in the performance of many existing control and communications systems.

The model was designed to match the actual biological data at certain known "points." It is hoped that the model will be capable of interpolating or extrapolating the data between these points. The model can, however, relate the behavior of the "clock" to various stimuli more accurately than an intuitive idea of the "clock's" functioning could.

The electronic model of the biological clock is also valuable because of its role as an organizer. The model collects together a large variety of facts into a compact diagram and interrelates them, one to another. The model serves to consolidate the experience obtained thus far with the biological prototype and simplifies the task of formulating new biological experiments.

As the model is evolved through successive prototypes, the functional correspondence will become better. It is expected that the acquisition of new biological information will lead to even greater refinements in the model.

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A Transducer for the Continuous External Measurement of Arterial Blood Pressure*

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***Summary*—The objective of the research was to develop a transducer to measure arterial blood pressure. It was required that the transducer provide a continuous measure of blood pressure, that it not encumber the subject and that it not require cannulation. Two basic techniques were investigated both analytically and experimentally.**

First, an indirect measurement of blood pressure based on arterial deflection was attempted. Difficulties of calibration and sensitivity to physiological changes of skin and tissue around the artery led to the decision to attempt a more direct measurement of arterial blood pressure. In this second approach, arterial deflection is restrained by the transducer and the resultant restraining force is measured.

A mathematical model of the transducer artery system was developed and was used as a guide for the design of the experimental prototype transducers. Tests performed on these experimental transducers gave results consistent with the predictions of the model. These transducers have been used to measure blood pressure at large superficial arteries, with results comparable to sphygmomanometer determinations.

TWO TECHNIQUES are commonly employed to obtain arterial blood pressure. The sphygmomanometer provides only intermittent measurements of the systolic and diastolic levels. Intra-arterial catheterization requires surgical procedures that limit its application to clinical environments. Attempts have been made to measure blood pressure by other methods, including the measurement of arterial distention,[1] of pulse wave velocity,[2]

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[1] R. W. Correll, "Theoretical Analysis and Preliminary Development of an Indirect Blood Pressure Recording System," M.S. thesis, Massachusetts Inst. of Tech., Cambridge, Mass.; June, 1959.
[2] G. Weltman, "The Continuous Measurement of Pulse Wave Velocity," M.S. thesis, University of California, Los Angeles; 1959.

of ophthalmic artery occlusion and the use of finger-type sphygmomanometers. The first two are still in the experimental stage. While equipment for the last two is commercially available, these methods usually provide intermittent systolic measurements only.

This paper describes methods for obtaining a continuous measurement of the arterial blood pressure by external means. A transducer for physiological monitoring on active subjects is described. This transducer provides a measurement which is relatively insensitive to normal subject movements. Application of the measuring device does not appreciably interfere with subject activity.

Initially, work was directed toward measurement of superficial effects of arterial distention. Experiments were conducted in an effort to relate displacement of skin above an artery to the blood pressure causing arterial distention. Relative arterial distention was determined by measuring the elevation of the skin surface directly above an artery. Measurements of this type have been described by R. W. Correll.[1] Both strain gage and capacitance devices were developed, each using the principle of a differential measurement; the skin surface deflection above the artery was measured relative to skin deflection immediately adjacent to the artery (Fig. 1). Therefore, large-scale movements of the skin surface such as those due to muscle activity do not cause spurious gage outputs.

The springs shown in Fig. 1 must be sufficiently flexible to allow the skin to react as if no measuring device were present. In practice, the spring constants were less than 1/10 the apparent spring constant of the skin-artery system.

It soon became apparent that skin displacement effects could not be related to blood pressure to the exclusion of other physiological influences such as skin tension and muscle tone. For example, certain drugs reduce the blood pressure, but the effect on tissue tone of the patient is such that the arterial distention increases. The displacement gage would then register an increase instead of a decrease in blood pressure.[3] Because of this limitation, work on this technique was dropped in favor of the direct force method described below.

To obtain a more accurate visualization of the nature of a blood pressure measuring system, it is useful to consider a mechanical model of the system characteristics. Although a model cannot completely represent all the factors in a physical system, even rudimentary models may provide insight into the nature of the measurement problems and may, as in this case, provide clues to improved techniques.

The model is based on the assumption that the deflections are so small that nonlinearities are insignificant. Therefore, compressible and extensible tissue are represented as linear springs. The artery is assumed to rest on a firm base, to have elastic walls and to be surrounded by uniform tissue. The transducer is assumed to have side structures that allow it to rest on the skin and a center structure that responds to arterial pressure (see Fig. 2).

The mathematical model is developed as follows (see Fig. 3). Arterial pressure is represented by the force F_a; k_4 is determined by the tangential elasticity of the artery walls, k_5 by the compressibility of the tissue immediately adjacent to the artery and k_6 by the compressibility of the tissue directly above the artery. These springs represent the physiological parameters, which are variable and are generally unknown. In the transducer, k_1 represents the transducer mounting and F_1 the force used to hold the system against the skin surface.

Ideally, k_1 should be infinitely stiff to restrain the transducer rigidly with respect to the bone structure and hence to a fixed position relative to the artery. In practice, it was found that such a system was impractical, if not impossible. An alternative approach using a pneumatic loading principle (described below) was developed to maintain F_1 constant and reduce k_1 to nearly zero. This, in essence, accomplishes the same as the ideal system except for relaxation of tissue represented by spring k_5.

The masses of the various components both in the transducer and artery system are ignored because of the relatively low driving frequency of the system. This assumption has been proved valid for the transducer and is be-

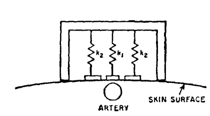


Fig. 1—Diagram of the arterial distention gage. The difference between outer spring (k_2) and inner spring (k_1) deflection is measured by either a capacitance or strain gage system.

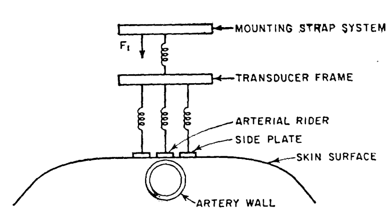


Fig. 2—Schematic diagram of the transducer artery system.

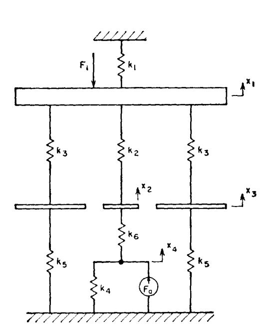


Fig. 3—The mechanical model of the physical system shown in Fig. 2. Only vertical force and deflection components are considered; for this reason the horizontal skin surface (Fig. 2) does not appear in the model. The quantity $x_2 - x_3 = \delta$, the parameter measured; k_2 and k_3 are spring rates of the transducer structure; x_1, x_2, x_3 and x_4 are displacements.

[3] R. Adams, R. W. Correll and N. H. Wolfeboro, "Cuffless, noncannula, continuous recording of blood pressure," *Surgery*, pp. 46–54; January, 1960.

lieved to be reasonable for the artery. All skin tension effects are ignored because the skin is forced to lie flat under the transducer, perpendicular to the measured displacement; thus it cannot exert vertical components of force on the transducer measuring plate.

The equations for this model follow.

Center plate forces are:

$$k_6(x_4 - x_2) = k_2(x_2 - x_1). \tag{1}$$

Forces at x_4 are:

$$F_a = k_6(x_4 - x_2) + k_4 x_4. \tag{2}$$

Forces on top plate are:

$$k_2(x_2 - x_1) = F_1 + \left(k_1 + 2\frac{k_3 k_5}{k_3 + k_5}\right)x_1. \tag{3}$$

Forces on side plate are:

$$0 = k_5 x_3 + k_3(x_3 - x_1). \tag{4}$$

These equations give the following result:

$$x_2 = \frac{-k_e F_a + k_2\left(\frac{k_4}{k_6} + 1\right)F_1}{k_2^2\left(\frac{k_4}{k_6} + 1\right) - k_e\left[k_4 + k_2\left(\frac{k_4}{k_6} + 1\right)\right]} \qquad 5)$$

$$x_2 - x_3 = \frac{\left(-k_e + \frac{k_3 k_2}{k_3 + k_5}\right)F_a + \left[Ak_2 - \frac{k_3}{k_3 + k_5}(k_4 + Ak_2)\right]F_1}{Ak_2^2 - k_e[k_4 + Ak_2]}, \tag{6}$$

where

$$A = 1 + \frac{k_4}{k_6} \tag{7}$$

$$k_e = k_1 + k_2 + \frac{2k_3 k_5}{k_3 + k_5}\cdot \tag{8}$$

The quantity $x_2 - x_3$ is δ, the measured displacement. Under the basic assumptions governing validity of this model, this displacement must be very small; in practice it is approximately 12 microinches. Direct control of k_2, k_3, F_1 and k_1 is possible and these can be selected to minimize the effect of variations in k_4, k_5 and k_6. The optimum results are obtained by making $k_3 \to \infty$ and $k_1 \to 0$; the equation for δ then becomes

$$\delta = \frac{F_a}{k_4 + (A + B)k_2} - \frac{BF_1}{k_4 + (A + B)k_2}, \tag{9}$$

where

$$B = \frac{k_4}{2k_5}\cdot \tag{10}$$

Eq. (9) is the best result that can be gotten by manipulation of the transducer springs only. Under these conditions δ will be a reliable indication of F_a if $k_2 \gg e_4$ and $A + B$ is either constant or close to unity. Since

$$A + B = 1 + \frac{k_4}{k_6} + \frac{k_4}{2k_5}, \tag{11}$$

it is necessary to assume that both k_4/k_6 and $k_4/2k_5$ are constant or (more reasonably) that they are both very much less than unity. Although k_4/k_6 may be small because of the thin layer between skin surface and artery, there is no reason to believe that the same is true for $k_4/2k_5$. The measurement would therefore be sensitive to variations in physiological parameters, as well as to variations in the loading force F_1.

The artery and flesh elasticities are nonlinear for large displacements and can therefore be controlled to some extent by external manipulation. Considering this, the result of reducing k_4 is examined. For $k_4 \to 0$, the term A [(9)] approaches unity and the term B in this equation approaches zero. The equation for δ now becomes

$$\delta = \frac{F_a}{k_2}\cdot \tag{12}$$

The measure of the arterial pressure under these conditions is provided by δ, which is directly proportional to the force due to arterial pressure acting on the center plate area.

The reduction of k_4 (the artery elasticity) can be accomplished by pressing the transducer against a superficial artery with sufficient loading force to flatten a portion of the artery (see Fig. 4). The tangential arterial wall tension due to k_4 should not affect the vertical force measured by the transducer. The partial compression of the artery required to produce the flattening is consistent with the assumption that the artery rests on a firm base, since the loading of the artery will cause recession and very little compression until the firm base is attained.

The force F_a is derived from arterial blood pressure acting over the surface area of the transducer center plate; *i.e.*, $F_a = PA$. The measured output δ will thus be strictly proportional to pressure only as long as the flattened area A remains constant.[4] This is best achieved by making the

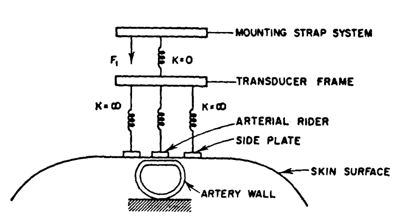


Fig. 4—The transducer artery system after the steps required to eliminate the unwanted parameters have been taken. Particularly important is the flattening of the arterial wall under the transducer.

[4] R. Stuart MacKay, "Fast, automatic ocular pressure measurement based on an exact theory," IRE Trans. on Medical Electronics, vol. ME-7, pp. 61–67; April, 1960.

area of the transducer center plate smaller than the flattened area of the artery so that it will always be entirely covered by flattened arterial wall.

The design of the direct force-measuring transducer is based on consideration of this mechanical model. Since the validity of the measurement is dependent on the skin surface and artery being flattened below the transducer, it is essential that the transducer spring constant k_2 be made much greater than the effective spring constant which normally determines arterial deflection. Measurements with an arterial distention gage have shown that the unrestrained motion of skin over the temporal artery is approximately 120 microinches. To make the transducer sufficiently rigid to flatten skin and artery and provide a true force measurement, k_2 is made about ten times the value for the artery-skin system so that the resulting deflection is on the order of 12 microinches.

An additional requirement for this technique is that the central measuring plate or arterial rider be small. Preferably it should be somewhat narrower than the artery, so that it can be easily positioned entirely on the artery. It should also be reasonably short, to minimize the difficulties of alignment with the artery. The resulting small area of the arterial rider limits the amount of force available from arterial blood pressure.

Although a great variety of techniques are available for force measurement, all known methods involve the sensing of a displacement caused by the force to be measured. In the blood pressure transducer, the extremely low displacement complicates the selection of the basic sensing element. A strain gage was chosen for use in the first transducer designs. The availability of semiconductor strain gages permits the use of low strain levels but provides sufficient electrical output signal.

Other design considerations involve the external and internal geometry of the transducer. Fig. 5 is a basic diagram of the strain gage transducer configuration. The side plates are integral with the transducer body, so as to maximize k_3; k_1 is made negligibly small and F_1 is constant because it is produced by a pneumatic loading system. Pressure in the air chamber is maintained at a constant value by a pressure regulator and the air box is forced down over the skin surface by straps or similar means. So long as the diaphragm (rubber membrane) lies tangent to the skin surface at all edges of the transducer, *i.e.*, when it is flattened against the skin, it can exert no tension normal to the skin and, therefore, cannot influence transducer loading. Therefore, the constant pressure acting over the area of the transducer provides the sole loading force. A change in strap tension may radically change the force exerted on the air box, but this will only result in a change in the flattened area of the diaphragm to provide the necessary reaction force.

The pressure in the air box is increased until sufficient artery flattening occurs as determined by observation of pulse amplitude. The amplitude increases with pressure until flattening is complete; then no further increase occurs as air box pressure rises. The amount of pressure required will depend on the over-all transducer surface area and the resistance of the underlying tissue.

A part of the transducer configuration given in Fig. 5 is not represented in the mathematical model of Fig. 3. It is the necessary gap between the arterial rider and the side plates. This gap should be made as small as possible, since there will be a tendency for skin to bulge into it, as shown in Fig. 5(a). The bottom surfaces of the arterial rider and the side plates should be at the same level, since deviations will cause a distortion of the skin surface [Fig. 6(b)]. Both these effects violate an important requirement of the design theory, namely, that the skin surface lie flat under the transducer. The effect of skin surface distortion is shown in Fig. 6(c). A component of force (F_T) resulting from skin tension is transmitted to the arterial rider; the magnitude of this force depends on the skin tension (T) and the angle of incidence of the skin surface at the edge of the arterial rider (θ):

$$F_T = 2T \sin \theta. \tag{13}$$

The effect of reducing the gap and making the surfaces level is to minimize θ, thus reducing F_T to the point where F_T/F_a is less than the desired accuracy of the transducer. In practice, the gap is made less than 2 mils and surface flatness is kept to within 0.1 mils (by lapping).

An essential part of all measuring systems is the method of calibration. The most convenient for this system would be an absolute calibration performed on the instrument

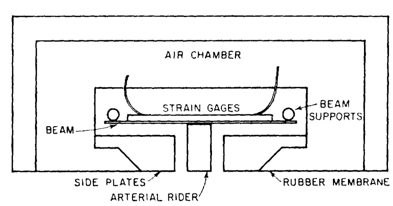


Fig. 5—Basic diagram of the strain gage approach to the transducer design.

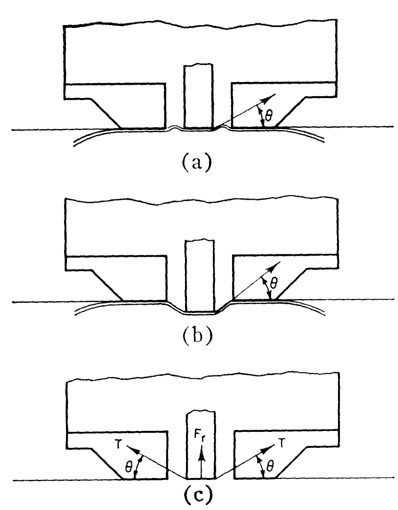


Fig. 6—The effect of rider alignment and gap length on the skin surface. (a) The effect of excessive gap between rider and side plates. (b) The effect of rider surface misalignment with side plate surface. (c) Forces introduced by skin surface deformations.

before it is used and remaining constant regardless of the subject or the location. This type of calibration appears feasible with a highly developed form of the high spring rate direct-force transducer. The requirements of the measuring system would be:

1) The artery surface below the transducer must be flattened.
2) The flattened area of the artery must cover the entire measuring surface (center plate) of the transducer.

Under these conditions, the transducer can be calibrated by means of an "artery simulator," which consists of an extensible rubber membrane that can be pressurized to a known pressure level. The transducer is mounted against the rubber membrane and the output signal is recorded as a function of applied pressure.

If condition 1) is satisfied but condition 2) is not because of uncertainty in transducer placement, the calibration must be performed when the transducer is in place on an artery. One method of doing this is to use the sphygmomanometer as a standard; the systolic and diastolic points are then obtained and (assuming linearity) can be used to calibrate the transducer. Unfortunately, this method of calibration is influenced by the accuracy of the sphygmomanometer. Since individual interpretations of the same readings on the sphygmomanometer may vary as much as 20 per cent, it is necessary to take a number of measurements and use the average value for the calibration.

If the transducer is located on a remote artery, such as the temporal artery, the value of blood pressure measured on the arm may not accurately reflect the pressure in the more remote artery. Also, instances have been reported where sphygmomanometer readings were consistently lower than direct intra-arterial measurements using the catheter. If the transducer is used on the radial artery, the sphygmomanometer cuff cannot be applied to the arm on which the gage is mounted, because occlusion of the brachial artery will stop circulation to the radial artery, making it impossible to obtain simultaneous readings. Simultaneous readings are required because of semi-periodic variations in blood pressure. These variations, which have been observed during transducer tests, are probably the result of respiration. They may be as large as 10 to 15 mm Hg.

Another proposed method of calibration consists of obtaining a zero point by upstream occlusion of the measured artery and obtaining the slope of the curve by changing the elevation of the measuring point with respect to the heart. This method has been tried with limited success. Occlusion of an artery does not necessarily result in zero (atmospheric) downstream pressure because of the availability of alternative paths. For example, occlusion of the radial artery does not cut off completely the blood flow to the vessel because of the alternative path through the volar arch from the ulnar artery. Even if the brachial artery were occluded by local pressure, alternative paths are still provided by anastomosis around the brachial artery. Large-scale occlusion, like that provided by a cuff around the arm, also blocks venous flow and prevents reduction of downstream pressure. This effect is demonstrated by records taken on the radial artery during sphygmomanometer readings on the same arm (see Fig. 7).

The change in elevation of the measuring point provides a known change in mean blood pressure determined by the pressure head of blood in the vertical distance between the two positions. As the measuring point is raised above the heart, the mean pressure is lowered. This lowering has been observed in experiments with the prototype transducer. The relationship appears to be approximately of the right magnitude; however, the change in position must be made rapidly to avoid vasomotor compensation.

If neither condition 1) nor condition 2) can be obtained with certainty a calibration can still be performed; however, the assumption of linearity may not be valid and the effects of physiological parameters as given by (9) may not be entirely eliminated. The quality of the measurement is determined by the degree of flattening obtained.

The direct force measuring system is considered to be one in which it is at least theoretically possible to achieve an absolute calibration. Measurements obtained by careful placement of the calibrated direct force transducer agreed with the readings of the sphygmomanometer used for comparison, within its limits of accuracy.

Transducers built from two experimental designs are shown in Figs. 8 and 9. The model shown in Fig. 8 used two strain gages (gage factor = 110) and a simply supported strain gage beam and had a rider area of 0.0156 square in (0.25×0.065 in). The signal output was 10.4 μv per mm Hg, with 4-v excitation to the strain gage bridge circuit. The model of Fig. 9, which is a refinement of the earlier design used four strain gages (gage factor = 110) and a double cantilevered strain gage beam; it had a rider area of 0.00375 square in (0.125×0.030 in). The signal output was 27.5 μv per mm Hg with 2.5-v excitation to the bridge.

The first model was highly sensitive to temperature variations; this resulted in an unstable base line in the measurements. However, the transducer demonstrated the ability to display the correct pulse waveforms (see Fig. 10) and to follow pronounced variations in blood pressure, like

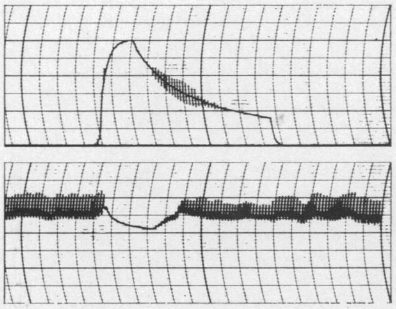
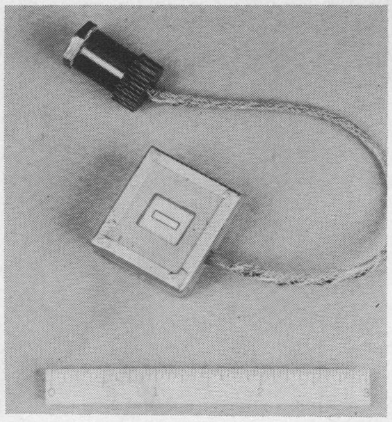
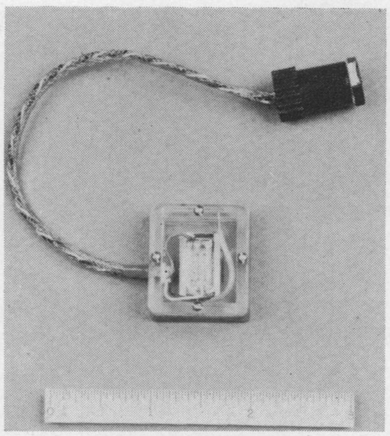


Fig. 7—Occlusion of the brachial artery during a sphygmomanometer determination of blood pressure. The transducer is located on the radial artery. The upper trace is the record of sphygmomanometer cuff pressure with Korotkov sounds superimposed. The lower trace is the transducer output.



(a)



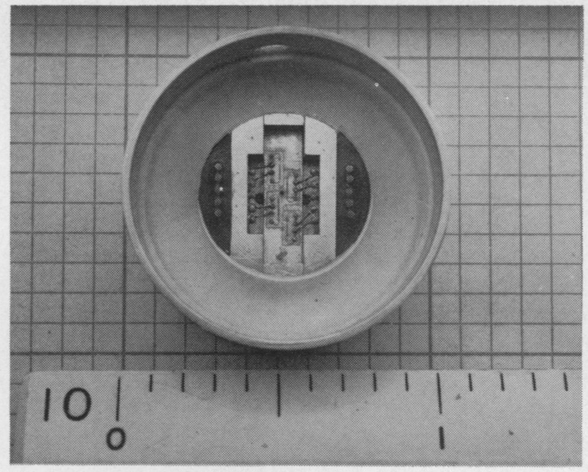
(b)

Fig. 8—The first experimental transducer. (a) Front view. (b) Rear view.

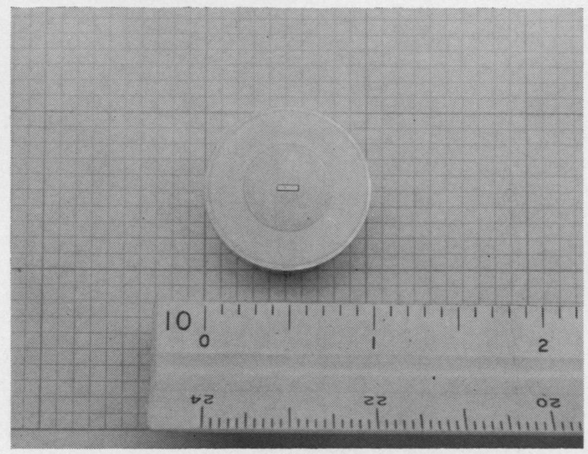
those indicated in Fig. 10(c). Here, the subject performed a Valsalva maneuver and the transducer recorded the blood pressure variations at the radial artery.

The construction details of the second model are given in Fig. 11. The use of four strain gages in this model greatly reduced the effect of temperature and made static calibration possible. A rubber membrane 0.010 in thick covering a pressurized cavity supplied a test pressure to the transducer for calibration. Although this apparatus is by no means a true model of the human artery, it satisfies the basic requirement that the transducer should respond to pressure changes to the exclusion of membrane characteristics. The results of this calibration are given in Fig. 12.

To obtain an estimate of the transducer's dynamic response, the rubber membrane apparatus was equipped with a Statham Model PM-60TC $\pm$ 5-pressure transducer to measure dynamic pressures acting on the membrane to load the blood pressure transducer. The pressure was increased from 0 to 200 mm Hg in $1/5$ sec, with a maximum rate of pressure rise of about 2000 mm Hg/sec. Both transducer outputs were fed to an X-Y oscilloscope to obtain the dynamic calibration curve shown in Fig. 13. The reference (Statham) transducer output is recorded on the X axis and the SRI transducer output on the Y axis.

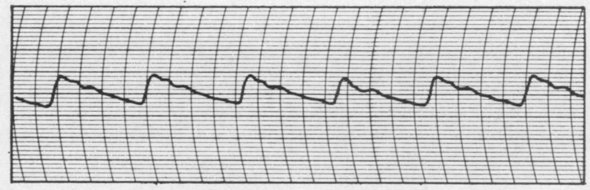


(a)



(b)

Fig. 9—The second model experimental transducer. (a) The use of four miniature strain gages on the cantilever beam. (b) Reduced rider size and gap.



(a)

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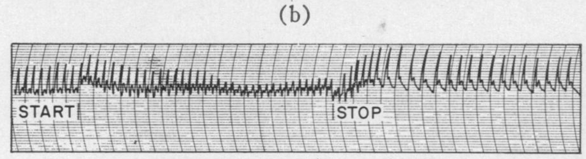

(c)

Fig. 10—Response of the first model blood pressure transducer. (a) Pulse waveform at the radial artery. (b) Pulse waveform at the temporal artery. (c) The response of a Valsalva maneuver as measured on the radial artery. The chart speed for (a) and (b) is 25 mm/sec and for (c) is 5 mm/sec.

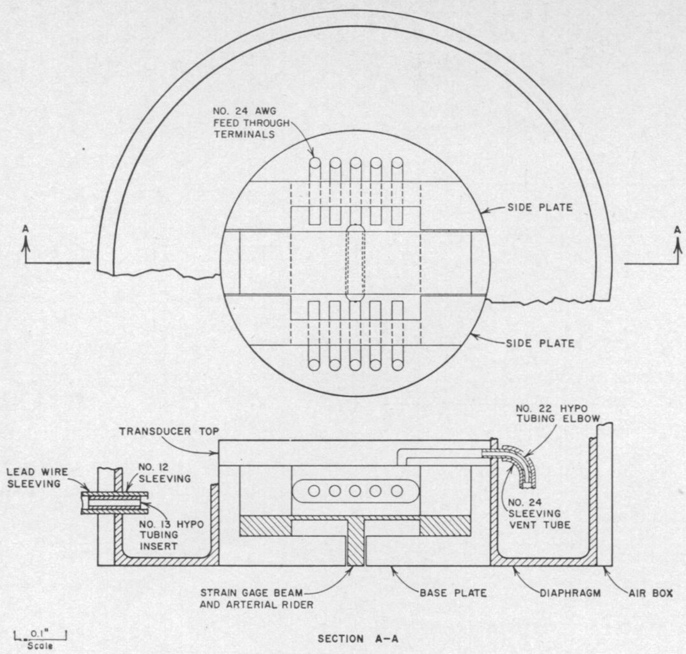

Fig. 11—Construction drawing of the improved transducer. The beam, beam supports and rider are one piece. The four strain gages are mounted on top of the beam and are positioned so as to receive maximum average values of strain, with two gages in compression and two gages in tension. Since the rider gap is likely to be sealed when the transducer is in contact with the skin and the transducer case must be sealed against tne box pressure, a vent tube is provided from the transducer interior to the atmosphere. This provides the local ambient pressure as the reference. The lead wires and the air box air supply are brought in at the same point which is the air tube containing the wires. The diaphragm is molded from RTV-11 silicone rubber. The construction material for air box, transducer body and components is 24ST aluminum alloy.

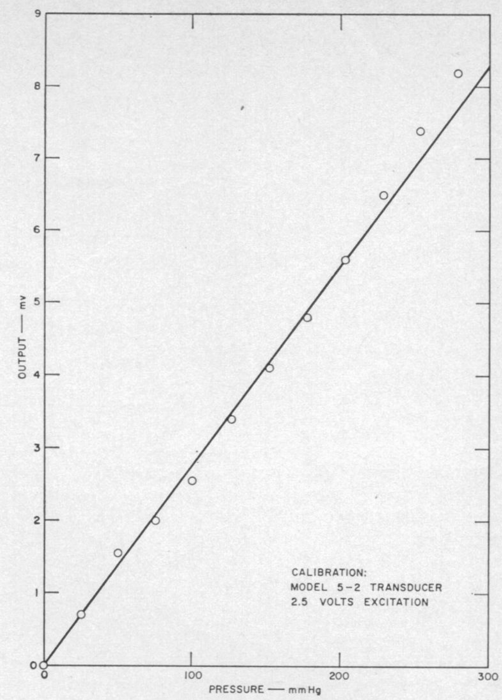

Fig. 12—Transducer static calibration curve.

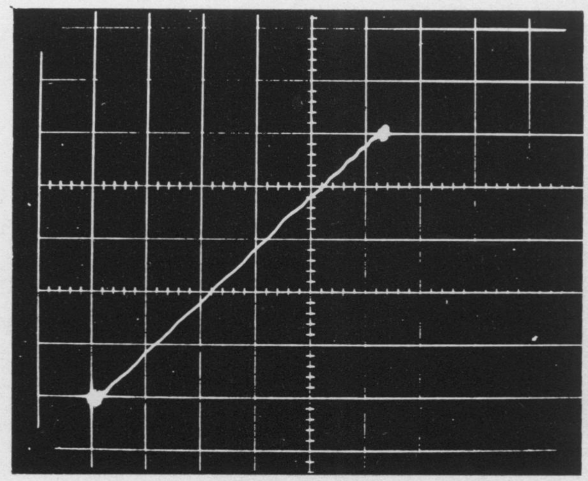

Fig. 13—Dynamic response curve.

A series of tests was performed to compare the blood pressure transducer readings with systolic and diastolic readings taken with a sphygmomanometer. The transducer was carefully positioned on the left radial artery to obtain the maximum pulse wave amplitude output and the air box pressure was increased until constant pulse wave amplitude was obtained (usually from 40 to 100 mm Hg box pressure was required—the value is not critical). A standard sphygmomanometer and an electronic stethoscope were used to record systolic and diastolic pressure in the right arm. These readings were taken simultaneously and recorded on a two-channel recorder. A typical recording is shown in Fig. 14. The upper trace shows cuff pressure with Korotkov sounds superimposed. The lower trace is the precalibrated blood pressure transducer output. The results of these tests are given in Table I on the following page.

The differences between the two measurement methods are attributed to the following factors: 1) error in sphygmomanometer readings and in interpretation of stethoscope recordings, especially in the diastolic measurement, 2) difference in blood pressure between the two arms, 3) error in positioning of the transducer and 4) arterial rider too wide with respect to artery.

During these tests, the transducer was found to be sensitive to position. It was necessary to hold the transducer in position rather than to rely on a strap.

The technique of searching for the highest pulse output by changing position and box pressure proved to be practical for the original positioning of the transducer.

Some preliminary comparisons with direct intra-arterial measurements have been attempted, using animals and applying the transducer to exposed arteries.

Indirect pressure readings were obtained from the exposed femoral artery. Direct measurements were made with a Statham pressure transducer connected to a cannula in the contralateral femoral artery. The femoral artery of the cat appeared to be approximately 0.050 in in ex-

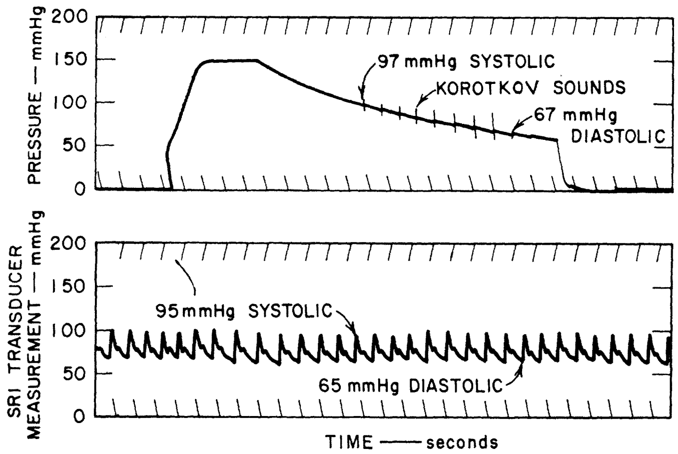

Fig. 14—Typical comparison to sphygmomanometer. The upper trace is the cuff pressure; an electronic stethoscope is used to pick up the Korotkov sounds, which are superimposed electrically onto the pressure trace. On this record the systolic and diastolic readings and their time of occurrence can be determined. The lower trace is the transducer output.

TABLE I

Comparison of Transducer to Sphygmomanometer

Subject	Age (years.)	Height	Weight (pounds.)	Sphygmomanometer		SRI Transducer		Per Cent Difference	
				Systolic (mm Hg)	Diastolic (mm Hg)	Systolic (mm Hg)	Diastolic (mm Hg)	Systolic	Diastolic
A	23	6'2"	170	97	67	95	65	−2.1	−3.0
B	30	5'9"	164	125	90	115	88	−8.0	−2.2
C	31	6'1"	155	125	77	125	68	0.0	−11.7
D	27	5'8"	150	110	70	100	75	−9.1	+7.1
E	35			120	87	110	85	−8.3	−2.3
F	28	5'9"	155	110	80	105	65	−4.5	−18.8
G	39	5'11"	140	115	82	112	85	−2.6	+3.5
H	41	5'11"	170	112	85	107	82	−4.5	−3.5
I	25	6'0"	175	111	80	102	75	−8.2	−18.8
J	29	5'6"	125	108	80	100	83	−1.9	+3.8
K	41	5'7"	140	117	67	100	67	−14.5	0.0

ternal diameter; therefore, it was difficult to position the tranducer. The results follow.

Pressure in Femoral Artery of a Cat

(mm Hg)

	Cannula (Right Femoral)	SRI Transducer (Left Femoral)
Systolic	120	98
Diastolic	105	77
Mean	110	84

Because of the discrepancies between the two sets of readings, additional measurements were made using larger arteries. Readings were obtained from the carotid artery and abdominal aorta of the cat, both of which are larger than the femoral artery. Summarized results follow.

Artery	Cannula (Right Femoral)	SRI Transducer	Estimated Distance from Cannulation Point
Carotid	120/99-106	125/95-105	12 inches
Abdominal Aorta	120/95-103	125/90-102	4 inches

Further testing, both on animals and humans, is being planned.

The positioning of the transducer has been shown to be a critical factor in obtaining readings corresponding to a fixed calibration. Therefore, it is necessary that the design of a blood pressure measuring system includes a means of positioning the transducer accurately. When it is positioned manually, observation of the pulse amplitude is used as a guide, the maximum reading occurring when the transducer is directly over the artery.

If the transducer is to be used continuously for long periods, a means of mounting must be provided. It has been found impossible to position the transducer manually before the mounting device is applied, since the installation procedure invariably moves the transducer. Manual positioning after mounting is very difficult because of the force with which the transducer is held against the body. A satisfactory transducer mounting device should include a means of lifting off and moving the transducer without disturbing the mounting system. (The lifting off is desirable to allow the zero level of the transducer to be checked.) With such a mounting device, the transducer could be positioned after being mounted on the body, using pulse amplitude as a criterion.

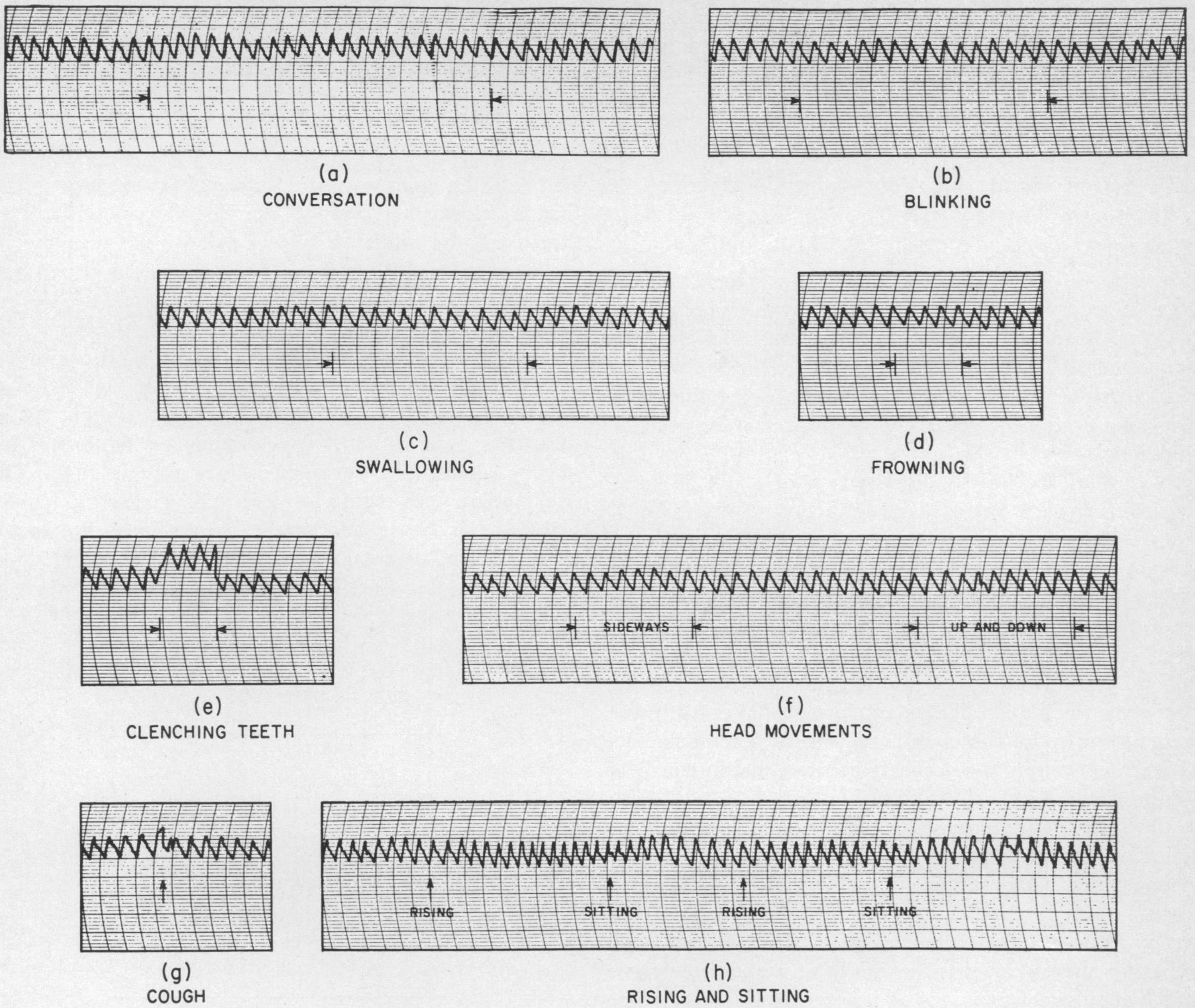

Fig. 15—The effect of subject activity on the pressure recordings. The transducer (the one shown in Fig. 8) is mounted on the superficial temporal artery's frontal branch, using a head band. The activities selected involved facial movements which were considered likely to interfere with the measurements.

A number of simple experiments were performed to test the transducer's sensitivity to artifacts. For these tests the temporal artery was used as a measuring point. The results are shown in Fig. 15. They were obtained with the first transducer design; no significant change was noted in the newer design. Fig. 15(a) shows the output of the transducer during normal conversation. During the period indicated on the record the subject was wearing the pressure transducer while talking in a normal tone of voice. Fig. 15(b) shows the transducer response while the subject is blinking his eyes rapidly. In Fig. 15(c) the subject is swallowing and in Fig. 15(d) he is frowning at the points marked. Figure 15(e) shows the effect of clenching teeth. Figure 15(f) shows the results of head movements. In the first period, as marked, the subject was turning his head approximately plus or minus 45° sideways at a rate faster than one turn/sec. In the second marked period the subject was moving his head up and down approximately plus or minus 45° at the same rate. Figure 15(g) shows the effect of a cough. In Fig. 15(h) the subject is rising and sitting down while wearing the transducer.

At the present stage of development the results obtained with the experimental transducer are consistent with the assumptions upon which the mechanical model is based. The transducer has demonstrated usefulness as a precalibrated absolute measurement of arterial blood pressure when applied to large superficial arteries. Design improvements are being considered which will reduce the arterial rider size, improve positioning techniques and increase the measuring system spring constant k_2 to more closely approach the theoretical condition of equivalence of transducer reading and arterial pressure. Additional work is planned which will incorporate these improvements in new transducer designs and additional studies are in preparation which will provide a comparison with direct intra-arterial measurements in animals.