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THE TRANSMISSION CHARACTERISTICS OF LARGE AND SMALL PRESSURE WAVES IN THE ABDOMINAL VENA CAVA

by

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SUMMARY

The mechanical behavior of the abdominal venae cavae of anesthetized dogs has been studied by measuring the speed, attenuation and changes in wave form of various kinds of artificially induced pressure signals. The propagation of large amplitude pressure waves is shown to be affected by reflection interference and pronounced nonlinear phenomena. For pressure signals exceeding a few mm Hg the speed increases with amplitude and the wave front steepens during propagation as in the early phases of the formation of a shock wave. By inducing distension waves in the form of finite trains of sine waves with amplitudes less than 20 mm H_2O the dispersion and attenuation were determined without requiring Fourier transform computations. The vena cava was found to be only mildly dispersive for frequencies between 20 and 100 Hz and the logarithmic decrement appears to be independent of frequency. Irrespective of the amplitude and shape of the pressure signals, their speeds varied along the vena cava and also with respiration. In addition the speeds generally increased under the influence of the chemical and electrical stimuli applied and with rising transmural pressure.

INTRODUCTION

Extensive studies have been made on the mechanical behavior of larger arteries of various species, but little information is available regarding the dynamic elastic properties of veins even though their role in maintaining and controlling circulation is certainly no less important than that of arteries. This lack of knowledge may in part be attributed to the difficulties encountered in obtaining quantitative data on the distensibility of veins. These difficulties are primarily due to the inherently high flexibility of veins, their complex anatomy and the low transmural pressure that prevails in most parts of the venous system. Even small variations in the transmural pressure produced for example by respiration, changes in posture and muscular contraction may cause substantial changes in the geometry of the vessels, in their state of stress and in the elastic or viscoelastic properties of their walls, which all can strongly affect the distensibility of the vessels. Since these variations occur naturally and are not necessarily small, it must also be expected that veins exhibit a more complicated behavior than arteries.

It is well known that the mechanical behavior of individual blood vessels can be studied indirectly in terms of their wave transmission characteristics. This approach is particularly convenient since it does not require any trauma of the vessel wall and yields essentially local and instantaneous values for the changing elastic and viscoelastic parameters. Also, the availability of ultrasound sensors has made it possible to adapt the wave transmission techniques validated in animal experiments to the noninvasive study of large arteries and veins in the extremities of man. This should facilitate extension of present knowledge of the behavior of individual superficial limb veins which has been obtained by measuring the pressure rises induced by the injection of known volumes of fluid into temporarily isolated segments of veins *in situ*.[1) However, before a meaningful quantitative interpretation of the wave transmission properties of these vessels is possible, we must establish a mathematical theory that predicts these properties with sufficient accuracy as functions of the geometric and physical features of the vessels.

Medical researchers have measured transmission properties of naturally occurring and artificially induced pressure waves in arteries as well as veins.[2-9] The data obtained in these studies and related work, however, do not lend themselves to an incisive assessment of the relevance and accuracy of the various mathematical models that have been postulated for the mechanical behavior of the blood vessels. In most cases the shapes of the pressure signals were too restricted or the amplitudes too large to allow for a broad evaluation of basic wave transmission properties such as dispersion and attenuation.

Aiming for the development of a noninvasive method of determining the elastic behavior of individual arteries and veins and the physiological control mechanisms that affect it we began systematic theoretical [10,11,12] and experimental[13,14] investigations of the dispersion and attenuation of pressure, axial and torsion waves in large blood vessels. Our formal reports on the results of these efforts have so far largely dealt with the propagation of small amplitude waves. In view of the significance of certain nonlinear phenomena affecting the transmission of large amplitude pressure signals[4,13,15] we are presenting here our findings on the propagation of both large and small pressure waves in the abdominal vena cava of anesthetized dogs. As will be shown, the nonlinear behavior must be taken into account in the interpretation of much of the published data and in the design of future experiments directed to the elucidation of the mechanisms mediating changes in the distensibility of blood vessels.

EXPERIMENTAL METHODS

Waves in blood vessels can be characterized by either wall displacements, intraluminal pressure perturbations or flow fluctuations. Depending on the type and frequency of the waves considered we may find that only one or two of these variables are of sufficient magnitude to allow for accurate measurements. For example, distension waves are easily sensed in terms of their pressure or flow fluctuations, while for the detection of axial waves it is most convenient to monitor the corresponding axial wall displacements.[14] Obviously, the choice of variables to be measured depends not only on the nature of the waves but also on the sensitivity of available transducers and on whether the waves are to be recorded by an invasive or a non-invasive technique. Pressure or distension waves can be readily induced in arteries and are observable over distances that are relatively large compared to the vessel radii when the frequency is less than 100 Hz.[13,14] Guided by these facts we have generated artificial distension waves in the vena cava of anesthetized dogs.

The pressure perturbations in the vena cava were monitored by highly sensitive catheter-tip manometers utilizing Bytrex pressure cells Model HFD-5*). One of these transducers is shown in Figure 1. It has a diameter of 3 mm and a sensitivity of 80 to 100 μV/mm Hg with an excitation of 25 volts and therefore allows for a resolution of a fraction of one mm H_2O. The natural frequency of such pressure cells was found to be above 60 kHz in air and their response is linear within 1% from 0 to 300 mm Hg. The signal from the solid state strain gauge bridge used in these transducers is amplified by Astrodata amplifiers Model 885**) and recorded on a Honeywell Visicorder Model 1108 equipped with galvanometers whose frequency response is flat up to 3300 Hz.

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By injecting into a blood vessel a given volume of blood or saline from a spring-loaded syringe one can produce a pressure pulse whose amplitude and shape depend on the volume and injection time as well as the distensibility of the vessel. In our experiments we induced such signals with the syringe shown in Figure 2. The amplitudes of the resulting pressure signals were generally between 30 and 60 mm Hg for injections of 0.5 cm^3 and as much as 225 mm Hg for 1 cm^3. A tracing of a typical pressure pulse generated in the abdominal vena cava of an anesthetized dog is illustrated in Figure 3. The speed of the pressure pulse is usually defined by the transmission time Δt of a characteristic point of the wave front over a given distance Δx. As a characteristic point we have chosen here the intersection of the tangent in the inflection point of the wave front with a line approximating the unperturbed venous pressure. For this point Δt could be measured with a high degree of repeatability in contrast to other admissible choices for the characteristic point, which displayed a rather wide scatter in the measurement of the transmission times.

While spring-loaded syringes are convenient tools for inducing transient pressure waves, they do not allow for accurate control of the shape of the pulse. For example, slight changes in the orientation of the cannula at the point of injection can produce marked differences in the pulse shape. Moreover, in order to determine the wave velocity as a function of the frequency from such pulse waves it would be necessary to evaluate the Fourier spectra of the signal at different points along the vessel. These laborious Fourier transform computations can be avoided by generating a sinusoidal pressure signal. In some of our experiments this was done by the pump shown in Figure 4. It consists of a motor-driven reciprocating piston and was connected to the vena cava by a flexible plastic cannula that was inserted through a femoral vein. Typical tracings of sinusoidal pressure waves induced by the pump are shown in Figure 5. As illustrated in this figure, the speed of the signals over a known distance Δx was evaluated from the transmission time Δt of the points of intersection of the tangents to the sine wave at two successive inflection points.

With the pump used it was not possible to induce sine waves of small amplitudes over a frequency range wide enough for an extensive study of dispersion. Also, the inertia of the pump and of the fluid contained in it was sufficiently large to cause some distortions of the sine waves during the starting phase of the pump at essentially all frequencies between 4 and 40 Hz. By inserting an electrically driven piston (Figure 6) directly into a femoral vein and positioning it near the bifurcation, sine waves could be generated with frequencies between 15 and 100 Hz. The piston consists of a lightweight aluminum rod that is attached to the core of a solenoid and is protected by a brass cylinder which also prevents any direct contact of the moving rod with the vessel wall. The sinusoidal displacements of the piston are induced by means of an electronic oscillator and a high fidelity amplifier.

At high frequencies one can avoid noticeable reflection interference even when the amplitudes of the waves are not necessarily very small[13)] and when no dissipative mechanisms are present. This is accomplished by generating finite trains of sine waves with the aid of a tone burst generator. For sufficiently short trains the waves can be recorded before the reflections arrive at the sites of the transducers. Tracings of representative pressure signals of this type are given in Figure 7.

The attenuation of these signals was determined by comparing the peak-to-peak amplitudes of corresponding waves at the two pressure recording sites. The amplitude is defined as shown and denoted by A_o at the proximal transducer and by A at the distal transducer.

For mildly dispersive media, i.e. those in which the phase velocity does not vary by more than a few per cent whenever the frequency is changed by 10%, the speed of a finite train of sine waves can be interpreted as a good approximation of the phase velocity corresponding to the frequency of the sine waves. This is evident from the Fourier spectra of finite trains of sine waves shown in Figure 8. Such signals are dominated by harmonic components whose frequencies are close to that of the sine waves in increasing propertion to the length of the train. Therefore, if the vessel is only mildly dispersive the variation of the phase velocity with frequency can be determined simply by measuring the speeds of trains of sine waves of different frequencies without having to perform extensive Fourier transform computations. Likewise, the attenuation characteristics of the vena cava can be obtained directly by measuring the amplitude ratio A/A_o for wave trains of various frequencies.

EXPERIMENTAL PROCEDURE

The animals used in the experiments were mongrel male dogs of uncertain age and weighed between 15 and 40 kg. They were anesthetized intravenously with 30 mg/kg sodium pentobarbital (Nembutal). The dogs were kept in supine position throughout the experiment. As an indicator of the general physiologic state of the animals their arterial blood pressure was continuously recorded from a Statham manometer P23Dd connected to a cannula which was inserted into the left carotid artery or a femoral artery. In addition, their respiration was usually monitored by a mercury-filled silastic tube stretched across the chest. The variations in gauge resistance with breathing were recorded to allow for the study of the effects of respiration on the wave transmission characteristics of the abdominal vena cava. Besides this, the venous pressure at the bifurcation was continuously measured with a Statham P23BB transducer and a saline manometer with the cannula inserted through a femoral vein. The outputs from the Statham gauges were amplified with Honeywell Accudata Amplifiers Model M-104 and recorded on the Visicorder.

The general experimental arrangement is schematically illustrated in Figure 9. To detect small pressure fluctuations two Bytrex catheter-tip manometers were inserted into the right external jugular vein and positioned in the abdominal vena cava with the aid of an X-ray fluoroscope. The catheter-tip manometers, the Statham gauges, amplifiers and recording equipment were calibrated as a system for each experiment. Depending on recording speed and amplitude of the signals we could evaluate Δt with an error of 0.3 to 1.0 milliseconds.

When the artificial pressure or distension waves were generated by the spring-loaded syringe or by the sinusoidal pump, the cannula from either device was inserted through a femoral vein and placed near the bifurcation. When the electrically driven vibrator was used, the brass sleeve which encloses the piston and has an outside diameter of 6.2 mm was disconnected from the vibrator, filled with heparnized saline and inserted through a femoral vein into the common iliac. The vibrator piston was then guided into the brass sleeve. This procedure was adopted to overcome the difficulties in manipulating the vibrator, which weighs 10 kg, and to facilitate the positioning of the relatively large brass cylinder.

The distance between the tip of the cannula or the tip of the vibrating piston and the nearest catheter-tip manometer was usually 3 to 7 cm, while the distance separating the two manometers ranged in general from 2 to 7 cm. These distances were determined radiographically with an accuracy of .1 cm by using an image intensifier and a radio-opaque grid with a mesh size of 1 cm and taking into account the effects of parallax.

Theoretically the wave transmission properties of arteries and veins can be affected by changes in the transmural pressure and in the geometry of the vessels.[10,11,16] For accurate estimates of the elastic properties of blood vessels from their wave propagation characteristics it is therefore desirable to know the geometry of the vessel and the prevailing transmural pressure. To determine the dispersion and attenuation of waves in the abdominal vena cava in presence of a well-defined transmural pressure, the abdomen of the dog was opened by a midline incision from the lower end of the sternum to the pubis and the vena cava was exposed to atmospheric pressure by displacing the viscera. This surgical process also enabled us to observe the vena cava during the experiment and to measure with calipers its external diameter and wall thickness. Variations in the level of the transmural pressure were achieved with the aid of a tilt-table or by occluding the vena cava above the hepatic branches with a balloon catheter.

RESULTS

A. Waves Generated by Spring-loaded Syringe

To assess the feasibility of utilizing wave transmission phenomena in large veins for the purpose of determining their mechanical properties under various conditions we induced relatively large pressure pulses in the vena cava of 20 anesthetized dogs with the spring-loaded syringe shown in Figure 2. The speed of the pressure signals normally ranged from 300 to 700 cm/sec and was found to depend strongly on the amplitude, the instantaneous transmural pressure and the general physiologic state of the animals. The amplitude itself depended on the volume of saline injected, the injection time, the orientation of the syringe cannula, the venous pressure and, of course, the local distensibility of the vessel. A typical example of the variation of the pulse speed with amplitude is illustrated in Figure 10 which shows that the signal induced by injecting 0.5 cm^3 saline has an amplitude of 60 mm Hg and travels at a velocity of 500 cm/sec while the signal produced by 1 cm^3 saline has an amplitude of 225 mm Hg and a speed of 774 cm/sec. As a rule, pressure signals travel at a greater speed when they are superimposed on a higher transmural pressure. This behavior has been consistently observed in experiments in which the transmural pressure was increased by as much as 200 mm H_2O by tilting the animal. Representative variations in the arterial and venous pressures and in the wave speed with tilt are given in Figure 11.

By varying the location of the transducers we observed that the amplitude generally decreased with increasing distance from the cannula except in the neighborhood of the renal branches. The speed also varies as indicated in Figure 12. We attribute such changes in wave speed to local variations in the distensibility caused for example by the branch structure of the renal veins. We also noted changes in wave shape in this region which are presumably due to reflections at the renal branches.

In the presence of a mean flow the velocity of pulse waves propagating in the direction of the flow is predicted to be approximately equal to the wave speed at zero mean flow plus the average stream velocity,[17)] which generally is expected to be rather small in the vena cava compared with the propagation velocity of pressure pulses.[2,18)] We observed that the speeds of pressure signals generated during inspiration were consistently greater than those induced during expiration or at the resting phase of the respiratory cycle. In some cases the difference exceeded 200 cm/sec as indicated in Table I. By measuring continuously the total and static (end and lateral) venous pressure we found that the blood flow velocity increased during inspiration by approximately the difference in the pulse wave speeds. Since we are primarily interested in studying the mechanical properties of the vena cava in terms of its wave transmission characteristics at known or negligible blood flow rates we have avoided these respiration-induced flow effects on the wave speed by consistently selecting data obtained during the resting phase of the respiratory cycle.

Significant increases in wave speed were also produced by vagal stimulation, fibrillation of the heart and by injection into the vena cava of massive or lethal doses of various drugs such as epinephrine, nembutal, KCl, NaCl, $MgCl_2$, $CaCl_2$ and KCN. In some cases the speed increased by more than 400%. These findings are currently under detailed investigation, with particular emphasis on venous motor control mechanisms including the effects of changes in venous pressure.

With a reduction in the injection volume it is possible to generate sufficiently small pressure pulses by the spring-loaded syringe to minimize the observed nonlinear behavior with amplitude. But to obtain the dispersion and attenuation we would still have to determine the Fourier spectra of the signals.

B. Waves Generated by Pump

Initial results on the dispersive nature of the vena cava were obtained by measuring the speeds of pressure signals of the form of long trains of sine waves induced by a pump. The frequency and amplitude of the sine waves were systematically varied. For peak-to-peak amplitudes less than 3 mm Hg no evidence of reflection interference could be observed between the bifurcation and the renal branches. This was concluded from the fact that the signals retained their sinusoidal character during propagation independent of frequency. In the exposed vena cava the speeds of such sine waves with frequencies between 4 and 40 Hz ranged from about 60 to 400 cm/sec depending on pressure and the general physiological state of the animals. Representative dispersion curves are illustrated in Figure 13. In each case shown the speed varied by no more than $\pm$ 15% with frequency, which confirms that the exposed vena cava is not strongly dispersive.

For sinusoidal wave trains of sufficiently small amplitude the velocity was not dependent on the selection of the characteristic point and could be determined with greater certainty than in the case of single pressure pulses such as those generated by the spring-loaded syringe. Also, the signals clearly retained their sinusoidal form during propagation and no "forerunners"[19)] were observed. These facts lend further support to the conclusion that the vena cava is not strongly dispersive and that effects of reflections are not noticeable. Consequently, when the amplitude of a sinusoidal pressure wave induced by the pump is small, the speed of such a signal can be interpreted as the phase velocity corresponding to the frequency of the sine waves.

An example of the manner in which the wave shape can be affected by the amplitude is illustrated in Figure 14. As the peak-to-peak amplitude is increased from about 3.5 to about 10 mm Hg the signal no longer retains the form of true sine waves during propagation and the up-stroke of the sine wave steepens markedly, i.e., the rise-time becomes progressively shorter while the decay- or downstroke-time increases. This amplitude phenomenon may be attributed to a pressure dependence of the wave speed. Whenever the speed increases markedly with pressure the peaks of the sinusoidal pressure waves travel faster than the valleys. This is further illustrated in Figure 15 in which the velocities of the wave peaks and valleys are given as a function of wave amplitude. We note that for sufficiently large pressure signals the speed depends strongly on the amplitude. The vessel apparently behaves in a nonlinear fashion with respect to large pressure pulses which precludes the possibility of performing a harmonic analysis of such signals for the purpose of determining dispersion and attenuation.

The extent to which reflections can alter the apparent wave speed of an induced pressure signal is illustrated in Figure 16.

C. Waves Generated by Vibrating Piston

By generating finite trains of sine waves with the aid of the electrically driven piston it was possible to obtain extensive dispersion and attenuation data for waves in the exposed vena cava. Representative tracings of trains of different frequencies generated in this manner are illustrated in Figure 17, and 3 typical dispersion curves are given in Figures 18 and 19.

The speed of the induced sine waves ranged from 100 to 400 cm/sec for these experiments and exhibited only mild dispersion which was usually more pronounced for frequencies below 50 Hz. For frequencies between 20 and 100 Hz the wave speed generally increased with increasing frequency. The significant differences in speed from animal to animal observed in our experiments may be attributed to differences in vessel geometry and to the physiological state of the animals which includes such factors as transmural pressure, depth of anesthesia and surgical trauma. An additional cause for these variations may be the location of the transducers relative to the bifurcation or renal branches. Figure 20 illustrates the dependence of wave speed and dispersion on transducer location in the vena cava of a single animal. This behavior was investigated in 3 dogs and appears to be the result of significant variations in the local wave transmission characteristics of the vessel.

Most of the pressure waves generated with the vibrating piston had peak-to-peak amplitudes less than 20 mm H_2O and there was no significant or systematic variation in the speed of different wave peaks or valleys within the same wave train. It appears that for distension waves with amplitudes less than 20 mm H_2O and frequencies between 20 and 100 Hz the wave speed is independent of amplitude. For such small waves the vena cava behaves essentially in a linear fashion. For frequencies above 100 Hz the stroke of the vibrating piston used was too small to generate sufficiently strong pressure waves in the vena cava to allow for accurate measurements while below 20 Hz the shape of the signals deviated too much from that of true sine waves.

Data regarding the attenuation characteristics of the vena cava with respect to sinusoidal pressure waves were obtained by comparing again the peak-to-peak pressures for various frequencies at the two transducer sites. Attenuation curves corresponding to the dispersion data given in Figures 18 and 19 are illustrated in Figures 21 and 22. As in the case of the thoracic aorta [13)] the amplitude ratio A/A_o for pressure waves in the vena cava with frequencies between 20 and 100 Hz decays exponentially with distance traveled. Also, high frequency waves are dissipated much more rapidly over a given distance than low frequency waves. Knowing the wave speed and frequency we can plot A/A_o as a function of the distance traveled in wavelengths and we find again that the amplitude ratio decays in the same exponential fashion for all frequencies:

$$A/A_o = e^{-k \frac{\Delta x}{\lambda}}$$

where k is the attenuation coefficient or logarithmic decrement. Data from 6 experiments indicate that k may vary from about 1 to 2.5 for the vena cava as compared with a range from 0.7 to 1.0 for the aorta.

The vibrating piston also permitted us to investigate the effects of transmural pressure on the propagation velocity of small amplitude waves in the vena cava. Figures 23 and 24 are based on initial results and illustrate the typical relationship between wave speed and transmural pressure for signals of various frequencies. The venous pressure was increased by occluding the vena cava below the heart with a balloon catheter inserted through the right external jugular vein and positioned in the inferior vena cava between the heart and the hepatic branches. By inflating the balloon for a period of 15 to 30 seconds the venous pressure would rise from its normal value of 50 to 100 mm H_2O to as much as 200 or 300 mm H_2O. In all cases the wave speed increased with increasing transmural pressure and a marked linear relationship between the two quantities was apparent. Linear regression coefficients calculated from the data ranged from about 1.5 to 2.5 cm/sec/mm H_2O which indicates a rather strong dependence of the wave speed on pressure. Some of the pertinent results of these pressure studies are summarized in Table II.

DISCUSSION AND CONCLUSIONS

The propagation of large amplitude pressure waves induced by the spring-loaded syringe or by the pump appears to be affected by reflection interference and pronounced nonlinear phenomena. For waves with amplitudes exceeding a few mm Hg the speed increases markedly with signal amplitude and the wave front steepens during propagation. We also find an increase of the signal speed by raising the level of the unperturbed transmural pressure. Therefore the observed amplitude dependence of the speed and the steepening of the wave front can be interpreted as nonlinear phenomena attributable to the fact that the wave speed increases with pressure. Accordingly the steepening of the wave front may correspond to the early phases of the formation of a shock wave. This implies that a sufficiently strong pressure pulse may generate a shock wave in the vena cava which would manifest itself in the form of a sharp and audible sound. Clinical evidence supporting the possibility of the actual development of shock waves is established by the occurrence of the venous pistol shot.[20)]

The wave transmission data obtained with signals induced by the spring-loaded syringe yielded valuable information regarding the gross behavior of the vena cava including the effects of pressure, respiration, various drugs, vagal stimulation and ventricular fibrillation. But these data did not allow for the actual determination of the dispersion and attenuation of pressure waves because of the strong nonlinearities that were observed.

With the vibrating piston we could induce only pressure waves of small amplitude. The maximum volumetric displacement of the piston was about 200 mm^3 for frequencies below 70 Hz and decreased rapidly with increasing frequency above 80 Hz. For displacements of about 150 mm^3 the peak-to-peak pressure amplitudes were generally of the order of 20 mm H_2O at normal transmural pressure levels and at a distance of 4 cm from the piston. The frequencies and shape of the waves generated in this fashion could be controlled with a high degree of accuracy and repeatability. Only distension waves were observed; no attempts were made to record axial or torsion waves. Nonaxisymmetric waves[10,11] if present did not exhibit sufficiently strong pressure fluctuations at the transducer sites to be recognizable. This was verified by placing the two transducers at various points of the same cross section (by twisting the catheter leads) and comparing the corresponding pressure variations. The fact that both transducers recorded the same signal with zero phase shift independent of transducer position suggests that the pressure waves were axisymmetric.

From our experimental results it follows that the abdominal vena cava of anesthetized dogs does not exhibit much dispersion of distension waves with frequencies between 20 and 100 Hz. Although the dispersion may be considered as mild, it appears to vary markedly with location as indicated in Figure 20. The absence of strong dispersion of small axisymmetric distension waves is clearly in agreement with theoretical predictions.[10,12] However, the observed changes in the dispersive nature with location have not been anticipated. This may be due to the fact that the local variations in geometry and in the elastic and viscoelastic properties of the vessel wall and the surroundings are usually disregarded in theoretical analyses. It should be noted that when the wave speeds are measured over longer distances (Figure 19) the fluctuations in <u>local</u> dispersion characteristics may no longer be evident since the speed of the pressure waves between any two points of the vessel is an average measure of the mechanical properties.

As in the aorta[13] small pressure waves in the vena cava appear to decay exponentially with distance traveled in wavelengths independent of frequency. This dissipation of signal energy may be attributed primarily to viscoelastic damping in the vessel wall since theoretical considerations[12] indicate that the effects of blood viscosity are insignificant for the range of frequencies and type of waves investigated.

Furthermore, radiation of energy into the surrounding medium cannot be a major contributing factor since the vessel was exposed during the experiments.

Differences in attenuation are also observed in various segments of the vena cava of each animal studied, but this aspect has not yet been systematically investigated. Before any definitive statement can be made regarding the accuracy of a mathematical model postulating the relationship between the dispersion, attenuation and mechanical properties of the vessel, we need additional information on the wave transmission characteristics of the vena cava. For example, the dispersion and attenuation of axial and torsion waves would allow for the evaluation of the degree of anisotropy in the elastic and viscoelastic behavior of the vessel wall.

Although the wave velocity itself can serve as a measure of the distensibility of a blood vessel, it is commonly used to estimate the effective Young's modulus of the vessel wall from the Moens-Korteweg equation:

$$c^2 = \frac{Eh}{2\rho_f a}$$

where c is the signal velocity, ρ_f the density of the blood, h/a the thickness-to-radius ratio and E the effective Young's modulus. Recent theoretical studies[12,21,22] indicate that the Moens-Korteweg equation constitutes a meaningful approximation for the speed of low frequency pressure waves for parameter values associated with large veins such as the vena cava. Applying this formula we have computed the effective Young's modulus of the exposed abdominal vena cava from data obtained with the vibrating piston. For transmural pressures between 75 and 100 mm H_2O, signal frequencies from 20 to 100 Hz and $\frac{a}{h}$ values between 30 and 45, the effective Young's modulus ranges from about 5×10^5 to 5×10^6 dynes/cm^2. These values for E_{eff} for the vena cava wall are of the same order as those found for the aorta.[8] The distensibility of the veins however is much higher because of their large values for the radius to wall thickness ratio.

An even greater range of values for the elastic modulus would be obtained from the speeds of large pressure waves induced by the spring-loaded syringe or the pump. However, the actual significance of these data is questionable in the light of the observed nonlinear behavior and apparent reflection phenomena. In fact, even the effective Young's modulus derived from the speed of small amplitude pressure waves is relevant only when referred to the conditions under which the wave speeds were measured, which includes the respiratory phase and the transmural pressure. The relatively large variations in the effective Young's modulus from animal to animal at comparable transmural pressures is probably due to differences in the physiologic state of the animals.

The wave transmission approach presented here will be useful in the continued study of the distensibility of parts of the vascular capacitance system and its regulation. The suggested use of transcutaneous ultrasound sensors should prove valuable in the determination of the relative significance of different components of the reservoir system and in the evaluation of the range of variation in the distensibility of blood vessels in different individuals under normal and pathologic conditions as well as in states of stress.

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TABLE I

EFFECT OF RESPIRATION ON THE SPEED OF PRESSURE PULSES IN THE ABDOMINAL VENA CAVA INDUCED BY THE INJECTION OF 0.5 CM^3 SALINE WITH A SPRING-LOADED SYRINGE

Experiment	P_v mm H_2O	Δx cm	n	Velocity, cm/sec Inhale	Resting	$V_{in} - V_r$ cm/sec
36	150	4.4	6	593	490	103
37	75	9.4	4	582	459	123
47	120	5.8	6	591	383	208
63	140	3.6	7	626	547	79
67	90	4.9	10	437	385	52
68	90	4.5	13	495	349	146
73	60	4.4	4	776	739	37
93	50	5.1	3	637	564	73
123	-	6.0	5	575	480	95
132	120	6.0	2	533	425	108

P_v transmural pressure

Δx distance between transducers

n number of wave speed measurements during inhalation and corresponding resting phase

V_{in} wave speed at inhalation

V_r wave speed during resting phase

TABLE II

EFFECTS OF TRANSMURAL PRESSURE* ON THE SPEED OF SMALL AMPLITUDE SINE WAVES IN THE ABDOMINAL VENA CAVA

Experiment	Δx cm	Freq. Hz	P_v*, mm H_2O Initial	Max.	Velocity, cm/sec Initial	Max.	$\frac{\Delta V}{\Delta P_v}$ $\frac{cm/sec}{mm\ H_2O}$
294	4.0	30	60	200	230	500	1.59
	4.0	30	60	235	213	597	1.75
	4.0	50	65	235	265	635	2.13
	4.0	40	60	235	234	615	1.85
	3.0	50	70	240	256	612	1.97
	3.0	50	60	210	236	638	2.34
298	4.0	35	60	155	239	406	1.62
319	3.0	40	45	195	200	643	2.43
	3.0	55	97	215	273	486	1.69

* The transmural pressure was increased by occluding the vena cava for 15 to 30 seconds with a balloon catheter positioned near the diaphragm.

ΔP_v change in transmural pressure

ΔV change in wave speed associated with pressure variation ΔP_v

$\frac{\Delta V}{\Delta P_v}$ linear regression coefficient

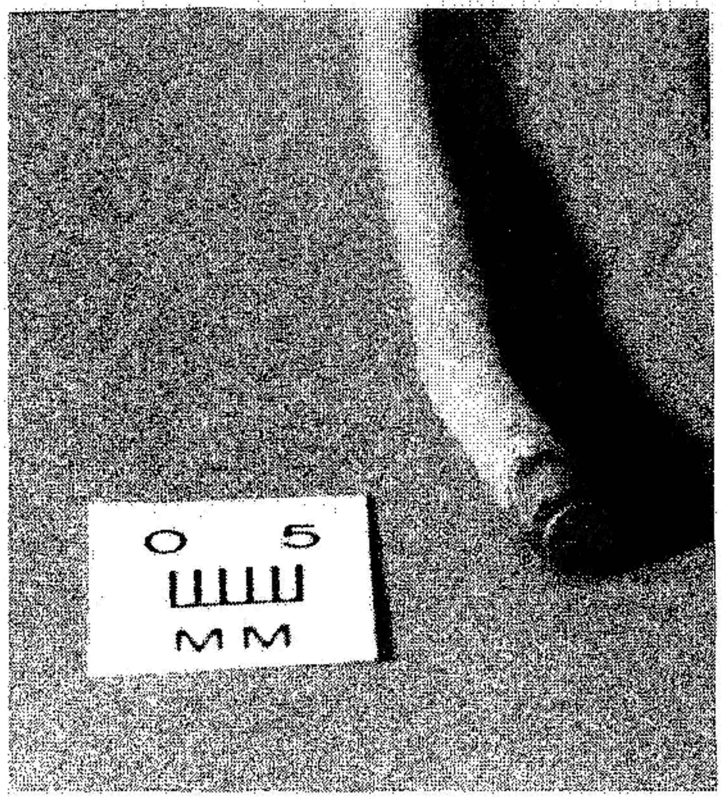


Figure 1. Schaevitz-Bytrex pressure transducer model HFD-5 adapted for use as a catheter-tip manometer. The shielded leads and the venting tube are protected by a heat-shrinkable plastic tube forming a leak-proof flexible catheter. The pressure cell itself is gold plated to reduce the corrosive effects of body fluids. The manometer is sufficiently sensitive to resolve pressure fluctuations of less than 1 mm H_2O.

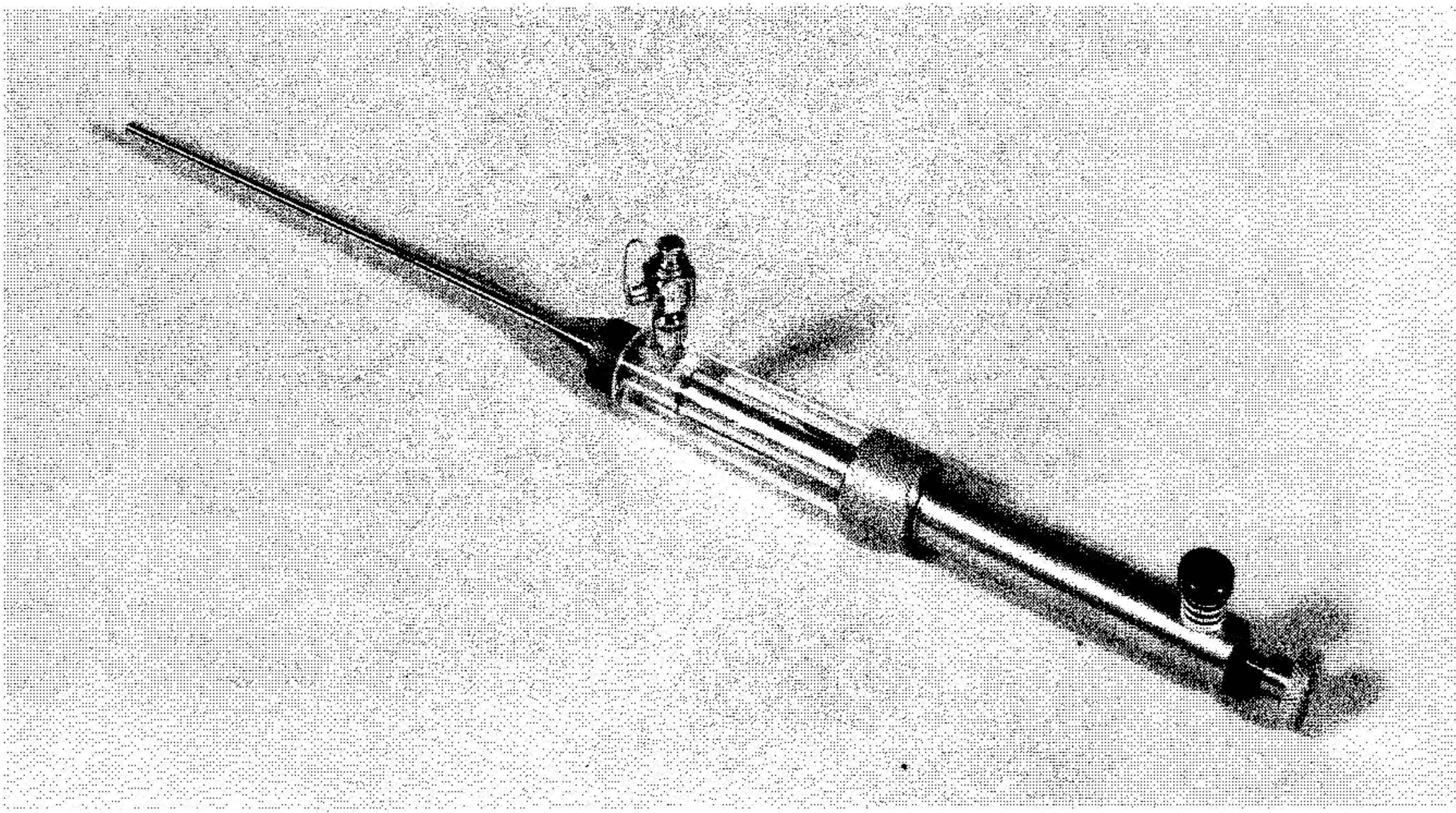


Figure 2. Spring-loaded syringe used to generate large amplitude pressure pulses in blood vessels. A volume of 0.5 or 1.0 cm^3 of blood or saline is injected in 10 to 20 msec, depending on the stiffness of the spring and the length and diameter of the cannula. Amplitudes of the pressure pulses ranged from 30 to 225 mm Hg.

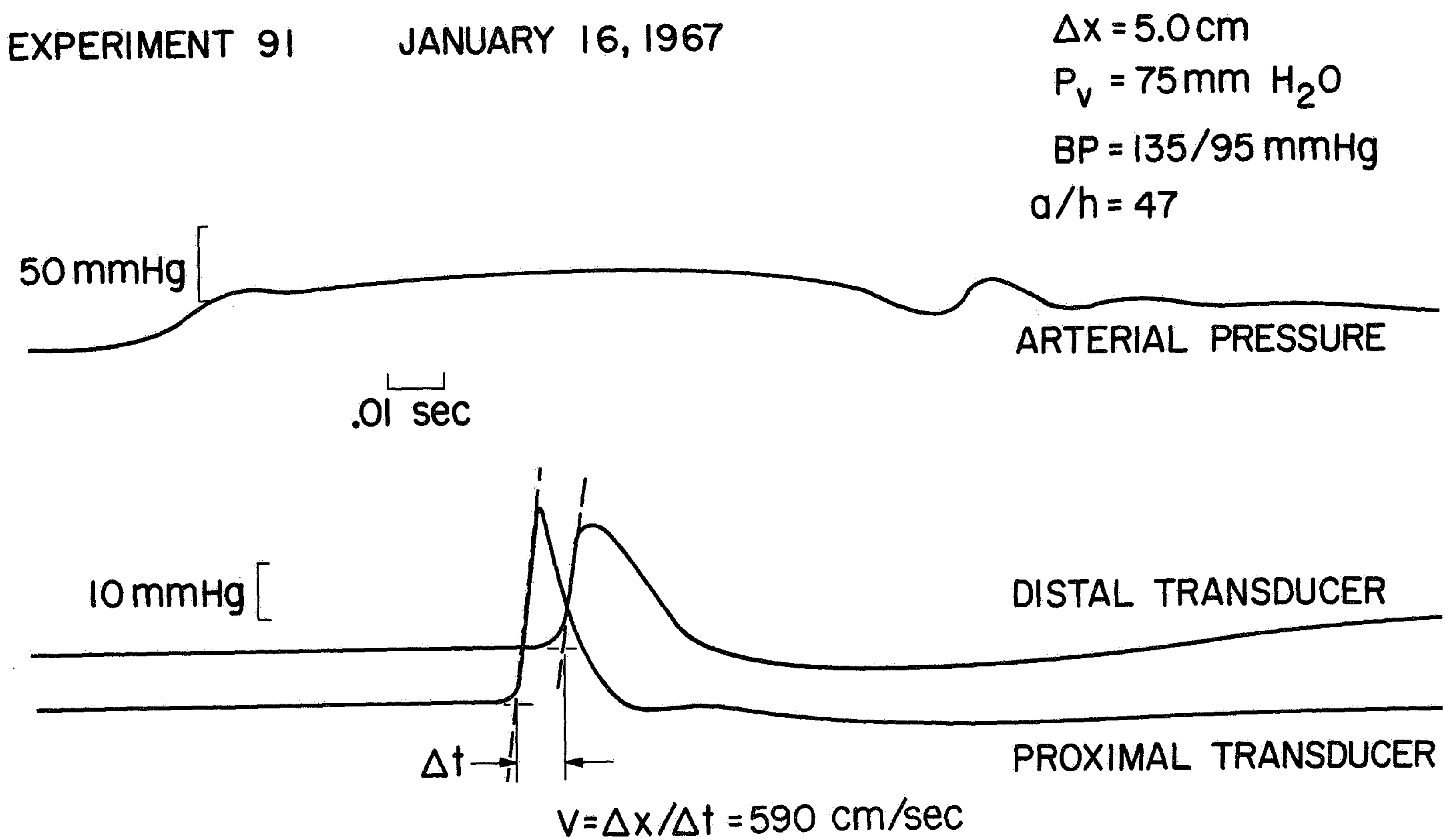


Figure 3. Representative tracings of pressure fluctuations in the carotid artery and in the abdominal vena cava of an anesthetized dog during the transmission of a pressure pulse induced by the injection of 0.5 cm^3 of saline into a femoral vein. In this instance the arterial blood pressure (BP) was 135/95 mm Hg while the unperturbed venous pressure (P_V) was 75 mm H_2O. The distance between the two catheter-tip manometers (Δx) was 5.0 cm and the radius to wall thickness ratio (a/h) for the vena cava was 47. The relative scales for the venous and arterial pressure tracings and time are indicated. Each of the two venous pressure curves has a different zero point since they were separated for illustration purposes. The signal speed v is determined from the transmission time Δt of a characteristic point of the wave front over the distance Δx. The characteristic point is defined by the intersection of the tangents to the inflection point of the wave front and the unperturbed venous pressure.

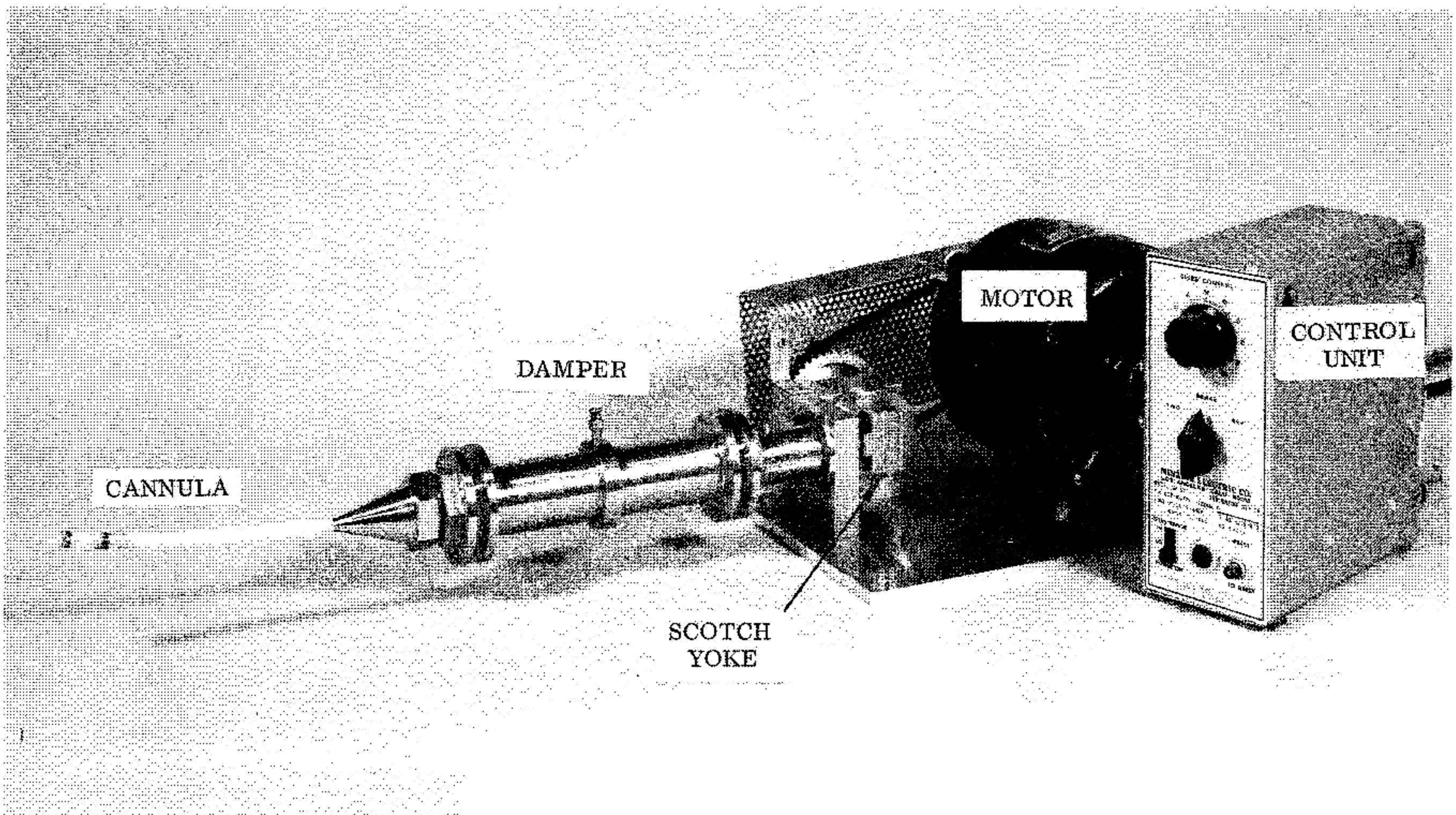


Figure 4. Pump with frequency control unit used to generate sinusoidal pressure waves in blood vessels. The waves are induced by a reciprocating piston connected to a 1/8 hp D.C. motor by means of a timing belt and scotch yoke to enforce pure sinusoidal motions of the piston. The stroke volume and frequency of the pump can be varied from 0 to 5.0 cm^3 and 0 to 40 Hz respectively. To eliminate high frequency pressure fluctuations due to inherent structural vibrations of the system, a damping device was installed between the cannula and the pump cylinder. Damping is provided by an air cushion that is separated from the fluid by a rubber membrane and whose pressure can be varied to optimize the damping characteristics. Since no check valve is used the net flow of the pump is zero.

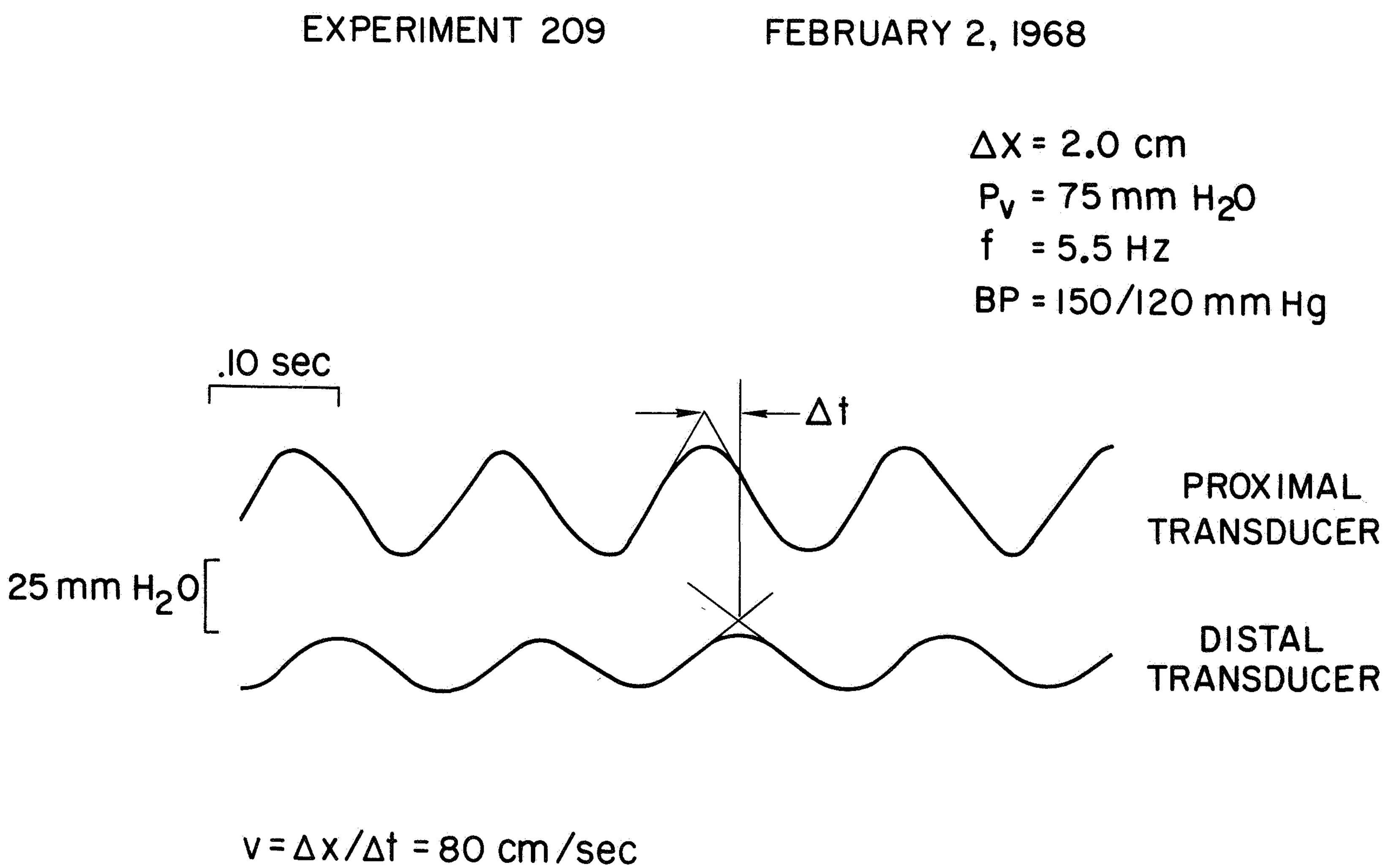


Figure 5. Trains of sinusoidal pressure waves in the abdominal vena cava produced by the pump. The transmission time is determined graphically as illustrated. As characteristic point we have chosen the intersection of the tangent to the sine wave at two successive inflection points.

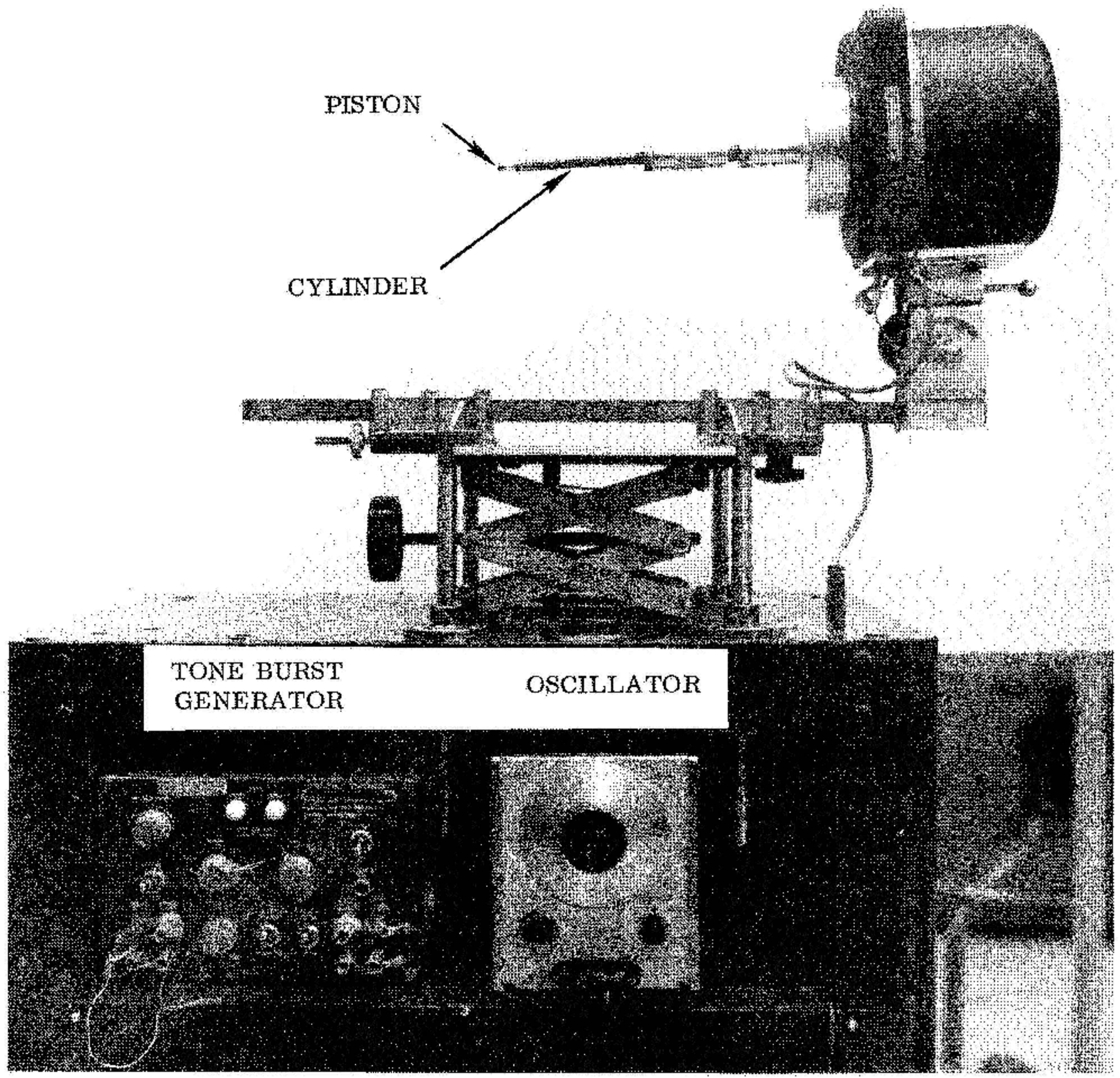


Figure 6. Electrically driven piston with tone burst generator and oscillator. The piston is enclosed in a brass sleeve which is inserted directly into a femoral vein and positioned 1 to 5 cm below the bifurcation. The piston diameter is 5.5 mm and its maximum travel is about 6.5 mm for frequencies between 20 and 100 Hz.

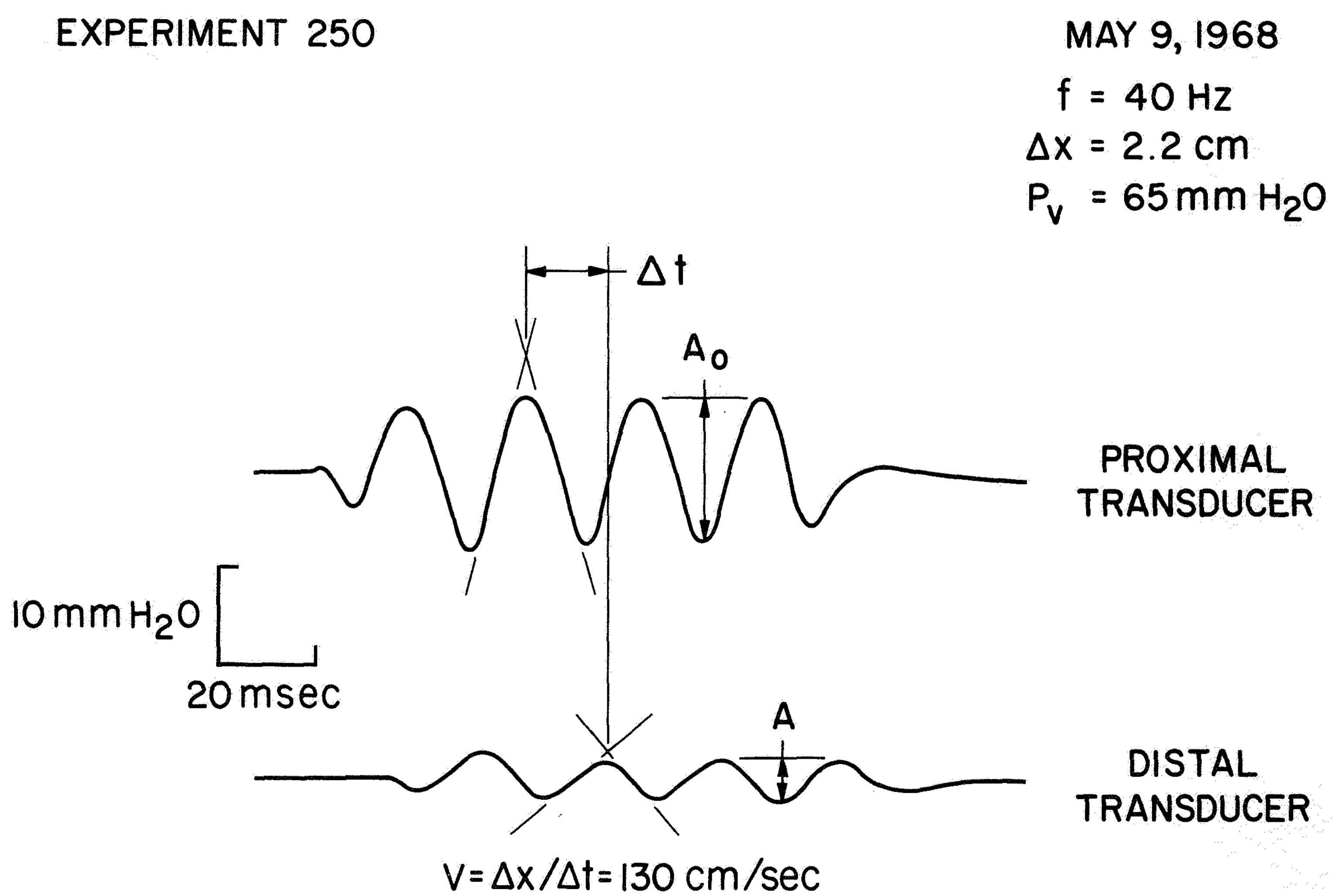
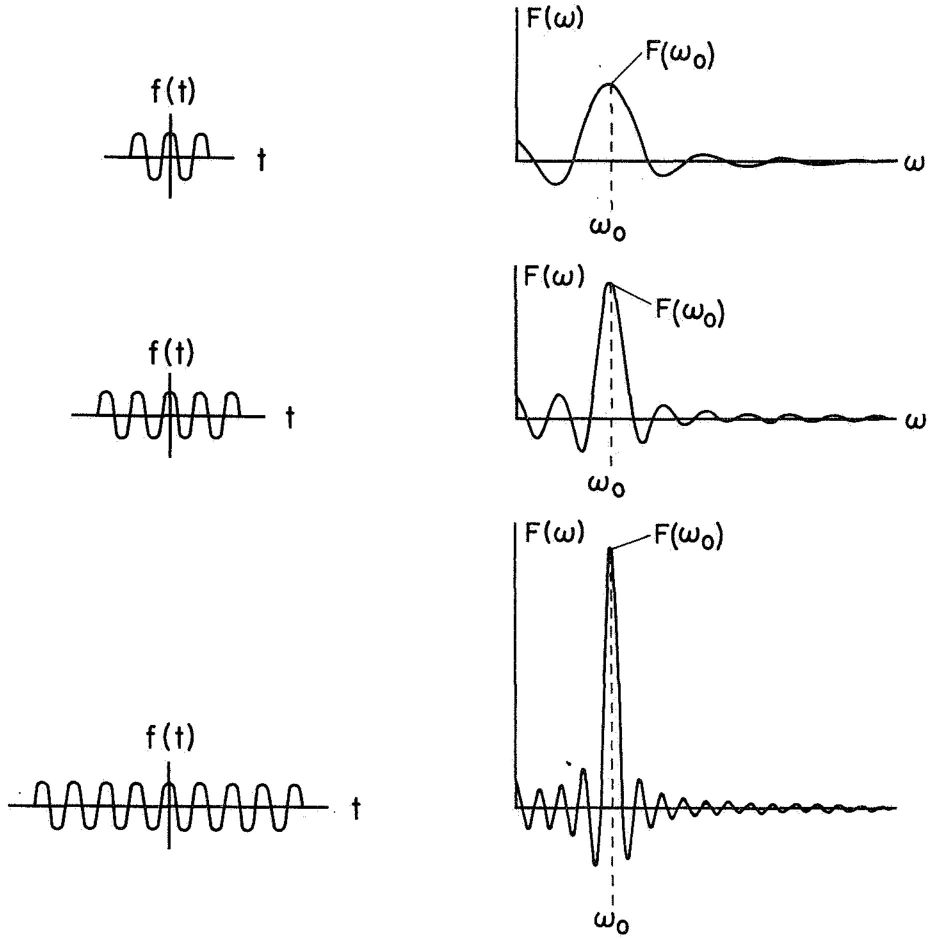


Figure 7. Recording of a finite train of sine waves produced by the vibrating piston. The transmission time and amplitude are measured as indicated. Attenuation is defined by the peak-to-peak amplitude ratio A/A_o. Note that the amplitude A_o at the proximal transducer is of the order of 15 mm H_2O.



$$f(t)=\frac{1}{2\pi}\int_{-\infty}^{\infty} F(\omega)\,e^{i\omega t}\,d\omega \qquad F(\omega)=\int_{-\infty}^{\infty} f(t)\,e^{-i\omega t}\,dt$$

$$f(t)=\begin{cases}\cos(\omega_0 t), & |t|<\frac{\tau}{2}\\ 0, & |t|>\frac{\tau}{2}\end{cases} \qquad F(\omega)=\frac{\tau}{2}\left[\frac{\sin\left(\frac{\omega-\omega_0}{2}\right)\tau}{\left(\frac{\omega-\omega_0}{2}\right)\tau}+\frac{\sin\left(\frac{\omega+\omega_0}{2}\right)\tau}{\left(\frac{\omega+\omega_0}{2}\right)\tau}\right]$$

Figure 8. Fourier transform of finite trains of sine waves. The uppermost pair of graphs illustrates a train of 2 1/2 sine waves of circular frequency ω_0 (left) and the corresponding Fourier transform $F(\omega)$ (right). The pairs of graphs in the center and at the bottom correspond respectively to trains of 4 1/2 and 8 1/2 sine waves of the same circular frequency ω_0. In each case the Fourier transform $F(\omega)$ has a distinct maximum at ω_0. With increasing length of the train, $F(\omega)$ peaks more sharply at $\omega = \omega_0$, and $F(\omega_0)$ dominates the spectrum in increasing proportion. In the limiting case of an infinitely long train $F(\omega_0)$ would be infinite at $\omega = \omega_0$ and zero everywhere else. The progressive dominance of $F(\omega_0)$ with increasing length of the sine wave train implies that in a mildly dispersive medium such signals should have a speed of propagation which is a close approximation to the phase velocity corresponding to ω_0.

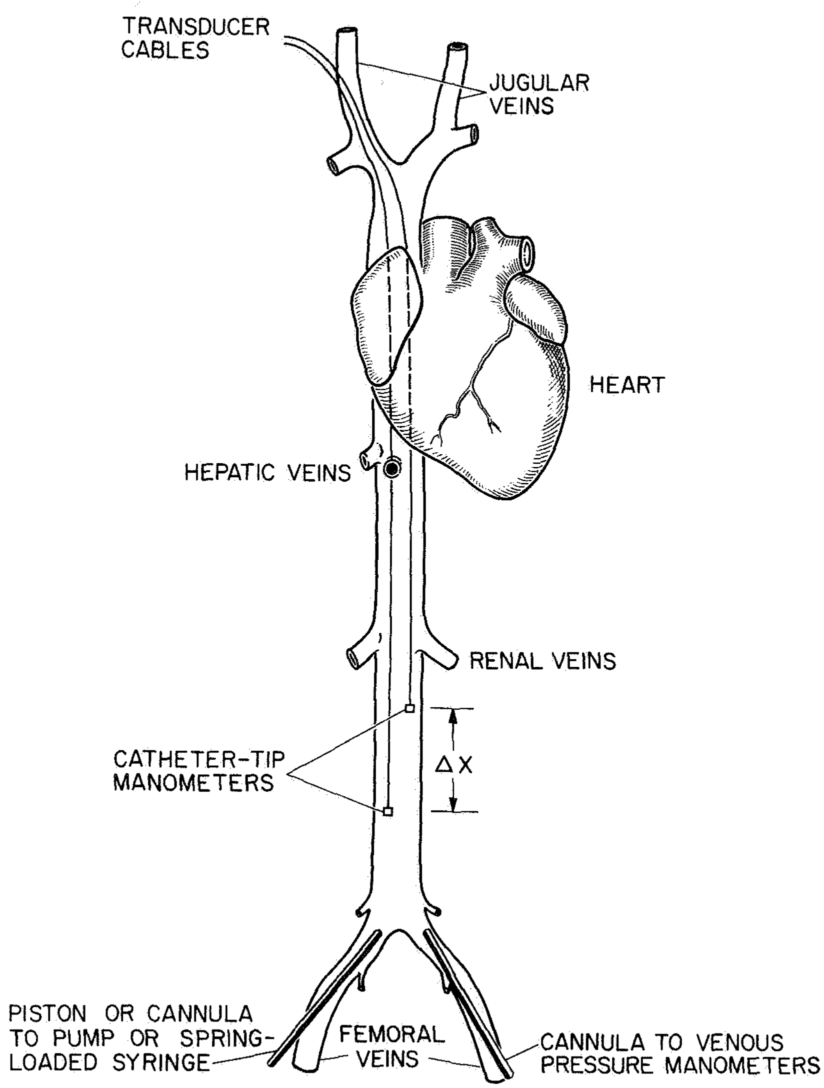


Figure 9. General experimental arrangement for studying the transmission characteristics of artificially induced pressure waves in the abdominal vena cava of anesthetized dogs. Three types of wave generators were employed in the investigation: spring-loaded syringe, pump and vibrating piston. A Statham P23BB pressure transducer and a water manometer were used to monitor venous pressure.

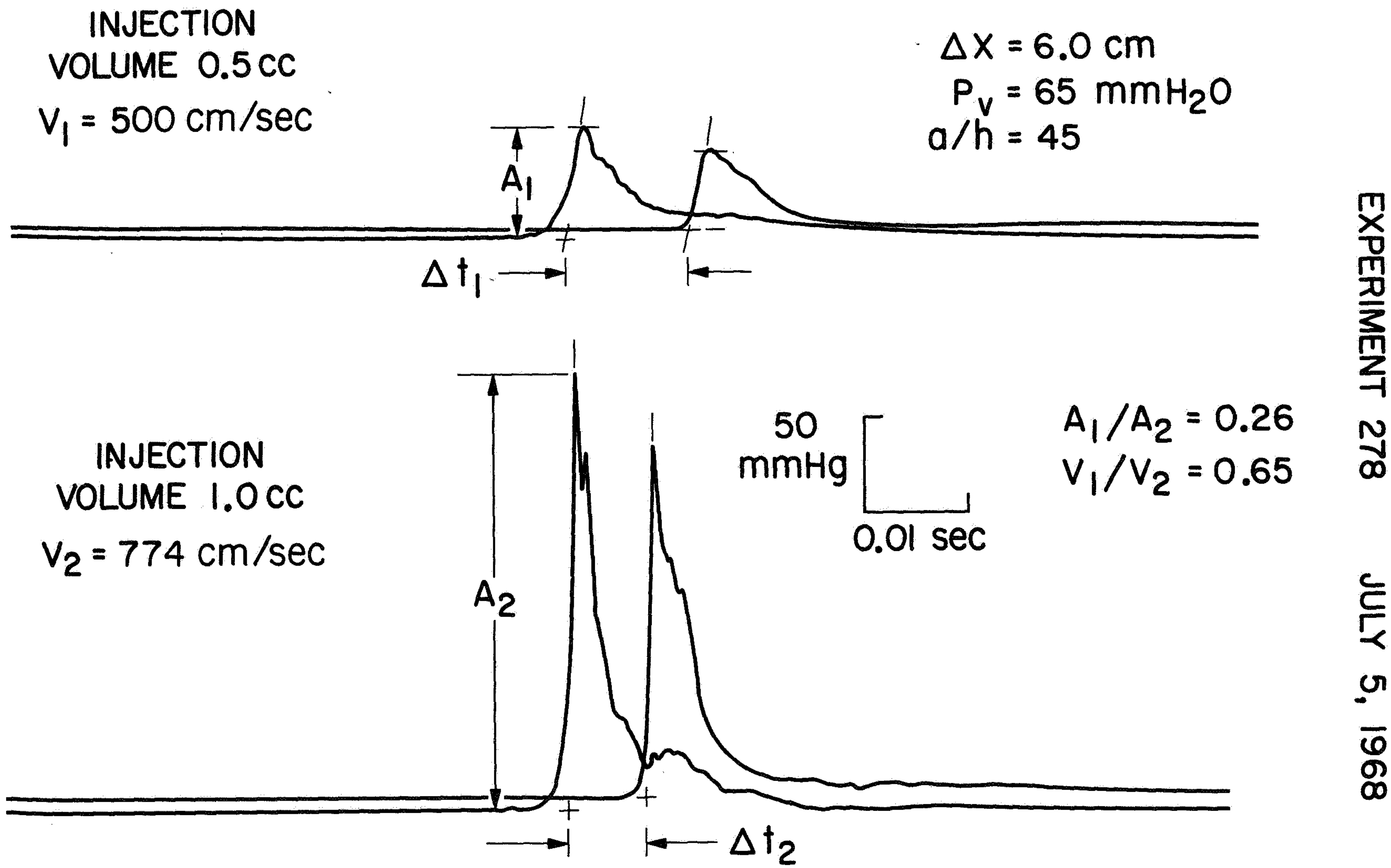


Figure 10. Variation in wave speed with signal amplitude for pulses produced in the abdominal vena cava by a spring-loaded syringe. Volume injections of 0.5 and 1.0 cm^3 produced signal amplitudes of approximately 60 and 225 mm Hg respectively. The marked increase in wave speed with signal amplitude indicates that the vena cava exhibits nonlinear behavior with respect to large pressure signals.

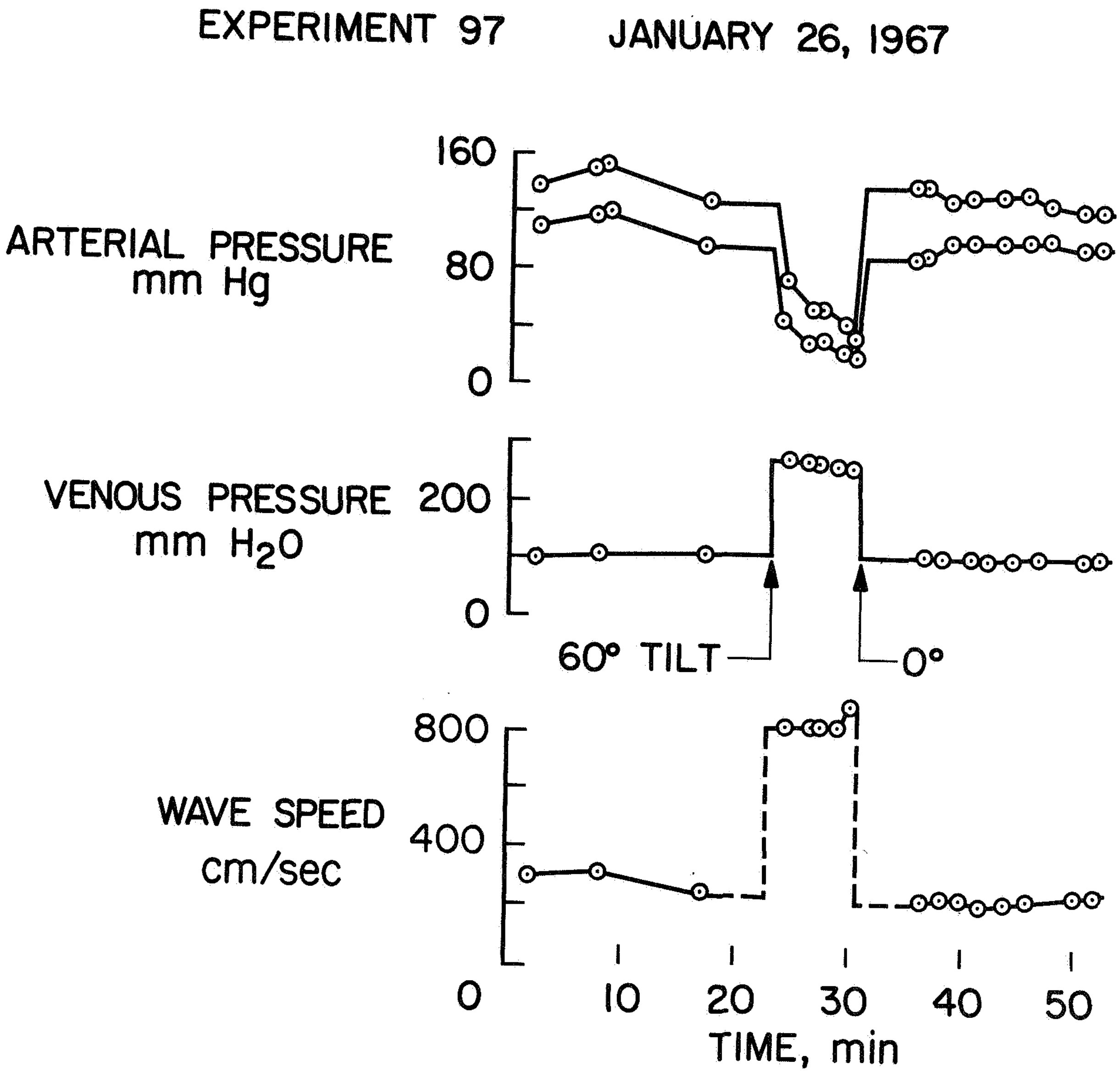


Figure 11. Typical example of the effects of tilting on the arterial and venous pressures and the speed of large pressure pulses in the abdominal vena cava. Arterial pressure was measured in the carotid artery and venous pressure in the abdominal vena cava at the bifurcation. Pressure signals were produced by the injection of 0.5 cm^3 of warm saline into a femoral vein. Tilting the animal immediately produced an elevation of the venous pressure and a substantial increase in the speed of large amplitude pressure signals. While the animal was kept at 60^o tilt for a period of 8 minutes the arterial pressure continuously decreased while the wave speed and venous pressure exhibited no major changes. Data obtained 4 minutes after returning the animal to its original position do not deviate substantially from the pre-tilt values.

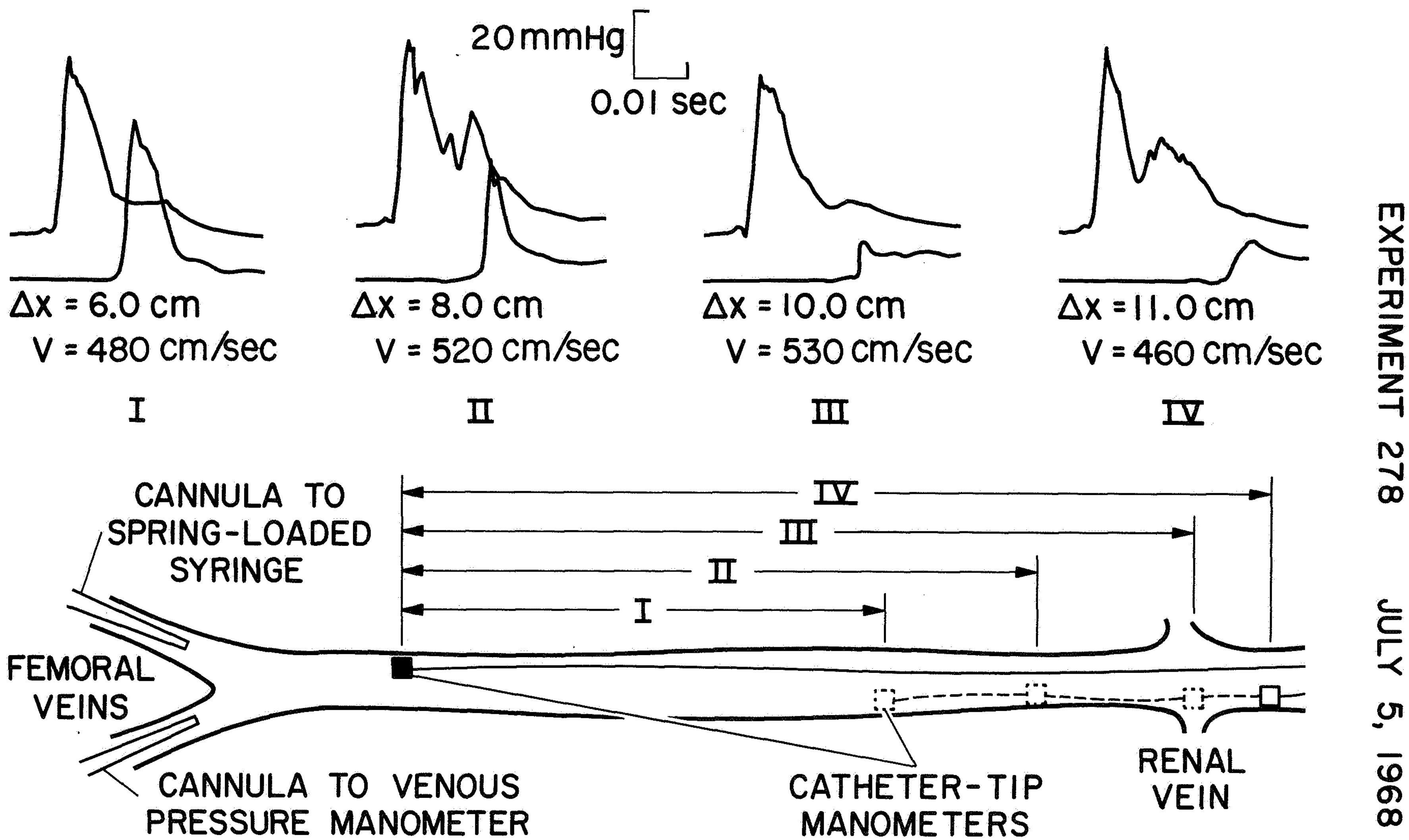


Figure 12. Variations in the shape and speed of pressure pulses at different locations in the abdominal vena cava. The distance between transducers Δx was systematically increased from 6.0 to 11.0 cm by retracting the distal transducer. Typical tracings of pressure waves induced by the spring-loaded syringe and recorded at four different relative locations (I-IV) are illustrated. In each case the maximum amplitude of the pulse at the proximal transducer is about the same while, as expected, the amplitude at the distal site progressively decreases with distance traveled as a result of dissipative mechanisms. Differences in the wave speed v indicate variations in the distensibility of the vena cava which are attributed to either geometrical changes or to variations in the elastic properties of the vessel wall. Variations in the wave shape at the proximal transducer may be due to slight alterations in the orientation of the cannula at the point of injection.

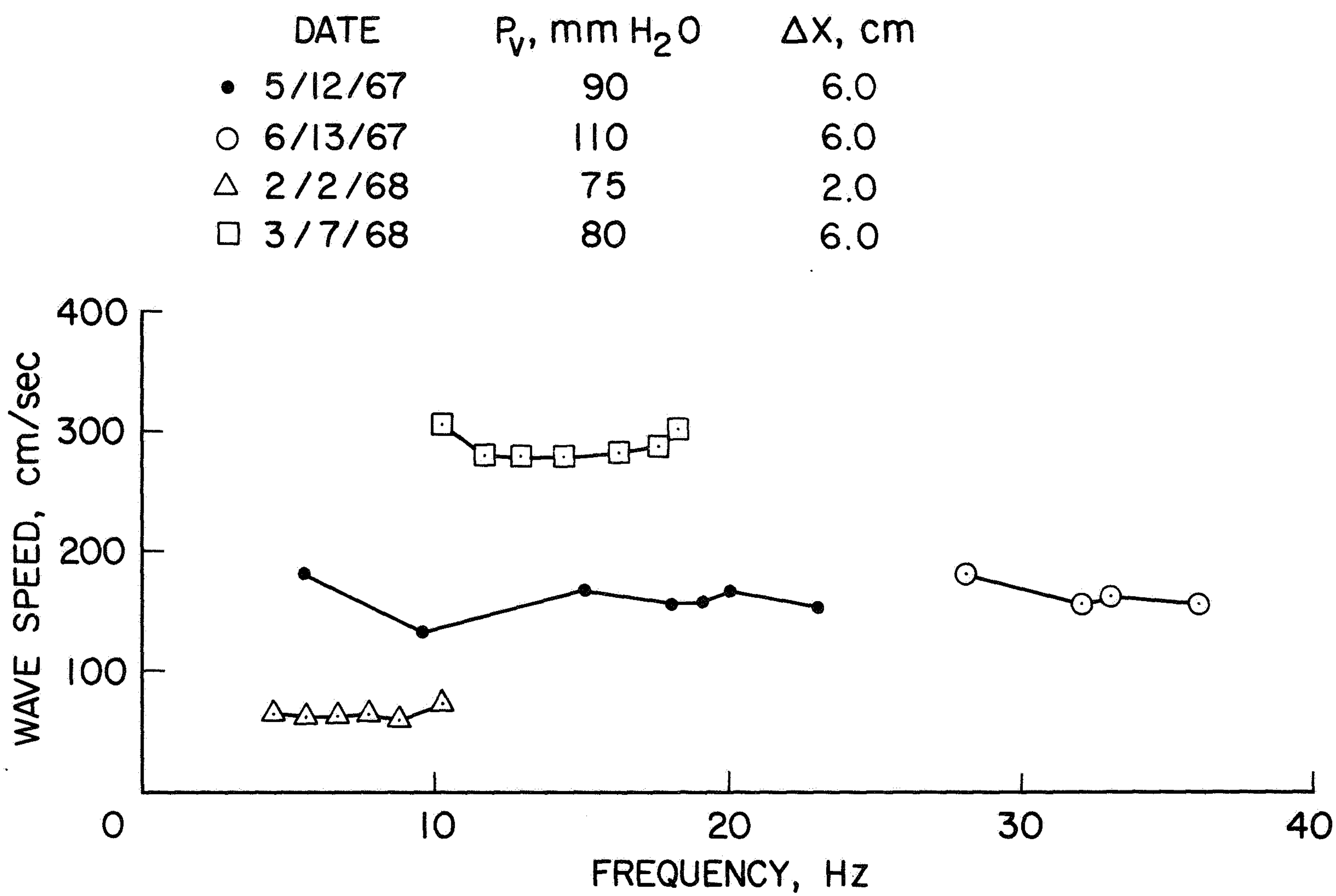


Figure 13. Representative dispersion curves for continuous pressure waves produced by the pump. While the wave speed differs considerably from animal to animal, there is only mild variation of the wave speed with frequency. Each data point represents an average of 9 to 12 speed measurements in a series of wave trains induced during the resting phase of the respiratory cycle. The peak-to-peak amplitude of the waves at the proximal transducer was 2 to 4 mm Hg.

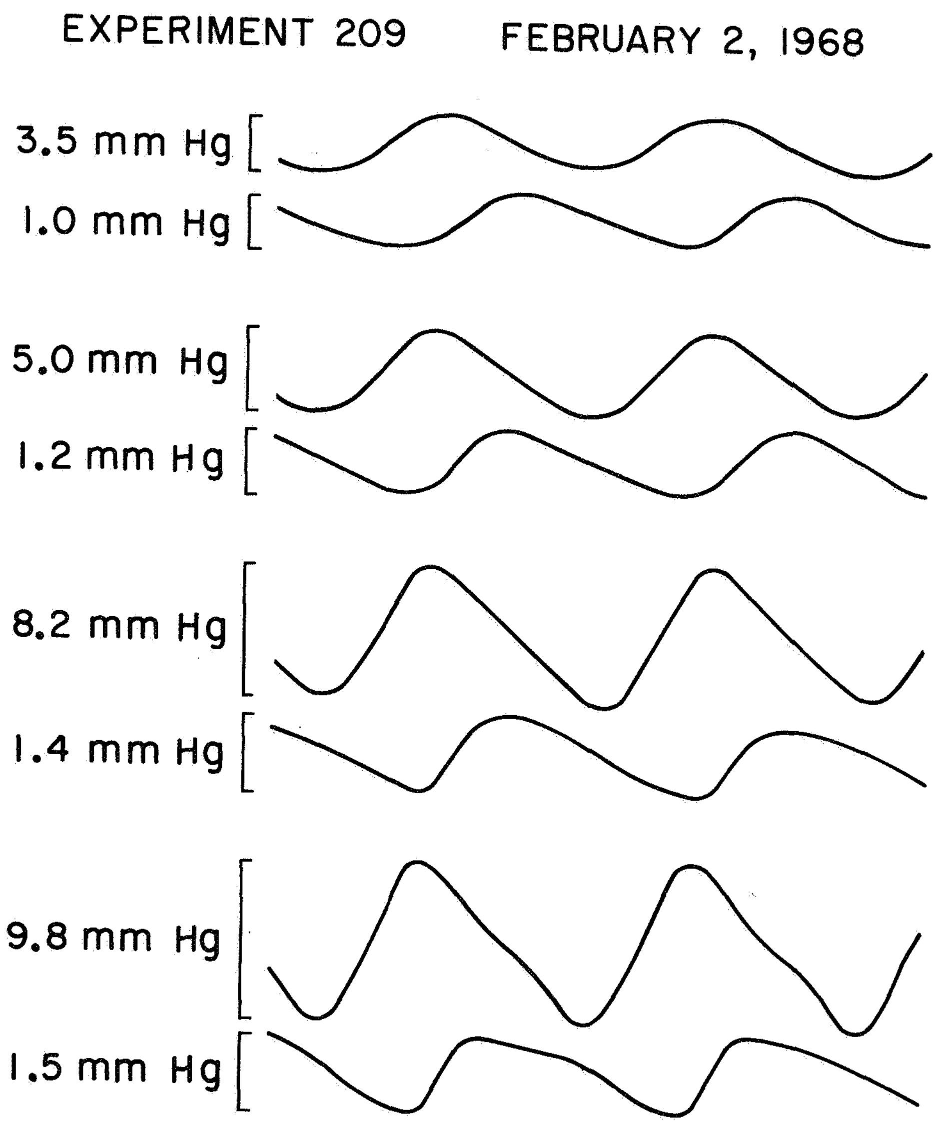


Figure 14. Changes in wave shape due to increasing signal amplitude. The four pairs of tracings illustrated represent the pressure fluctuations induced by the pump for different stroke volumes at two fixed stations in the abdominal vena cava. The distance between the transducers was 4.0 cm and the frequency was 7.5 Hz. As the stroke volume of the pump is successively increased the peak-to-peak amplitude at the proximal site rises in this case from 3.5 to 9.8 mm Hg and the pressure waves become increasingly non-sinusoidal. One can readily discern a relative steepening of the wave front with respect to the phase of diminishing pressure as the amplitude of the wave increases. Such a steepening of the wave front could, for example, be expected whenever the wave speed increases markedly with pressure.

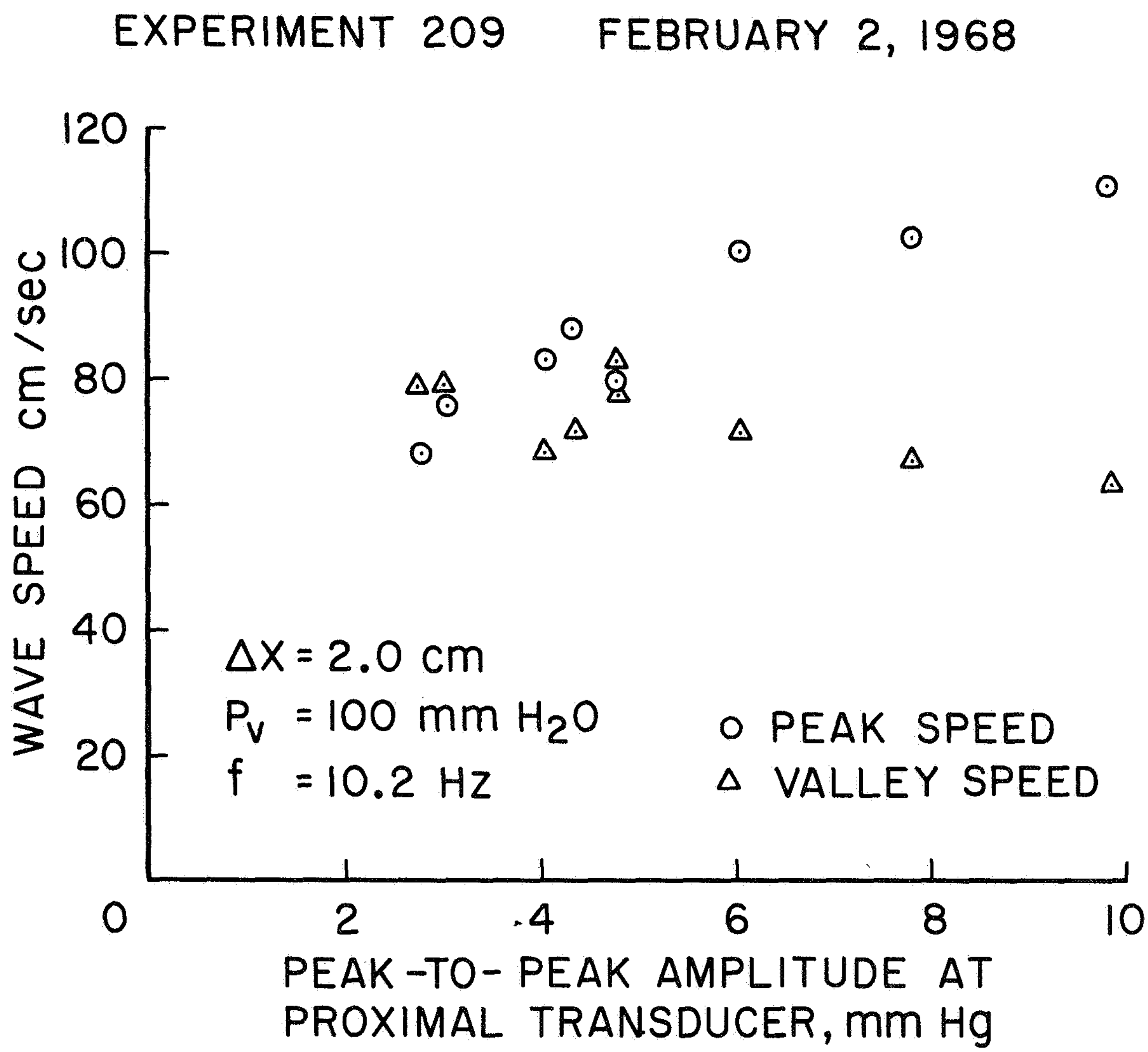


Figure 15. The effect of a large signal amplitude on the speed of the peaks and valleys of near sinusoidal pressure waves produced by the pump. The steepening of the wave front with increasing signal amplitude illustrated in Figure 14 can be effectively demonstrated by plotting the speeds of the wave peaks and valleys as a function of the amplitude. For peak-to-peak amplitudes larger than about 5 mm Hg the speed of the wave peaks is noticeably greater than that of the valleys and the difference increases progressively with amplitude. Each data point corresponds to 9 to 12 measurements of the transmission time as described in Figure 5.

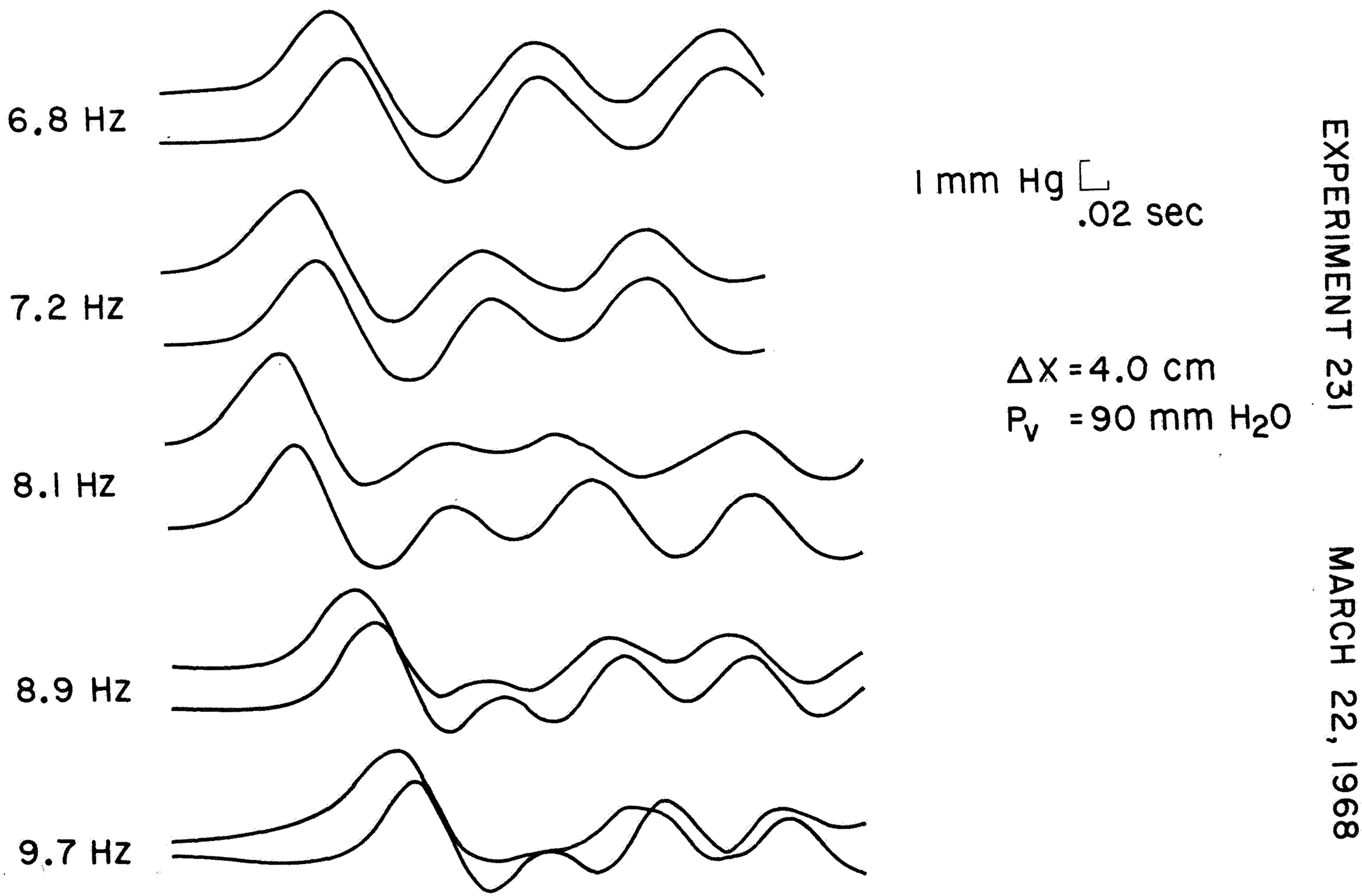


Figure 16. Indication of reflection interference associated with the propagation of large amplitude pressure waves produced by the pump. Each of the five pairs of tracings was obtained for identical transducer locations. Two possible manifestations of reflections can be discerned from the tracings. At 6.8 Hz the first peak and valley appear to arrive at the distal transducer site with a definite time lag. However, the second peak seems to reach both transducers simultaneously. As the frequency is increased the relationship between the wave patterns beyond the first peak changes dramatically. For example, the second peak completely disappears at the proximal transducer for 9.7 Hz.

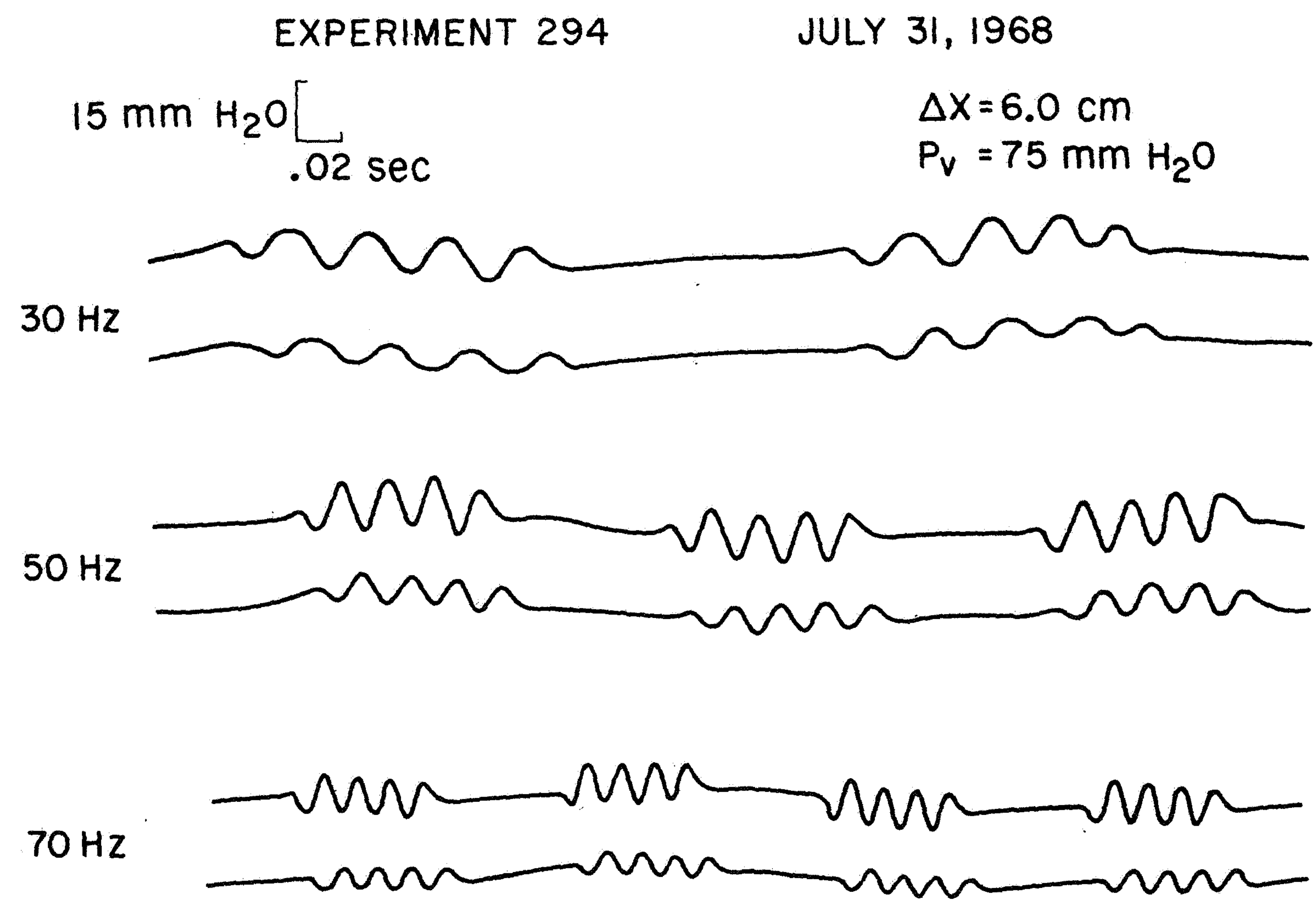


Figure 17. Sample tracings of finite trains of sinusoidal pressure waves of different frequencies generated by a vibrating piston and recorded in the vena cava. Each of the tracings has a different zero point since they were separated for illustration purposes. Note that the signals are attenuated during propagation but retain their sinusoidal character for all frequencies.

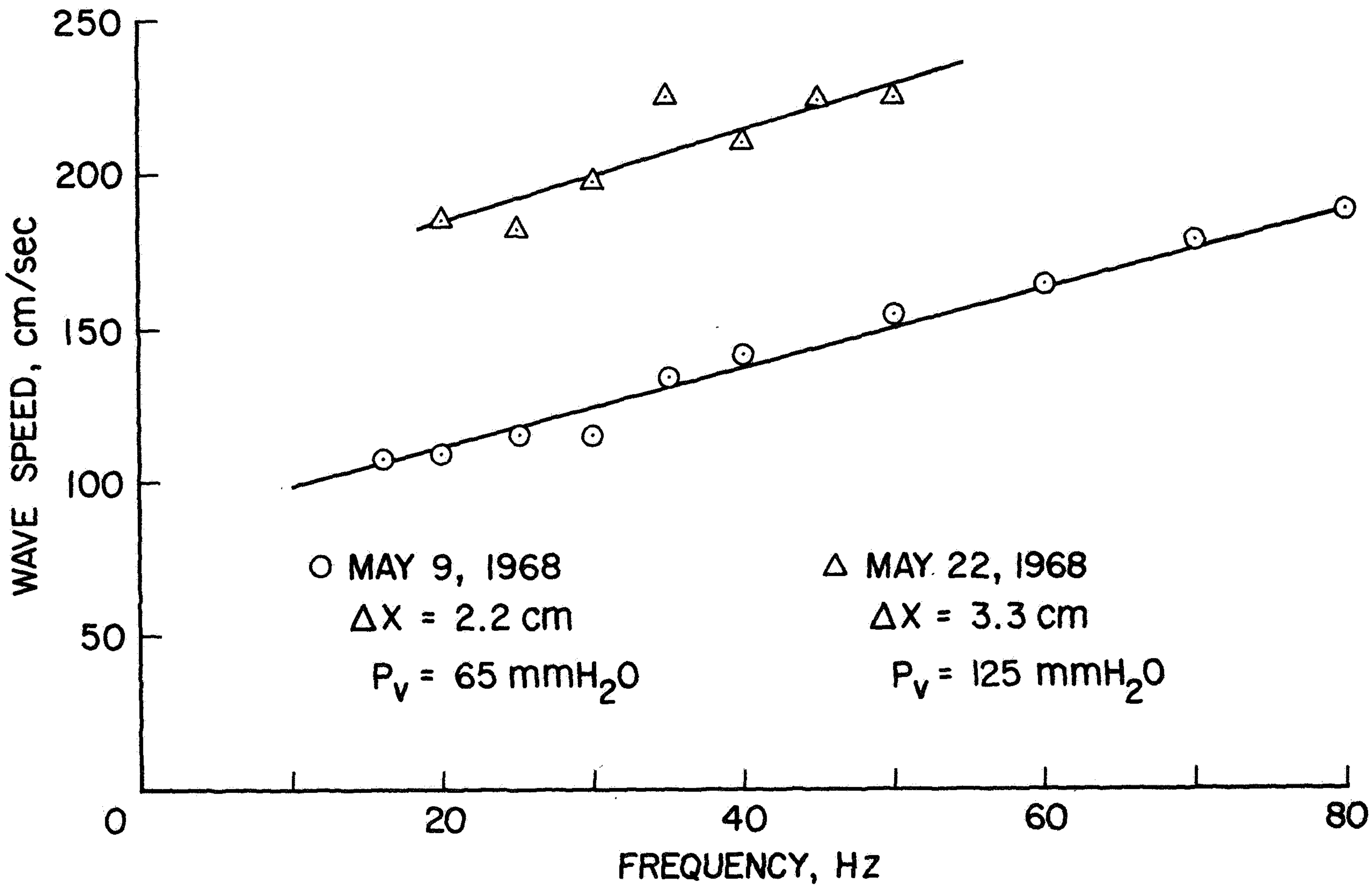


Figure 18. Representative dispersion curves for small pressure waves produced by the vibrating piston during the resting phase of the respiratory cycle. The amplitude of the signals was less than 20 mm H_2O and each data point represents an average of 9 to 12 wave speed measurements from recordings taken over a period of less than 5 seconds.

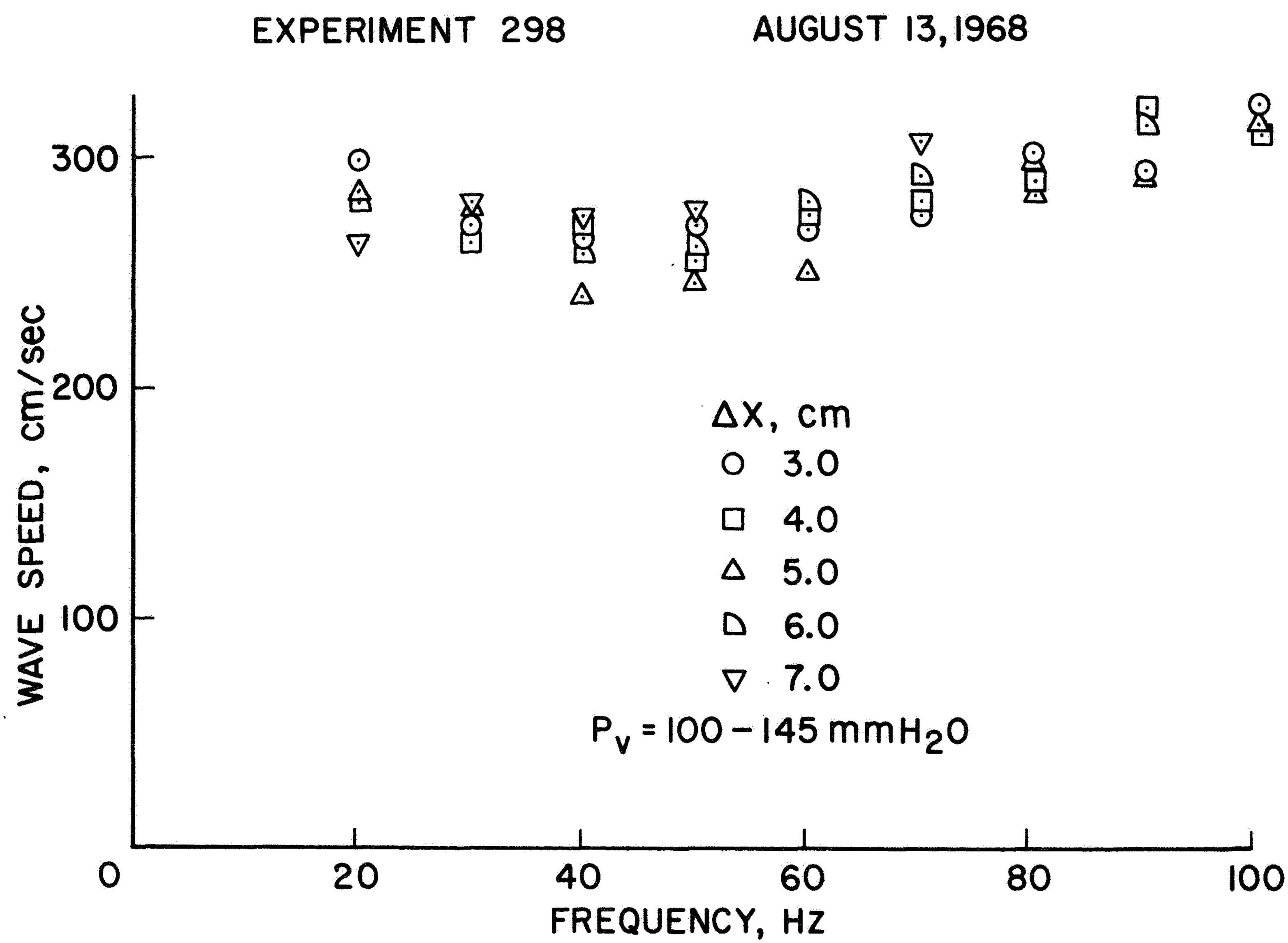


Figure 19. Dispersion of small pressure waves produced by the vibrating piston for various locations of the distal transducer and fixed position of the proximal transducer.

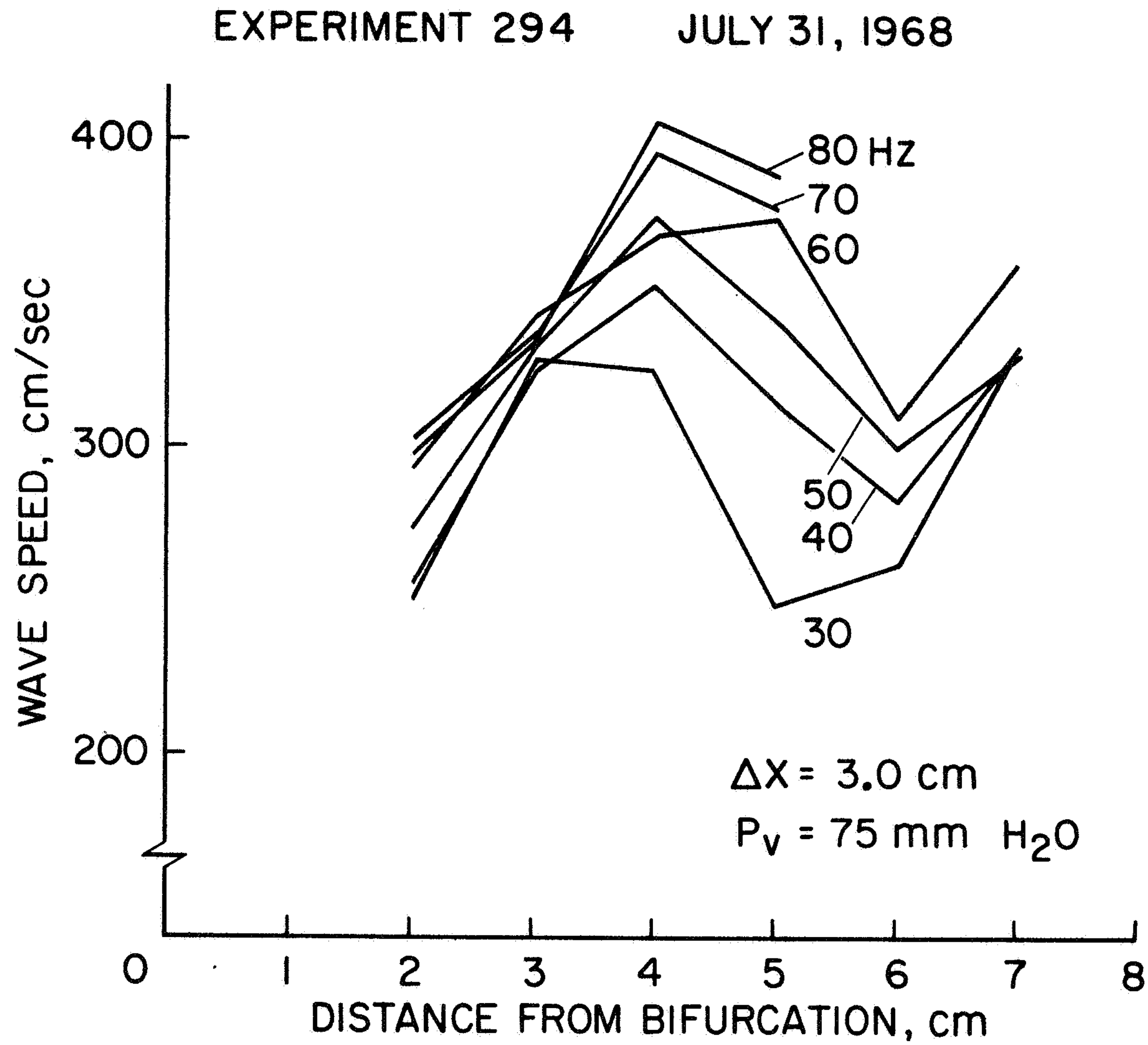


Figure 20. Variation in the speed of small pressure waves with frequency and transducer location. The distance between transducers Δx was constant while their position relative to the bifurcation was systematically changed. Note the marked variations in wave speed and dispersion with distance from the bifurcation.

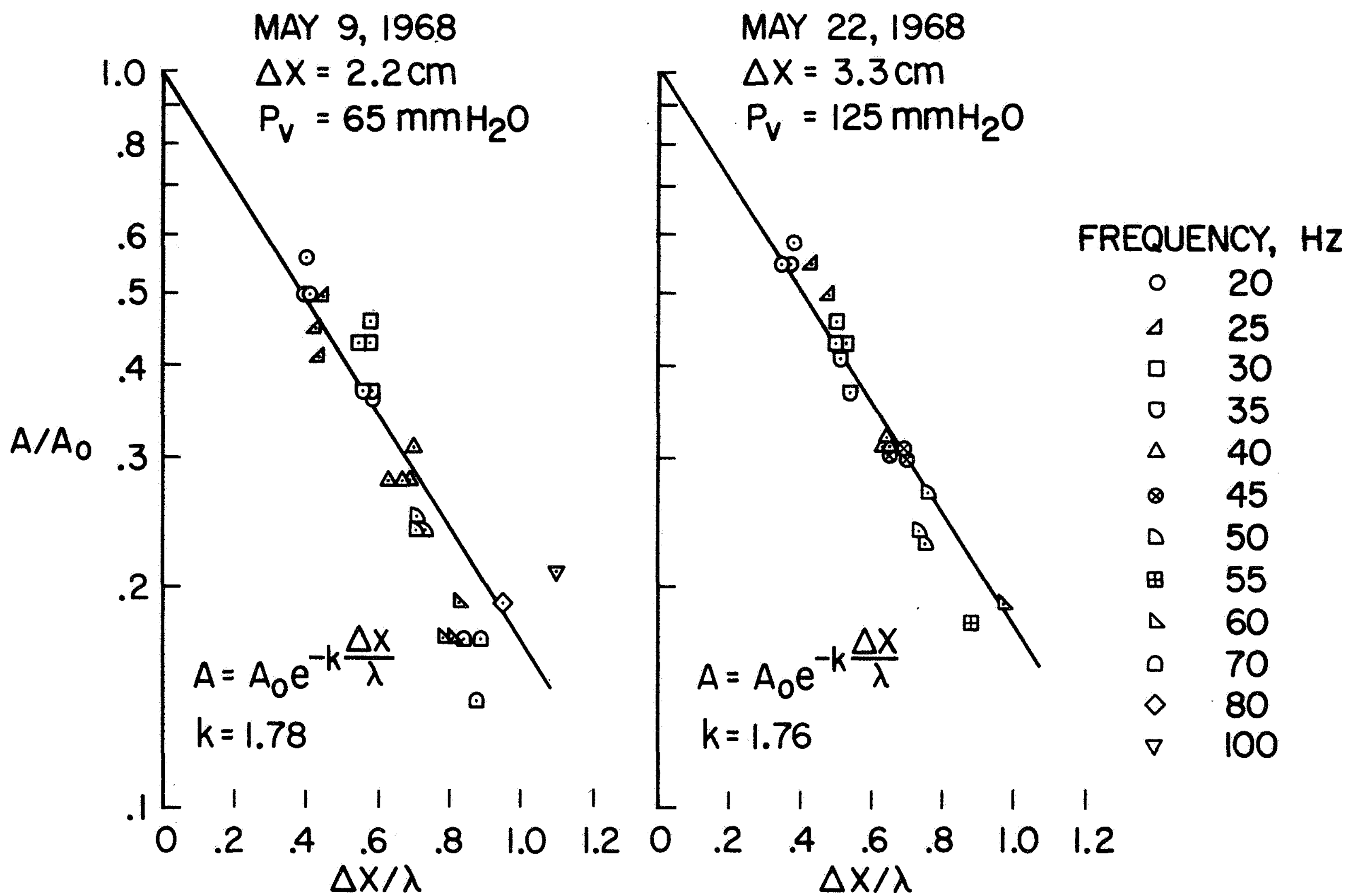


Figure 21. Attenuation curves corresponding to the dispersion data given in Figure 18. The amplitude ratio A/A_o is plotted on logarithmic scale vs. the distance traveled by the signal in terms of wave length. The logarithmic decrement k has been calculated in each case. The attenuation pattern and logarithmic decrement are independent of frequency.

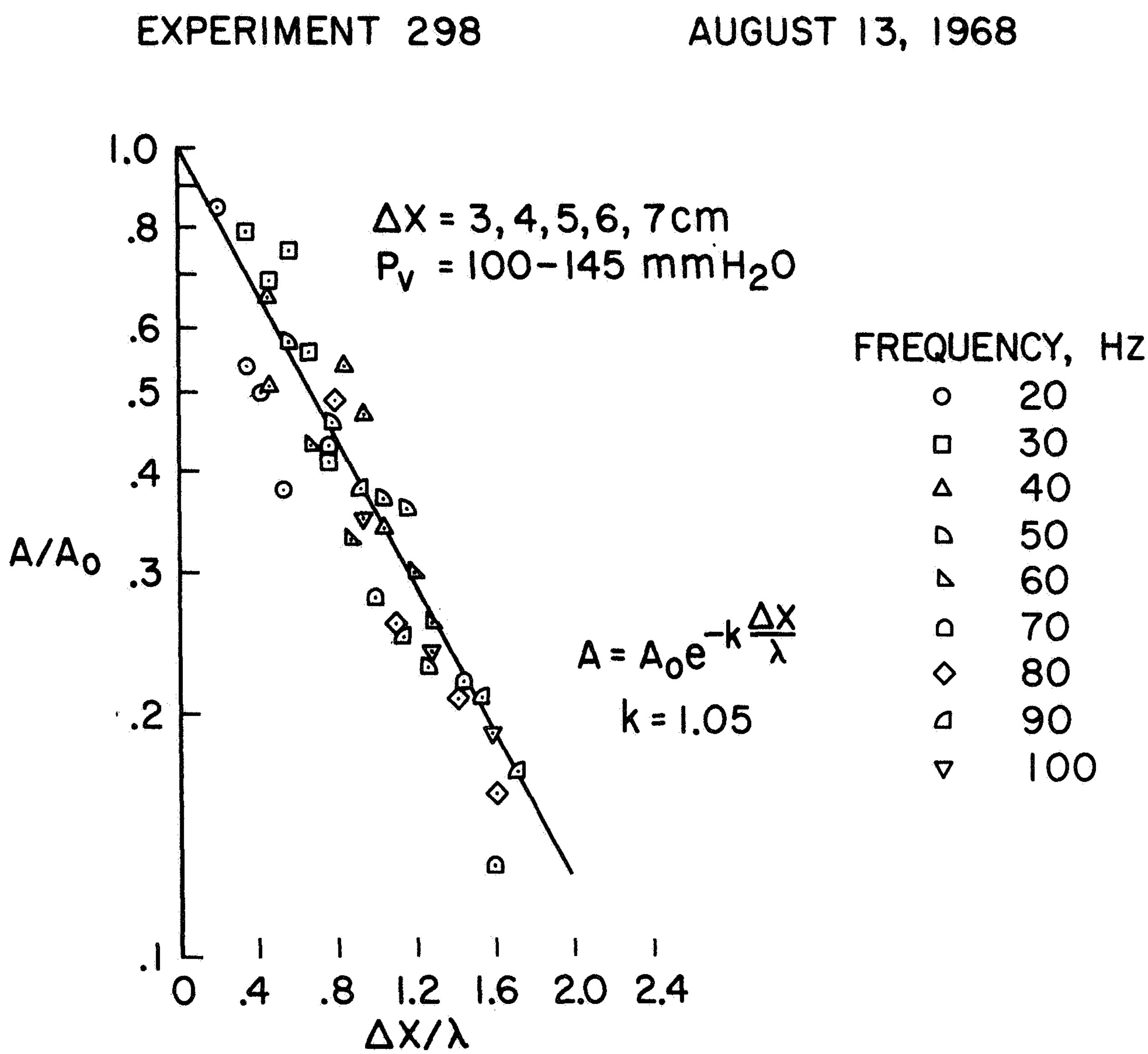


Figure 22. Attenuation curve corresponding to the dispersion data given in Figure 19.

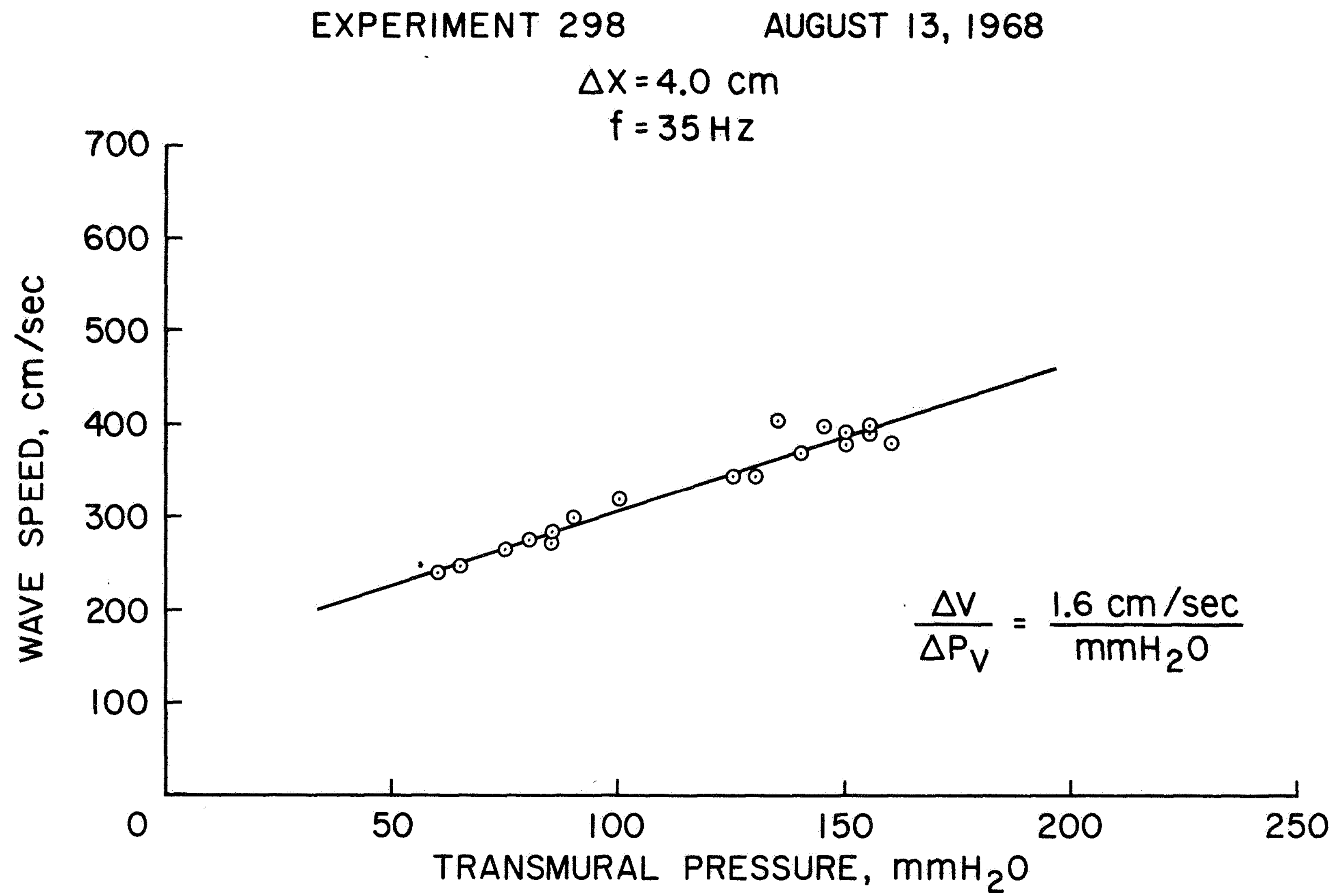


Figure 23. Variation in the speed of small amplitude sine waves with transmural pressure in the vena cava. The signals were induced by the vibrating piston and had peak-to-peak amplitudes less than 20 mm H_2O. The pressure was increased from 60 to 160 mm H_2O in a period of 15 seconds by occluding the vessel with a balloon catheter. Assuming a linear relationship between the wave speed v and transmural pressure P_V, the slope of the curve calculated by the method of least squares is 1.6 cm/sec/mm H_2O. The strong dependence of the wave speed on pressure serves to explain some of the nonlinear behavior observed in the transmission characteristics of large amplitude signals.

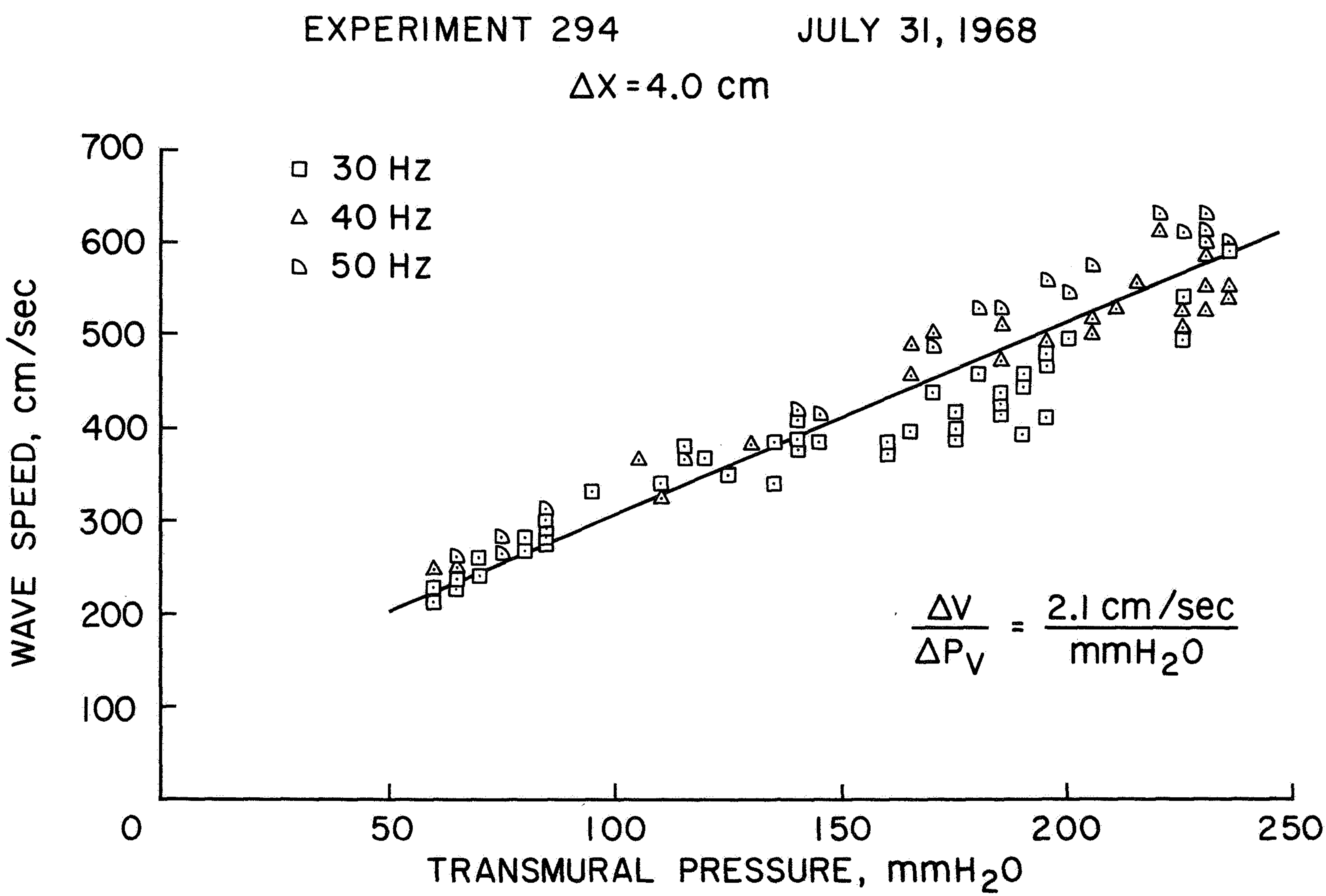


Figure 24. Variation in wave speed with transmural pressure for small amplitude sine waves of different frequencies. Although the complete data were collected over a period of approximately 8 minutes, the data corresponding to a given frequency were obtained in less than 20 seconds during a single occlusion of the vena cava. A linear regression line was calculated for the combined data and was found to have a slope of 2.1 cm/sec/mm H_2O.