free has a speed of propagation of 4938 ft/sec. By just exposing the water to the open atmosphere, the speed of sound is altered by 1.5% (57). Furthermore, a salt solution at constant temperature and pressure has a speed of sound variation of 4% for an increase in concentration from 10% to 15%. (57) During most operative procedures patients receive saline solutions that may contain from fractions of 1% to 3% NaCl. Thus a very significant alteration in salt content can result.

The "sound" source must have a frequency that is high enough to "follow" pressure variations as high as 100 cps. It would appear that the source frequency would have to be at or just beyond the threshold of human hearing, thus of the order of 20,000 cps. Gregg (59) states that due to ultra-sonic energy sources, biologic effects can be major. In fact, ultrasonic energy sources are used for cell manipulation for the treatment of some diseases. This treatment utilizes the heat generated by the hysterisis.

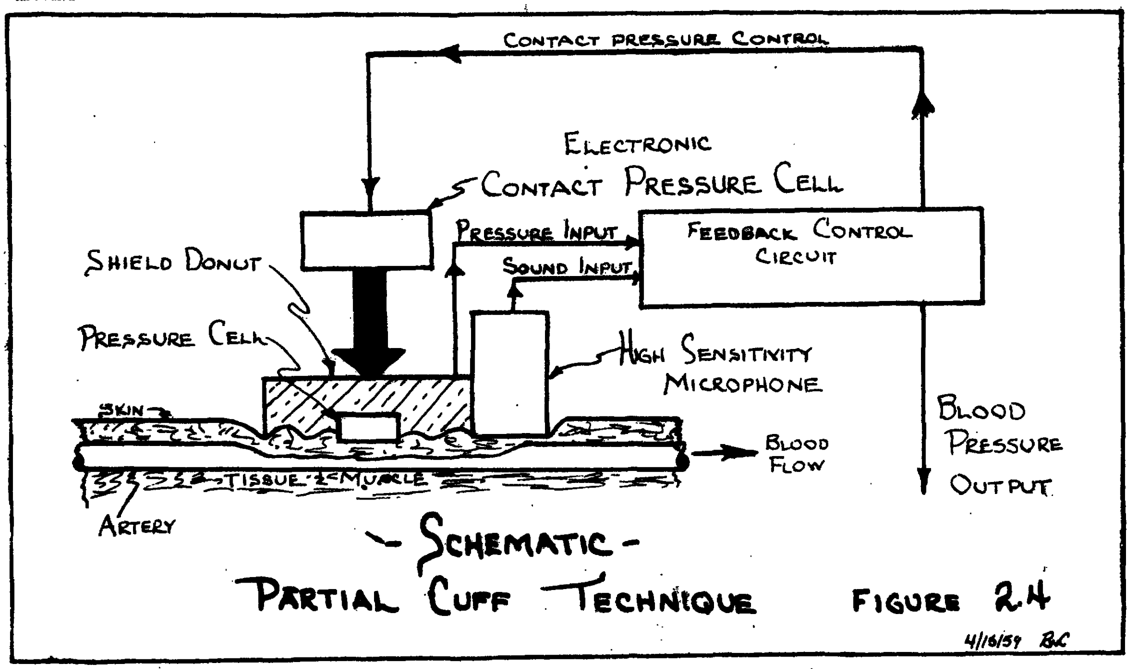
For some of the reasons contained herein, it was felt that though this system might very well be a possible method for indirect measurement of blood pressure, its complexities would appear to limit the potentialities. The most limiting factor seemed to be the need for arterial exposure, and that by our criteria this would not be acceptable. Though we did not explore this technique to any great extent for this reason, it still might very well have real possibilities for future development and exploration.

Still another application of the same basic concept, would involve the direct measurement of pressure wave velocity by employing differential tambors to a pulsative artery. Studies by Klip (60) indicate that this is a reasonable method for measuring wave velocity. Unfortunately the system would only yield data regarding mean blood pressures. For this reason, no further consideration was given to this idea.

2.6 Partial Cuff Technique

Like investigators who preceded us, we felt that the Korotkow method still had some reasonable possibilities. The logic that is basic to this particular application of the Korotkow method, is the elimination of the total cuff and the application of a partial cuff.

The concept of the system is simple in theory. The partial cuff is a circular donut surrounding a pressure cell. The cuff is placed over an artery of a patient and a force is applied to the cuff to increase the

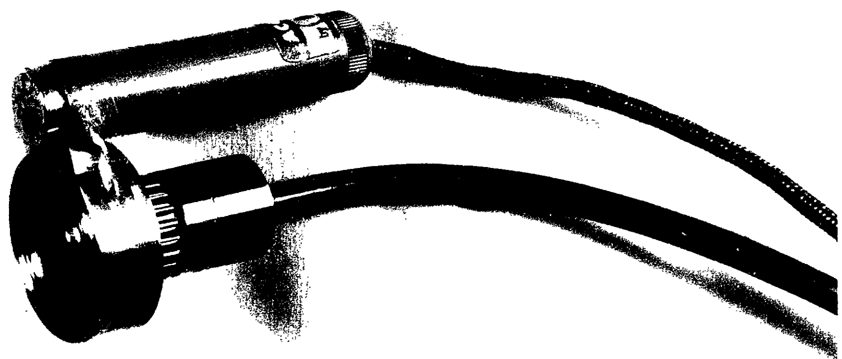


- Schematic -
Partial Cuff Technique Figure 2.4

interface pressure between the cuff and the arm. During this time, the artery is slowly constricted until the flow of blood ceases. It is theorized that the interface pressure will be just equal to the internal blood pressure when the flow is just constricted. The interface pressure will vary somewhat over the surface of the partial cuff, however, the variation will be minimal at the center of the cuff. This is the reason behind the incorporation of a pressure cell at the center of the cuff. The interface pressure variations will become significant toward the outward edges of the cuff. These variations are not important since the system is insensitive to these magnitudes and directions. The important factor is that the interface pressure over the surface of the pressure cell must be constant and normal to the cell face. If this is true, then the pressure recorded will be the pressure to which the artery is subjected. The characteristics are monitored by a highly sensitive condenser microphone. As the occluding interface pressure drops just below the blood pressure, a high velocity jet stream is created; associated with it is a low frequency audible sound. This sound can be heard by a stethoscope but is more clearly defined by a sensitive microphone. Similarly at the low pressure point in the artery, the jet stream is ultimately eliminated and the characteristic sound is altered. These two "sound" points can be easily detected by the output of the microphone. The corresponding interface pressure will be the maximum and minimum pressures in the artery.

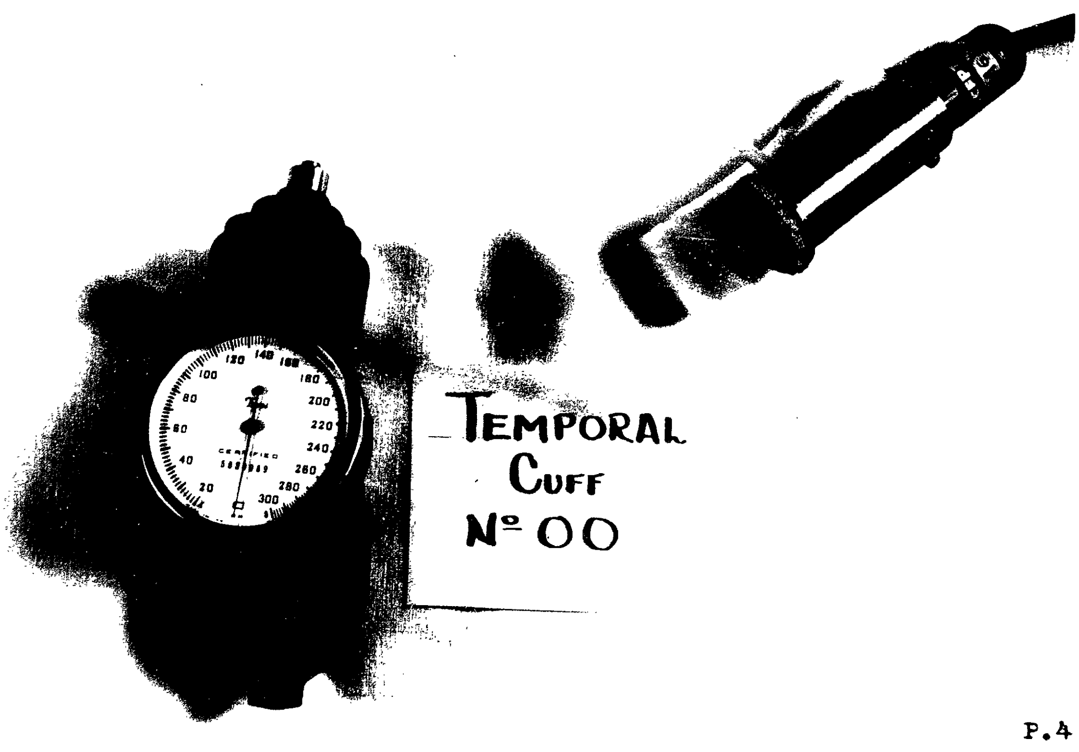
Experimentally the pressures were recorded by the polygraph system using a strain gage type transducer for the pressure cell. (See P. 3) The characteristic sounds were monitored by an Altec Microphone, amplified by a Stromberg-Carlson AV-70 Power Amplifier and recorded on tape for further analysis. The sounds created by the jet stream were visually analyzed on an oscilloscope while the audio effects were followed on a low frequency speaker.

As depicted in Figure 2.4, a system of this nature could be electrically programmed to maintain an interface pressure in accordance to a characteristic sound. A system of this nature would be able to maintain



PARTIAL
CULL

P.3



P.4

a recording of just one level of pressure, either systolic or diastolic. By using two such devices, the maximum and minimum pressures could be recorded.

The total cuff can not be used continuously for the same reason that a tourniquet must be periodically released. The partial cuff appears not to be restricted under certain conditions by this type of problem. The human body maintains cross flow systems such that the tissues are supplied with blood even though a particular artery is partially or totally restricted.

As a preliminary study, it was decided that a prototype unit would be built and tested. The results of a few of the trials are shown in Figures 2.5 and 2.6. This data shows the results of the application of the unit to six different individuals, ranging in age from 3 years to 55 years of age. The pressures attained by this method were compared to total cuff pressures obtained by an individual who had worked his technique against the intra-arterial system. His standard deviation of error on systolic pressures was approximately ± 3.6% and on the diastolic pressures was ± 8.1%, or a total standard deviation of ±5.8%. Using this basis of comparison, the partial cuff technique is equally as accurate as the total cuff technique. The partial cuff system appeared to have possibilities; even though the data shown is fairly reasonable, the system is attended by some problems. First it would require two monitoring cuffs, one on each arm. Secondly, it is extremely sensitive to external noises. Thirdly, the necessary equipment for automatic programing would be extremely elaborate and fairly delicate to operate effectively. Fourthly, it appears to be only as accurate as the total cuff. Finally, it might well be more traumatic than our preliminary studies would indicate.

PARTIAL CUFF TECHNIQUE DATA

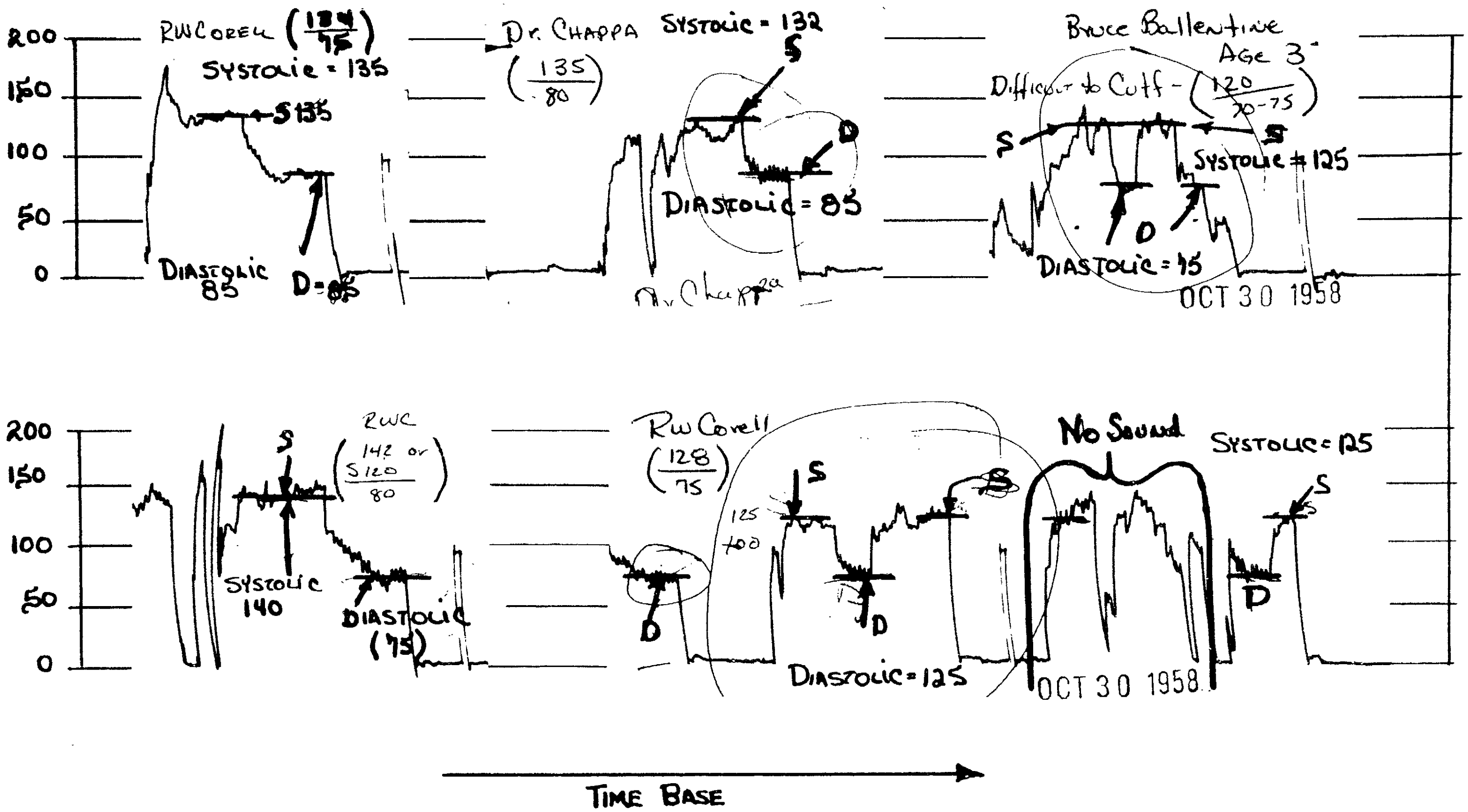


FIGURE 2.5

PARTIAL CUFF TECHNIQUE DATA

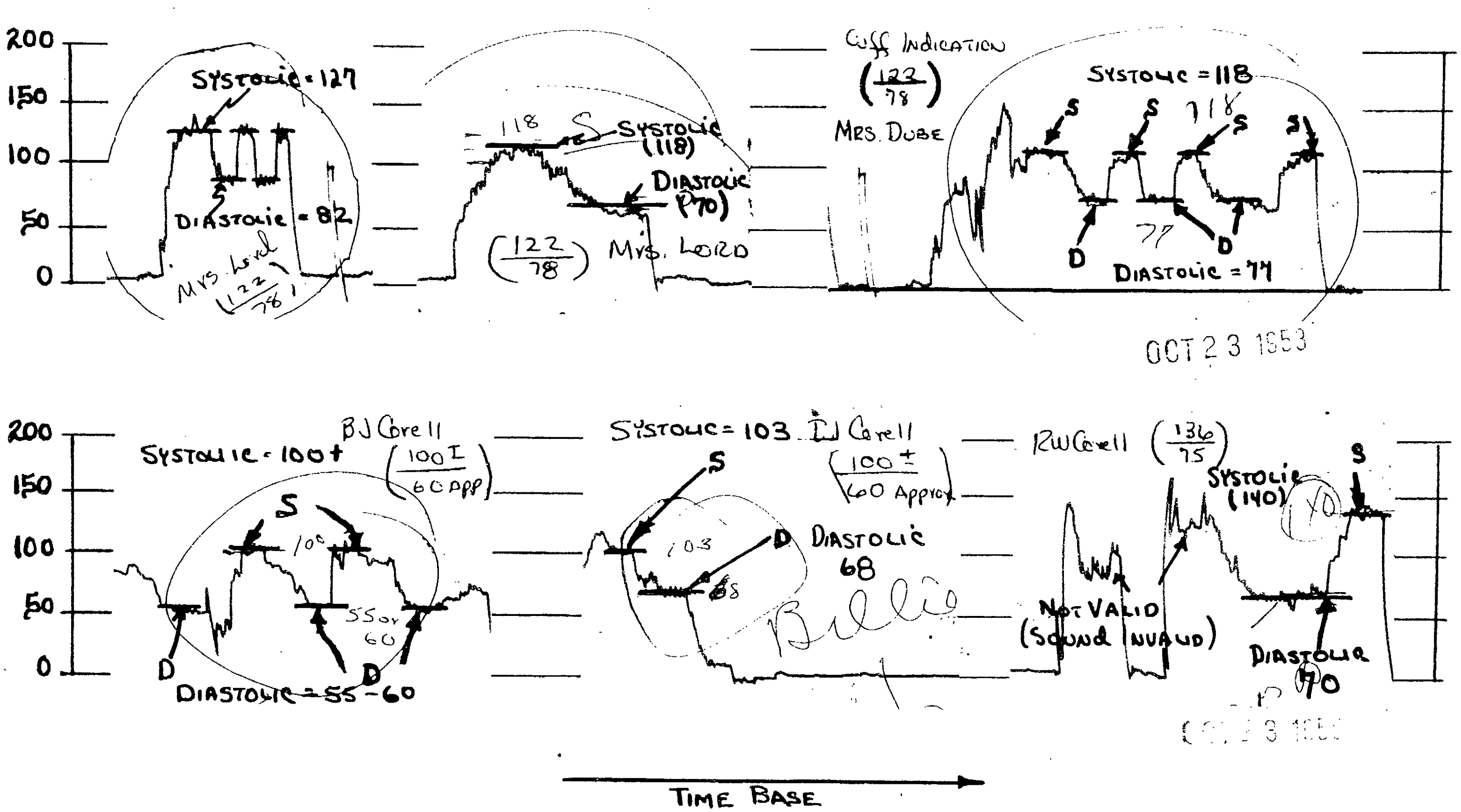


FIGURE 2.6

For these reasons, this system was not deemed sufficiently acceptable for further development, and consequently it was dropped as a possibility. Upon further refinement and development, one might very well find that it has potentialities.

The partial cuff technique was also applied to the temperal artery. A picture of the transducer used is seen in the photograph (P.4). This unit was based on the same theory as the unit above, however, it was applied to the temperal artery instead of the brachial or radial artery on the arm. In practice, the idea was less fruitful than the above, and was more difficult to operate successfully. Further development was considered unnecessary and therefore dropped without full analysis.

2.7 Plethysmography

Plethysmography has been a field in which the medical sciences have worked for some time. (81). Basically a plethysmograph is a rigid chamber surrounding an organ or extremity of the body. This chamber is connected to a volumetric recording device which senses the changes in volume of the extremity due to physiological changes. The system has been used for various reasons. In one case it is used to diagnose certain physiological disorders, such as poor peripheral defusion, organ swelling rate, etc. It was thought by the investigator that since this plethysmographic technique was always oscillatory in nature and closely in phase with the blood pressure, it might be related to arterial blood pressure.

A plethysmograph was built and tried under various conditions. Unfortunately it has many disadvantages. The most troubling is the fact that the chamber is sensitive to all volume changes and not just those changes due to distention of the arteries. This causes the output signal to vary according to all the volume changes and one can not completely sort out the changes due only to artery distention. Under controlled situations, one can obtain tracings like that shown in Figure 2.7. In this tracing one notes the reversal of the wave form. This particular tracing is the result of plethsymographic recording on a finger and it is theorized that there exists an integral effect due to the small capillary flow in the extremity. This is merely a theory, but in a conversation with Dr. Lion of MIT, he tended to agree with this logic. The result remains that the form of the signal is significantly different than the intra-arterial pressure form.

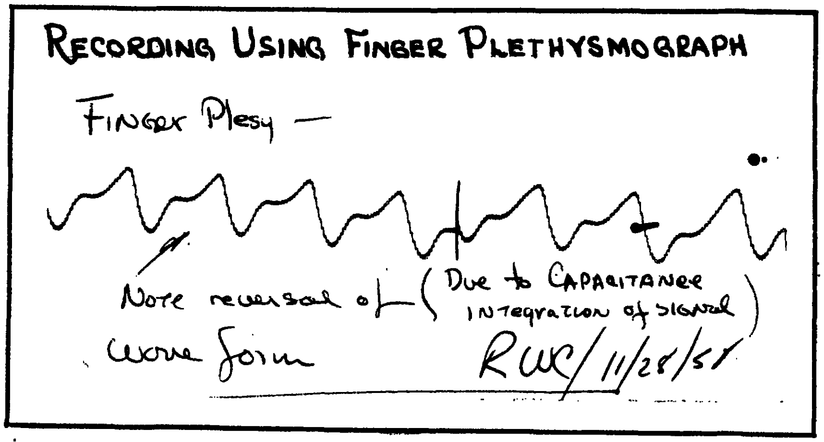


Figure C.7

For reasons that are obvious, this plethysmographic technique was not studied any further.

2.8 Re-Evaluation of Other Possible Approaches

Since the Korotkow method had been used so extensively, we felt that we must re-evaluate its full potentialities. First, we studied the correlation between the intra-arterial pressures and the sphygmomanometer and found that under controlled conditions and after perfection of individual technique, a doctor can obtain reasonable accuracy with the cuff and stethoscope. As indicated in earlier parts of the thesis, many attempts have been made at utilizing this basic concept for indirect recording. Since most of the attempts have only limited application, and since a good deal of engineering know-how was incorporated in those attempts, it was decided to look elsewhere for ideas.

Still another method which might provide an approach to the measurement of blood pressure is the critical analysis of the sound characteristics of the cardiovascular system. It was thought that the sound characteristics might be related to the pressure level within the system. These sound characteristics are a function of the natural frequencies of the body system, and these frequencies might be altered by the pressure level of the system. The frequency analysis of the cardiovascular sounds provides still another possible approach to this problem of blood pressure measurement. Preliminary studies in which we analyzed the frequency characteristics of the vascular sounds on an oscilloscope, strongly indicated that a correlation would be most difficult if not impossible. The db level of the sound and the frequency varied from individual to individual, and within a particular individual, no direct correlation could be noted.

Therefore, only a limited program of study was undertaken, the results of which strongly suggested that this would not provide a fruitful avenue of approach to the problem.

2.9 Pressure Variable Parameters

Since the numerous methods which were investigated proved to be nonproductive, it was thought that possibly there was some parameter within the vascular system that varied in accordance with the blood pressure. If one could measure a parameter of this nature, then a correlation might be drawn between the pressure level and this parameter.

To answer the question as to whether or not such a parameter exists, the author studied numerous sources of physiological information. The main body of this dissertation is based on the results of this search.

2.10 Summary of Preliminary Investigation and Analysis

In the course of attempting to find a satisfactory method for the continuous measurement of blood pressure by indirect techniques. the author studied various possibilities. Studies proved that most of these systems would not provide an adequate solution to the problem based on the knowledge obtainable at the present time. Photoelectric techniques provided one of the most provocative avenues of approach. It has proved not to be a blood pressure technique, but it may provide physiological information which may be of greater importance than blood pressure itself. Other techniques such as the partial cuff, ultra-sonics and plethysmography were deemed unsatisfactory in light of the studies. A system based on a pressure variable parameter was most promising and thus will be analyzed and developed in the remainder of the thesis.

— SUMMARY —

CRITERIA	PHOTO-ELECTRIC	ULTRA-SONIC	PARTIAL CUFF	PLETHYS-MOGRAPHY	OTHER APPROACHES	P. VARIABLE PARAMETERS
1. RECORD CONTINOUSLY ON A TIME BASE.	✓	✓	X	✓	X	✓
2. NON-TRAUMATIC TO THE PATIENT.	✓	X	✓	✓	✓	✓
3. AS ACCURATE AS THE INTRA-ARTERIAL SYSTEM	X	O	X	X	X	O
4. WITH FUNCTION UNDER VARIABLE CONDITIONS	X	O	✓	X	✓	✓
5. WILL MONITOR SOME CARDIAC FUNCTIONS	✓	X	X	✓	X	✓
6. WILL YIELD RESULTS UNDER SAME CONDITIONS AS INTRA-ARTERIAL.	X	X	X	X	X	X
7. SIMPLE TO OPERATE	✓	X	✓	✓	✓	✓

FIG. 2.8

KEY:
✓ MEETS CRITERIA
O MEETS CRITERIA, BUT NOT KNOWN HOW WELL
X DOES NOT MEET CRITERIA

III. ESTABLISHMENT OF A BASIC STANDARD FOR COMPARISON

Before any new method for the registration of blood pressure can be fully understood and evaluated, there must be a standard against which that system can be compared. Over the past fifty years there have been significant improvements in the methodology for direct measurement and recording of pulsatile pressures in the living body. Among the first of the continuous recording devices was the Hürthle (61) recording membrane manometer. Hürthle utilized mechanical leverage to amplify the pressure responses of a dilative membrane. Since that first attempt in 1888, there have been numerous improvements and developments; ranging from the adaptation of optical principles to recording techniques, through the multiplicity of electrical devices up to the electronic devices of the present.

The direct method of recording pulsatile pressures implies the use of some type of cannula which is introduced subcutaneously. This intravascular cannula is often a small lumen polyethylene tubing or a hypodermic needle. The cannula forms a small pressure probe which can sense pressure variations within the cardiovascular system of a patient. The cannula, which forms a quasi-pitot tube, is connected directly to a dynamic pressure sensing device. For the purposes of this study, the dynamic sensing device used was a strain-gage-type pressure transducer manufactured by Statham Instruments, Inc. (Model P23G).

3.1 Test Set-up for Standard Measurements

The physical set-up for the continuous registration of blood pressure is schematically shown in Figure No. 3-A. There are some important considerations in the design of this particular test configuration. There are a number of surgical procedures for the cannulation of arterial sections. The procedure used throughout this investigation involved the direct

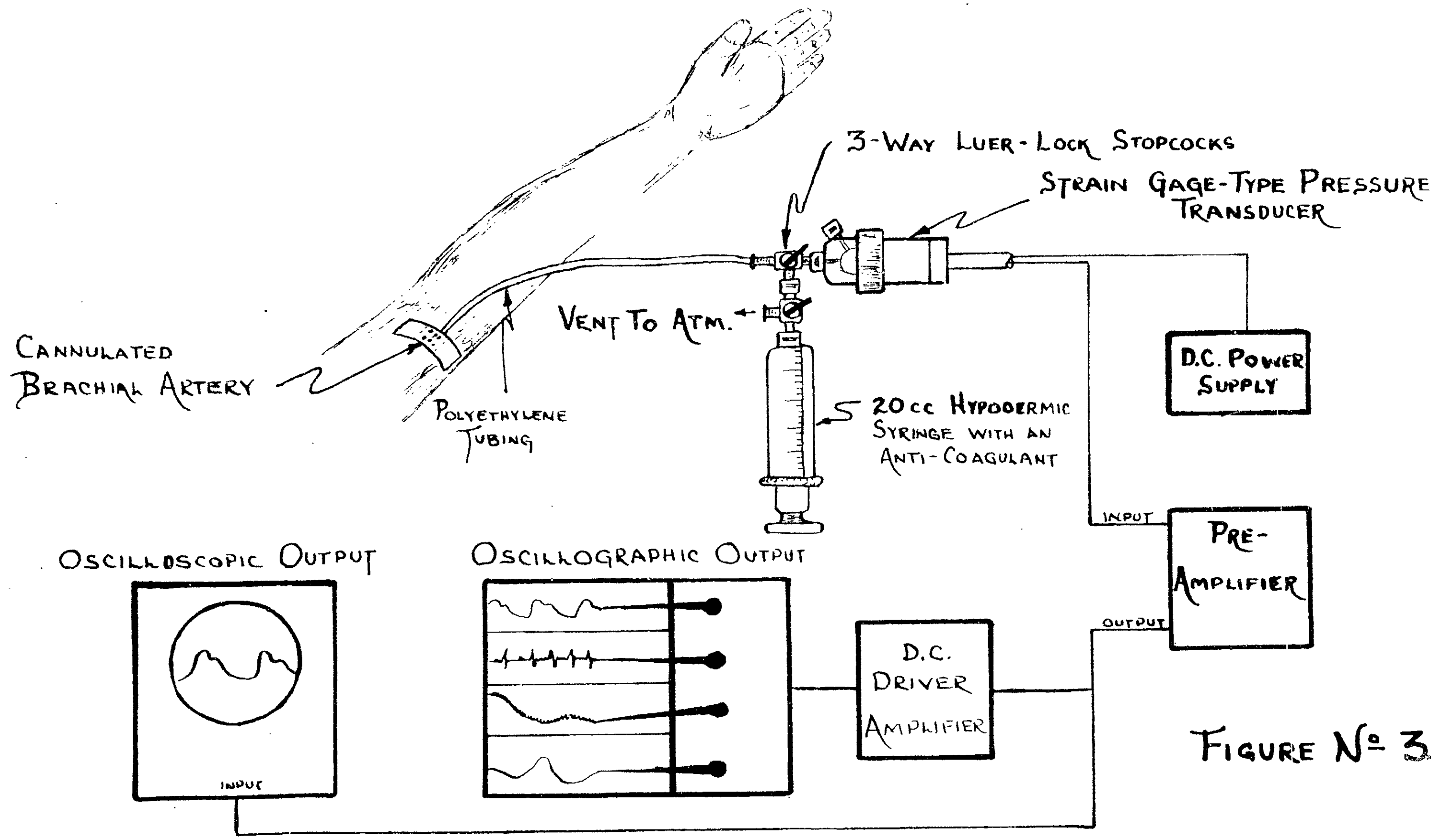
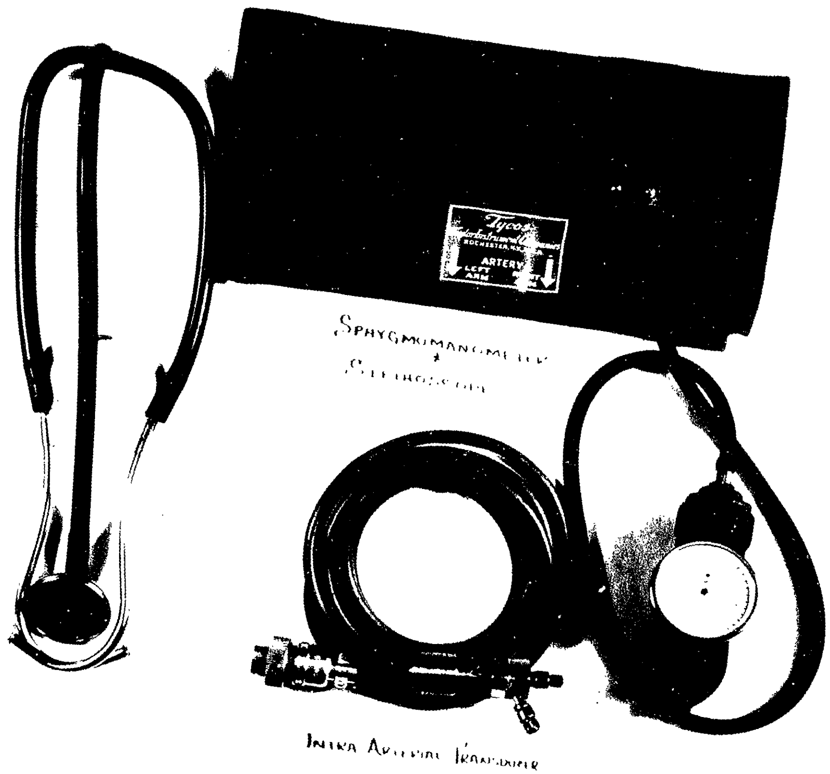


Figure № 31

Continuous Registration of Blood Pressure By Direct Method


Tycos
ARTERY
LEFT ARM
Sphygmomanometer
&
Stethoscope
Intra Arterial Transducer

2.

exposure of the artery to be cannulated. The investigators found that for most accurate results, the artery must be exposed by what is commonly referred to as a surgical "cut-down". With the artery exposed, the cannula then can be placed so that its sensing abilities are not adversely affected by improper location in the artery. Without arterial exposure, the clinician can only randomly position the cannula. Experience has proven that random positioning of the cannula can significantly effect the dynamic characteristics of the pressure sensing probe.

The cannula was secured by surgical procedures to hold it in place and to guard against loss of blood during the recording of the pulsatile pressures. The cannula was connected to a polyethylene tubing which was in turn connected to the input of the pressure transducer through a three-way stopcock. The function of the stopcock was to permit the injection of an anti-coagulant solution like heparinized saline. The function of the anti-coagulant solution was to reduce the clot formation in the cannula system as well as to aid in the preventation of thrombosis and embolism. The solution was introduced into the system periodically to "flush out the system". This is necessary to reduce the viscous clotting which alters the damping characteristics. Intermediary to the anti-coagulant reservoir (20cc Syringe) was another stopcock so that the system can be vented to atmosphere for calibration purposes.

The excitation source for the transducer was a D.C. power supply. Since the transducer was basically an unbonded strain-gage-type unit. this power supply excites a Wheatstone bridge composed of the four strain-gage components. Obviously, the transducer relies upon the modulus of elasticity of the force-summing diaphragm to transfer the diaphragmatic dilatation into a change in electrical resistance of the strain-gage elements.

The potential unbalance of the Wheatstone bridge, caused by the resistance changes due to pressure variations, was fed into a direct-coupled pre-amplifier for power pre-amplification. The output signal of the pre-amplifier was fed directly into a Type 530-Series Oscilloscope manufactured by Tektronix, Inc. (For characteristic see Appendix A). Furthermore, the output of the pre-amplifier was power amplified by a Model 5 D.C. Driver Amplifier manufactured by the Grass Instrument Company. The output of the Driver Amplifier controlled a Model 5 Polygraph oscillograph, for pen and ink registration of the signal. (For characteristics see Appendix A).

The test configuration allows for oscilloscopic or oscillograph registration of the pulsatile pressures. The oscilloscopic output was used because of possible frequency limitations of the oscillograph, since the investigators are of the opinion that there are frequencies within the blood pressure wave form that can exceed an estimated output of 100 cycles per second. The oscillograph may not adequately follow these frequency variations, so it was deemed necessary to study the wave form characteristics on the oscilloscope.

3.2 Validation Under Static Conditions

A transducer-electronic system is a secondary standard; however, it can be calibrated against a primary standard, such as a mercurial manometer or an oil dead-weight tester. It should be noted that a calibration procedure of this nature is purely statistical. It has been the experience of the manufacturer (62) that the repeatability of the instrument is of the order of ± .1% and the non-linearity and hysteresis error is ± 1%. The calibration curve taken from our instrument is shown in

Figure 3.3.

This calibration is a total input-output calibration of the recording system as depicted in Figure No. (3.3). All readings were within the readability of the output signal of the Polygraph. The accuracy and precision of the transducer is better than the readability of the polygraph output. The readability of the polygraph is of the order of $\pm$ 1.0 mm of Hg and this is within the desired accuracy of the test.

3.3 Dynamic Characteristics of the Recording System

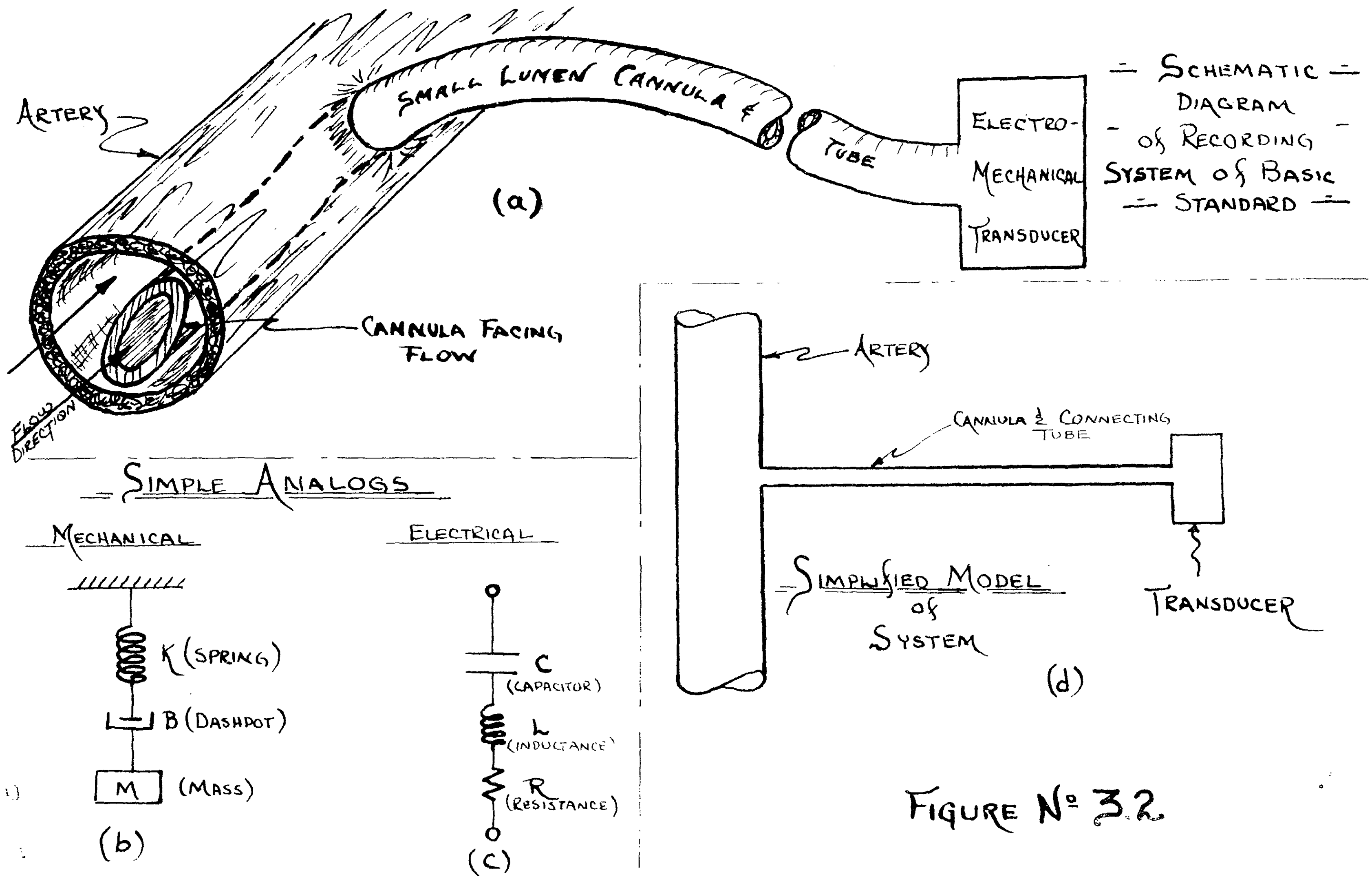
In the continuous registration of the pulsatile pressure of the cardiovascular system, it is necessary to evaluate the standard system under dynamic conditions. It is commonly accepted that the strain-gage-type pressure transducer is an acceptable device for recording pressure variations. The question that must be answered is whether it is an acceptable device under the conditions of this investigation.

The system which is under analysis is schematically represented in Figure No. 3.2a. (Simple analogs can be seen in 3.2b and 3.2c) It consists of an artery through which a fluid is flowing under variable pressures. The operating pressure range is of the order of 35mm of Hg to 250 mm of Hg.

For the purposes of analysis, a simplified model was chosen as indicated in Figure No. 3.2d. This simplified system can be thought to behave somewhat like the mechanical analog in Figure No. 3.2b which consists of a spring, damper, and mass system. Under sinesoidal forced vibration with viscous damping, the differential equation of motion is of the form

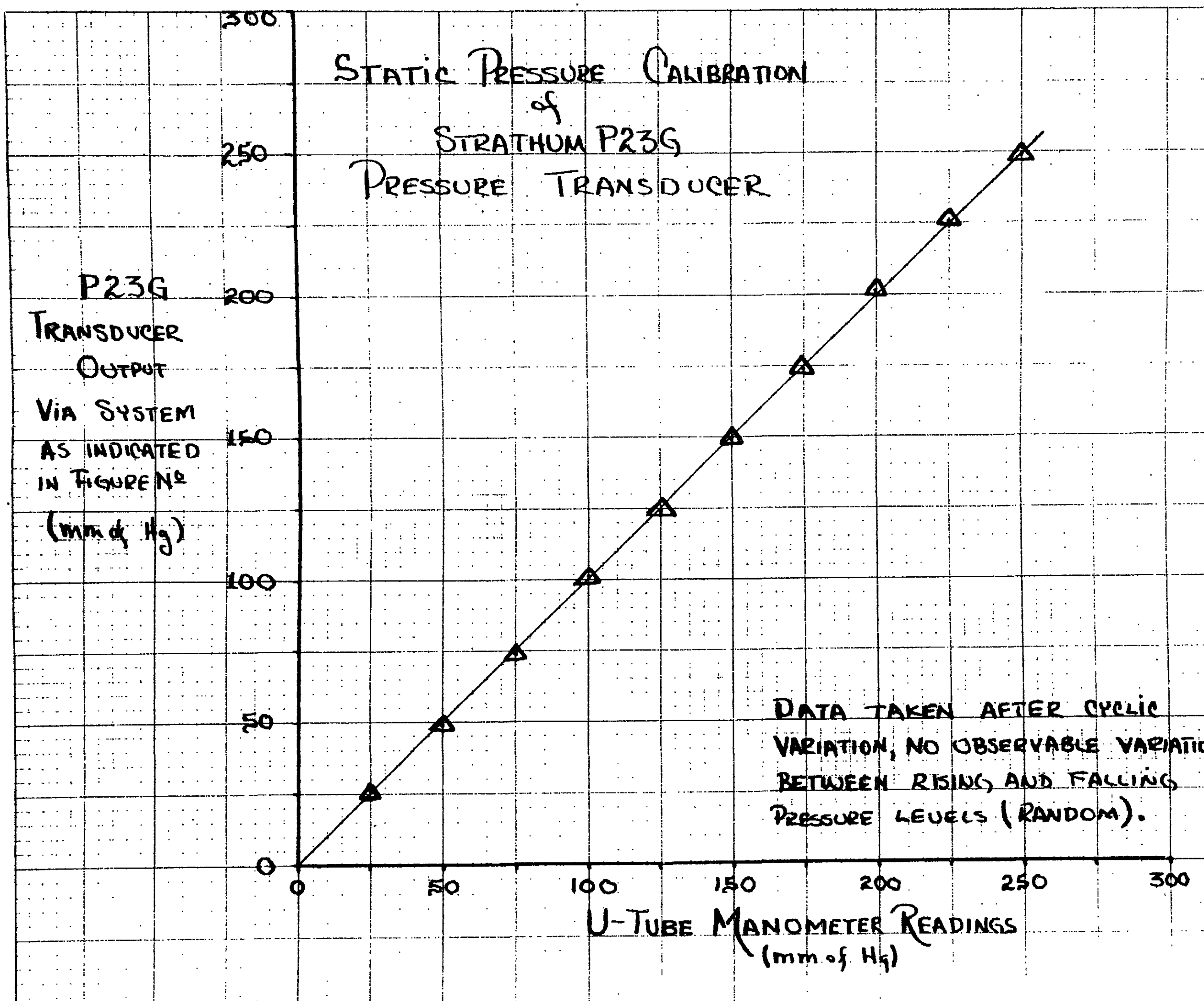
$$m\ddot{x} + B\dot{x} + Kx = P_o \sin(\omega t) \qquad (3.1)$$

where $P_o \sin(\omega t)$ is the forcing function. (M) is the total effective mass of the portion of the system in motion. (B) is the effective



— Schematic —
Diagram
— of Recording —
System of Basic
— Standard —

Figure Nº 32



DATA (TAKEN IN mm of Hg)	
U-TUBE	P23G
0	0
25	25+
50	50-
75	75-
100	100+
125	125
150	150
175	175-
200	200+
225	225+
250	250-

NOTE: ALL READINGS WITHIN ABILITY TO READING FROM THE GRASS POLYGRAPH.
(DATA: 2/21/59)

FIG. 3.3

RW CORELL

damping coefficient of the viscous forces, and (K) is the effective spring constant of moving system. The undamped natural frequency is:

$$f_n = \frac{1}{2\pi}\sqrt{\frac{K}{m}} \qquad (3.2)$$

whereas the damped natural frequency is of the form:

$$f_{nd} = \frac{1}{2\pi}\sqrt{\frac{K}{m} - \left(\frac{B}{2m}\right)^2} \qquad (3.3)$$

It is obvious that if the damping coefficient (B) is relatively small, the natural is approximately the same for damped and undamped motion. An analysis of this sort is very limited and over simplified since the variables of the system are difficult to measure and often the differential equation is non-linear due to variations in the coefficients. This makes the theoretical analysis most difficult, even though in theory the physical analysis is quite simple, a mathematic solution is beyond the scope of this dissertation (65-67).

The natural frequency and the damping ratio are the factors that completely define the dynamic characteristics of a mass subjected to a linear restoring force, viscous damping and constraint to a single degree of freedom (63-64). The response characteristics of an instrument as supplied by a manufacturer can be completely meaningless for a particular application since the instrument characteristics are generally taken independently of the system to which it might be applied. The natural frequency of a transducer alone in air can be calculated to a fairly good approximation by equation (3.2). This same transducer when connected to a system consisting of tubing and fluid of greater density than air, will have a natural frequency significantly lower than equation (3.2) would yield. In fact, the mass of the transducer may become insignificant in the determination of the natural frequency of the new system.

The calculation of the natural frequency and the damping ratio of a liquid-filled elastic line plus a transducer system is usually very difficult. Even in a simplified analysis (65) it is necessary to know precisely the parameters of the pressure transducer and the fluid connections. Some parameters of the liquid connections (i.e., density, temperature, diameter, tube length, etc.) are usually known. However, in the case of blood, other parameters are most difficult to evaluate, for example, the viscosity and isotropy of the blood, in fact, blood is an isotropic in behavior. The transducer is also hard to evaluate completely on physical grounds; the spring constant, the <u>effective</u> area of the diaphragm, the mass of the <u>effective</u> diaphragm, etc. are difficult to obtain or may even be unknown. Even though there are difficulties in an analysis, there have been some very good theoretical analyses of pressure systems. Iberall (66) developed a theory which yields design parameters for system design. His theory accounts for compressibility, finite pressure variation, fluid acceleration, and finite tube length. He also considered heat transfer effects. Barton (67) developed a simplified theory which yielded natural frequencies and damping ratios as a function of statistically determinable parameters. Parnell, Beckman and Peterson (53) studied various transducer systems and the response characteristics of each. They found transducers which when studied alone in air, had frequency responses of the order of 1000 cycles per second. While under the influence of a liquid filled system, the response varied by a factor of 100; namely down to 10 cycles per second.

In the data provided by Stratham Laboratories (See Figure No. 3.4) on a transducer similar to the one used in this study, the natural frequency is approximately 180 cycles per second, using a system consisting of a #20 hypodermic needle 2 inches long with the system filled with distilled water. It can be noted that the response is flat up to 25 cycles per second. The same transducer connected to a #5 cardiac catheter 100 cm long reduces the natural frequency to 47 cycles per second while reducing the flat response to about 10 cycles per second. It is obvious from this data, as well as from the theoretical analysis by Iberall (66), that the tubing length should be as short as possible and the diameter as large as possible in order to retain the best response characteristics. The reason for the use of distilled water instead of blood is obvious. It has been our experience that anti-coagulants in connecting the tubing increases the frequency response above that for blood alone. The major reason for this, other than the elimination of blood from the transfer tube, is the reduction of clot formation in the area of the cannula. By filling the connecting polyethylene tube with heparinized saline, the comparison between the above data and the test becomes realistic, since the heparinized saline is approximately 99% distilled water, the remainder being the heparin and sodium chloride. With short connection tubing, the natural frequency will be of the order of 125 cycles per second, and the damping coefficient will be of the order of .16.

3.4 Some Considerations of the Dynamics of the Cardiovascular System

There are some considerations in regard to the cardiovascular system that must be understood before this direct method of recording the pulsatile pressure is acceptable as a standard for comparison.

The first consideration is the velocity effect as related to the measurement of pressure by conventional pitot tube techniques. Studies

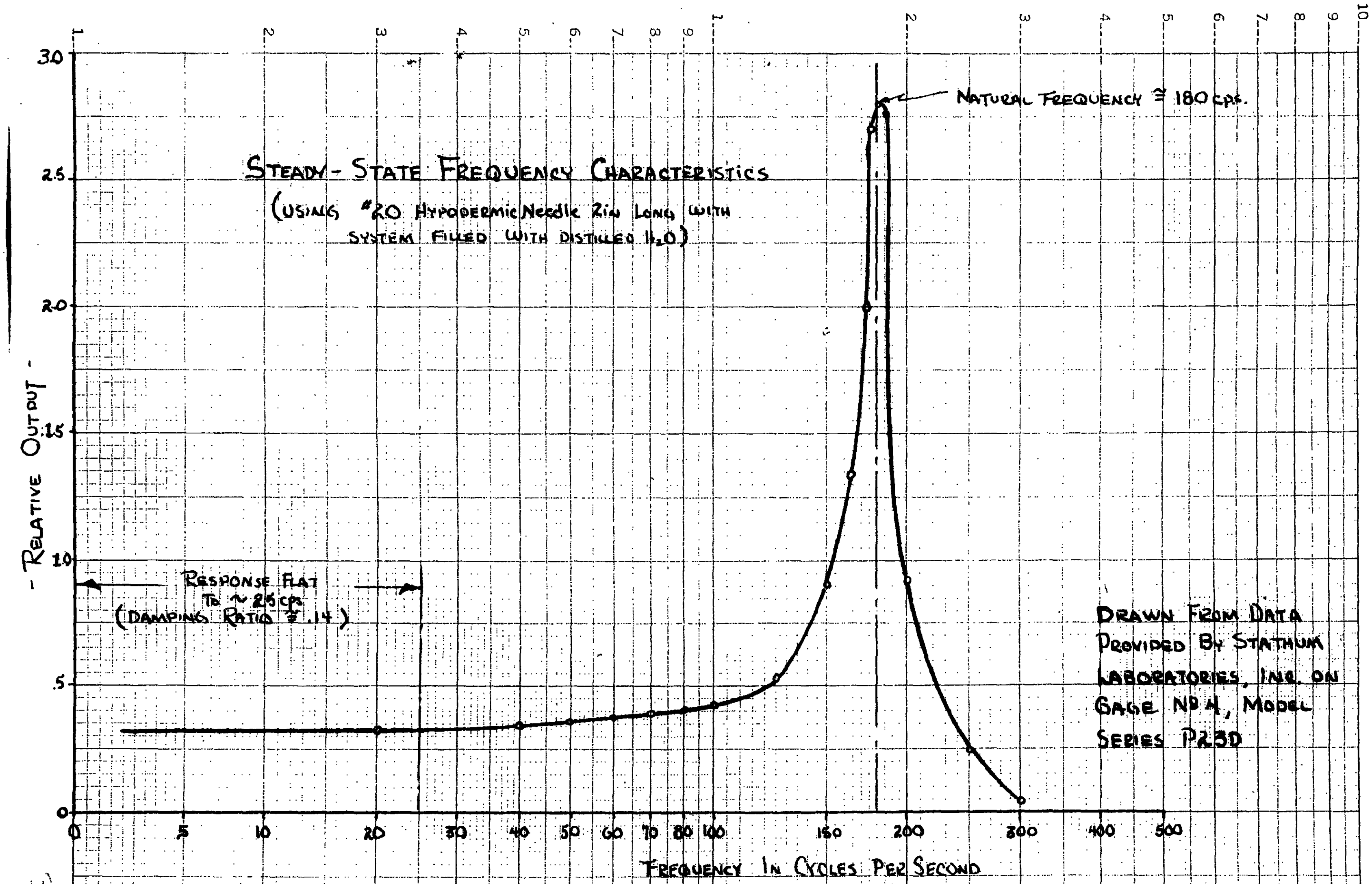

STEADY-STATE FREQUENCY CHARACTERISTICS
(USING #20 HYPODERMIC NEEDLE 2IN LONG WITH SYSTEM FILLED WITH DISTILLED H_2O)
NATURAL FREQUENCY ≅ 180 CPS.
RESPONSE FLAT TO ~25 CPS
(DAMPING RATIO ≅ .14)
DRAWN FROM DATA PROVIDED BY STATHUM LABORATORIES, INC. ON GAGE NO 4, MODEL SERIES P23D
- RELATIVE OUTPUT -
FREQUENCY IN CYCLES PER SECOND
DRAWN BY RWC 4/4/59

FIG. 3.4

by Broemser (68) show that the maximum velocity of the blood occurs in the aorta (just outside the heart in the main supply line of the cardio-vascular system), where it is approximately .65 ft/sec (20 cm/sec). This velocity is not affected greatly until the sharp reduction of velocity in the arteriolar and capillary districts. Since this analysis is dealing with the flow in the larger arterial sections, let it be assumed that the velocity of flow is 20 cm/sec. With a mean pressure of 100 mm of Hg, the total error due to disregarding the velocity pressure is of the order of 0.15%. This is a negligible engineering error.

The next consideration is that of the frequency response necessary to adequately follow the pulsatile pressure variations. Studies by Wiggers (69) suggest that a recording device must respond to the tenth harmonic of the fundamental frequency in order that the continuous registration be valid. To accomplish this, it is necessary to have the natural frequency of the recording device at least three to five times the frequency of the tenth harmonic. This means that the damping ratio must be of the order of (.33) to (.2). If the pulse rate is 120 beats per minute (a liberal approximation) then the natural frequency should be of the order of 60 or 50 Cycles per second. It has been the experience of the investigator, that even better registration can be obtained if the natural frequency is 100 cycles per second or better. This is obviously an empirical result, but it is the firm belief of the author that with a complete wave analysis, characteristic frequencies in excess of 100 cps may very well be present in the wave.

Finally, the wave propagation speed should be considered. Parnell and associates (53) have found that wave propagation speed is of the order of 1000 meters per second. Therefore, the wave length at a frequency of 100 cycles per second is about 10 meters. Purely from a wave propagation point of view, it is suggested that the catheter length be shorter than 1 meter for minimal phase lag. The length must be small for reasons of frequency response, thus propagation speed effect will be negligible.

3.5 Summary of Basic Standard Analysis

On a statical basis the recording system was repeatable to within $\pm$ .1% with a total nonlinearity and hysteresis error of $\pm$ 1%. From a hemodynamic point of view, the recording system must have a natural frequency of 70 cycles per second as a minimum. The transducer and fluid system used in the analysis has a natural frequency of the order of 125 cycles per second provided the cannula diameter is of the order of a No. 18 hypodermic needle and the length of connecting tubing is less than 50 cm. Furthermore, the hemodynamics suggest that there are characteristic frequencies up to the tenth harmonic of the fundamental frequency, which for a 120 pulse rate, means frequencies of the order of 22 cycles per second. The recording system used in the analysis is linear within 2% up to approximately 40 cycles per second. This is the system to be utilized as the standard for comparison in all future analysis in this dissertation.

IV. THEORETICAL ANALYSIS OF A PRESSURE VARIABLE PARAMETER

4.1 Background

A study program was undertaken, aimed at a fundamental appraisal of physiological inter-relationships. Two broad questions arise with reference to physiological and technological effects. What are the effects? What are the variables? If one finds a pressure variable parameter, one must then analyze the effects and all possible variables.

The logic behind the pressure variable parameter system is simply this; if a parameter like ph, volume, or density were to vary in a known manner with pressure, then one could possibly measure that parameter with greater ease than one can measure blood pressure. Once knowing the variations of the parameter with time, then a correlation can be drawn regarding pressure.

4.2 The Relationship Between Pressure and Arterial Volume

It has long been known that the majority of the tissues within the body are long chain polymers. The high polymers of body cells are for the most part long protein macromolecules whose structure may vary according to the numerous possible arrangements of their structural units, amino acids. These long molecules occur in a variety of forms, progressing from randomly twisted threads to crystalline networks. In general, the properties of the various tissues are attributed directly to their molecular structure, and it is expected that variations in orientation and crystallinity go hand in hand with variations in function (70).

Proteins form one of the largest groups of naturally existing elastic-polymers, called elastomers. Since body tissues behave as elastomers, it becomes necessary to understand elastic behavior of body tissue. Elastomers are rubber-like in behavior, therefore one can learn of the behavior of tissues by studying rubber-like materials in general.

The kinetic elasticity of rubber can be shown by a simple thermodynamic analysis. The thermodynamics of rubber behavior is analogous to a thermodynamic analysis of a perfect gas. Let us consider a kinetic theory of rubber elasticity by first looking at a perfect gas and then drawing an analogy.

The equation of state for a perfect gas is given by:

$$PV = RT$$

P - Pressure
V - Total Volume
R - Universal Gas Constant
T - Absolute Temperature
Q - Heat
U - Internal Energy
S - Entropy
W - Work
F - Force
L - Extension Length

The first law of thermodynamics can be written as:

$$\delta Q = dU + p\,dV$$

or rewritten in a partial derivative form it is:

$$\delta Q = \left(\frac{\partial U}{\partial T}\right)_V dT + \left[\left(\frac{\partial U}{\partial V}\right)_T + p\right] dV$$

But through the second law for a reversible process, the equation can be rewritten in the following form:

$$ds = \frac{1}{T}\left(\frac{\partial U}{\partial T}\right)_V dT + \frac{1}{T}\left[\left(\frac{\partial U}{\partial V}\right)_T + p\right] dV$$

However: $$ds = \left(\frac{\partial s}{\partial T}\right)_V dT + \left(\frac{\partial S}{\partial V}\right)_T dV$$

and by Maxwell Relationships where these equalities exist:

$$\left(\frac{\partial s}{\partial V}\right)_T = \left(\frac{\partial p}{\partial T}\right)_V \qquad \left(\frac{\partial s}{\partial T}\right)_V = \frac{1}{T}\left(\frac{\partial U}{\partial T}\right)_V$$

yields upon substitution

$$\left(\frac{\partial U}{\partial V}\right)_T = T\left(\frac{\partial P}{\partial T}\right)_V - P \qquad 4.1$$

Where $\left(\frac{\partial U}{\partial V}\right)_T = 0$ meaning the potential energy is zero, which is a definition of a perfect gas.

Now consider the ideal elastomer, where the PdV work is expressed by $dW = -FdL$

Equation 4.1 can be written by an analogy as

$$\left(\frac{\partial U}{\partial L}\right)_T = F - T\left(\frac{\partial F}{\partial T}\right)_L$$

4.2. now

if $\left(\frac{\partial U}{\partial L}\right)_T$ where zero, then rubber would be elastic in a kinetic sense as a perfect gas. Experiments (71) have shown that such is the case over a good portion of the extention curves. Thus a conclusion can be drawn that elastomers for the most part, are kinetically elastic. This type of elastic behavior now can be explained in terms of statistical mechanics. King (72) applied statistical mechanics to elasomeric polymers, from his work one can draw some conclusion about the elastic behavior of body tissue. Applying statistical mechanics to the behavior of cylindrical tubes under conditions where the radius and the length are changed by the same factor, King obtains the following relationship for the effect of pressure on the volumetric distention of elastomeric cylindrical tubes:

$$P - P_0 = B\left[\left(\frac{r_0}{r}\right)^{\frac{1}{2}}\left\{L^{-1}\left(\beta\frac{r}{r_0}\right)/L^{-1}(\beta)\right\} - \left(\frac{r_0}{r}\right)^{\frac{B}{2}}\right]$$

where:

$B = e_0\, p_0/2r_0$, $\beta = q_e e_0$, L^{-1} = Langevin Function

p = internal pressure, P_o = Atm. pressure, r_o = Initial radius

r = radius variable, g_e = molecular constant, e_o = wall thickness

Even though the properties of the human arterial tissue are anisotropic, it is convenient to study a theoretical model where the structure is isotropic, the wall thickness is uniform, and the material is homogeneous. Under these conditions, one can apply King's formula to arterial sections. King (54) found that using appropriate values in the formula for the age dependent contants, that he was able to obtain some characteristic curves. (See Figure 4.0)

Upon further investigation, King found that the theoretical characteristic curves agreed with experimental results to a remarkable degree of accuracy. He superimposed his theoretical curves on the results of experimental work by Hallock and Benson (See Figure 4.1) and found agreement within approximately 4%.

The following curves are theoretical in natural, but they agree with experimental results. A great deal of experimental work (56, 73-77) has been done on the pressure-volume relationships for human organs, particularly with reference to the cardiovascular system. Included in Appendix D are some of these experimental curves. This includes some work in this area by the author. (See P.5)

The fact that suggests an approach to blood pressure measurement is the surprising linearity of the relationship over a fairly wide range. This range of linearity is well within the range of "normality" for pressure measurements. Furthermore, the only apparent variable is the effect of aging, namely, a change in the slope of the P-V relationship.

Theoretically, these facts present a possibility for utilization

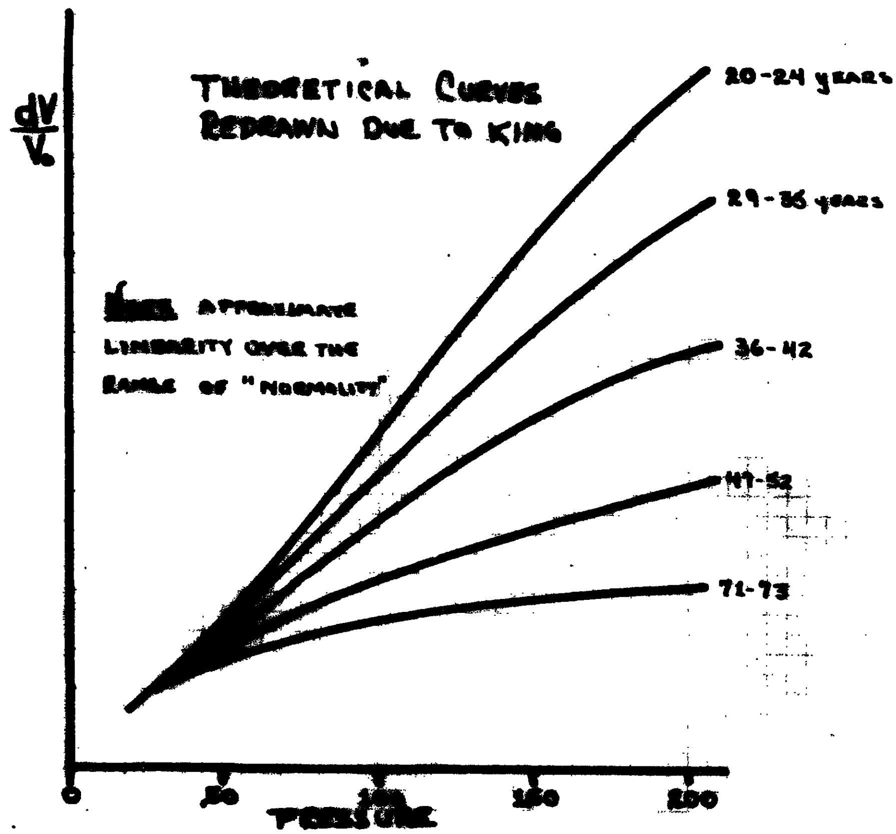


Figure 4.0

in a system for measuring blood pressure. They suggest that, if the change of volume of an arterial section can be measured, the pressure changes within the artery can likewise be correlated in a linear manner.

<u>4.3 Calibration Techniques</u>

A question immediately arises. One may have some knowledge about the general characteristics of the pressure-volume relationships, but what is the precise relationship of a given individual? Put another way, one must have some method of knowing the precise slope of the pressure-volume curve for each individual.

Since a near-linear relationship does exist, then all that is needed is one point on the curve and the slope to be fully defined for each patient. There are two relevant facts which bear directly on this problem.

1. The human blood pressure in an arterial section can be reduced to atmospheric pressure (0 mm of hg.) by proper application of medical techniques.
2. An induced pressure change can be created in the body by basic consideration of hydrostatics; i.e., an elevation change of the arm by 13.2 inches induces an associated pressure change within the artery of 25 mm of mercury.

The above two facts are theoretically sound and have been experimentally verified. First, it is known (78) that the veinous side of the vascular system returns to the heart where the pressures are below atmospheric by a 3-5 mm of hg. In the process of flowing from the capillary network, the pressure drop is of the order of a few millimeters of mercury. This fact that the capillary pressure level is near atmospheric is easily demonstrated. If the cutaneous surface is cut near a capillary network, bleeding is very slow and without significant velocity, therefore the pressures must be approximately atmospheric.

If one were to occlude an artery, then the downstream pressure would assume the level of pressure in the capillary beds, namely atmospheric. This fact has been experimentally demonstrated using an intra-arterial transducer, and every time, without fail, the downstream pressure would be within one or two millimeters of atmospheric pressure during an occlusion.

Secondly, the induced pressure by level variations can be calculated by simple application of hydrostatics.

PRESSURE-VOLUME RELATIONSHIPS OF HUMAN ARTERIES*

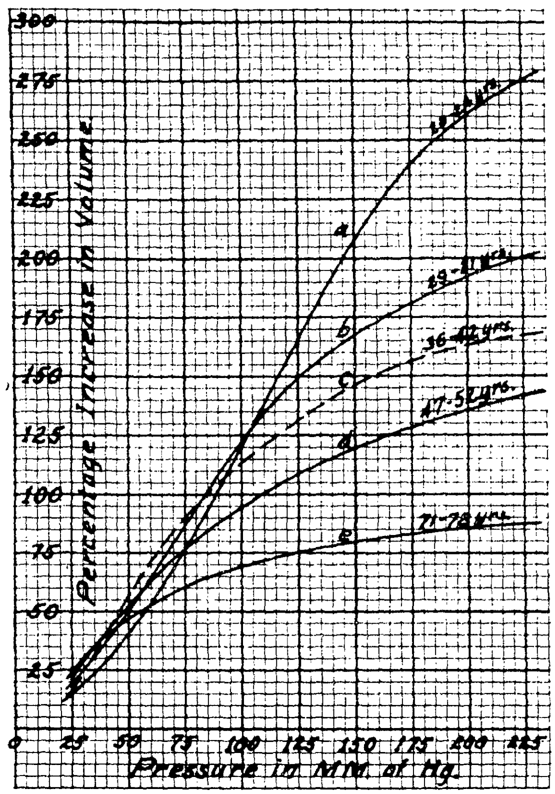


FIG. 4.1 MEAN VOLUME-ELASTICITY CURVES OF THORACIC AORTAS AT VARIOUS AGE GROUPS

These curves show the relation of percentage increase in volume to increase in pressure for five different age groups and were constructed from the mean values obtained from a number of aortas excised at autopsy.

*Obtained from article by P. Hallock and J.C. Benson. Journal of Clinical Investigation 16:597,1937

DYNAMIC CHARACTERISTICS OF PRESSURE-VOLUME RELATIONSHIPS FOR HUMAN ARTERIES*

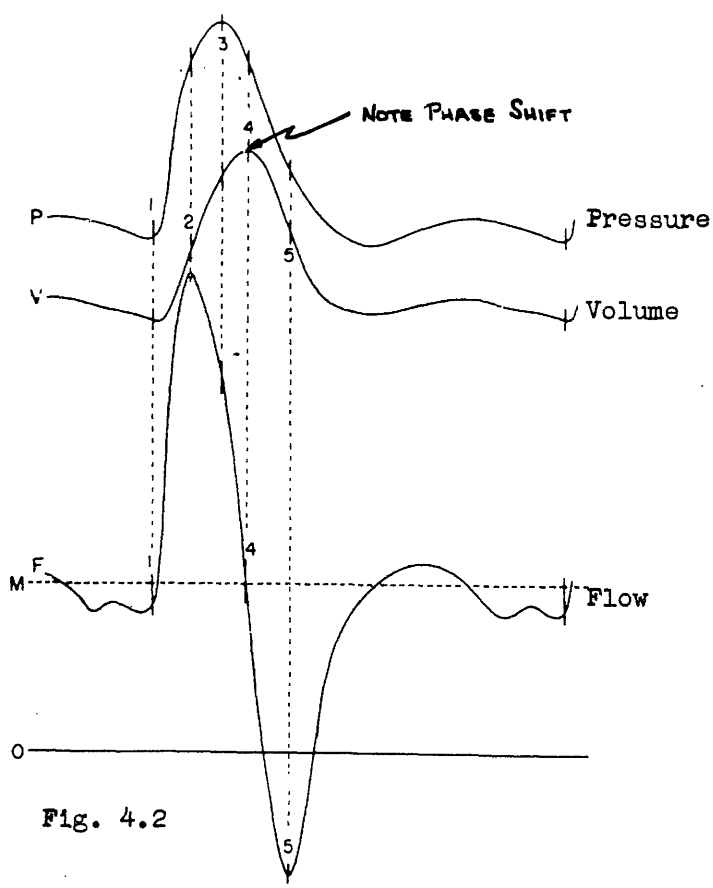


Fig. 4.2

Curves showing inter-relationship between aortic volume distension and the associated intra-aortic pressure.

*Obtained from article by R.F. Shipley, D.E. Gregg, and E.F. Schroeder. American Journal of Physiology. 138: 723, 1943

Pressure Change (P) = Weight density (ρ) x Level change (h) with Blood having a specific weight of 1.057 we obtain the following relation:

$\Delta P =$ .74 mm of Hg/cm of level change, thus a level change of 33.8 cm induces a pressure change of 25 mm of Hg. This is shown experimentally in Figure 4.3 and in the table that follows.

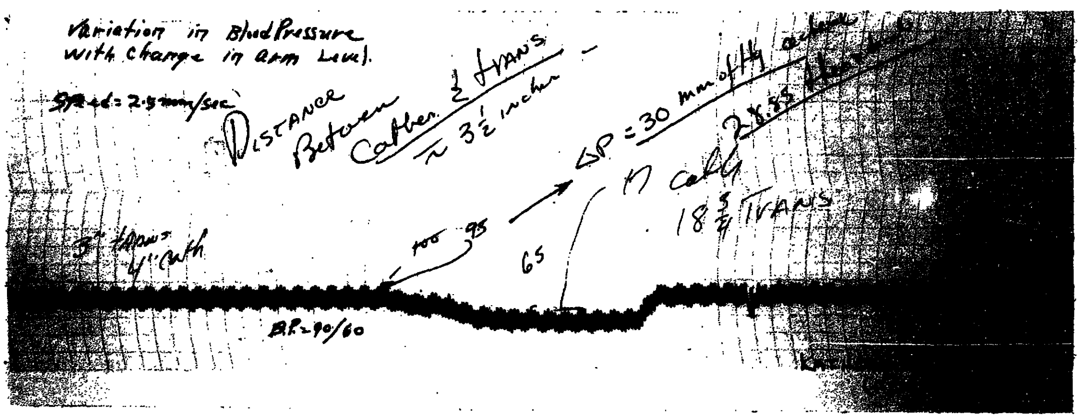


Figure 4.3

An actual tracing showing the hydrostatic level variation inducing a pressure variation, using a strain gage transducer for blood pressure measurement.

Some Other Data Collected Shows:

Trial	Theoretical	Actual
1	23.5 mm of Hg	22 mm of Hg
2	28.8 mm of Hg	29 mm of Hg
3	23.8 mm of Hg	25 mm of Hg
4	24.9 mm of Hg	23 mm of Hg

Figure 4.4 schematically shows the effects of the above two principles on an intra-arterial pressure. In the down position, the arm is lower in a hydrostatic sense and therefore assumes the higher pressure.

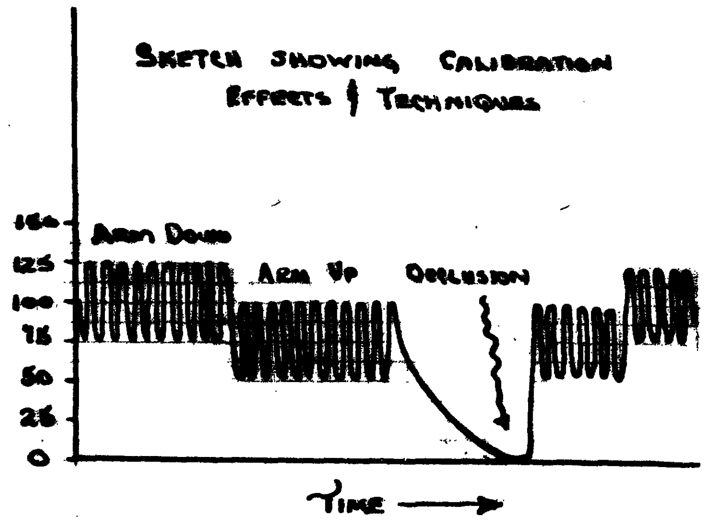


Figure 4.4

In the up position, the artery is higher in respect to the heart, and therefore, the pressure is reduced. The point to note here is that the amplitude of oscillation is <u>not</u> reduced; only the base line is effected.

It is a known fact that the body resists any changes in pressure by compensating the change with vasomotor constriction, which acts as an in-line pump. In our studies, the effects of arterial vasomotor control did not appear to effect the above calibration procedure. We believe the effect to be minimal because of the magnitude of the change in pressure. This vaso-action is more effectual in the veinous return than in the arterial system.

These two factors give us a method for calibrating a pressure device. One merely occludes an artery that gives a base line of zero gage pressure. By level changes, a given pressure can be induced and the associated volumetric change measured; the ratio of these two give the slope of the linearized pressure-volume curves.

In summary, with a known calibration setting for an individual, one can postulate that a volumetric recording device can indeed be correlated directly to intra-arterial pressure. This of course is only possible within the limitation of a linearized theory.

4.4 A Theoretical Model of a System

With these pressure-volume relationships as background, one then seeks a system to measure the volumetric distention of an artery. In developing a theoretical model, it is noted that a volumetric distention of the artery causes a volumetric deformation of the skin surface. Is there a relationship between the volumetric distention of the artery and this movement of the skin surface? Figure 4.5 (a) depicts an idealized cross-section of the human arm. If (d) is the diameter of the artery and (D) is the diameter of the arm section, then the d/D forms a dimensionless ratio of diameters.

If the surface of the skin is made flat as pictured in Figure 4.5 (b), by suitable means, then the relationship of the artery to the skin surface becomes theoretically analyzable by two dimensional stress analysis. Since the d/D ratio is about 0.05 for most individuals and since the skin surface is flat, the model that is to be considered is depicted in Figure 4.5 (c). The model we will use is a semi-infinite elastic medium with a flat surface open to the atmosphere under which is a cylindrical tube. It can be shown (See Appendix B) that deformations within this cylindrical tube are linearly transmitted to the surface. With this model, we assume a quasi-elastic medium and a small d/D ratio. Furthermore, we are assuming an isotropic, homogeneous elastic medium. These assumptions are broad in scope and may prove to be invalid, but

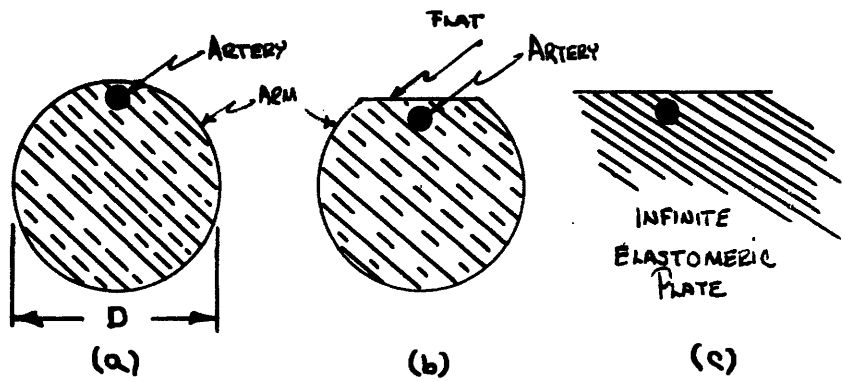


FIGURE 4.5

they do suggest a method of approach. The assumption of a quasi-elastic medium can be given some validity by the analysis in Section 4.2, where we show that body tissue is elastomeric in behavior. One must realize that the organic material between the arterial walls and the skin surface is complex in molecular structure, and surely is not homogeneous. In this model, it is postulated that the tissue characteristics will be relatively constant even though they are not completely homogeneous.

With this background, one seeks a sensing unit which will measure the "deflection" of the surface of the skin in a linear fashion. The functional approach is to develop an instrument which can sense these volumetric deformations of the skin under variable pressures in the artery.

The theoretical model, which makes some broad assumptions about the quasi-elastic nature of muscle and skin tissue, in actuality may cause serious problems. Not only does it assume a quasi-elastic medium, but it assumes a lack of vasomotor control over the artery. It was our belief, however, that the volumetric distention of larger lumen arteries is affected by vasomotor control to only a negligible effect, if at all; conversely, in the capillary bed, vasomotor control has a major effect. Another assumption is that muscular reaction can be minimized at the point of application of a transducer. However, this can prove to be a troublesome problem in the application of the system. The assumption which eliminates the effect of muscular movement is that the interface pressure between the sensing transducer and the surface of the skin must be at all times constant. If this interface pressure is constant, then the transducer will always be in the same relationship to the artery as it was at the initial introduction of the transducer. This assumption of constant interface pressure is basic to the whole procedure, for without it, gross movements due to muscular activity become larger in magnitude than is the deflection caused by the arterial volumetric distention.

4.5 Theoretical System

Since theoretically there exists a direct linear relationship between internal pressure and volumetric distention of human arteries and since there is theoretical grounds for assuming a relationship between subcutaneous volume change and surface deformation, we can postulate a system to monitor blood pressure by indirect means.

In block diagram form, Figure 4.6 shows the basic concept behind the linearized system under study. The intra-arterial pressure is linearly related to the volumetric distention of the artery, which is in turn linearly related to the surface deformation. If the deformation can be linearly recorded, then the output signal will be linearly related to the intra-arterial blood pressure. The next question, is of course, can such a linear recording device be developed.

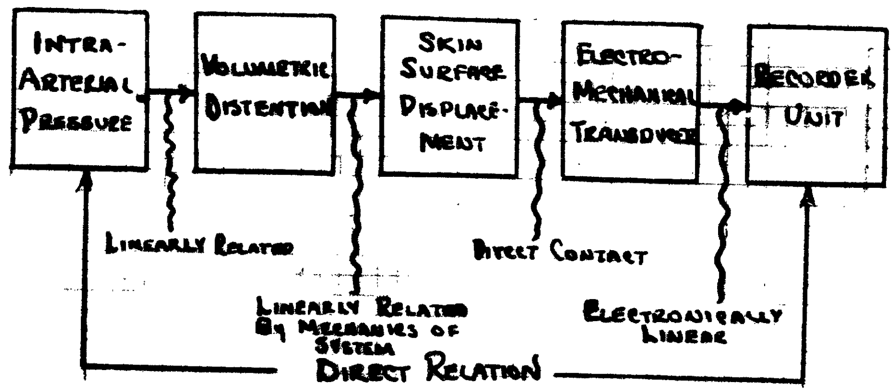


Figure 4.6

4.6 Summary

A theoretical system for monitoring blood pressure by indirect means is to be based on the following:

1.) A nearly linear relationship between blood pressure and arterial distention.

2.) A theoretical relationship between subcutaneous deformation and the deformation of a semi-infinite surface plate of elastomeric composition.

3.) A calibration procedure for evaluating the slope and base line characteristics of each patient's dV/dp curve.

4.) A constant interface pressure to eliminate extraneous gross movements.

V. INITIAL DEVELOPMENT AND TRANSDUCER DESIGN

5.1 Basic Concepts

There are five basic concepts that must be considered before any transducer configuration can be built. Each of these concepts are related to the instrumentation as applied to blood pressure measurement.

1.) The first question that arises is: where do we find an arterial section that meets the requirements. It appears that vasomotor control has its major effect in the smaller lumen vessels of the vascular system. Wiggers (79) states that vasoconstrictive nerves activate muscular elements of the arterioles, metaterioles and possibly the Rouget cells of the capillaries. It has been felt by the medical profession that vasomotor control is limited in effect in the large lumen arteries. There does not appear to be any real common agreement on this fact. A basic function of vasomotor control is to maintain proper oxygen supply for the brain. If there is some duress which causes reduced blood supply, then vasomotor control reduces the flow in the extremities, thus tending to divert the flow toward the brain. One theory is as follows: If the flow is reduced by constriction of the capillary network, then restriction of the larger lumen vessels will not occur because the flow is already reduced by valvular action further downstream. This is only a hypothesis, however, the fact of vasomotor control is known, but its entire action is not fully understood.

If we postulate that vasomotor control is limited in effect in the larger lumen arteries, then we will be justified in using the brachial artery for study. (The brachial artery is the main artery of the arm.) The action of vasomotor control strongly suggests using the temporal artery because it is the artery least affected by vasomotor control. For two reasons we will not study the system at the temperal artery. First, experimentally it was found that application of a transducer to the temperal artery was most difficult. Secondly, the calibration procedures would not be possible at the temporal artery. Thus, for the study contained herein, we will restrict the investigations to the brachial artery, at both the elbow and the wrist.

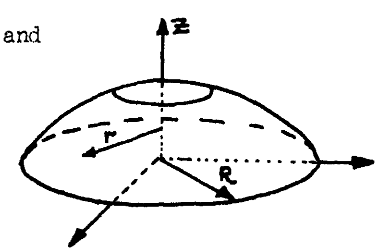
It will be necessary to evaluate the deformation characteristics of the skin over an artery. (Appendix F) If the deformations can be measured with reasonable accuracy, the assumptions of the system can be evaluated.

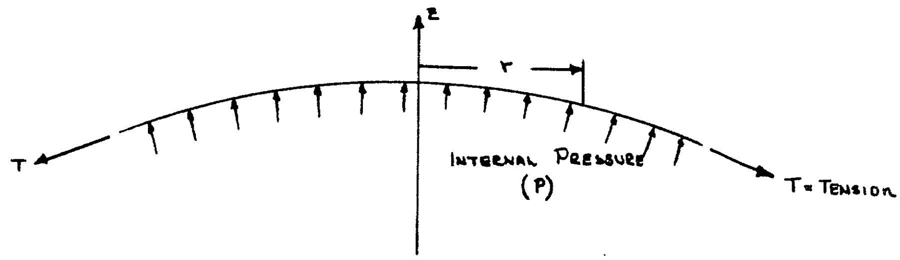
2.) The next question which arises is: how to measure these deformations. There are undoubtedly many possible methods for measuring the deformation. However, it is necessary to measure the deformation of the skin surface. If one were to use a membrane-type material that assumed the "same" characteristics as the skin surface, then this diaphragm could form a dilative element of another system, a system that would be sensitive to these dilative changes. This diaphragm must be different in theory than conventional transducer diaphragms. Most pressure transducer diaphragms rely on the modulus of elasticity of a force summing diaphragm. Here we do not wish to have the diaphragm resist the movements of the skin surface. In fact, the surface of the skin would be an ideal diaphragm. This, by definition, requires a diaphragm material that is more flexible than the skin. This is important for two reasons. First, a resistance to movement may introduce some non-linearity. This effect can be seen in Appendix D, as applied to a stainless steel diaphragm. Secondly, any resistance to movement of the skin will reduce the sensitivity of any transducer. The skin movements are of the order of magnitude of 0.0003 in., so obviously it is not wise to reduce the sensitivity.

3.) With a concept for diaphragm construction, the next problem to arise is the nature of the geometry of a diaphragm. For reasons of practicality and ease of construction, we will look only at two geometries, namely; circular and rectangular,

a.) Circular Membrane Diaphragm:

Assuming small deflections and constant tension in the membrane, we can, by symmetry cut out a concentric circle and analyze the conditions of equilibrium.





Horizontal equilibrium is satisfied by symmetry.

Vertical equilibrium:

$$p\pi r^2 = -T \frac{dz}{dr} 2\pi r$$

OR $$-\frac{dz}{dr} = \frac{p}{2T} r$$

Integrating and putting in the boundary conditions [at $r = R$, $z = 0$

yields $$z = \frac{P}{4\pi}\left(R^2 - r^2\right)$$ at center, $z = \frac{P}{4\pi} R^2$

the volume under the membrane is

$$V = \int_{r=0}^{r=R} z \, \pi \, 2r \, dr = \frac{\pi}{8} \frac{P}{T} R^4$$

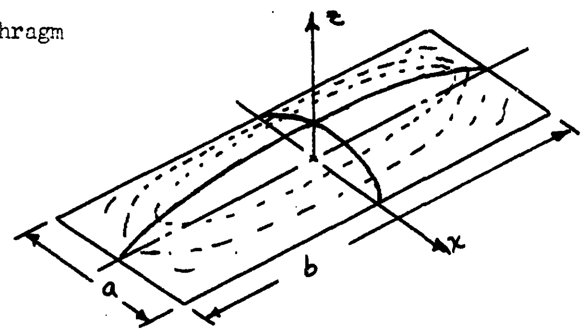
or the volume is a direct function of P.

$V = KP$ where $K = \frac{\pi R^4}{8T}$ where $\rightarrow$ (T constant)

Similarly the deflections are a direct function of Pressure.

b.) Rectangular Membrane Diaphragm

Again assuming small deflections and constant tension and $a \ll b$



A similar analysis will yield a vertical force balanced such that

$$\frac{dz}{dx} = -\frac{P}{T}x$$

and integrating and putting in boundary conditions $\left(x = \frac{a}{2} \text{ @ } z = 0\right)$ yield the deflection to be linear with pressure.

$$z = \frac{P}{2T}\left(\frac{a^2}{4} - x^2\right)$$

or at the center of diaphragm. $(x = 0)$

$$z = \frac{P}{2T}\left(\frac{a^2}{4}\right)$$

The volume under the diaphragm will be found by integration.

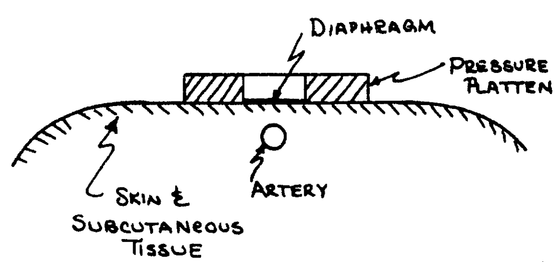
Volume $= \dfrac{Pa^3b}{12T}$ and again the volume is directly proportional to the internal pressure.

Volume $= K'P$ where $K' = \dfrac{a^3b}{12T}$

In conclusion, the volumes and center point deflections are proportional to the internal pressure for both diaphragm configurations.

4.) Another point to consider is the effect of interface pressure on the diaphragm characteristics. The surface of the skin can be disturbed by muscular activity as well as by blood pressure. It is obvious that the transducer must remain in the same relationship to the artery at all times. If the relationship changes, then the constants in the quasi-linear relations will be altered.

The relationship between the transducer and the skin will be unaltered if the interface pressure is held constant. Since it is essential to the linear theory that the interface be flat and since the interface pressure is to be held constant, the following configuration will be used.



If the surface area of the pressure platten is about four times the area of the diaphragm, it has been found experimentally that the muscular movements will cause the whole transducer to ride up and down, while the diaphragm remains in the same relative position to the artery. The holding fixture must permit this transducer movement while still retaining the constant interface pressure.

There are three ways of doing this, and each will be discussed later in the paper.

5.) The final concept to be analyzed, is the effect of the earth's gravitational field on the transducer. If the transducer were to remain stationary at all times, then the gravitational effects would not be a factor. However, the calibration techniques involve moving the arm through a specified distance to induce pressure changes. During this movement, the transducer interface changes its relationship to the earth. Since the transducer has "weight", it contributes to the interface pressure. The contribution of the weight will be a function of the relationship of the center of gravity to the interface. If the interface is horizontal, then the full effect of the weight will be felt at the interface, however, if the interface is at some angle, then the component of the weight will go as the cosine of the angle, or the weight as felt on the interface will be reduced. There are two ways of circumventing this problem. First the calibration can be carried out such that in the two extreme calibration positions, the weight effects are the same. This is shown schematically in Figure 5.1. The second method is to design the transducer so that there are no gravitational effects. The author has proceeded under both techniques. As will be seen in subsequent discussion; this later technique involves designing the transducer holding device such that the pivot point is also the center

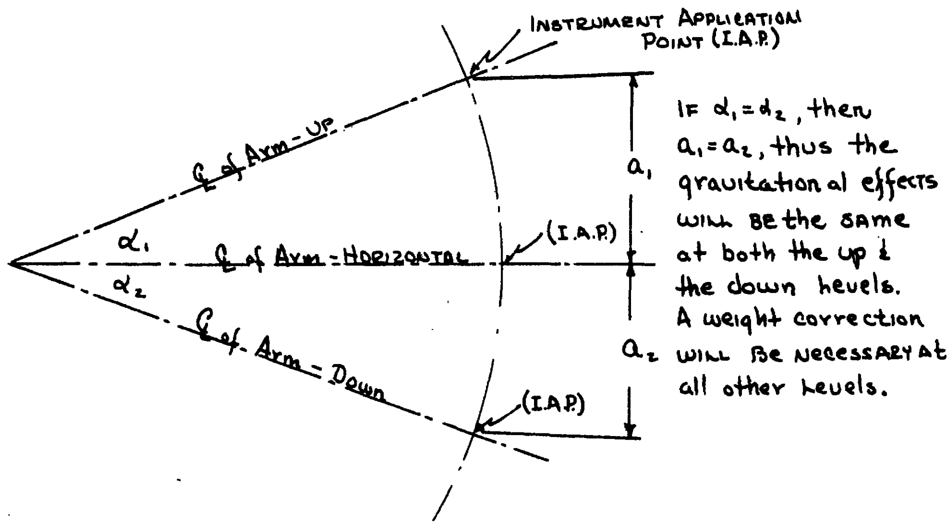
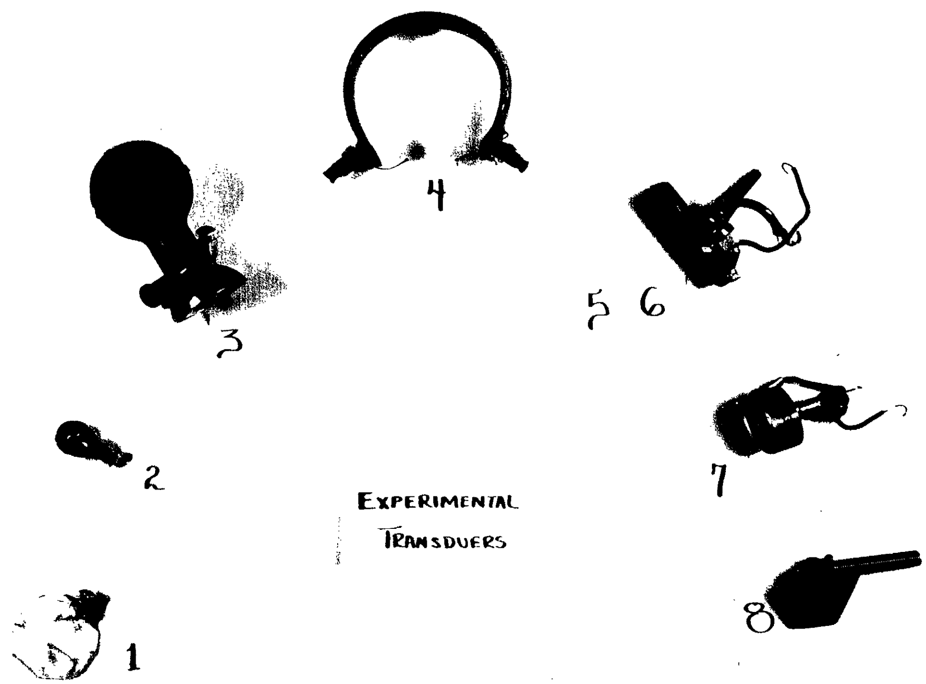


Figure 5.1

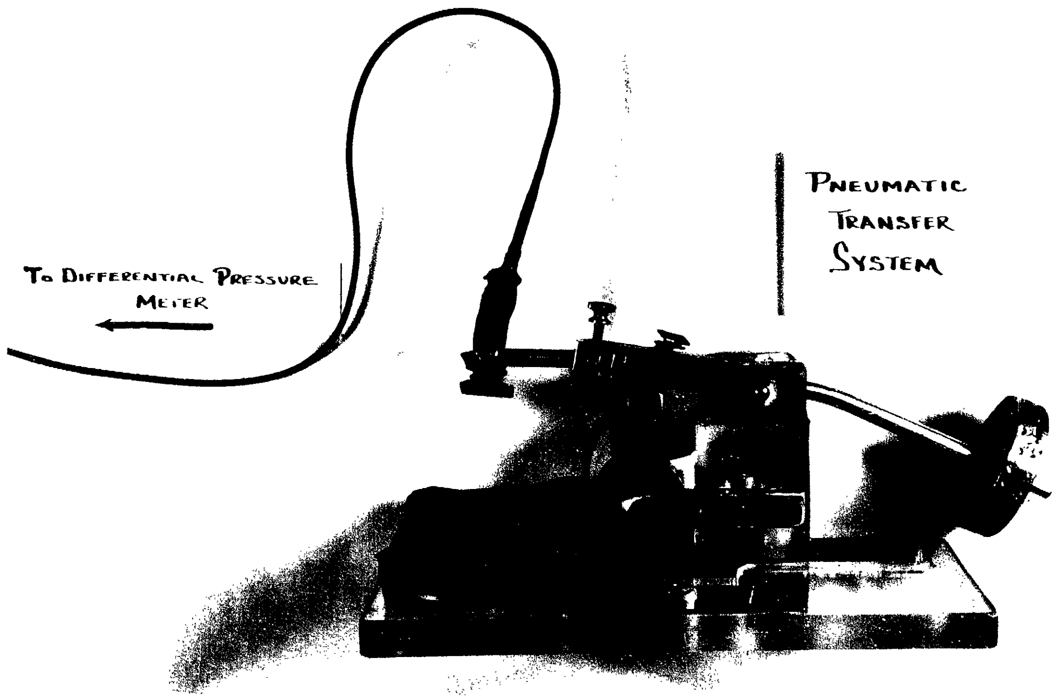
of gravity.

5.2 Pneumatic Transfer System

The analysis related to the membrane theory in Section 5.1 suggested a system in which the membrane diaphragm is a dilative element of a closed pneumatic system which is sensitive to volumetric changes. Going under this assumption, a number of experimental transducers were built, each incorporating improvements, both technical and medical. Some of these can be seen on the Photograph P.6. The resulting transducer of this development can be seen as (8) in that photograph. This transducer yielded very valuable results, as will be seen in the evaluation section of this paper. The system consisted of the transducer with a thin rubber membrane connected to a closed system which terminated at a highly sensitive differential pressure meter. The meter used was manufactured by the Decker Aviation Corporation and the specifications and characteristics are listed in Appendix A. As one can note, the



F.6


To Differential Pressure
Meter
Pneumatic
Transfer
System

F.7

sensitivity is very high, thus very small volume changes can be recorded very easily.

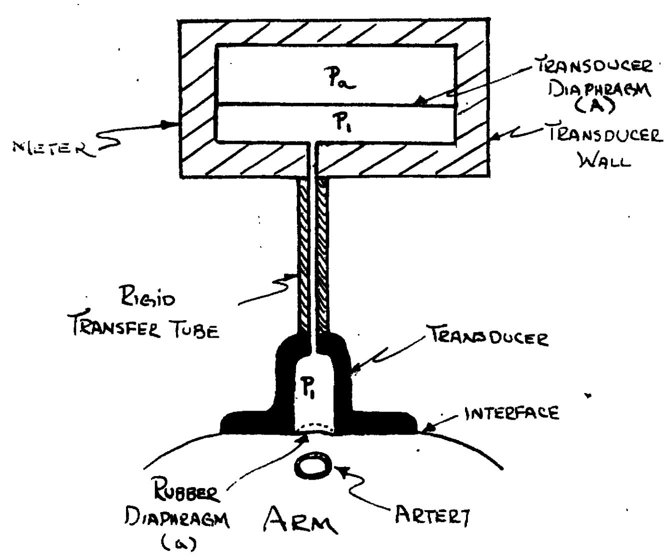


Figure 5.2

A schematic diagram of the system is depicted in Figure 5.2. A surface deformation results in a deflection of the diaphragm, and results in a pressure change (Pl) in the closed system. This pressure change is "sensed" by the differential pressure meter, and yields an electrical potential output, which can be graphically recorded. As will be seen in the evaluation section, this device is temperature sensitive, thus it was necessary to develop still another transducer system, which eliminated this temperature problem.

5.3 Air Gap Capacitance System

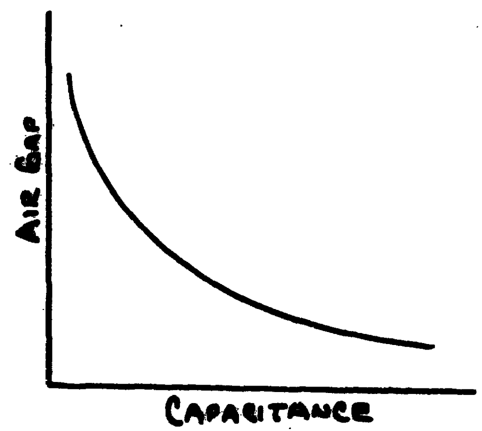
The primary problem with the pneumatic system was its temperature sensitivity. The obvious thing to do was to either compensate for the temperature effects or to eliminate the pneumatic transfer system. The most practical solution was to eliminate the pneumatic aspects of the system by placing the entire sensing element upon the arm, in effect, bring the entire pressure meter down on the arm. Since there were no commercially available transducer designs with the assumptions of this system, it was necessary to develop a transducer for this application.

There are many methods for sensing volumetric changes, ranging from optical methods to strain gages, and from mechanical leverage systems to inductance variations. Each has its advantages and disadvantages. A transducer was designed based on a variable air gap capacitor. This type of variable parameter was chosen because of the high output sensitivity obtainable, and because of the similarity of design.

The method of sensing the capacitance variation is extremely important. A dynamic capacitance measuring device, first suggested by Lion (80), has been commercialized by the Decker Aviation Corporation. The monitoring unit, for sensing small variations in capacitance, consists of a small ionization chamber containing a rarefied noble gas. A 250KC oscillator applies a signal across this chamber by means of internal electrodes. This potential ionizes the gas within the chamber. A small portion of this high frequency excitation appears across two other electrodes to which external measuring capacitors are attached. The induced A.C. voltage on these measuring capacitors varies as the capacitance is varied. When an A.C. voltage exists between the two electrodes, a difference in migration of the electrons in the gas occurs, thus giving rise to a D.C. potential difference between the electrodes. This D.C. signal will vary in accordance with the variation of external measuring capacitance, thus giving rise to a sensing system for external variable capacitors.

This Delta Unit, as it is called, can be used with differential capacitors, and thereby yielding a linear relation between air gap variation and output potential signal. A single parallel-plate variable air-gap capacitor varies in a non-linear fashion.

Either of these configurations is possible, however, we are always limited by the fact that one surface of a plate of a capacitor must be free to be placed on the surface of the patient's skin. This limits the various possible combinations, and for reasons of simplicity, it is therefore necessary to use a single parallel plate capacitive system. The problem on non-linearity is still to be considered.

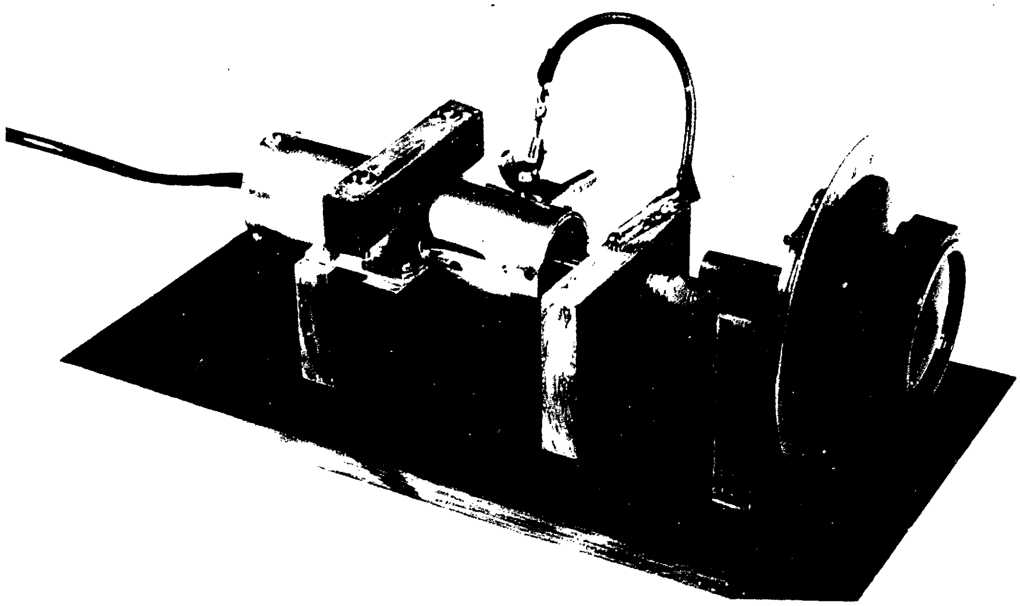


Single Parallel Plate
Capacitor Characteristics

The first transducer to be constructed was a non-linear device, but it was felt that an experimental study program could prove or disprove the validity of the basic concept even though there were electronic non-linearities. The first transducer consisted of a rubber diaphragm with a floating electrode imbedded in the diaphragm. As the diaphragm deflected, the electrode's position with respect to a fixed electrode would vary, and thus the capacitance would vary. This system yielded instructive information.

An important development would be a linearized system. In discussions with various individuals, it became apparent, that quasi-linearized transducers might be realistic. A single parallel plate capacitor is essentially linear (2%) if the air gap variation is of the order of ± 10% of the total air gap. Furthermore, by application of the Brook-Smith principle (82) single parallel plate capacitors can be linearized over a limited range. This is accomplished by placing a material of high dialectric strength in parallel with the air gap.

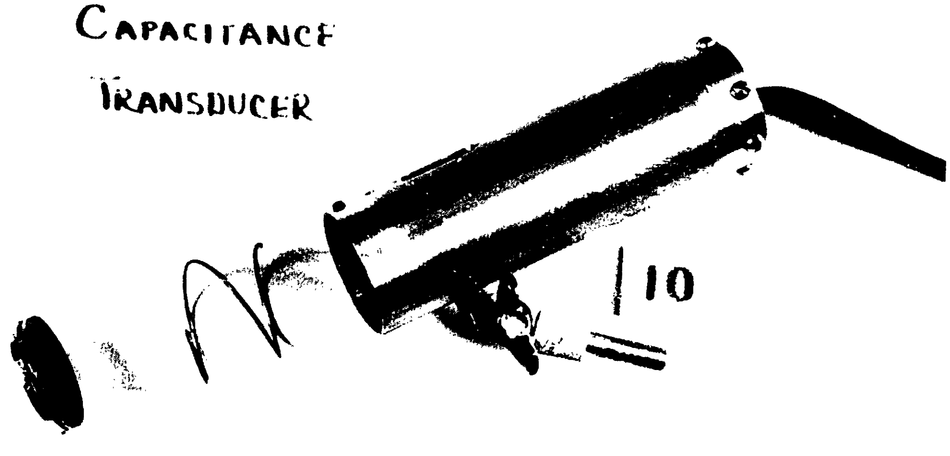
A new transducer was built so that various diaphragms could be tested for linearity. Numerous materials and configurations were tested. A linearity device was also constructed to test these transducers. The test fixture is shown in photograph P.8. It consisted of a micrometer screw which deflected the diaphragm through known increments. The micrometer screw head had a flat surface which was placed in direct contact with the diaphragm. This is somewhat different than is the actual case, however, it was noted in the membrane theory, that the center point deflection was a direct function of pressure; thus the linearity is evaluated on that assumption. Using this assumption for linearity evaluation, numerous diaphragms (examples pictured in photograph P.9) were tested and data was obtained. This data was plotted for each, and curves like the following one were obtained. The various diaphragm materials tested included. Beryllium Copper (thicknesses ranging from .00015 to .001), various thicknesses of surgical rubber, and mylar. The non-conductive materials were made conductive by various techniques. The most productive technique was made possible by the National Research Corporation. It involved vacuum coating aluminum on 25 gauge mylar sheet. This mylar sheet provided the necessary characteristics; first it was flexible like rubber, but it was conductive. Secondly, by placing the aluminum surface outside and with the mylar side adjacent to the fixed electrode, the Brook-Smith principle was applied. It was found that significant differences in linearity existed between

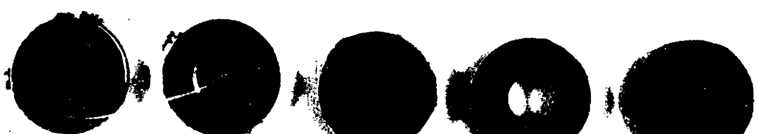


LINEARITY METER

P.

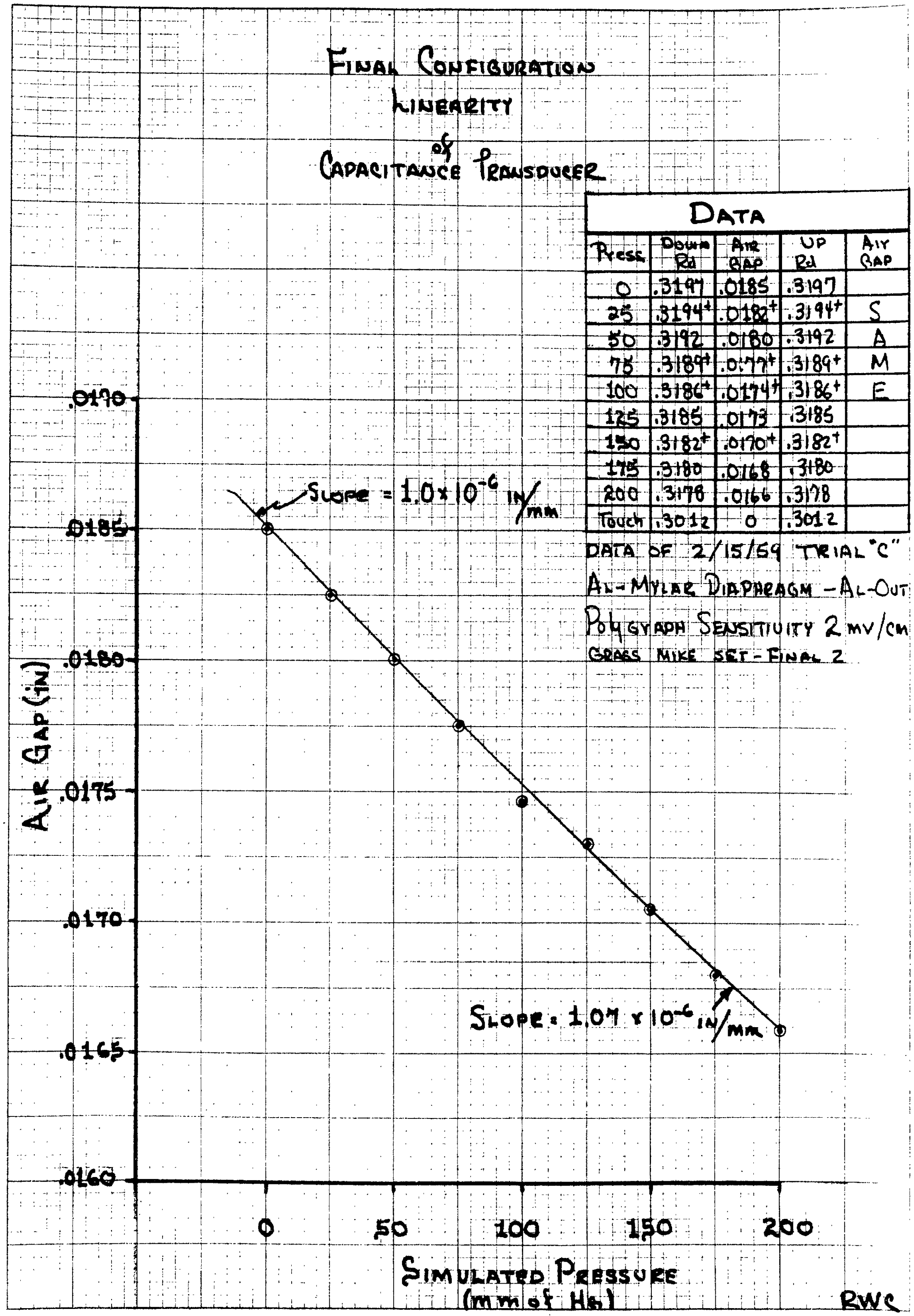
CAPACITANCE
TRANSDUCER





1 2 3 4 5

P.9



Data				
Press.	Down Rd	Air Gap	Up Rd	Air Gap
0	.3197	.0185	.3197	
25	.3194+	.0182+	.3194+	S
50	.3192	.0180	.3192	A
75	.3189+	.0177+	.3189+	M
100	.3186+	.0174+	.3186+	E
125	.3185	.0173	.3185	
150	.3182+	.0170+	.3182+	
175	.3180	.0168	.3180	
200	.3178	.0166	.3178	
Touch	.3012	0	.3012	

Data of 2/15/69 Trial "C"

Al-Mylar Diaphragm – Al-Out

Polygraph Sensitivity 2 mv/cm

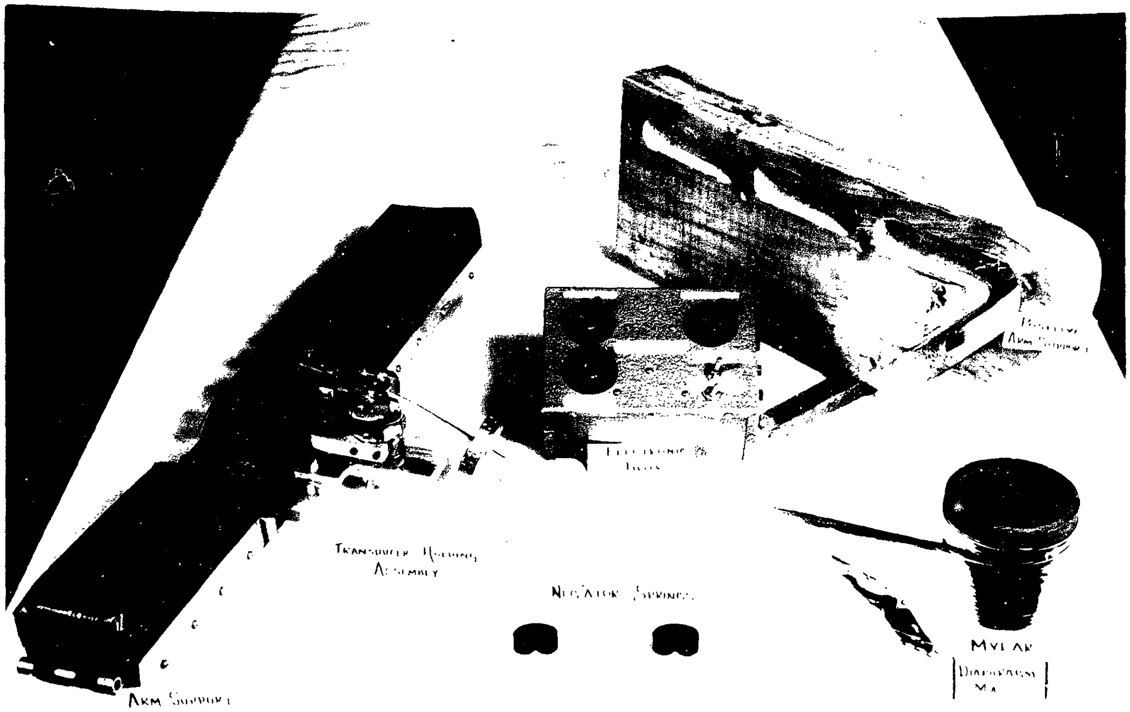
Grass Mike Set – Final 2

FIGURE 5.3

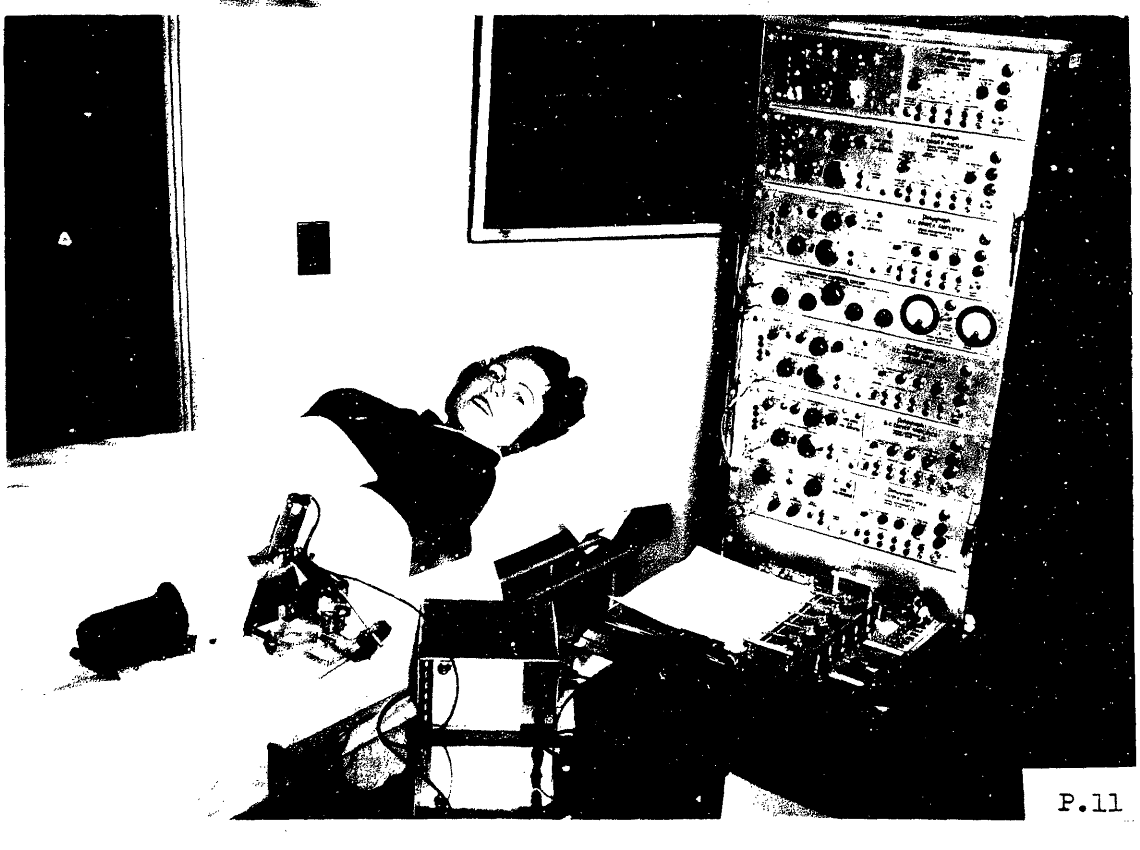
having the aluminized side inside or outside. The most linear responses were obtained with the mylar between the aluminum and the fixed electrode, this is in principle the concept proposed by Smith and Colls (82). The characteristics of this configuration are shown in Figure 5.3. The linearity is measured by the variation of the slope of the characteristic curve. With the final configuration, the slope varied by $\pm$ 3% which induces an error of $\pm$ 2 mm of Hg over the entire range of pressures. Thus, in theory, one has an electronic system which is linear to within $\pm$ 2 mm of Hg, and therefore we are ready to apply the system to blood pressure measurement.

5.4 Transducer Positioning

As indicated previously, it is extremely important to maintain at a constant level the interface pressure between the transducer and the patient. The transducer is positioned on the arm of a patient by means of a pivot cantilever arm. The holding assembly is pictured in photograph P.7. The cantilever arm is pivoted about needle point bearings for reasons of anti-friction. The rod and mass extending to the left is a means of setting the center of gravity on the axis of the pivot rotation. The patient's arm is placed between the transducer and the arm support. The interface pressure can be maintained effectively constant by three methods. The first method considered was a pneumatic cylinder placed between the transducer arm and the base. The pressure to be maintained constant within the cylinder. It became obvious that such a system would be attended by many problems of friction and control. For reasons of practicality, it was discarded as a possible method for maintaining constant interface pressure. The next idea considered and applied, was the placing of a low force coefficient spring between the transducer arm and the base.


Transducer Holding Assembly
Neg'ator Springs
Mylar
Arm Support

P.10



P.11

It was thought that since the arm movements were of the order of a few hundredths of an inch, that the variation in the force due to the spring would be small. This idea provided valuable information concerning the value of the interface pressure. The pressure level for ideal signal transmission proved to be about 5 psi, however, after repeated applications of the transducer at that pressure, we found that it was medically not desirable because it caused some trauma. With further experimentation, the pressure level which gave the best signal transmission while not adversely affecting the patient, proved to be about 2.5 psi.

The spring principle for interpressure maintenance, proved to be undesirable because the pressure did vary significantly. The next idea that was tried was the application of a weight suspended in order to maintain constant interface pressure. This too was an improvement, however, it was clumsy and impractical in application.

The most practical solution to the constant interface pressure problem was the application of "constant force" springs manufactured by the Hunter Spring Corporation. They are known as "Neg'ator" Springs. They consist of a pre-stressed coiled spring which is allowed to rotate freely about the coil axis. The work done on the spring during uncoiling, is a function of the reverse bending of this pre-stressed material. The reverse bending occurs over 90° of the coil and is in the same relative position at all times, thus the force created is essentially constant during the uncoiling action. There is a thickness effect, but it is minimal for small deflections.

Using this type of spring, a constant force of 8.6 lbs. $\pm$ .2 lbs. was obtained for the entire range of operation. During a particular application, the force varies by approximately 0.05 lbs. or 0.6%.

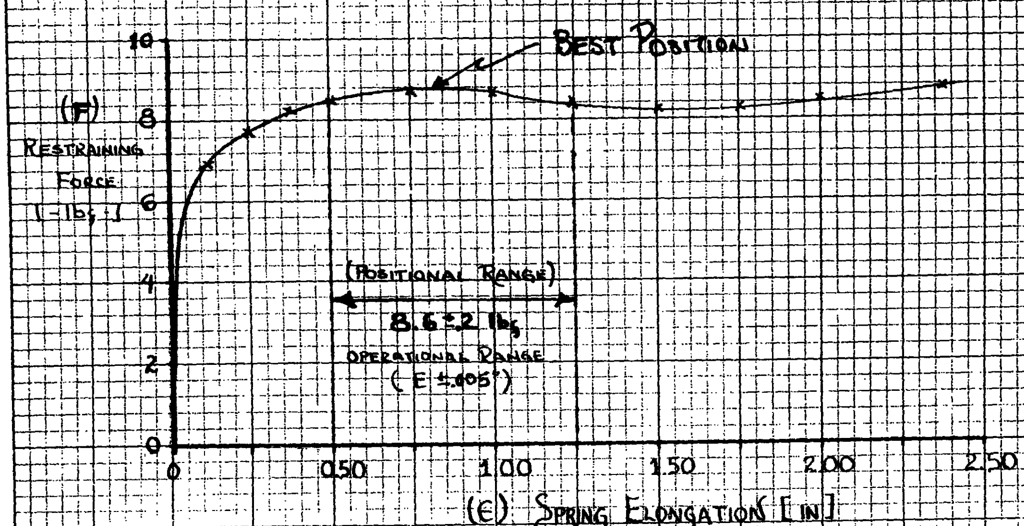
The Neg'ator Spring as applied is shown in P.7 and the calibration curves appear on the following page. The axis of coil is free to rotate on miniature precision ball bearings.

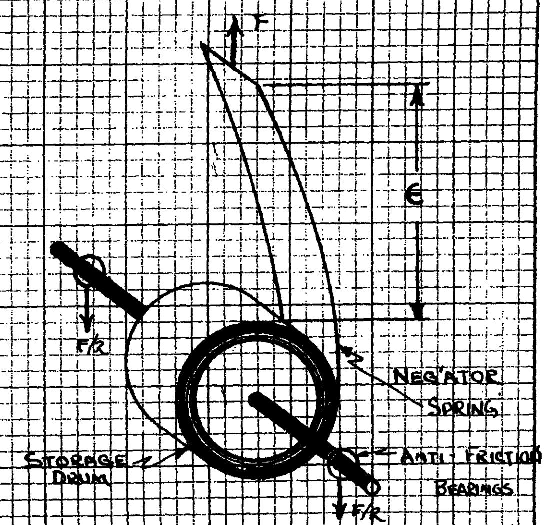
5.5 Summary

The concepts for transducer design are discussed. These concepts, when applied to a prototype model, yields electronic input-output non-linearities of the order of ± 2 mm of mercury pressure, or expressed in terms of characteristic curve slope variation ± 3%. This transducer is based on the variation of the capacitance of a simple air gap parallel plate capacitor.

The total unit as applied to a patient is depicted in P.11. The transducer and holding assembly are seen applied to the patient and the Decker Delta Unit is in the foreground. The oscillograph is in the background.

FORCE - ELONGATION CURVE
for
CALIBRATION
of
NEG'ATOR SPRING*





ELONG.	1	2	3	AVERAGE
.125	6^{15}	$6^{15\frac{1}{2}}$	6^{15}	6.95
.250	7^{13}	7^{12}	$7^{12\frac{1}{2}}$	7.78
.375	$8^{5\frac{1}{2}}$	$8^{4\frac{1}{2}}$	8^{5}	8.31
.500	8^{8}	8^{8}	$8^{8\frac{1}{2}}$	8.52
.750	$8^{12\frac{1}{2}}$	8^{13}	$8^{12\frac{1}{2}}$	8.79
1.000	8^{11}	8^{12}	$8^{12\frac{1}{2}}$	8.74
1.250	8^{7}	8^{7}	$8^{6\frac{1}{2}}$	8.43
1.500	8^{4}	8^{5}	8^{5}	8.29
1.750	$8^{3\frac{1}{2}}$	8^{4}	$8^{4\frac{1}{2}}$	8.25
2.000	8^{7}	$8^{6\frac{1}{2}}$	8^{7}	8.43
2.375	8^{12}	8^{12}	$8^{11\frac{1}{2}}$	8.74

DATA of 3/21/59
for PART No 12H3K

* Mfg by HUNTER SPRING Co.
LANSDALE, PA.

FIG. 5.4

RWCORELL

VI. PRELIMINARY EVALUATION

The objective of this whole investigation is to consider a new technique for recording human blood pressures. We wish first to consider the evaluation from a microscopic viewpoint before a complete and detailed analysis can realistically be undertaken. The objective is to evaluate the logic and feasibility of a concept. For this reason, and for reasons of practicality, much of the evaluation is directed toward the concept. It is obvious that a more detailed program of study will be necessary before the practicality of the system can be predicted.

From this background, the evaluation will be chronological in character. This will permit an analysis, pointing out some of the problems encountered, and how they were overcome.

6.1 Evaluation of the Pneumatic Transfer System

The pneumatic transfer system was described in detail earlier in this paper. Briefly, it consisted of a highly dilative rubber diaphragm which created a pressure change in a closed volume system. An electronic differential pressure manometer was used to record the associated pressure changes due to the change in volume.

This system is essentially linear in behavior provided the temperature remains constant. Since the surgical rubber is essentially a membrane, the volumetric dilatations of the diaphragm are linearly related to the skin surface deformations, see section 5.3. The closed pneumatic system consists of two dilative elements, namely, the measuring diaphragm of the manometer and the membrane diaphragm. The deflection of both diaphragms affects the character of the relationship between pressure and volume. The measuring diaphragm of the manometer is of the conventional type, namely, a force summing unit relying on the elasticity of the material of which the diaphragm is constructed. The deformation of the measuring diaphragm is linear with pressure according to the manufacturer. The deformation of the measuring diaphragm is of the order of micro inches; whereas the deflection of the rubber diaphragm is of the order of 0.0003, which means that there is about two orders of magnitude difference between the two deflections. Even though the deflection of both diaphragms is linear with pressure, it is possible that in combination there would exist a non-linear coupling. If such coupling did occur, it would be a second order effect due to the difference in the deflection characteristics of the two diaphragms. Essentially, the system consists of one dilative element with "large" volumetric changes and another dilative element consisting of "small" volumetric changes. Theoretically, one should know precisely the character of these effects, but since the system proved to have other serious affects, such an evaluation was not obtained. Then, the salient point, assuming a perfect gas and constant temperature, is seen by the equation of state. Changes in volume are directly related to pressure changes.

$$-\frac{dV}{V} = \frac{dP}{P}$$

Using this system under highly controlled conditions of an anesthesized patient. no disturbances, and temperature stability, tracings like the one shown in Figure 6.1 (a) were obtained. One can see the character of the form is very much like the intra-arterial pressure. This stable recording was possible only on a few cases. There always tended to be some modulation, in arandom manner, as seen in Figure 6.1 (b). This was not totally understood at first. Preliminary analysis suggested that these movements were due to muscular instability. However, this was not the entire answer.

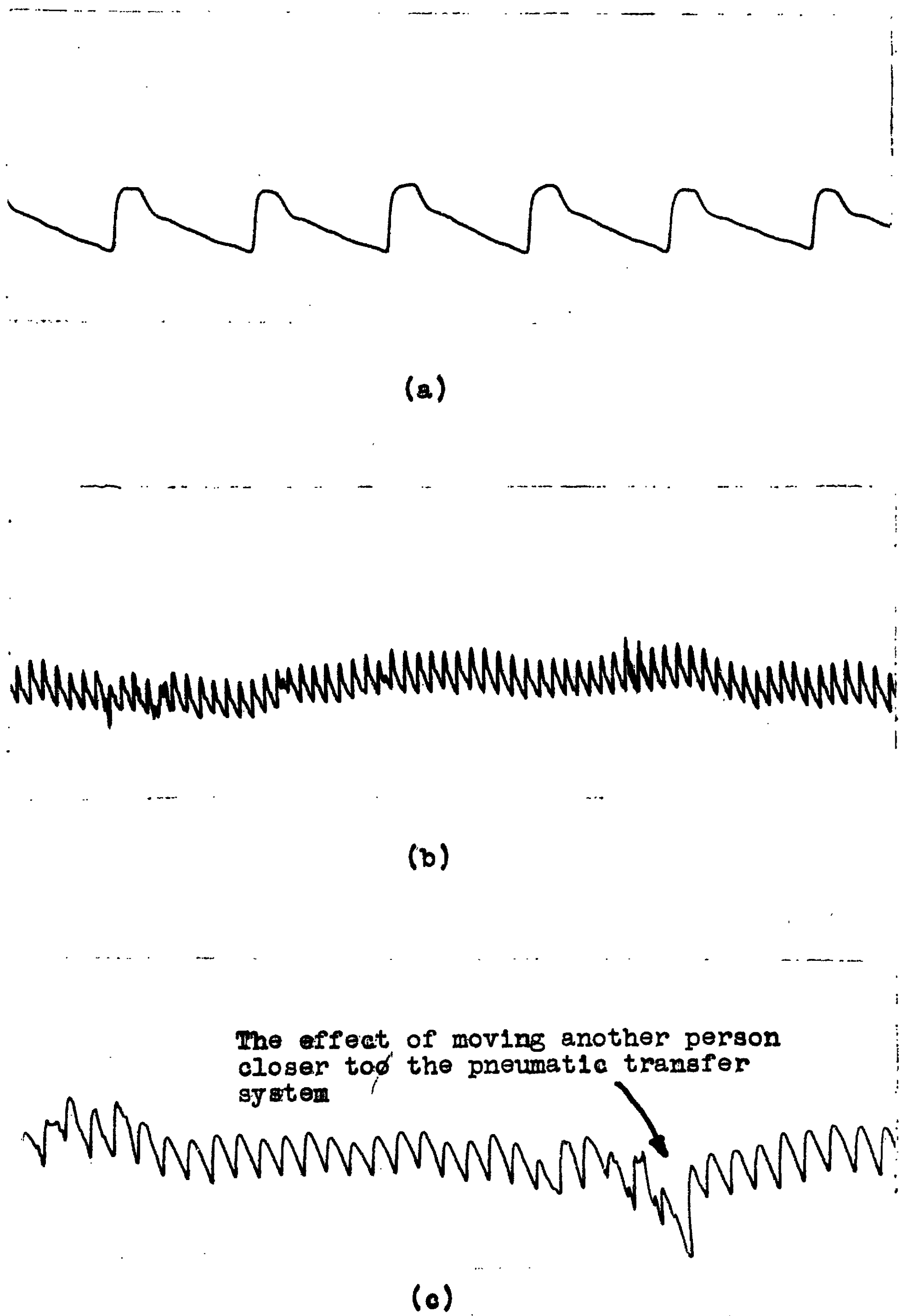

(a)
(b)
The effect of moving another person closer toø the pneumatic transfer system
(c)

FIGURE 6.1

It was assumed that for all practical purposes the temperature remained constant. This seemed like a fair assumption in light of controlled hospital temperatures, however, this proved to be a great source of error.

Looking at the equation of state again, it becomes clear.

Equation of State:

$$pV = RT$$ (see page 50 for symbols)

Take logarithmic derivatives yields:

$$\frac{dp}{p} = \frac{dT}{T} - \frac{dV}{V}$$

The entire volume (v) of the system was approximately 6000 cubic millimeters. The change in volume (dV) was approximately 1.5 cubic millimeters. Thus dV/V was approximately 1.5 parts in 6000. If the temperature was constant, then the pressure change would be of the same order, however, since room temperature is approximately 540° R, temperature variations of 0.1° F would be of the same order of magnitude as dV/V, and thus the system is extremely sensitive to temperature changes. This effect can be seen in Figure 6.1 (c) where an individual came close to the transfer system; the effect was to change the base line by a significant amount.

It was thought that insulating the system would help, however, the effect still remained while the insulation complicated the system.

The pneumatic transfer system was positioned on a patient during a major operation. Simultaneously, the intra-arterial catheter was positioned in the lower end of the aorta. After precautions to permit stable recordings, the output signals from both recordings were super-imposed on an oscilloscope. The two signals, when super-imposed, were identical within the limits of the electron beam. Unfortunately, the photographic equipment was not set-up to photograph this for a permanent record. However, this comparison was noted by the entire surgical staff and provided a real impetus for continued work on this system.

The near-identical nature of these two tracings strongly indicates that there is no significant phase shift with this system. Furthermore, if there were significant non-linearities, superposition would undoubtedly indicate their presence. This superposition technique has provided valuable information about the characteristics of the system.

Even though, the pneumatic transfer system provided valuable informations, it was not the answer to the problem because of its high degree of temperature sensitivity. For this reason, other techniques were studied.

6.2 Non-Linear Capacitive Transducer

A non-linear transducer was analyzed and developed because it would yield the desired result, yet would not be too prohibitive from a development point of view. This is important, for if the basic theory were invalid, it would be clarified by a relatively simple technique without an extensive transducer development program. Once the theory has been evaluated, then a more detailed program of development should be undertaken.

This non-linear transducer had a diaphragm of surgical rubber within which was embedded a floating ground electrode. The non-linearities of this probe were due to a number of factors.

1. It was a single parallel plate capacitor, and the characteristics are inherently non-linear.
2. The relative size of the two parallel plates was such

REPEATABILITY OF CALIBRATION - ESTABLISHING BASE LINE

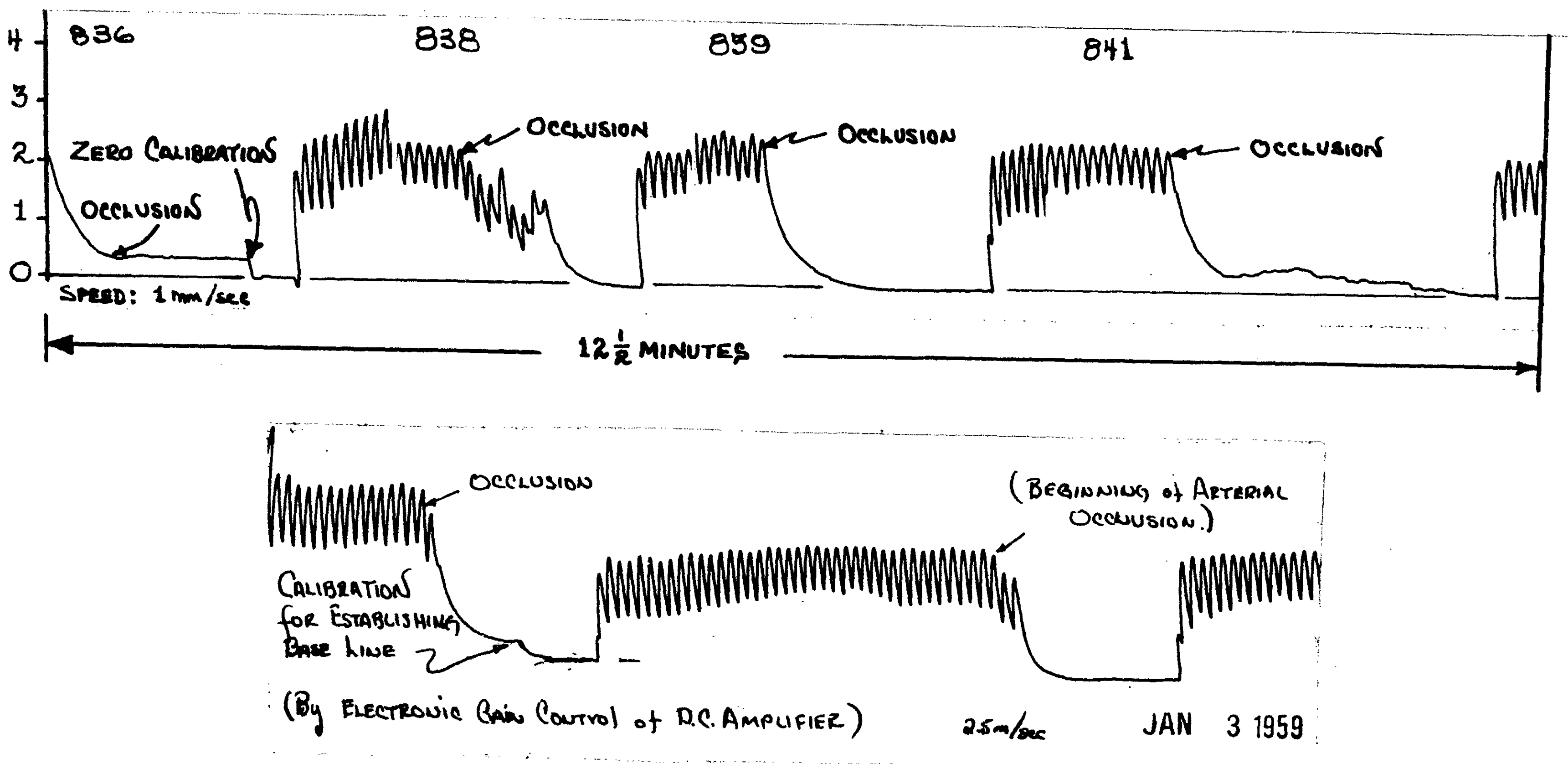


FIGURE 6.2

that stray capacitive effects were major.

3. The air gap variance was greater than ten percent, and thus the inherent non-linearities were large.

4. The positive electrode shape was such that stray fields could adversely affect the characteristics.

5. Unknown to us at the time, the matching impedence between the polygraph and the delta unit was such as to create electronic non-linearities. However, upon subsequent experimentation, these non-linearities did not amount to a significant factor.

Even though the transducer was attended by such inherent design problems, for a given situation, the characteristic response should be invariant. Thus, for a given patient, we could analyze the potentialities of the system.

The first concept to be checked was the calibration procedure to establish a base line of atmospheric pressure. The tracings showing this effect are given in Figure 6.2. In the top tracing of that figure, the initial chart shows the setting of the calibration by adjustment of the grid voltages of the power amplifier. This procedure positions the potential output of the electronic system so that it reads at a pre-determined level, namely atmospheric pressure. Once that is set, we checked the repeatability of the calibration procedure. In this particular trial, over a twelve minute period, the base line returned to the pre-set position within 2 or 3 mm of Hg. This procedure was carried out on numerous patients, and in every case, this procedure was repeatable within the limite given.

Due to the non-linearities, this procedure can lead to a misguiding result, for as the pressure is reduced, the output sensitivity of the capacitive transducer is also reduced. This would permit relatively large errors to be undetectable because of this loss in output sensitivity. However, with a number of different patients under study, and with the output sensitivity varied, we were still able to obtain good repeatability. This evaluation of the base line calibration procedure provided encouraging data. The next effect studied was the effect of hydrostatic level variation.

This effect was extremely difficult to evaluate, for the transducer was extremely sensitive to positional effects. During the hydrostatic level changes, there are significant movements that can alter the initial positional relationship. Thus only under highly controlled condition were we able to check this procedure.

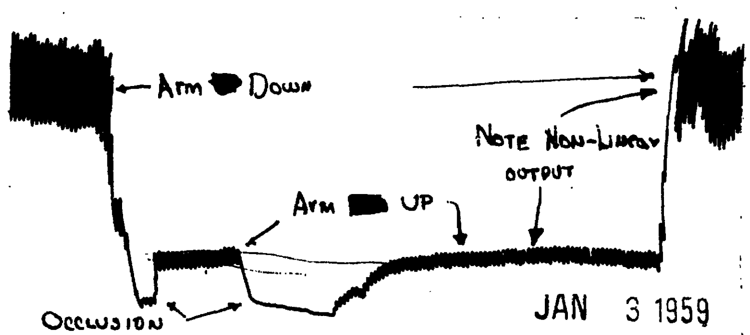


Figure 6.3a

The above tracing, taken over a two minute period, shows all the calibration effects. With the arm down, the pressures are higher. With the arm up, the pressures are lower, and during occlusion the output reduces to approximately atmospheric pressure. Similar tracings were obtained from numerous other patients, and for the most part they are identical to the above. The degree of non-linearity varies from patient to patient because the transducer is operating over a different portion of characteristic curve. This presented a real problem since one wishes to know whether or not the device can be calibrated. Continued study on this calibration technique, analyzing tracings like the one below, convinced the author that the hydrostatic level variation was repeatable over short periods of time (15 minutes) and that with perfection of the instrumentation, this repeatability time limit might very well be extended. The reason that one can not obtain repeatability for longer periods of time is that the transducer is so sensitive to position and orientation, that the slightest movements completely alter the initial calibration.

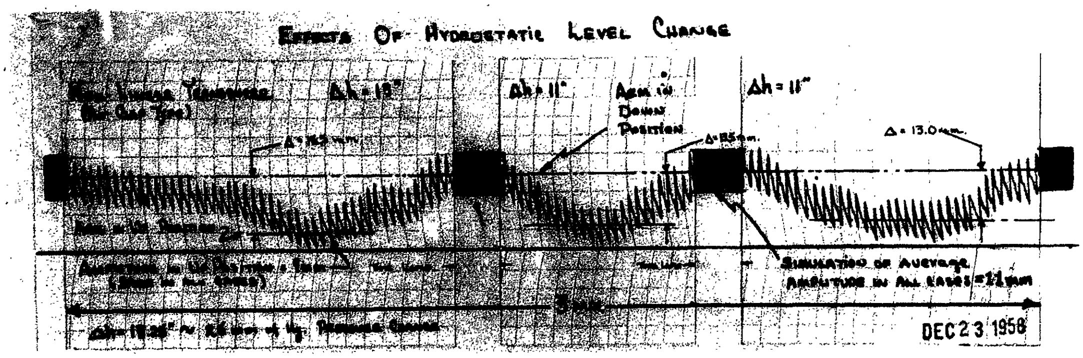


Figure 6.3b

The evaluation of the calibration techniques involved plotting the output characteristic error on a time basis and from this information, determining the cut-off point for repeatability of $\pm 5\%$. A comparison with the theoretical is not possible since the degree of non-linearity is an unknown factor. If the actual instrumentation non-linearities were known, then the non-linear effects could be analyzed. Unfortunately, there is a coupling of the non-linearity due to the particular patient under study. The stray capacitive effects for each patient are highly variable and unanalyzable by conventional methods. For this reason, we were only able to show that the calibration techniques were good in theory, but still can not use them for calibrating an instrument. Since the calibration procedures appeared to have some basic validity in theory, and since significant information had been obtained with the pneumatic transfer system, it became apparent that a new transducer should be constructed to eliminate the problems encountered with the two earlier systems. The basic problems to overcome were:

1.) Make the transducer and electronic system linear to within a few percent.

2.) Eliminate the effects of stray capacitance fields.

3.) Make certain that the dielectric constant of air is not appreciably affected by temperature or pressure.

4.) Improve the effects of gross movements and positional effects if possible.

6.3 Linearized Capacitance Pressure Transducer

Any new transducer must overcome the problems indicated above. The stray capacitive fields can be eliminated by making the entire outside surface of the transducer a common ground. By this method all stray fields are immediately grounded, and consequently can not effect the internal operation of the system. The transducer was constructed to accomplish this end.

The transducer as developed is seen in Figure 6.3c. The difference between the new unit and the old unit, lies in the relationships of the elements of the transducer. From practical consideration, it is a known fact that in this type of single element capacitor, the measuring electrode (Positive Electrode) must be small in cross section as compared

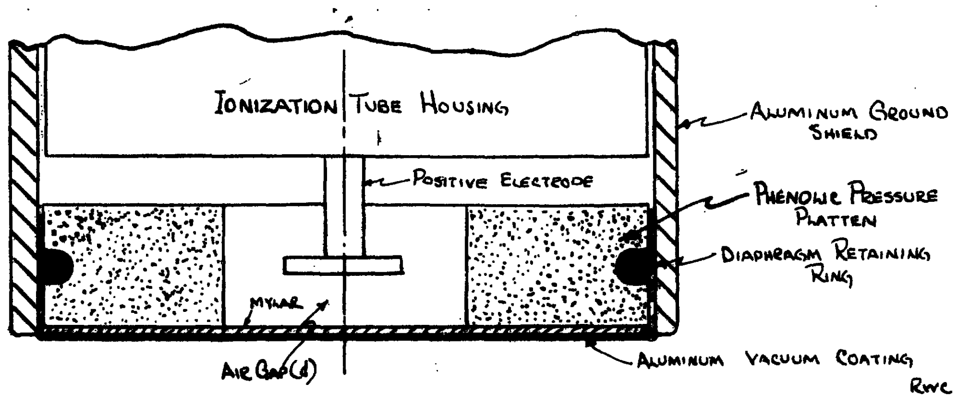


Figure 6.3c

to the dilative diaphragm. To assure this, the diaphragm was made so that its area was 250% of the positive electrode area. The Aluminum coating of the diaphragm was used for two important purposes. First, it provided a continuous external ground for the system. Second, it formed a dilative diaphragm which could serve as a reference for the positive measuring electrode.

The pressure platten was made of phenolic plastic to minimize the capacitive fields in the region near the measuring electrode. The linearity of this transducer was discussed earlier, and it was found to be of the order of 3-5%, within the limitations of the linearity assumptions. This transducer is shown in an exploded view in Figure No. 9.

The dielectric strength (ϵ) of dry air is a function of pressure and temperature. It is given by

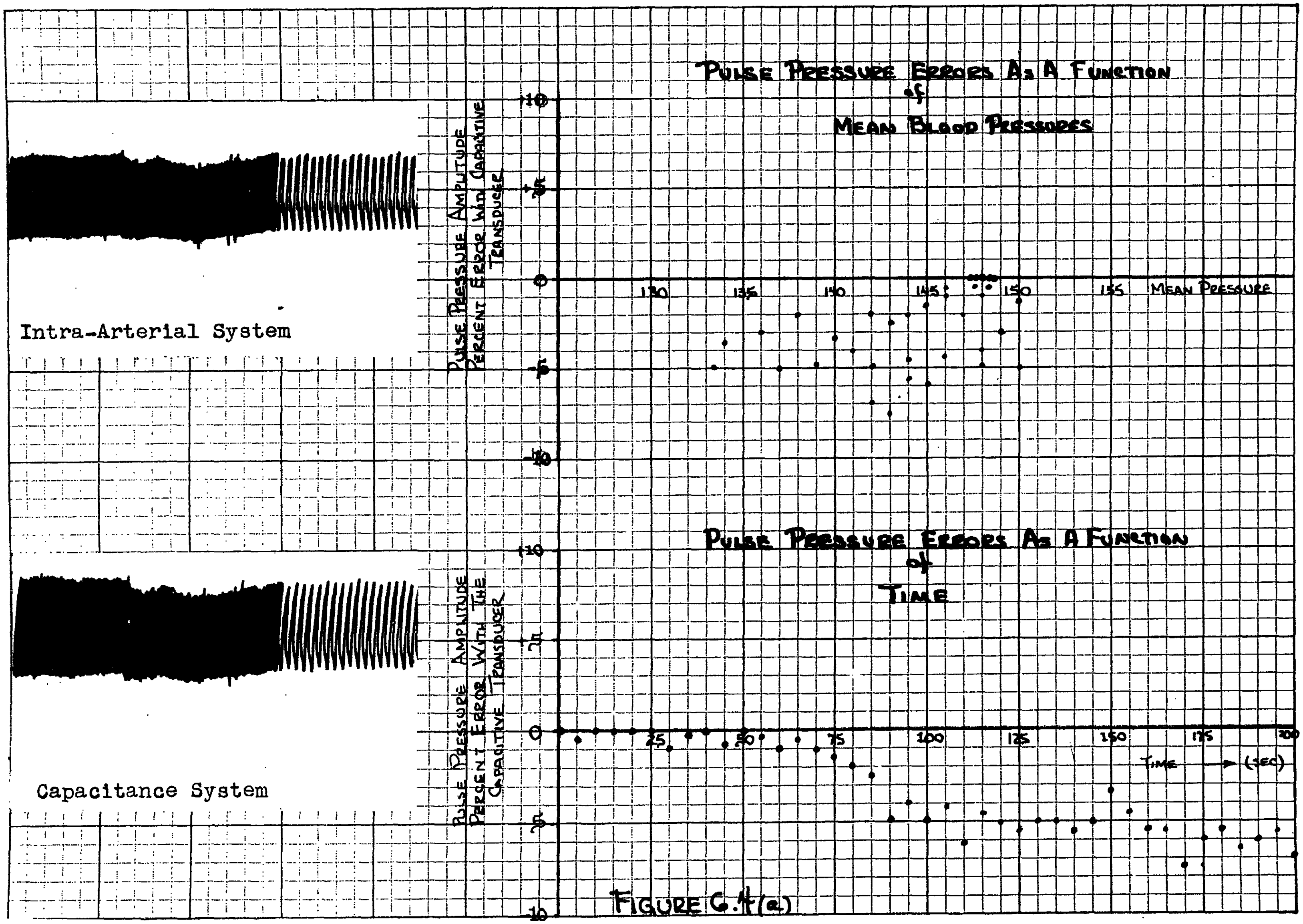
$$(\epsilon - 1) = (\epsilon_o - 1)\frac{P}{P_o(1 + 0.003411(t-20))}$$

where P & P_o are pressures & t = temperature in degrees (C).

where, $\epsilon_o = 1.000537$ for standard conditions, it is obvious that for ordinary temperature and pressure variations this is constant.

Various methods of evaluation were used to indicate the effectiveness of the linearized theory. One of the problems that creates difficulties is that we are dealing with life and with the human body. We have felt that it is essential to study the overall problems under the conditions that will be met in practice. When we wish to evaluate a particular effect, we must first wait until there exist a clinical operation which by its nature demands accurate intra-arterial blood pressure data. These conditions are infrequent, and therefore, it is a lengthy process to obtain the conditions for evaluation. Once having the justification for obtaining intra-arterial data, we may not obtain the data that is necessary. For example, if we wish to study the new system under variable pressure conditions, we must continually record data by the two measuring systems until that particular case arises whereby variable pressure data is obtained. Many operations proceed without significant variation in blood pressure level. Only a few cases does there exist the variations of the order of 25 or 50 mm of Hg. This obviously presents an evaluation process of some duration.

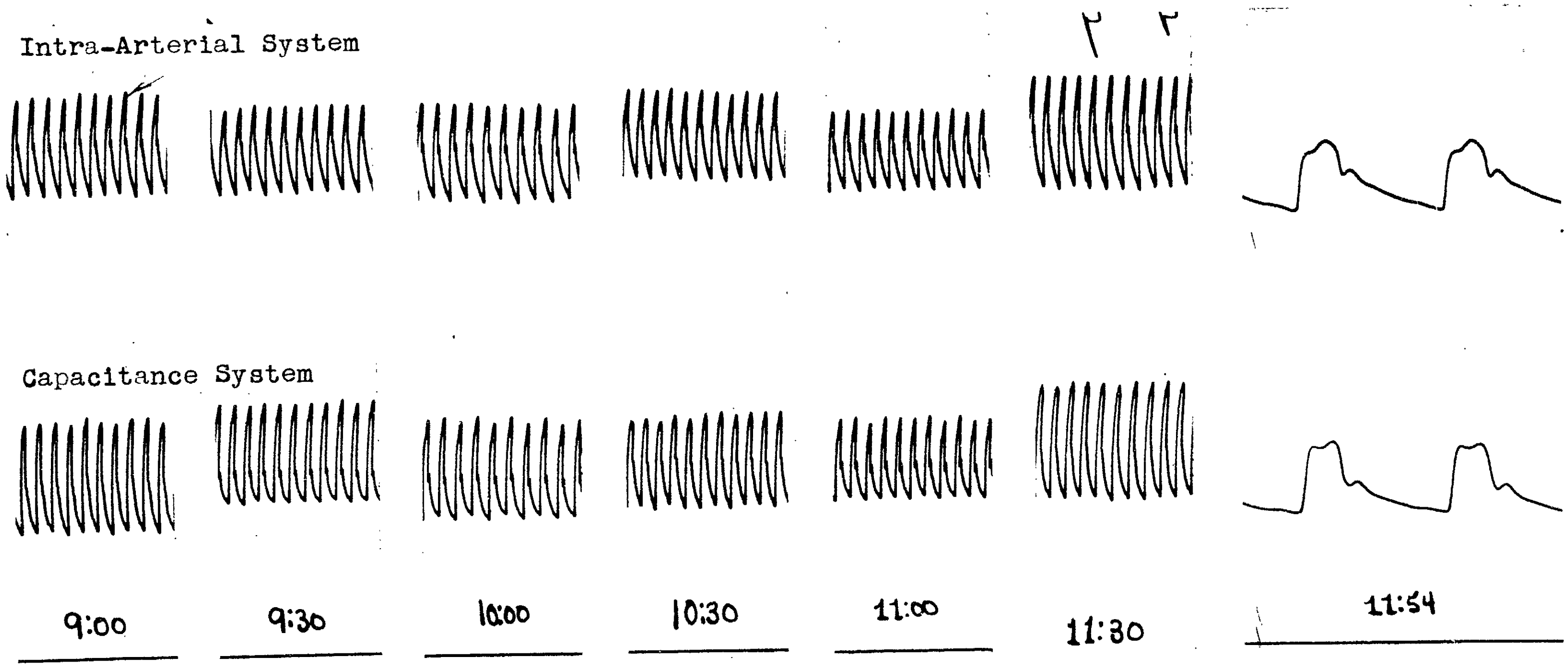
When those few cases do arise, of course valuable data is obtained. Numerous types of evaluations when used, only two are given here as an example of the errors encountered: First, the errors in pulse pressure are plotted against pressure levels. This provided no real answers as to the source of error. (See Figure 6.4 (a)). In general these errors were totally random. Next, this same data when plotted as a function of time (Figure 6.4 (a)), the real source of error is seen. The errors are in general time dependent. Similar curves were plotted for the mean pressure levels, this gives us information about the nature of the base line shift. For these analyses, the graphical tracings were



PULSE PRESSURE ERRORS AS A FUNCTION of MEAN BLOOD PRESSURES
PULSE PRESSURE AMPLITUDE PERCENT ERROR WITH CAPACITIVE TRANSDUCER
10
5
0
-5
-10
130
135
140
145
150
155
MEAN PRESSURE
Intra-Arterial System
PULSE PRESSURE ERRORS AS A FUNCTION of TIME
PULSE PRESSURE AMPLITUDE PERCENT ERROR WITH THE CAPACITIVE TRANSDUCER
10
5
0
-5
-10
25
50
75
100
125
150
175
200
TIME → (SEC)
Capacitance System

FIGURE G.4(a)

COMPARATIVE STUDY 3 HOUR OPERATION



photographically enlarged to permit better resolution of the data. Follow this procedure, along with other types of data analyses, there always appeared a time depend source of error. These errors, though time dependent, were not repeatable. In one case, as in Figure 6.4 (a) the errors were negative and of the order of 8% within $3\frac{1}{2}$ minutes. In another case, the pulse pressure errors were only 6% negative after 20 minutes. While still another case, had a pulse pressure error of plus 8% over a period of four minutes. One of the most interesting cases, was one where an analysis of the data was taken at intervals over a one and one-half hour period. At the beginning the data as analyzed, and the pulse pressure errors were of the order of 4% high, during the analysis of the data, this drifted from a positive error to a negative error and back to positive again. The errors at the end point, were only about 3% high. The base line shift during this time was of the order of 5%. The interesting aspect of this particular data, (See Figure 6.4 (b)) is that during this time the patient was given medication which limits the effectiveness of the device. In particular, certain vaso-active drugs activate the vasomotor system such that our assumption of a purely elastic system of tissue is invalid. In this patient, the capacitance recording system did not follow the actual blood pressure during medication, however, after the drug effects wore off, the recording came back to within 4% on the pulse pressure readings while the base line shift was of the order of 5%. The effects of vaso-active drugs can be seen Figures 6.5 and 6.6. The vaso-relaxant drugs tend to decrease the central pressure by relaxing the arterial wall. This has the opposite effect on the capacitance recording device, since the "elastic" constants are decreased by this "relaxation", the surface deformations are increased in magnitude. The converse effects

Effects of Vasorelaxant Medication on the System *

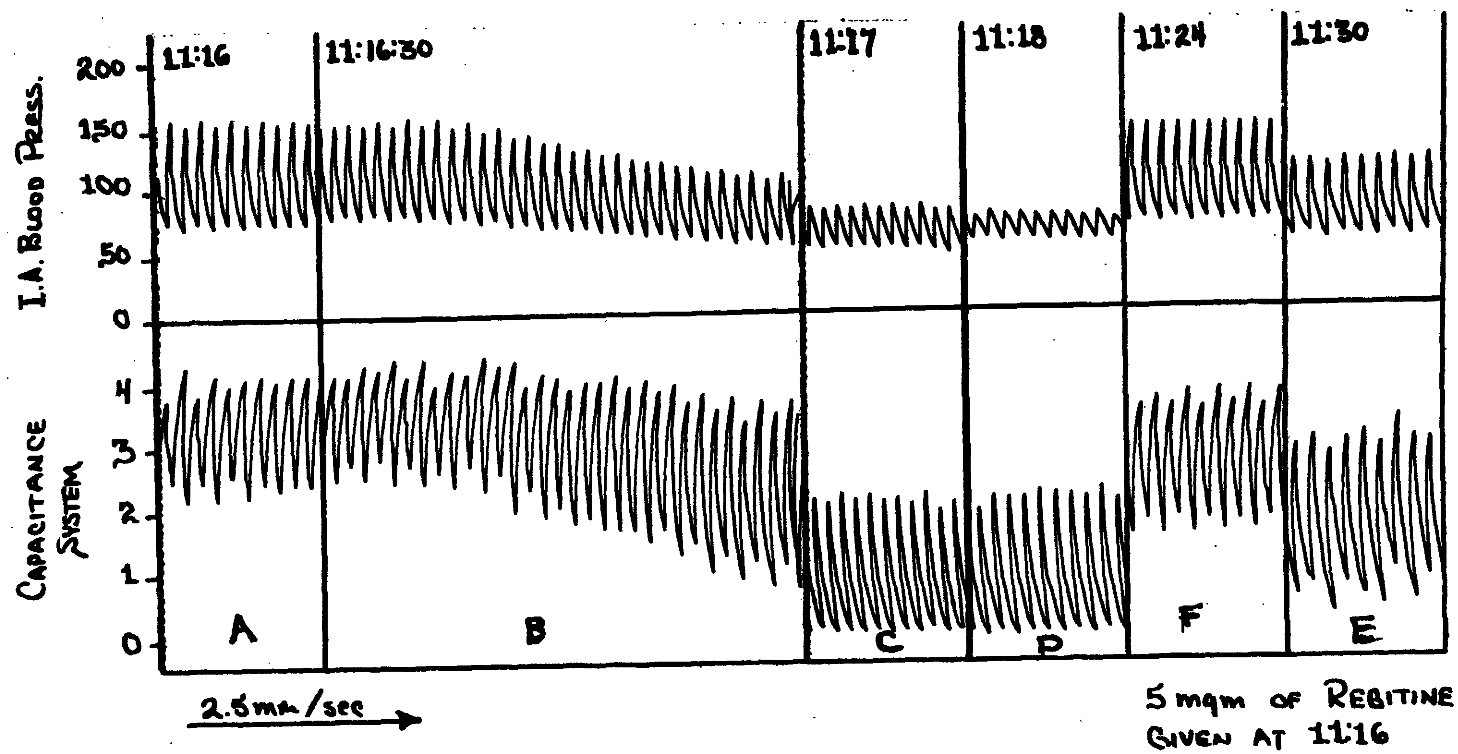
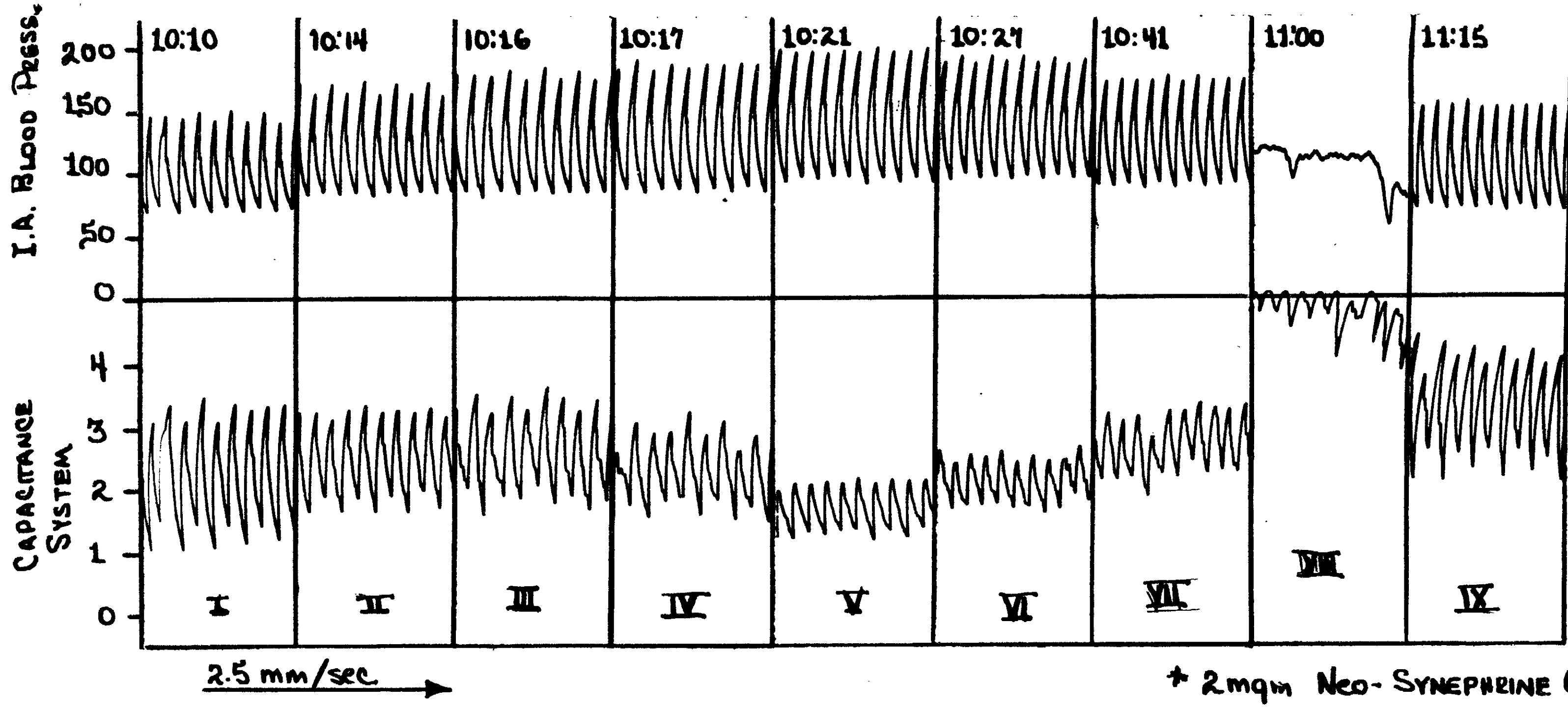


Figure 6.5

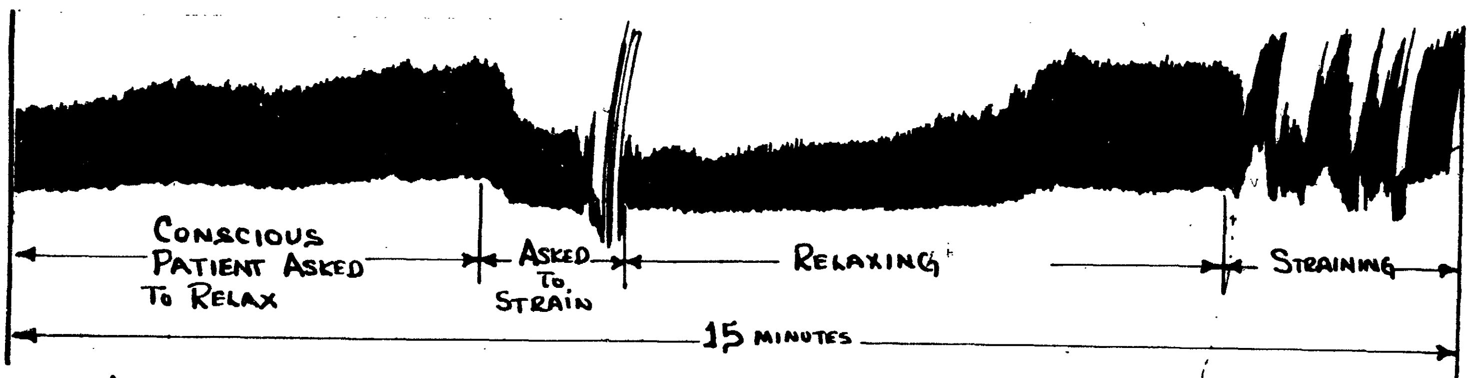
Effects of Vasopressor Medication on the System*



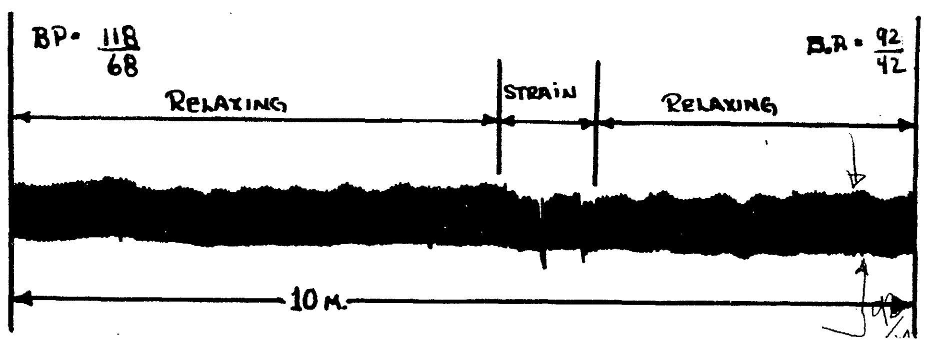
* 2 mgm Neo-Synephrine Given to patient at 10:10

Figure 6.6

STABILITY UNDER TWO CLINICAL CONDITIONS



A CONSCIOUS PATIENT IN SUPINE POSITION - NOTE EFFECTS OF MUSCULAR RELAXATION

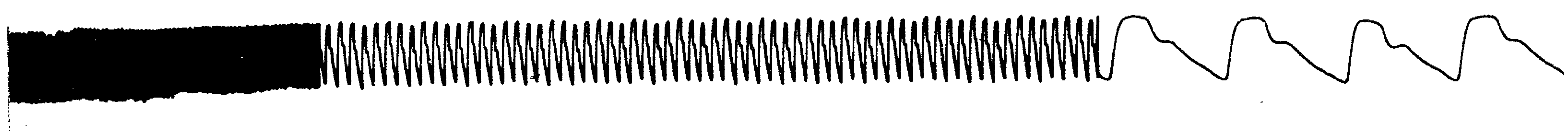


A CONSCIOUS PATIENT IN SUPINE POSITION - NOTE MINIMAL ADVERSE EFFECTS

FIGURE 6.7

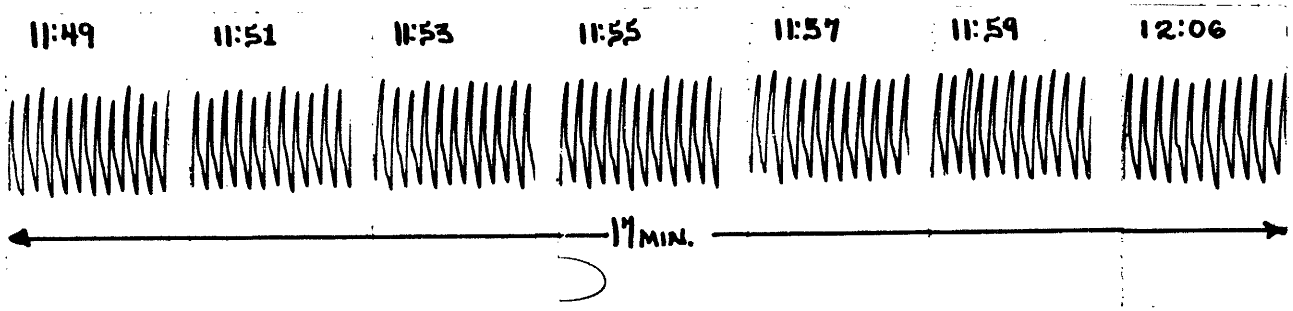
Stability Under Two Clinical Conditions

APR 1 2 1959



Case # 4980

An Anesthetized Patient In Supine Position - During Operative Procedures



An Anesthetized Patient In Supine Position - No Relaxation Effects are seen for vaso-pressor drugs. In this later case, the vaso-motor system is constricted so as to increase the central pressure and this action tends to increase the "elastic" constants so that reduced surface deformations are realized.

The time dependent errors are still not completely understood. The reasons for these time dependent errors are complex. There appear to be two major sources for this type of error. The first is what we are calling "muscular relaxation". In terms of the system, it means that the transfer coefficients of the elastic network are varying with time. Their variance is a complex function of the status of the physiology of the body. These effects can be seen in Figure 6.7. In this tracing we have two conscious patients under the same physical conditions, yet the responses are entirely different. In the upper tracing the recording varied in relation to the self-induced straining of the patient, and in the lower tracing. the variance is greatly reduced during straining. The net effect is to alter the output characteristics of the capacitance recording. The precise way in which these characteristics are altered, is of course very difficult to evaluate without further and more extensive investigations. Two other clinical conditions are shown in Figure 6.8. Here the patients are both under an anesthesia, and even though there are significant operative procedures in process, there appears to be little if any muscular relaxation effects. With further work, this effect will undoubtedly be more fully understood. It should be noted here that significant straining does alter the blood pressures in the extremities. In some cases, the muscular straining will completely invalidate any data obtained from an intra-arterial pressure transducer. This effect is shown here in Figure 6.9.

The second source of an error which can be time dependent, is a problem in transducer positional changes. These effects are major, and are very significant in the over-all performance of the device. Since we are relying on some transfer characteristics to give a blood pressure reading which is proportional to surface deformation, the recording transducer must be in the proper

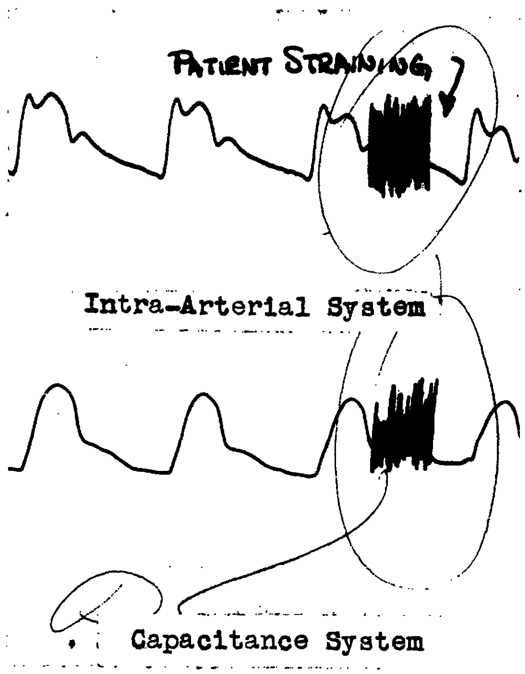


Figure 6.9

relationship to the pulsatile deformations in order that the transfer characteristic be linearly related to blood pressure. The physical orientation of the transducer to the artery appears to be very critical. It has been found that a trial-and-error method of positioning the transducer is the only method which can yield positive results. It is obvious that the transducer must be "right" over the artery, but there is more to it than that. Figure 6.9 * shows this effect very well. In tracing set I, the device is recording the pressures quite well, then the transducer was mearly lift and returned to what seemed to be the same position as tracing I. The result of this change is seen in tracing set II. The transducer was then reset, with the result shown in tracing

* The EKG tracings are given to show the actual correlation with heart response characteristics. The intra-arterial blood pressure was not obtained in this case, but the cuff pressures are given.

Effects Of Interface Position On The Characteristics Of The Pressure Monitoring System

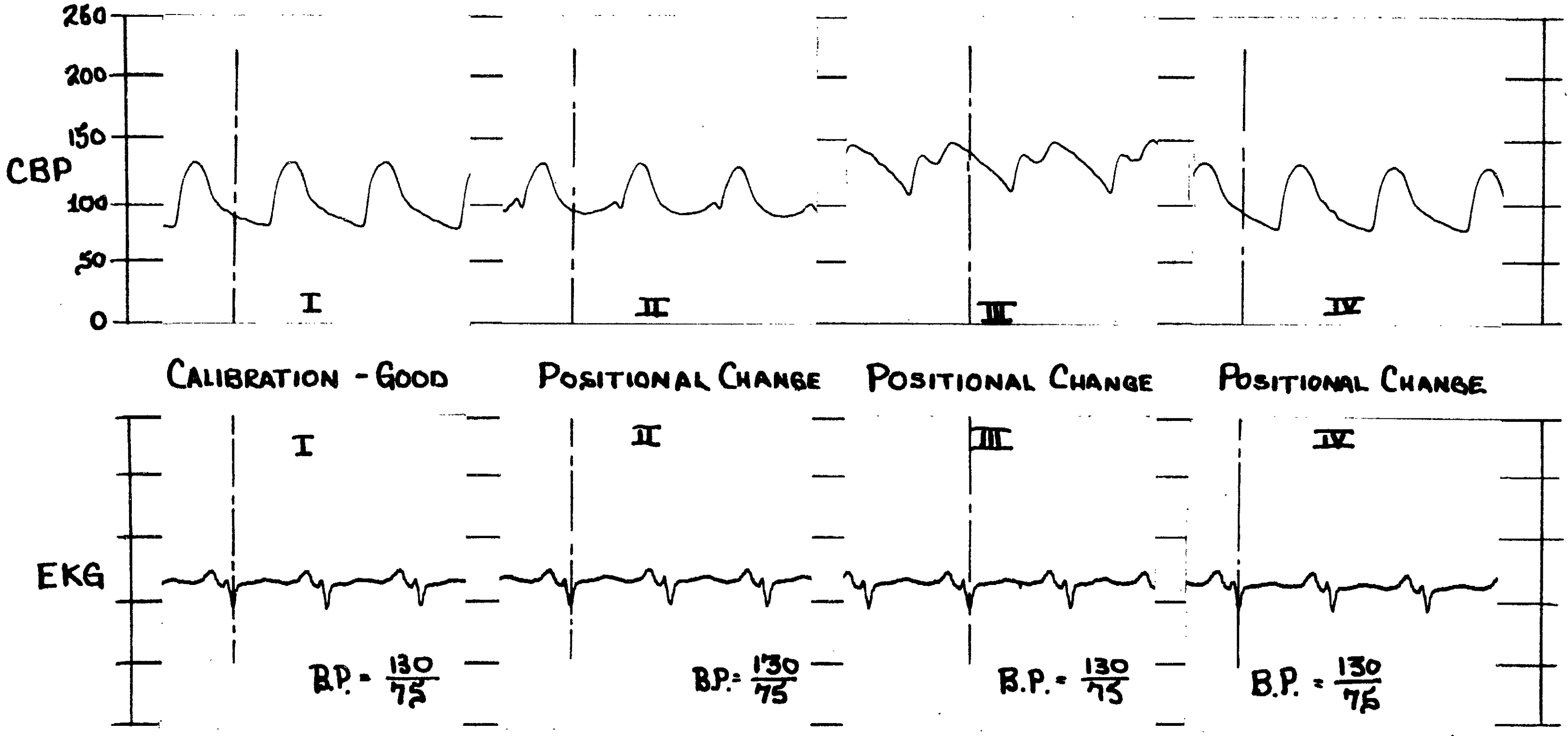


Figure 6.9

III. The transfer characteristics are altered completely by this position orientation. After a trial-and-error method, the tracing was restored to a likeness of tracing set I as shown in set IV. These positional effects are not easily controllable, in fact we have not been able to position the transducer by any other method except by this trial-and error technique. One of the reasons for this problem is that there are mechanisms within the body which operate to protect the vascular system from trauma. When a foreign pressure is exerted over an artery, the muscle tissue tends to "move" the artery in such a manner so as to reduce the foreign pressure. In effect, what you have is a greased pig which you are trying to hold down. One method to reduce this movement is to apply the corregated pressure platten as shown previously for the partial cuff. This method has not been applied to the present transducer, however, it is felt that it would aid in the reduction of this effect.

The capacitance transducer system was comparatively studied with the aid of an oscilloscope. Figure 6.10 shows some of these tracings. This type analysis permitted us to evaluate the order of magnitude of the phase shift in the transfer characteristics. This shift appears to be of the same order of magnitude as the isometric contraction of the heart, namely 0.05 seconds, or 34°. For research purposes this would be significant, however, for clinical purpose this shift is undetectable at the tracing speeds of the recordings. Superposition, gives some strong evidence about the damping characteristics of the system. To give any quantitative information about this damping, it will require a frequency analysis of both recordings. Since the actual blood pressure wave form contains numerous frequencies, it would be necessary to pin point those frequencies at which the response is damped. Qualitatively, the frequencies, at which damping effects are significant, can be seen.

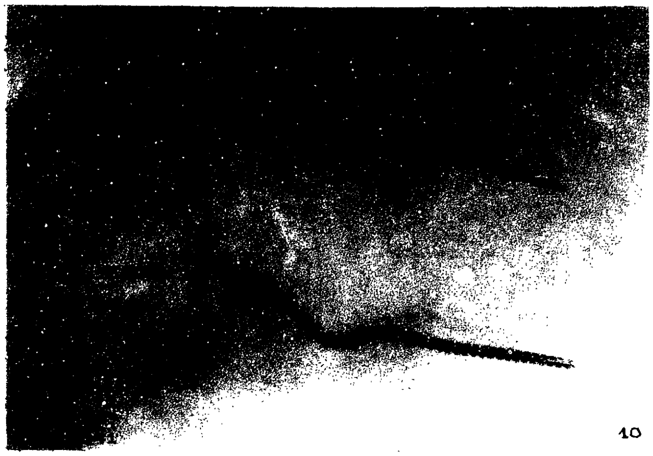
In the form of the diacrotic knotch where three inflection points exist, and relatively high frequencies exist (as compared to the 1.6 cycles/second for a complete pulse). Since a wave analysis is not presently possible, the only thing that can be said is that the damping of the system is such that the capacitance system is not able to follow all the frequency changes within the system. The diacotic knotch appears in a reduced manner, but differing with each particular patient studies. The types of variations in wave forms are shown in Figures 6.11 and 6.12.

One of the important tests of a system is to evaluate its response to variable clinical conditions. We have discussed some of the errors involved, but still we are faced with the problem of being unable to test the characteristic over a complete pressure range under clinical conditions. The only thing that we felt we can do for this pilot study, is to continue to take clinical data, and analyze the errors for each case, and from a great many cases deduce the kind of accuracy we can expect. There are analyses which will have to be pursued ultimately but they are beyond the scope of this microscopic investigation. Figures 6.13, 6.14 and 6.15 are included to indicate the kind of potentialities that may exist within this concept for blood pressures. Each of these recordings, when photographically enlarged, yield data to determine some of the inaccuracies.

6.4 Gravitational Effects

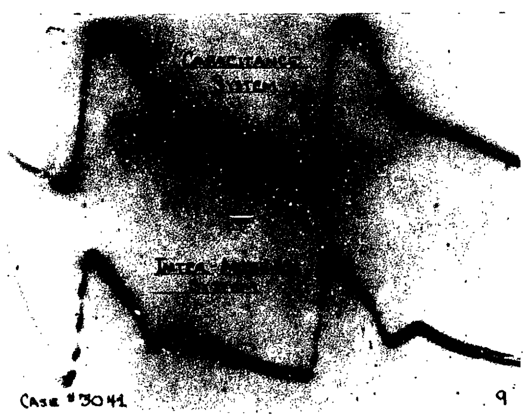
One of the major problems with the linearized transducer is that the calibration techniques have not been evaluated. Every attempt at quantitative evaluation has been attended by errors of an erratic nature.

OSCILLOSCOPIC COMPARATIVE STUDY

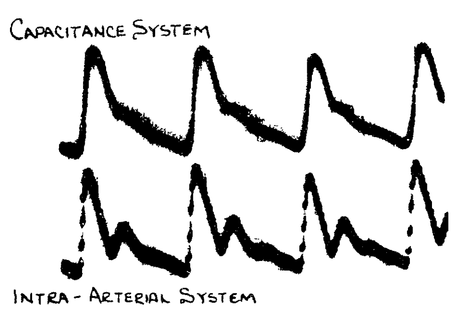


10

Oscilloscope Tracing No. 10



Oscilloscope Tracing No. 9



Case #3041

6

Oscilloscope Tracing No. 6

FIGURE 6.10

SAMPLE DATA

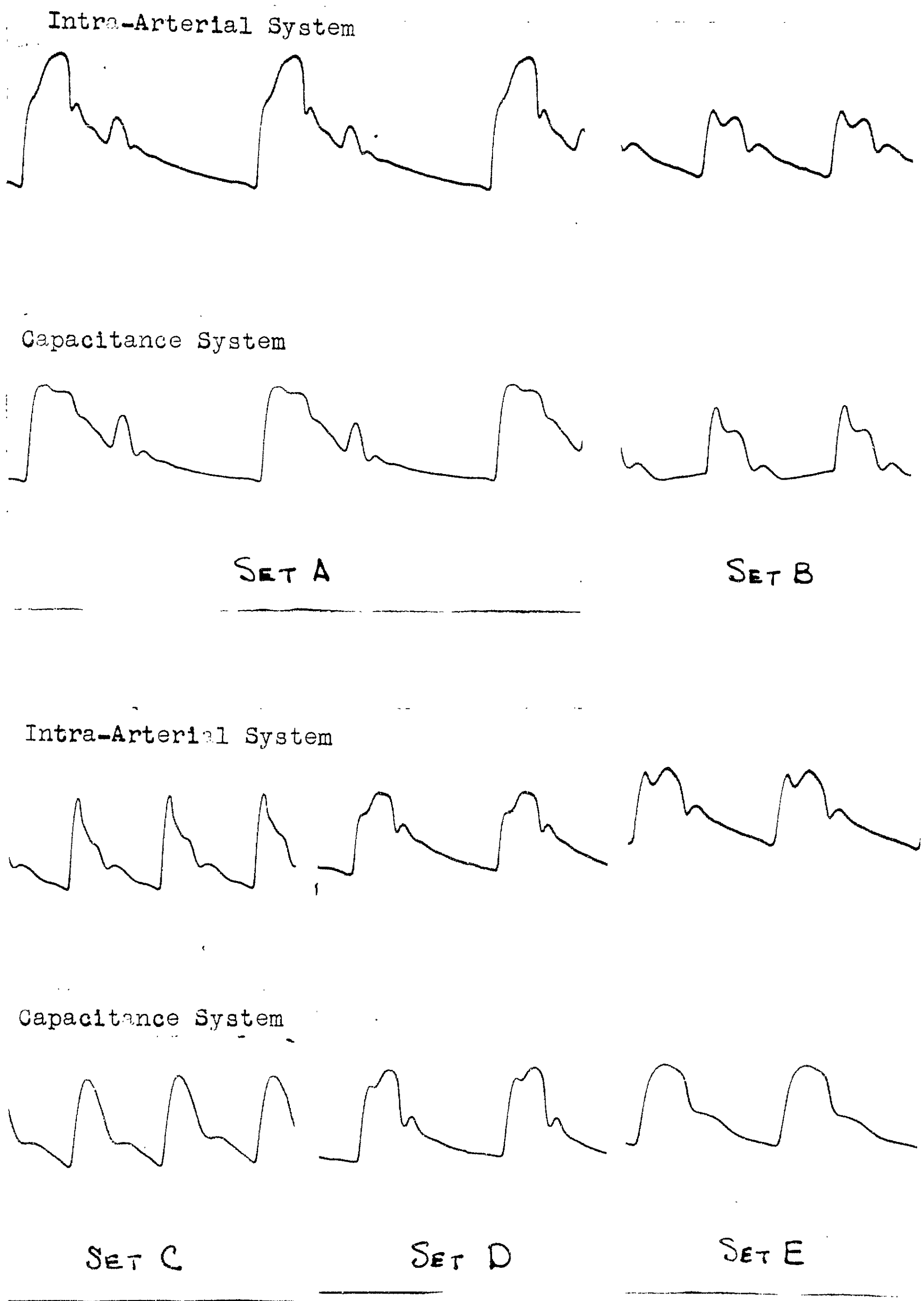


FIGURE 6.11

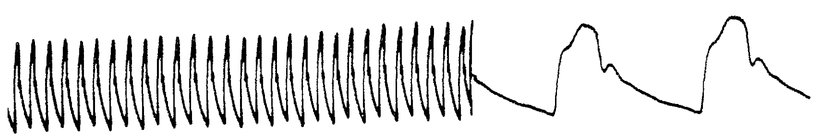
SAMPLE DATA

Intra-Arterial System

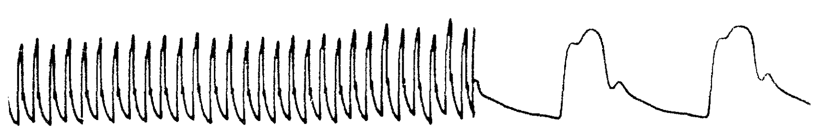
Capacitance System

SET F

INTRA-ARTERIAL BLOOD PRESSURE



CAPACITANCE BLOOD PRESSURE



SET G

CASE #6963

FIGURE 6.12

DATA TAKEN DURING VARIABLE CLINICAL CONDITIONS

Intra-Arterial System

Capacitance System

Intra-Arterial System

Capacitance System

Figure 6.13

DATA TAKEN DURING VARIABLE CLINICAL CONDITIONS

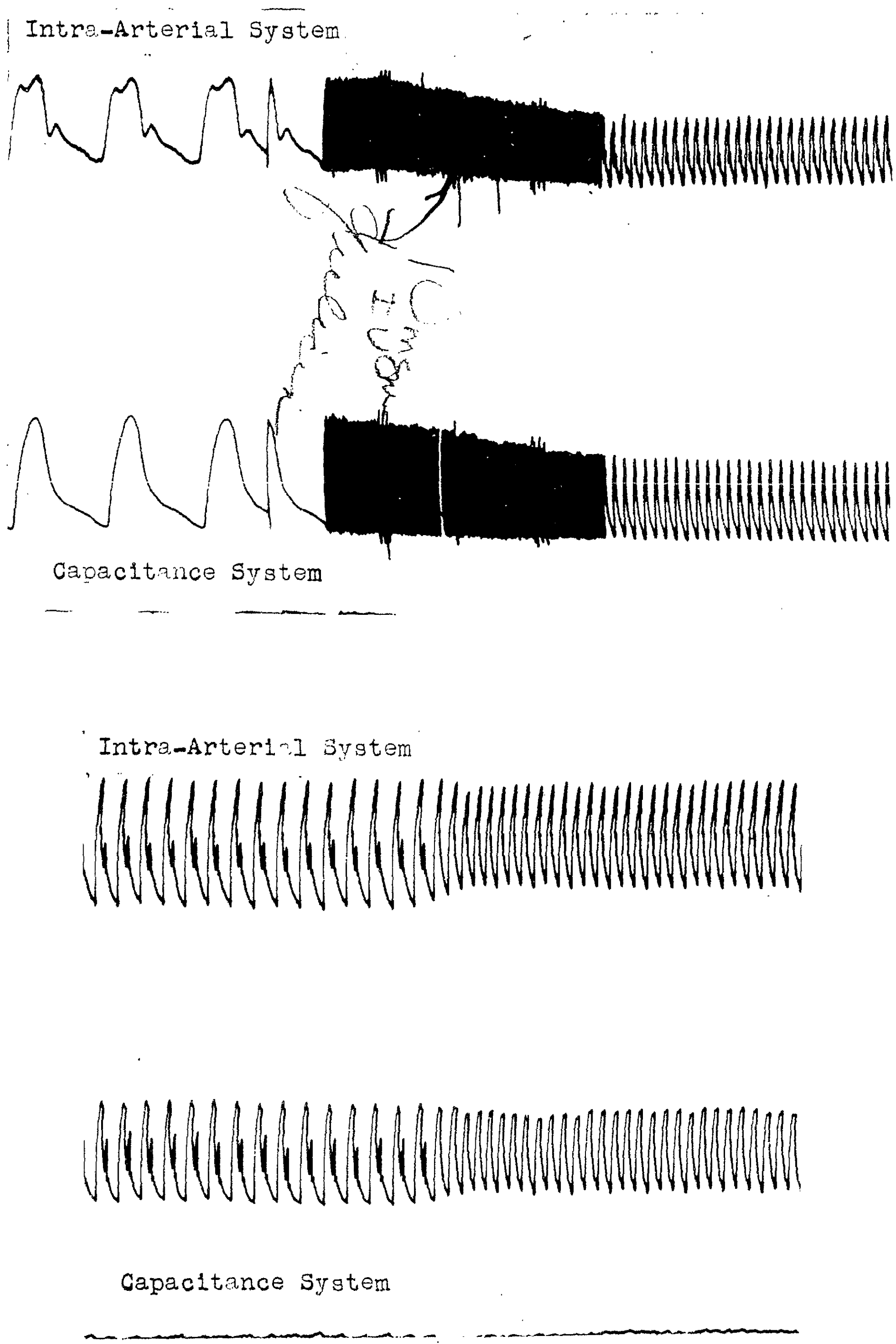


Figure 6.14

DATA TAKEN DURING VARIABLE CLINICAL CONDITIONS

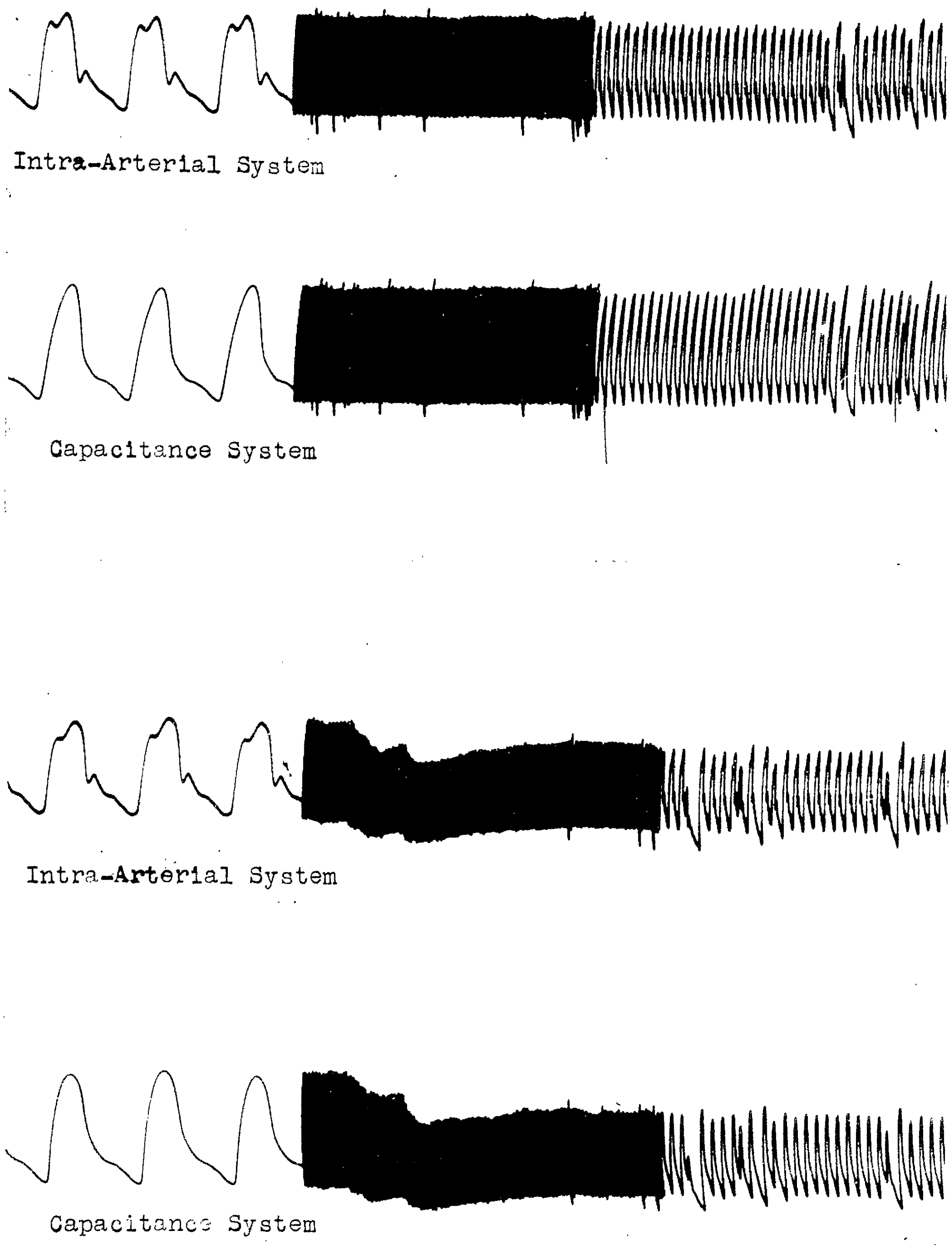
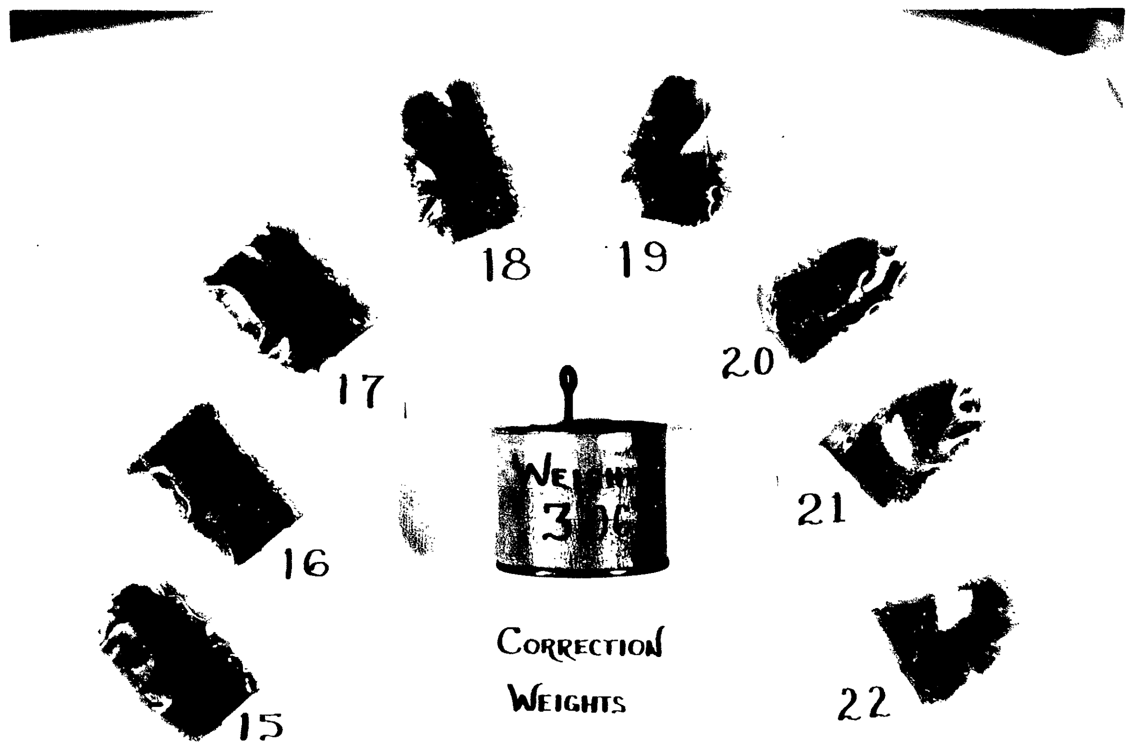


Figure 6.15

The transducer which weighs .474 pounds in total, contributes significantly to the instability of the system. At the outset, the transducer was attached to the holding assembly by a cantilever arm. The center of gravity of the system was located at 3.7 inches from the pivotal point. This meant that during an elevation change, the gravitation effects on the interface pressure were altered by the cosine of the angle of elevation. Various techniques were tried to overcome this gravitation field problem. First, the arm was positioned so that during elevation changes, the elevation effect was a plus and minus term about a horizontal axis. (See Fig. 5.1) This procedure did not work well. A free body diagram of the system yielded information which permitted us to make up special correction weights (P.12) to add just enough weight to offset the loss in interface pressure. This procedure likewise did not prove fruitful. Finally, the counter-balance was added so as to move the c.g. to the pivotal point. This procedure makes considerable improvements in the workability of the system. Unfortunately, it did not permit a quantitative evaluation of the calibration procedures. Qualitatively, the calibration procedure did cause the system to respond in the directions predicted by the theory, but other errors caused the system to operative ineffectively.

The source of error during calibration appears to be, of positional origin. The mass of the present system is such that it requires considerable holding fixtures to maintain stability. This stability is obtained by mounting the transducer on an arm holding board, (see P.11) and this board is in turn fastened to the operating table. So, in effect, the transducer is held stable by the operating table. This presents another aspect of the problem, during calibration procedures,



P.12

the arterial section under consideration must be moved so that its elevation is altered by a predetermined distance. In doing this, however, the patient's arm and the transducer must be moved to accommodate this level change. In moving, the relationship between the transducer and the patient is inevitably altered, and consequently the positional effects mentioned earlier are major. The present transducer mass will not permit a change in the holding method to eliminate this effect. The author has tried many methods for effecting the level change, and none have proved completely effective. We have tried everything from new holding fixtures, to numerous techniques of level changes, like hydraulically controlled arm boards for closely controlled level changes. It appears that the calibration procedure will have to be evaluated with a transducer system of entirely different design. With an extremely light transducer system, the holding device can be arm bound instead of table bound, thus the transducer will follow the arm and it is believed that the characteristic will not be affected by any relative movements between the patient and that upon which he is resting.

6.5 General Limitations and Problems of the System

It would appear that there are four basic limitations of the system as it exists.

1.) Positional Adjustments: It is subject to unpredictable response characteristics due to the manner in which the system is positioned to the subject. This is a very serious problem which must be more thoroughly understood. In one position it will function in a most acceptable manner, in another position, will not function. There have been cases where a great deal of time has been spent just attempting to find that position for effective operation. For the most part, given enough time and patience, the system can be properly adjusted, but unfortunately the parameters for this "proper" adjustment are still not fully understood.

2.) Vaso-active effects: There have been enough cases where induced vaso-activity has significantly altered the characteristic responses, that one must raise serious questions about the possibility of self-activated vasomotor activity. At this point, this fact is still under study, but no conclusive data has been obtained. The time dependent drift could easily be affected by this type of activity. Since vaso-motor affects may be present, only with a much more extensive evaluation program will this question be answered. There have been enough cases where stability was obtainable, that this limitation should not be considered in a completely negative manner. Further research will yield the answers to these most difficult questions.

3.) Gross Movements: There have been considerable limitations imposed by extraneous movements. With application of the same basic engineering design analysis, this problem can undoubtedly be eliminated. These gross movements, due to muscular activity, were the most limiting problem early in the program, however, they are very minimal at present. With such improvements as the constant force spring, proper interface pressure, and stable holding assemblies, this problem has virtually been eliminated. It must be always regarded as a possible source of error, however.

4.) Non-linearities: The system has some inherent non-linear behavior. These non-linearities must all be completely identified. We have identified these errors to some extent, but a more thorough program is necessary. Like most research programs, there is a long bridge between the concept and the total evaluation of that concept. This question of non-linear behavior may not be completely settled until some of the more limiting problems are solved.

VII. CONCLUSIONS

7.1 The Future

One of the next steps should be the construction of a new transducer, Some of the technical problems encountered, can most likely be overcome by proper design. One of the most serious problems is the effect of the mass on the system. Earlier we discussed the significant improvements that were realized when a counter-balance was added. This improvement in the response characteristics is felt to be directly related to the reduction of gravitational effects. The analysis given did not indicate all the effects attributable to the mass of the transducer. The changes in the angle of elevation contribute errors in two dimensions. The center of gravity lies 3.7 inches from the pivot point. and 1.25 inches above the arm interface. This geometry can adversely influence the stability of the system. The elevation changes cause a reduction of the interface pressure by 5% or 10% depending upon patient's arm length. Secondly, a moment is created which is proportional to the sine of the elevation angle. The moment is of the order of 0.1 in-lb which can have an adverse affect on the initial calibration position.

Another influence, is the $m\ddot{x}$ term in the dynamics analysis of the muscular movement absortion system. Recalling that the purpose of the pressure platten is to maintain constant interface pressure by causing the entire transducer to oscillate in accordance with the muscular disturbances. Obviously, there are accelerations in this process, and the response to an acceleration is a d'alembert force proportional to $m\ddot{x}$.

Still another source of difficulty arises due to the necessity of a table-bound holding assembly. This is necessitated by the magnitude of the mass of the transducer.

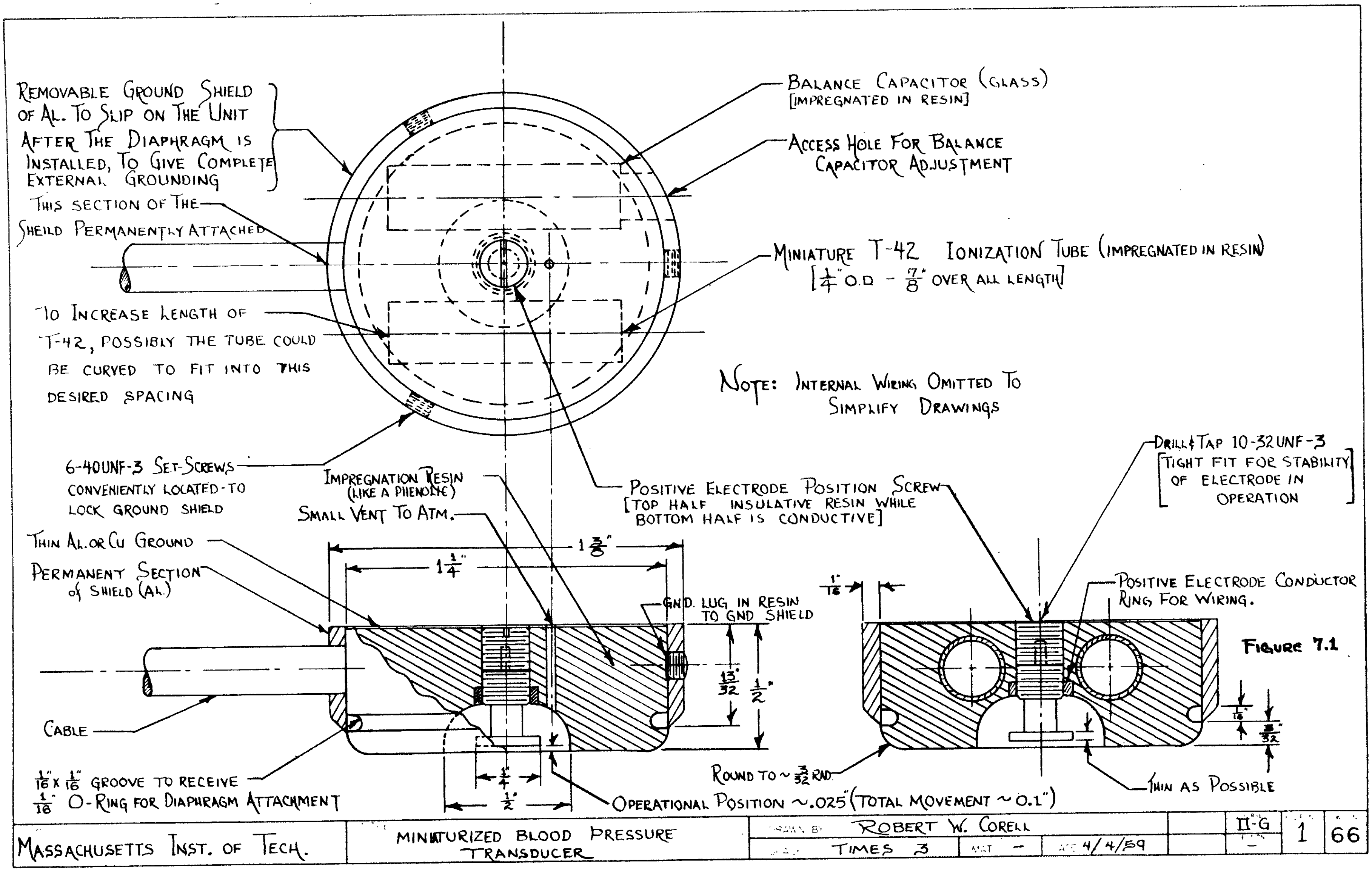

Removable Ground Shield of Al. To Slip on The Unit After The Diaphragm is Installed, To Give Complete External Grounding
This section of The Sheild Permanently Attached
To Increase Length of T-42, possibly the tube could be curved to fit into this desired spacing
Balance Capacitor (Glass) [impregnated in resin]
Access Hole For Balance Capacitor Adjustment
Miniature T-42 Ionization Tube (impregnated in resin) [1/4" O.D - 7/8" over all length]
Note: Internal Wiring Omitted To Simplify Drawings
6-40UNF-3 Set-Screws conveniently located to lock ground shield
Impregnation Resin (like a phenolic)
Small Vent To Atm.
Positive Electrode Position Screw [top half insulative resin while bottom half is conductive]
Drill & Tap 10-32UNF-3 [Tight fit for stability of electrode in operation]
Thin Al. or Cu Ground
Permanent Section of Shield (Al.)
1 3/8"
1 1/4"
Gnd. lug in resin to gnd shield
13/32"
1/2"
1/16"
Positive Electrode Conductor Ring For Wiring.
Cable
1/16" x 1/16" groove to receive 1/16" O-Ring for Diaphragm Attachment
1/4"
1/2"
Round to ~3/32 rad.
Operational Position ~.025" (Total Movement ~0.1")
1/16
3/32
Thin as Possible
Massachusetts Inst. of Tech.
Miniaturized Blood Pressure Transducer
Robert W. Corell
Times 3
4/4/59
II-G
1
66

Figure 7.1

All of these problems would appear to be reduced in magnitude if the mass of the transducer were reduced. Furthermore, the c.g. of the system should be as close to the interface as possible. This suggests a philosophy for design.

Incorporating the design philosophy suggested by the problems of mass interactions, and by considering some of the electronic variables, a new transducer was designed. It is shown in Figure 7.1.

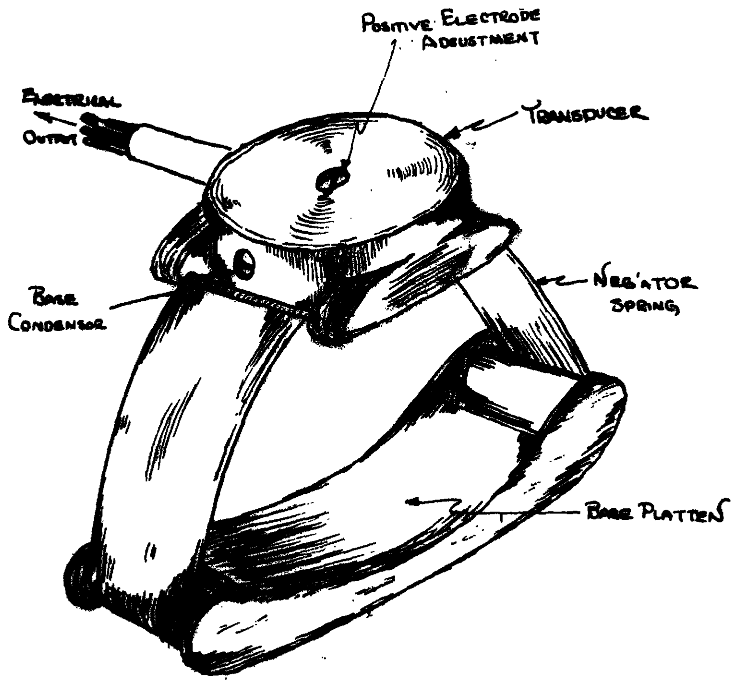


Figure 7.2

With this design, it is felt that significant progress can be made to permit a more complete evaluation of the theory. The transducer and holding assembly are sketched in Figure 7.2. The application is to be totally arm-bound, resembling a wrist watch.

With this transducer, the following evaluations would be essential before any broad conclusion can be made about the potentialities of the system.

1.) <u>Calibration</u>: The calibration procedure must be fully checked.

The theoretical calibration procedure is still in question. Another calibration procedure is available provided absolute accuracies are not necessary. For relative pressure changes, the transducer can be calibrated against the Sphygmomanometer. This will yield a calibration which is in error, however, the relative changes in blood pressure can readily be seen. This later procedure has been used with considerable success.

2.) Dynamics: The dynamic response of the system must be analyzed more thoroughly. This will involve a wave analysis to determine the actual damping ratio and phase shift for the total system, including the arm. Furthermore, the natural frequency, damping ratio and other dynamic characteristics of the transducer must be known.

3.) Linearities: The non-linearities of the physiology should be analyzed. Once the precise characteristics of the instrumentations are known, then an input-output study can yield some of these physiological unknowns.

4.) Stress Function: In the analysis of the transfer characteristics between the artery and the arm surface, we only proved that the relationship was linear. Future work on this project should include a more detailed analysis of this consideration. If possible, the stress function Φ should be found and the full deflection characteristics identified.

5.) Clinical Application: The system should be continually applied to clinical conditions such that some of the unknowns can be studied and possibly identified. It is felt that much is to be gained by this kind of study.

6.) Vaso-activity: The effects of vasomotor control must be totally identified. If these effects are major, obviously the system is ineffective. If vasomotor effects are significent in the extremities, then the transducer should be applied to the temporal artery where vasomotor effects are known to be very minimal. Recording from the temporal artery will create calibration problems, but they will have to be faced.

7.) Re-Analysis: Should the majority of the previous analysis yield positive results, then the system should be subjected to a complete re-analysis of a very extensive nature. This is necessary to prove the absolute practicality of the system for clinical use beyond a research oriented program.

7.2 Potentialities

This pitot study has indicated potentialities under controlled conditions. At present, the system is quite inadequate in application and design. Some of the data is encouraging and remains as evidence that there is a significant potential to this concept. At the moment, it is completely impractical, however, with continued study, these impracticalities may be only due to poor application of the theory and not inadequate theory. Others should analyze the theory and consider different methods of applying the theory to ultimately realize a workable instrument for recording human blood pressures.

If the instrumentation is ultimately developed and proved adequate, then it is conceivable that other potentialities exist in a research and clinical sense. Present results encourage one to believe that new information about the volumetric distention of human arteries in living persons, as opposed to studies of sections taken at postmortem, may now be possible. A plot of pressure changes versus volumetric distention in the human body can conceivably be made, using the proposed theory. Furthermore, information about the elastic nature of an artery and of the skin that lies between the artery and the surface may be extracted from the matrix of uncertainty which has engulfed previous efforts to separate these components from those of muscle. Although I am not proposing to indicate precisely how these studies should be done at the moment, they are in need of answer in the future and the fundamentals and techniques collateral to the approach here described makes such analyses theoretically and practically possible. The investigators are sure, that if there is validity to the basic approach in this system, then there are still other avenues of application, as yet unexplored, for solving medical-technological problems in the instrumentation and recording of physiological phenomena in the human body.

APPENDIX A

Equipment Characteristics

STRATHUM PRESSURE TRANSDUCER (Model P23G, Serial No. 1736)

Pressure Range: 0 to 75 cm of Hg.

Sensitivity: 46.5 microvolts per cm of Hg.

Natural Frequency: With distilled Water @ 75°F and #20 x 5 cm needle, approximately 185 cps.

Critical Damping Ratio: 0.07 under same conditions as f_n.

GRASS 5P1 PRE-AMPLIFIER (Supplied by Manufacturer)

Frequency Response: Linear from D.C. to 40 cycles per second.

Drift: Less than 3 microvolts per hour-random.

Sensitivity: Accurate to within 1%.

Noise Level: referred to the input at 200,000 ohms is not more than 3 microvolts.

Calibration Reference: by 2 millivolt D.C. manually generated signal.

Balance Voltages: Accurate to within 1%.

GRASS MODEL 5 DRIVER AMPLIFIER AND WRITER COMBINATION

Frequency Response: Linear with 5% from 0 to 45 cycles per second.

Peak to Peak deflection: 60 Millimeters.

Linearity of deflection: Linear with 2% over range of 50 Millimeters peak to peak.

Chart Speeds: 100, 50, 10, 5, 2.5, 1.0, 0.5, and 0.25 mm/sec.

Chart: Curvilinear

TEKTRONICS TYPE-530 OSCILLOSCOPE

Frequency Response: D.C. to 10mc (Rise time 0.03 usec.)

(With dual signal input)

Incremental Accuracy with .2% full scale

Sweep Rate: Variable from .1 usec/cm to 12sec/cm

Accuracy of Sweep Time: 1%

SORENSON MODEL 1000S VOLTAGE REGULATOR (FOR ELECTRONIC EQUIPMENT)

Voltage: 118 Volts ± 0.10 Volts

Wattage: 1000

RCA WV-98A VTVM (For Voltage Calibration of All Equipment)

Full Scale Accuracy: 3%

Frequency Response: Flat to 30 cycles per second.

DECKER MODEL 902-1 Delta Unit

Output Signal: Max. ± 30 Volts.

Sensitivity: 2 Volts/mmf at balance point (C 10mmf).

Frequency Response: D.C. to 750 cycles/second with 42" cable

Drift: Less than .03% of full scale per hour random, with .05% full scale per 24 hours.

Stability: .1%

Linearity: With ideal capacitor .1%, typical capacitor 1%.

Noise Level: 10mv in 0-10,000 cycles per second freq. band.

ALTEC MODEL M-20 MICROPHONE SYSTEM

Frequency response: 2 cps to 18 Kcps

Rise time: 50 microseconds

Sensitivity: -56 db re 1 volt/dyne/cm^2

STROMBERG-CARLSON MODEL AV-70 AMPLIFIER

Power: 3 watts

Sensitivity: .0035 volts

Frequency response: Compatable with Altex system

<u>DECKER AVIATION CARDIODYNAMETER (Sensitive Pressure Meter)</u>

Natural Frequency: Approximately 100 cps (in the pneumatic system in the study contained herein)

Damping Ratio: Approximately .15

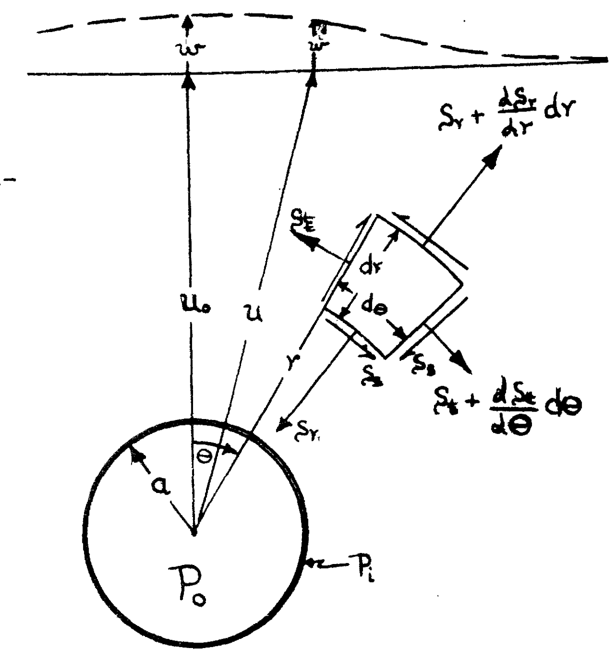
Frequency Response: ± 5% Amplitude and ± 5° Phase distortion throughout .05 to 75 cps

Sensitivity: 7.5 millivolts/micron of Hg.

APPENDIX B

LINEAR DEFORMATION THEORY -

Consider the theoretical model sketch at the right. We postulate a homogeneous, isotropic elastic medium. The semi-infinite flat plate is subject to surface deformations due to internal pressure Po. The assumptions are:



1.) Po Blood Pressure creates an arterial deformation Proportion to Po with radial variations (a). Deformation theory due to King (54).

2.) Arterial Deformations create radial pressures Pi between the artery and the elastic tissue media. This pressure Pi is proportional to Po.

Now, consider the $dr d\theta$ element and write the equations of equilibrium.

RADIAL EQUILIBRIUM:

The net unbalance force due to S_r is given by:

$$+ \frac{\partial}{\partial r}\left[S_r r d\theta\right] dr = \left[S_r + r\frac{\partial S_r}{\partial r}\right] dr d\theta$$

The net unbalance force due to S_s is given by:

$$+\frac{\partial S_s}{\partial \theta}\, dr\, d\theta$$

The net unbalance force due to S_t is given by:

(Assuming small angles and neglecting second order effects)

$$- S_t\, dr\, d\theta$$

Then the summation of forces in radial direction $\left(\sum F_r = 0\right)$ is:

$$S_r - S_t + r\frac{\partial S_r}{\partial r} + \frac{\partial S_s}{\partial \theta} = 0 \qquad \text{(a)}$$

TANGENTIAL EQUILIBRIUM:

The net unbalance force due to S_t is given by:

$$+\left(\frac{\partial S_t}{\partial \theta}\right) d\theta\, dr$$

The net unbalance force due to S_s is given by:

$$\frac{\partial}{\partial r}\left(S_s\, r\, d\theta\right)dr = \left(S_s + r\frac{\partial S_s}{\partial r}\right) dr\, d\theta$$

Then the summation of forces in radial direction $\left(\sum F_t = 0\right)$ is:

$$\frac{\partial S_t}{\partial \theta} + 2 S_s + r\frac{\partial S_s}{\partial r} = 0 \qquad \text{(b)}$$

After Airy (*), we can define stress function Φ, such that S_r, S_t, and S_s can be defined in terms of Φ. Φ is purely a function of (θ) and (r). In polar coordinates, the three stresses can be defined by:

$$S_t = \frac{\partial^2 \Phi}{\partial r^2} \qquad \text{(c)}$$

(*) See A more complete analysis in Advanced Strength of Materials, J. P. Den Hartog, McGraw-Hill, New York, 1952.

$$S_r = \frac{1}{r}\frac{\partial \Phi}{\partial r} + \frac{1}{r^2}\frac{\partial^2 \Phi}{\partial \theta^2} \qquad \text{(d)}$$

$$S_s = -\frac{\partial}{\partial r}\left(\frac{\partial \Phi}{r\,\partial \theta}\right) \qquad \text{(e)}$$

The equation of compatibility for Airy's function is given by:

$$\left(\frac{\partial^2}{\partial r^2} + \frac{1}{r}\frac{\partial}{\partial r} + \frac{1}{r^2}\frac{\partial^2}{\partial \theta^2}\right)^2 \Phi = 0$$

which is a biharmonic function in r & θ . With the boundary conditions, there may exist a function Φ which satisfies these equations. However, we are only interested in the nature of the solution. If we can show that surface deformation (w) is purely a function of Pi or Po, then a linear relationship exists.

The boundary conditions are:

at $r = a$:

$S_r = P_i$

$S_s = 0$

at $r = \dfrac{u_0 + w}{\cos \theta}$

$S_r = 0$

$S_s = 0$

Taking these conditions, and substituting into either equations (a) and (b), or into the stress function relationships (c) (d). and (e), one can

Satisfy the boundary conditions:

at $r=a$

$$S_r = P_i = \frac{1}{a}\frac{\partial \Phi}{\partial r} + \frac{1}{a^2}\frac{\partial^2 \Phi}{\partial \theta^2} \tag{f}$$

$$S_s = 0 = -\frac{1}{a}\left[\frac{\partial^2 \Phi}{\partial r \partial \theta}\right] \tag{g}$$

at $r = \dfrac{u_o + w}{\cos\theta}$

$$S_r = 0 = \frac{\cos\theta}{u_o+w}\frac{\partial \Phi}{\partial r} + \frac{\cos^2\theta}{(u_o+w)^2}\frac{\partial^2 \Phi}{\partial \theta^2} \tag{h}$$

$$S_s = 0 = -\frac{\partial}{\partial r}\left[\frac{\cos\theta}{u_o+w}\left(\frac{\partial \Phi}{\partial \theta}\right)\right] \tag{i}$$

Now we define a new function ϕ such that $\phi = \frac{\Phi}{P_i}$

From this we obtain new boundary conditions all of which are a function of ϕ. (Note P_i is independent of r & θ)

From this we could obtain a solution for w

Such that $w \sim P_i$, since the boundary conditions are only a function of ϕ, which is in turn linearly related to Φ by P_i

Thus:

$$w = P_i \left(\text{Geometrical Factors}\right) \tag{j}$$

Where the Geometrical Factors are defined in terms of the Φ function. We see from this limited analysis that (w) is linear with Pi, and Pi is linear with Po. One can see this almost without further calculations in equation (f).

APPENDIX C

PART 1 - Some Results of a Study of the Infra-Red Pulse

In the body of this dissertation, it was noted that a program of evaluation was undertaken to study the relationship between the I.R. Pulse and intra-arterial blood pressure. Contained herein are a few examples of the data which ultimately led to the conclusion that I.R. Pulse data is not directly correlated to blood pressure.

Figures C.1, C.2, and C.3 are typical examples of a data analysis program which shows the error in I.R. Pulse as compared to the I. A. (intra-arterial) blood pressure. Figure C.1 shows a trend of error defined by the angle α. In this case the trend was definable by an approximate angle of ten degrees, whereas in Figure C.2, a trend cannot be defined and there is no correlation between the two signals. In Figure C.3, there again exists a trend, but the approximate definition angle is very different. Similar studies were undertaken on a number of cases, and the general results of this type of study indicated no definable relationship between the two outputs.

Another type of study which was most fruitful, is typically indicated in Figure C.4. In this type of study, the comparison of the two signals was done on a time basis. Here one can see the effects of clinical procedures on both the I.R. Pulse and on the I.A. blood pressure. The difference between good correlation and poor correlation lies in the clinical condition of the patient under study. In the cases where the patient retains a condition of homeostasis throughout the operation, the I.R. Pulse and the blood pressure correlate very well. In cases where the condition of the patient is not clinically good, one sees the kind of results depicted in Figure C.4. Various clinical factors are shown to effect the I.R. Pulse and the blood pressure in different ways. A vasoconstrictive drug like Neo-synephrin has a substantial effect on the blood pressure and little if any effect on the I.R. Pulse. On the other hand, a vasorelaxative drug such as Regitine has a similar effect on both parameters. Other major effects can also be seen. Case studies like this one led to an analysis of how other variables affected the condition of the patient. Further evidence is shown in the Figures C.5 through C.12. Each bit of data indicating some aspect of the nature of the I.R. Pulse, shows increase in the lack of correlation between the I.A. blood pressure and the I.R. Pulse.

Continued study is leading to some very important conclusions. Some of these condlusions are recorded in Part 2 of this Appendix.

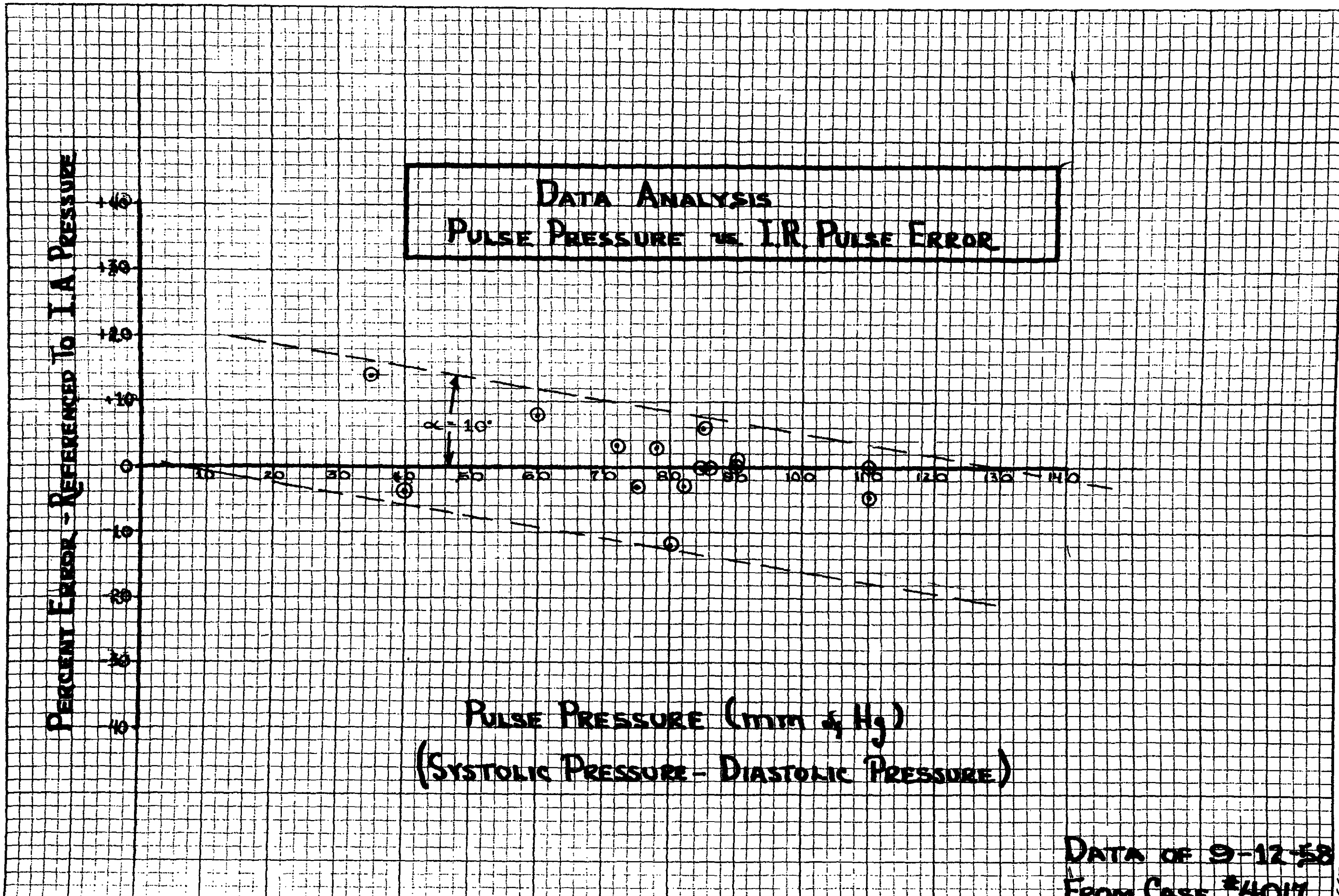

DATA ANALYSIS
PULSE PRESSURE vs. I.R. PULSE ERROR
PERCENT ERROR - REFERENCED TO I.A PRESSURE
+40
+30
+20
+10
0
-10
-20
-30
-40
10
20
30
40
50
60
70
80
90
100
110
120
130
140
α = 10°
PULSE PRESSURE (mm of Hg)
(SYSTOLIC PRESSURE - DIASTOLIC PRESSURE)
DATA OF 9-12-58
FROM CASE #4017
RW CORELL

FIGURE C.1

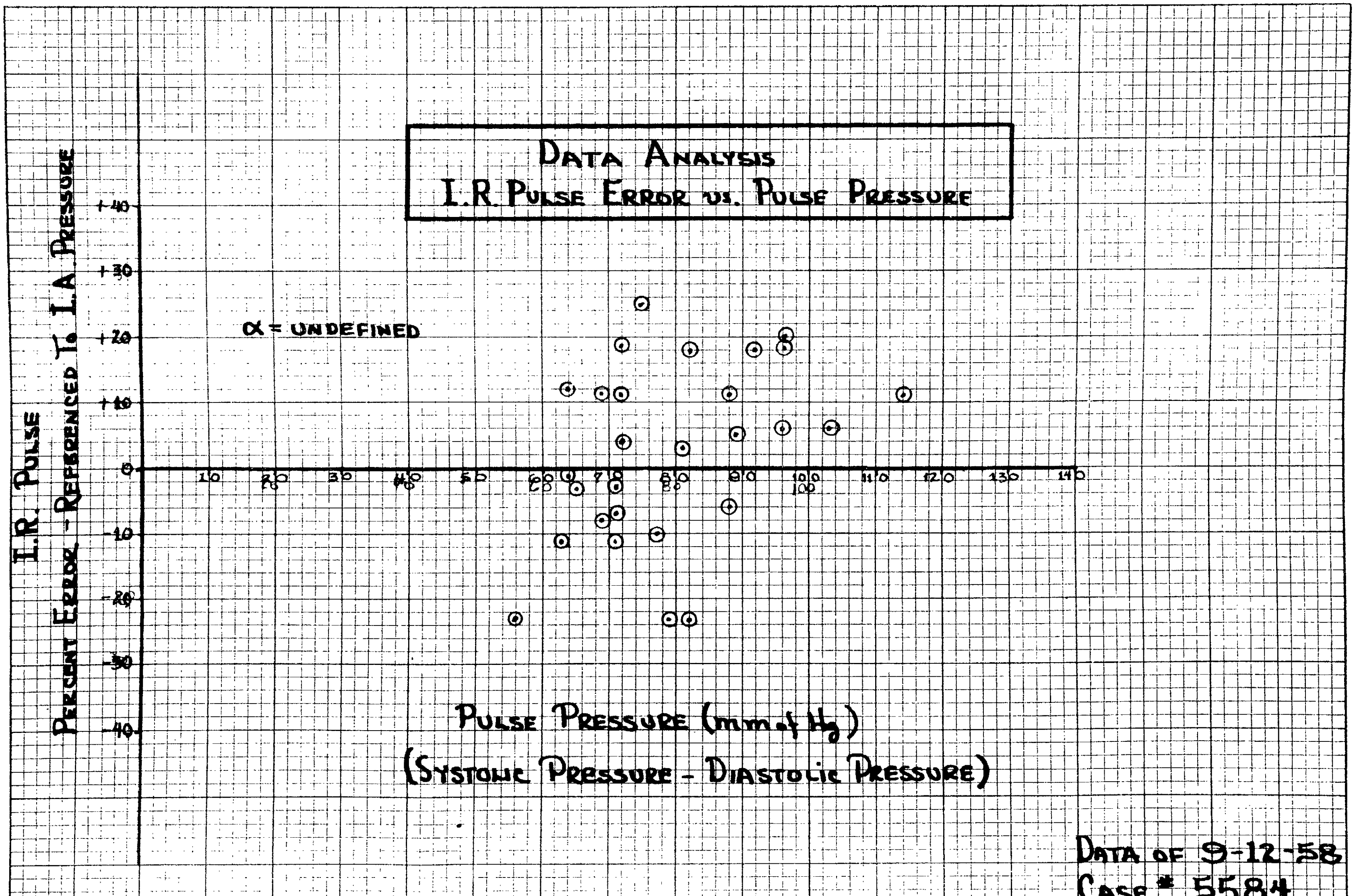

Data Analysis
I.R. Pulse Error vs. Pulse Pressure
I.R. Pulse
Percent Error - Referenced To I.A. Pressure
+40
+30
+20
+10
0
-10
-20
-30
-40
10
20
30
40
50
60
70
80
90
100
110
120
130
140
α = UNDEFINED
Pulse Pressure (mm. of Hg)
(Systolic Pressure - Diastolic Pressure)
Data of 9-12-58
Case # 5584
RUD CORELL

Figure C.2

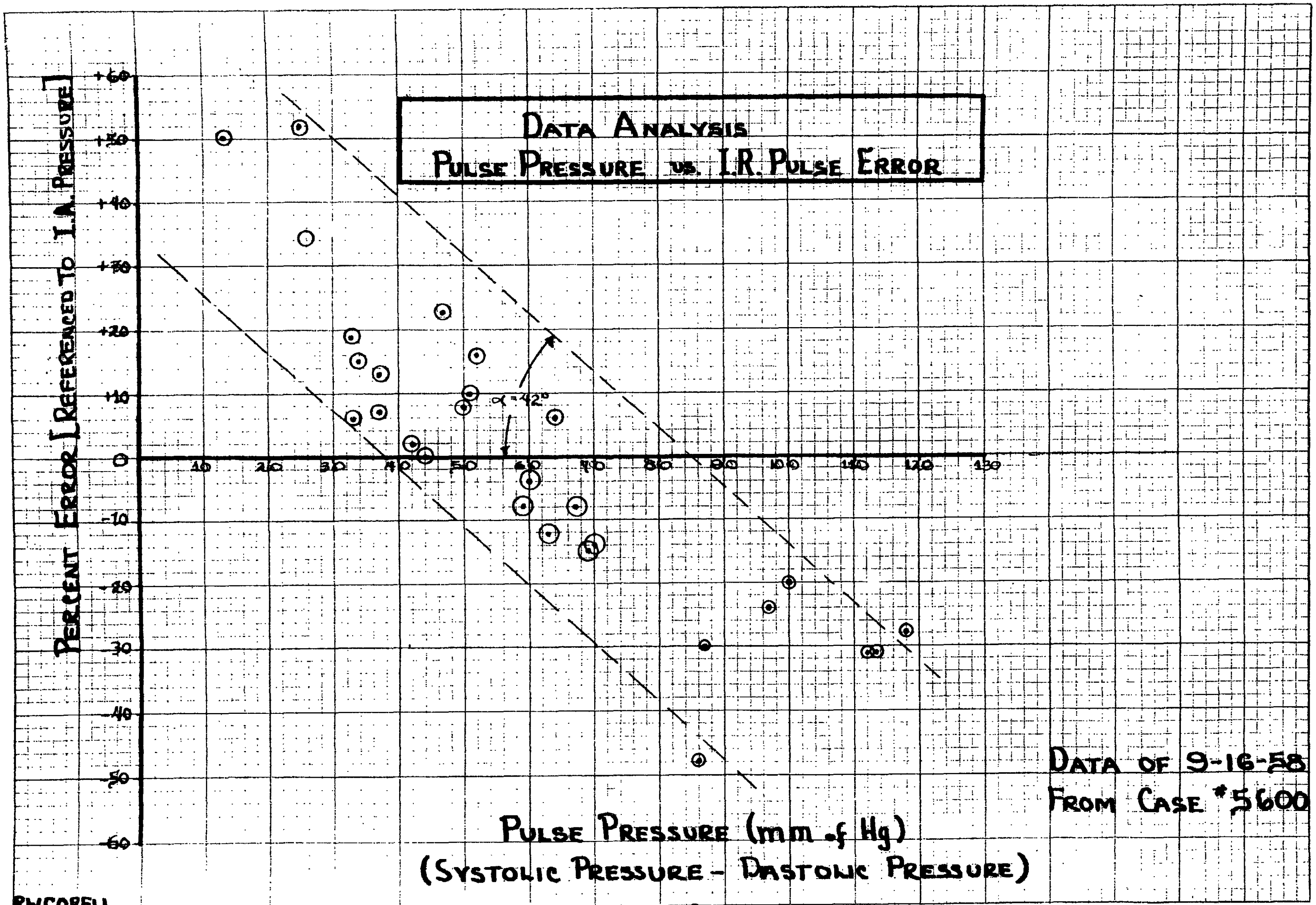
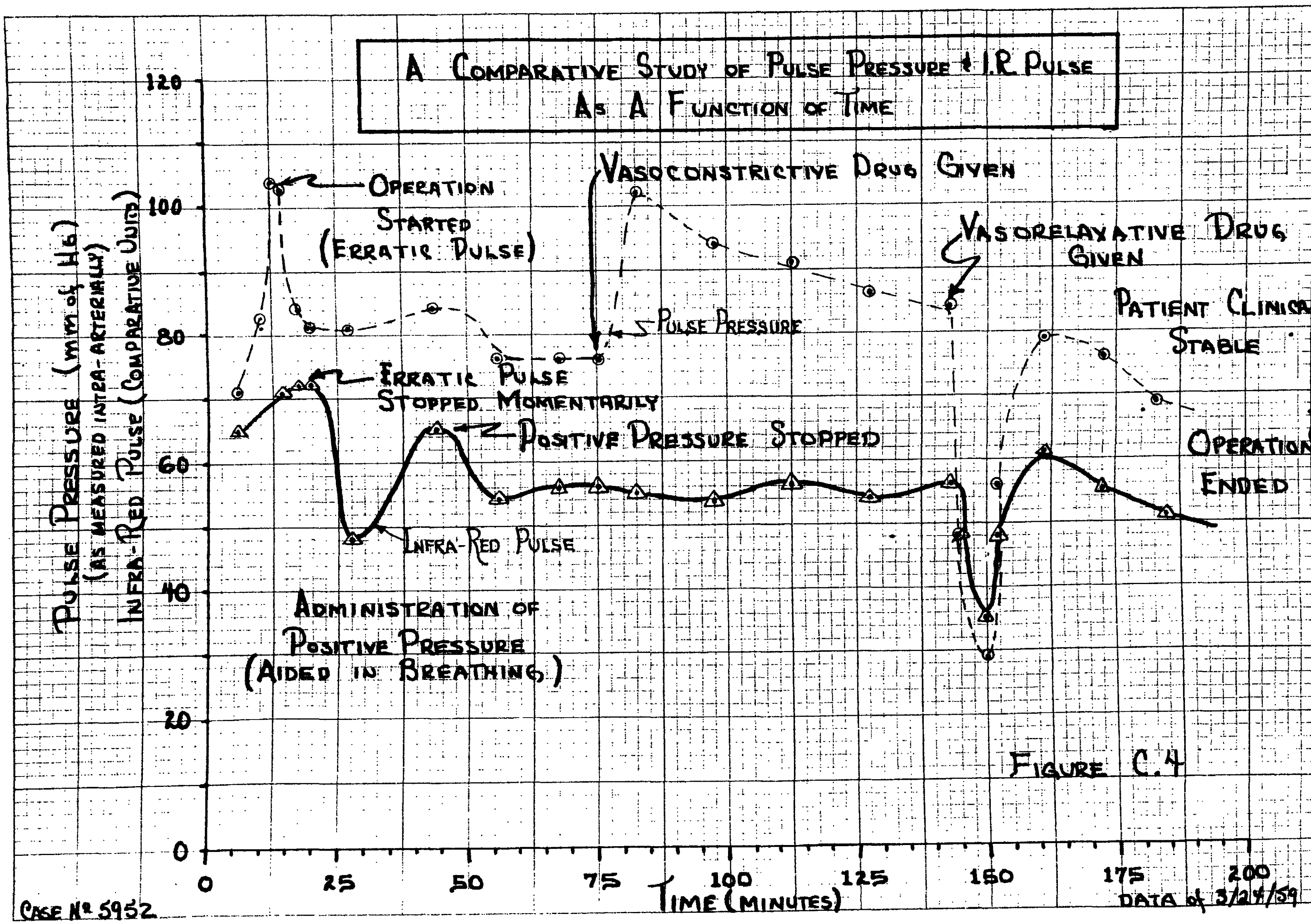

DATA ANALYSIS
PULSE PRESSURE vs. I.R. PULSE ERROR
PERCENT ERROR [REFERENCED TO I.A. PRESSURE]
+60
+50
+40
+30
+20
+10
0
-10
-20
-30
-40
-50
-60
10
20
30
40
50
60
70
80
90
100
110
120
130
α = 42°
PULSE PRESSURE (mm of Hg)
(SYSTOLIC PRESSURE - DIASTOLIC PRESSURE)
DATA OF 9-16-58
FROM CASE #5600
RWCORELL

FIGURE C.3


A Comparative Study of Pulse Pressure & I.R. Pulse
As A Function of Time
Pulse Pressure (mm of Hg)
(As measured intra-arterially)
Infra-Red Pulse (Comparative Units)
120
100
80
60
40
20
0
0
25
50
75
100
125
150
175
200
Time (Minutes)
Operation Started (Erratic Pulse)
Erratic Pulse Stopped Momentarily
Positive Pressure Stopped
Infra-Red Pulse
Administration of Positive Pressure (Aided in Breathing)
Vasoconstrictive Drug Given
Pulse Pressure
Vasorelaxative Drug Given
Patient Clinical Stable
Operation Ended
Figure C.4
Case No 5952
Data of 3/24/59

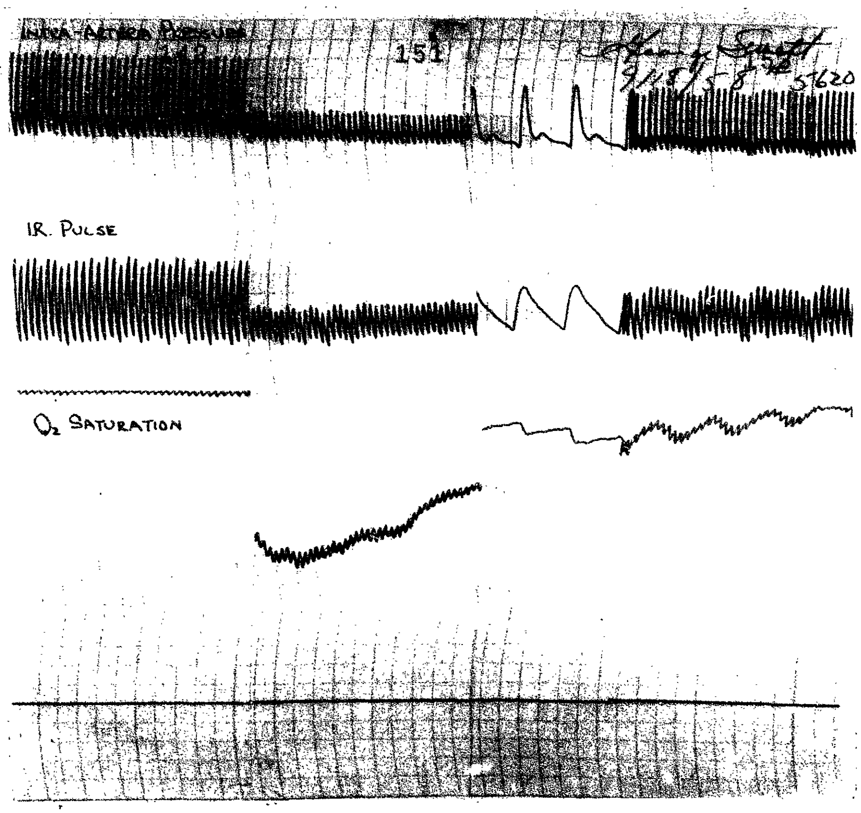


Figure C.5 This photograph shows the type of clinical condition which led to the assumption that the I.R. Pulse might be an indirect method for recording blood pressure. For High pressure, low pressure, and modulatory pressure, there exists, in this case, a similarity in form.

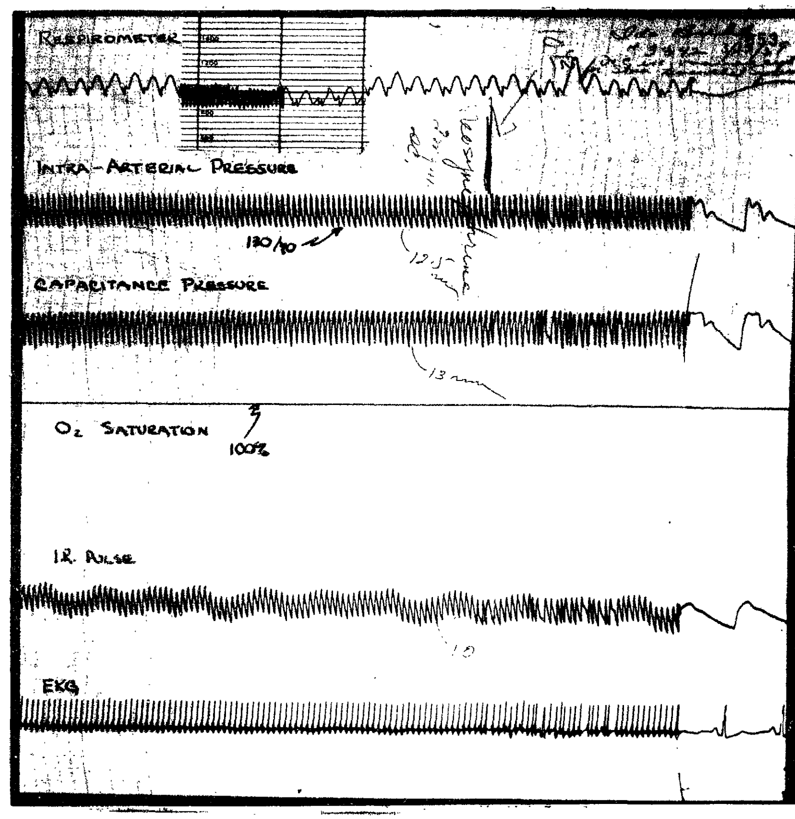
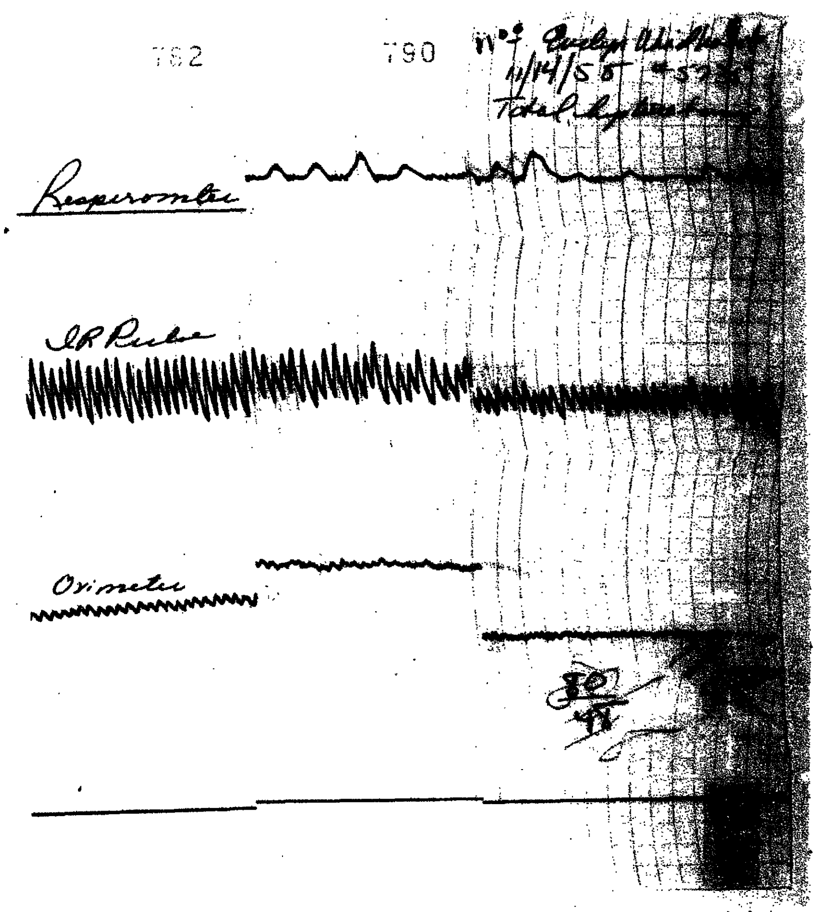
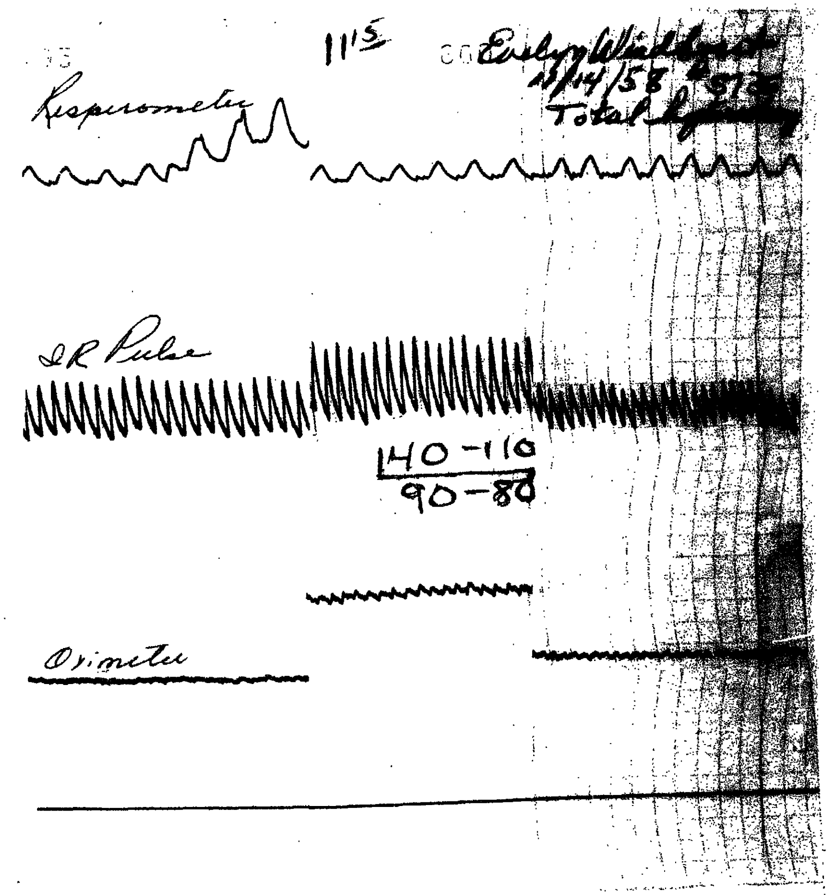


Figure C.6 This photograph shows the conditions under which the I.R. pulse does not correlate with the intra-arterial blood pressure. The I.R. pulse is reflecting an alteration in the clinical status of the patient even though the blood pressure remains steady. Also shown is the stability of the capacitance blood pressure technique. Note the similarity in the characteristic wave form of the two tracings.



<u>Figure C.7</u> This tracing shows some of the effects of clinical procedures on the characteristics of the I.R. pulse. 782 shows the pre-operative conditions, 790 shows the effect of insertions of the intratracheal tube, and the last chart shows the prolonged effects of hypotension.



<u>Figure C.8</u> This tracing shows the effects of aided breathing upon the I.R. pulse even though the blood pressure remained fairly stable at 140/90. 795 shows the I.R. pulse before positive pressure, 800 shows the response during positive pressure, and the last chart shows the condition upon cessation of aided breathing.

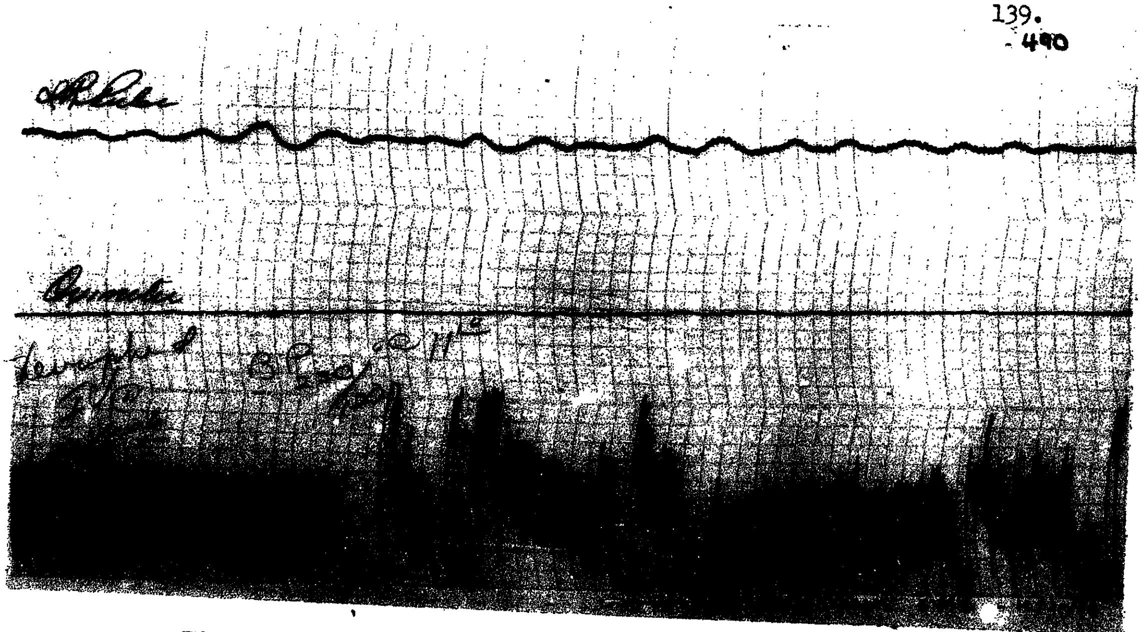


Figure C.9 This tracing was taken just prior to cardiac failure. Note the small amplitude of the I.R. pulse even though the blood pressure was very high, namely, 230/120.

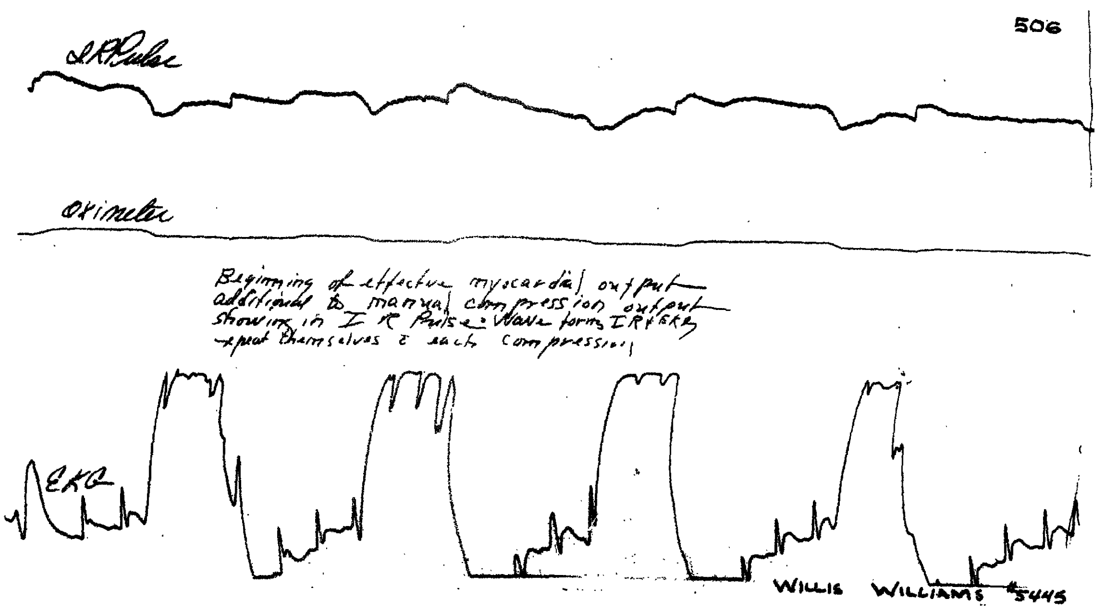


Figure C.10 This tracing was taken during cardiac failure. It shows the irregularity of the I.R. pulse under these conditions of extreme duress. The major activity of the I.R. pulse is during manual compression and not during the myocardial output. During this time no accurate knowledge of the blood pressure was obtainable.

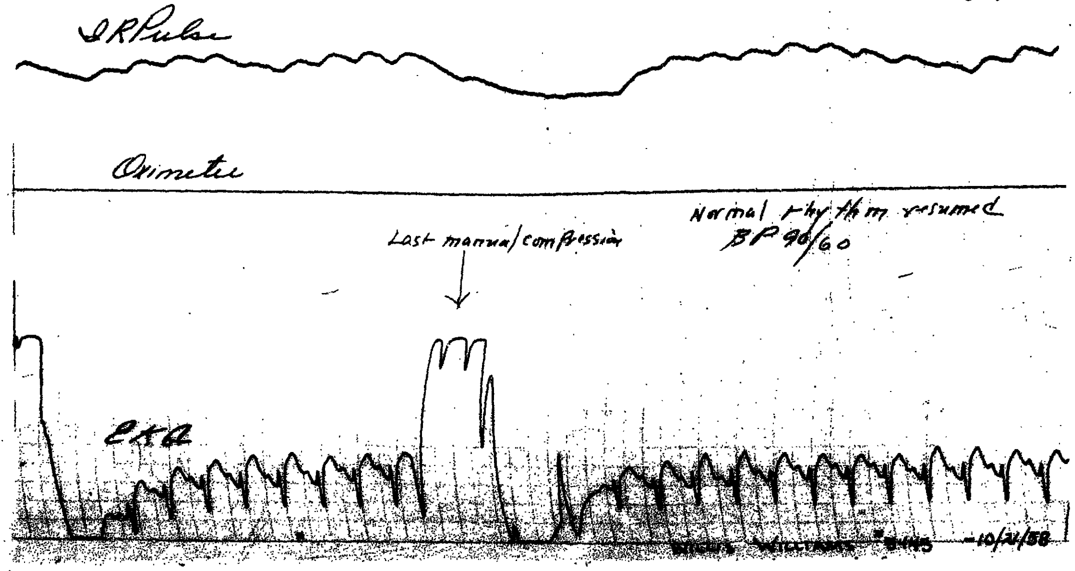


Figure C.11 This tracing was taken during the onset of normal rhythm The blood pressure was 90/60 and the I.R. pulse was still somewhat irregular in nature. The effect of the last manual compression is every obvious as well as the ineffectual muscular activity of the heart as reflected by the electrocardiograph(EKG).

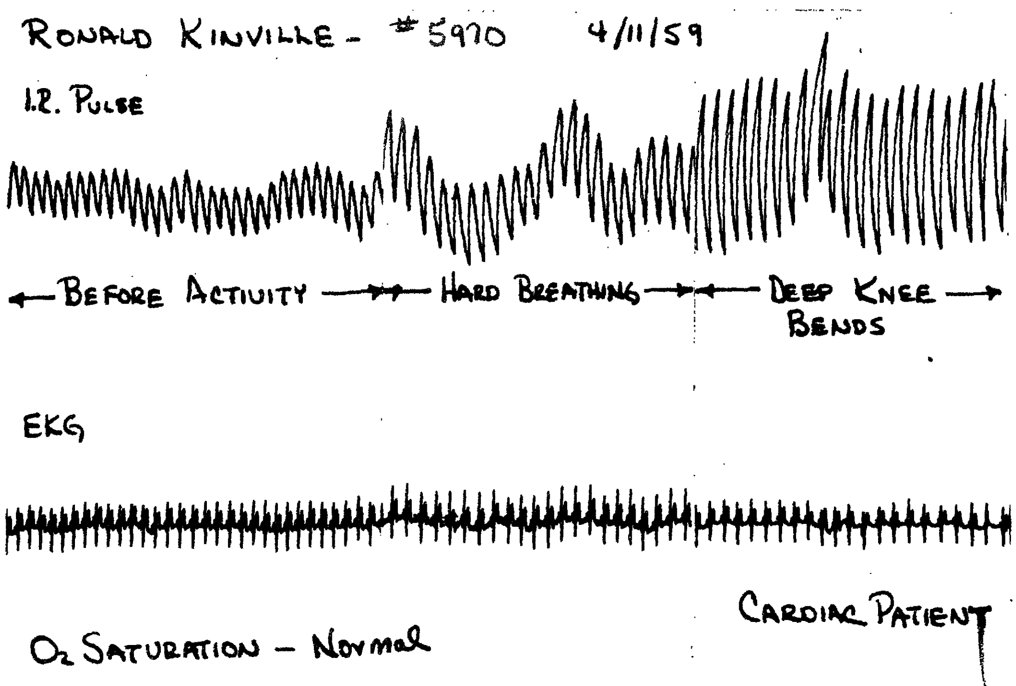


Figure C.12 This tracing shows the effect of minor activity on the I.R. pulse of a cariac patient.

APPENDIX C

PART 2 - Preliminary Study into the Nature of the Infra-Red Pulse

The infra-red pulse appears to have much broader implications than the sensing of volumetric change, because of its responsiveness to various "known" conditions in which the clinical situation can be assayed with fair accuracy, even though the circulatory function, "peripheral diffusion", cardiac output, etc., is variable. It has been suspected of being the stroke volume output, which it may be, but about which we are not yet quite sure. Whether or not it is stroke volume output, one can be fairly confident that it either is or comes close to being an equivalent representation of the stroke volume output arriving at the vessels under the photoelectric cell, and possibly as modified by vasomotor control. Since the IR instrumentation may very well be sensitive, not only to physiologic changes, but also to chemical changes in the body, it is important to determine the extent to which it can indicate effectiveness of circulatory, respiratory, or other functions. Correlation with many other items of information, such as levels of carbon dioxide in the end-alveolar air, the blood, the pH, the blood pressure, the tidal volume, and the minute volume exchange, with the clinical course of the patient are possible and revealing. We have developed confidence in the prognostic indications of the I.R. Pulse wave, and have not as yet seen any other indicator of circulatory function, or reflector of satisfactory ventilatory exchange, that e ualled this single test in sensitivity or accuracy.

Theories are proposed by the investigators in explenation but it is too early yet to know whether full validation has been achieved.

In analyzing the I.R. Pulse, one must gain some understanding as to the nature of the pulse wave. Obviously, a physical piece of electronic equipment is recording a physical parameter arising out of physiologic circumstances. That which we are measuring and calling the I.R. Pulse seems to be the optical density in the ear. The nature of this optical density is the important question and the following comments are directed toward a possible answer. The ear presents a physical makeup of skin, subcutaneous tissue, muscle fibers, and vessels. Within the small blood vessel walls is the circulating blood. The factors which might effect the optical density are variable. They may include the skin, but probably the skin is not changed in terms of optical density. The color of the skin might change, but its optical density would remain the same. The tissues that lie between the skin and the arterial walls themselves also are probably constant in terms of optical density. Additional factors, such as heat or edema, which might change optical density do not seem to play a part, but one must remain open minded to further analysis of this possible variable. It seems that optical density mainly will be effected by changes in the arterial wall and its muscle fibers, and in the blood itself.

The following factors may change optical density. A change in volume is a most important change of significance. The tremendous influence of vasomotor control over peripheral volume, flow, and distention is well recognized. Less well known or appreciated is the integral relationship between flow and distensibility in production of peripheral diffusion. A change in volume would change the passageway volume for blood, and consequently, the volume of the blood under the cell at a particular instance would be greater or less depending upon the pattern. One must then ask which affects optical density the most, pressure or volume flow? The rate of volume flow is significant but this factor can be seen in the pulse wave itself, so that the question reduces to that of the relative importance of pressure and of volume.

The IR cell output seems to be proportional to the stroke output of the heart. This situation can perhaps be better understood by analogy. Vasomotor control may clamp down the arterioles under the photoelectric cell at the ear and prevent flow, even when there is a recognizable or satisfactory level of blood pressure centrally. If there be no vasomotor effect, then the I.R. Pulse at the ear is probably equivalent in its volumetric distention to the pressure change indicated proximally by an intra-arterial cannula. In a similar manner, one might say that a plethysmogram on a finger can be equivalent in its representation to the blood pressure, if there be no vasospasm. When vasomotor control has expressed itself in constriction, the volumetric distention of the finger may become nearly 0 even though a central blood pressure persist at near normal levels.

There are other factors than pressure or volume which may effect the I.R. Pulse characteristics. Moving away from consideration of the purely physical facts, - that the I.R. Pulse is a measure of the optical density in the ear and that any deviation is clearly a reflection of the optical density change, one has reasons for believing that the optical density is affected by volume mainly, but not exclusively. The number of particles carried by the blood will affect the optical density, just as water will have a different optical density from blood. This can be observed in the different height in pulse wave in the anemic person as distinguished from the plethoric person. It is now recognized that density of the red cells and white cells, as reflected by anemia or plethor , affects not only the physiologic and medical condition of the patient involved, but also the optical density characteristics on which measuring efficiency of the vascular system by this means are based.

One is now ready to define what is meant by volumetric deficiency. We define volumetric efficiency as the ratio of the total volume that the heart is capable of extruding into the vascular system to what the heart is actually doing at any given instant of time. If the heart is normally operating, volumetric efficiency is 100%. If the stroke decreases, the ratio decreases, and the I.R. Pulse likewise decreases in its amplitude and one can recognize the qualitative relationship between the I.R. Pulse and volumetric efficiency. It is apparent from the physical data that has been obtained from seriously ill patients in the operating room, that the I.R. Pulse is definitely correlated to the condition of the patient, and that it reflects very sensitively the status of the patient at any particular instant of time. The I.R. Pulse has been recognized by us as a valuable tool for measuring the changes that occur in the preoperative, operative, and postoperative situations to which patients are subjected.

APPENDIX D

A Collection of Pressure-Volume Relationships for human arterial sections. Some of the data herein was obtained through Journals and other publications, while some limited data was collected by the author.

PRESSURE-VOLUME RELATIONSHIPS

OF HUMAN ARTERIES*

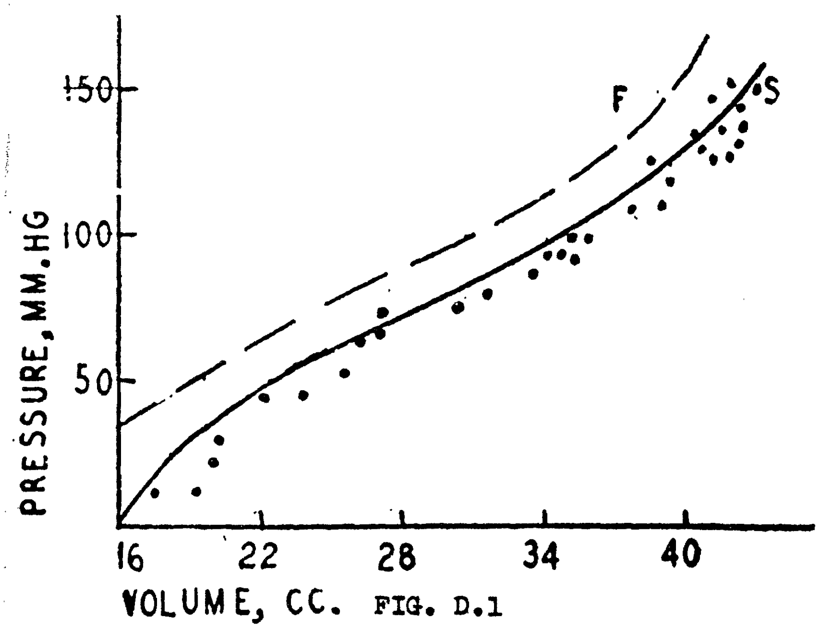


FIG. D.1

Obtained from an article by J.W. Remington, W.F. Hamilton, and P.Dow. American Journal of Physiology. 144: 540, 1945

PRESSURE-VOLUME RELATIONSHIPS OF ANIMAL ARTERIES*

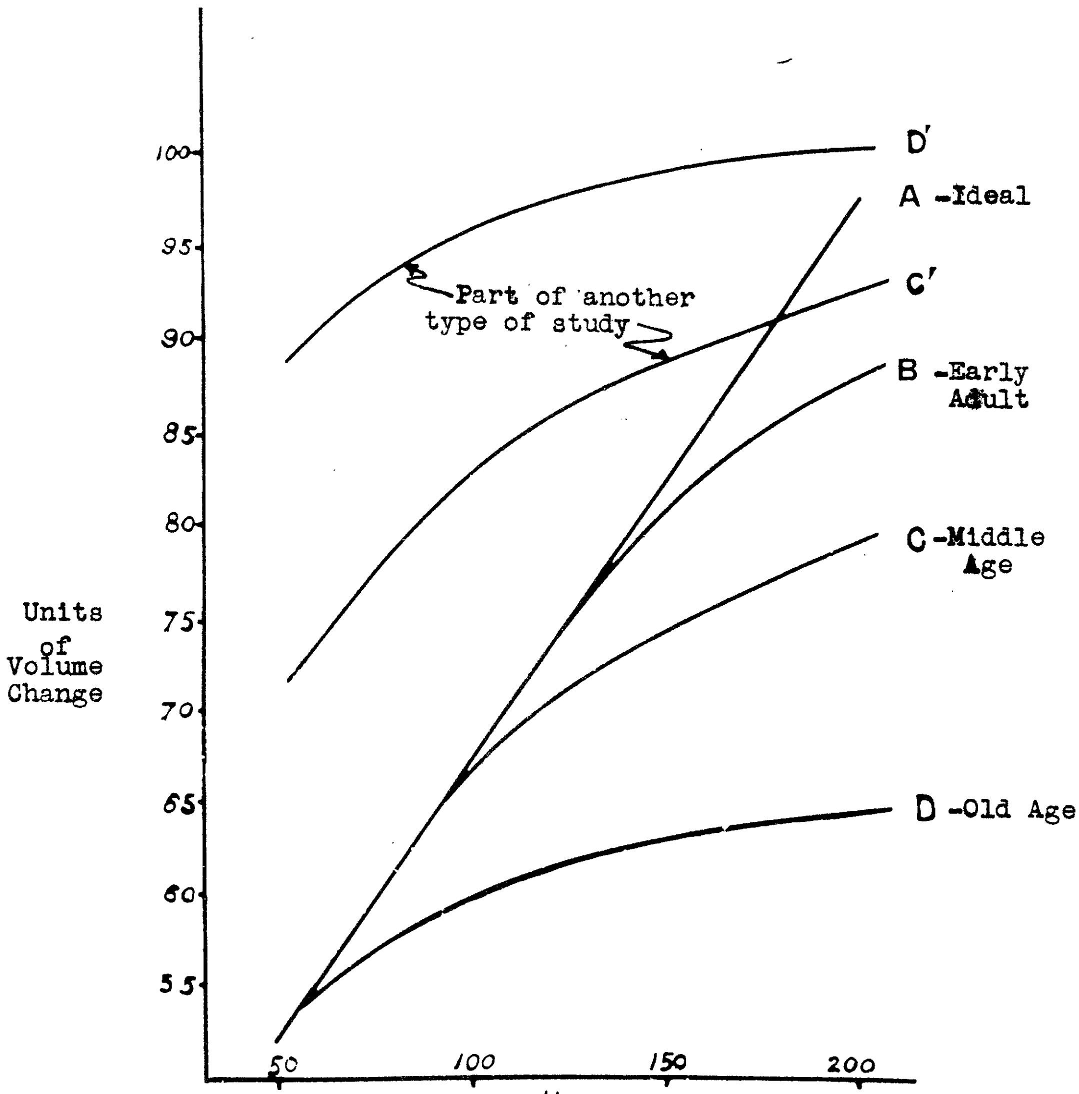


FIG. D.2

Pressure-Volume relationships from the entire intact aorta of dogs that have recently died.

*Obtained from article by C.J. Wiggers in the American Journal of Physiology. 123:644,1938

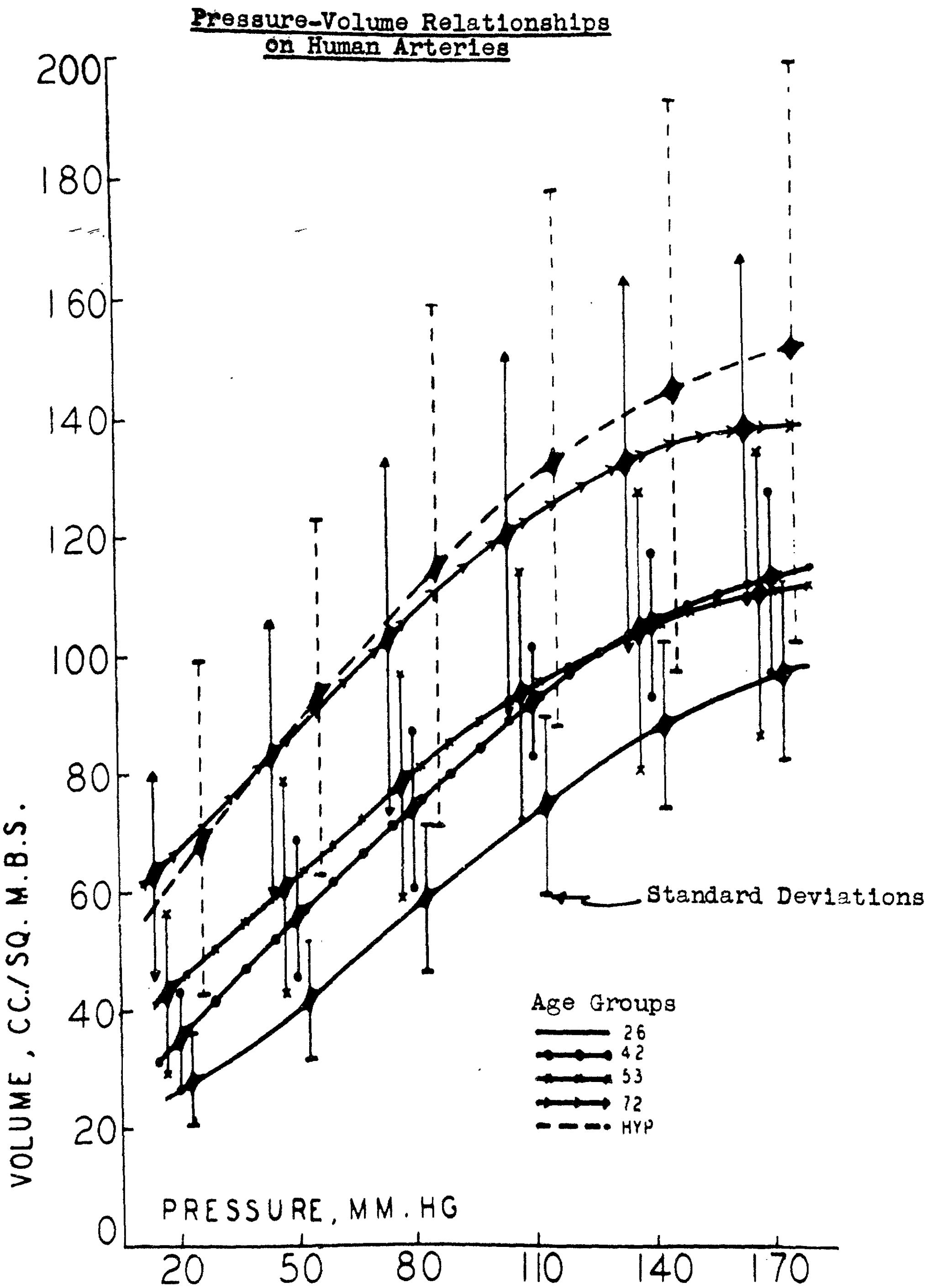
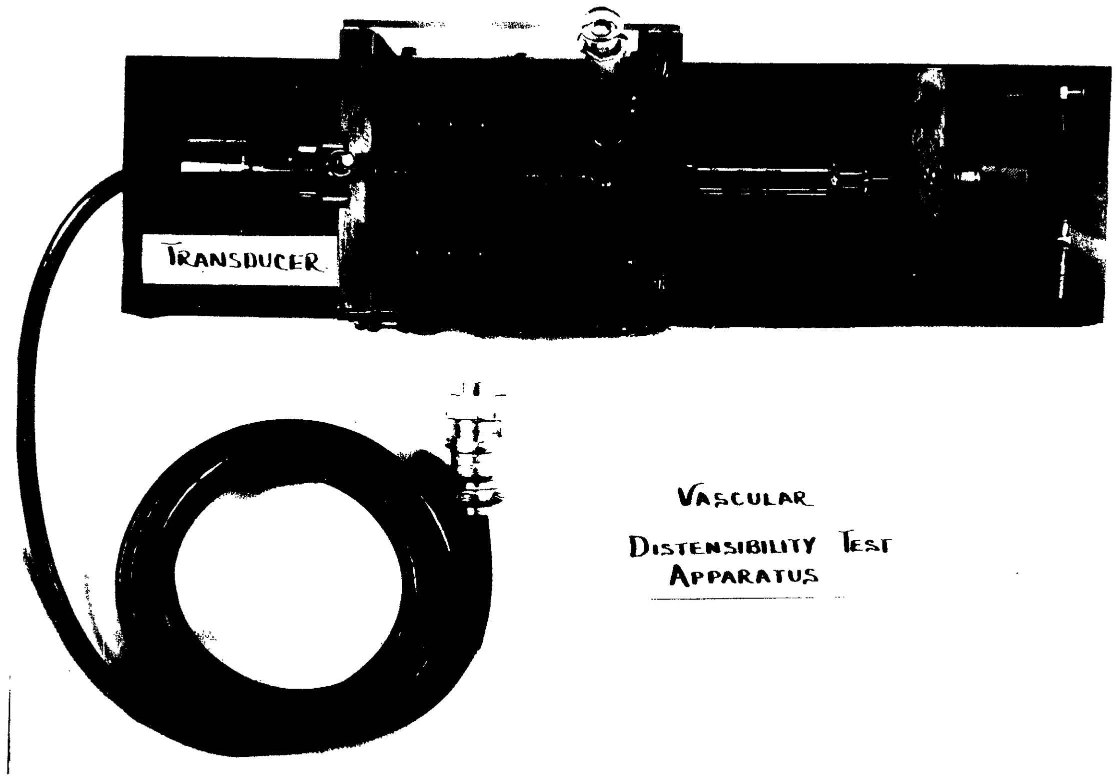


FIGURE D.3

Obtained from American Journal of Physiology in Article by J.W. Remington, C.R. Noback, W.F. Hamilton, and J.J. Gold, Vol 153, Page 300, 1948



VASCULAR
DISTENSIBILITY TEST
APPARATUS

P.5

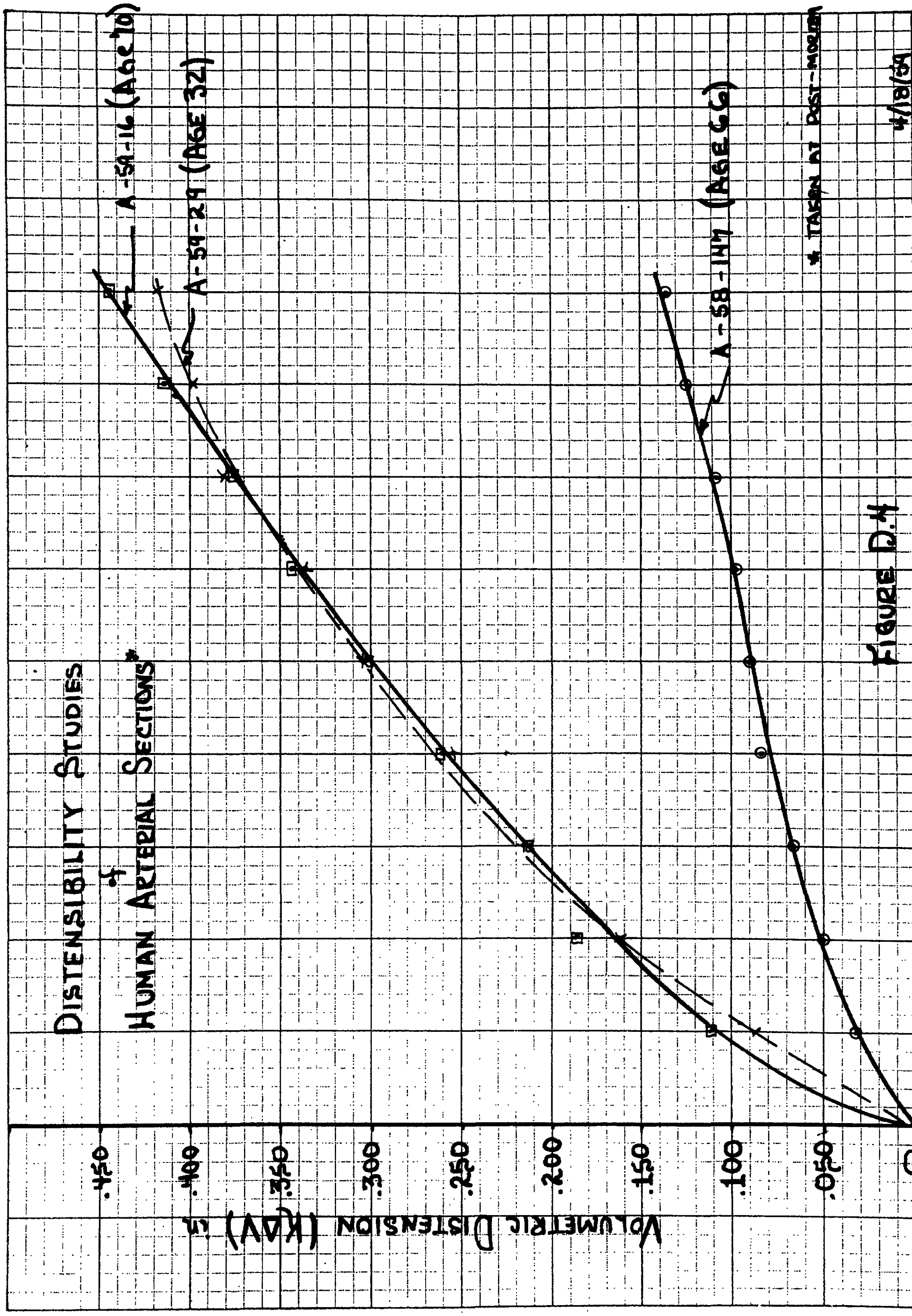

DISTENSIBILITY STUDIES
of
HUMAN ARTERIAL SECTIONS*
A-59-16 (AGE 10)
A-59-29 (AGE 32)
A-58-147 (AGE 66)
* TAKEN AT POST-MORTEM
4/18/59
PRESSURE (mm of Hg)
0
25
50
75
100
125
150
175
200
225
250
VOLUMETRIC DISTENSION (KΔV) in
0
.050
.100
.150
.200
.250
.300
.350
.400
.450

FIGURE D.4

HUGGINS HOSPITAL NEW HAMPSHIRE
DEPARTMENT OF Surgery

~~Course~~ 5cc Syr. Experiment No. B Date 4/18/59

~~Section No.~~ Observers RW Corell

~~Recorder~~ Mass. Memorial Hospital Case No [A-58-147; A-59-29; A-59-16]

Apparatus Distensibility Studies [Average Data of Five Runs / Artery]

Pressure (mm of Hg)	Diaphragm Correction	Average of Five Data Runs A-58-147	A-59-29	A-59-16	Corrected Data A-58-147	A-59-29	A-59-16
0	0	.0	0	0	0		
25	-.0007	.0320	.0890	.1120	.0310	.0880	.1110
50	-.0016	.0520	.1650	.1913	.0500	.1630	.1897
75	-.0022	.0680	.2200	.2180	.0660	.2180	.2158
100	-.0030	.0850	.2580	.2632	.0820	.2550	.2602
125	-.0038	.0930	.3110	.3071	.0890	.3070	.3033
150	-.0044	.1000	.3410	.3503	.0960	.3370	.3459
175	-.0048	.1120	.3870	.3823	.1070	.3820	.3775
200	-.0052	.1300	.4020	.4220	.1250	.3970	.4168
225	-.0056	.1430	.4250	.4531	.1370	.4190	.4475

~~Instr. No.~~

Speciman Sizes: (Arterial Sections taken @ Post-Mortem, stored at -78°F)

A-58-147 - 3/8" O.D - 1.5" Length

A-59-29 - 3/8" O.D - 1.25" Length

A-59-16 - 3/8" O.D. - 1.25" Length.

HUGGINS HOSPITAL NEW HAMPSHIRE
DEPARTMENT OF SURGERY

~~Nurse~~ Experiment No. A Date 4/18/59

~~Section No.~~ Observers RW CORELL

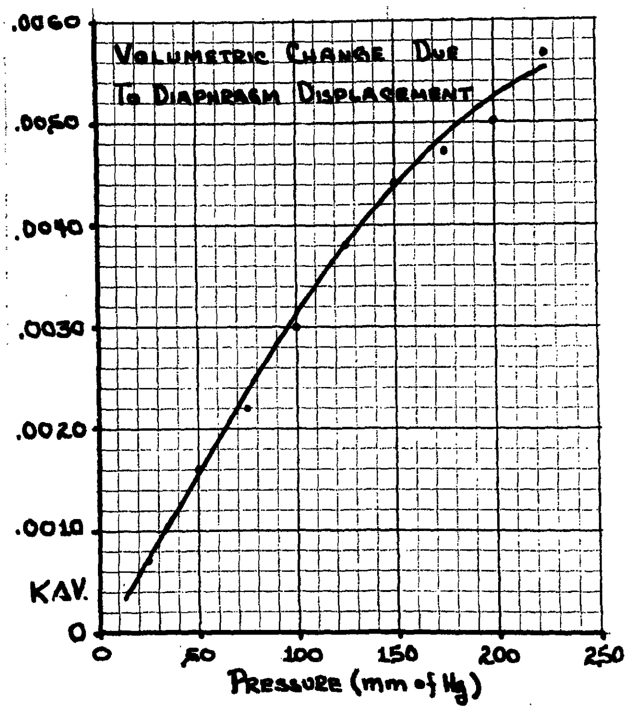
Recorder

Apparatus DISTENSIBILITY STUDIES - RIGID SYSTEM - VOLUMETRIC DISPLACEMENT of DIAPHRAGM

SUMMARY of DATA
(Average Data of 5 Runs)

Inst. No.

PRESSURE MM of HG	KΔV (in)
0	0
25	.0007
50	.0016
75	.0022
100	.0030
125	.0038
150	.0044
175	.0047
200	.0050
225	.0057



APPENDIX E

BIBLIOGRAPHY

1. Wiggers, C.J. Physiology in Health and Disease. Philadelphia: Lea and Febiger, 1949. Page 694.

2. Busygin, V.E., Nefedov, I.V. E. Prolonged and Uninterrupted Registration of Arterial Pressure Using Bloodless Method. (Russian) Biulleten Eksperimental Noi Biologii I Meditsiny. 39:74-76, Feb. 1955.

3. Caster, W.O., Lentz, G.A., Poncelet, J., and Armstrong, W.D. Indirect Determination of Systolic and Diastolic Pressures in the Rat. J. Applied Physiol. 8:664-666, 1956.

4. Fry, D.L., Noble, F.W., Mallos, J.M. An Evaluation of Modern Pressure Recording Systems. Circulation Research. 5:40, 1957.

5. Green, H.D. Circulation, Medical Physics. Edited by Otto Glasser. Chicago: Yearbook Publishers, 1944 Vol. I, page 208: Vol. II, page 208.

6. Hamacher, J., Vater, W. Sound Guided Cuff Pressure Recording in Closed Blood Pressure Measurements as a New Recording Principal in ECB - Direct Recorder. (German), Arch. Exp. Path, Berl., 19:135-137, 1955.

7. Hansen, A.T. Pressure Measurement in the Human Organism. Aeta Physiol. Scandinav. (Supp. 68) 19:1-230, 1949.

8. Henschel, A., DeLaVeba, F., and Taylor, H.L. Simultaneous Direct and Indirect Blood Pressure Measurements in Men at Rest and Work. J. Applied Physiol. 6:506-508, 1954.

9. Karpman, J.L. A Method for Indirect Pulse Recording, and Its Use in the Diagnosis of Aortic Stenosis. American Heart Journal, 56:

799-803, 1958.

10. Piper, H.P. Registration of Phasic Changes of Blood Supply By Means of a Catheter Type Flowmeter. Review of Scientific Instr. 29:965, 1958.

11. Roy, O.F. and Charbonneau, J.R. Transistor Unit Monitors Blood Pressure: Electronics 31:(Aug) 82, 1958.

12. Wood, E.H. Special Instrumentation Problems Encountered in Physiological Research Concerning The Heart and Circulation in Man. Science. 112:707 1950.

13. Zuidema, G.D. Edelberg, R., and Salzman, E.W. A Device for the Indirect Recording of Blood Pressure, Wright Air Development Center - WACD, Tech. Note No. 55-427, 1955.

14. Circulation Research. All available copies from 1953 - 1958 were reviewed and little was found that had a significant relationship to this project.

15. Current List of Medical Literature. All copies from 1953 - 1958 were reviewed and significant references were obtained.

16. Quarterly Index. All copies from 1954-1955 were reviewed and little was found that had a significant relationship to this project.

17. Chemical Abstracts. Various copies were reviewed and little was found that had a significant relationship to this project.

18. Review of Scientific Instruments. All available copies were reviewed (1940-1958) and little was found that had a significant relationship to this project.

19. Journal of Biophysics. (Russian), Reviewed all available copies and little was found that had significant relationship to this project.

20. Glasser, O. Medical Physics. Has excellent references on blood pressure measurement. 1:231, 2:221, 1950.

21. Many other references were reviewed and they are not listed here since they served only as background material and thus were not specifically related to the basic project.

22. Arynchin, N.I., Zonkevich, I.S. Half Century of Application and Further Development of Korotkow's (Russian spelling - Krotkov's) Sound Method for the Determination of Human Blood Pressure. Fiziol, Zhur. SSSR. 43 (1): 92-95, 1957. Referat Zhur. Biol. 1957, No. 70741. (trans. by U.S. Joint Publ. Res. Serv.)

23. Bazett, H.C., and Laplace, L. B. Studies on the Indirect Measurement of Blood Pressure. Am. Jr. Physiol. 103:48, 1933.

24. Bazett, H.C., and Laplace, L.B. The Pressure Changes Induced in the Vascular System as the Result of Compression of a Limb, and Effects on the Indirect Measurement of Lateral Pressures. Am. Jr. of Physiol. 112:182, 1935.

25. Berry, M.R., The Mechanism and Prevention of Impairment of Auscultatory Sound During Determination of Blood Pressure of Standing Patients. Proc. Staff Meet., Mayo Clinic. 15:699, 1940.

26. Hamilton, W.F., Woodbury, R. A. and Harper, H.T. Physiologic Relations Between Intrathoracic Intraspinal, and Arterial Pressures. J. A.M.A. 107:853, 1936.

27. Kotte, J.H., Iglaver, A., and McGuire, J. Measurement of Arterial Blood Pressure in Arm and Leg. Am. Heart Journal, 28:489, 1944.

28. Parnell, J., Beckman, E.L. and Peirce. Conventional Methods Compared with an Absolute Method for Pressure Transducer Response Measurement. U.S. Naval Air Development Center - report No. NADC -

MA - 5206, Johnsville, Pa. 1953.

29. Ragan, C., Bordley, J. Measurement of Blood Pressure. Bull. Johns Hopkins Hosp. 69:526, 1941.

30. Steele, J.M. Measurement of Arterial Pressure in Man. J. Mt. Sinai Hosp. 8:1049, 1941-1942.

31. Von Bonsdorff, B., Zur Methodik Der Blutkruck Messung. Acta Med. Scandinau Suppl. LI: 1932.

32. Van Bergen, F.H., Comparison of Indirect and Direct Methods of Measurement of Arterial Blood Pressure.10:481-490, 1954.

33. Wigger, C.J., Chairman to Revise Standardization of Blood Pressure Readings. Recommendations for Human Blood Pressure Determination by sphygmomanometer, Official Publication of Am. Heart Association as Published in Circulation . 4: 1951. J. A.M.A. 1951.

34. Hales, S. Statistical Essays. From Fulton's Selected Readings in the History of Physiology. Springfield: Charles C. Thomas 1930.

35. Poiseuille, J. Researche Sur La Force Du Coeur Aortique. These: Paris, 1828.

36. Gilson, W.E., Personal Letter of Sept. 16, 1958. Regarding an Instrument Currently Manufactured, "Its only fault being that so much mechanism has been added to eliminate the human factor that the machine itself is slightly neurotic."

37. Clarridge, R.E., Personal Letter of Sept. 12, 1958. (Taylor Instrument Co.)

38. Gilford, S.R. and Broida, H.P. Physiological Monitoring Equipment for Anesthesia and Other Uses. National Bureau of Standards, No. 3301, Washington, D.C. 1954.

39. Gilson, W.E., Automatic Blood Pressure Recorder, Electronics, 1942.

40. Lanza, K. A Recording Sphygmomanometer: Annals of Internal Med., 18:13, 1943.

41. Ornberg, A.C. Apparatus for Recording Systolic Blood Pressure. R.S.I. 7: 1936.

42. Stokvis, B. Uninterrupted Automatic, Bloodless Registration of Blood Pressure in Man. Nederl, Tydscher, V. Genosk 1937.

43. Doupe, I., Newman, H.W., and Wilkins, R.W., Method for Continuous Recording of Systolic Arterial Pressure in Man. J. Physiol. 95: 239-243, 1939.

44. Royce, P., Personal letter of Sept. 16, 1958: (Cambridge Instrument Co.), "Although we did manufacture an automatic blood pressure recorder-------(we) are no longer producing the instrument."

45. Rossiter, C.A., Personal Letter of Sept. 26, 1958. (Coleman Instruments, Inc.)

46. Martens, M. Personal Letter of Sept. 10, 1958. (Gilford Instrument Laboratories).

47. Wright Air Development Center, WADC Report Regarding Blood Pressure Monitoring Sept., 1956. (Technical Note No. 55-427 and Astia Document No Ad 110576)

48. Wood, E.H., Knutson, J.R.B., and Taylor, B.E. Measurement of blood content and arterial blood pressure in human ear. Proc. Staff Meeting, Mayo Clin. 25:398, 1950.

49. Schotz, S., Bloom, S.S. Helmsworth, F.W., Dodge, H.C., and Birkmire, E.L. The Ear Oximeter as a Circulatory Monitor. Anesthesiology. 19: 386, 1958.

50. Wood, E.H.. Oximetry. Medical Physics - Edited by Otto Glasser. Vol. II, 664-680, Chicago: Yearbook Publishers, 1950, Vol. II,

664-680.

51. Zijlstra, W.G. A Manual of Reflection Oximetry, Assen, Netherlands. Van Gorcum, 1958.

52. Perkins, J.F. The Photoelectric Oximeter, Modern Medicine, Aug: 117-124, 1956.

53. Parnell, J., Beckman, E.L., and Peterson, L.H. The Evaluation of Pressure Transducer Systems. Report No. NADC - MA - 5206, U.S. Naval Air Development Center, U.S. Navy, Johnsville, Pa., 1953.

54. King, A.L. Pressure-Volume Relations For Cylindrical Tubes with Elastomeric Walls. Journal Applied Physiology. 17:501, 1946.

55. King, A.L. Velocities of Pulse Waves in Large Arteries. Journal Applied Physiology. 18: 595, 1947.

56. Hallock, P. and Benson, I.C. Studies on Elastic Properties of Human Isolated Aorta. J. Clinical Invest. 16: 595, 1937.

57. Hodgman, C.D. Handbook of Chemistry and Physics. Cleveland: Chemical Rubber Publishing Co. 40th Edition, 2499.

58. Gregg, E.C. Ultrasonics, Medical Physics, Edited by Otto Glasser, Chicago: Yearbook Publishers II: 1132, 1950.

59. Klip, W. Difficulties in the Measurement of Pulse-Wave Velocity. Am. Heart Journal 56: 806, 1958.

60. Hurthle, K. Beitrage Zur Hamodynamik. I. Zur Technik der Untersuchung des Blutpruckes. Arch. f.d. gcs. Physiol. 43: 399 1888.

61. Cowherd, E.W. Personal Letter from Statham Instrument, Inc. Oct. 29, 1958.

62. White, G.E. Response Characteristics of a Simple Instrument. Statham Laboratory Instrument Notes, No. 2.

63. White, G.E. The Meaning of Natural Frequency. Statham Laboratory Instrument Notes, No. 12.

64. White, G.E. Liquid Filled Pressure Gage Systems. Statham Laboratory Instrument Notes, No. 7.

65. Iberall, A.S. Attenuation of Oscillatory Pressures in Instrument Lines. Journal Research, National Bureau of Standards 45:85, 1950 (Paper RP 2115).

66. Barton, J.R. A Note on the Evaluation of Designs of Transducer for the Measurement of Dynamic Pressures in Liquid Systems Statham Laboratory Instrument Notes No. 27, 1954.

67. Broemser, V.P. Der Differential sphygmograph. Z6sch. f. Biol. 88: 264, 1928.

68. Wiggers, C.J. The Pressure Pulses in the Cardiovascular system. London: Longmans Green and Co., 1928.

69. King, A.L. and Lawton, R.W. Elasticity of Body Tissues. Medical Physics, Edited by Otto Glasser, Chicago: Yearbook Publishers, 2:303, 1950.

70. Houwink, R. Elasticity, Plasticity and Structure of Matter. New York: Dover 1952, Page 194.

71. King, A.L. Law of Elasticity for an Ideal Elastomer Amer. J. Phys. 14: 28, 1946.

72. Wiggers, C.J. Personal Letter of Dec. 2, 1958.

73. Wiggers, C.J. Amer. J. Physiol. (Distensibility Studies in Dogs) 123:644, 1938.

74. Frank, O. Fitzungher d. Besell f. Morph. U. Physiol. Munchen 37:23, 1927. Biol 71: 270, 1920.

75. Hockrein, C. Munchen, Med., Wehnsehv 73:1512, 1936.

76. Remington, C.R., Am. J. Physiol. 144:536, 1945.

77. Guyton, A.C., The Venous System and Its Role in the Circulation, Modern Concepts of Cardiovascular Disease. Am. Heart Assoc. 27:483, 1958.

78. Ibid (1) Page 615.

79. Lion, K. Mechanic - Electric Transducer. Review of Scientific Instruments, 27: 222-225, 1956.

80. Nyboer, J. Plethysmograph. Medical Physics, Edited by Otto Glasser, Chicago: Yearbook Publishers, Vol. II, 1950, Page 736.

81. Smith, B and Colls, J.A. Journal Scien. Instr. 60:361, 1937.