Acceleration Ballistocardiography
Design, Construction, and Application of a New Instrument

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The construction of a linear accelerometer for use in ballistocardiography is presented including the relationship between acceleration, displacement, and velocity. The sensing element described is new in the field of ballistocardiography, being based on an electrochemical principle. A means of easily calibrating the instrument is described, thereby allowing quantitative measurements of various components of the ballistocardiogram. Tracings of one abnormal and two normal individuals are illustrated to show the relationship between displacement and acceleration.

Ballistocardiography has been established as a valuable diagnostic aid in the evaluation of a great number of cardiovascular conditions. Many types of recording apparatus are now available. They fall into two general categories, namely, the moving table type and the portable, direct body type. The moving table types have proved to be accurate instruments, but their great disadvantage is that the apparatus is bulky and complex. Most of the direct body types operate on the displacement or velocity principle and yield tracings which are excellent for qualitative analysis. However, because of the very principle upon which these instruments develop their electrical current, which is in turn amplified and recorded as the familiar ballistocardiogram, they fail to measure quantitatively the component of the ballistocardiogram which is theoretically the most important, namely, the force of movement imparted to the entire body by the recoil of the heart and the impact of the blood ejected from the heart.

The velocity pickup generates its electrical current by the passage of electromagnets through a fixed magnetic field, and the electrical current generated is directly proportional to the velocity of the body. Consequently, the amplitude of any of the waves is more a measurement of the time interval over which the force is applied and is not an accurate measurement of force. Similarly, with the displacement type pickup, the amplitude of the waves obtained is dependent upon how far the body is moved by the force but does not measure accurately the force which caused the movement. Standardization of either form of ballistocardiograph is difficult and at present is not available for general use.

Arbeit and Lindner and Smith and Bryan have recently devised an apparatus with which they differentiate velocity electrically into acceleration and integrate velocity into displacement, obtaining the so-called differentiated acceleration tracing.

When acceleration is differentiated from velocity, and velocity is difficult to standardize, it appears that the calibration of differentiated acceleration would be very difficult and perhaps unreliable. There are as yet no other feasible methods of obtaining acceleration tracings, except for the instruments based on the seismic principle. Since accurate measurements obtained by seismic instruments depend on a rather large mass, delicately suspended in a sphere or box, their size precludes use directly on the body, and they are not sensitive in the range of forces which are encountered in ballistocardiography.

Since force can be calculated from acceleration it seems that the ideal instrument for general use in ballistocardiography would be an accelerometer which is rugged, portable, inexpensive, and which could record on an electrocardiograph as does the Dock (electromagnetic) Ballistograph. This instrument should be capable of accurately measuring acceleration entirely independently of velocity.
or displacement. It should be sensitive to forces in the range from 0.00 to 0.02 g and have the ability to follow changes in direction of acceleration (from head to foot, side to side, or from anterior to posterior) very rapidly. In addition, the apparatus should be linear in its response and should be capable of measuring acceleration rather than merely changes in acceleration. It should be capable of simple calibration, preferably by the use of the 1 millivolt standardization of the electrocardiograph machine, thereby eliminating additional equipment which is often cumbersome and subject to its own critical standards. With such an instrument anyone possessing an electrocardiograph could take ballistocardiograms directly from a patient, and by presetting the sensitivity to a previously determined 1-millivolt deflection, could read the ballistocardiogram in terms of force. This would mean an accurate quantitative evaluation of the ballistocardiogram in a comparable manner to that now used in electrocardiography.

We have been able to construct an instrument that is capable of a flat response from 0 to 3000 cycles per second (without capacitance in the circuit) and shows a linear response from 0 to 0.015 g forces and above. It reproduces the typical ballistocardiogram in its gross contour and is capable of recording consistent small deflections in the systolic pattern not seen in the smooth record of the velocity or displacement type apparatus. These changes will be described in more detail below.

This accelerometer is an electrokinetic device whose operation is determined by electrical changes at the surfaces of contact of a mercury-sulfuric acid interface, the so-called "U effect." The general theory of the "U effect" is given in several papers by Ueda and others12, 13 and is summarized briefly in an interesting manner by Yeager and Hovorka.14 Although the theory is beyond the scope of this paper, the construction of this apparatus is amazingly simple. The actual pickup consists merely of a capillary tube with alternate layers of metallic mercury and 1 normal sulfuric acid, thereby creating multiple interfaces of mercury and electrolyte solution. At each end of the capillary a mercury seal is employed into which a small copper or platinum electrode is inserted and in turn sealed to the capillary tube. Upon movement such an element produces electrical voltage by changes of the surfaces of the mercury and electrolyte solution, the voltage being proportional to the change of contour of these interfaces. The element responds maximally to acceleration in its longitudinal axis, and this electrical output can be transferred directly to the recording apparatus (electrocardiograph) through the right arm and left arm electrode leads with the selector switch on lead I. A filtering system can be used in the circuit to filter out high frequency waves, but this is not necessary. This element is mounted on a cross bar between the shins of the patient with the element lying in an exactly level position in the longitudinal axis of the body. The standard leg block, as used in the velocity and displacement type pickups, is utilized in facilitating the mounting of this apparatus, and, of course, a solid table for the patient is essential as in the other portable type pickups.

The element is easily calibrated with respect to its response to a known force by means of a pendulum. By adjusting the sensitivity control of the electrocardiograph to a desired valley-to-peak deflection, and then in turn noting the deflection of the stylus upon the introduction of 1 millivolt into the electrical circuit, a constant reproducible record of force can be obtained at will by merely presetting the sensitivity control to this desired deflection of the stylus.

Polarization of the element is readily accomplished during standardization on the pendulum while periodic motion is taking place. By arbitrarily calling one direction of the pendulum "headward" and the other "footward," the deflection of the recording stylus can be noted with each direction. It must be noted, however, that when properly polarized in this fashion the pendulum direction called "headward" causes the stylus to show a downward deflection, which, on first consideration would lead one to the assumption that the apparatus is polarized in reverse. Upon careful consideration, however, it must be recalled that this element is a true accelerometer, and it is the electrical output of the mercury and electrolyte interfaces that
we wish to record and not the movement of the capillary tube. Consequently, a “headward” movement of the tube causes a “footward” movement of the interfaces because of the inertia of the liquid within the capillary, and actually we record the mirror image of displacement or a minus sine wave pattern in contrast to the sine wave of displacement. With the apparatus calibrated in this manner, the typical gross pattern of the systolic component of the ballistocardiogram is recorded from the body with the J wave being the prominent headward deflection and the K wave being the prominent footward deflection.

**Design, Construction, and Calibration**

The above described capillary tube with the mercury and sulphuric acid interfaces is mounted perpendicular to a cross piece containing a series of capacitors in parallel leading to a suitable connection for the electrode leads from the electrocardiograph. A rotor switch allows selection of capacitances varying from 0 to 0.009 microfarad for use as desired in filtering out high-frequency vibration.

**Construction of Capillary Element**

The exact inside diameter of the capillary tube is relatively unimportant since each element is calibrated according to its output characteristics when subjected to known forces. However, the stability of the element is proportional to the diameter of the capillary tube. Tubes of smaller diameter are more stable than those of large diameter because of the capillary attraction phenomenon. The ideal inside diameter appears to be in the vicinity of 0.3 to 0.5 mm., and such capillary tubes are easily drawn by hand from soft glass tubing. Thick-walled capillary tubing can also be used if the inside diameter is not too great. In addition, to the stability factor, the smaller the diameter (within certain limits), the greater the electrical output per number of interfaces.

After cutting off one end of the glass tube, leaving at least 20 to 25 cm. of capillary attached to the other end of the glass tube, a 90 degree angle is made near the distal tip of the capillary tube. The capillary is now ready for “loading.” We have found it best to “load” capillaries in a horizontal position through the small tip by suction with a syringe from the large end of the tubing. A 1 cc. tuberculin syringe, mounted on a stand, is satisfactory for this. A fine control of the suction facilitates loading, and this can be accomplished by bolting a plate to the plunger of the syringe and in turn connecting this plate to a micrometer screw. The large end of the capillary tube is attached to the tip of the syringe by means of plastic or rubber tubing, care being taken to approximate the glass tubing and the tip of the syringe as closely as possible, thereby allowing little play in this connection. At the tip of the capillary tube, a small cup half filled with metallic mercury and half filled with normal sulfuric acid is placed. By gently swinging the tip of the capillary tube alternately in and out of the mercury meniscus, and simultaneously creating a negative pressure in the capillary tube by the use of the micrometer screw, alternate layers of mercury and sulfuric acid can be drawn into the capillary tube.

It is best to preload the syringe and capillary tube with normal sulfuric acid to eliminate the compression and expansion properties of a gaseous medium which tend to make negative pressure difficult to control. When the tube has been “loaded” to the desired length, and with the desired number of mercury cellules, a sealing wax seal is applied to the tip of the tube following its removal from the cup. (A length of capillary tube approximately 13 to 15 cm. in length, containing approximately 25 to
40 cells of mercury, appears to be adequate for ballistocardiography.) When this has been accomplished, the "loaded" tube may be cut at its desired length near the proximal end, and a small copper wire electrode inserted into the large mercury cellule and again sealed with sealing wax or suitable substance. The distal limit of the element is then measured, cut, and the above process of inserting a copper electrode is repeated, and the element is finished. (See fig. 1, diagram A.)

Construction of the Acceleration Ballistocardiograph

The cross piece can be designed of metal, plastic, or wood. Additive capacitors in parallel are embedded in the cross piece, and connected to a rotor switch, which allows selection of the desired capacitance ranging from 0 to 0.009 microfarad with 0.0015 or 0.003 microfarad gradations. A three pole plug is mounted for connection of the two electrode leads from the electrocardiograph and a grounding wire. The electrical circuit is diagrammed in figure 1, diagram B.

The capillary tube is next enclosed in a cylindrical aluminum or copper shield, which is connected to a ground wire. This shield is advisable to prevent the possibility of picking up alternating current interference from surrounding electrical fixtures and also affords protection against breakage of the glass tube. Grounding has been found necessary because the capillary element produces an electrical potential and is capable of acting as an antenna for electrical frequencies if unshielded and exposed. The shielded element is mounted to a cross piece on a hinged plate, with a tension spring and set screw adjustment at the end of the plate opposite from the hinge for exact leveling of the element in the horizontal plane. The cross piece is grooved on its bottom surface to cradle comfortably on the shins and is strapped into a fixed position in a snug, but not tight manner by means of rubber tapes (fig. 2).

Calibration of the Acceleration Ballistocardiograph

As mentioned above, an accurate calibration can be accomplished with regard to g forces by the use of a pendulum. The pendulum can be constructed by use of a heavy solid platform (approximately 6 by 8 by ¼ inch steel or brass) mounted from each of its four corners by fine piano wire. The ballistocardiograph is placed on the pendulum with the element in a horizontal position with its long axis in the longitudinal swing of the pendulum. A vernier scale, mounted at one end of the platform, allows accurate measurement of the arc of the swing.*

Acceleration thus varies sinusoidally with time for small amplitudes of swing, and the maximum amplitude determines the maximum acceleration. Since the pendulum we use has a length of 25 inches, the displacement of ¼ inch will give an acceleration of 0.005 g. Knowing this, by use of the vernier scale we can release the pendulum at ¼ inch from its 0 position, to develop a force of 0.005 g; if the pendulum is released ¼ inch from its 0

* The pendulum we use has a length of 25 inches.

If T represents the tension on the supporting wires, mg the weight of the pendulum (the product of mass and acceleration due to gravity), the angle of the supporting wire from the vertical position, and mg tan φ the component of the forces in the horizontal direction, we can set up the following equations for calculation of the force of any swing of the pendulum:

\[ f = ma = mg \tan \phi \]

Therefore \( a = g \tan \phi \). But for small values of \( \phi \) (when \( \phi \) is expressed in radians) \( \tan \phi = \phi \). Therefore, \( a = g \phi \). Since \( \phi \) is very nearly equal to \( d/l \) (where \( d \) represents distance the pendulum moves from its zero position and \( l \) represents the length of the pendulum) we have the following: \( a = (d/l)g \).
Fig. 3. Response of the acceleration ballistocardiograph to known forces (as delivered by the pendulum). Note the linear response as the force increases. The output is increased with a capacitance of 0.009 microfarad in the circuit, but this in no way affects the linear response. A stylus deflection, from 1 millivolt, of 16 mm. was used in all tracings. The sensitivity setting should be made with the instrument in the circuit. Measurements are made from valley to peak of the seventh cycle; dividing this measurement by two gives the correct deflection from any given force, since the 0 point of the pendulum is midway between the valley and peak deflections.

position, it develops 0.010 g; if the pendulum is released ½ inch from its 0 position, it develops 0.015 g; it will continue to develop an increasing force of 0.005 g for each ½ inch added to its swing. (This linear response applies only within certain limits, but these limits far exceed the forces we measure in ballistocardiography.)

Tracings obtained from this calibrating apparatus, are shown in figure 3. All measurements were taken on the valley-to-peak swingings of the stylus at the seventh cycle of the pendulum. It will be noted that there is a linear response of the acceleration ballistocardiograph with the increasing g forces. It must be noted, however, that calibration of any one element is not the same when different capacitance is used in the circuit. The response for any given capacitance in the circuit does not in any way affect its linearity, but the response from 0.005 g with a 0 microfarad setting and the response to the same force with a 0.009 microfarad setting will differ slightly. This is due to the fact that the element produces its own electrical current, and, with increased capacitance in the circuit, current in the circuit is increased. This fact affects the calibration only slightly on various microfarad settings and can be corrected by merely checking the output of the ballistocardiograph on all microfarad settings, constructing a table of the output differences with each setting, and using this table as a reference when measuring the ballistocardiograms. Once the ballistocardiograph has been calibrated on the pendulum and its response to various forces recorded, no further calibration is necessary by use of the pendulum.*

* The instrument illustrated in figure 2 has maintained a constant output to a given force for a period of eight months, and shows no signs of deterioration.
All acceleration tracings illustrated in this paper, were taken with a ballistocardiograph standardized at a valley-to-peak deflection at the seventh cycle of 36 mm. with 0.015 g with a 0.009 microfarad capacitance in the circuit, and the sensitivity control set to a 16 mm. deflection with 1 millivolt. The valley-to-peak measurement represents a total force of 0.030 g, and, therefore, from the 0 point of the pendulum, 0.015 g is equivalent to 18 mm. deflection of the stylus. Therefore, a 1 mm. deflection is equivalent to 0.00085 g. It is a simple matter to convert this measurement into dynes when the weight of the subject is known. 

\[ F = ma \] (where force is in dynes, \( m \) is the weight of the subject in grams and acceleration is 980 cm. per second^2). Since \( g \) may be substituted for \( a \) and each millimeter deflection is 0.00085 g, our equation is 

\[ F = 0.833. \]

If we had a subject weighing 154 pounds (70 Kg.) we would have the following equation:

\[ F = 7000 \text{ (Gm.)} \times 0.833 \]
\[ F = 58,310 \text{ dynes per millimeter deflection of the stylus} \]

If we want to measure the force of \( J_a \) and its amplitude is 10 mm., it is simply \( 5.83 \times 10^4 \) times 10 or \( 58.3 \times 10^4 \) dynes. Such measurements can be calculated for either headward or footward deflections.

**Discussion**

By using the above described instrument on normal individuals and on patients with known cardiac disease we have obtained interesting tracings. Various headward and footward motions that are not seen in displacement and velocity tracings are consistently seen in acceleration tracings made with this instrument. A problem of nomenclature arises in the use of true acceleration ballistocardiograms, not allowing use of the present lettering system, as described by Braunstein. The Committee on Ballistocardiographic Terminology has presently deferred this problem, allowing the possibility of confusion to exist regarding this.

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At this writing. The stability of the element depends on several factors (the pH of the acid, the number of interfaces remaining constant, and airtight seal), but these are technical problems that can be overcome with better construction techniques now under investigation. The use of soft glass is not advisable, since the acid will change in pH after several weeks and the output will noticeably diminish. Elements of Pyrex tubing have been found to be relatively stable.

In an attempt to avoid further confusion, we have used the nomenclature used by Arbeit and Lindner of labeling the prominent headward deflection of acceleration as \( J_a \). The subscript, \( a \), would likewise be appropriately used on \( H, I, \) and \( K \) waves that correspond in gross contour to waves so labeled in displacement and velocity tracings. In such a system of labeling, it is to be noted that \( J_a \) occurs simultaneously with \( I \) (of displacement). Consequently, it must be kept in mind that the genesis of \( J_a \) and \( J \) are not the same as would be assumed by such a lettering system.

Since all systolic complexes are not identical, even in the same person, we feel it inadvisable to compare anything other than simultaneous displacement and acceleration ballistocardiograms. We accomplish this by mounting acceleration and electromagnetic ballistographs on a patient’s legs at the same time and recording their output simultaneously on a double-channel, direct writing electrocardiograph. We obtained displacement tracings by using a 20 microfarad capacitance in the circuit of a Dock (electromagnetic) Ballistograph, thereby integrating (almost completely) velocity input to a displacement output. The R wave of the electrocardiogram is superimposed for time reference.

Notching of \( I_a \), in subjects considered normal, is a fairly consistent finding, as is notching to biphasic components of \( H_a \). High speed (50 mm. per second) tracings frequently demonstrate notching or slurring of the \( I_a-J_a \) limb in normal subjects. Accentuation of these findings is often marked in subjects with heart disease. A clear-cut biphasic component of \( J_a \) is at times evident, indicating a very fast change in the body motion (frequently not detected by displacement or velocity tracings). We have further labeled such biphasic waves using the numbers 1 and 2 as subscript.

Of particular value is the ability to quantitate various headward and footward deflections of the acceleration ballistocardiogram. These measurements are made directly from the tracings and allow tracing comparisons from one person to another. It appears that quantitative comparison of wave amplitudes will
lead to the establishment of ranges of normality and perhaps be of value in the interpretation of ballistocardiograms.

By quantitative analysis of the cases illustrated, the force in any phase of cardiac systole can be appreciated. If we consider the normal tracings to be representative of expected limits, we find a range of Jn amplitude to vary from $70.6 \times 10^4$ to $98.6 \times 10^4$ dynes in inspiration, and from $51.3 \times 10^4$ to $67.5 \times 10^4$ dynes in expiration. By comparison, in the abnormal tracing illustrated, we find the Jn amplitude to represent only $34.4 \times 10^4$ dynes in inspiration and $13.7 \times 10^4$ dynes in expiration.

Because components of the acceleration
Fig. 5. A. P., 23-year-old white male, weighing 77 Kg. Displacement and acceleration ballistocardiograms are considered normal. The entire systolic complex is of smaller amplitude than in the preceding case, far smaller in terms of actual force than would be expected by comparing displacement tracings on both individuals. The $J_s$ amplitude in inspiration, represents a force of $70.6 \times 10^4$ dynes, far less than in the preceding case. Note the slight irregularity of the $I_s$-$J_s$ limb in the acceleration tracing not seen in the displacement tracings. High-speed tracings reveal this irregularity to be actually a slight notching, still not evident in the displacement tracings.

Tracings are multiple and occur very rapidly, we have taken recordings at both standard speed (25 mm. per second) and double speed (50 mm. per second). The latter gives a "slow motion" tracing allowing easier differentiation of the various components of the complexes. Because of the rapid speed of the paper traveling at 50 mm. per second, quantitative comparison between these tracings and tracings taken at 25 mm. per second are not valid, since we attempted no calibration of our ballistocardiograph on high speed tracings. These
Fig. 6. Simultaneous displacement and acceleration tracings of A. H. Note that the 1 millivolt standardization is 16 mm. (as in all acceleration tracings illustrated) yet most of the $J_a$ waves fail to represent more than $34.4 \times 10^4$ dynes and many are far less than that. Of interest is the definite biphasic component of the $J_a$ wave following a compensatory pause after a premature ventricular contraction (labeled as $J_m$ and $J_m$ in the acceleration tracing). A notch is noted in the displacement $J$ wave of the same origin, but the acceleration $J_a$ wave clearly demonstrates the forces, as well as their direction, far more vividly. The high-speed tracing demonstrates notches of the $I_a$ and $J_a$ waves, not even suspected in the displacement $I$ and $J$ waves.

Tracings are used merely to visualize more easily qualitative differences between velocity and acceleration tracings.

Figures 4 and 5 are tracings obtained from healthy males of ages 24 and 23, weighing 74 Kg. and 77 Kg., respectively, who show no evidence of any cardiovascular abnormality on physical examination or by electrocardiogram. They represent what we have found to be a representative range of amplitude and are considered to be normal ballistocardiograms. Figure 6 shows acceleration and displacement
ballistocardiograms in the case of a 51 year old white female weighing 55 Kg., six years after a myocardial infarction. Electrocardiograms revealed a left bundle branch block and an occasional premature ventricular contraction. She is on a full maintenance dose of digitalis and remains in borderline decompensation in spite of restricted activity and full digitalization. The high speed tracing (50 mm. per second) vividly demonstrates the notching of the Jₐ wave in acceleration as discussed above.

**Summary and Conclusions**

Acceleration is a valuable measurement in ballistocardiography. We have designed and constructed a portable, body-type, acceleration ballistocardiograph that is capable of directly sensing acceleration. It is easily constructed, free from complex electrical circuits, and yields consistent results. This accelerometer has a flat frequency response (without capacitance) from 0 to 3000 cycles per second and is linear well beyond the forces encountered in ballistocardiography.

The described accelerometer allows quantitative evaluation of ballistocardiograms, since it is readily calibrated by use of a simple pendulum, and once it has been calibrated on a pendulum its response can be standardized by use of the 1 millivolt standardization of the recording apparatus (electrocardiograph). Ballistocardiograms can be read quantitatively in terms of force allowing valid comparisons of tracings from one patient to another.

Examples of the clinical use of the acceleration ballistocardiograph have been presented revealing wave components in certain instances not seen in displacement ballistocardiograms.

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**Sumario Español**

La construcción de un acelerómetro linear para el uso en ballistocardiografía se presenta incluyendo la relación entre aceleración, desplazamiento y velocidad. El elemento sensorio descrito es nuevo en el campo de la ballistocardiografía y está basado en un principio electroquímico. Una manera de calibrar el instrumento se describe, de esta manera permitiendo medidas quantitativas de los varios componentes del ballistocardiograma. Trazados de un individuo anormal y dos normales se ilustran para mostrar la relación entre desplazamiento y aceleración.

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