Studies Made by Simulating Systole at Necropsy

XII. Estimation of the Initial Cardiac Forces from the Ballistocardiogram

By Isaac Starr, M.D.

In a preceding communication I have suggested that we are now in a position to look more closely at the higher dynamic aspects of cardiac function in our patients. From an analogy with common experience in automobiles, it is reasonable to expect that the first sign of myocardial weakness will manifest itself in diminished ability to accelerate the blood, that is, in diminished cardiac forces. In this study we aim to establish a clinical method of detecting abnormalities of the initial cardiac forces by means of the ballistocardiogram.

In the studies based on our cadaver experiments we have previously concerned ourselves with clinical methods of measuring the heart’s output and its work, aspects of cardiac function that could be measured, when systole was simulated, with an accuracy that cannot be attained during life. To complete this line of attack we needed a similar study of the cardiac forces, which, as far as I am aware, have never been measured during life. Such a study was of especial interest to me since the ballistocardiogram, as used in this laboratory, is a recorder of force.

From the first it has been realized that the ballistocardiogram does not give a true record of the cardiac forces throughout systole. This is because, as systole progresses, secondary forces arise due to changes in direction and velocity of blood in the vessels. These secondary forces either compete with or reinforce the forces arising directly from the heart; so the ballistocardiogram secured during the later part of systole is the resultant of many forces. But one would expect that during a brief interval between the onset of ejection and the arrival of important amounts of blood at the aortic arch and the curve of the pulmonary artery, the ballistocardiogram would be related solely to the cardiac forces occurring simultaneously. This paper is concerned with developing a quantitative method of estimating these initial cardiac forces from certain aspects of that record.

Several considerations encouraged us to make this study. First, our theoretical views of cardiac dynamics indicate that the forces concerned with the initial acceleration of the ejected blood should have great clinical importance in judging the strength of the heart; they produce what one might call the jerk of the contraction. Second, experience suggested that such a method would provide information of great clinical interest, for the I wave of the ballistocardiogram, occurring simultaneously with these forces, varies greatly from one patient to another, tending to be low or absent in many patients with heart disease. Finally, if the cardiac forces could be successfully estimated in our experiments, we would have a test of our method against known amounts—the ideal of all quantitative methodology—and we had every reason to expect such a comparison would at once disclose errors in our present conceptions regarding the genesis of the ballistocardiogram and permit us to seek for means of correcting them. Thus we could hope to define more exactly the influence of noncardiac factors, such as the height of the blood pressure, the distensibility of the vessels, and the size and structure of body tissues, on the ballistocardiogram, for there was already reason to believe that these might have some effect on the record.

The consummation of such a plan required the development of a quantitative method of
INITIAL CARDIAC FORCES IN THE BALLISTOCARDIOGRAM

estimating the initial "cardiac" force applied in each of our 76 experiments. The "cardiac" forces could then be compared by statistical analysis with the forces registered by that part of the ballistocardiogram that was recorded simultaneously, and with other features of this record.

Material and Methods

The technic of the cadaver experiments has already been described. A high-frequency table ballistocardiograph was used and displacement was recorded, giving what is properly called a "force ballistocardiogram."

The clinical and autopsy findings of 6 of the 7 subjects used in this study have already been recorded. The other, a woman aged 50, height 159 cm., weight 41.5 Kg., had died from advanced scleroderma. We were thus provided with a unique opportunity to discover whether marked change in the physical properties of the tissues would have a noteworthy effect on the ballistocardiogram. The other 6 subjects were normal in this respect, and they exhibited considerable differences in body size and habitus.

Five subjects were perfused with blood and 2 with water, providing a total of 76 simulated systoles for study and analysis. The remainder of our data secured on subjects perfused with water we disqualified for 1 of 2 reasons: either these data fell into the period when we were having technical difficulties getting a proper I wave on the recorded ballistocardiograms, or they were secured in early experiments in which the speed of the moving film was too slow to permit measurement of the slopes as accurate as that which could be secured in later experiments in which the film moved twice as fast.

Of the data obtained from the 7 subjects studied we omitted those secured in only 1 systole, E.S. no. 11, because the time record showed that the film had slipped while systole was being simulated, and the slopes could not be accurately measured.

Measurement of the Record

Both the measurements and calculations are most easily explained by examples. Thus the record of curve 10 of subject E.S. has been redrawn in figure 1. The first step was to draw a line tangent to the ejection curve 0.08 sec. after the onset of ejection, and measure the angle (θ) of this constructed line with the horizontal.

After placing the base line of the ballistocardiogram the dimensions of the I wave were measured, its maximum depth being recorded to the nearest millimeter. A line was then drawn to coincide with

![Diagram](http://circ.ahajournals.org/)

**Fig. 1** Top. A scale drawing of the record secured on subject E.S. curve 10 to show the measurements made on all records.

**Fig. 2** Bottom. Diagram of the aortic cannula with the fluid in it divided into 3 parts. The movement in two of these parts makes a contribution to the forces when systole is simulated. During the experiment the terminal limb of this cannula lies in the left ventricle's outflow tract and its tip is tied into the mouth of the aorta.

the initial part of the H-I segment, and its angle with the horizontal was measured. These measurements were converted to absolute values by the corresponding calibrations, for example in curve 10 of E.S. we found

angle H-I = 70°\[\tan H-I = 2.75\]

Calibration of the ballistocardiogram: 280 Gm. = 12.65 mm.

Time calibration: 1 mm. = 0.02 sec.

To bring this to our standard calibration, 280 Gm. = 10 mm., we correct the slope as follows:

2.75 × \frac{10}{12.65} × \frac{1}{0.02} = 109 mm. per sec., \(a\)

which is the slope expressed in the relative units we routinely use.
An estimate of the errors involved in measuring the slopes of our records was sought by remeasuring a sample of 22 records after the lapse of 2 months had removed all recollections of the value first secured. The 2 measurements of the same slope agreed within 1° in 82 per cent, within 3° in 91 per cent. The 2 remaining II-I segments were flat and rounded and it was more difficult to draw with accuracy the lines tangent to the initial part of the curves. However the angles measured were so small that their tangents differed very little, and the larger error of measurement did not introduce a larger error into the result.

Calculation of the “Initial Cardiac Forces”

To calculate the forces employed in our various simulations of systole, one must consider the fluid in each of the 2 glass cannulas, one tied into the mouth of the aorta, the other into that of the pulmonary artery, to be divided into 3 parts, as is shown for the aortic cannula in figure 2. The movement of fluid in part A (fig. 2), as well as that of the contiguous fluid upstream in the glass tubing connecting the cannulas and syringes, and the movement of fluid within the syringes themselves, does not concern us, as all this fluid was moving at right angles with the vector in which the ballistocardiograph was recording the forces.

Forces of the Terminal Limb

The calculation of the forces concerned with the movement of fluid in the terminal limbs of the 2 cannulas (fig. 2C) is best considered first because it presents no difficulties. In the record of E.S. 10 we found

\[ \text{angle } \theta = 64^° \text{ and } \tan \theta = 2.05 \]

Calibration of the flow record: 1 mm. = 2.29 ml.
Calibration of the time record: 1 mm. = 0.02 sec.
The flow velocity at 0.08 sec. after start of ejection is therefore

\[ 2.05 \times \frac{2.29}{0.02} = 234 \text{ ml/sec}. \] (b)

And since there was no velocity at the start of the ejection, the average acceleration of flow during the first 0.08 sec. is

\[ \frac{234}{0.08} = 2930 \text{ ml/sec}^2. \] (c)

This fluid was being delivered through glass cannulas the diameter of which was 2 cm. Therefore the average linear acceleration of the injected fluid in both cannulas in 0.08 sec. in this experiment was

\[ \frac{2930}{1^2 \times 3.14} = 933 \text{ cm/sec}^2. \] (d)

The volume of the terminal limb of the aortic cannula, the limb which passed through the left ventricle and was tied into the root of the aorta, was 20.4 ml. That of the corresponding limb of the pulmonary artery cannula was 29 ml.

The total mass of blood concerned was therefore 49.4 Gm., the specific gravity of blood being close to 1.

The initial force concerned with motion of blood in the terminal limbs (fig. 2C) was therefore

\[ 933 \times 49.4 = 46,100 \text{ dynes approximately } (r) \]

Forces of the Turn

The movement of the fluid as it changes direction in the cannulas, at B of figure 2, is of great importance. Mr. George Peirce suggested that it could be calculated from our data as follows. The basic concept stems from the general formula

\[ \frac{\text{volume}}{\text{time}} \times \frac{\text{distance}}{\text{time}} \times \frac{\text{mass}}{\text{volume}} = \text{force} \] (f)

For example from the record of E.S. 10 we found that at 0.08 sec. after the start of ejection the flow velocity is 234 ml/sec., since the radius of the tube is 1 cm, the linear velocity at 0.08 sec. is

\[ \frac{234}{1^2 \times 3.14} = 74.5 \text{ cm/sec}. \] (g)

Since the density of blood is approximately 1, the third term of equation (f) can be neglected and we have

\[ 234 \times 74.5 = 17,440 \text{ dynes} \] (h)

Since the force at the start of ejection was zero the average turning force over the period of 0.08 sec. is approximately

\[ \frac{17,440}{2} = 8,720 \text{ dynes} \] (i)

The turning force in both cannulas is twice this amount.

I must admit that I was slow to see the rationale behind this simple calculation of turning force, which takes no account of the characteristics of the turn, whether sharp or gradual. Accordingly, with the help of Dr. Askovitz I recalculated the forces of turning by a much more elaborate method taking account of the radius of the turn. But the result I secured was the same as that given above; so I was reconciled. I had failed to grasp the fact that if the turn is sharp the acceleration is great but the volume turning at any instant is small. If the turn is gradual the reverse is true. So the total turning force is independent of the radius of the curve.

The average forces applied during the initial 0.08 sec. of the simulation of systole in E.S. 10 was therefore the sum of the 2 groups of forces calculated above:

\[ 46,100 + 17,440 = 65,540 \text{ dynes} \] (j)

The initial forces applied in all the other systoles were calculated similarly, and the method should
give a good approximation of the average forces applied in our experiments during the first 0.08 sec. of ejection.

Forces Recorded by the I Wave of the Ballistocardiogram

By measurements on the photographic record we find in curve 10 of subject E.S.:
Distance from I wave tip to base line = 5 mm.
The calibration of the ballistocardiogram showed 12.65 mm. = 280 Gm. Therefore the force recorded at the I wave tip is:
5 mm. = 112 Gm.
= 112 × 980 = 109,760 dynes (k)
The I wave is so nearly triangular that it can properly be assumed to be exactly so. Therefore:
I wave force = ½ I wave depth × I wave duration (l)
and the average force during the I wave:
\[
\frac{I \text{ wave depth} \times I \text{ wave duration}}{2 \times I \text{ wave duration}} = \frac{109,760}{2} = 54,880 \text{ dynes} (m)
\]
The same calculation was made in each of the other experiments.

RESULTS AND DISCUSSION

As far as I am aware, this paper marks the first attempt to standardize a method of detecting the cardiac forces by means of a comparison with known forces. But it should be noted that I felt competent to do satisfactory quantitative work only where the situation is at its simplest, at the very beginning of ejection, when, for a brief period, the relation between cardiac forces and ballistocardiogram is, in theory, that of Newton’s third law of motion. The primary aim of this study was to see how nearly this ideal situation obtained in our experiments and presumably would obtain during life. It was, therefore, necessary to set a time limit and we chose the first 0.08 sec. of ejection because this is the usual duration of the I wave in ballistocardiograms, which facilitates the comparison of force applied with force recorded.

Relation between Cardiac Force Applied in 0.08 sec. and Force Recorded in I Wave. Figure 3 shows the data secured in each of the 7 subjects, and in the whole group. The regression of all 76 experiments is shown in relation to the line of perfect agreement in

![Figure 3. Relation of the cardiac force applied in the first 0.08 sec. of ejection to the force represented by the I wave of the ballistocardiogram. Subjects E.S. and E.I. were perfused with water, the others with blood. The solid lines are the calculated regressions for each subject. At the bottom the regression of all the data, 76 simulations of systole, is shown as a solid line for comparison with the line of perfect agreement (broken line). The correlation coefficients and standard deviations are in table 1.](http://circ.ahajournals.org/)

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PULSE WAVE VELOCITY M/SEC

Fig. 4. Three ballistocardiograms which accompanied the same initial cardiac force but under conditions of greatly different pulse wave velocity. All from subject E.I., the curves, from left to right, are nos. 10, 11, and 9. The initial cardiac force was calculated to be 160,400, 159,600, and 160,400 dynes respectively in each. The corresponding pulse wave velocities are given under each of the records. The stroke volumes were 83, 63, and 55 ml. respectively; the femoral blood pressures 80/44, 203/106, and 287/153 mm. Hg; the duration of ejection 0.31, 0.22, and 0.29 sec. Note the diminution of the depth and breadth of the I wave as P.W.V. increases, but the slope of the H-I segment remains the same.

The lowest part of figure 3. The agreement between the 2 slopes is extremely impressive for the 2 lines are almost exactly parallel, that representing the estimates being somewhat lower than the actual forces applied. The regression equation is

\[
\text{Average initial force applied (dynes) } = 30,500 + 1.02 \times (\text{Average force of I wave, dynes})
\]

(143)°

The standard deviation about the regression is 32,200 dynes and the correlation coefficient 0.79 whereas a value 0.29 is significant for \( p = 0.01 \). So there is no doubt that the relationship is a very strong one. This part of the ballistocardiogram is obviously in large measure determined by the cardiac forces acting simultaneously.

It is to be noted, however, that the recorded forces are lower than those applied, and one would expect this for several reasons. First, the glass cannulas lying in the outflow tracts of the 2 ventricles do not point directly headward but at an angle of about 30° from the longitudinal axis of the body. The recorded forces would be somewhat smaller than those applied for this reason alone. Second, our limit of 0.08 sec. does not exclude all the opposing forces due to blood movement in the aorta, as will be demonstrated later. Finally there is every reason to expect that, in passing through material such as the body tissues, mechanical energy would in part be converted into heat and so fail to appear in our force record. The difference, therefore, between the recorded and applied forces is readily accounted for; I was surprised that it is as small as it turned out to be.

Factors in the Scatter of the Data. Turning one’s attention to the scatter shown in figure 3, one must first recall that I-wave depth was measured only to the nearest millimeter, and since these waves ranged from 0 to 14.9 mm. in depth, the error of measurement from this source alone is considerable. Therefore it seems evident that in the data of subjects J.W., E.S., R.R., and M.L., the scatter about the regression is so small that the known errors of measurement are ample to account for most of it. But in the data of H.Z. and E.I. the scatter is very much greater. Searching for a reason for this difference, we discovered 3 experiments on subject E.I. in which the forces applied were almost identical but the I-wave depth varied 3-fold. The ballistocardiograms of these 3 experiments have been redrawn (fig. 4) to illustrate the differences among them. When the I wave was small the pulse wave velocity was several times that present when the I wave was larger. This suggests an explanation that can be tested in our data.

Do Differences in Vascular Elasticity Influence the Size of the I Wave? Such differences can be judged by the changes in pulse wave velocity which accompany them, so the answer to this question was sought by arranging the data into pairs, and 11 such pairs

*Regression equations of general interest have been given numbers in series with those obtained before from the data secured in these cadaver experiments. Other equations have been given letters.

*The apparent exception to this statement is due to a misprint. In subject R.R., Curve 9, the I-wave size, given as 17.5 mm. in table 3 of a previous publication, was actually 7.5 mm.
are shown in Table 1. In each pair the subject is the same, the force applied very nearly the same, the flow velocity of injected fluid at 0.08 sec. after the onset of ejection is very similar, but the pulse wave velocities and I-wave sizes differ widely. By concerning ourselves with differences between the members of each pair we eliminate the major effect of the forces on the I wave and can hope to uncover minor effects such as might be caused by differences in vascular elasticity.

If the vascular elasticity has an effect on the amplitude of the I wave, the differences in pulse wave velocity shown in Table 1 should be related to corresponding differences in I-wave size. The correlation between these 2 columns of differences in Table 1 has a coefficient of 0.83, a level of 0.60 being significant for \( p = 0.05. \) Therefore these results cannot be explained by chance and indicate that a noncardiac factor, the vascular elasticity, does influence the size of the I wave.

By another arrangement of the data of Table 1 we can learn more about this relationship. For each of the 11 pairs we compute a difference in I-wave force per unit change of pulse wave velocity, and then ask ourselves whether these quotients bear a relation to the average flow ejected into the great vessels. This correlation is significant, \( r = 0.66, \) and the regression crosses the zero line at a flow velocity of 234 ml./sec. which is only a little below the mean flow velocity 261 ml./sec. used in these experiments. This means that differences in vascular elasticity have little or no effect on the I wave when the cardiac ejection velocity is small; but abnormally stiff vessels constitute a possible source of error in interpreting cardiac function from the ballistocardiogram when the stroke volume is large and flow velocity correspondingly high.

Therefore, while our results indicate that a factor other than the initial cardiac force may influence the size of the I wave, in com-
slope of its record represents the first derivative of force, which is known to physicists as the jerk. As far as I am aware, the cardiac jerk has never been studied from the physiologic viewpoint and its quantitation presents many difficulties; our data are not exact enough to permit us to calculate the jerk with enough accuracy to make the effort worth while. So we cannot directly compare the slope of the H-I segment in each of our records with the aspect of "cardiac" performance which we believe to have brought it about. But our data do permit us to compare the H-I slope with the initial cardiac forces themselves and this relationship seemed well worthy of careful study.

Estimations of Initial Cardiac Force from Slope of the H-I Segment. At first thought an aspect of the record which represents the derivative of the recorded forces does not seem like a good starting point for a simple arithmetical method of estimating the forces applied; ordinarily one needs the calculus to define such a relationship accurately. But there is another way of looking at our problem. The deflection of the ballistocardiogram from the base line at any instant represents the force at that instant, so let us take as our starting point a measurement of I-wave depth made, not at the wave tip, but at a point on the H-I segment a very short time \( T \) after the beginning of ejection. Let us use the same \( T \) for this measurement in each experiment.

Then

\[
\text{Depth of the I wave at } T = (\text{slope of the H-I segment}) \times T \quad (n)
\]

We plan to explore our data by regression technics and could use either side of this equation as the starting point of a method of estimating cardiac force. When one uses the right side, one may omit the time factor \( T \) because it is a constant throughout all the experiments and so its value would be automatically taken care of in the calculation of the regression equation. Therefore, in this particular case, measurements of the slope of the H-I segment make as good a starting point for a method of estimating cardiac forces, as
would measurements of I-wave depth made at a predetermined time.

For several other reasons, the H-I slope makes a much better starting point than does a measurement of I-wave depth made at a given time. Any such measurement must be made from the record’s base line so the error of placing this base line is included in the unavoidable errors of measurement. In some records the base line is difficult to place with certainty and this is far more often true in records secured in the clinic than in those of our cadaver experiments. By using the slope of the H-I segment the error involved in placing the base line is altogether avoided. The considerable error inherent in identifying the exact time of onset of ejection, if one must place the time of the observation, is avoided also.

Figures 5 and 6 show the relationship between the initial forces applied and the slopes of the H-I segments of the resulting ballistocardiograms. The use of the H-I slope instead of the area of the I wave improves the correlation in every subject and also in the group as a whole (table 2). The improvement is especially conspicuous in the data from subjects E.I. and H.Z., on which the method based on I-wave area did very poorly. In each of the 7 subjects shown in figures 5 and 6 the scatter is now so small that it seems likely that the unavoidable errors of measurement account for it. The correlation between change of pulse wave velocity and change of H-I slope in the pairs illustrated in table 1 is not significant; so, if any error due to differences in vascular elasticity remains, we are unable to demonstrate it.

In figure 7 the regressions found in the data of each subject have been placed for comparison with that of the whole group of 76 experiments. This regression equation of the whole, when, in the vertical dimension of the record, 10 mm. = 280 Gms., is as follows:

Initial cardiac force applied, dynes =
18,000 + 603 (slope of the H-I segment, mm./sec.)

(144)

The standard deviation around this regression is 28,100 dynes and the correlation coefficient is 0.85 whereas a value of 0.29 is significant for $p = 0.01$. The regressions of the individual subjects are very nearly parallel with the exception of that of M.L., which, pulled off by 1 divergent point, is not signifi-
cantly different from the others. So the slopes of the regressions seem as equal to one another as we have a right to expect considering the methods at our disposal.

Possible Factors in the Scatter. Obviously a large part of the scatter of the group as a whole is due to differences between one subject and another. Before seeking an explanation for this, we should first point out a difficulty inherent in our experiments. The position of the glass cannulas tied into the mouths of the great vessels in each subject is very critical. When, in certain early experiments, these cannulas were passed too far into the great vessels, simulation of systole produced no I wave at all. In later experiments, we made every effort to tie each cannula exactly in the mouth of its vessel, but it seems likely that there was enough difference of technic from subject to subject to have some effect on the I wave's size. Such a technical error would manifest itself in our results as a difference between one subject and another. For this reason the differences found between subjects in our experiments may well be larger than those to be expected during life.

The fact that some of the differences found may be due to technical difficulties does not relieve one of the duty to explore the data for factors related to body size and quality, which might account for the remaining differences between one subject and another. Subject E.I. died of advanced scleroderma and the physical properties of much of her skin differed greatly from those of our other subjects. The results secured on this subject were not conspicuously different from those of our other subjects but the omission of her data does slightly improve the scatter of the remainder, as is shown in table 3. Whether the subjects were in rigor mortis or not made no demonstrable difference to our results.

Looking for effects of differences in body size, we first investigated the correlation between the forces applied and 3 products: H-I slope and body weight, body surface area, and height. The results are given in table 3. We made no significant progress in reducing the scatter by this method, a result which might have been anticipated by noting how parallel were the regressions in figure 6.

Another type of adjustment for body size can be made by calculating multiple regres-

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**Table 2.—Comparison of Two Methods of Estimating Initial Cardiac Force from the Ballistocardiogram**

<table>
<thead>
<tr>
<th>Subjects</th>
<th>No. of exps.</th>
<th>Estimations from I wave $\sigma_r$, dynes</th>
<th>$r$</th>
<th>Estimations from H-I slope $\sigma_r$, dynes</th>
<th>$r$</th>
</tr>
</thead>
<tbody>
<tr>
<td>J.W.</td>
<td>12</td>
<td>6,100</td>
<td>0.96</td>
<td>5,500</td>
<td>0.96</td>
</tr>
<tr>
<td>M.M.</td>
<td>7</td>
<td>15,900</td>
<td>0.83</td>
<td>8,100</td>
<td>0.96</td>
</tr>
<tr>
<td>R.R.</td>
<td>15</td>
<td>20,300</td>
<td>0.93</td>
<td>21,900</td>
<td>0.92</td>
</tr>
<tr>
<td>H.Z.</td>
<td>12</td>
<td>30,300</td>
<td>0.67</td>
<td>21,500</td>
<td>0.85</td>
</tr>
<tr>
<td>M.L.</td>
<td>8</td>
<td>26,100</td>
<td>0.79</td>
<td>18,100</td>
<td>0.91</td>
</tr>
<tr>
<td>E.I.</td>
<td>11</td>
<td>29,500</td>
<td>0.43</td>
<td>18,000</td>
<td>0.84</td>
</tr>
<tr>
<td>E.S.</td>
<td>11</td>
<td>25,200</td>
<td>0.90</td>
<td>15,900</td>
<td>0.96</td>
</tr>
<tr>
<td>All</td>
<td>76</td>
<td>32,200</td>
<td>0.79</td>
<td>28,100</td>
<td>0.85</td>
</tr>
</tbody>
</table>

Standard deviation of cardiac forces used around their own means 55,900.
sion equations which employ the H-I slope and the subject's height or body surface area to estimate the cardiac force applied. These were calculated by Dr. Schild and are as follows:

Initial cardiac force applied, dynes =

\[-164,000 + 580 (\text{slope of H-I segment, mm./sec.}) + 1100 \text{ height, cm.}\]  \hspace{1cm} (145)

Initial cardiac force applied, dynes =

\[-57,300 + 585 (\text{slope of H-I segment, mm./sec.}) + 44,000 \text{ body surface area, M.}^2\]  \hspace{1cm} (146)

The standard deviations about these regressions have been given for comparison with the other results in table 3. The gain in accuracy secured by correcting for differences in body size seems too small to be worth while at the present stage of our knowledge.

Despite our inability to improve the method further at this time, it seems obvious that the slope of the H-I segment affords a rough clinical method of estimating cardiac forces with an accuracy more than sufficient to permit one to take the first quantitative step; that is, to divide the field into three parts; normal, above normal, and below normal; and to identify the position of any subject in this field with a chance of error reasonably small. When interested in changes in single subjects we should do much better than this. And one should point out that this method works well despite large differences in other physiologic functions. Thus in the data from the 76 simulated systoles analyzed for this study the stroke volume varied from 83 to 18 ml., the blood pressure from 282/139 to 42/29 mm. Hg, great differences in the elasticity of the great vessels are indicated by pulse wave velocities varying from 13.0 to 3.1 M./sec., the cross section area of the aortas varied from 10.8 to 2.5 cm.\(^2\), the viscosity of the injected fluid varied from that of blood to that of water, 2 subjects had very sclerotic aortas, and in 3 the aortas were altogether normal; and these profound physiologic and anatomic differences had little if any effect on the relationship studied.

Relation of New Data to Previous Findings.

Better estimation of the forces inherent in our simulations of systole has thrown new light on one of our previous findings. That the amplitude of ballistocardiograms is closely related to the acceleration of flow of the ejected blood was discovered early in these studies, and we naturally concluded that this record was determined by the cardiac forces. But in these early studies the square root of the ballistocardiogram amplitude, or of the areas of the I and J waves, was even more closely related to the acceleration of flow. Prof. H. C. Burger first pointed out to me that this finding was hard to reconcile with the obvious fact that the mechanical construction of our ballistocardiograph was such that one would expect it to record the forces directly, and so have a linear relation to them.

The explanation for this discrepancy is now at hand. In our former studies we had assumed that the maximum flow velocity, the integral of the acceleration of the blood ejected in our simulations of systole, had a direct linear relation to the forces applied. Estimates of the forces themselves, described in this paper, demonstrate that this is true of only part of the force applied, that due to

<table>
<thead>
<tr>
<th>Subjects</th>
<th>No.</th>
<th>Basis of method</th>
<th>Type of regression</th>
<th>or dynes</th>
</tr>
</thead>
<tbody>
<tr>
<td>All</td>
<td>76</td>
<td>H-I slope</td>
<td>Simple</td>
<td>28,100</td>
</tr>
<tr>
<td>All</td>
<td>76</td>
<td>H-I slope × Body Surface Area</td>
<td>Simple</td>
<td>28,300</td>
</tr>
<tr>
<td>All except E.I.</td>
<td>76</td>
<td>H-I slope × Body Weight</td>
<td>Simple</td>
<td>31,900</td>
</tr>
<tr>
<td>All except E.I.</td>
<td>76</td>
<td>H-I slope × Height</td>
<td>Simple</td>
<td>27,400</td>
</tr>
<tr>
<td>All except E.I.</td>
<td>65</td>
<td>H-I slope</td>
<td>Simple</td>
<td>26,000</td>
</tr>
<tr>
<td>All except E.I.</td>
<td>65</td>
<td>H-I slope and Height</td>
<td>Multiple</td>
<td>24,200</td>
</tr>
<tr>
<td>All except E.I.</td>
<td>65</td>
<td>H-I slope and Body Surface Area</td>
<td>Multiple</td>
<td>23,700</td>
</tr>
</tbody>
</table>

Standard deviation of initial cardiac forces used (except E.I.) around their own mean 49,000.
movement of fluid in the terminal limb of the cannulas; another part, the force of turning, varies with the square of this velocity. So our previous assumption, close to the truth when the forces applied were small or moderate, led to a serious underestimation of the forces when they were maximal, and taking the square root of measurements made on the ballistocardiograms, by having its chief effect on the largest, improved the correlation. In the data described in this paper taking the square root of measurements made on the ballistocardiograms does not improve the correlation with the forces that originated them.

In a previous study\(^4\) of the pulse we reported very strong correlation \((r = 0.92)\) between a product, the maximum ejection velocity of the aortic blood \(\times\) mean aortic blood pressure, and the maximum slope of the femoral pulse wave front. We have now tried the effect of substituting the initial cardiac forces for the maximum ejection velocity in the relationship with the pulse wave front given above. The correlation coefficient of this second relation was 0.85, which, when tested by Fisher’s \(z\) transformation,\(^8\) is found not to be significantly different from the value \((r = 0.92)\) obtained before. Obviously, the slope of the advancing pulse wave front is closely related both to the maximum velocity of the ejected blood and to the forces which bring about this velocity.

In a preliminary report\(^9\) of the present study on the ballistocardiogram we concerned ourselves with the relation between the slope of the II-I segment and the initial acceleration of the injected blood, and, after an adjustment for differences in body size, found very strong correlation between the two \((r = 0.88)\). In this presentation we have concerned ourselves with the forces that bring about this initial acceleration. These are also closely related to the slope of the II-I segment and, without an adjustment for differences in body size, the correlation is 0.85, which is not significantly different from that of the relationship studied before.

**General Considerations.** Many more doctors will interpret ballistocardiograms from inspection than from measurement, and for these our results can be summarized as follows. Under most clinical conditions the normality of initial cardiac forces can be judged from the depth of the I wave. However, if the cardiac output is large and the pulse wave velocity is unusually great, the I wave may be diminished in amplitude although the cardiac forces are strong. Fortunately the likelihood of an error of interpretation is small, for this unusual situation can be recognized at a glance; in our records an I wave, small despite large initial cardiac forces, is always followed by a large J wave. In contrast, when the I wave is reduced by cardiac weakness, the J wave is reduced also.

If one bases one’s judgment of the initial cardiac forces on the slope of the II-I segment, the error caused by differences in vascular elasticity is either minimized or avoided. So in reading ballistocardiograms routinely it is important to examine the record carefully for the sharp footward break of the II-I segment, which is such a conspicuous feature of normal records. If this sharp footward break is not found at the time of the onset of ejection, or if its usual steep slope is replaced by a more gradual one, this is strong evidence that the heart under study is not contracting with the jerk characteristic of hearts in good health.

**Comparison between Pulse and Ballistocardiogram as Method of Determining Heart Strength.** Some of our studies made by simulating systole at necropsy have been concerned with the pulse, others with the ballistocardiogram. In the last of those concerned with the pulse,\(^1\) I pointed out the drawbacks inherent in any pulse wave method of estimating cardiac function. In this paper, which may well be the last of the long series, I should do the same for the ballistocardiogram. Both methods give important information about cardiac function, and there are certain advantages and disadvantages inherent in each of them.

We have been unable to establish satisfactory quantitative relations between pulse
wave and cardiac function from records taken by apparatus pressing on the skin over an artery, because such transducers are so difficult to calibrate in terms of pressure within the vessel. So we have been forced to puncture an artery to do quantitative work, a technic far less satisfactory for routine clinical work than is that of securing a ballistocardiogram.

Under certain conditions the ballistocardiographic method encounters difficulties which the pulse method avoids. These are chiefly concerned with uncertainties which arise concerning the contribution of the two sides of the heart to the ballistocardiogram. While the contribution of the normal left heart to the ballistocardiogram outweighs that of the right by a ratio of 5 or 6 to 1, in disease this ratio might not hold. However asynchronism of the forces of the two sides of the heart can be readily recognized by abnormal notching of the ballistocardiogram and this should put one on one's guard. In contrast, the pulse is concerned with the function of the left heart alone, and so this confusion does not arise.

Considerations related to the size of the cardiac chambers and the energy expended before ejection begins affect both methods equally. An obstruction at the aortic valve would cause both pulse and ballistocardiogram to fall short of indicating the full strength put forth by the ventricle. When the obstruction is further out the ballistocardiogram has an advantage; for example, in coarctation of the aorta the pulse in the lower parts of the body may be abolished, but only the later systolic parts of the ballistocardiogram are influenced, leaving the I wave and the H-I segment, the parts of that record from which cardiac strength can be most safely judged, apparently altogether unaffected. And this part of the ballistocardiogram is likewise unaffected by the more peripheral obstructions, which may diminish or abolish the pulse wave in any artery.

Differences in the elasticity of the vessels have an effect on the I wave of the ballistocardiogram only if the vessels are unusually rigid and the cardiac output is high, and this is a minor disadvantage in judging myocardial function from that record; if one bases his judgment on the slope of the H-I segment, the difficulty is avoided. In contrast, knowledge of vascular pressure or elasticity is a sine qua non of every quantitative judgment of cardiac forces made from observations of the pulse.

Observations of the pulse, the blood pressure, and the ballistocardiogram comprise the only methods that at this time give any promise of affording doctors a routine estimate of the cardiac strength of their patients. They should not be considered as rival methods, but as methods complementary to one another. Neither is perfect and we will do well to secure all the information we can concerning the strength of the heart of our patients.

**Summary**

In 76 simulations of systole made on 7 cadavers, 5 perfused with blood and 2 with water, the "initial cardiac force" of the first 0.08 sec. of ejection has been estimated. By statistical analysis this force has been compared with the force recorded in the longitudinal axis of the body by the I wave of the ballistocardiogram taken simultaneously, and also with the slope of the H-I segment.

There is very strong correlation between the initial cardiac force applied and the average force recorded in the I wave of the ballistocardiogram. The regression is almost exactly parallel to the line of perfect agreement, but the force recorded in the I wave is smaller than the force applied. This difference can be readily accounted for.

By moving the point of observation closer to the onset of ejection the conflict with opposing forces is minimized, so the initial slope of the H-I segment affords a better measure of the initial cardiac forces than does the depth of the I wave.

Attempts to improve the estimation of the initial cardiac force from the ballistocardiogram by consideration of the subject's height and weight resulted in only a small reduction of the scatter; so body size is evidently a minor factor.
The initial footward deflection of the ballistocardiogram, the H-I segment, represents the reaction to the initial cardiac forces which accelerate the blood headward. Thus it is in essence an example of Newton’s third law of motion, and the qualifications that must be made to this statement are minor in nature.

From the slope of the H-I segment one can estimate the initial cardiac forces, which cause the initial acceleration of the ejected blood, with an accuracy equal to that of many good clinical methods.

Certain limitations of the ballistocardiographic method are discussed, and compared with those inherent in quantitative estimations of cardiac function from the pulse.

**ACKNOWLEDGMENT**

I am indebted to Mr. George Peirce for a very important suggestion which led to great improvement in the estimation of the “cardiac” force applied in these experiments. I am also indebted to Dr. Albert Schild for the major part of the statistical analysis of these data. Dr. Anna Corbascio collaborated on the rest.

**SUMMARIO IN INTERLINGUA**

In 76 simulationes del systole effectuate in 7 cadaveres—5 perfusionate con sanguine e 2 con aqva—le “fortia cardiac initial’’ del prime 0,08 secundas de ejection esseva estimate. Per medio de analysees statistic iste fortia esseva comparate con le fortia registrate in le axe longitudinal del corpore per le unda I del ballistocardiogramma obtenite simultaneamente e etiam con le inclination del segmento H-I.

Il existe un multo forte correlazione del applicate fortia cardiac initial con le fortia medie registrate in le unda I del ballistocardiogramma. Le regression es quasi exactemente parallel al linea de accordo perfecte, sed le fortia registrate in le unda I es inferior al fortia applicate. Iste differentia non es difficile a explicar.

Si on move le puncto de observation plus proxime al declaration del ejection, le conflictio con fortias opposites es reducete, de manera que le inclination initial del segmento H-I provide un melior mesura del fortia cardiac initial que le profundor del unda I.

Essayos de meliorar le estimation del fortia cardiac initial ab le ballistocardiogramma per prender in consideration le altor e le peso del subjecto resultava in solmente un leve reduction del dispersion, de manera que il deveni evident que le dimensiones del corpore non es un factor important in iste calculationes.

Le initial deflexion verso le pede in le ballistocardiogramma, le segmento H-I, representa le reaction al fortias cardiac initial que accelerar le sanguine verso le capite. Assi nos ha hic in essentia un exemplo del tertie lege de motion de Newton, e le modificationes que debe esser applicate a iste assertion es de natura minor.

Ab le inclination del segmento H-I on pote estimar le fortias cardiac initial, le quales causa le acceleration initial del sanguine ejeite, con un grado de accuratia equal a illo de multe bon methodos clinic.

Certe limitationes del metodo ballistocardiographic es discutite e comparate con le limitationes inherente in estimationes quantitativa del function cardiac ab le pulso.

**REFERENCES**


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INITIAL CARDIAC FORCES IN THE BALLISTOCARDIOGRAM


The authors report the results of mitral commissurotomy on 120 surviving patients of their first 131 operations. In this group 72.5 per cent either maintained an excellent result or continued to be significantly improved. Restenosis of the mitral valve occurred in 10 patients (8.3 per cent). Six of these patients were improved after the second operation. Arterial embolization occurred in 4 patients during the postoperative period of observation. This incidence represents one fourth the preoperative incidence. The functional status of the patient, his age, and the cardiac rhythm all play a role in the operative results. However, the anatomic status of the valve itself was the most important factor. When the valve was pliable and competent 17.5 per cent of the patients had a poor result, in contrast to a poor result in 47.5 per cent of those cases with an immobile valve or significant regurgitation. By present standards it is now recognized that many of the early operations did not accomplish complete opening of the valve. In particular, it is important to open the posteromedial commissure whenever indicated. At present, closed mitral commissurotomy still seems to be the procedure of choice for mitral stenosis. Perhaps when a satisfactory prosthetic replacement for the mitral valve is available, the patients with calcified immobile valves with regurgitation may become candidates for open-heart repair with the use of extracorporeal circulation.

Krause
Studies Made by Simulating Systole at Necropsy: XII. Estimation of the Initial Cardiac Forces from the Ballistocardiogram
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