KOROTKOFF SOUNDS: EFFECT OF VESSEL ELASTICITY ON SOUND FREQUENCY

A Thesis

Presented to

the Faculty of the Department of Mechanical Engineering

University of Houston

In Partial Fulfillment

of the Requirements for the Degree

Master of Science in Mechanical Engineering

by

Ben Joseph Brookman Jr.

January 1969

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ABSTRACT

The clinical determination of blood pressure is commonly made by the auscultatory technique, i.e., listening to Korotkoff sounds. Although the accuracy of the auscultatory method has been adequately documented, the factors which produce the Korotkoff sounds have not. Some investigators believe that the sounds are due to blood vessel wall vibrations, others believe that they are due to blood flow disturbances. Many believe that these two factors can not be separated. Therefore, to shed light on the factors that underlie the cause of the sounds, this investigator elected to determine if there is a relationship between the frequency of Korotkoff sounds and the modulus of elasticity of the artery. This investigation and its results form the subject of this thesis.

In order to determine the contribution of vessel elasticity to the production of Korotkoff sounds, an experimental investigation of the effect of vessel elasticity on Korotkoff sound frequency has been conducted. A model was constructed to simulate conditions in the upper arm during blood-pressure determination. The experiment was performed using both cylindrical rubber tubes and actual carotid, femoral, and brachial canine arteries. The mean diameter of these tubes and arteries was very close to 3.0 millimeters (mm.) for all specimens, while their length was 25.0 mm. The modulus of elasticity of the tubes and arteries was measured with a special compliance-measuring apparatus. The moduli

for the rubber tubes ranged from 300 to 3600 pounds per square inch(p.s.i.)(or 15,500 to 186,000 mm. of Hg). The canine artery elasticities varied from 50 to 200 p.s.i.(or 2590 to 10,300 mm. of Hg). The sound waves they produced were recorded using a piezoelectic crystal transducer connected to a storage oscilloscope via a type 2A61 Tektronix differential amplifier. These sounds were monitored aurally as they were being recorded by use of a preamplifier, amplifier, and loud speaker.

The experimental investigation showed that, for the rubber tubes, the sound frequency range was 400 to 1110 Hertz (Hz.). For the canine arteries the frequencies varied from 196 to 250 Hz. Semilogarithmic graphs of vessel elasticity versus Korotkoff sound frequency were plotted. These plots indicated that the frequency of the sounds varied logarithmically with elasticity. In addition it was found that a four-fold increase in the test chamber volume did not significantly alter the sound frequency. Also a five-fold increase in the test-fluid mean pressure did not alter the frequency of the sounds.

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CHAPTER I

INTRODUCTION

Blood pressure measurement has been increasing in importance since 1903 when Harvey Cushing [1] * added blood pressure to the previously introduced fever chart of hospital patients. Arterial blood pressure is an important measure of cardiovascular performance. Although blood pressure is not constant throughout the human body, the accepted meaning of "blood pressure" is the pressure in the aorta; it is however, frequently measured in the brachial artery in the upper arm. It can be measured directly, by arterial cannulation, or indirectly, by techniques applied to the outside of the body. The accurate and reliable results of direct blood pressure measurement are more than offset by patient discomfort and risk of infection. As a result, indirect methods have been adopted.

The most widely accepted indirect blood pressure measurement method is the auscultatory technique which is an extension of a method devised by Riva-Rocci in 1896, in which he measured systolic(maximum)blood pressure by measuring the pressure in a pneumatic cuff required for obliteration of the arterial pulse distal to the occlusion. The cuff and its pressure indicator became known as a spygmomanometer.

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*Numbers in brackets refer to the bibliography.

The auscultatory method of blood pressure measurement, which measures both systolic and diastolic(minimum)pressures, was proposed by N.S. Korotkoff [2] in 1905. A description of this method follows: The spygmomanometer cuff is placed securely around the upper arm of the subject. To minimize gravitational affects the cuff is placed at approximately the same vertical elevation as the subject's left ventricle. The cuff is then inflated to a pressure higher than systolic blood pressure, thus occluding the brachial artery and stopping blood flow in the arm. The spygmomanometer cuff pressure is then gradually reduced(2-3 mm. of Hg. per heart beat). As the cuff pressure falls below the systolic value, the artery momentarily opens and blood spurts through the opening. Because the blood pressure is pulsatile, the artery again closes as the arterial blood pressure falls below cuff pressure. With continued deflation of the cuff, this sequence of events occurs periodically until cuff pressure falls below the diastolic value. This sequence of events is illustrated in Figure 1. At each opening of the artery a sound is produced which can be heard with the use of a stethoscope placed just distal to the occluding cuff. The American Heart Association and the Cardiac Society of Great Britain [3] in 1939 recommended that the systolic blood pressure be read as that cuff pressure corresponding to the first sound heard through the stethoscope. The diastolic pressure should be read as the cuff pressure corresponding to the last sound heard, or,

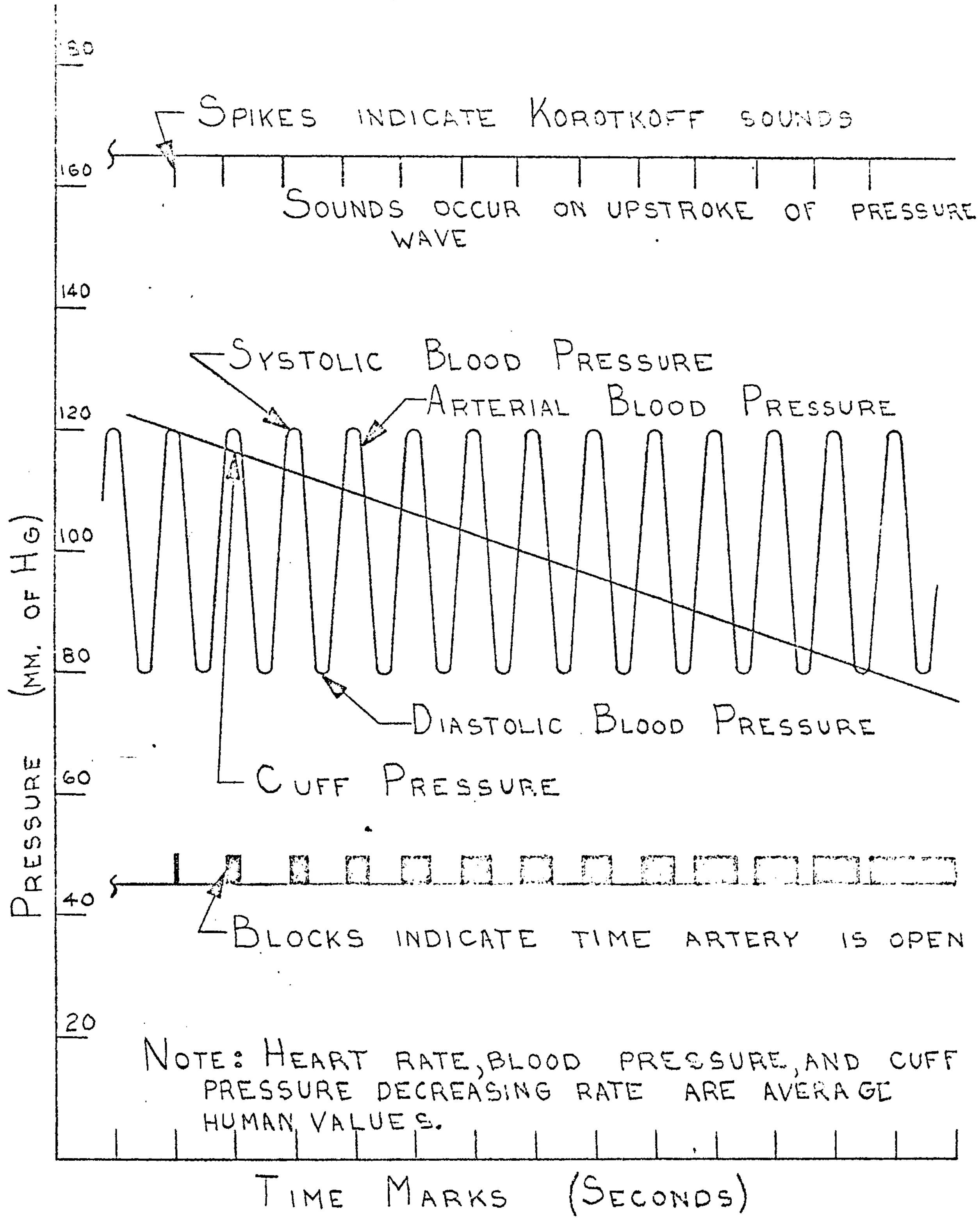


FIGURE 1. PRESSURE VERSUS TIME
(ARTERIAL BLOOD PRESSURE, CUFF PRESSURE, AND KOROTKOFF SOUNDS)

if the sounds do not disappear, the point of muffling of the sounds should be read. Both groups recommended pressure-cuff sizes that provided the most accurate pressure measurements. The recommendations were on publications up to that time. Studies comparing direct and indirect blood-pressure recordings include those of Steele [4] and Van Bergen et. al. [5] .

Although the auscultatory indirect blood pressure measurement method now gives fairly accurate results, the cause of the Korotkoff sounds has never been adequately determined. The origin of Korotkoff sounds was first investigated by Korotkoff [2] himself. He suggested that the mechanism for sound production was the turbulence of the blood as it spurted through the partially occluded brachial artery. Kositskii [6] thought that the sounds were due to the impact of the blood flowing under the cuff, on the stationary column of blood just distal to the occlusion. MacWilliam and Melvin [7] , from studies on thin rubber tubes, concluded that the sounds were due to a rapid change in the shape of the vessel produced by the difference in blood pressure and cuff pressure. Korotkoff sound origin was believed, by Erlanger [8] , to be due to water hammer. Chungcharden [9] stated that Korotkoff sounds were due to turbulence alone. All of these theories and others have been reviewed by Geddes, Hoff, and Badger [10] and McCutcheon and Rushmer [11] . The conclusion most often reached is that no single theory adequately explains the cause of Korotkoff sounds. However, the origin of these sounds can be separated into two distinct categories, flow-associated and vessel-associated. Possible flow-associated causes include fluid characteristics, turbulence, vortex shedding, water hammer, and cavitation. Fluid characteristics include the non-Newtonian behavior of blood. Possible vessel-associated causes include vessel elasticity and system elasticity. We can readily see that vessel elasticity would be important, if the sounds could be attributed to sudden vessel-shape changes.

In this study, vessel elasticity was varied while system volume, pulsatile pressure, test fluid, and vessel length were held constant. This investigation was conducted to determine to what extent change of vessel elasticity would affect the Korotkoff sound frequency. This investigation was undertaken with the hope of better understanding the mechanism of Korotkoff sounds.

Presented within this report are elasticity data for thin rubber tubes and canine arteries. The elasticities were obtained using a compliance-measuring apparatus. Also presented are sound frequencies from both thin rubber tubes and arteries as measured on a storage oscilloscope. The rubber tubes were chosen with the intent of obtaining a large range of elasticities while keeping the wall thickness as small as possible. The carotid, femoral, and brachial arteries were obtained from four dogs. The arteries were extracted by the author within one-half hour after death of the animals.

CHAPTER II

EXPERIMENTAL APPARATUS AND METHOD

Elasticities of the rubber tubes and arteries were measured with the use of a compliance-measuring apparatus which permitted measuring of the volume-pressure relationship. Since undisturbed human or canine arteries within the body are extended about 20% of their unstretched length, the arteries used in this test were stretched by the same amount. It was desirable to stretch the rubber tubes 20% also, but, due to their greater stiffness, this was very difficult. As a result the rubber tubes were extended 10% of their unstretched length for the elasticity measurement. A complete description of the compliance-measuring apparatus and the method of use is given in the appendix.

An experimental apparatus(see Figure 2)was constructed to simulate the upper arm during blood pressure determination. A plexiglass tube 4.4 centimeters(cm.) in diameter with a volume of 220 cm^3 and a number ten rubber stopper in each end was used as the test chamber. The rubber tube or artery was mounted to tubes which passed through the rubber stoppers so that the test specimen was in the center of the test chamber with the connecting tubes protruding from the chamber through the rubber stoppers. Pulsatile flow was provided to the system by a positive displacement Randolpf roller pump. Since the blood in the

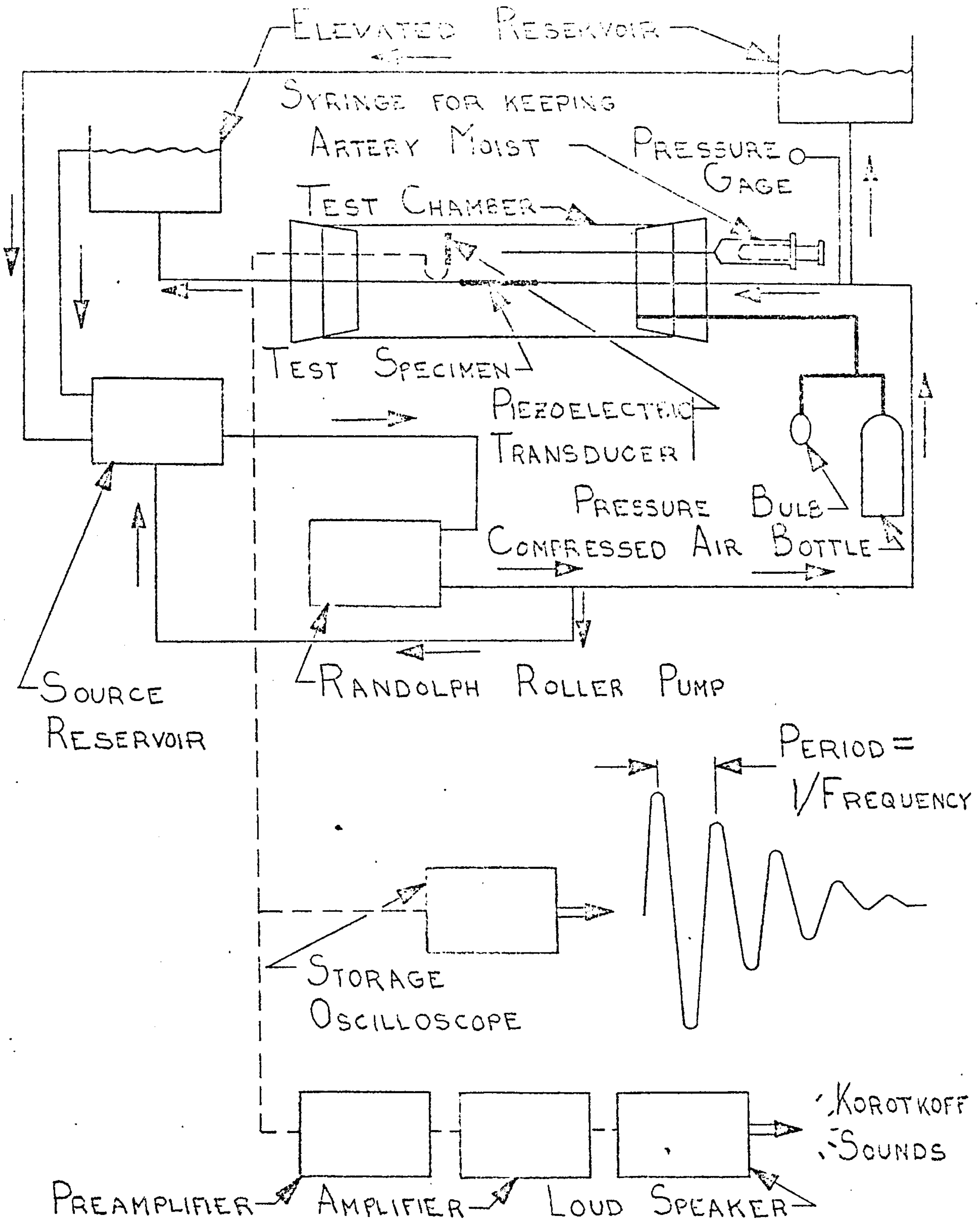


FIGURE 2. EXPERIMENTAL APPARATUS

human body flowing through a brachial artery is about 1/16 of the heart output, a bypass line was utilized on the pump to simulate this condition. Thus the test fluid would begin in a source reservoir, flow through the roller pump, and split into two lines with 1/16 of the total flow being pumped through the test chamber and back into the source reservoir. The remaining fluid leaving the pump would flow directly into the source reservoir. The roller pump speed was variable and was set to approximately 60 pulses per minute. With the bypass line and the pump speed set as described above, the flow rate of fluid through the test chamber was found to be about 180 milliliters per minute which is the same as the value reported by Wade and Bishop [12] , who pointed out that this value is only an approximate one.

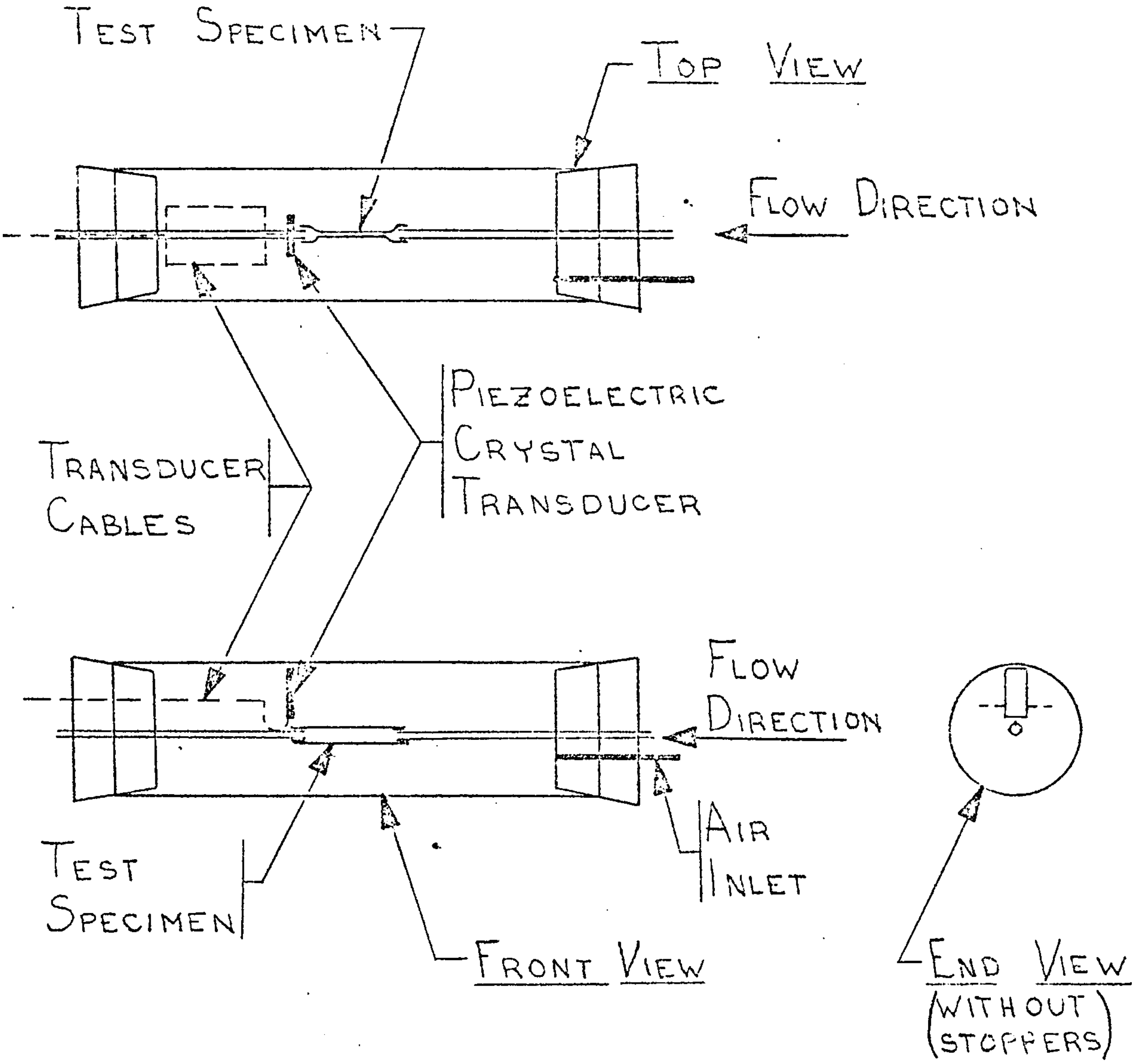
The systolic and diastolic pressures were 130 and 80 millimeters mercury(mm. of Hg) respectively. The test chamber was equipped with an air inlet so that the chamber could be pressurized and the test specimen occluded. The occluding pressure was obtained by pumping a pressure bulb by hand or by use of a compressed air bottle. The air bottle was necessary to occlude the high elasticity specimens.

When the brachial artery in the upper arm is occluded, the mean blood pressure does not rise significantly proximal to the occlusion. To simulate this condition in this model, an elevated reservoir was connected to the upstream chamber fluid inlet. When the rubber tube or artery was occluded, this reservoir was allowed to overflow into the source reservoir, thus keeping the upstream mean pressure from rising significantly. Also, when the brachial artery is occluded, the mean blood pressure distal to the occlusion does not drop to zero, but to about 30 to 50 mm. Hg. This condition was simulated by adding an elevated reservoir on the downstream side of the test chamber. Next it was noticed that when an artery is in an animal or human, it is kept moist on its outer circumference by body fluids. Thus a water-filled syringe was injected into the model test chamber such that water could be dropped onto the artery from time to time. This was not done for the rubber tubes. The chamber used was air-filled since the theoretical analysis of Morgan and Ferrante [13] and Womersley [14] indicate that the effects of surrounding tissues are probably small. A schematic of the experimental apparatus is shown in Figure 2.

As in the elasticity determination the arteries were extended 20% of their original length and the rubber tubes were stretched 10% of their original length. The original test length for all specimens was 25 mm. The fluid used to simulate the blood was a solution of glycerin and water with a dynamic viscosity of .03-.05 poise which compares favorably with that of blood.

Also shown in Figure 2 is the electronic equipment used to record the sounds. A piezoelectric crystal transducer was used to detect the sounds. The crystal dimensions were 0.5 x 2.5 x 10.0 mm. approximately. The sounds caused the crystal to bend slightly, creating a minute voltage difference between the electrodes on the two sides of the crystal. This voltage difference was applied to a storage oscilloscope. A preamplifier, amplifier, and loud speaker were used to monitor the Korotkoff sounds as they were being recorded. The audible sounds were very helpful in identifying extraneous room noises. The oscilloscope sweep was triggered by the Korotkoff sound signal by using the output of the preamplifier. This enabled the sound to be recorded on the oscilloscope starting from the beginning of the sound. A triggered signal as seen on the oscilloscope is shown in Figure 2.

The location of the piezoelectric crystal transducer was found to be extremely critical. Studies were performed to determine the location for maximum sound amplitude. After both upstream and downstream locations inside and outside the test chamber were tried, it was evident that the optimum crystal location was just distal to the test specimen inside the chamber. The transducer was therefore taped to the connecting tube at this site. Different size and material connecting-tubes were also tried. It was found that the diameter of the connecting tubes did not affect the sounds appreciably. On the other hand, the downstream connector was found to yield the greatest amplitude sounds when it was made of polyurethane. The upstream connector did not affect the sound amplitude. Thus the upstream connector was made from stainless steel tubing for rigidity, while the downstream connector was made from polyurethane. The connectors were about 15.0 cm. long with an inside diameter of 3.1 mm. This diameter was chosen to approximate the canine carotid, femoral, and brachial artery size. Initially it was found that almost all sounds could be heard with little difficulty, but nearly all sound traces contained noise which made the sound frequencies immeasurable. The position finally used was such that the transducer was mounted perpendicular to the connecting tubes while the lead wires were taped to the connecting tubes. This insured that the transducer was touching nothing except its lead wires. This kept the transducer from rubbing against anything and producing undesired noises. It was also found necessary to place that portion of the lead-wires that was between the taped-down portion and the rubber stopper in the shape of a rectangle. This is shown in Figure 3. This kept most of the connector-tube lateral movement from being transferred to axial transducer movement, where axial refers to the test-chamber centerline axis. Another mounting requirement is also shown in Figure 3. This is the orientation of the occluded test specimen. Best results were acquired with the specimen in the position shown, probably because lateral movement looking at the top view in Figure 3 could be best damped by the rectangular configuration as shown.



NOTE: TRANSDUCER LOCATION WITH RESPECT TO THE OCCLUDED TEST SPECIMEN IS IMPORTANT (COMPARE FRONT AND TOP VIEWS).

FIGURE 3. PIEZOELECTRIC CRYSTAL TRANSDUCER INSTALLATION

The procedure for running a test was to measure the modulus of elasticity of the test specimen. The next step was to turn on all electronic equipment. Next the piezoelectric transducer and test specimen were installed as shown. The roller pump was then started and the clamp on the upstream water reservoir tube was removed. The test specimen was then occluded by applying pressure from the pressure bulb or the compressed-air bottle depending upon whether large or small elasticity specimens were being studied. The chamber pressure was then decreased and a series of Korotkoff sounds occured as depicted in Figure 1. One of these sounds was recorded on the oscilloscope and the frequency was measured. Approximately 10 sounds were recorded for each specimen to insure consistency. Elasticities of the test specimens were then checked to make sure they were unchanged.

CHAPTER III

RESULTS OF THE EXPERIMENTAL INVESTIGATION

The mean radii of the rubber tubes ranged from 1.15 to 2.39 mm. The thickness varied from 0.38 to 1.66 mm. These dimensions, along with the data from the compliance-measuring apparatus, showed that the elasticities varied from 297 to 3610 p.s.i.(or 15,350 to 187,000 mm. Hg). The frequency range obtained was 400 to 1110 Hertz(Hz.) In the case of the canine arteries, the mean radii ranged from 1.53 to 1.89 mm., while the thickness varied from 0.42 to 0.67 mm. Along with the compliance data, these dimensions yielded elasticities of 57.6 to 205.0 p.s.i.(or 2,980 to 10,600 mm.Hg). The frequencies recorded were 196 to 250 Hz. These values are presented in tabular form in Tables I and II. Figures 4 and 5 show semilogarithmic plots of modulus of elasticity versus frequency for rubber membranes and canine arteries respectively. Both figures indicate that the frequency is a logarithmic function of the elasticity. For the rubber tubes, we have $F = -711. \text{Log}_{10} E + 2850$, and for the canine arteries, $F = -107.2 \text{ Log}_{10} E + 439$. These equations were obtained using a least-squares fit for the data. The data were analyzed on a PDP8 Digital Systems digital computer at the Baylor University College of Medicine. In the above two equations, the frequency, F, is in Hz., and the modulus of elasticity, E, is in p.s.i.

TABLE I

DIMENSIONS, ELASTICITY, AND SYSTEM FREQUENCY
FOR VARIOUS RUBBER TUBES

Mean Radius (mm.)	Thickness (mm.)	Elasticity (p.s.i.)	Frequency (Hz.)
1.52	0.38	3610	400
1.43	0.50	3200	410
1.81	0.68	2130	416
1.98	0.58	607	684
1.15	0.78	1060	714
2.01	0.83	1190	835
2.39	1.66	452	961
1.86	0.60	345	1000
1.88	0.78	297	1110

Tube Test Length - 27.5 mm.
(Stretched 10% of original length)

TABLE II

DIMENSIONS, ELASTICITY, AND SYSTEM FREQUENCY
FOR VARIOUS CANINE ARTERIES

Mean Radius (mm.)	Thickness (mm.)	Elasticity (p.s.i.)	Frequency (Hz.)
1.84	0.56	57.6	250
1.82	0.53	72.3	243
1.86	0.60	102.0	222
1.53	0.44	108.0	211
1.81	0.50	155.0	208
1.89	0.67	144.0	207
1.77	0.42	205.0	196

Artery Test Length - 30.0 mm.
(Stretched 20% of original length)

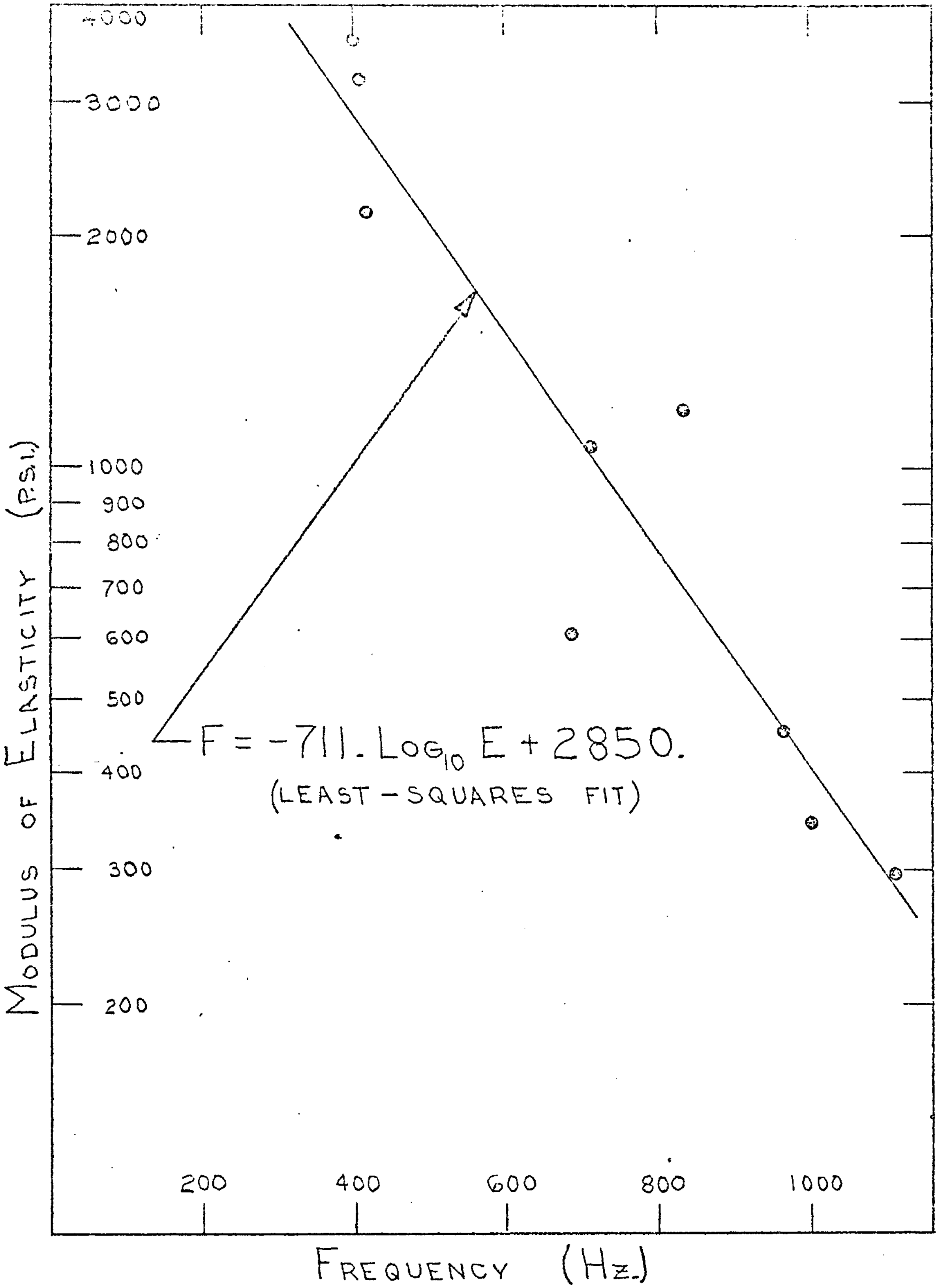


FIGURE 4. MODULUS OF ELASTICITY VS. FREQUENCY (RUBBER TUBES)

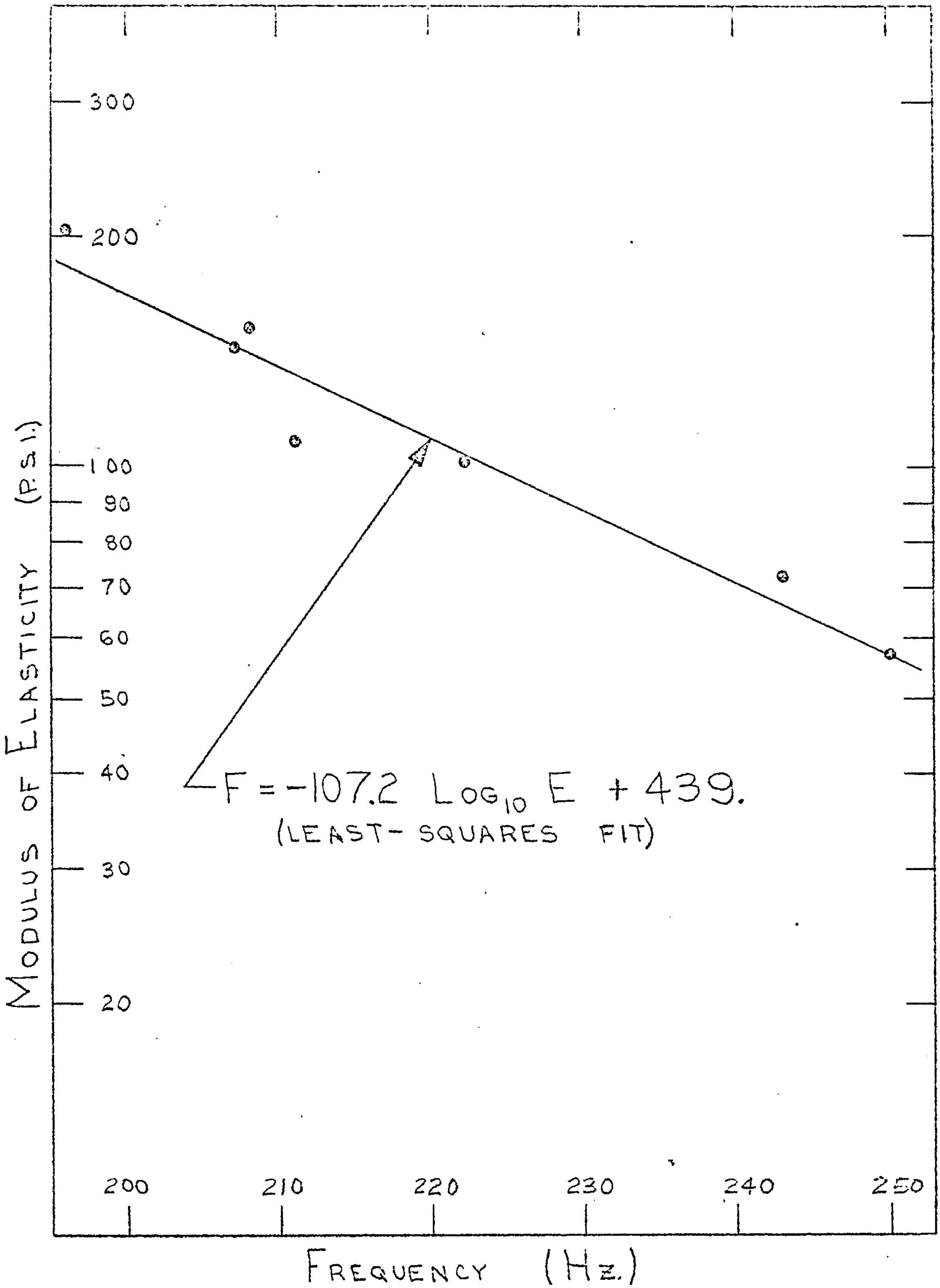


FIGURE 5. MODULUS OF ELASTICITY VS. FREQUENCY (CANINE ARTERIES)

During the portion of the test dealing with rubber tubes, additional information was obtained. The sound frequency of a tube was recorded using a 950 $cm.^3$ test chamber as well as the previously described 220 $cm.^3$ test chamber. There was no measurable frequency change when only the volume of the test chamber was changed. This test was repeated with other tubes, and the results were the same; no frequency change resulted from the chamber volume increase. Since the results obtained with the rubber tubes were so convincing, arteries were not tested in this manner.

Also noted was the fact that a change of mean fluid pressure from 60 to 300 mm. of Hg did not alter the sound frequency. This was shown to be true for arteries as well as rubber tubes. The tube connecting the upstream elevated reservoir to the system could be clamped, with the effect of greatly increasing the mean fluid pressure when the test specimen was occluded. This also did not alter the sound frequency, as recorded with the elevated reservoir tube open.

Therefore, in the range tested, it has been shown that the system frequency is independent of mean fluid pressure and test chamber volume, while it is a logarithmic function of the vessel elasticity. However, it should be noted that as vessel elasticity changed, other factors could have changed at the same time. The fluid properties remained essentially constant, but there is no guarantee that turbulence generation and vortex shedding, if they in fact occur, remain constant as elasticities are varied. In fact, one would suspect that they would be influenced by such an elasticity change. This points to the need for additional work and additional constraints. Also, for the rubber tubes, the mean radius varied from 1.15 to 2.39 mm., which is greater than a 100% change. The tubing thickness ranged from 0.38 to 1.66 mm., which is over a 300% change. No correlation could be obtained between sound frequency and either of the above factors. For the canine arteries, the mean radius varied only about 24% and the thickness only varied by about 60%. Again no correlation was found between sound frequency and either thickness or mean radius. It is encouraging to note that, even though the changes in mean radius of 100% and 24%, and the changes in thickness of 300% and 60% are very much different, the plots of elasticity versus frequency are very similar in that they are both logarithmic.

Elasticity, as determined in this experiment, involved measurement of one other variable. This was $\frac{\Delta P}{\Delta V}$ for each specimen. The elasticity formula used is $E = \frac{\Delta P}{\Delta V} \frac{R^3}{T} L$ and is derived in the Appendix. R is the mean radius, T is the thickness, L is the length, and $\frac{\Delta P}{\Delta V}$ is the reciprocal of the change of volume resulting from a pressure change divided by the pressure change.

The logarithmic plots shown in Figures 4 and 5 indicate that the grouping of variables as given in the elasticity formula do adequately represent the data. It should be pointed out that the range of frequencies and of elasticities for the arteries is much smaller than that for the rubber tubes.

CHAPTER IV

SUMMARY AND CONCLUSIONS

The relationship between vessel elasticity and the frequency of Korotkoff sounds has been investigated experimentally. A model was constructed simulating the upper arm and brachial artery during auscultatory blood pressure measurement. It was desired to vary only vessel elasticity, but vessel mean-radius and thickness also varied considerably. It is probable that turbulence generation and/or vortex shedding could also have varied. The moduli of elasticity of various rubber tubes and canine arteries were measured using a special compliance-measuring apparatus. This apparatus enabled the change in volume corresponding to a change in pressure to be measured. The tubing and artery thickness and inside diameter were determined using a micrometer and rods of known diameter, respectively. The elasticity was then calculated using the formula $E = \frac{\Delta P}{\Delta V} \frac{R^3}{T} L$. These arteries and membranes were then placed one at a time inside the test chamber. Pulsatile flow was provided to the system and Korotkoff sounds were produced. The frequency of each corresponding specimen was then measured with the use of a storage oscilloscope. With rubber tubes the mean fluid pressure was varied and the test chamber volume was increased. Graphs were then plotted of elasticity versus Korotkoff sound frequency.

The results obtained indicate that for the system described, frequency varies logarithmically with vessel elasticity. Also when the mean fluid pressure or the chamber volume is changed, the frequencies remain essentially unchanged. Although the elasticity has been changed and this change correlates with the sound frequency change, we must note that no attempt was made to observe the fluid flow to determine changes in turbulence or vortex shedding. The fact that similar graphs were obtained when very different vessels were used indicates that further study with more specimens and higher accuracy could verify and go beyond the present work.

The conclusion of frequency dependence upon vessel elasticity and independence upon test-chamber volume and mean fluid pressure could be of direct clinical importance. Blood pressure values are of primary importance in the diagnosis of many diseases and are a direct measure of cardiovascular performance. Goodman and Howell [15] suggested that changes in Korotkoff sound frequency might be important to clinical investigations. If, in addition to blood pressure, the frequency of Korotkoff sounds was recorded, then a change in the frequencies could be directly related to elasticity, which, in turn could be linked to possible cardiovascular abnormalities and diseases. This fact alone is an adequate incentive for further work on the mechanism of Korotkoff sound generation and further work in the area of changes of sound frequency with elasticities.

Areas of further study should include:

1. The use of an improved system to study more intimately the relationship between frequency and vessel elasticity.

2. The consideration of different tubing and artery lengths as a possible cause for frequency changes.

3. The development of another method for sound recording so as to shorten and simplify the transducer installation procedure.

4. The construction of a scaled up model for photographic studies of the flow patterns during sound production.

5. The repetition of the investigation of this report keeping the tube mean radii and thicknesses constant and varying the elasticity by means of material change.

6. The development of a method to measure, indirectly, Korotkoff sound frequency in man.

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APPENDIX

MODULUS OF ELASTICITY OF THIN WALLED TUBES

Consider $E = \frac{\text{STRESS}}{\text{STRAIN}} = \frac{F/A}{\Delta L/L}$

Where:

E . . . . . Young's modulus of elasticity

F . . . . . Applied force

A . . . . . Area over which force is applied

ΔL . . . . Change in length due to the force

L . . . . . Original length.

This formula assumes that the specimen initially has zero force applied to it. Looking at the more general case, we can see that this formula would become $E = \frac{(F_2 - F_1)/A}{\Delta L/L}$. It is noted that if F_1 is zero then this equation is the same as the original equation.

Now consider a thin-walled tube with mean radius R, thickness T, length L, inside pressure P_i, and outside pressure P_o. If a longitudinal section through the tube centerline is taken, and only half of the tube is considered, we can see that:

1. the stretching pressure is $P_i - P_o$ or P_i since $P_o = 0$
2. the bursting force is P_i(Area) or $P_i(\pi RL)$
3. the longitudinal crossectional area of the tube is 2LT.

Recall that stress is $(F_2-F_1)/A$. Then $F_2 - F_1$ is $\pi RL(P_{i2}-P_{i1})$ if $R_1 \simeq R_2$. Thus, stress is $\frac{\pi RL\Delta P}{2LT}$ or $\frac{\pi R\Delta P}{2T}$.

With attention focused on strain, the tube is observed as a flat plate of length $2\pi R$. Thus, it is noted:

1. the increase in length is $2\pi(R+\Delta R)-2\pi R$ or $2\pi\Delta R$

2. the fraction increase in length is $\frac{2\pi\Delta R}{2\pi R}$ or $\frac{\Delta R}{R}$.

Thus, E is $\frac{\text{STRESS}}{\text{STRAIN}}$ or $\left(\frac{\pi R\Delta P}{2T}\right)\left(\frac{R}{\Delta R}\right)$ or $\frac{\pi R^2\Delta P}{2T\Delta R}$.

Now we write ΔR in terms of ΔV (volume),

$\Delta V = \pi(R+\Delta R)^2 L - \pi R^2 L = \pi L\left[R^2+2R\Delta R+(\Delta R)^2 - R^2\right]$

If $(\Delta R)^2 \simeq 0$ then, $\Delta V = 2\pi LR\Delta R$ or $\Delta R = \Delta V/2\pi LR$. Substituting this value into the above equation yields $E = \left(\frac{\pi R^2\Delta P}{2T}\right)\left(\frac{2\pi LR}{\Delta V}\right) = \left(\frac{\Delta P}{\Delta V}\right)\left(\frac{\pi^2 R^3 L}{T}\right)$ where $\frac{\Delta P}{\Delta V}$ is the pressure change inside the tube caused by a volume change divided by this volume change.

Thus, the above is an equation for elasticity, all the terms of which can be easily measured. In this investigation inside diameter was measured by measuring the diameter of a rod that would slip inside of the tube to be measured without creating circumferential tension, but still fit snugly. Different size rods were machined prior to the investigation. The thickness of the tubes was determined by measuring the outside diameter with a micrometer caliper while the tube was still pulled over the inside diameter measuring rod. $\frac{\Delta P}{\Delta V}$ was measured using the apparatus shown in Figure 6. A micrometer syringe was connected to the specimen to be tested such that a small volume could be added accurately. A pressure tap was placed between the syringe and the test specimen so that pressure changes could be

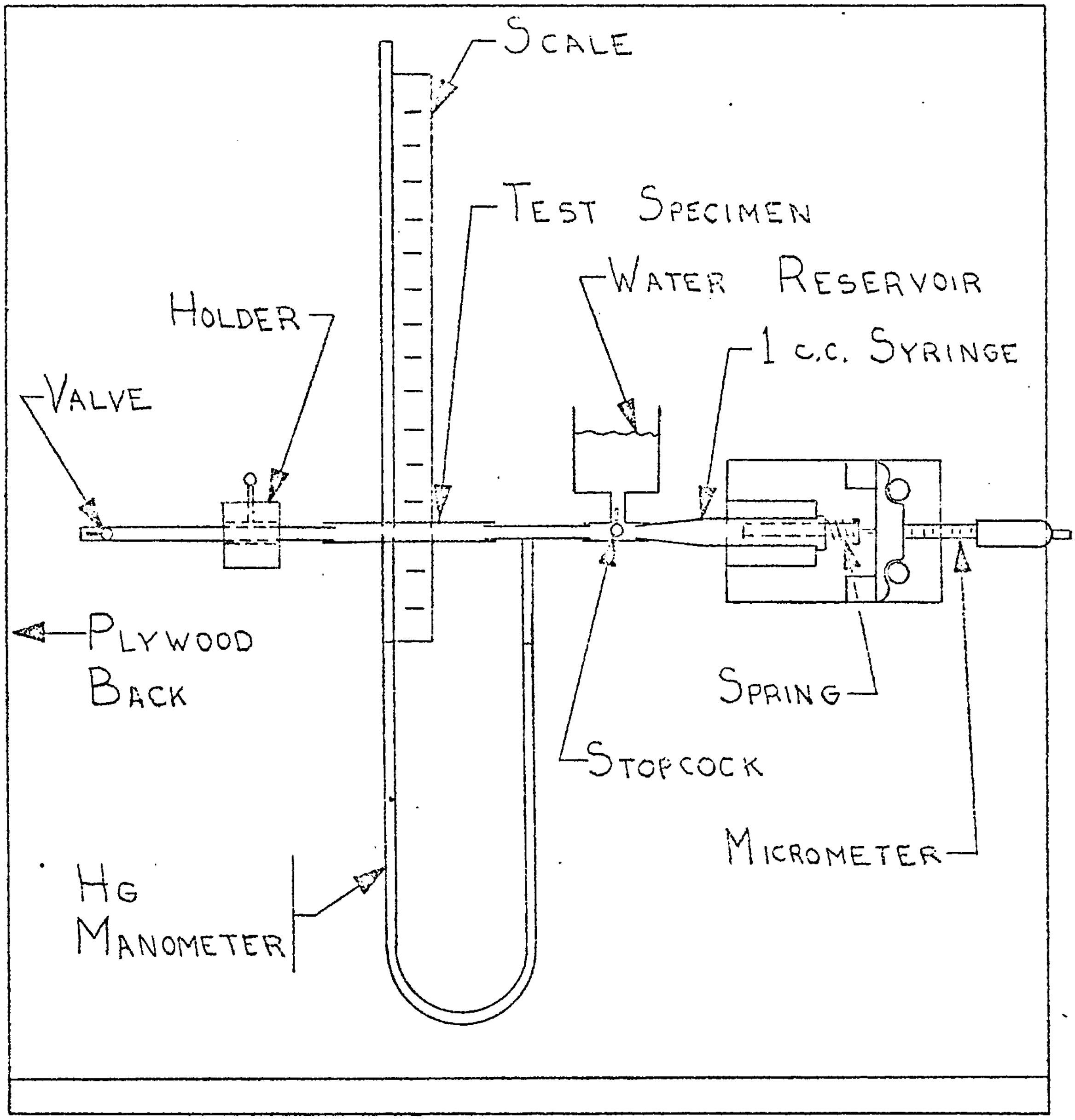

FIGURE 6. COMPLIANCE-MEASURING APPARATUS

simultaneously recorded with volume changes. The pressure was measured with a mercury manometer. The syringe, test specimen, and one side of the manometer were completely filled with water while the remaining side of the manometer was open to the atmosphere. A water reservoir was placed between the test specimen and the syringe such that water could be added when needed. The procedure was to attach the test specimen to the system using surgical silk, fill the system with water, and then record volume changes corresponding to pressure changes for pressures of 0 to 200 mm. of Hg. Next the specimen was removed and replaced by a rigid tube. Again pressure changes corresponding to volume changes were recorded. This latter procedure yields $\frac{\Delta P}{\Delta V}$ for the system while the former yields $\frac{\Delta P}{\Delta V}$ for the system and the test specimen. If, at each pressure, the volumes are subtracted, the result will be the change in volume of the test specimen at that pressure. Figures 7 and 8 show typical pressure versus volume curves for a rubber tube and a canine artery, respectively. Figure 9 shows pressure versus volume curves for the system and for the system plus the test specimen. A correction was made in the pressure readings to account for the water column on one side of the manometer. With no correction, when the manometer pressure reading was 200 mm. of Hg, the actual pressure would have been 200 mm. of Hg minus 100 mm. of water or 200 - 7.35 or 192.65 mm. of Hg. Since 7.35 is about 4% of 192.65 the error in the pressure

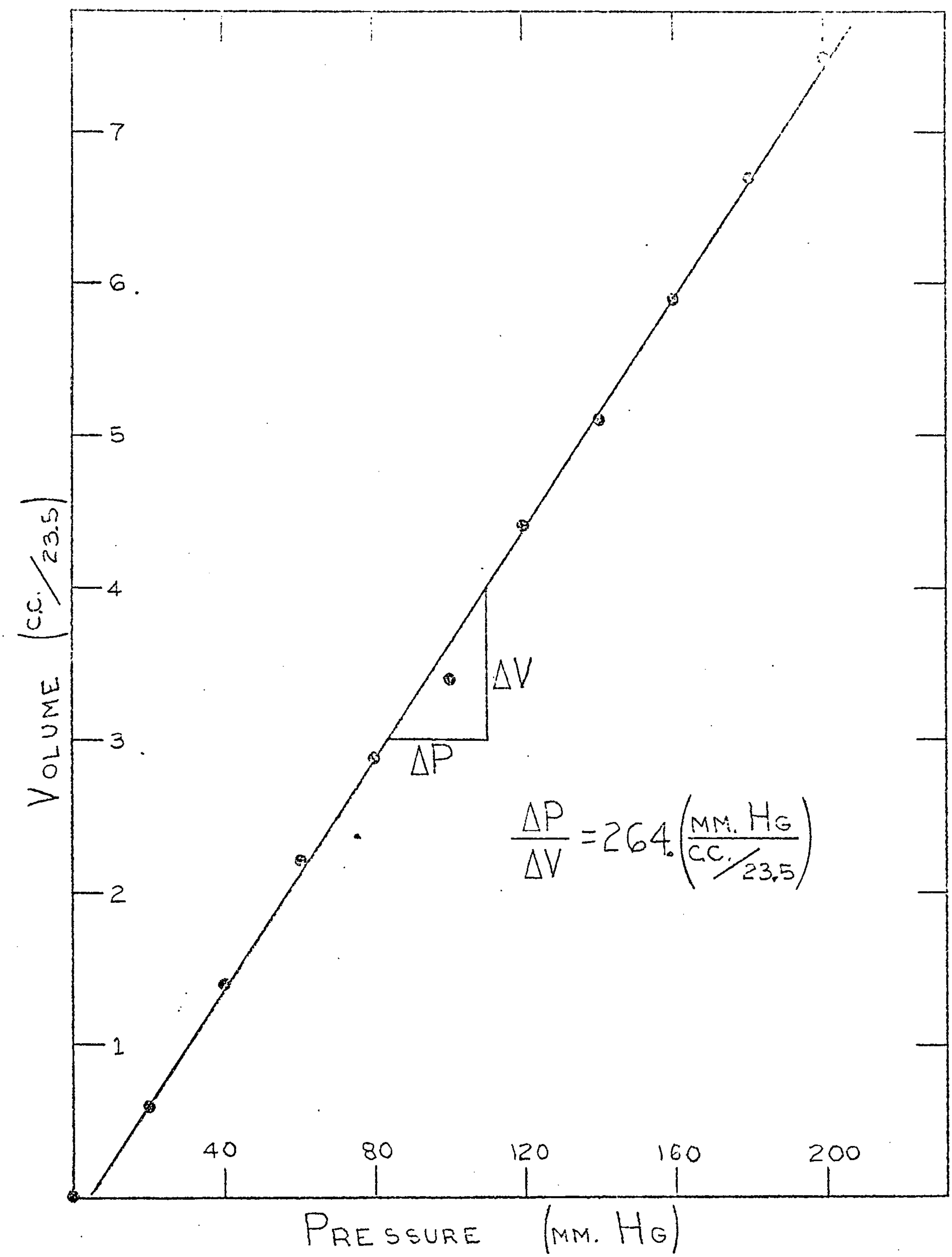

FIGURE 7. PRESSURE VS. VOLUME
(RUBBER TUBES)

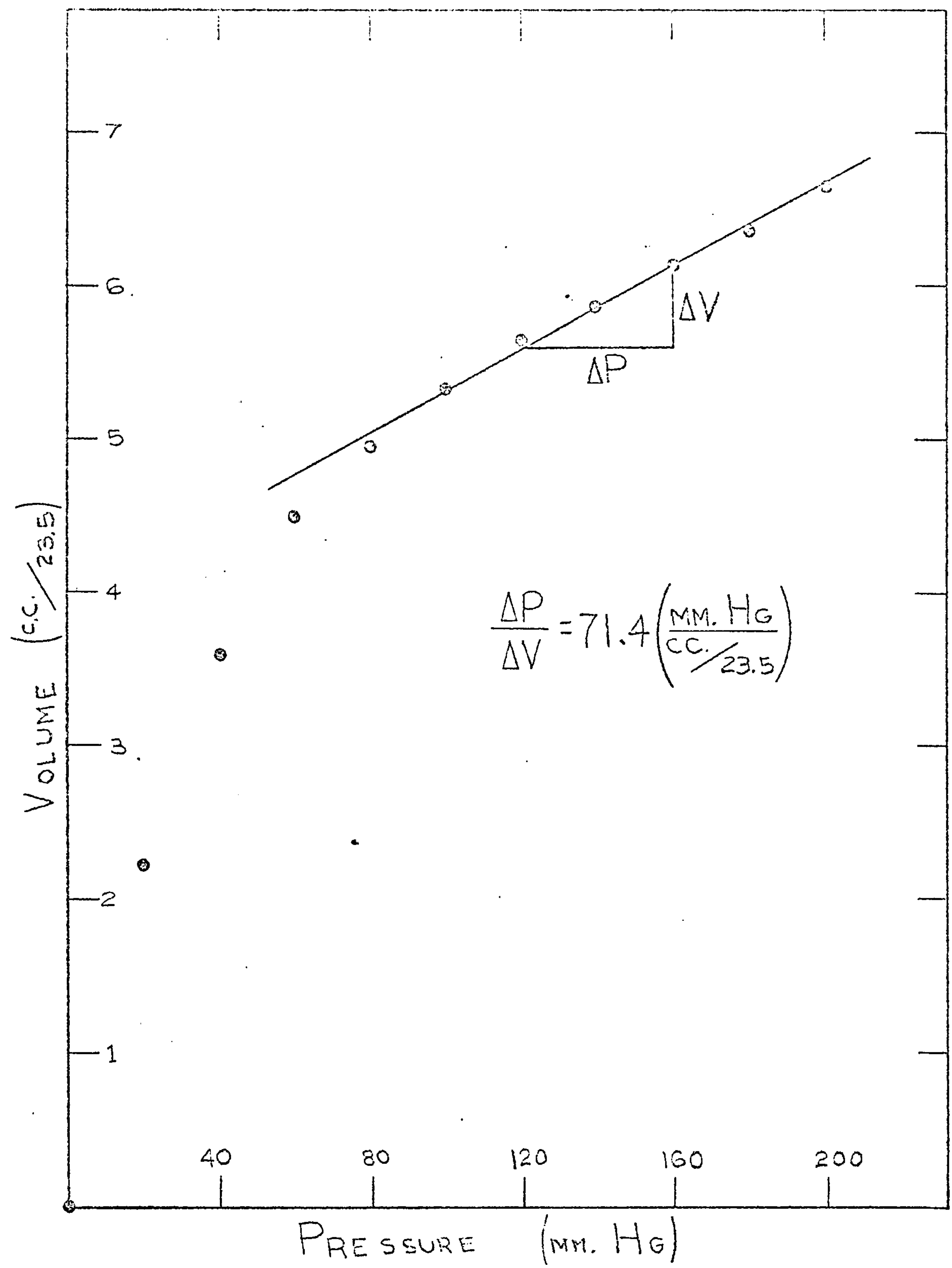

FIGURE 8. PRESSURE VS. VOLUME (CANINE ARTERIES)

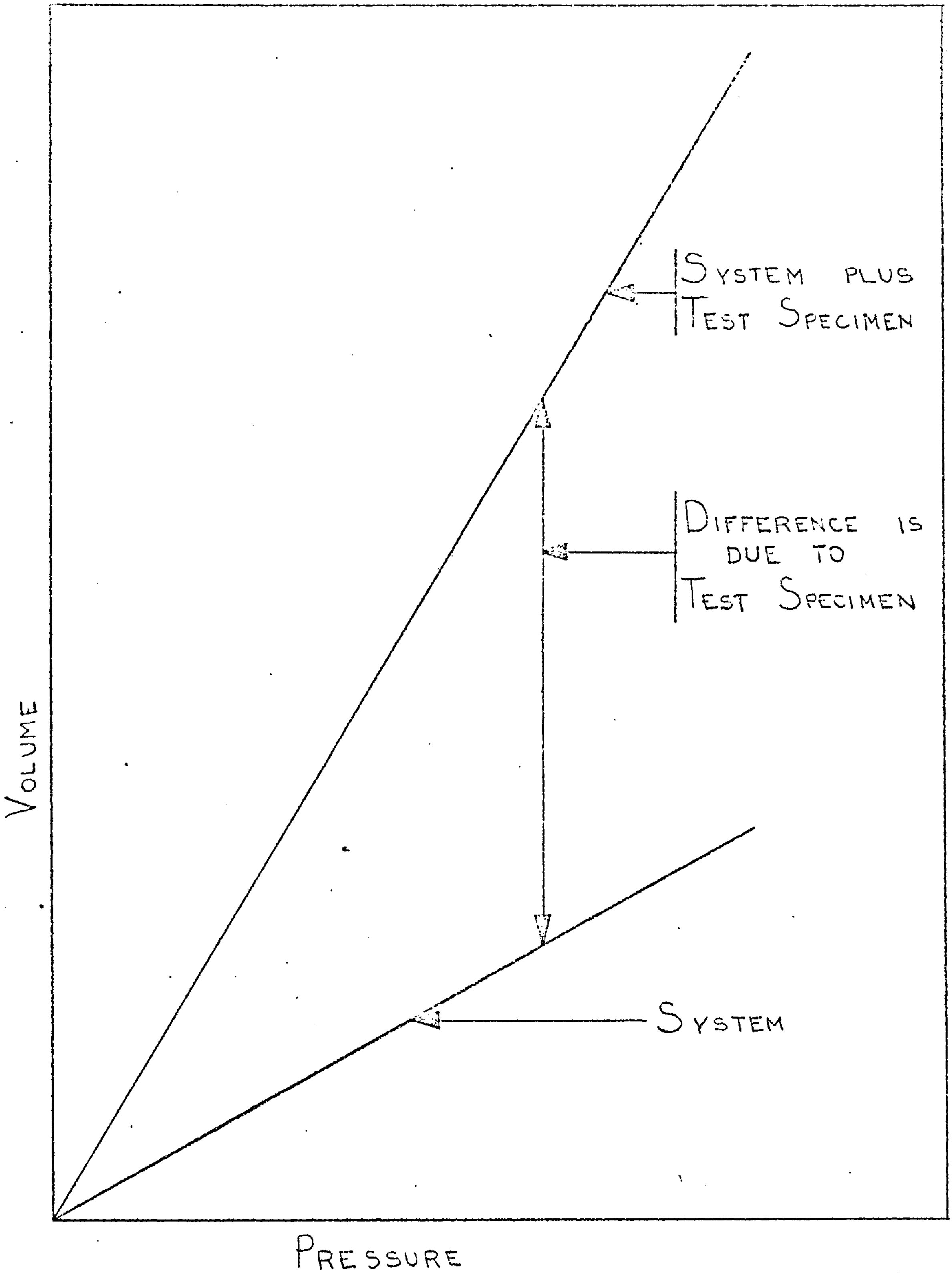

FIGURE 9. PRESSURE VS. VOLUME
(SYSTEM AND SYSTEM PLUS
TEST SPECIMEN)

reading could have been approximately 4%. Similarly, with no correction, and a manometer reading of 100 mm. of Hg, the error would still be 4% since when the pressure reading is cut in half, the water level is also reduced by half. Also to be noted in Figure 8 is the fact that the entire pressure versus volume curve is not linear. This is a representative curve since all arteries behaved in the same way. The $\frac{\Delta P}{\Delta V}$ figure was taken in each case from the linear portion of the graph as shown by the straight line in Figure 8. This nonlinearity at low pressures is due to the difficulty in starting the test at the actual initial volume of the artery. An artery has no rigidity, except when its internal pressure is greater than its external pressure. As a result the arteries tested did not fully assume the shape of a cylinder until the internal pressure was from 60 to 80 mm. of Hg. The large volume-increases prior to this time resulted from the expansion of the artery to its cylindrical shape. Thus the nonlinearity shown in Figure 8 is due to the measurement system and not to the artery itself.