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Diagnostic applications of ultrasound in diseases of the heart

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DIAGNOSTIC APPLICATIONS OF ULTRASOUND IN DISEASES OF THE HEART

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PHYSICAL PROPERTIES OF ULTRASOUND

The experiments described in this paper illustrate the use of reflected ultrasound in the diagnosis of abnormal cardiac conditions. This technique for recording cardiac motion was introduced by Edler and Hertz in 1954.[1] It consists of the emission of many short pulses of high frequency sound waves into the chest from a transducer applied to the chest wall. Between transmissions the transducer is silent and is thus capable of receiving the echoes from within the chest which have been produced by reflections of the ultrasound impulse at tissue interfaces. As echoes from deep within the chest require a longer time period to return to the ultrasound source than do superficial reflections, the distance between a given structure and the skin is easily discerned. For convenience these echoes are displayed on an oscilloscope screen. By altering the apparatus so that continuous recordings are obtained, such intrathoracic motions as the beating of the heart may be analyzed.

Ultrasound is the term applied to sound waves above the frequency of audible sound waves. For the human ear this audible limit is 20,000 cycles per second. The laws which govern the behavior of sound in the audible range also hold true for the ultrasonic range, and thus the distinction between frequencies above and below 20,000 cps is arbitrary from a physical point of view. Nevertheless, the exceedingly small wavelengths achieved with

waves of such high frequency permit the utilization of phenomena which are not apparent at lower frequencies. For example, ultrasound may be directed in a straight beam with considerably less scatter than audible sound. In addition, it may be reflected or refracted under suitable conditions. In the following pages the physical bases for these properties will be examined in detail.

Sound waves are longitudinal vibrations (primarily) ~~with~~ which travel through a conducting medium with alternate compression and rarefaction. The velocity of sound in any such medium is determined by the density and compressibility of the material. A fundamental relationship for mechanical waves is that the velocity (v) is equal to the product of the frequency (f) and the wavelength (λ), i.e. $v=\lambda f$.

The transmission properties of a sound conductor are best expressed by its acoustical impedance, which is simply the product of its density (p) and the velocity (v) of the sound passing through it. The velocities and acoustical impedances of various conducting media have been listed in table 1.

When sound travels through a medium with constant acoustical impedance, its intensity becomes progressively diminished as a result of absorption and scattering. If, after traveling through a homogeneous medium, it encounters a second medium with acoustical impedance different from that of the first conductor, part of the sound wave

will be reflected or refracted.

Absorption means that sound loses some of its energy in the form of heat conduction due to internal friction in the conductor. The production of this heat has been widely employed as diathermy in physical medicine. The residual intensity I_r of a sound wave of initial intensity I_o after traveling x distance is described by

(1.1) $I_r = I_o \; e^{-2af^2x}$ a is the amplitude absorption coefficient

From equation 1.1 it may be seen that the penetrating power of sound diminishes with increasing frequency. Unfortunately, high frequencies are necessary to obtain good resolution, so that the ultimate choice of frequency becomes a balance between the desire for high resolution and deep penetration. 2.5 megacycles per second has been found to be a convenient compromise for studies of heart motion. The distance through which sound must travel to lose one half its original intensity in various tissues (at high frequencies) is listed in table 2.

Sound waves begin to drift apart from each other as the distance from their source increases; this is called scattering. For a short distance in the vicinity of the ultrasound generator no scatter takes place. For the cylindrical generator used in these experiments this distance (D) is

(1.2) $D = \frac{r^2 \, . \, f}{v}$ r=radius of generator.

Equation 1.2 states that the distance free from scatter increases directly with increasing frequency.

After the distance D sound waves begin to diverge from the horizontal at an angle θ,

$$(1.3)\ \theta = \sin^{-1} \frac{0.61\ v}{r.f}$$

so that the greater the frequency, the more defined and intense the beam. At 2.5 megacycles per second with a cylindrical generator of 1/2 cm., D is 60 mm. and θ is 4^{o}.

Reflection occurs when sound encounters an interface between two acoustically uniform media having different acoustical impedances. The proportion of total sound which is reflected when the train of impulses reaches an interface at right angles is given by the equation

$$(1.4)\ R = (p_1v_1 - p_2v_2/p_1v_1 + p_2v_2) \quad \begin{matrix} p_1p_2 \text{ respective densities} \\ v_1v_2 \text{ respective velocities} \end{matrix}$$

The percentage of sound waves reflected increases as the difference in magnitude between the two acoustical impedances of the two tissues widens. For water-steel the reflection is 86%, for water-glass 66%, and for bone-muscle 30%. Air and tissue have such widely different impedances (as do air and ultrasound generators) that it becomes necessary to abolish these interfaces if ultrasound is to be transmitted into tissue. This is easily accomplished by the use of air free coupling agents such as lanolin or glycerol.

In crossing an interface sound undergoes refraction in accordance with Snell's law:

$$(1.5)\quad \frac{\sin \theta_1}{\sin \theta_2} = \frac{v_1}{v_2}$$

θ_1 is the angle of incidence
θ_2 is the angle of refraction
v_1, v_2 are the respective velocities

Generation of ultrasound: That ultrasonic waves could be produced in air was shown by König in 1899[2] who employed small tuning forks with prongs only a few millimeters long. Frequencies up to 90,000 cycles per second can be produced in this way. These waves are, however, of extremely small energy.

The most frequently used method at present for generating ultrasound is by piezoelectric crystals. This produces the highest frequencies now possible (10^9 cycles per second).*

The Piezoelectric effect: Crystals such as quartz, zinc, cane sugar, and sodium chlorate develop electric charges on their surfaces when subjected to pressure or tension. This effect (pressure-electric) was discovered in 1880 by the Curie brothers. Their experiments showed that the number of charges set free is proportional to the

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* Since as the frequency of sound increases, the wavelength correspondingly decreases, the wavelength eventually approaches inter-atomic distances. For gases this wavelength corresponds to a frequency of 10^9 cycles per second and for solids 10^{14} cycles per second. At higher frequencies, the motions of particles are more logically discussed in terms of kinetics, rather than waves.

amount of mechanical pressure (or tension). The piezoelectric phenomenon is not a property of all crystals but only of those without a center of symmetry. The sign of the charges is reversed when compression is changed into tension, but otherwise there is no electrical change.

The reverse piezoelectric effect was predicted by Lippmann in 1881. He pointed out that not only will vibrating the crystal mechanically produce electric charges, but that placing a piezoelectric crystal suitably in an alternating electric field will produce mechanical vibrations. Furthermore, when the sign of the applied charge is changed, contraction changes into expansion. The same crystal may then be used to turn high frequency electric oscillations into powerful mechanical oscillations and vice versa. These two effects form the basis of many ultrasonic systems.

TABLE 1*

Tissue	Velocity of sound (meters/sec)	Acoustical impedance ($gr/cm^2/sec$)
Fat	1476	1.36×10^{-5}
Muscle	1568	1.63×10^{-5}
Bone	3360	6.1×10^{-5}
Water	1497	1.49×10^{-5}

TABLE 2

Tissue	Half Value Layer (cm.)
Plasma	100
Fat	6.9
Muscle	3.6
Brain	2.5
Skull	0.23

* Adapted from Edler, I.: The use of ultrasound as a diagnostic aid. Acta Scandinavica suppl. 370.

DIAGNOSTIC APPLICATIONS OF ULTRASOUND

An enormous variety of different medical applications have been imagined for ultrasonic energy in the past twenty years. No tissue has escaped either diagnostic or therapeutic irradiation.

The destructive aspects of high frequency sound were the first to be appreciated. Edler[2] describes the experience of Langevin with the first underwater generator of ultrasound. Fish died when they swam into the beam; a colleague experienced an excruciating pain when he dipped his hand in the water. This property of disruption of tissue has been observed repeatedly; Nelson[5] was led to conclude that ultrasound was too dangerous an agent to be used therapeutically.

The severe disruption of tissue exposed to ultrasound energy is a product of the heating that occurs at interfaces and of the violent oscillations into which tissue particles are thrown. In the previous section the intensity of sound was shown to diminish as it penetrated tissue. The energy absorbed by the body is largely in the form of heat. The application for 20 seconds of sound of frequency 1 megacycle per second at 4 watts/cm^2 will produce a rise in temperature of 5 degrees from skin surface to a depth of 3 cm. These thermal effects have been widely used in physical medicine as diathermy.[6] In the 1940's this method was widely employed in Europe for various muscular and arthritic complaints. Apparently the lack of sufficient knowledge of the properties of ultrasound led to serious

thermal injuries.

Although ultrasonic generators are available with outputs of 60 watts or greater, an output limit of 3 watts/cm^2 minimizes the danger of excessive dosage. When carefully used by trained physical therapists, the technique is extremely valuable, primarily because of its high depth of penetration with good beaming and its higher absorption in muscle than fat.

The second effect of ultrasound on tissues is mechanical disruption. Tissue particles in conducting the sound beam are thrown into oscillation at 10^6 times persecond. Such a particle will be accelerated more than 1000 times the acceleration of gravity. These powerful positive and negative pressures lead to cavitation. Lehmann *et al.*[8] describe the effects of plant roots exposed to 110 watts/cm^2. Complete and uniform destruction of the root centers was observed. Although factors other than those already described were postulated, the cavitation and cellular destruction were considered consistant with those of heating and mechanical disruption. Bell[9] has described the necrosis of liver cells exposed to intensities of 1.5 watts/cm^2 to 40 watts/cm^2. Fry and Dunn[10] have described the ease with which nervous tissue may be irreversibly damaged and report their experience in the production of intracranial lesions.

In adapting ultrasonics to diagnosis, methods of overcoming

the disruptive properties of sonic energy had to be devised so that advantage could be taken of the transmission properties. In some of the earliest experiments short exposures were used in living tissue, but most of the experiments were conducted in isolated specimens. The introduction of the high repetition pulse wave technique made possible continuous application of ultrasonics to tissue without injury. The apparatus used in this method delivers ultrasound impulses of exceedingly short duration. Between pulses there is a long period of silence.

French et al. (1951)[11] were early experimenters with this technique. They used 15 megacycles per second frequencies. Their generator delivered 100 watts when the sound was applied continuously. However, when pulses of 5×10^{-7} seconds duration were used with intervening silent periods of 10^{-3} seconds, (i. e., 7 1/2 vibrations were emitted into the tissues after which 1298 vibrations were skipped, then 7 1/2 were emitted, etc.) the effective power was found to be only 0.05 watts. Animals irradiated for 30 minutes showed no ill effects when examined 24 hours later. Apparently the mechanical effects of ultrasound are minimized when the pulse is of short duration; the long periods of silence probably permit the heat to dissipate before the next series is introduced.

At present, thousands of human subjects have been exposed

to pulsed ultrasound and there is as of yet no recorded instance of harmful effects. In addition after ten years there has been no report of late sequelae. Hertz reports he has carried out repeated ultrasoundcardiography on himself over the last seven years without any untoward effect (quoted by Edler[2]).

Many techniques have been devised to employ the transmission properties of ultrasound in diagnosis. Dussik[12] is credited with having made the first such attempt. He directed parallel beams of ultrasound at the side of an intact skull and recorded them at the other side, much as x-rays are allowed to pass through tissue and are recorded on a plate behind the subject. It is apparent from the discussion in the previous section that such a method takes no advantage whatsoever of the unique properties of ultrasound, but rather relies on their worst aspects, namely, poor penetrance, increasing scatter and energy loss. It appeared at first that he had demonstrated the ventricular system. Güttner, Fiedler and Pätzold[13] were, however, able to reproduce these results with a skull filled with water and considered the method to be therefore valueless.

Keidel[14] used practically an identical method to record heart motions. A continuous beam of ultrasound was passed through the thorax. The residual intensity fluctuated synchronously with the heart beat but more specific details especially accurate volume

changes, were not able to be evaluated.

A second diagnostic approach was introduced by Satomura[15] and makes use of the Doppler effect. A detecting device receives both the transmitted pulse and the reflected wave from the heart. If a Doppler frequency shift has occurred, there will be a detectable difference in frequency between the transmitted pulse and the reflected pulse and this difference will be recorded. Many Japanese workers, in particular Yoshida,[16] have reported characteristic beats which when correlated to the electrocardiogram seem to represent motion of the atrial and ventricular walls as well as other signals which are said to represent valve motion. The difficulty with the technique is that a continuous recording of motion is not produced, and the distance between the reflecting surface and the transducer is obscured.

What has become the most fruitful diagnostic method is the pulse reflection technique, which in essence is precisely the technique used by Navy research teams in sonar exploration of the ocean floor. As was discussed in the previous section, the ultrasonic pulse is partially reflected by any interface where the specific acoustic impedance of the medium changes. Tissue is composed of numerous such interfaces. As the ultrasound beam penetrates the body, a fraction of it is reflected at each interface, while the remaining energy passes deeper. The reflections are recorded at

skin surface either by a second transducer (when the reflections occur at other than 90^{o}) or as in the experiments described in a later section by the same transducer during its silent phase. The echoes which return latest are interpreted by the machine as having come from the deepest structures.

This method is used routinely in industry for detecting flaws in long beams. It has even been used by hog raisers to measure the thickness of the hog's outer fatty layer, the interface of major importance being the fat-muscle border.

In 1949 Ludwig and Struthers[17] found that they could detect foreign bodies that they had buried in dog's muscles but concluded from their experiments that soft tissues produced such inexact reflections that they could not be distinguished from one another. Wild,[18] however, (1950) concluded that echoes could be used to detect tumor masses from soft tissue. He investigated tracings from three layers of small intestine of a freshly killed dog and was able to detect the presence of each layer. Tracings were also made from a carcinomatous ulcer of human stomach and the thickness of the ulcer was measured fairly exactly. Together with Reid in 1956[19] he reported the results of examination of 77 palpable masses in human breast tissue. The tumors were said to be clearly divisible on the basis of ultrasonic recordings into malignant and non malignant

tumors with a high degree of accuracy. The 27 malignant tumors all gave a characteristic echo (7 benign tumors also gave this pattern) while 43 of the 50 benign tumors produced a second type of echo. In carrying out these investigations the experimenters used two separate transducers, one which emitted the impulse and one which searched for the cho. By moving the transmitting crystal and exposing the same photographic plate only when each returning echo had been found, they were able to build up on one picture a two-dimensional view of the tumors.

In 1954 Edler and Hertz[2] began to use this method for examination of the heart. As their experiments form the basis for the experiments carried out in this paper, and as they have already become classic papers in the field of ultrasoundcardiography, they will now be described in detail.

In 1953 and 1954 Hertz and Edler undertook a series of experiments to determine whether the heart wall-blood interface could be detected. Using isolated heart preparations they demonstrated how sharply defined such interfaces were and how easily they could, in fact, be located. The thickness of the heart wall and the presence of thrombi in the heart chambers were also well described.

Applying this technique to the living heart, they obtained a series of pulsating echo signals when the transducer was placed over

the fourth left interspace near the midline. By changing their method of recording so that continuous records could be built up, characteristic echoes were recorded from many human subjects. One echo in particular was found to occur on the records of every patient when the transducer was applied to the left third or fourth interspace 1-4 cm lateral to the sternal margin. This echo signal represented the motion of a structure lying 6-8 cm. from the chest wall and 2-4 cm. anterior to the rearmost echo from the heart. It moved much more rapidly than any other echo. Originally it was assumed that it arose from the anterior surface of the left atrium and it was noted that in mitral stenosis the echo assumed a different pattern.

In order to elucidate the exact nature of this echo Edler and Christensson[20] carried out studies on cadavers. They passed a needle through the chest wall in exactly the same path as that assumed to have been taken by the ultrasound beam. After piercing the anterior wall of the right ventricle the needle passed through the right ventricular outflow tract, the intraventricular septum, the upper part of the left ventricle, the anterior leaflet of the mitral valve, and out through the posterior wall of the left atrium.

The echo producing structure was noted to approach nearest the crystal early in ventricular diastole and be farthest from the crystal during ventricular systole. Furthermore, the behavior of the

echo source could be well correlated with atrial activity. In the same study, Edler and Christensson perfused isolated cow hearts and produced motion of the mitral valve by cyclically varying the pressure in the left ventricle. Echoes of characteristic movement were obtained, and needles passed in the direction of the ultrasound beaming encountered the same structures as in the cadaver studies. During the experiment the anterior leaflet of the mitral valve was immobilized by forceps, whereupon the echo signal became flat. This was considered ultimate proof that the source of the fast moving echo signal was the anterior leaflet of the mitral valve.

Joyner[21] has carried out examination of the mitral valve in over a hundred patients and recently published his experience. Twenty-five normal patients showed the characteristic double-peaked curve described earlier by Edler. In contrast, 90 patients with mitral stenosis demonstrated a distinctive, abnormal pattern in which the peaks were replaced by a plateau. Although the pattern of 35 patients who were studied both before and after surgery changed after valvulotomy, none of these patients ever developed a normal pattern.

Ultrasound has been used for a variety of diagnostic purposes. References to many of these other techniques appear at the end of this paper for the convenience of readers interested in other aspects of ultrasound diagnosis.

PROTOCOL: MATERIALS AND METHODS

RESULTS

DISCUSSION AND SUMMARY

Materials and Methods:

The equipment consists of two ultrasound reflectoscopes made available to the Yale Department of Radiology by the Sperry Division of Automation Industries. As can be seen in the accompanying block diagram, a short pulse of electrical energy is delivered to a barium titanate crystal transducer producing an ultrasound pulse having a frequency of 2.5 megacycles per second, and a duration of one microsecond. The transducer immediately becomes silent for the next 999 microseconds and then emits another pulse. Thus 1000 such pulses are emitted every second but the transducer emits sound for only 1/1000th of a second. The acoustic power output is well below the limits discussed in the previous section.

After returning from the patient, the echo is transformed again by the transducer into electrical energy and displayed on an oscilloscope screen. For convenience there are two oscilloscope screens which are related in the following manner.

The first screen is connected directly to the amplifier. As the electron scan sweeps horizontally across the scope face, returning echoes cause vertical deflections (or peaks) the height being in proportion to the magnitude of the echo. This is the "A" type oscilloscope. A typical pattern is seen in figure 1. The second oscilloscope is connected to the first and is of the intensity type.

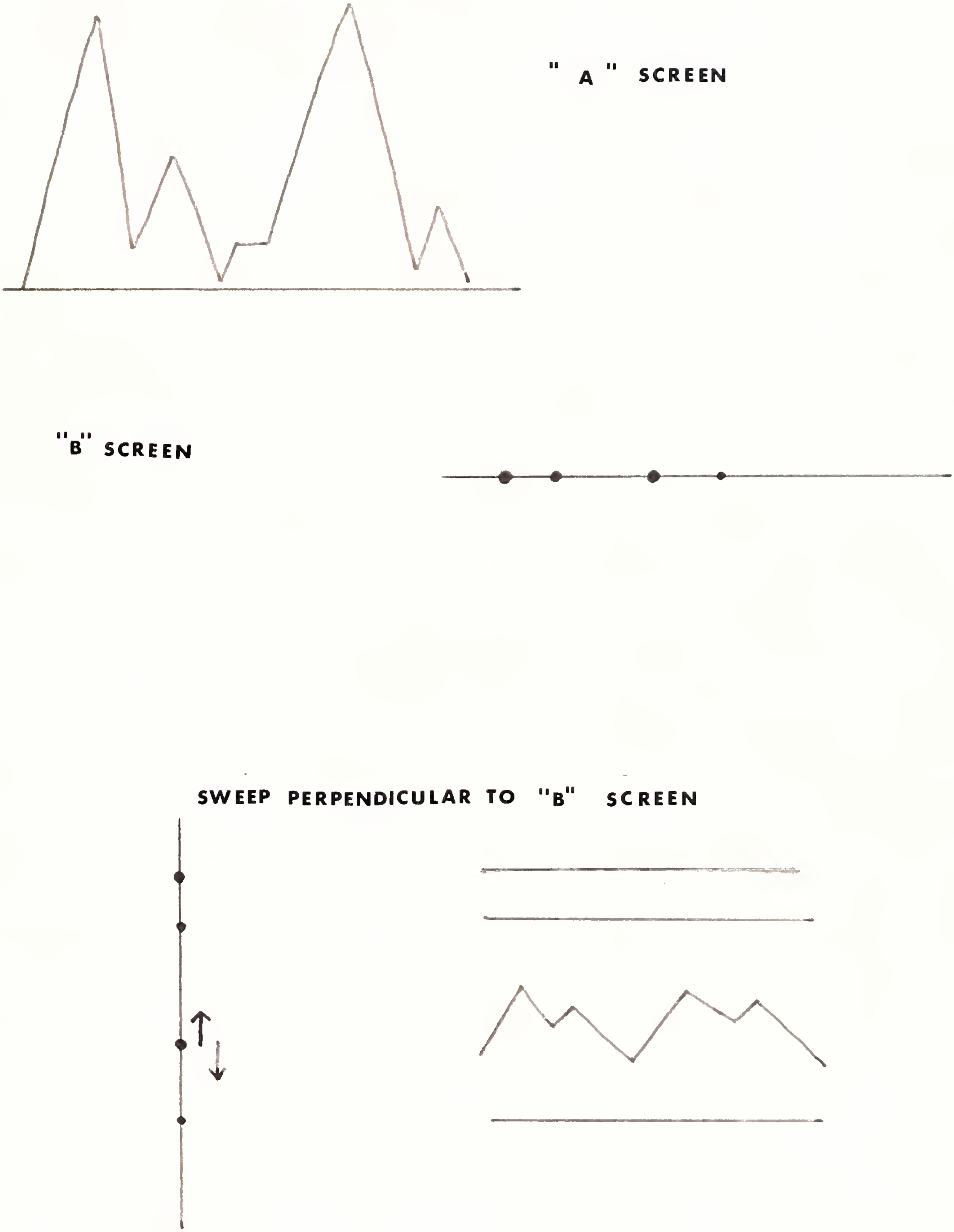


FIG. 1

Peaks on the "A" scope are reproduced as dots on the "B" scope. These dots increase in intensity with increasing height of "A" scope peaks. A reject is provided so that the minimal height at which an "A" scope dot will make its appearance on the "B" scope can be varied.

As the heart beats, its components alternately come closer to and recede from the skin. These echos can be seen as moving peaks on the "A" scope and as oscillating dots on the "B" scope. With a little practice the distinctive movements of the myocardium and mitral valve can be recognized on the "A" scope.

Once these have been detected, a slow sweep perpendicular to the base line of the "B" scope is begun. This allows the dots to be displayed as lines and thus a two-dimensional (depth, time) picture is obtained. Permanent records are obtained by means of a Polaroid camera with time exposure.

To obtain a recording, the patient is placed in the supine position and the skin surface from the third through the fifth interspace on the left 1-3 cm. from the left sternal border is coated with a thin layer of lanolin. The transducer is placed firmly on the chest over this area and adjusted until characteristic echoes are seen. In most instances this has been in the fourth interspace 2 cm. from the left sternal border. Occasionally recordings are made from the

fifth or third interspaces. Best results are obtained in the majority of instances when the transducer is placed firmly on the skin and angled slightly upwards so that the contact is finest at the lower margin of the interspace.

The "B" scope is a dual beam oscilloscope, the second channel being used to transmit the electrocardiogram. The "A" scope is equipped with a square wave marker which can be calibrated to read depth. Unfortunately, the "B" scope does not, as yet, have such a calibrating wave. The measurements that are recorded on the individual prints appearing in this paper were obtained from the "A" scope calibrating wave and were visually estimated at the time of the examination.

The sweep time may be varied from 2 seconds to 4 1/2 seconds. The timings that accompany the photographs were also noted (by a watch) at the time they were recorded. (Most of the early photographs have 3 seconds of exposure, while the later ones are generally 4 1/2 seconds.)

Results:

Ten normal subjects were studied and all showed the characteristic echo pattern originally described by Edler and Effert. Three of these tracings are shown in Fig. 2. The horizontal axis represents a time span of 4.5 seconds (except where marked otherwise). The

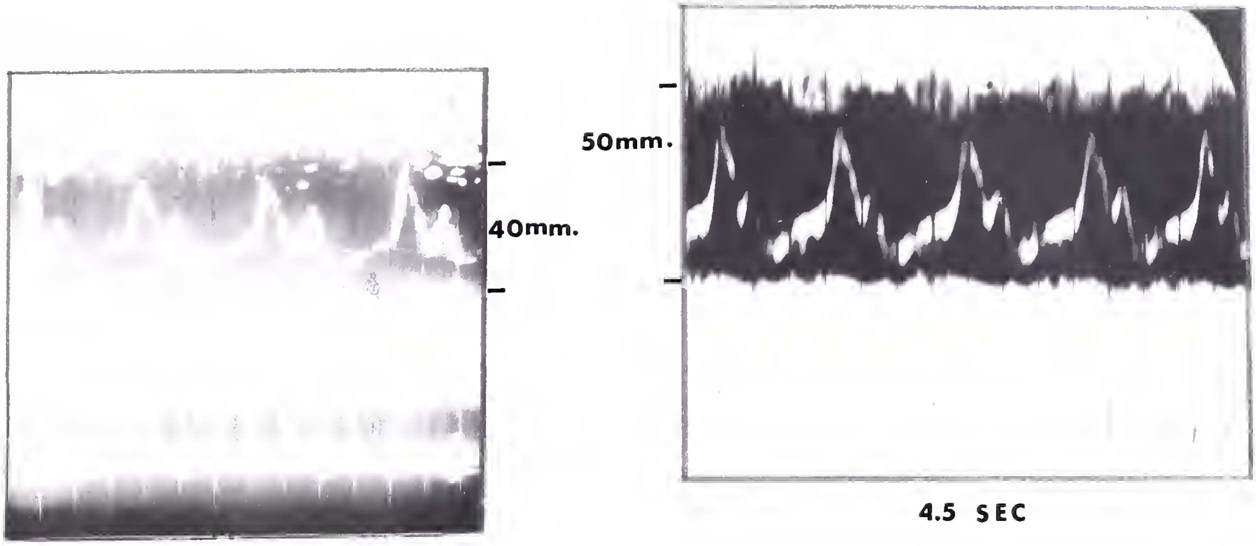


Fig. 2

Normal Mitral Valve Pattern

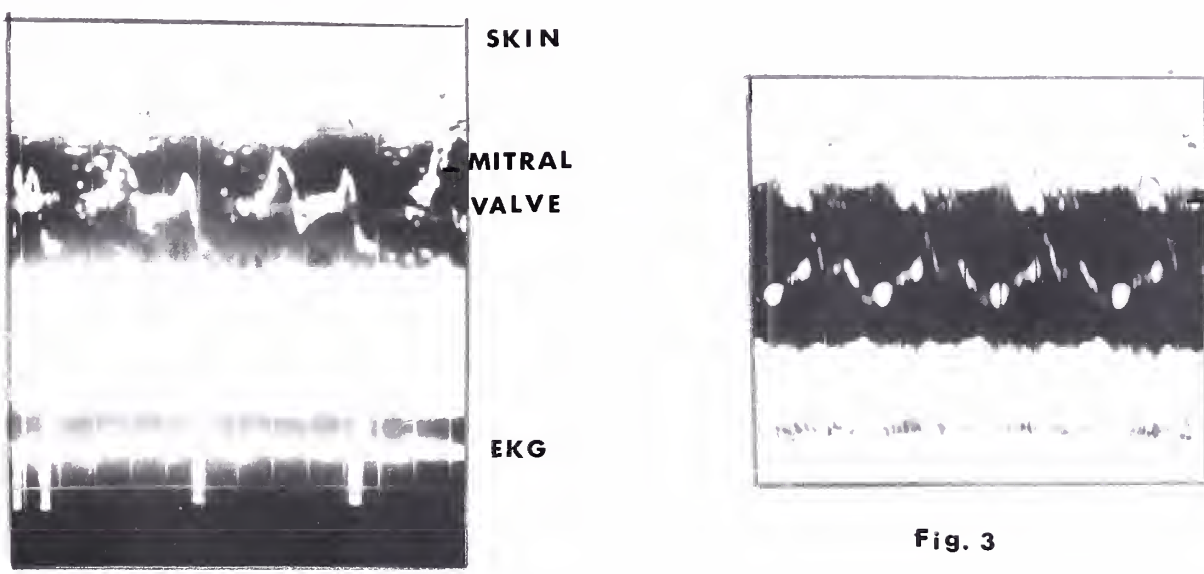


Fig. 3

top of the recording represents the echo returning the anterior chest wall. The echoes returning from successively deeper structures are so numerous that they coalesce to form the broad zone of white seen at the top. On some of the records the anterior heart wall is distinct from this zone, and has been noted separately as in Fig. 3. Below the echo from the anterior heart wall there is a blank space of approximately 40 mm. width containing a single fast moving echo source. This tracing is from the anterior leaflet of the mitral valve. The black area represents the blood within the heart chambers. As there are no interfaces within blood, the sound is conducted without reflection and no additional echoes occur. Below the mitral valve recording there is a second broad white expanse which contains the echoes returning from the posterior wall of the heart and structures lying deeper within the posterior mediastinum.

As was discussed in the last section the ultrasound crystal is placed in the fourth left interspace from 1-4 cm. lateral to the left sternal border. In its passage through the thorax it encounters the anterior wall of the right ventricle, the right ventricular outflow tract, the intraventricular septum, the anterior leaflet of the mitral valve, and the posterior wall of the left atrium. The experiments of Edler showed that the tracing obtainable from the mitral leaflet oscillated between 5.5 and 8.5 cm. depth from the anterior chest wall.

In Fig. 4 the mitral tracing has been numbered in accordance with the designations proposed by Effert. Upward deflections of the record represent movement of the mitral leaflet anteriorly and hence opening of the valve, while downward deflections represent closure.

The mitral tracing consists of two peaks (numbered 4 and 1). These are the points at which the anterior mitral leaflet is closest to the ultrasound generator on the chest wall. Point 2 is the portion of the curve at which the mitral leaflet is furthest from the crystal.

Normally point 1 occurs about 0.08 - 0.12 seconds after the start of the P-wave on the electrocardiogram. After point 1 the valve tracing falls to its lowest point (2) early in ventricular systole. A slight plateau (3) follows after which the curve rises to its highest point (4) which is always the most anterior portion of the recording. Effert[2] found that among the normals he examined the interval between the second heart sound and point four was 0.11 - 0.16 seconds with an average value of about 0.13 seconds. The curve then falls off sharply between point four and five. The slope of segment 4-5 is normally quite steep; as will be seen below, this is radically altered in mitral stenosis. From 5 the tracing rises to return to point 1. Once again it can be seen that the P-wave falls between 5 and 1.

The results of examination of ten normal subjects is shown in Fig. 7. The average amplitude of motion from the most anterior

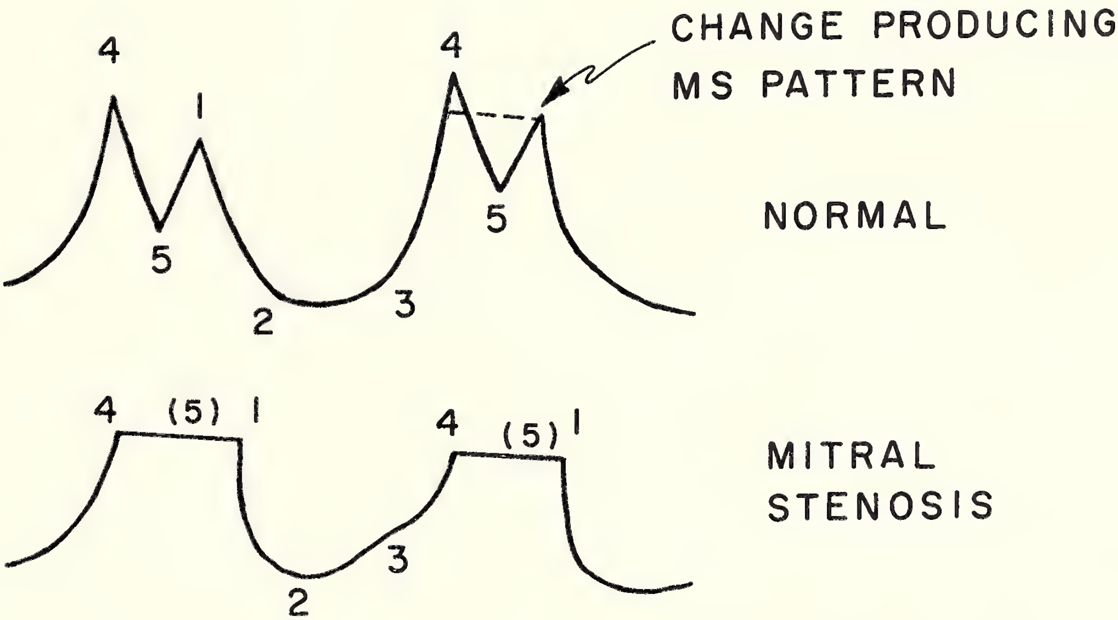


FIG. 4

(SEE TEXT FOR DETAILS)

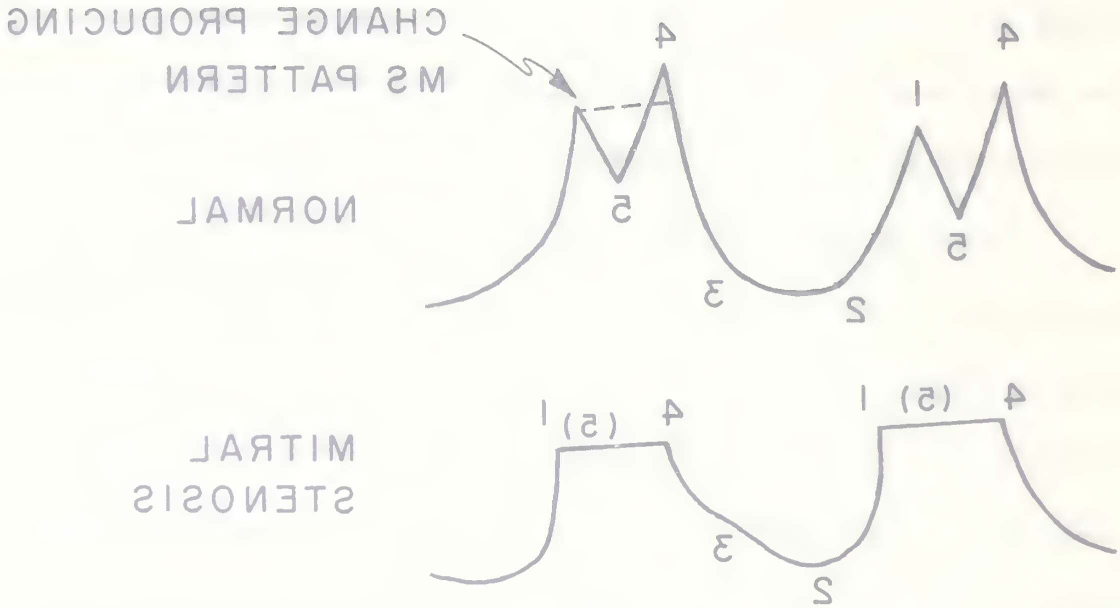


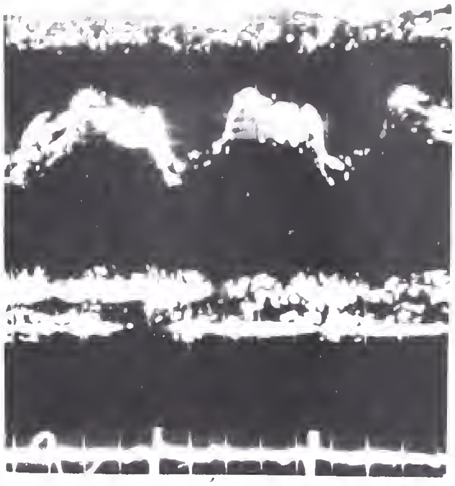
FIG. 4

(SEE TEXT FOR DETAILS)

point (4) to the most posterior point (2) was 25 mm. with a range of 20-32 mm. The slope of the curve between 4 and 5 averaged 140 mm. per second posteriorly with a range of 100-160 mm/sec. The time interval for this period was generally between 0.16 - 0.20 seconds. (One subject showed a 4-5 interval of only 0.11 seconds but the slope was still in the range of normal, at 160 mm/sec.)

Twenty-eight subjects with mitral disease were studied. Ten of the records were omitted from the study for the following reasons. The first six records were made when little experience had been gained with the apparatus and are technically so poor that it was felt best not to consider them further. There were four patients of the remaining 22 from whom recordings could not be obtained. All of these were women with a large amount of breast tissue which apparently absorbed and scattered so much of the ultrasound that no meaningful tracing could be recorded. This failure to obtain recordings in four of the 22 patients (four of 37 people examined including normals) is less favorable than the failure to record in one of every 15 cases published by Joyner in his study of over a hundred patients.

Of the 18 patients from whom recordings were obtainable 12 had mitral stenosis as the predominant lesion and six had mitral insufficiency. All but three of these were proved at operation. The other three were not operated upon but have supporting data from cardiac



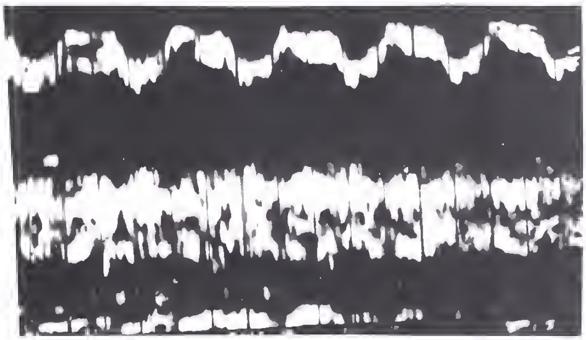
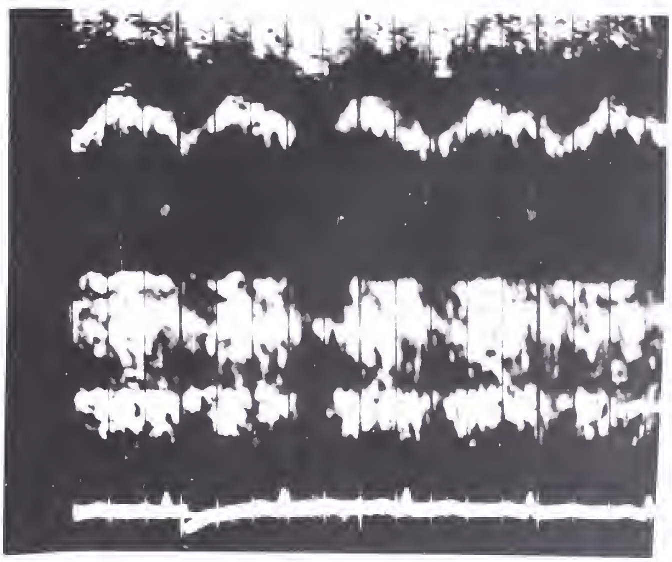


Fig. 5

MITRAL
STENOSIS



catheterization. (All the patients in the series were catheterized.) Five of the patients were examined both pre-operatively and post-operatively. Two of these patients had Starr valve replacements.

The typical mitral stenosis pattern is seen in Fig. 5. It will be noted that the striking difference between this recording and a normal one is the replacement of the double peak by a single flat top recording. What has happened is that the excursion to peak four is only about one half the normal value and the tracing has practically remained at a fixed level until the onset of ventricular systole (just after peak 1, normally.) The average maximum excursion for the mitral leaflet was 25 mm. in normals and 15 mm. in the stenosis group. Even more striking is the slope of the curve posteriorly from point four. It was mentioned above that in the normals this slope averaged 140 mm/sec. In the stenosis group the range was 0-40 mm/sec. with most tracings falling in the 10-20 mm/sec. range. The post-operative records show some improvement in the motility of the valve leaflets, but as can be seen in Fig. 6, these never approach the normal values. Fig. 7 summarizes quantitatively the difference in mitral motility between normals and subjects with mitral stenosis. Size of the mitral orifice as estimated at operation (or at catheterization when no operation was performed) is plotted against the speed of descent of the mitral tracing between points four and five. It can be seen that

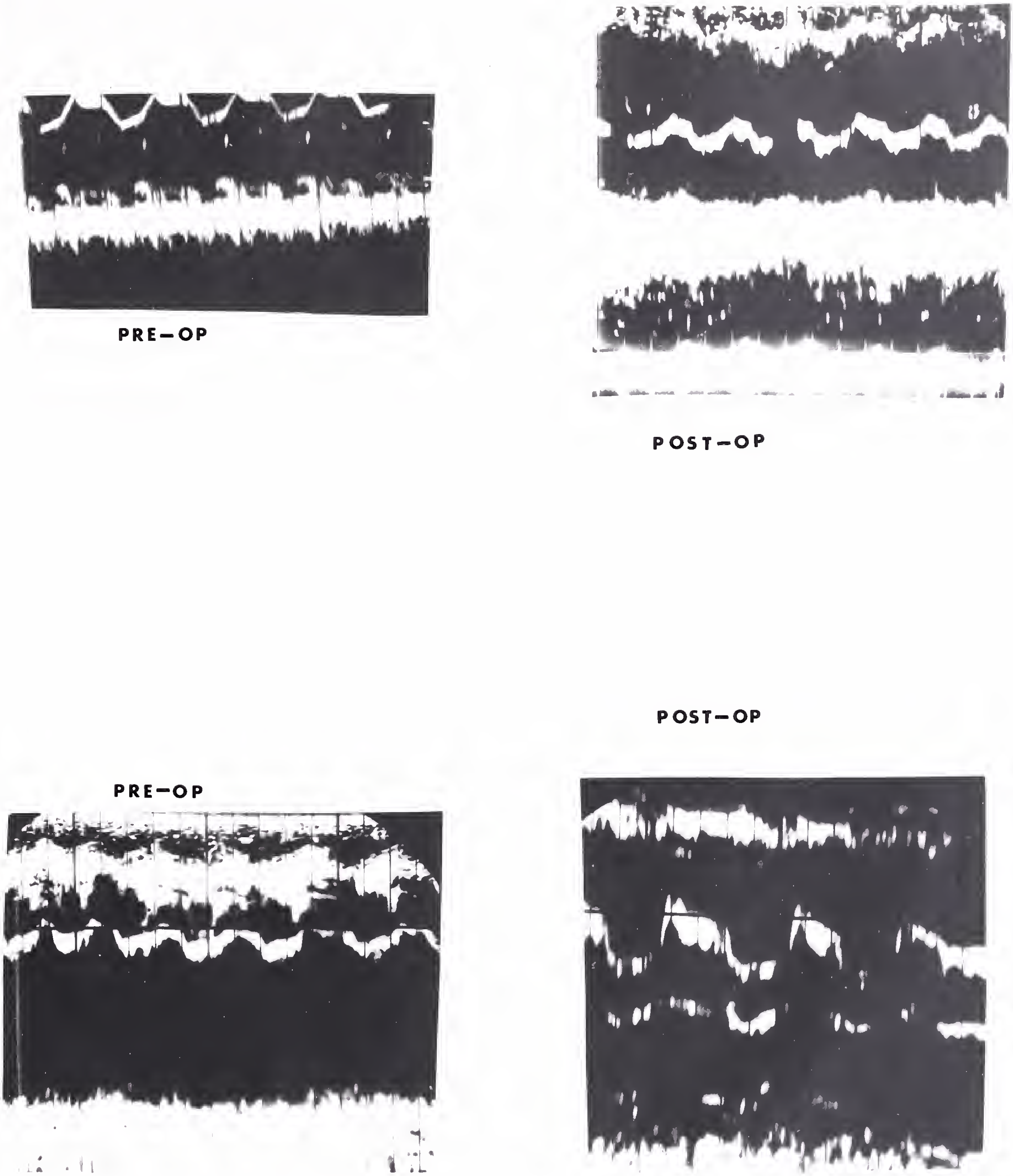


FIG. 6 MITRAL STENOSIS

no normal patient had a velocity slower than 100 mm/sec. and no patient with mitral stenosis had a pre-operative velocity faster than 50 mm/sec. and that most fell in the range of 10-20 mm/sec. Furthermore, after operation all patients showed a velocity well below that of the normal group. The pre- and post-operative records of two patients are reproduced in Fig. 6. Note that even where there is some increase in the velocity of the curve after point four that the tracing is still qualitatively that of stenosis, i.e. there is a plateau rather than a double peak.

Of the five patients with predominantly insufficient valves, two showed a completely normal pattern while three were abnormal in that they demonstrated only one peak. Nevertheless even these three recordings represent highly mobile leaflets and all have posterior deflections from point four in, or close to, the normal range. Two of these patients had Starr valve replacement. Echo recordings from these prosthetic valves were unsatisfactory.

Summary:

1. The mitral valves of ten normal subjects and of 18 patients with known mitral disease were studied with ultrasonic reflection techniques.

2. In each of the normals a characteristic and reproducible echo pattern was obtained.

3. The patients with mitral stenosis all exhibited a characteristic echo pattern which was different from that of the normal group.

4. Patients undergoing mitral valvulotomy had increases in their velocity of valve motion post-operatively; nevertheless, these rates still fell far short of the normal range.

5. Patients with mitral insufficiency could not be distinguished from normal subjects.

These results are in agreement with those of Joyner and Reid.[21] Echocardiography appears to be a reliable guide in the diagnosis of mitral stenosis.

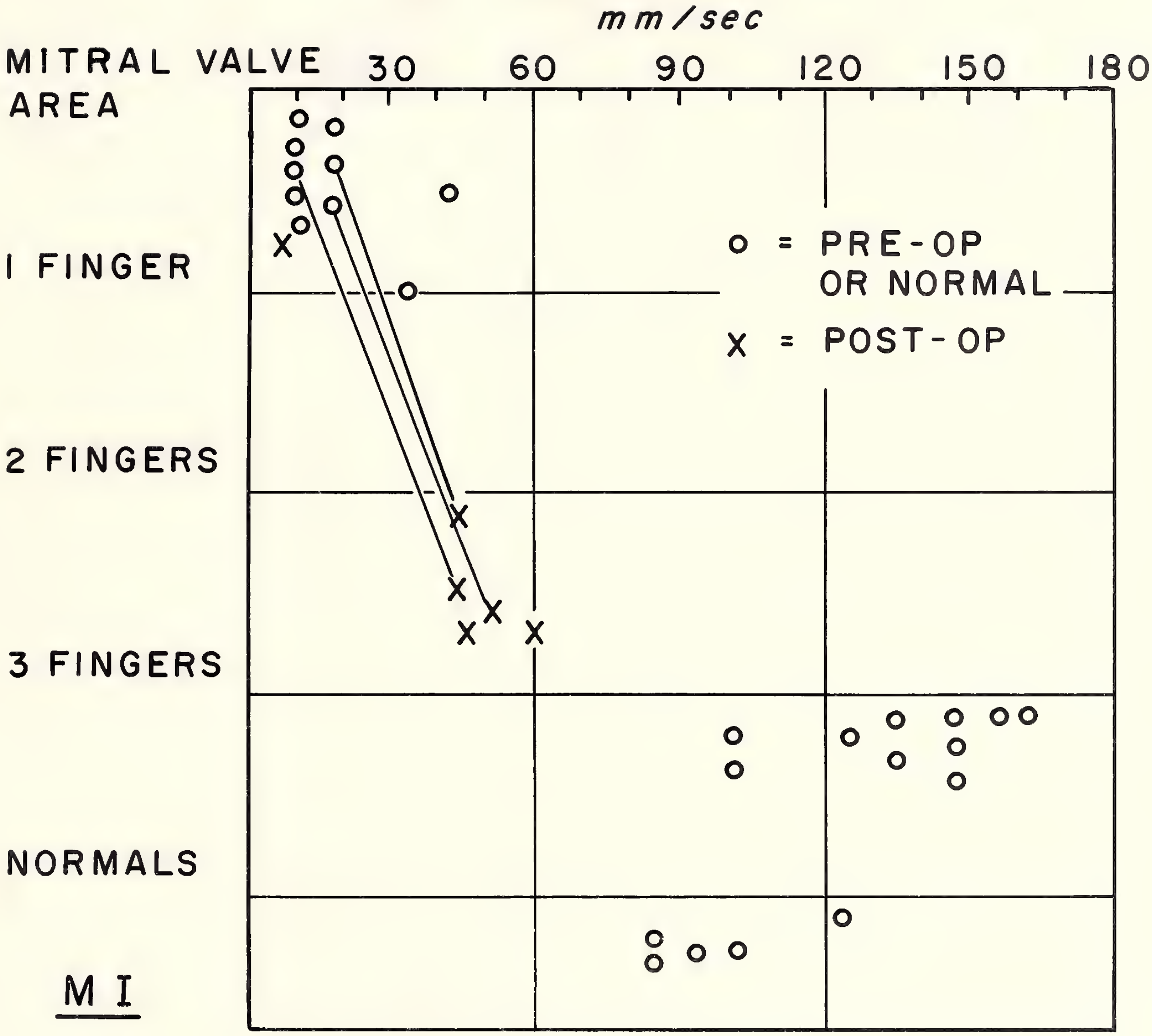

mm/sec
MITRAL VALVE AREA
30
60
90
120
150
180
1 FINGER
2 FINGERS
3 FINGERS
NORMALS
MI
o = PRE-OP OR NORMAL
X = POST-OP

FIG. 7

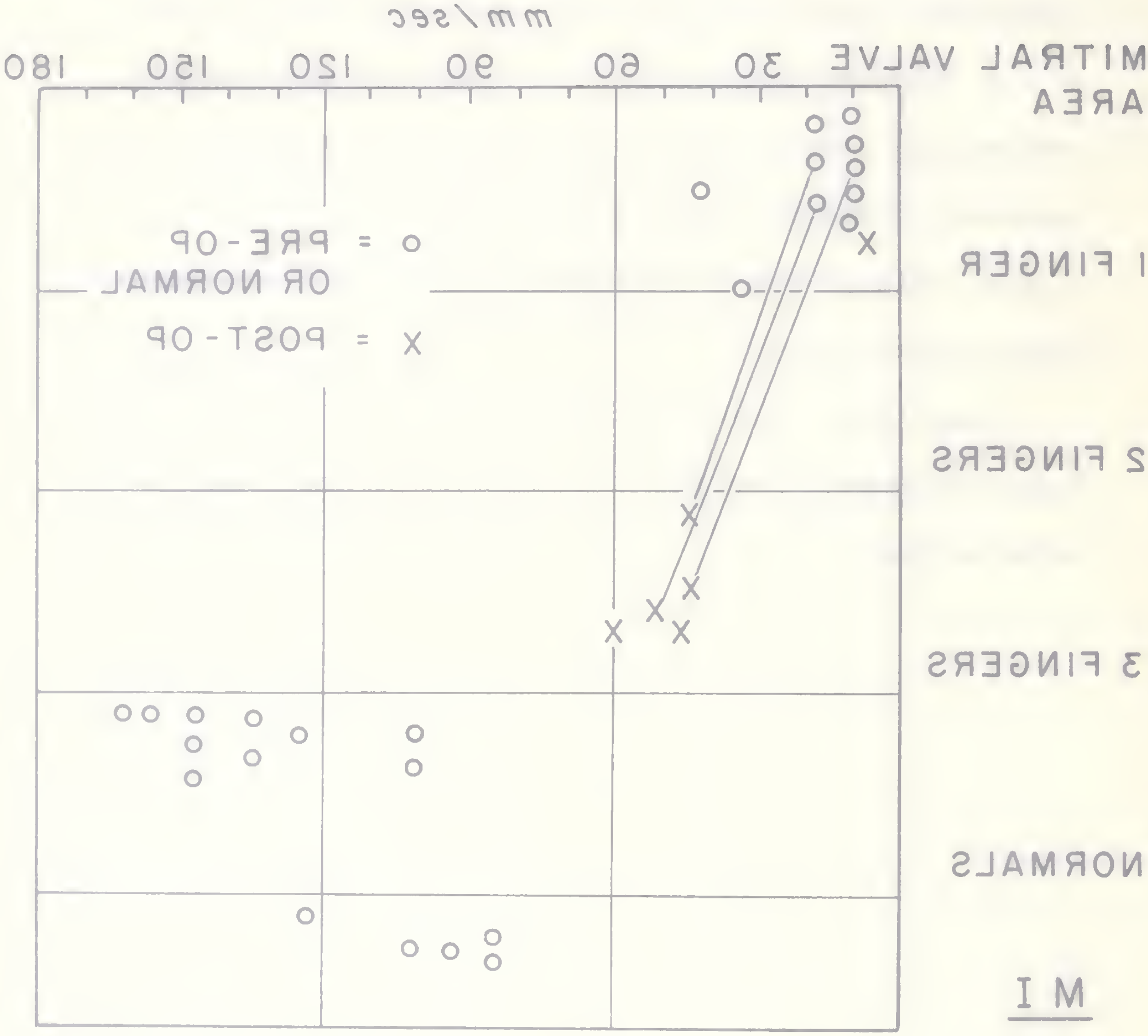

mm/sec
180
150
120
90
60
30
MITRAL VALVE AREA
o = PRE-OP OR NORMAL
X = POST-OP
1 FINGER
2 FINGERS
3 FINGERS
NORMALS
MI

FIG. 7

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APPENDIX

The unique ability of ultrasound to detect tissue interfaces has suggested its use in the diagnosis of pericardial effusions. The differentiation of cardiac dilatation from pericardial effusion is often difficult, special radiographic procedures are necessary. They are discomforting to the patient and are not without risk. An excellent review of available methods and their limitations has recently appeared.[1]

In the normal or dilated heart the myocardium is closely applied to the pericardium and that in turn to the chest wall. As has already been described this uniform sound density results in ultrasonically indistinguishable tissue planes, which in normal recordings of the heart becomes a coalesced zone of while (see Fig. 2). Occasionally the anterior wall of the heart can just barely be distinguished as in Fig. 3.

With pericardial effusion, however, the fluid lying between the pericardium and the myocardium is of sufficiently different sound density from the myocardium so as to produce a clearly defined pair of interfaces. To test this hypothesis and to determine

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1. Burch, G. E. and Philips, J. H: Methods in the diagnostic differentiation of myocardial dilatation from pericardial effusion. Am. Heart Journal. 64:2, 1962.

the amount of effusion that must be present for ultrasonically detection, studies were first undertaken in cadavers.

Because tissue rapidly loses its sound transmissibility after death, cadavers less than four hours old were used. With the thorax intact fluid was introduced into the pericardial sac through the diaphragm. After 100 cc. had been infused a sharp black band appeared between the echoes returning from the anterior chest wall and those from the anterior heart wall and represents the region of the effusion.

To date only two patients with suspected pericardial effusion have been studied. One patient was seen while recovering from open heart surgery. Effusion appeared to be present by ultrasound but no confirmatory studies were undertaken. Several days later the apparent effusion had disappered.

The second case is that of a young woman with Hodgkin's disease with marked enlargement of the cardiac shadow on chest film. The ultrasound study is produced in Fig. 8. Note that a black band appears between the anterior heart wall and more superficial structures. A CO_2 study was performed in this patient which unfortunately was technically unsatisfactory, causing observers to disagree over the presence or absence of an effusion.

Although the dearth of clinical material has prevented any

convincing study concerning the reliability of this technique, on both theoretical grounds and the results of studies with cadavers, there is reason to believe that ultrasound will become a most sensitive method for the detection of pericardial fluid, and may replace other radiographic procedures now in use.

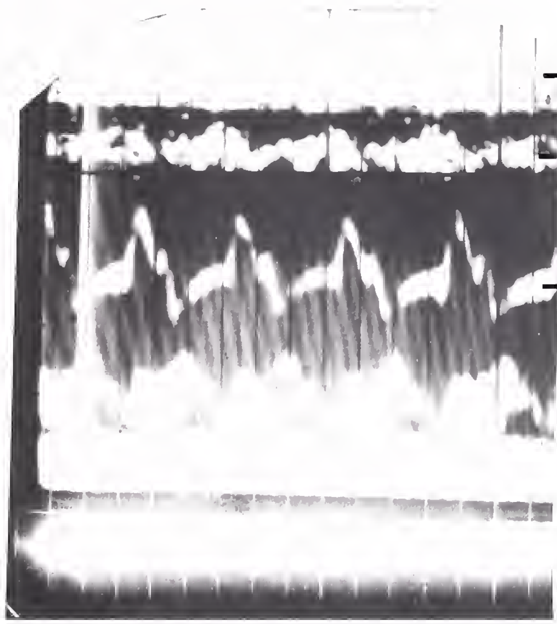

FIG. 8

PERICARDIAL EFFUSION