Cardiovascular Research, 1973, 7, 703-712.

Instruments and techniques

Transcutaneous measurement of blood velocity profiles and flow[1]

MICHAEL B. HISTAND, CHARLES W. MILLER, and FRANCIS D. MCLEOD, JR.
From the Departments of Mechanical Engineering, Physiology and Biophysics, and Radiology and Radiation Biology, Colorado State University, Fort Collins, Colorado 80521, and the Institut für Biomedizinische Technik der Universität Zürich und der Eidgenössischen Technischen Hochschule Zürich, Switzerland

AUTHORS' SYNOPSIS A comprehensive report is presented of the application of a pulsed ultrasound Doppler velocity meter for transcutaneous measurement of time varying velocity, velocity profiles, and instantaneous flow in arteries of anaesthetized dogs. The procedure used to provide direct velocity and flow calibration using the Doppler equation is outlined. Typical transcutaneous recordings obtained from the femoral artery, abdominal aorta, and carotid artery are illustrated. The results compare favourably with data obtained by invasive means such as electromagnetic cuff flowmeters. The possibility of high resolution, non-invasive haemodynamic measurements on dogs is demonstrated and the application to conscious human subjects suggested.

A long-standing goal of medical researchers has been a facility for high resolution measurement of blood flow and velocity distributions transcutaneously. Numerous applications exist for the new techniques described in this paper, but our particular interests are haemodynamic measurements germane to diagnosis, localization, and study of the progression of peripheral artery disease. The research provides a non-invasive, non-traumatic method for recording intraluminal velocity distributions, velocity profiles, and instantaneous volume flow in subcutaneous arteries.

Medical ultrasound instrumentation developed during the past decade permits less traumatic measurement of blood flow than the electromagnetic flowmeter or hot film anemometer (Strandness, 1969). A continuous wave Doppler flowmeter was originally conceived by Satomura (1959) and later used by a number of investigators to locate subcutaneous arteries transcutaneously and evaluate vascular patency (Strandness, Schultz, Sumner, and Rushmer, 1967), to measure instantaneous mean blood flow in a vessel transcutaneously or transmurally (Rushmer, Baker, and Stegall, 1966; Reagan, Miller, and Strandness, 1971), or to monitor the fetal heart beat *in utero* (Bernstine and Litt, 1971). Frequently *in vivo* timed fluid collections are required for instrument calibration. These calibrations requiring surgery are usually not warranted in man and may distort phenomena studied experimentally in animals. Some investigators contend that the Doppler equation may be applied directly to calculate flow (Franklin and Van Citters, 1967; Van Citters, 1968); however, the difficulty involved in accurately measuring the vessel diameter introduces error into this procedure. Because of this problem, the transcutaneous continuous wave Doppler velocity detector is more error prone than its application as a surgically implanted cuff. The most recent advancement in ultrasound flowmetry, the pulsed ultrasound Doppler velocity

[1] Supported in part by USPHS Grant No. HE 14737 and USPHS Contract CPE-R-70-0001.

meter (PUDVM), obviates many of the resolution and calibration problems of the continuous wave Doppler (McLeod, 1967, 1970, 1971; Baker, 1970, Peronneau, Hinglais, Pellet, and Léger, 1970). When integrated with tape acquisition systems and rapid computer data processing, the PUDVM provides a superior method for measuring blood velocity and flow transcutaneously. In contrast with the continuous wave Doppler, the PUDVM emits repetitive pulses of ultrasound which are backscattered by erythrocytes. Appropriate electronic gating of the returned signals provides velocity data from a small intraluminal sample volume rather than an average over the complete cross-section. This research was conducted to develop procedures using a PUDVM and to evaluate its performance in recording blood velocity and volume flow non-invasively at a number of arterial locations in dogs.

Method

Description of instrument

The ultrasound transducers (Fig. 1) used with the PUDVM were designed and made in our laboratory. The emitting-sensing portion of a transducer consists of a circular disk of lead titanate zirconate[2] (piezoelectric ceramic) silvered on its front and back surfaces and bound with a layer of epoxy to a stainless steel tube which supports and defines the axis of the transducer. The piezoelectric disks range in diameter from 1·5 to 3 mm. The size of the disk affects the emitted ultrasound beam geometry and consequently the sample volume (Baker, 1970). The size of transducer chosen for an experiment is related to the depth beneath the skin where one desires a recording. By controlled gating of the returned signals a single disk serves as both emitter and receiver.

The PUDVM emits pulses of 7–8 MHz ultrasound 4, 8, or 16 cycles in length at a repetition rate of 10, 20, or 40 KHz to drive the transducer. The choice of emitted ultrasound frequency is dependent on the optimization of transducer natural frequency and emitted beam characteristics. In these experiments at depths generally less than 1·5 cm, 2 mm diameter transducers were driven at a frequency of 7·5 MHz, 16 cycle pulse length, and 40 KHz repetition rate to obtain sufficient amplitude of the backscattered signal over the cross-sections of the subcutaneous arteries studied.

[2] Edo Western Corporation, Salt Lake City, Utah 84115.

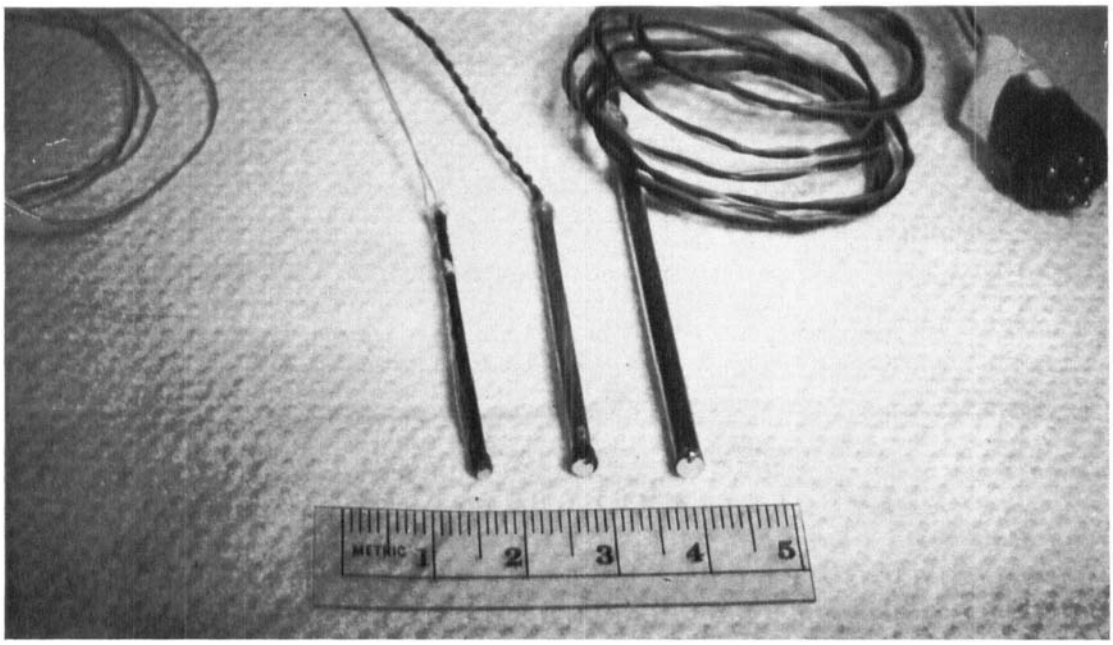

FIG. 1 *Ultrasound transducers used to make transcutaneous blood velocity and flow recordings. A single piezoelectric disk serves as emitter and receiver of ultrasound.*

During the latent period after pulse emission a transducer is in the receive mode, and an electronic gate permits the detection of backscattered signal during a short (1–2 μsec) interval after a preset delay after emission. The net result is the receipt of Doppler shifted signal from a small sample volume a known distance in front of the transducer disk (Fig. 2). The delay of the gate was increased in increments of 0·7 mm which provided a velocity scan of the medium in front of a transducer.

The gate length and beam divergence characteristics determine the size of the sample volume and resolution of the instrument. In our laboratory detailed evaluations of transducer beam characteristics by edge scans of a constant velocity mylar tape system and beam visualization using a Schlieren optical system demonstrate a sample volume of ~ 2 mm^3 for a gate of 2 μsec using a 2 mm diameter transducer (Morris, Histand, and Miller, 1972).

The backscattered signal from a sample volume is converted by the PUDVM to an analogue voltage proportional to the Doppler frequency shift. These signals are recorded on magnetic tape using a Sanborn 3907 FM tape recorder, digitized and processed by a digital programme to yield time varying velocities, velocity profiles, and flow rates. All data are calibrated by application of the Doppler equation

$$V = \frac{\Delta f c}{2 f_0 \cos \theta}$$

where Δf = Doppler frequency shift, c = speed of sound in tissue, f_0 = emitted ultrasound frequency, and θ = angle subtended by transducer axis and flow axis. Analogue Δf calibration signals are obtained from an internal frequency generator in the PUDVM. Quantitative data are therefore obtained directly. This PUDVM distinguishes forward and reverse velocity; therefore, one may record bi-directional flow directly.

Transcutaneous measurements on dogs

Transcutaneous ultrasound measurements were made on 20 mature beagle dogs of known age and medical history anaesthetized with sodium pentobarbital 25 mg/kg. Atropine sulphate, 0·07 mg/kg, was administered subcutaneously. The dogs were maintained in the supine position and the limbs restrained. An endotracheal tube was inserted to facilitate breathing. During the entire experiment a bipolar lead II electrocardiogram was continuously monitored and recorded. The following general protocol was applied for transcutaneous arterial measurements at different skin locations. A subcutaneous artery was palpated, and the general axial direction estimated and sketched on the skin with indelible ink. A stereotaxic holder (Fig. 3) designed especially for orienting ultrasound transducers was located over the skin site. The holder permits three degrees of lateral motion measurable to 0·01 mm with verniers, and calibrated angular motion in the plane of the vessel axis. A piezoelectric crystal 2 mm in diameter was positioned over the artery by maximizing the audio and monitor oscilloscope signals. The axis of the ultrasound transducer was moved laterally, maintaining a constant angle with the vertical until the maximum signal was obtained, thus establishing the transducer in the plane of the axis of the artery. Then the angle subtended by the vertical and the transducer axis was reduced until the centreline Doppler

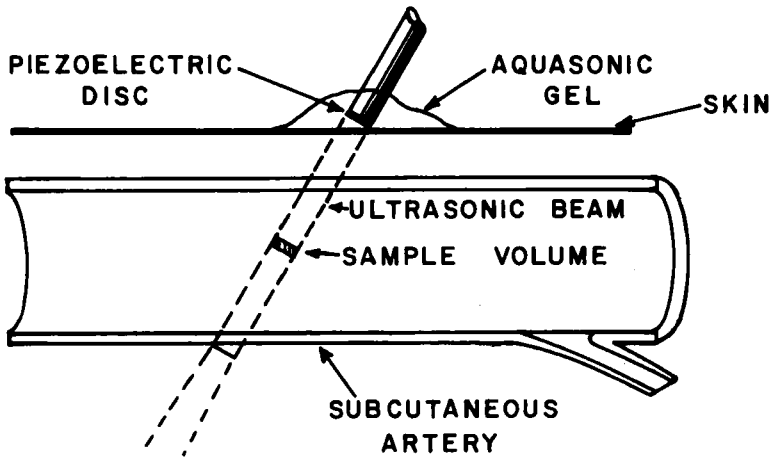

FIG. 2 *Illustration of the boundaries of the pulsed ultrasound beam through skin, tissue, and artery. By range gating the PUDVM velocity information from the small sample volume is obtained. At depths greater than 3 cm the ultrasound beam diverges appreciably when using transducers less than 3 mm diameter.*

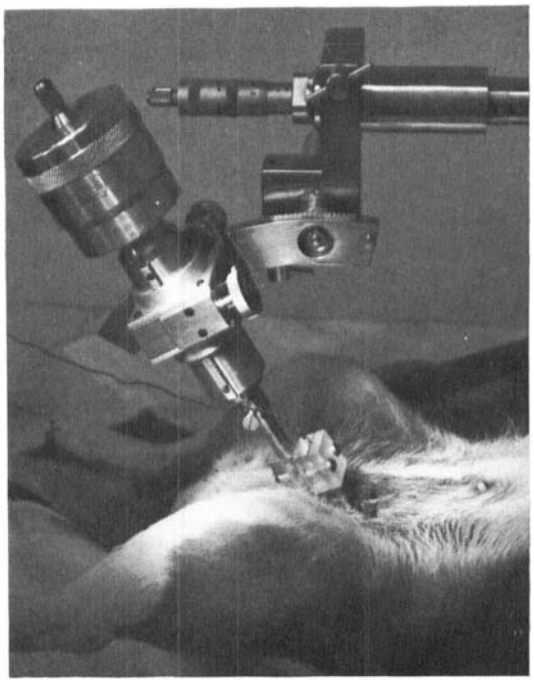

FIG. 3 *The stereotaxic holder used to position a transducer over a subcutaneous artery. The holder provides 3 degrees of lateral motion each with an accuracy of 0·01 mm, and provides an accurate measure of the angle of the transducer axis with respect to the flow axis.*

signal was minimized, thus establishing a perpendicular direction to the flow axis to be used subsequently to measure the transducer-flow axis angle required for calibration using the Doppler equation. The transducer was relocated at an accurately measured angle θ subtended by the crystal axis and vessel axis. The validity of this technique for establishing θ was corroborated during experiments on vessels surgically exposed after transcutaneous measurement. Aquasonic[3] gel was placed between the crystal face and skin surface to provide good acoustic coupling. The range gate of the PUDVM was increased in increments of 0·7 mm in a skew line across the lumen of the artery. At each intraluminal range position 20 flow velocity waves were recorded on a Sanborn 3907 FM tape recorder. Systemic arterial blood pressure was simultaneously recorded in the left femoral artery. A minimum of two lumen scans was made at each anatomical location. Transcutaneous recordings on each dog were made in the femoral artery, iliac artery, abdominal aorta, and carotid artery and on the thoracic aorta using an oesophageal probe. All data recorded on FM tape were digitized off line using a Redcor A-D conversion system triggering off the QRS complex of the EKG, and processed with a programme written for the CDC 6400 digital computer. Details of data reduction and analysis are in the following section.

Results

Time varying velocity recordings across an arterial lumen were made at a number of skin sites over subcutaneous arteries: the iliac and femoral arteries, the abdominal aorta, and the carotid artery. Figure 4 shows typical data plotted by the computer of the time varying velocities recorded in the femoral artery 5 cm distal to the aortic bifurcation. As the PUDVM was gated across the femoral artery from the near to far wall in increments of 0·7 mm (0·7 sin θ in actual

[3] Parker Laboratories, Inc., Irvington, New Jersey 07111.

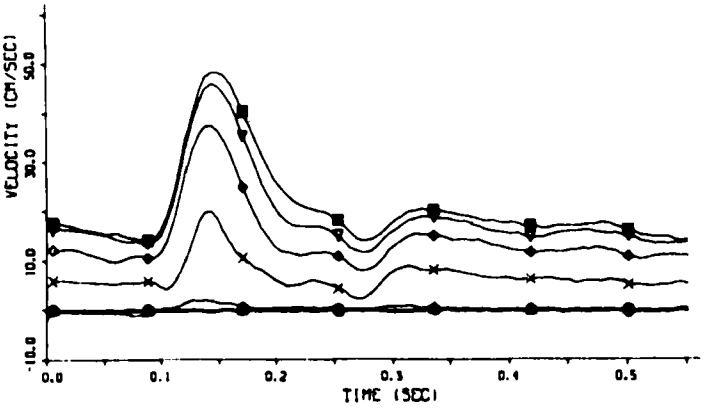

FIG. 4 *Seven time varying velocity waves recorded from the near wall* (bottom) *to the centreline* (top) *in radial increments of 0·56 mm in the femoral artery 5 cm distal from the bifurcation.*

cross-section), 20 velocity pulses were recorded at each intraluminal location. These pulses were digitized off line from the FM analogue tape and stored on digital tape files. A data processing programme written for the Control Data Corporation 6400 digital computer averaged 20 velocity pulses at each intraluminal position, calculated the variance, affixed calibration values by application of the Doppler equation for the known recording angle θ, and plotted the results on microfilm. The increase in peak velocity from a location at the vessel near wall to its maximum at a location along the centre line is clearly shown by the velocity pulses illustrated in Fig. 4. Seven time varying velocities are shown, and the additional waves recorded from the centreline to the far wall have been suppressed for clarity. During the complete cardiac cycle no reverse velocity was observed in the femoral artery. This was generally true for all the data recorded in the femoral arteries of the anaesthetized dogs.

At 21 equally spaced intervals during the cardiac cycle, velocity profiles were calculated from the velocity waves. Velocity as a function of range is shown in Fig. 5 for the same femoral artery location as Fig. 4. Each subsequent profile is separated by an added 15 cm/sec offset along the ordinate to make the shapes more apparent. From the profiles, vessel diameter can be measured and information about the symmetry and shape of the profiles at different locations can be obtained. The inside diameter of the femoral artery was approximately 4·3 mm at this location. For more exact location of the wall-blood interface smaller scan increments can be made. However, the finite size of the sample volume causes slight flattening of the velocity profiles at the walls. In straight, uniform segments of the arterial system the profiles were generally symmetrical as shown. However, near branch points, bifurcations, or sites of abrupt vessel curvature, the profiles can be distinctly skewed as shown in Fig. 6. The velocity profile skewing toward the near wall was presumably caused by curvature of the iliac artery at this location 2·5 cm from the bifurcation. This type of data is important in relating localized atherosclerosis to such haemodynamic factors as altered viscous shear stress or separation.

Assuming that a vessel segment at which symmetrical velocity profiles were obtained is straight, uniform, and circular in cross-section, time varying blood flow can be calculated, providing transcutaneous calibrated flow data. From a mathematical procedure consisting of summation of finite laminae (concentric cylinders) one can numerically integrate the velocity profiles and calculate the time varying flow at a particular arterial location. Data of this type obtained for the femoral artery location under discussion are shown in Fig. 7. There appears to be considerable flow throughout the cardiac cycle with a peak during systole. These data are in

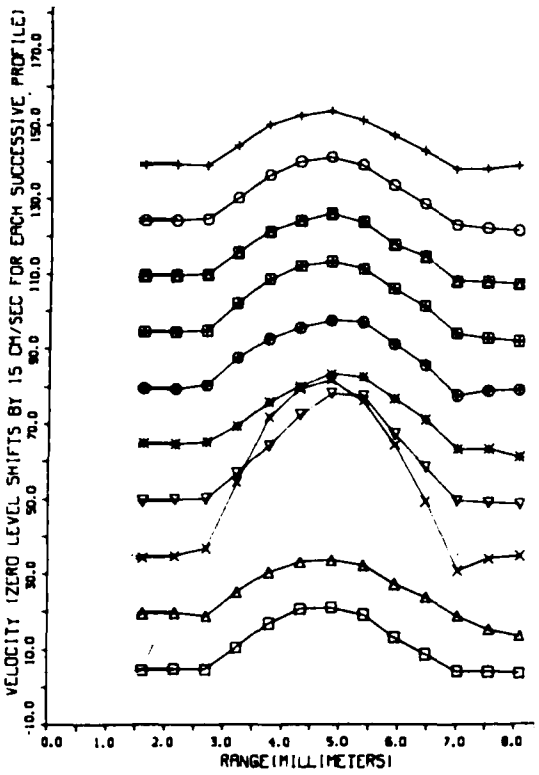

FIG. 5 *Velocity profiles calculated for 10 equally spaced time intervals during the cardiac cycle shown in Fig. 4. The bottom profile would be for $t = 0$ sec and the top for $t = 0{\cdot}55$ sec. The profiles are generally quite symmetrical if no branching or axial curvature occurs at the measuring site. The profiles are calculated from velocity data obtained by a scan diagonal to the flow. The data are then projected on the plane of a cross-section for easier interpretation.*

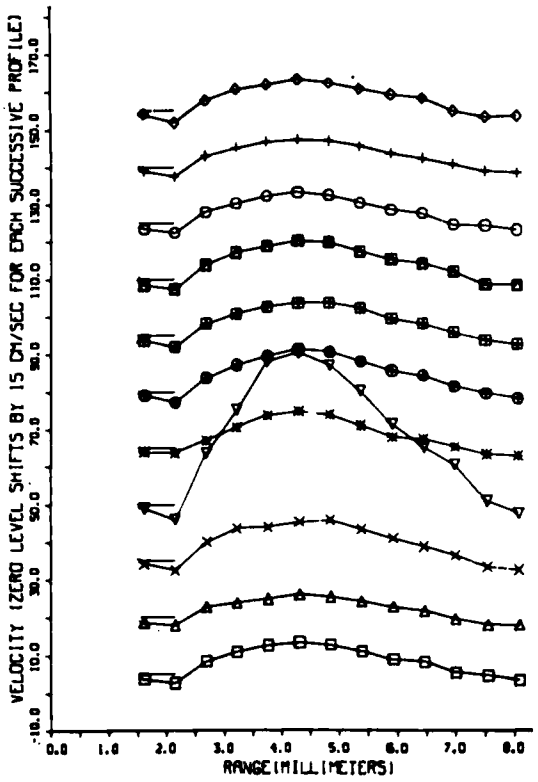

FIG. 6 *Velocity profiles skewed toward the near wall were recorded in the iliac artery of dog 8528. Presumably the skewing was caused by vessel curvature at this location 2·5 cm distal from the bifurcation as the iliac artery curves toward the leg.*

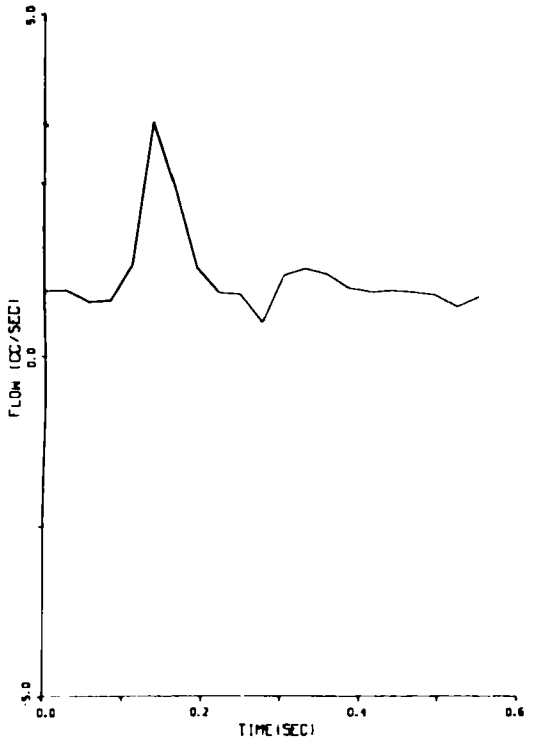

FIG. 7 *Time varying flow calculated by integrating the velocity profiles shown in Fig. 5. The peak flow was approximately 3·5 ml./sec and the mean flow somewhat in excess of 1 ml./sec.*

close agreement with flow data recorded with an electromagnetic flowmeter from 12 anaesthetized dogs (Attinger, Sugawara, Navarro, Riccetto, and Martin, 1966) indicating that the mean flow in the femoral artery was $1{\cdot}0 \pm 0{\cdot}1$ ml./sec.

Transcutaneous time varying velocity (Fig. 8), velocity profiles (Fig. 9), and flow (Fig. 10) were recorded in the terminal aorta. These velocity profiles are significantly different from those for the femoral artery. During peak flow the velocity profiles were distinctly flattened, in agreement with the results of Ling, Atabek, Fry, Patel, and Janicki (1968), obtained with a hot film anemometer in the aorta of the pig. In addition, during the immediate post-systolic period the profile exhibits some flow reversal at the wall although the centreline flow remains positive. Therefore, there is a phase difference between incipient reversal at different locations across the lumen. In general, one observes a small amount of post-systolic flow reversal in the abdominal aorta. The peak volume flow at this location 5 cm proximal from the bifurcation was approximately 10 ml./sec, which agrees closely with the peak value of 12 ml./sec recorded by Attinger *et al.* (1966) using an electromagnetic flowmeter. However, our values are about half those recorded by Rushmer, Franklin, Van Citters, and Smith (1961) with a continuous wave Doppler flowmeter on unanaesthetized dogs.

Additional transcutaneous time varying velocities (Fig. 11), velocity profiles (Fig. 12), and flow (Fig. 13) for the carotid artery are shown.

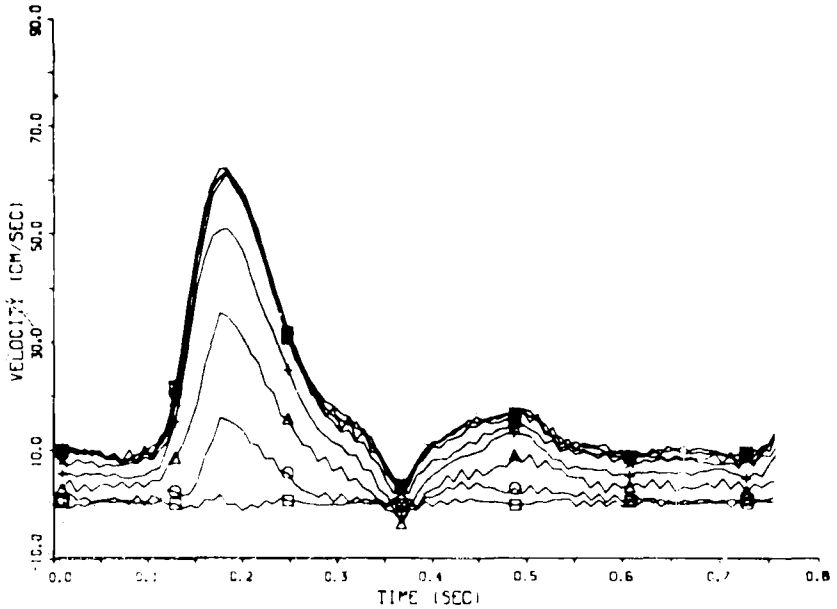

FIG. 8 *Time varying velocities recorded in a transcutaneous scan of the abdominal aorta 5 cm proximal to the bifurcation. For equivalent range increments more velocity waves than shown in Fig. 4 were recorded due to the larger inside diameter of the abdominal aorta.*

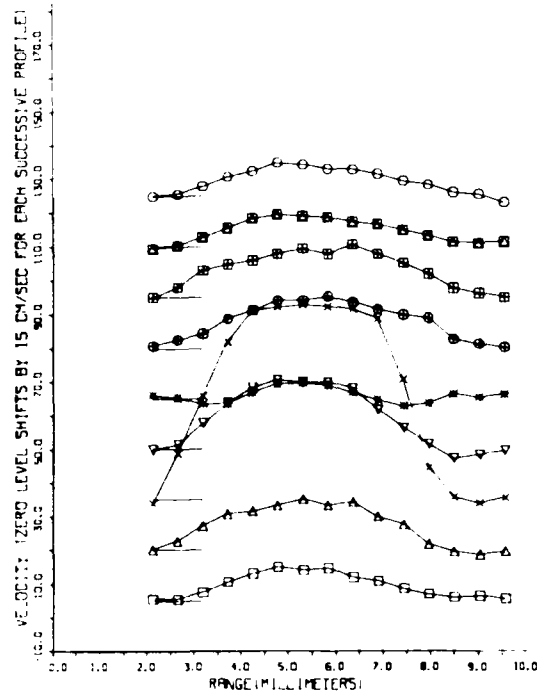

FIG. 9 *Velocity profiles calculated from the data shown in Fig. 8. During systole the profile is somewhat flattened. Also the profile (fifth from the bottom) at the end of systole shows interesting velocity reversal at the wall reminiscent of oscillatory flow in a tube.*

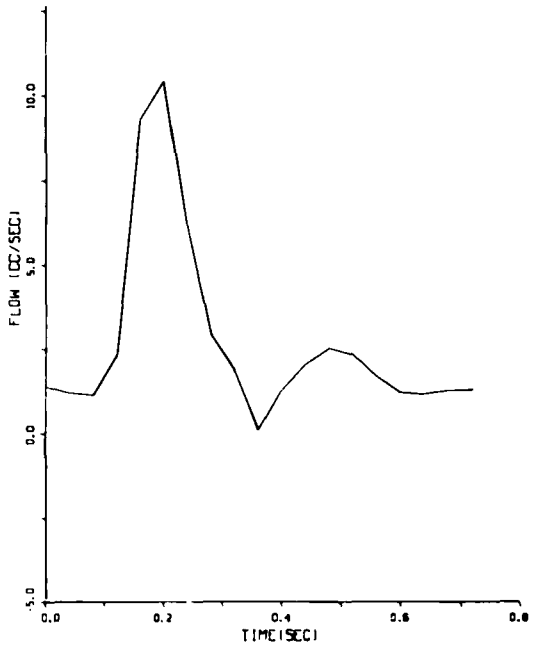

FIG. 10 *Time varying flow calculated by integration of the velocity profiles from Fig. 9. The peak flow rate 10 ml./sec was significantly higher than that generally recorded in the femoral artery.*

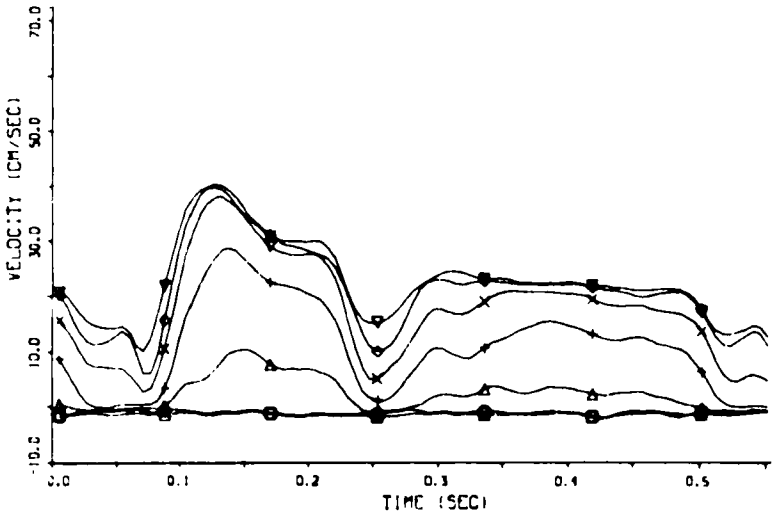

FIG. 11 *Time varying velocity waves recorded transcutaneously from the carotid artery. The waveforms during systole generally exhibited the flattened peaks.*

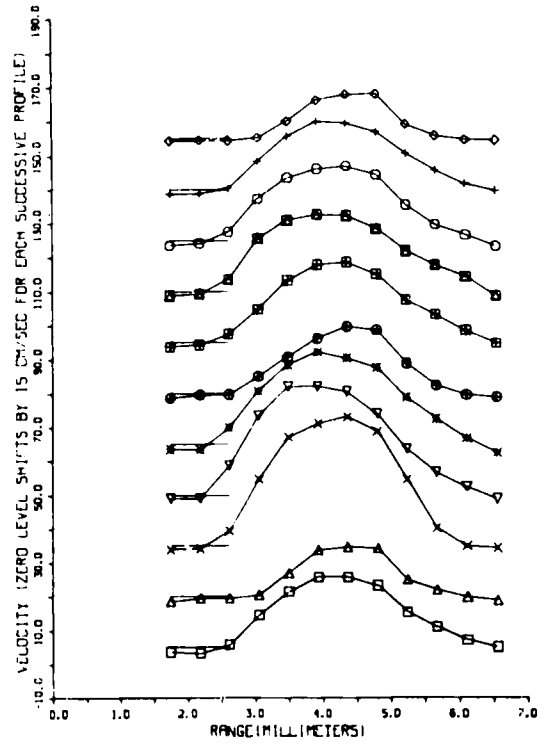

FIG. 12 *Corresponding velocity profiles for the carotid artery. Slight deviations of the axes of symmetry were caused presumably by slight curvature of the carotid artery in the neck.*

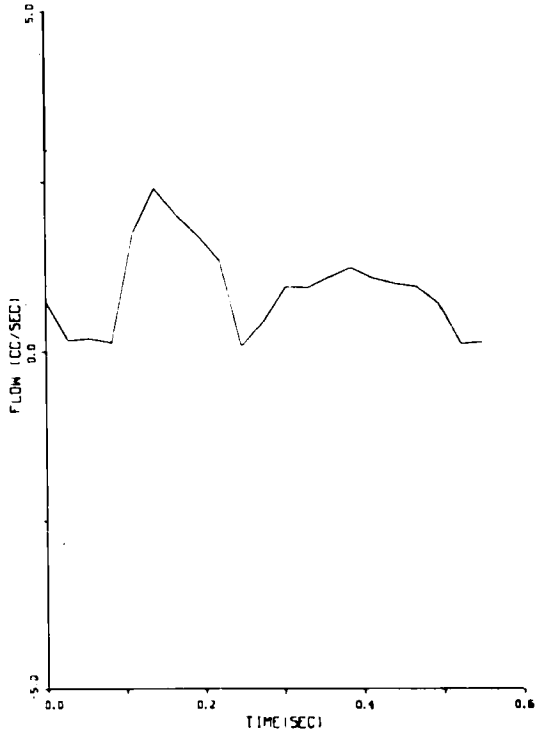

FIG. 13 *Time varying flow obtained transcutaneously in the carotid artery.*

The time varying velocity waves for the carotid were generally more flattened during the systolic interval.

Discussion

A high resolution technique for recording time varying velocity and flow transcutaneously has been described. By scanning the PUDVM sample volume across a subcutaneous artery, time varying velocities from the near to far wall were easily recorded. However, the finite resolution volume of the technique precludes recording the details of the flow in the boundary layer at the vessel wall. Although the data were averaged over 20 cardiac cycles, it was subsequently learned that only small data samples were required to yield similar velocity waves in the anaesthetized dog (pentobarbital) with a nearly constant heart rate. Averaging reduces the effects of slight changes in heart rate and blood pressure during the 2 min when one set of data is collected. A multiple gating system (McLeod, 1971) would alleviate the time problems associated with manual scans. The data show that zero velocity at the near and far walls is clearly delineated. A calculation of time delay and therefore distance between the near and far wall zero velocities provides a measure of the diameter of the vessel. Due to the finite sample volume (>1 mm^3), there is a slight broadening of the velocity profiles at the wall. This would introduce a slight error in the diameter measurement. Accurate calculations of the viscous shear stress at the wall would be very difficult due to the slight broadening of the profile at the wall. By careful sample volume correction one should be able to calculate viscous shear stresses from the transcutaneous measurements. This would be important during research relating mechanical stress to the development of atherosclerosis. Better ultrasound beam focusing techniques and smaller range increments should reduce these errors. Also, velocities are being recorded along a skewed diameter due to the requirement of a finite angle between transducer and vessel axis. A correction must be made to extrapolate to a velocity distribution in a single cross-section.

Velocity profiles were constructed by the computer from the time varying velocity data at 21 equally spaced intervals during a cardiac cycle. The number of profiles plotted can be increased if one desires more closely spaced time changes in the profiles. In straight, uniform segments of the vasculature the profiles are quite symmetrical, but at non-uniformities in geometry they often appear skewed and distorted. The non-invasive methods we describe offer distinct advantages for chronic study of the changes in haemodynamic patterns in animals or man. At the present stages of development transcutaneous measurements on deep arteries (>3 cm in depth) are difficult to obtain because of ultrasonic beam attenuation. To obtain information from the thoracic aorta, an oesophageal probe (Olson and Shelton, 1972) has been constructed, and preliminary results show the possibility of its application for measurements on thoracic vessels.

Assuming longitudinal flow in uniform vascular segments, the velocity profile may be integrated to give a measure of time varying instantaneous flow. The data presented closely correlate with the flow waves recorded by invasive techniques (Doppler ultrasound cuffs, electromagnetic cuffs). More closely spaced range gate increments would increase the accuracy of the calculated flow waves by only small amounts; but the additional time required to make the more detailed scans probably would reduce the accuracy of the data due to the gradual changes in heart rate or pressure. To be a useful technique for recording flow anywhere in the vasculature, corrections must be made to permit integration of asymmetrical profiles. It would be necessary to record a number of profiles at a given location from different orientations around the vessel circumference in order to prescribe the three dimensional surface of the velocity profile.

Conclusion

A transcutaneous method to measure time varying velocities, velocity profiles, and flow waves at a number of locations in the mammalian anatomy using a PUDVM has been described. This technique has particular significance in that it is non-traumatic and therefore applicable for measuring chronic changes in haemodynamics associated with drug studies, exercise, occlusive vascular disease, or heart disease. The method, applicable for both anaesthetized

and unanaesthetized patients provides recordings of blood velocity and flow which are far more detailed than those obtained using earlier methods. New opportunities for broadening the study of haemodynamics in both animals and man are offered. Further evaluation will lead to the application of these techniques on conscious human subjects.

References

Attinger, E. O., Sugawara, H., Navarro, A., Riccetto, A., and Martin, R. (1966). Pressure-flow relations in dog arteries. *Circulation Research*, **19**, 230–246.

Baker, D. W. (1970). Pulsed ultrasonic Doppler blood-flow sensing. *IEEE Transactions on Sonics and Ultrasonics*, SU-17, 170–185.

Bernstine, R. L., and Litt, B. D. (1971). Doppler ultrasound studies of the fetal animal. In *Recent Advances in Diagnostic Ultrasound*, pp. 133–142. Edited by E. Rand. Thomas: Springfield, Ill.

Franklin, D. L., and Van Citters, R. L. (1957). Blood flow in mesenteric artery of chimpanzee, monitored via radio telemetry. *Aerospace Medicine*, **38**, 926–931.

Ling, S. C., Atabek, H. B., Fry, D. L., Patel, D. J., and Janicki, J. S. (1968). Application of heated-film velocity and shear probes to hemodynamic studies. *Circulation Research*, **23**, 789–801.

McLeod, F. D., Jr. (1967). A directional Doppler flowmeter. In *Digest of the Seventh International Conference on Medical and Biological Engineering, Stockhom, 1967*, p. 213. Almqvist and Wiksell: Stockholm.

McLeod, F. D., Jr. (1970). Calibration of CW and pulsed doppler flowmeters. *Proceedings of the 23rd Annual Conference on Engineering in Medicine and Biology*, Washington, 1970. Vol. 12, 271.

McLeod, F. D., and Anliker, M. (1971). A multiple gate pulse Doppler flowmeter. 1971 *IEEE Sonics and Ultrasonics Symposium, Miami Beach, 1971*, p. 51, Miami.

Morris, R. D., Histand, M. B., and Miller, C. W. (1972). The resolution of the ultrasound pulsed Doppler for blood velocity measurements. *Journal of Biomechanics*. In press.

Olson, R. M., and Shelton, D. K., Jr. (1972). A nondestructive technique to measure wall displacement in the thoracic aorta. *Journal of Applied Physiology*, **32**, 147–151.

Peronneau, P., Hinglais, J., Pellet, M., and Léger, F. (1970). Vélocimètre sanguin par effet Doppler à émission ultrasonore pulsée. *L'Onde Électrique*, **50**, 369–384.

Reagan, T. R., Miller, C. W., and Strandness, D. E., Jr. (1971). Transcutaneous measurement of femoral artery flow. *Journal of Surgical Research*, **11**, 477–482.

Rushmer, R. F., Baker, D. W., and Stegall, H. F. (1966). Transcutaneous Doppler flow detection as a nondestructive technique. *Journal of Applied Physiology*, **21**, 554–566.

Rushmer, R. F., Franklin, D., Van Citters, R., and Smith, O. (1961). Changes in peripheral blood flow distribution in healthy dogs. *Circulation Research*, **9**, 675–687.

Satomura, S. (1959). Study of the flow patterns in peripheral arteries by ultrasonics. *Journal of the Acoustical Society of Japan*, **15**, 151–158.

Strandness, D. E., Jr. (1969). *Peripheral Arterial Disease*, Little, Brown: New York.

Strandness, D. E., Schultz, R. D., Sumner, D. S., and Rushmer, R. F. (1967). Ultrasonic flow detection. A useful technic in the evaluation of peripheral vascular disease. *American Journal of Surgery*, **113**, 311–320.

Van Citters, R. L. (1968). Blood pressure-flow studies in free-ranging primates and other animals. *Biomedical Sciences Instrumentation*, **4**, xvii–xxvii.