ULTRASONIC BLOOD FLOW DETECTION: DOPPLER TECHNIQUES FOR OBSTETRICS.

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This thesis describes work done by myself and was composed by myself.

Ronald McHugh
December, 1980.

CONTENTS

ABSTRACT

Ultrasonic Doppler techniques have been developed for the detection of uterine blood flow. The work was undertaken to provide a non-invasive method for the study of foetal haemodynamics.

The operation of the continuous wave and the pulsed wave Doppler instruments and the factors which influence their performance are discussed. The different types of Doppler signal extraction techniques which can be used with the pulsed wave Doppler are described. A design for a 2.5 MHz pulsed wave Doppler instrument is presented. The results of in vivo and in vitro trials with this instrument are presented.

A blood flow instrument specially designed for examining blood flow in the pregnant uterus is described. It consists of a real time ultrasonic scanner of rotating transducer design used in conjunction with the above types of Doppler instrument. In vivo evaluation of this equipment is presented.

A novel type of continuous wave Doppler instrument, the intersecting zone Doppler is described. This device overcomes the problem of lack of localisation normally associated with the continuous wave device.

A composite blood flow system incorporating all three Doppler techniques is described. Blood flow spectrograms from various sites within the pregnant uterus are presented.

CHAPTER 1

INTRODUCTION

The use of ultrasound in medicine is well established. The pulse echo technique is a useful diagnostic aid in the visualisation of human anatomical soft tissue sites. The use of the Doppler effect allows ultrasonic investigation of blood flow.

In obstetrics, the use of ultrasonic B-scanning and real time scanning offers a direct assessment of foetal growth and well-being. The foetus is an ideal subjectfor investigation. It lies floating in a bath of amniotic fluid allowing immediate ultrasonic access from most directions. Investigation of foetal heart is not hindered by obscuring lung or ribs as it is in the adult.

Little work has been done using ultrasonic Doppler instruments to investigate blood flow in the pregnant uterus. A review of the literature reveals little published work and no established technique of investigation. This is surprising when one considers the amount of effort that has been put into the development of Doppler techniques in such areas as the peripheral vascular clinic and cardiology.

Knowledge of foetal haemodynamics has been mainly derived from animal studies using invasive techniques. A few measurements of umbilical blood flow have been made in man after normal delivery or at therapeutic abortion. Flow in the human umbilical placental circulation falls rapidly after delivery and the relevance of such results is questionable.

Foetal placental circulation in the ovine species has been thoroughly investigated. These studies are all invasive and hence severely disturb the foetus and this may influence the results. Measurements of umbilical blood flow in foetal lambs at various gestational ages indicate flow rates from 130 to 200 ml $kg^{-1}min^{-1}$ for older foetuses, with higher flow rates being observed in younger foetuses.

Considerable variations occur in foetal circulation. Umbilical blood flow and foetal heart rate fall with increasing gestation. The foetal heart rate varies considerably both on a minute-to-minute basis and with normal foetal activity in utero. This must influence blood flow.

It has been well established experimentally that foreign substances introduced into the maternal circulation appear in the foetal blood, or have demonstrable effect on the foetus. The effect of stress on the foetus may be dangerous. It is important to measure the influence of these effects on foetal haemodynamics and in particular their duration and severity.

The study of human foetal haemodynamics requires a non-invasive technique which does not in any way disturb the foetal environment or modify foetal behaviour. It must be totally safe when used for long periods. Ultrasonic Doppler techniques seem to offer the key to such investigations.

1.1 THE CIRCULATORY SYSTEM IN THE FOETUS

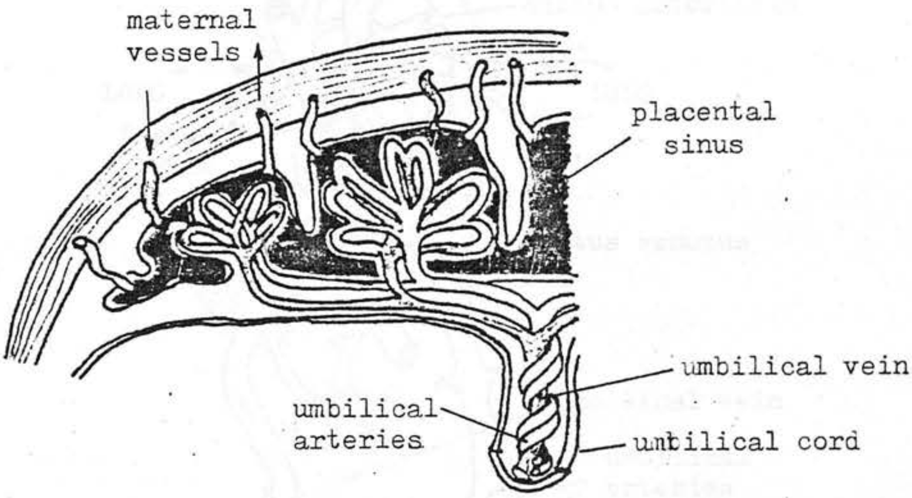
It is necessary as a medical physicist/engineer to have some understanding of basic foetal anatomy. A brief introduction is given below.

The placenta is a vascular structure, developed about the second month of pregnancy, which provides a means of transferring nutrients from the mother's blood and returning foetal waste to it for disposal. It is attached to the lining of the uterus and is part of the growing foetal tissues. Within the placenta are two distinct blood flow circulations, the maternal-placental circulation and the foetal placental circulation. The placenta acts as a semi-permeable membrane allowing diffusion of certain substances between the circulations but excluding others such as blood cells. Hence the description of the 'placental barrier'.

The mother's blood flows into large chambers called placental sinuses. The foetal portion of the placenta is composed of many small 'cauliflower-like' projections of tissue extending into these sinuses. Figure 1.1 illustrates the anatomical structure of the placenta.

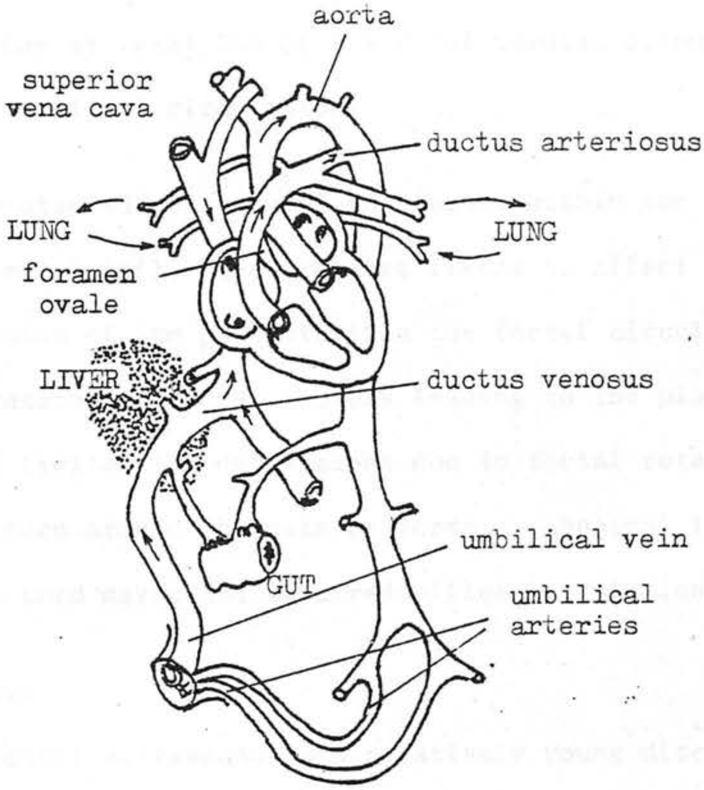
The umbilical cord connects the placenta and foetus. The cord consists of a single large umbilical vein around which are twisted the two umbilical arteries. Foetal blood flows through the umbilical arteries into the capillaries of the villi in the placenta where exchange of diffusible substances takes place. Oxygenated blood returns to the foetus via the umbilical vein and enters the foetal circulatory system through the ductus venosus. Figure 1.2 illustrates the foetal circulation.

Blood flow from the portal veins of the foetus enters the ductus venosus. The ductus venosus by-passes the liver and empties directly into the general venous system. The blood flow by-passes the fluid filled lungs. The umbilical arteries originate as major branches of the internal iliac arteries and are supplied with blood from the combined output of the right and left ventricles, with the right



Anatomical structure of the placenta

Figure 1.1



The foetal circulation

Figure 1.2

ventricle contributing the major portion. Accurate measurement of the proportion of the total cardiac output allocated to the placenta is not available in man. Data obtained using rhesus monkeys and sheep indicate that at least 50% of the total cardiac output is assigned to the umbilical circulation.

Because of the protected position of the foetus within the uterus, adverse environmental influences are more likely to affect the maternal circulation of the placenta than the foetal circulation. Mechanical compression of foetal vessels leading to the placenta are rare and are limited to compressions due to foetal rotation or wrapping of the cord around the neck or torso. Abnormal insertion of the umbilical cord may cause impaired villus circulation.

1.2 GENERAL REVIEW

The use of diagnostic ultrasound is a relatively young discipline, being introduced into obstetrics in the late 1950's (Donald et al., 1958). The use of ultrasonic cephalometry to determine foetal menstrual age is an established technique (Willocks et al., 1964). Ultrasonic placentography can directly assist in amniocentesis (C ampbell, 1974). Ultrasonic A-scan plus T-M techniques have been applied to the study of human foetal respiratory movements (Boddy et al., 1971).

The first reported use of Doppler ultrasound to measure blood velocity non-invasively is attributed to Satomura (1960). The first practical transcutaneous device was introduced by Baker (1964). Doppler ultrasound now finds application in the investigation of the central and peripheral circulations.

Strandness (1967), using a continuous wave Doppler instrument was the first to study the course of an artery in the diseased and healthy limb. The significance of changes in the blood velocity-time waveform shapes of arterial pulses at selected site on a limb was described by Woodcock (1970). Measurement of blood velocity in the thoracic aorta has been made by Light (1969). In cardiology, continuous wave (C.W.) Dopplers have been used for the detection and timing of heart valves (Lube et al., 1967). Continuous wave Doppler techniques have been used to detect tumours in breast scanning (Wells et al.,1977).

The continuous wave Doppler can be used to map blood vessels. Spencer et al. (1974) used a sharply focused probe attached to a two-dimensional position-sensing arm to translate the position of arterial flow onto an image screen. This technique has been applied to mapping the carotids (Coughlan et al., 1978).

Continuous wave Doppler instruments lack range resolution. Pulsed wave Doppler techniques allow Doppler recordings to be made at selected depths. Pulsed wave Doppler instruments were independently developed by Baker (1970), Peronneau (1969) and Wells (1969). Multiple range gate devices have been used to plot the velocity profile across vessels (Peronneau, 1972). This technique is restricted to the measurement of vessels close to the skin surface. Techniques for the study of the 'sample volume' of the pulsed wave Doppler instrument have been discussed by Baker (1972). Jorgensen et al. (1973) have shown that the size and shape of the 'sample volume' determines the sensitivity and accuracy of the velocimeter. Atkinson (1976) indicates the limitations of pulsed wave Doppler instruments operating at low ultrasonic frequencies.

Pulsed wave Doppler instruments have been used to record from the tricuspid, mitral, aortic and pulmonary valves and from within the four heart chambers (Brubakk et al., 1977 and Lorch et al., 1977). Pulsed wave Doppler mapping of blood vessels can give depth selective two-dimensional sections (Fish, 1975).

A pulsed wave Doppler instrument has been connected to a B-scan machine (Tremewan et al.,1976 and Atkinson et al., 1977). This combination allows vessel identification and accurate location of the pulsed wave Doppler's sample gate. Such a system has been used as an aid in the differentiation of dilated bile ducts from portal veins in the adult liver (Loh et al., 1978). A real time scanner and a pulsed wave Doppler have been combined (Barber et al., 1974).

Jethwa et al. (1975) describe a pulsed random signal Doppler system which uses broadband Gaussian noise as the transmitted signal. With this instrument the range resolution is determined by the bandwidth of the transmitted signal and the velocity resolution is determined by the bandwidth of the filter used to extract the Doppler frequencies.

It is often clinically important to be able to distinguish between forward and reverse flow. The first reported system for detecting the direction of flow using the Doppler technique was described by McLeod (1967). This system uses two zero crossing detectors in a time domain phase quadrature technique. It is prone to errors and can only be used with unidirectional flow (Lunt, 1975). Coughlan et al. (1976) summarise the five different directional techniques which can be used. These include three techniques based on phase quadrature detection, one on hetrodyne detection and one on single sideban detection.

1.3 REVIEW OF DOPPLER TECHNIQUES APPLIED TO OBSTETRICS

Ultrasonic Doppler instruments in obstetrics are largely restricted to use as foetal heart detectors. The movement of the foetal heart can be detected by the twelfth week of pregnancy (Bishop, 1966). Blood flow in the placenta has been recorded using a C.W. Doppler instrument (Hunt, 1969). Attempts at placental location using this approach have proved not to be reliable. The recording of intra-uterine foetal breathing movements using a C.W. Doppler has been reported by Gough et al. (1979). Such movements can also be recorded using a pulsed wave Doppler instrument (McHugh et al., 1978).

Measurement of human foetal blood flow in the aorta and intra-abdominal part of the umbilical vein has been reported by Eik-nes et al. (1980). This study used a combination of a real time linear array scanner and a pulsed wave Doppler instrument. Gill (1979) describes a technique in which a frequency-offset pulsed wave Doppler system is incorporated with the Octoson Scanner. Qualitative recordings of foetal umbilical vein blood flow have been made. McCallum et al. (1978) used a directional pulsed wave Doppler instrument to investigate the velocity waveform from flow in the umbilical artery within the cord. The angle independent pulsatility index, (P.I.), is used to characterise the velocity waveform. Measurable differences in the waveform shape, and P.I., between recordings taken from normal and some complicated pregnancies are reported. Stuart et al. (1980) and Fitzgerald et al. (1977) used a directional C.W. Doppler to record flow from the umbilical artery. They have shown that the A/B ratio, (the ratio of peak velocities during the systolic and diastolic periods), of such flow waveforms shows a decline with increasing gestational age. This was attributed to a reduction in placental resistance to blood flow with increasing gestation. Some patients with retarded intra-uterine growth showed uncharacteristic increased placental resistance.

CHAPTER 2

THE CONTINUOUS WAVE ULTRASONIC DOPPLER INSTRUMENT

All the currently available continuous wave ultrasonic Doppler instruments employ the same basic concept. A piezoelectric crystal is excited by an r.f. oscillator, (typically 1 - 10 MHz), and generates a plane acoustic wave of constant power and frequency. The probe is placed on the skin surface and reflected sound from tissue interfaces moving within the sound beam is backscattered towards the receiving transducer. The frequency of the back-scattered signal differs from that transmitted by an amount proportional to the velocity of these scatterers along the beam. The Doppler equation for a single point scatterer is given by

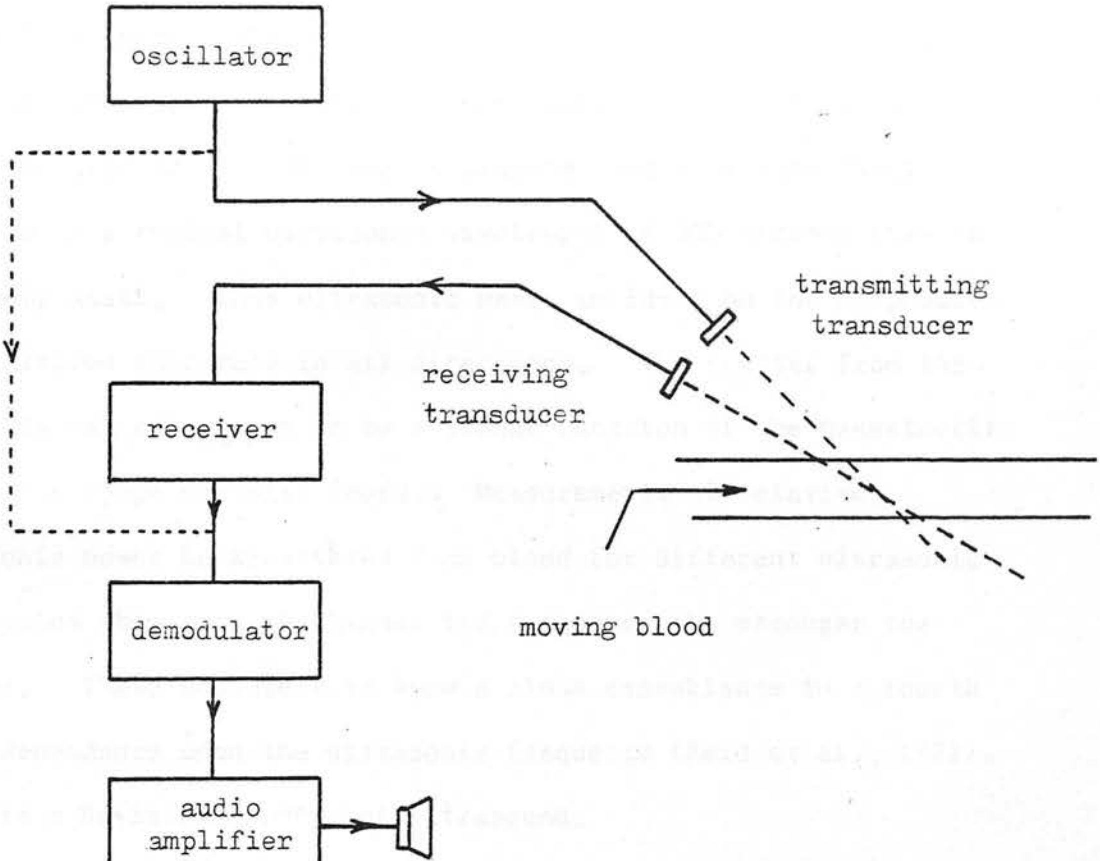
$$fd = \frac{2\ foV\ \cos\theta}{C}$$

where fd is the Doppler shift in frequency, fo is the emitted frequency, C is the velocity of ultrasound in blood, V is the velocity of the scatterer, and θ is the angle between the directions of the velocity vector and the ultrasound beam. A block diagram of the continuous wave Doppler instrument is shown in Fig. 2.1.

2.1 ULTRASONIC INSONATION OF BLOOD

In the design of any blood flow meter the scattering properties of blood must be taken into consideration. The ultrasonic power backscattered from blood is considerably less than that from tissues. The echo from a vessel wall at a suitable angle of attack is typically 20 dbs greater than the blood echo.

Blood may be considered, simply, as a suspension of red cells in



Block diagram of a continuous wave Doppler instrument

Figure 2.1

plasma (the white cells and platelets are negligible). The red cells occupy about 45% of the blood volume. A red cell is a biconcave disc about 7 microns in diameter and 2 microns thick. Compared to a typical ultrasonic wavelength of 300 microns this is extremely small. Thus ultrasonic waves incident on the corpuscles are scattered uniformly in all directions. The scatter from the red cells has been shown to be a linear function of the haematocrit within the range normally found. Measurements of relative ultrasonic power backscattered from blood for different ultrasonic frequencies show that the higher the frequency the stronger the scatter. These measurements show a close resemblance to a fourth power dependence upon the ultrasonic frequency (Reid et al., 1971). Blood is a Rayleigh scatterer of ultrasound.

The amplitude of the ultrasonic echo backscattered by blood varies with time. This arises from the random distribution of red cells. The rate of fluctuation is related only to the velocity of the scatterer and the dimensions of the resolution cell.

In blood flow through a vessel one would expect a distribution of velocities across the vessel ranging from zero flow to a maximum. In established pulsatile flow, the velocity can reverse in the outer zones whilst the central plug of forward flow is still decelerating. Laminar flow is usually regarded as having a parabolic velocity profile. The average velocity for parabolic flow is half the peak velocity.

2.2 DESIGN OF A CONTINUOUS WAVE DOPPLER INSTRUMENT

The main parameters which determine the functional characteristics of a continuous wave ultrasonic instrument are the ultrasonic frequency, the output power of the device, the transducer or probe, the Doppler shift detection circuitry, and the post-detection circuitry.

Ultrasonic frequency

In blood flow studies the choice of optimum ultrasonic operating frequency is usually decided by attenuation effects. The reflective power from blood obeys a roughly fourth power dependence on ultrasonic insonation frequency. The absorption of ultrasonic energy has an exponential dependence on range, typically 1 db/cm/MHz. Reid et al. (1971) have combined these two effects to show that a maximum power is reflected at a particular frequency which depends on the depth of penetration. As a rough guide the useful depth of penetration of Doppler instruments operating at ultrasonic frequencies of 2 MHz, 5 MHz and 10 MHz are respectively 10 cm, 4 cm and 2 cm. It is best to operate with as high an ultrasonic frequency as possible.

Output power

The power in the transmitted ultrasound beam will strongly influence the overall sensitivity of the system. In all biological uses of ultrasound the average intensity must be kept within safe limits. A figure of 5 mwatts/cm^2, or less, is often quoted as a recommended upper limit for continuous wave Doppler units.

The transducer

The overall response of the system is largely determined by the performance of the ultrasonic transducer. Continuous wave Doppler systems use separate elements for transmission and reception. Each element is acoustically and electrically isolated. For maximum sensitivity these elements are undamped piezoelectric crystals. An acoustic matching layer is often applied to the front face of a transducer to help match it acoustically to the soft body tissue.

Transducer size and geometry determine the effective beam region or sample volume of the device. This is the volume through which any moving particle will cause a detectable Doppler shifted signal. In most probes the transmitting and receiving assemblies are spaced apart or inclined to one another. The sample volume depends on the product of the transmitting and receiving zones. In the continuous wave device all the flow signals in the sample volume combine to form one signal.

<u>The Doppler shift detection circuitry</u>

The backscattered ultrasonic energy detected by the receiving transducer is amplified to a manageable level by r.f. amplifier circuitry. The frequency of the received signal will differ slightly from that transmitted, the difference being determined by the velocity of the moving scatterer. For simultaneous flow both towards and away from the transducer, the frequency spectrum of the received signal consists of a central carrier frequency at the crystal excitation frequency with sidebands above and below this frequency. The upper sideband is the result of motion towards the transducer, (conventionally referred to as positive flow), and the lower sideband of motion away from the transducer (negative flow). Separation of this velocity of motion information from the composite received spectrum is achieved by the signal processing circuits. The large signal energy at the excitation frequency is the result of leakage signals. The leakage is due to a number of effects, namely electrical coupling between transmitting and receiving circuits, acoustic coupling between the elements in the transducer, and as the result of reflected waves from static structures within the sound beam. The leakage signal caused by static structures is the most serious of these effects and can severely limit the overall performance of a device. Static reflectors can produce large fluctuating signals often many orders of magnitude greater than the blood flow signals.

There are three basic processing systems which can be used in ultrasonic Doppler instruments. The simplest is the envelope detector circuit. This finds almost universal use in the smaller portable instruments where a non-directional flow signal is required. It uses envelope or square law diode detection of the combined leakage and Doppler signal amplitudes. Its one major disadvantage is that it responds to variations in received signal strength and leakage signal variations.

The second system is the product detector. It is a coherent system in which both the amplitude and the phase information of the Doppler signal are used. A coherent reference signal is obtained from the ultrasonic transmitting circuitry. This coherent reference and the amplified Doppler returned signal are directly multiplied together to produce the Doppler shift signal. Detection by use of a single reference produces a non-directional composite Doppler signal. The advantage of this system is that if synchronous detection is carried out in two separate channels fed by phase quadrature references, these two channels can be combined to produce a directional Doppler flow velocity capability (Coughlan et al., 1976). The importance of being able to record the directional properties of flow is well established.

The third system is a coherent technique in which only the phase information of the return signal is used to extract the Doppler shifts. This coherent phase detector technique can find application in pulsed wave Doppler instruments and will be dealt with in more detail later. It is not used in continuous wave Doppler units.

Post-detection circuits

The Doppler shift frequency spectrum is proportional to the velocity of the moving reflectors. The simplest output device is an audio amplifier and loudspeaker or headphones. As the Doppler frequencies are all within the range of human hearing the ear is an ideal discriminator of flow state. The lower cut-off frequency of some machines is set to eliminate the first few hundred hertz of Doppler output. This low frequency contribution, mostly the result of vessel wall motion, can be of very large amplitude. Removing the very low frequency components allows the higher flow components to be more easily audibly discerned.

To record flow velocity on paper there are two types of instrumentation available. The first transforms the Doppler frequency spectrum into a voltage which can be recorded by an ordinary pen recorder. The most common of this type is the zero crossing detector. It produces an output voltage proportional to the root mean square frequency of the input signal. As it measures the root mean square and not the mean frequency a change in flow profile will affect the voltage output from this device even if the mean velocity remains constant (Arts et al., 1972). For this reason the zero crossing detector is little used in quantitative blood flow studies. An output voltage proportional to the maximum frequency in the Doppler spectrum can be obtained by using a maximum frequency detector (Callicot et al., 1979). Arts et al. (1972) have described a method in which a voltage proportional to the average frequency can be computed from the Doppler spectrum. This type of circuit has been used with a pulsed wave Doppler system to obtain volume flow (Gill, 1979).

The second approach is to spectrum analyse the Doppler flow signals. Full spectral analysis is a very powerful method for analysing the instantaneous velocity patterns. Off-line spectrum analysers have been commercially available for many years. They are limited in being only able to analyse short samples of signals and not in real time. Real time spectral analysis can be achieved using swept filter r.f. spectrum analysers with time compression processing of the input signal (Coghlan et al., 1974). The output is recorded as a spectrogram on a fibre optic chart recorder. Such analysers are very expensive. In situations where a lower resolution is acceptable a number of bandpass filters connected in parallel, each of equal bandwidth but increasing passband, can form a satisfactory system. The output of each filter is sequentially interrogated at a high sweep rate and the resulting signal used to z-modulate a fibre optic recorder to produce the spectrogram. An alternative to spectral analysis is the time interval histogram (Forster et al., 1978 and Baker et al., 1974). With this technique the inverse of the time interval between zero crossings of the Doppler signal is displayed as a function of time. Although such a representation does not result in a true continuous power spectrogram it can provide useful information about the peak frequency and bandwidth of the Doppler shifted signal.

CHAPTER 3

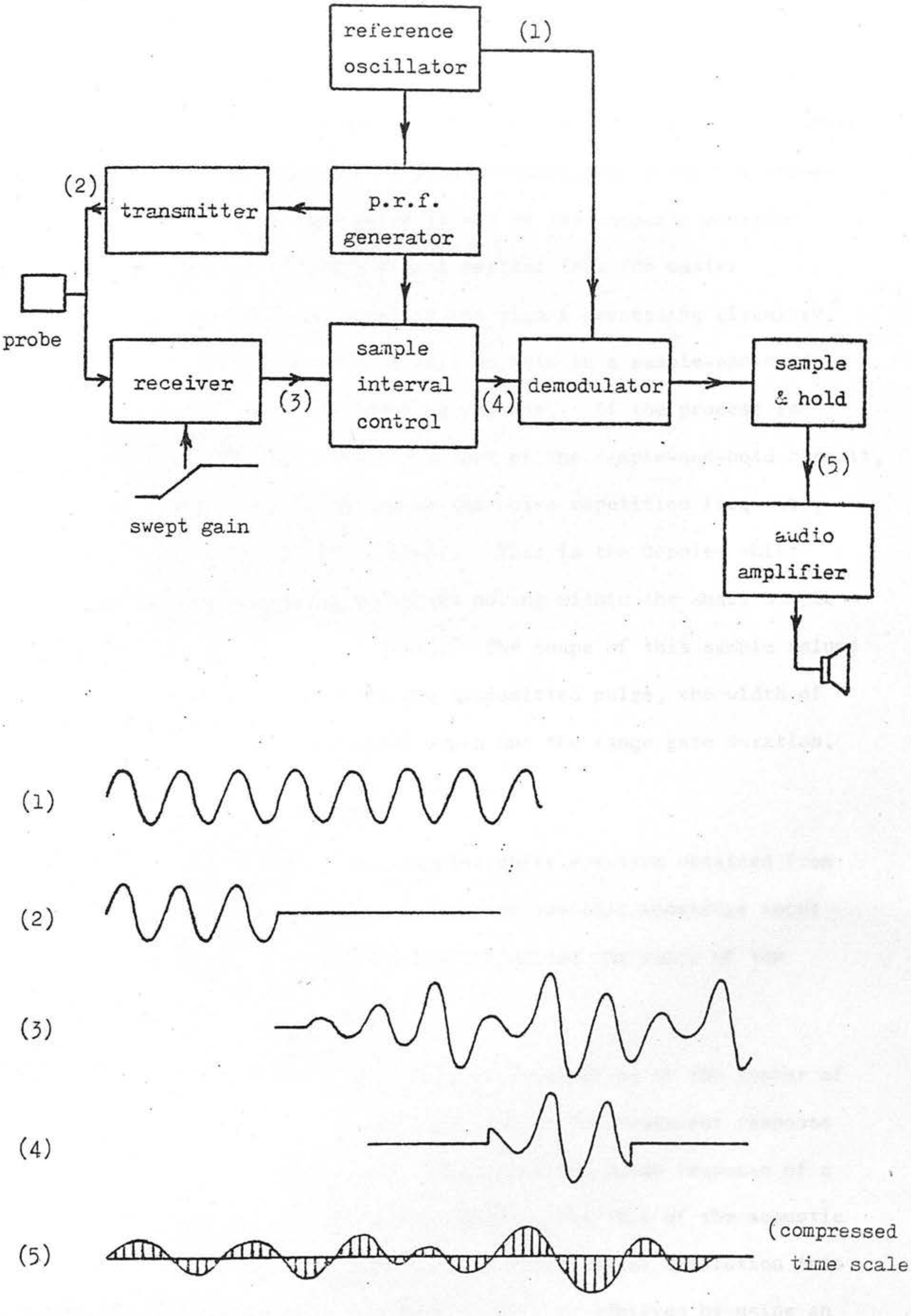
THE PULSED WAVE ULTRASONIC DOPPLER INSTRUMENT

To record blood flow within the pregnant uterus it is necessary to use a Doppler instrument capable of detecting flow to large depths, typically 10 - 15 cm from the skin surface. A 2.0 MHz continuous wave Doppler instrument, a Sonicaid D205 foetal heart detector, was found to give adequate penetration and was used initially to investigate foetal blood flow. As we wished to record flow at a specific site, a 2.5 MHz pulsed wave Doppler instrument seemed a more suitable tool. Such equipment is not available commercially. A pulsed wave Doppler instrument operating at 2.5 MHz was designed and constructed.

3.1 THE COHERENT PULSED WAVE DOPPLER

In contrast to the continuous wave device, this type of Doppler emits repetitive pulses of ultrasound at a pulse repetition frequency (p.r.f.). Between these emitted pulses, energy reflected by tissue interfaces and backscattered from moving blood is received by the transducer. Electronic gating allows echoes from a defined depth to be selected and their Doppler shift measured.

A block diagram of a coherent pulsed wave Doppler instrument is shown in Figure 3.1. The instrument uses a single damped transducer to act alternately as emitter and receiver. The transducer is excited by a short burst of electrical energy at the master oscillator frequency. Echoes arriving back at the receiver are amplified by r.f. amplifiers having a depth compensation or swept gain function.



Coherent pulsed wave Doppler instrument and associated waveforms

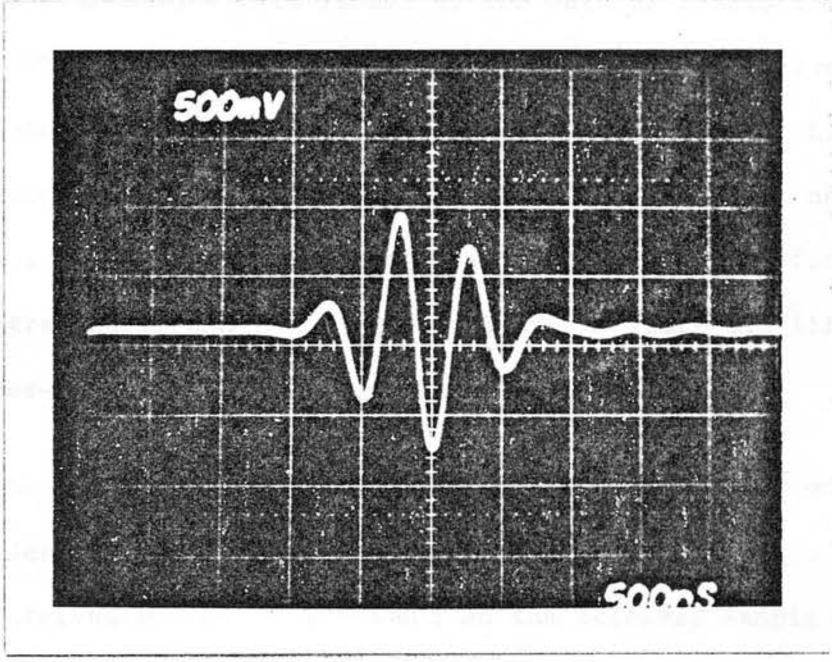
Figure 3.1

The amplified signal passes into the signal processing circuitry. Range information is obtained by range gating this signal. A sample gate closes for an instant at a predetermined time after the transmission pulse. This time delay is set by the sampling position control. This signal and a signal derived from the master oscillator are coherently mixed in the signal processing circuitry. The output voltage from this circuit is held in a sample-and-hold circuit between successive sampling periods. If the process is repeated at an adequate rate the output of the sample-and-hold circuit, after low pass filtering to remove the pulse repetition frequency, represents the Doppler shift signal. This is the Doppler shift due to the velocity of the particles moving within the small sample volume set at the range gate depth. The shape of this sample volume is influenced by the length of the transmitted pulse, the width of the ultrasonic beam at the given depth and the range gate duration.

3.2 THE ACOUSTIC SAMPLE VOLUME

Accurate interpretation of the Doppler shift spectrum obtained from a pulsed wave Doppler instrument required specific knowledge about the acoustic sample volume geometry throughout the range of the ultrasonic beam.

The axial length of the sample volume is controlled by the number of cycles of the transmitted pulse burst and by the transient response of the transducer. Figure 3.2 illustrates the pulse response of a typical pulse echo ultrasonic transducer. The tail of the acoustic burst is due to transducer ringing. For best axial resolution this ringing should be kept to a minimum. This is achieved by using an



Received voltage pulse from an
Aerotech 2·25 MHz pulse echo transducer

Figure 3·2

ultrasonically damped transducer.

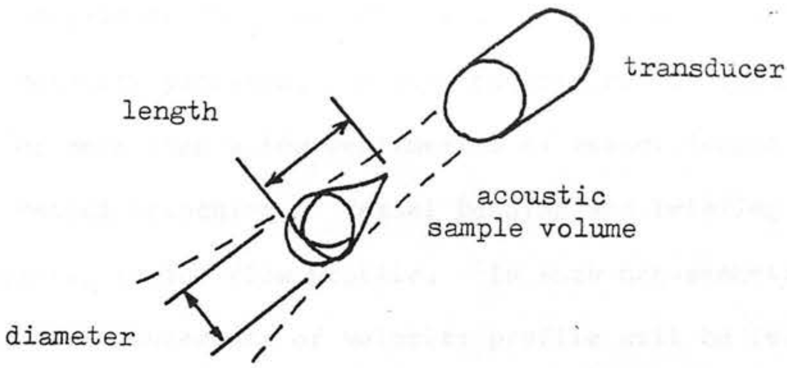
In addition to its finite length, the sample volume has a finite width or diameter. This is set by the lateral resolution of the beam. This is the distance off axis at which the received signal from blood is equal to the noise level of the system. Lateral resolution is typically quoted as the 20 db down point on the ultrasonic beamwidth. Beamwidth is a function of the focusing of the transducer and is range dependent. Figure 3.3 illustrates the three-dimensional geometry of the sample volume.

Range information is obtained by range gating the received signal at the desired depth. The actual spatial distribution of this gated received energy is dependent on the receiver sample duration and on the acoustic sample volume.

One effect of the finite volume of the acoustic sample is that velocity profile measurements across vessels show an overrun at the far walls of the vessel. This effect is illustrated by Figure 3.4 which shows velocity profile plots obtained across a vessel having a parabolic flow profile. Pulsed wave Doppler profiles with ratios of sample volume length to vessel diameter of 0.3 and 0.5 are shown. Such plots indicate a vessel larger than the true vessel by an amount equal to the sample volume length.

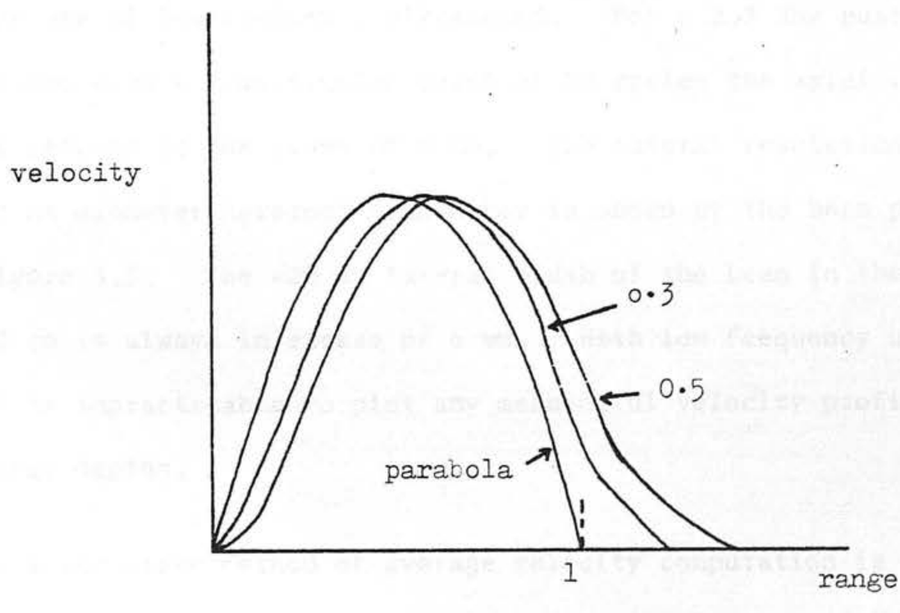
3.3 VOLUME FLOW

In order to compute volume flow in $cm^3\ sec^{-1}$ the instantaneous velocity of flow, averaged over the cross-section of the blood vessel, must be calculated. To calculate this average velocity one technique



Schematic of the sample volume geometry
(after Jorgensen et al 1971)

Figure 3.3



Parabolic profile and accompanying pulsed wave Doppler profiles with ratios of sample volume length to vessel diameter of 0·3 and 0·5.

(after Baker et al 1972)

Figure 3.4

would be to measure the velocity profile as a function of time. The blood vessels of the body present a whole range of very different velocity profiles. A flow pattern rarely remains constant for more than a few centimetres of vessel length due to continual vessel branching. Vessel bending and twisting will cause assymetry in the flow profile. In such non-symmetrical flow profiles measurements of velocity profile will be required on more than one axis through the vessel.

The ability of the pulsed wave Doppler to measure velocity profile depends on the relative size of the sample volume and the blood vessel. At large depths the sample volume can become quite significant. This is due to reduced lateral and axial resolutions, the result of the use of low frequency ultrasound. For a 2.5 MHz pulsed Doppler system with a transmission burst of 10 cycles the axial resolution is reduced to the order of 6 mm. The lateral resolution for a 2.5 MHz, 13 mm diameter Aerotech transducer is shown by the beam plot of Figure 3.5. The -20 db lateral width of the beam in the range 2 to 12 cm is always in excess of 6 mm. With low frequency ultrasound it is impracticable to plot any meaningful velocity profiles at large depths.

An alternative method of average velocity computation is to use the whole lumen averaging technique (Baker, 1972). Here the Doppler's transmitted pulse width and sample volume are made large enough to completely cover the vessel lumen. This ensures that the scattering power is composed of an equal contribution from all points across the vessel. In this case the average Doppler frequency shift is

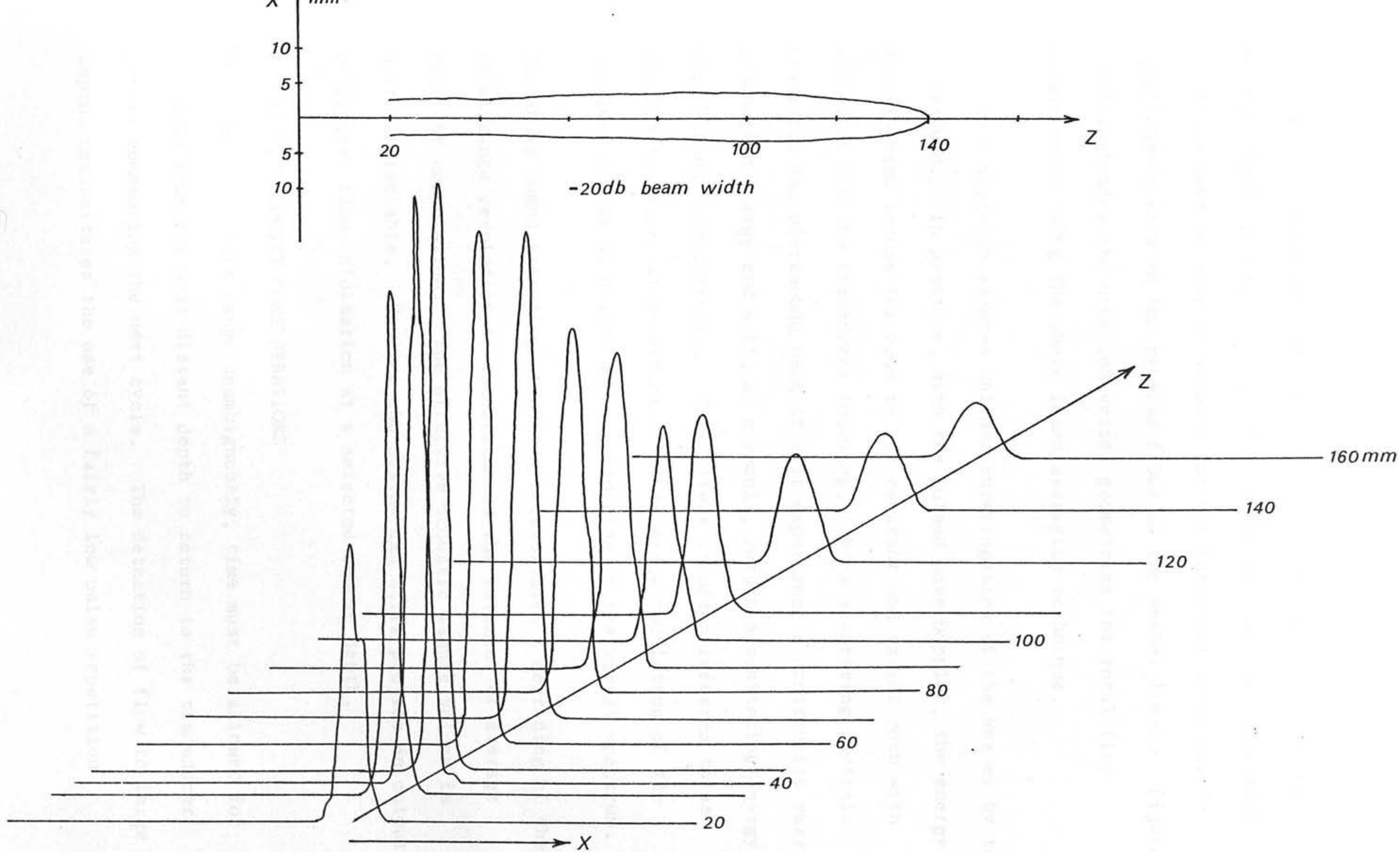


Figure 3.5 Beam plot of a 13 mm diameter 2·25 MHz pulse echo transducer.

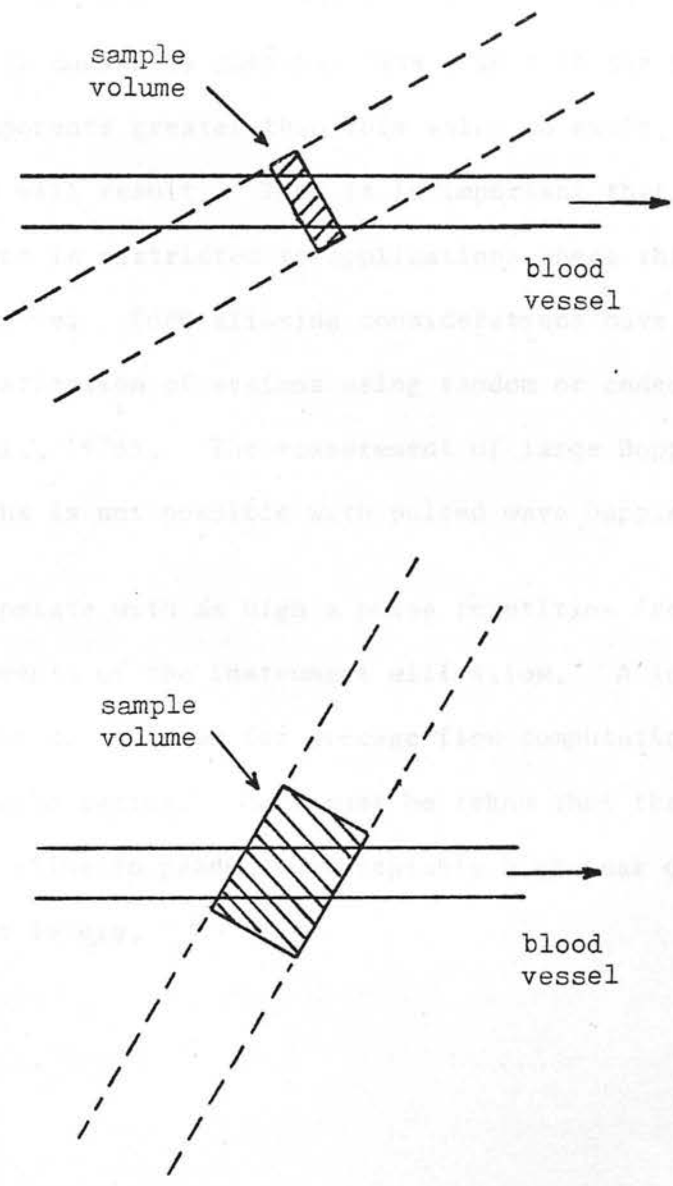
proportional to the average velocity regardless of the velocity profile (Brody et al., 1974). The signal or spectral processing technique must be able to account for the fact that there may be simultaneous forward and reverse flows in the vessel lumen. Figure 3.6 illustrates the beam and vessel geometries for total flow measurements using the whole lumen averaging technique.

The above approach assumes uniform interrogation of the vessel by the ultrasound. In practice, with the pulsed wave Doppler, the energy distribution across the beam is not constant and varies both with the depth and the transducer focusing. As a scattering particle traverses the ultrasound beam it will experience a continually varying insonation energy and will, as a result, return a scattering energy which changes accordingly. This effect is often referred to as the finite-transit-time effect. It causes a modulation of the received signal with time and a broadening of the signal spectrum.

The whole lumen averaging technique necessitates a degrading of the axial range resolution of the system in the interest of average velocity computation. The effective acoustic sample volume is quite appreciable. However the system can still provide an output of Doppler flow information at a selected sample depth.

3.4 RANGE/VELOCITY CONSIDERATIONS

In order to measure range unambiguously, time must be allowed for the echo from the most distant depth to return to the transducer before commencing the next cycle. The detection of flow to large depths necessitates the use of a fairly low pulse repetition



Beam and vessel geometries

Figure 3.6

frequency. As the pulsed wave Doppler is a sampling system the p.r.f. determines the maximum Doppler shift frequency which can be detected. This is nominally somewhat less than half the p.r.f. If Doppler shift components greater than this value do exist, erroneous recordings will result. Thus it is important that the use of this instrument is restricted to applications where this condition does not arise. Such aliasing considerations have prompted recent investigation of systems using random or coded signals (Jethwa et al., 1975). The measurement of large Doppler shifts at large depths is not possible with pulsed wave Dopplers.

It is desirable to operate with as high a pulse repetition frequency as the depth requirements of the instrument will allow. A long transmitter excitation is required for average flow computations and good signal to noise ratios. Care must be taken that the two factors do not combine to produce unacceptably high peak or mean ultrasonic power levels.

CHAPTER 4

CIRCUIT DESIGN FOR A PULSED WAVE DOPPLER INSTRUMENT

This chapter describes the electronic design for a non-directional pulsed wave Doppler instrument operating at an ultrasonic frequency of 2.5 MHz. Such an instrument is not available commercially.

The instrument can be considered as consisting of several distinct blocks of electronic circuitry as illustrated by Figure 4.1. Each of these electronic blocks with the exception of the demodulator circuitry will be considered in this chapter. Signal extraction techniques and the various types of demodulator circuit which can be used with the pulsed wave Doppler are dealt with in Chapter 5.

4.1 THE PULSED WAVE DOPPLER DESIGN

Reference oscillator (Figure 4.2)

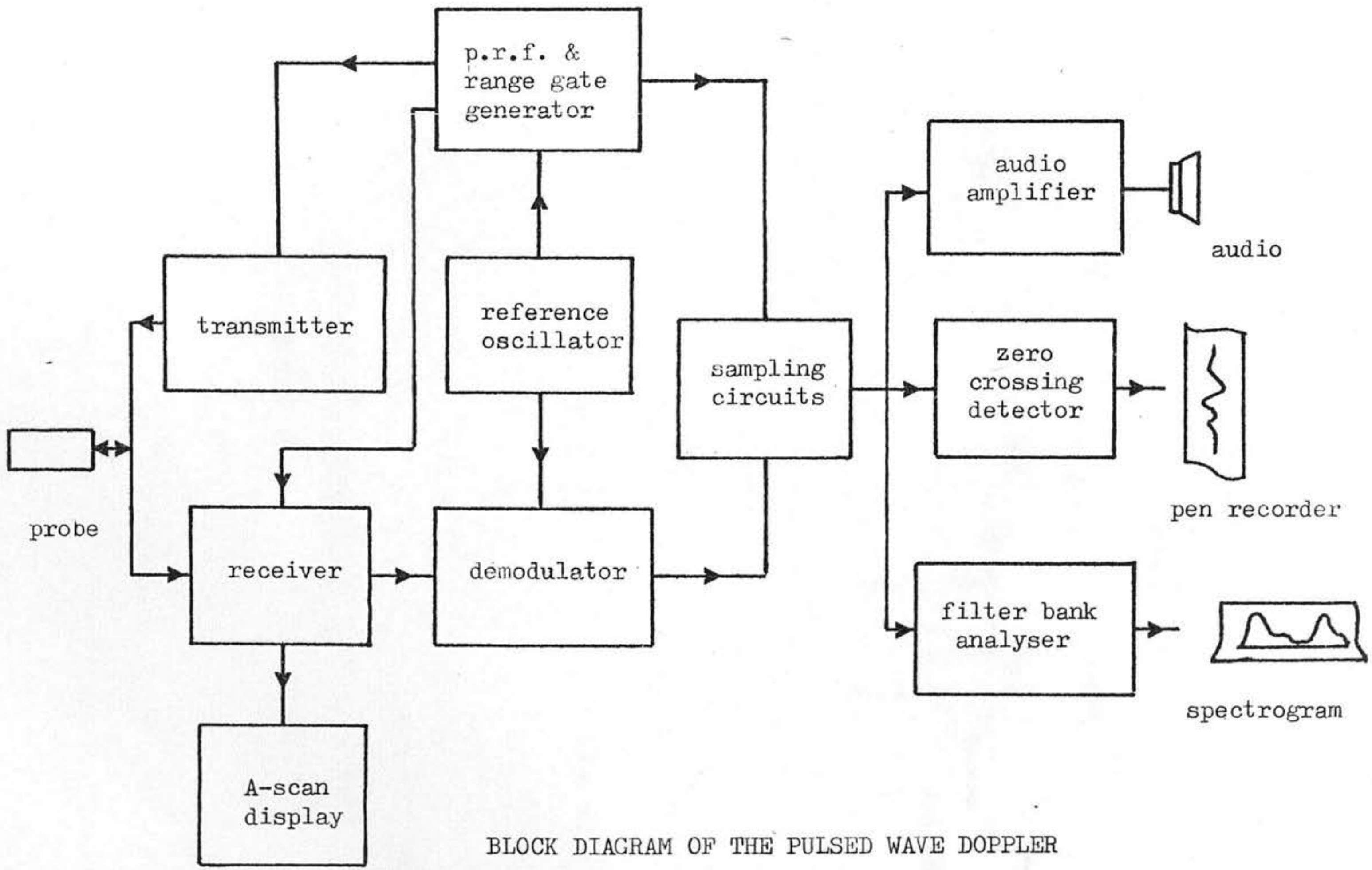
This circuit consists of a Colpits f.e.t. oscillator inductively tuned to the central resonant frequency of the ultrasonic transducer. This oscillator provides the continuous reference sinusoidal signal for coherent demodulation. A T.T.L. clock waveform derived from the zero crossings of this signal is also produced.

Pulse repetition frequency (p.r.f.) and range gate generation (Figure 4.3)

Any of three separate p.r.f's can be selected. These are:

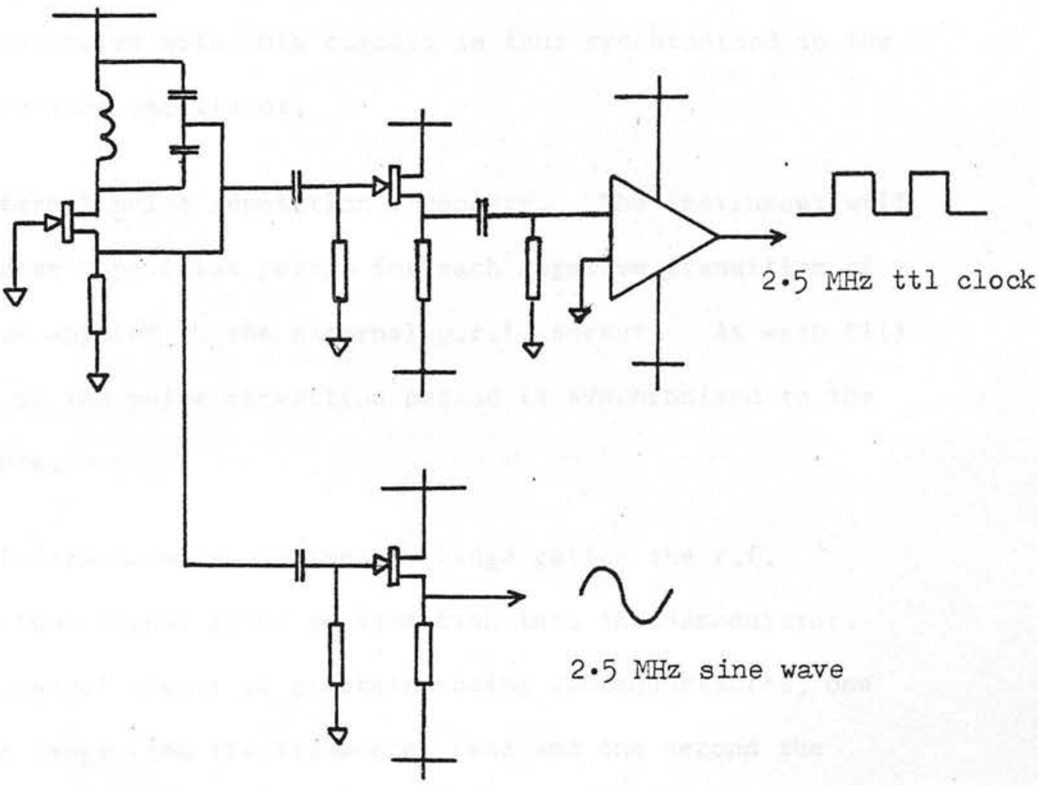
(i) A fixed clock of 3.90625 kHz. This clock is derived by dividing down the 2.5 MHz clock by a factor of 640 using T.T.L. integrated circuit decade and binary counters.

(ii) A highly stable, frequency adjustable clock oscillator.



BLOCK DIAGRAM OF THE PULSED WAVE DOPPLER

Figure 4.1



Reference oscillator

Figure 4.2

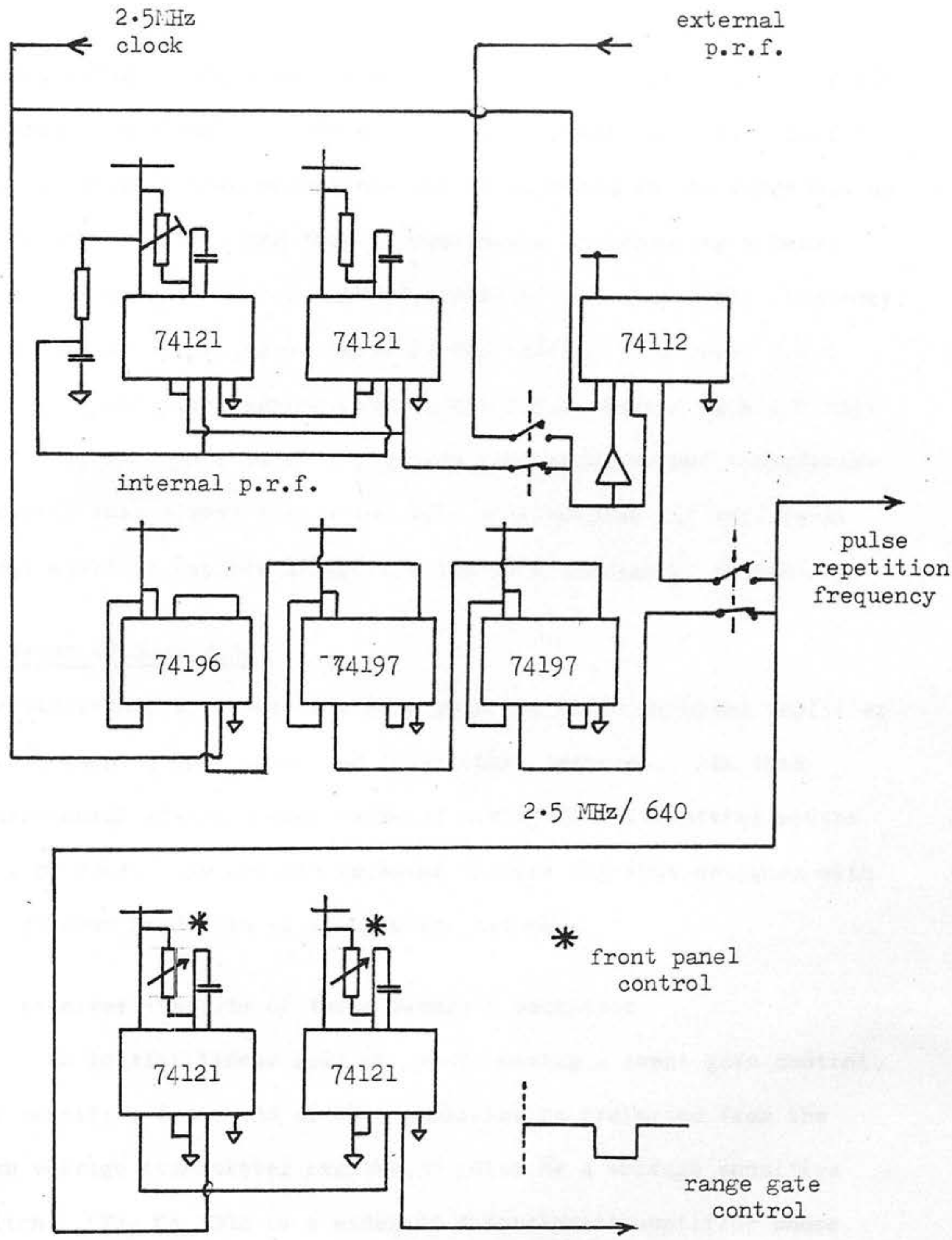
This circuit used two integrated circuit monostables to produce a pulse repetition frequency which can be adjusted to lie in the range 2.0 kHz to 4.0 kHz. The leading edge of this asynchronous waveform is locked onto the rising edge of the 2.5 MHz clock waveform using a D-type flip-flop. Each pulse repetition frequency generated with this circuit is thus synchronised to the 2.5 MHz reference oscillator.

(iii)An external pulse repetition frequency. The instrument will output a pulse repetition period for each negative transition of a T.T.L. pulse applied to the external p.r.f. socket. As with (ii) the timing of the pulse repetition period is synchronised to the 2.5 MHz clock.

The depth information is obtained by range gating the r.f. receiver output signal prior to insertion into the demodulator. The gate control signal is generated using two monostables, one setting the range from the transducer face and the second the duration of the sampling period. The sample depth can be adjusted in the range 1 cm to 20 cm by a front panel potentiometer. The duration of the sampling period can be adjusted to accommodate a tissue width of between 0.2 cm and 2.0 cm.

Transmitter (Figure 4.4)

In a coherent detection system each transmitted acoustic pulse must have an identical phase relationship to the reference oscillator. For this reason the transmitted pulse is derived directly from a gated version of the 2.5 MHz clock oscillator.



Pulse repetition frequency
and range gate control generation

Figure 4.3

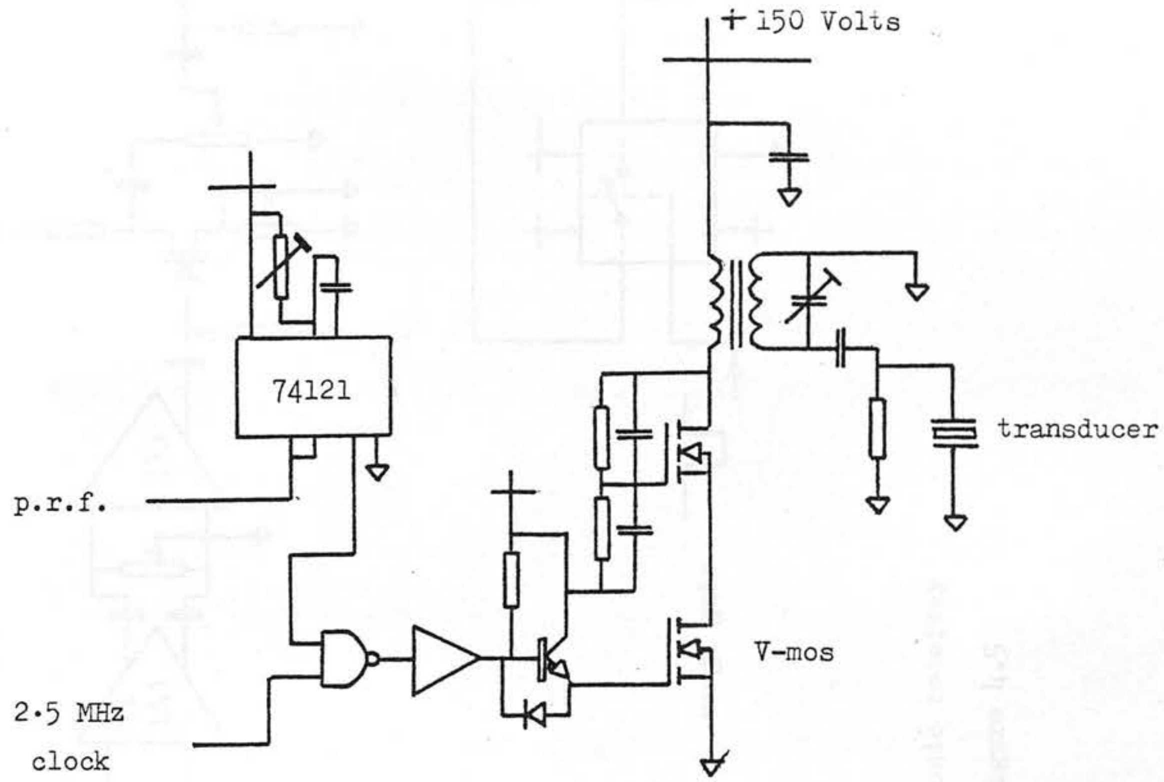
The duration of the transmitter excitation is determined by a T.T.L. monostable activated at the start of each pulse repetition period. The duration of this monostable can be adjusted in the range 0.4 μs to 12 μs. For 2.5 MHz this is equivalent to producing a burst length of between 1 cycle and 40 cycles of the ultrasonic frequency. This n-cycle T.T.L. clock burst is fed into a V-mos power f.e.t. switching amplifier which converts the T.T.L. pulses to a 120 Volt peak-to-peak square wave. A narrow band tuned output transformer converts this signal into a 100 Volt peak-to-peak 2.5 MHz sinusoidal waveform capable of driving the 50 Ω ultrasonic transducer.

Receiver (Figure 4.5)

The ultrasonic receiver is a high gain low noise wideband amplifier having swept gain control and logarithmic response. In this experimental system a wide range of ultrasonic transmitter widths will be used. An untuned receiver circuit was thus designed with a 3 db down bandwidth of 1.5 MHz to 5.5 MHz.

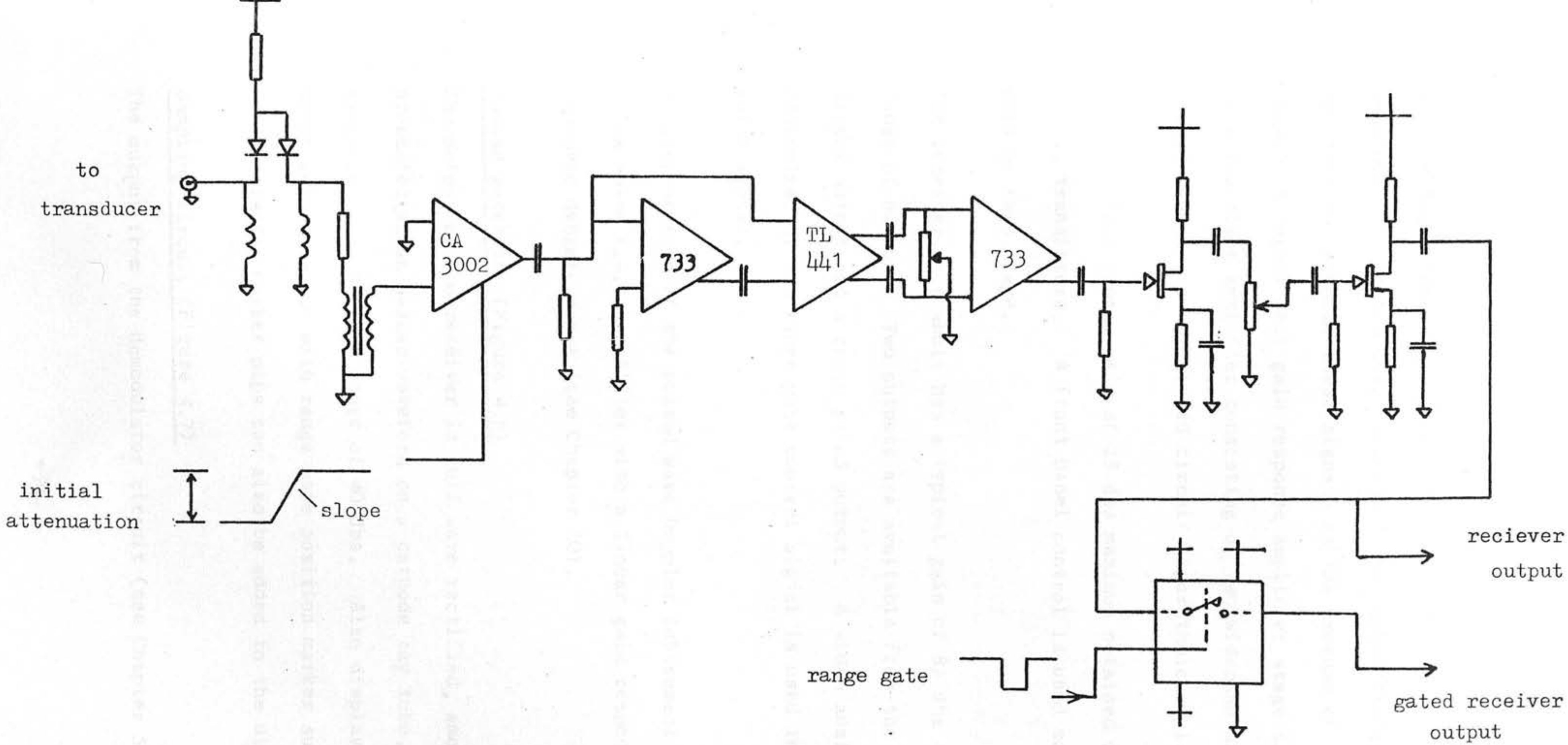
The receiver consists of three separate sections:

(i) An initial linear gain of 24 dbs having a swept gain control. The sensitive front end of this amplifier is protected from the high voltage transmitter excitation pulse by a voltage sensitive switch. The CA 3002 is a wideband differential amplifier whose gain can be adjusted by altering the D.C. bias applied to one of its inputs. The gain can be made to adjust linearly with voltage over a zero to 60 db range. A ramp type swept gain waveform is used. Two front panel potentiometers give control of initial attenuation and gain slope.



Ultrasonic transmitter

Figure 4.4



Ultrasonic receiver

Figure 4.5

(ii) A logarithmic amplifier. The Doppler information is mostly obtained in the low level echoes of the backscattered signal. In order to enhance these signals at the expense of the high level signals a logarithmic gain response amplifier stage is used. This is a low noise amplifier consisting of two wideband differential amplifiers and an integrated circuit logarithmic amplifier.

(iii) A final linear gain of 25 dbs maximum obtained using two f.e.t. transistors. A front panel control is used to adjust the gain of this stage.

The receiver as a whole has a typical gain of 82 dbs and a dynamic range of 60 dbs. Two outputs are available from the receiver, a direct output and a range gated output. A video analogue switch controlled by the range gate control signal is used to produce the gated signal.

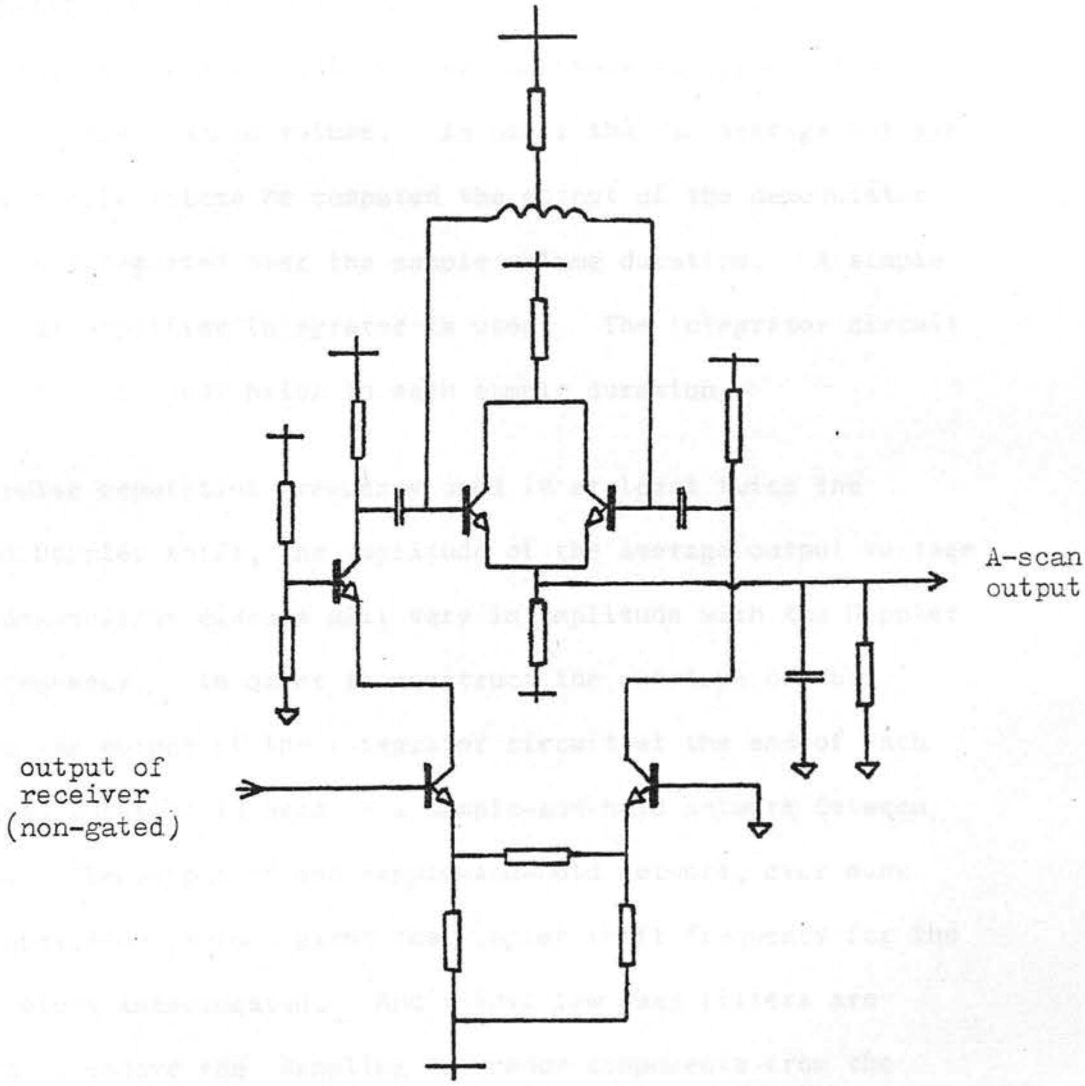
A later version of the pulsed wave Doppler instrument incorporates a low noise f.e.t. amplifier with a linear gain response and improved dynamic range (see Chapter 10).

<u>A-scan generator (Figure 4.6)</u>

The output of the receiver is full wave rectified, smoothed and presented as an A-scan waveform on a cathode ray tube. This video waveform has a dynamic range of 40 dbs. Also displayed is the swept gain waveform with range gate position marker superimposed. One centimetre marker pips can also be added to the display.

<u>Sampling circuit (Figure 4.7)</u>

The output from the demodulator circuit (see Chapter 5) is fed into



A-scan detector

Figure 4.6

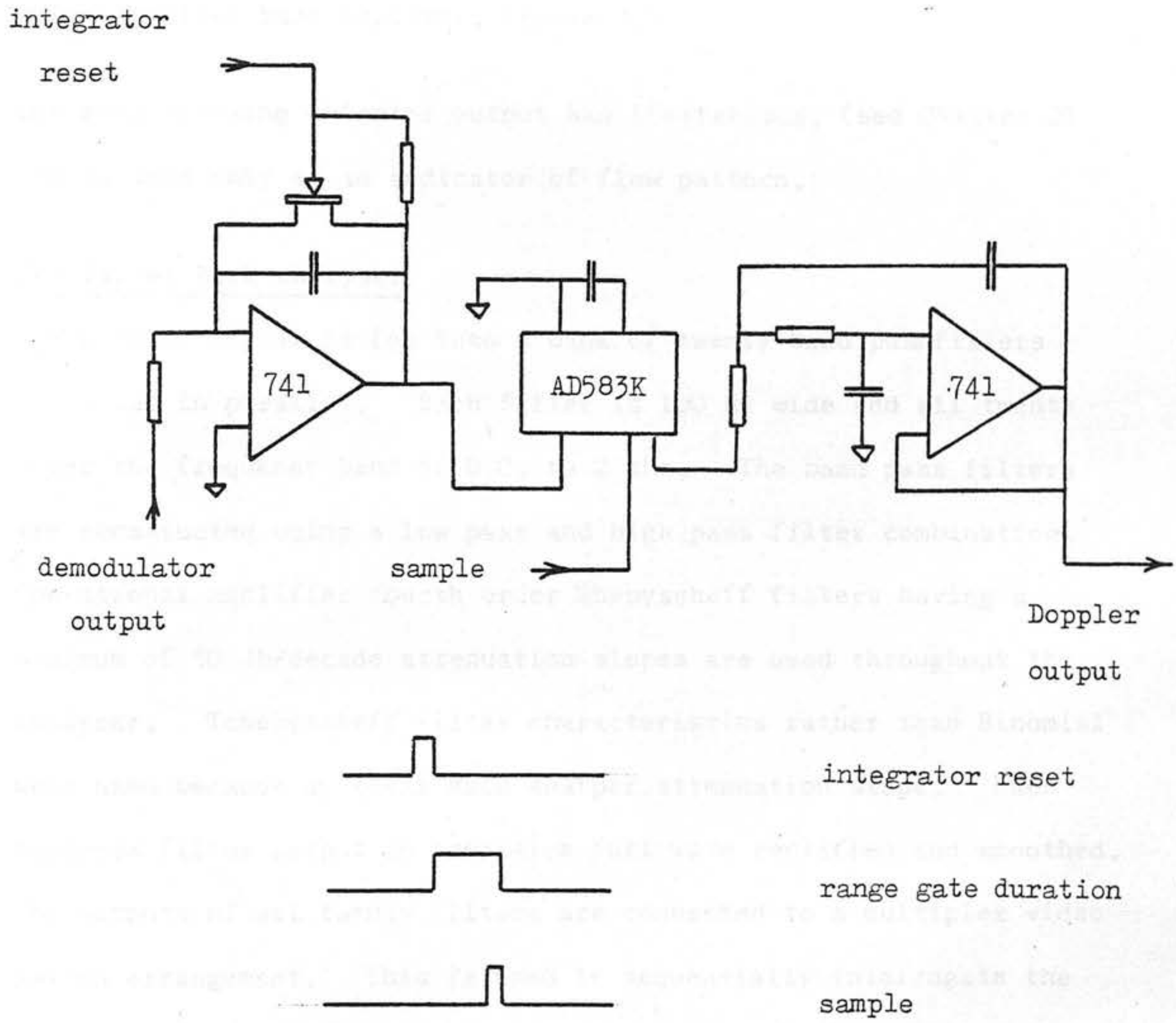
the sampling circuits. The demodulator output voltage, for a Doppler shifted signal, will vary in amplitude throughout the duration of the sample volume. In order that an average voltage over the sample volume be computed the output of the demodulator circuit is integrated over the sample volume duration. A simple operational amplifier integrator is used. The integrator circuit is reset to zero just prior to each sample duration.

As the pulse repetition frequency used is at least twice the expected Doppler shift, the amplitude of the average output voltage of the demodulator circuit will vary in amplitude with the Doppler shift frequency. In order to construct the envelope of this function the output of the integrator circuit at the end of each range gate duration is held by a sample-and-hold network between samples. The output of the sample-and-hold network, over many pulse repetition periods gives the Doppler shift frequency for the sample volume interrogated. Additional low pass filters are required to remove the sampling frequency components from the Doppler signal. Operational amplifier fourth order Tchebyscheff low pass filters having an 80 db/decade attenuation slope are used. For the p.r.f. of 3.906 kHz these filters are set to 1.75 kHz (0.45 x p.r.f.). When the 3 kHz clock is used, separate low pass filters are automatically selected.

4.2 RECORDING OF THE DOPPLER VELOCITY INFORMATION

As well as the audio channel the Doppler information can be presented in two other forms, namely

(i) as the voltage output from a zero crossing detector, or



Sampling circuits

Figure 4.7

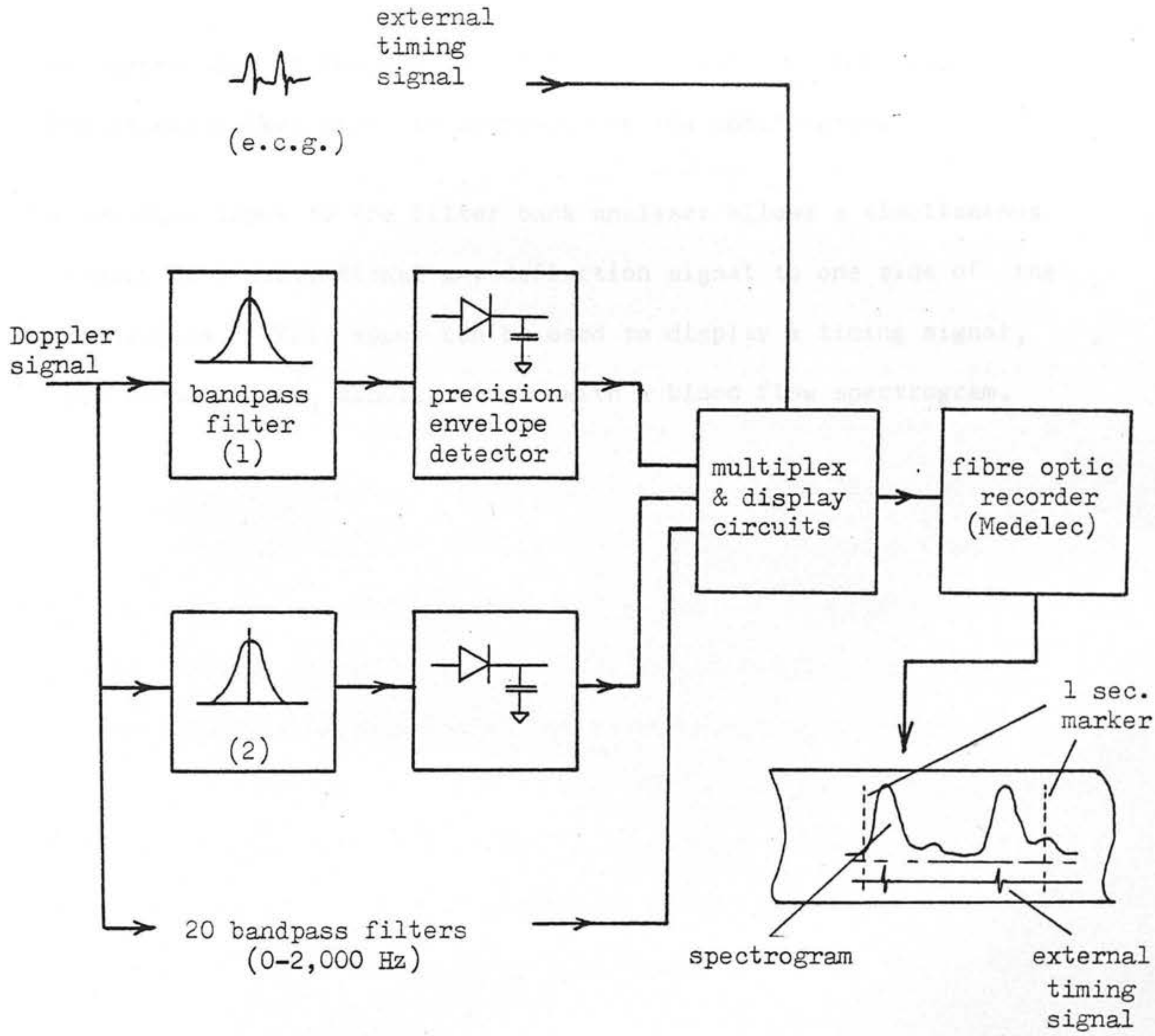
(ii) as a real time spectrogram displayed on a fibre optic recorder using a filter bank analyser, Figure 4.8.

The zero crossing detector output has limitations, (see Chapter 2) and is used only as an indicator of flow pattern.

<u>The filter bank analyser</u>

The Doppler signal is fed into a bank of twenty band pass filters connected in parallel. Each filter is 100 Hz wide and all twenty cover the frequency band of D.C. to 2 kHz. The band pass filters are constructed using a low pass and high pass filter combination. Operational amplifier fourth order Tchebyscheff filters having a maximum of 80 db/decade attenuation slopes are used throughout the analyser. Tchebyscheff filter characteristics rather than Binomial were used because of their much sharper attenuation slope. Each bandpass filter output is precision full wave rectified and smoothed. The outputs of all twenty filters are connected to a multiplex video switch arrangement. This is used to sequentially interrogate the outputs of each of the twenty filters. Each interrogation sweep produces a video waveform which represents the instantaneous Doppler spectrum. The sampling sweep has a repetition rate of 4.5 kHz.

The video waveform and a synchronised ramp waveform are fed, respectively, to a z-axis amplifier and the x-deflection amplifier of a Medelec fibre optic chart recorder. This recorder produces a spectrogram representing time in the x-axis and frequency content in the y-axis. The spectral power content is represented by the black ness of the displayed information. Approximately 6 grey shades can



Block diagram of filter bank analyser

Figure 4.8

be represented on the Medelec fibre optic chart recorder paper. One second marker pips are displayed on the spectrogram.

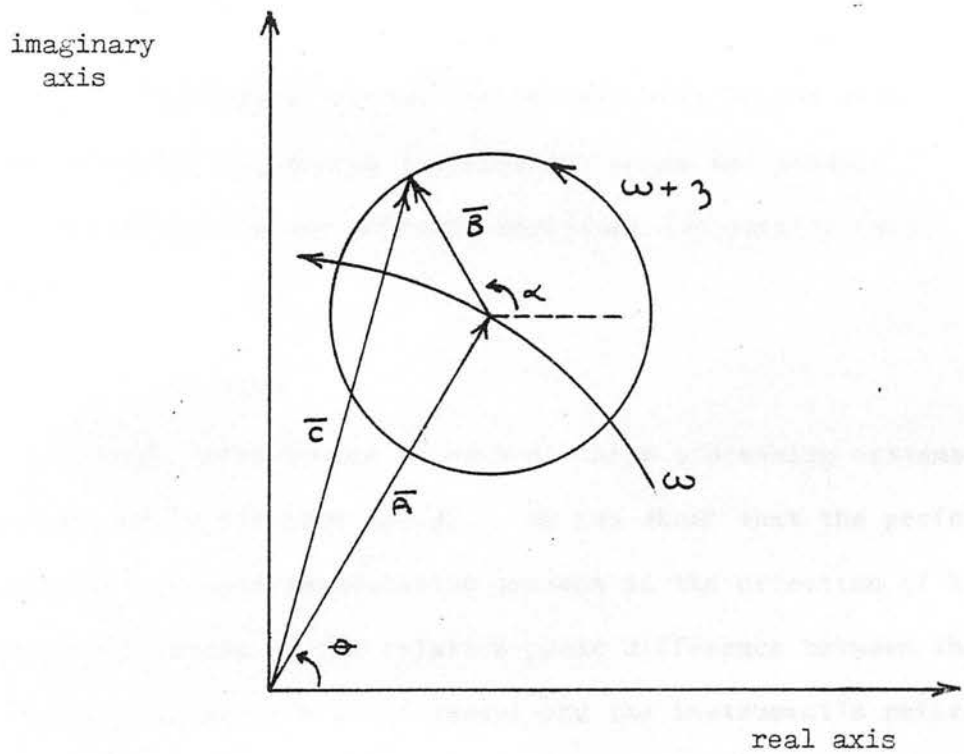
A separate input to the filter bank analyser allows a simultaneous display of a conventional x-y deflection signal to one side of the spectrogram. This input can be used to display a timing signal, such as an E.C.G., simultaneously with a blood flow spectrogram.

CHAPTER 5

SIGNAL EXTRACTION TECHNIQUES WITH PULSED WAVE DOPPLER INSTRUMENTS

The ultrasonic energy returned from the sample volume of a pulsed wave Doppler instrument has a composite frequency spectrum. A large part of the energy return is due to reflections from static structures and this will have a frequency, or frequency spread in the case of the pulsed spectrum, similar to the insonating ultrasound. A second contribution is due to energy reflected from moving structures and blood flow. This will be Doppler shifted either above or below the incident ultrasound spectra depending on the relative orientation of the ultrasound beam and the direction of blood flow. The Doppler shifted signal is typically some 40 - 50 dbs down on that due to non-Doppler shifted signals. Additional bandpass noise is generated in the amplification of the received signal and this may reach a significant level in some cases.

If the equipment generated noise is ignored, the backscattered signal can be represented in the form of a vector diagram, Figure 5.1. The vector $\underline{A}$ of amplitude A and phase angle θ, rotating at an angular velocity ω, represents the static echo from the wall of a blood vessel, for example, while the vector $\underline{B}$ of amplitude B and phase angle α, rotating at an angular velocity $\omega + z$, represents the contribution from the blood. The vector $\underline{C}$ represents the backscattered signal. In the typical instrument the Doppler shift component from blood will cause an amplitude modulation of about



Vector diagram describing the backscattered signal

(after Atkinson 1975)

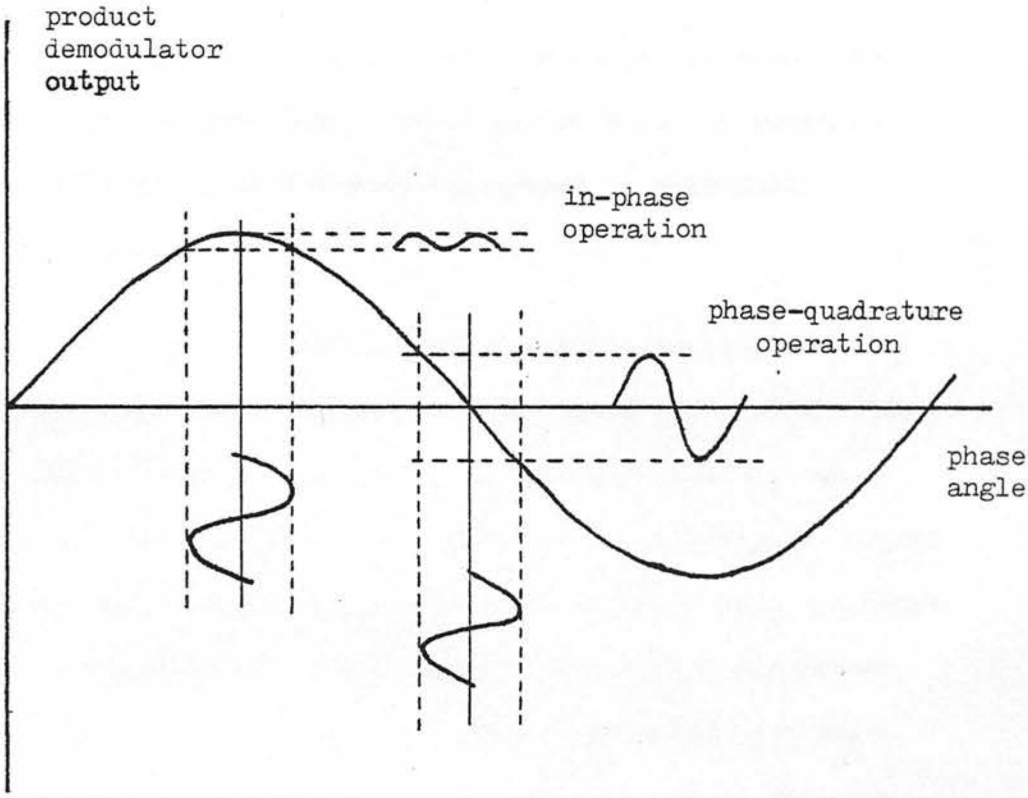
Figure 5.1

1% and a corresponding phase modulation of about 0.5^{o} at the carrier frequency.

Three basic processing systems can be used with pulsed wave Doppler instruments, namely the coherent phase and product demodulators and the non-coherent amplitude demodulator (see Chapter 2).

5.1 SIGNAL PROCESSING

The theoretical performance of each of these processing systems has been analysed by Atkinson (1975). He has shown that the performance of both the coherent demodulation systems in the detection of blood flow are influenced by the relative phase difference between the static wall echoes of a blood vessel and the instrument's reference signal. Figure 5.2 illustrates this effect. It shows a plot of the output voltage of a product detector plotted against the phase angle θ between two constant amplitude sinusoids. Also shown superimposed on this cosine curve is the change in output for small input phase variations at two points on the characteristic, namely at phase quadrature and at in phase conditions. When the back-scattered signal returns in phase quadrature with the reference, the output is a relatively large signal at the expected Doppler frequency. When the return signal is in phase the output is a low amplitude spurious signal. For the phase detector (i.e. product detection of the limited signal), the in phase signal is at a frequency of twice the expected value. Conditions can alternate between the in phase and the phase quadrature condition for each $^{\lambda}/_{8}$ of wall motion in the direction of the transducer. This results



product demodulator input-output characteristics

(after Atkinson 1975)

Figure 5.2

in a large amplitude low frequency output which is the Doppler signal originating from vessel wall motion. This low frequency signal must be removed by filtering, in order that the blood flow signal can be clearly detected. If it was possible to position the sample volume completely inside the vessel no such wall effects would occur.

The non-coherent envelope detector uses a wide transmitter excitation pulse and sample volume, both of which are greater than the vessel wall diameter. The echo signals generated by the vessel walls are used as the in situ reference. The Doppler signal is obtained by amplitude demodulation of the combined wall and blood flow signal. The Doppler shift signals due to tissue and vessel wall motions will in theory occur at lower frequencies with the envelope detector than would be generated with a product detector. This should enable more effective filtering of the low frequency components. With the envelope detector all motion is relative to the vessel walls and not the transducer face. Variations in the mixing ratio of the blood flow and vessel wall signals will result in erratic fluctuations in the detected Doppler signal.

5.2 DEMODULATOR CIRCUITS

It was decided to compare each of the three demodulator systems to determine which would be the most sensitive for this particular application. The three demodulator circuits are given below.

(i) Coherent Phase detector

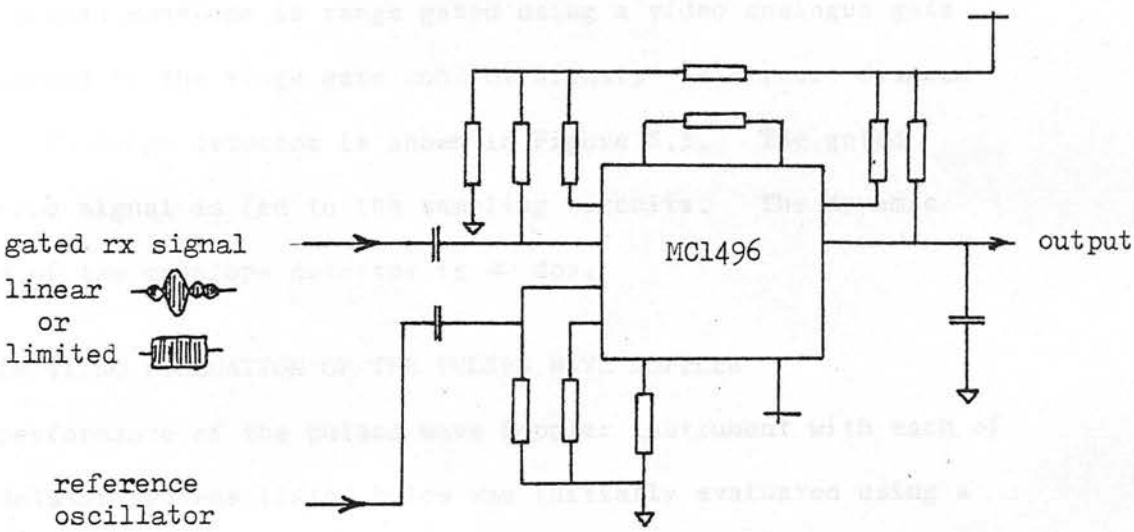
This type of detector operates by sensing the instantaneous phase difference between the gated r.f. signal and the 2.5 MHz reference sinusoid. Phase coherence between pulse repetition frequencies is maintained by using a transmitter excitation burst derived from the reference oscillator. Only the phase information in the received signal is used, hence the detector is insensitive to amplitude fluctuations in the received signal. Also, as the amplitude does not convey any information a large receiver gain can be used, saturation of large amplitude echoes being acceptable if the recovery time of the amplifier is adequate. It has been suggested that signal limiting in phase detectors can improve the signal to noise ratio of the detected Doppler signal (Baker, 1970).

The phase detector circuit has been used with both a linear and a limited receiver signal. Signal limiting is achieved by using an additional r.f. limiting amplifier, giving an additional 12 dbs of gain, in series with the receiver output.

A circuit diagram of the phase detector circuits is shown in Figure 5.3. It consists of a balanced modulator/demodulator integrated circuit (MC 1496) connected as a phase detector. This circuit provides a D.C. voltage output which is a function of the phase difference between the two input signals.

(ii) Coherent product detector

The product detector circuit performs multiplication of the gated received signal and the coherent reference sinusoid. Both the amplitude and the phase information of the received signal is used. A circuit diagram of the product detector circuit is shown in Figure 5.4. The signal at the drain of the f.e.t. contains frequency components at the carrier frequency plus the sum of forward and



Phase detector circuit

Figure 5.3

reverse Doppler flow signals. A low pass filter is used to remove the carrier frequency component.

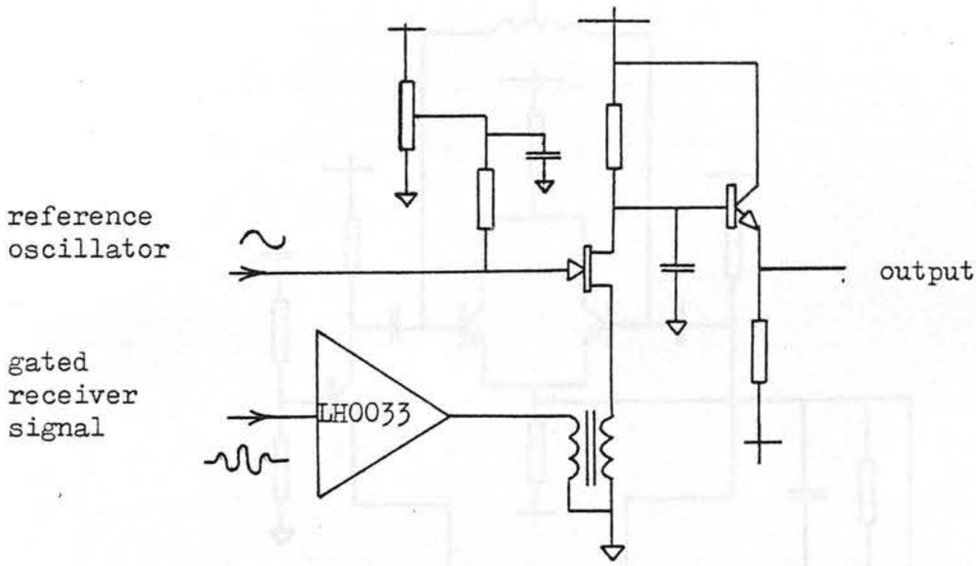
(iii) Non-coherent envelope detector

The ungated receiver output is full wave rectified and smoothed. This A-scan envelope is range gated using a video analogue gate controlled by the range gate control signal. A circuit diagram of the envelope detector is shown in Figure 5.5. The gated envelope signal is fed to the sampling circuits. The dynamic range of the envelope detector is 40 dbs.

5.3 IN VITRO EVALUATION OF THE PULSED WAVE DOPPLER

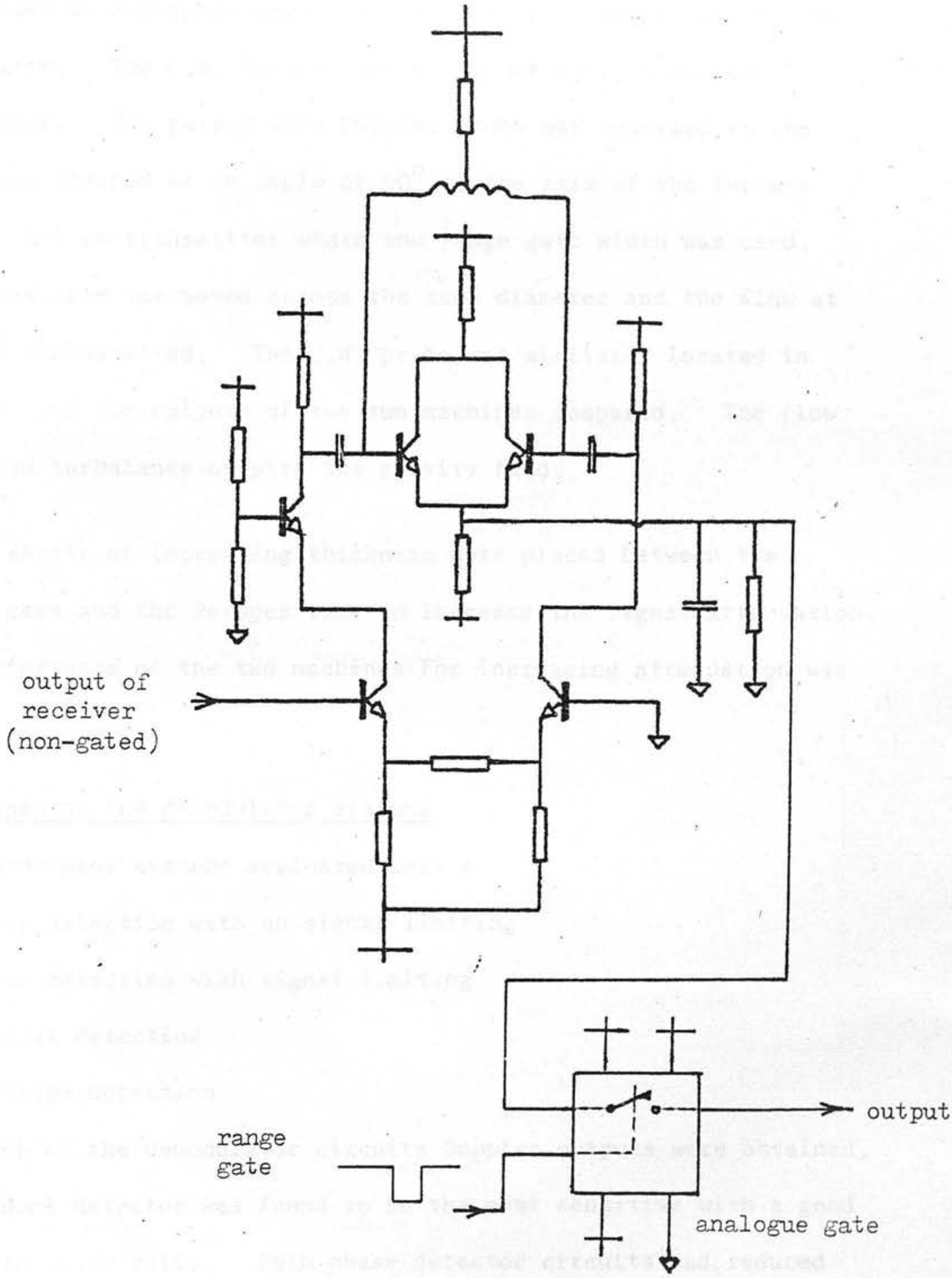
The performance of the pulsed wave Doppler instrument with each of demodulator systems listed below was initially evaluated using a simple test phantom. In each case, the performance of the pulsed wave Doppler was compared to that of a continuous wave device, a Sonicaid D205 instrument. Although there are obvious differences between the two types of Doppler, for example the sample volume, such a comparison is useful.

The phantom consisted of a fluid flowing in a thin walled Perspex tube (Figure 5.6). The tube was immersed in a water tank. Dilute milk was used to simulate blood. The suspended fat globules of milk have diameters smaller than the ultrasonic wavelength. The Perspex tube had an outer diameter of 11 mm and inner diameter of 10 mm and was 8 cm long. The milk was poured into an upper reservoir and allowed to drain through the Perspex tube into a calibrated lower reservoir. The flow rate was controlled by a valve. The volume per unit time collected in the lower reservoir indicated the flow rate.



Product detector circuit

Figure 5.4



Non-coherent envelope detector

Figure 5.5

The pulsed wave Doppler used a 2.5 MHz, 13 mm diameter pulse echo transducer. The C.W. device uses a 2.0 MHz split undamped transducer. The pulsed wave Doppler probe was immersed in the water and clamped at an angle of 60^{o} to the axis of the Perspex tube. A 5 µs transmitter width and range gate width was used. The range gate was moved across the tube diameter and the flow at various depths noted. The C.W. probe was similarly located in the tank and the outputs of the two machines compared. The flow exhibited turbulence despite the gravity feed.

Rubber sheets of increasing thickness were placed between the transducers and the Perspex tube to increase the signal attenuation. The performance of the two machines for increasing attenuation was noted.

Comparison of the demodulator systems

The demodulator systems evaluated were -

(a) phase detection with no signal limiting

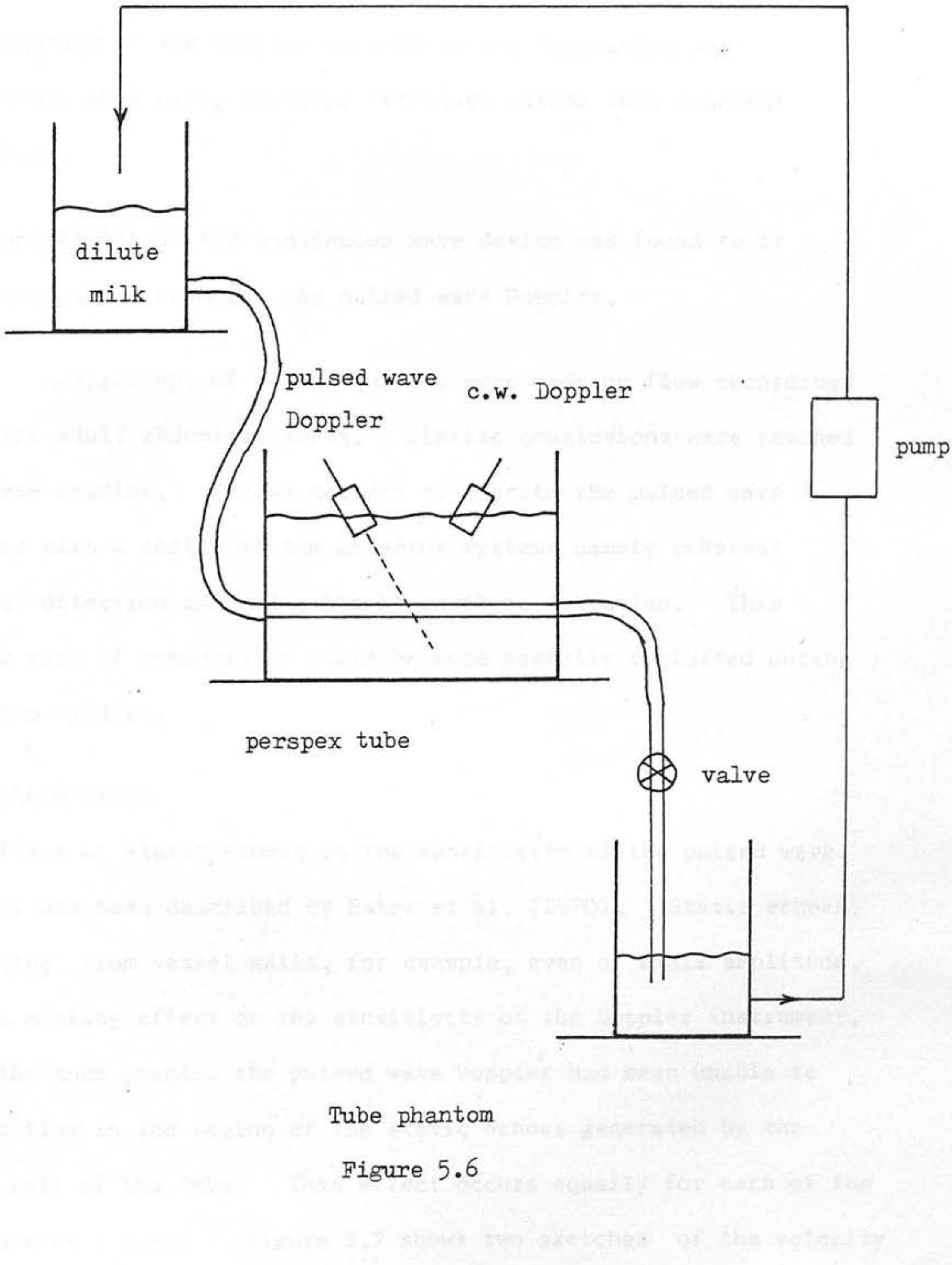
(b) phase detection with signal limiting

(c) product detection

(d) envelope detection

With each of the demodulator circuits Doppler outputs were obtained. The product detector was found to be the most sensitive with a good signal to noise ratio. Both phase detector circuits had reduced signal to noise ratios. No detectable difference was noted between the performance of the phase detector with and without signal limiting.

The envelope detector produced a slightly more noisy signal than the coherent detectors, even when using a wide transmitter and range



Tube phantom

Figure 5.6

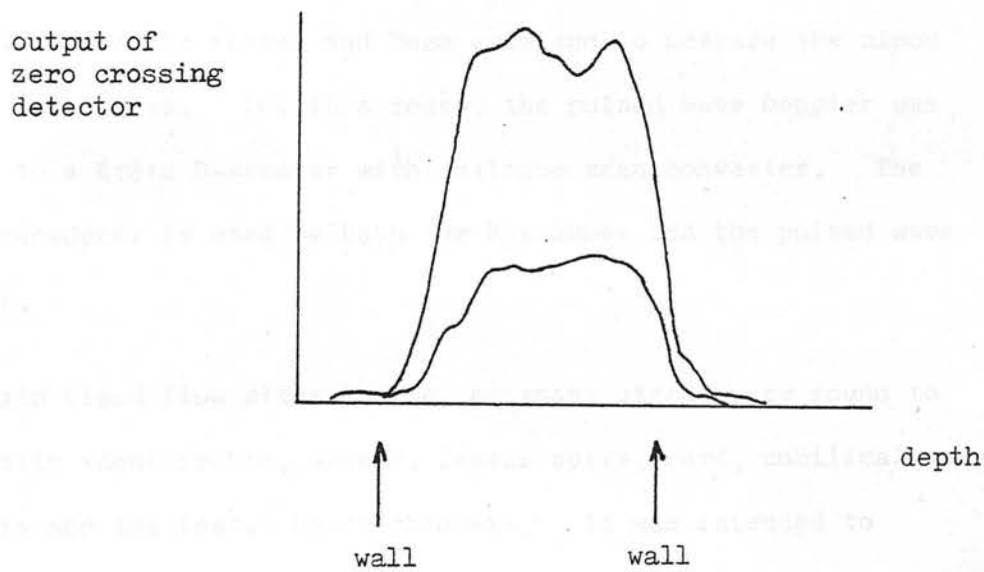
gate width (up to 12 μs). There was little noticeable reduction in frequency of the Doppler components due to phantom wall vibrations when using envelope detection rather than coherent detection.

The sensitivity of the continuous wave device was found to be slightly higher than for the pulsed wave Doppler.

Similar comparisons of signal quality were made on flow recordings from the adult abdominal aorta. Similar conclusions were reached in these studies. It was decided to operate the pulsed wave Doppler with a choice of two detector systems namely coherent product detection and non-coherent envelope detection. This latter type of demodulator would be more usefully evaluated during in vivo studies.

5.4 WALL EFFECTS

The effect of static echoes on the sensitivity of the pulsed wave Doppler has been described by Baker et al. (1970). Static echoes resulting from vessel walls, for example, even of small amplitude, have a masking effect on the sensitivity of the Doppler instrument. With the tube phantom the pulsed wave Doppler had been unable to detect flow in the region of the static echoes generated by the near wall of the tube. This effect occurs equally for each of the demodulator systems. Figure 5.7 shows two sketches of the velocity profile obtained by moving the range gate of pulsed wave Doppler across the tube phantom. The output of the zero crossing detector is plotted against depth. Velocity plots for two different flow rates are shown.



velocity profiles obtained by moving range gate across the perspex tube.

Figure 5.7

CHAPTER 6

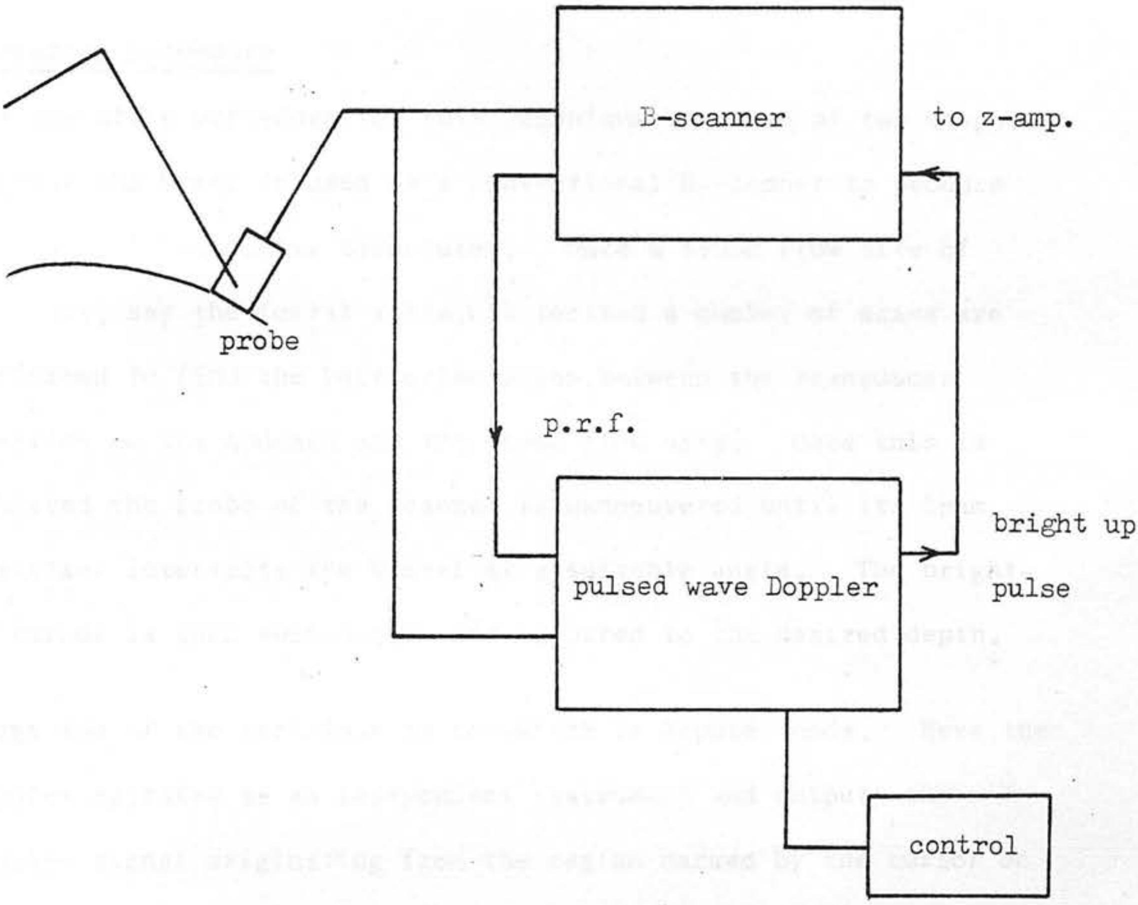
PRELIMINARY TRIALS WITH THE PULSED WAVE DOPPLER INSTRUMENT

Obstetrical blood flow studies cannot be undertaken using a Doppler instrument searching blind. It is essential to be able to identify clearly the blood flow site of interest, to determine the relative orientation of the vessel and beam axis and to measure the blood vessel lumen area. For this reason the pulsed wave Doppler was linked to a Kretz B-scanner with analogue scan converter. The same transducer is used by both the B-scanner and the pulsed wave Doppler.

Five main blood flow sites in the pregnant uterus were found to be readily identifiable, namely, foetal aorta, cord, umbilical vein, placenta and the foetal heart chambers. It was intended to concentrate on these main blood flow sites and investigate the feasibility of recording flow at each. Normal pregnancies of 28 weeks menstrual age or over were chosen.

6.1 B-SCAN AND PULSED WAVE DOPPLER COMBINATION

Figure 6.1 shows a block diagram of the B-scan and pulsed wave Doppler combination. Mode selection, range gate display control and range gate position are all controlled from a remote hand-held control box. Two modes are selectable, namely B-scan mode and pulsed wave Doppler mode. During the B-scan mode, pulses from the Kretz p.r.f. generator are fed into the external p.r.f. input of the pulsed wave Doppler. These pulses enable the Doppler range gate circuitry to output a sample volume bright-up pulse. When



B-scan / pulsed wave Doppler combination

Figure 6.1

selected, this pulse produces on the scan converter screen a bright-up cursor which exactly identifies the position of the pulsed wave Doppler range gate.

Operating procedure

The operating procedure for this technique consists of two steps. Firstly the Kretz is used as a conventional B-scanner to produce an image of the foetal structures. Once a blood flow site of interest, say the foetal aorta, is located a number of scans are performed to find the best orientation between the transducer position on the abdomen and the blood flow site. Once this is achieved the probe of the scanner is manoeuvered until its beam direction intersects the vessel at a suitable angle. The bright-up cursor is then switched on and adjusted to the desired depth.

Stage two of the technique is to switch to Doppler mode. Here the Doppler operates as an independent instrument and outputs the Doppler signal originating from the region marked by the cursor on the scan converter screen. This assumes no movement of the blood flow site between the time of imaging and the commencement of Doppler recording. The Doppler information is recorded on an f.m. audio tape recorder. Vessel diameter and transducer to blood vessel inclination is recorded by taking a photograph of the scan converter screen. The B-scan machine's electronic calipers can be used to measure the vessel diameter.

6.2 PRELIMINARY TRIALS

Adult abdominal aortic flow

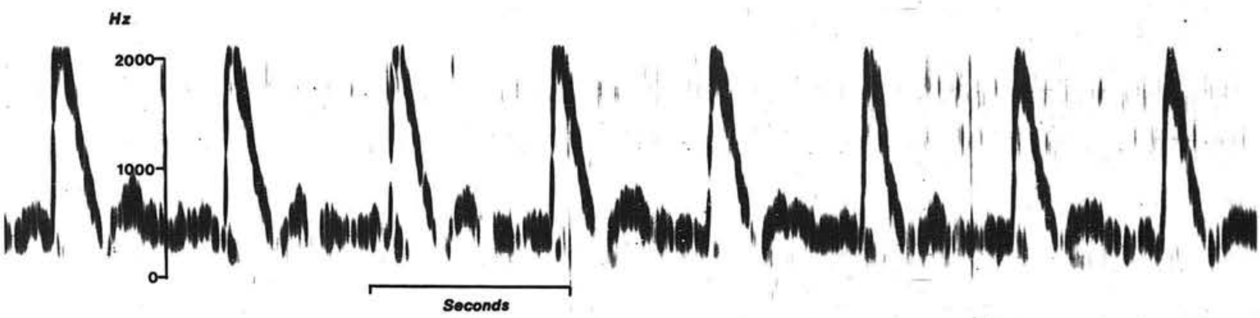
This B-scan and Doppler combination was initially tested on a group of adult male volunteers, scanning the abdominal aorta and vena cava. The p.r.f. was 3.9 kHz.

Characteristic aortic and venous blood flow signals were obtained. When a small transmitter and range gate width (3 μs) was used and the sample volume set to lie within the vessel the blood flow signals obtained were very noisy. Low frequency vessel wall motions were also recorded. Breathing motions and vessel pulsations make it difficult to maintain the sample volume completely inside the blood vessel.

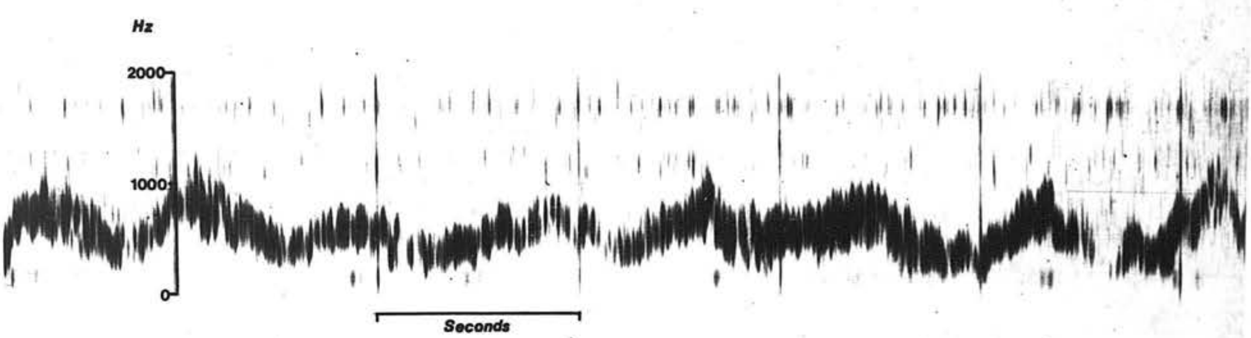
Improved signal to noise ratios were obtained when a wide transmitter excitation pulse, typically 10 μs, was used and the sample volume was set to straddle the vessel. In this case the range resolution is about 1 cm. The vessel wall motions contribute a very large amplitude low frequency Doppler component (up to about 300 Hz). The product detector consistently produced better quality Doppler signals than the envelope detector. Figure 6.2 shows typical sonograms obtained from the adult aorta and inferior vena cava using the product detector.

<u>Obstetrical trial</u>

A short series of five volunteer patients was examined. Doppler shift signals from within the foetal heart and from cord were obtained. The foetal heart signals were predominantly of a high frequency flicking nature attributed to valve motion and not blood flow. The cord signal was discontinuous as would be expected due to cord motion. Flow in the foetal aorta was not successfully



(a)



(b)

Pulsed wave Doppler spectrograms from (a) adult aorta and (b) adult inferior vena cava.

Figure 6.2

recorded. Umbilical vein was not located. Within the placenta no meaningful signals were recorded.

The placenta

To record blood flow from within the placenta is one of the most challenging aspects of this study. Characteristic placental sounds have been recorded with a C.W. Doppler (Hunt, 1969), but this method does not confirm the site within the placenta at which such sounds are heard. It is conjecture that these C.W. recordings are from the foetal placental circulation at sites close to the placenta-cord junction. Within the placenta foetal-placental and maternal-placental flow rates are expected to be very much lower due to the large volume of the placental bed.

Within the placenta there is no recognisable blood vessel structure. The combination of very low velocity blood flow in a homogeneous structure which will produce large static echoes, makes blood flow detection within the placenta a difficult technical problem.

A placental test phantom

In view of the difficulties encountered in the use of both the continuous wave and the pulsed wave Doppler when probing a structure such as the placenta, a special test system was constructed. It consisted of a Perspex drum 15 cm in diameter and 5 cm high, with polythene sheeting tightly stretched over the top and bottom surfaces. Fluid flow through the drum was by two 15 mm diameter pipes attached to the drum sides diametrically opposite each other. The inside of the phantom was loosely packed with long thin plastic fibres. Milk was fed through the phantom at a slow rate using the gravity feed technique as before. The phantom was immersed in a water tank.

Both the continuous wave and the pulsed wave Doppler instruments were equally effective at detecting the fluid flow at the input and output of the phantom. As the sample volume of the probes was moved towards the centre of the phantom it became progressively more difficult to record flow. At the centre of the phantom neither Doppler recorded flow. The C.W. device was slightly more sensitive than the pulsed wave device. The C.W. Doppler incorporates a low frequency cut-off filter which may have reduced its low frequency response. The transmitted pulse duration and sample gate duration of the pulsed wave Doppler were individually adjusted in the range 1 - 12 μs. This produced little change in the performance of the device although there was a slightly better signal-to-noise ratio with the wide transmitter pulse. With the placental phantom, the masking effects due to static wall echoes may have a deleterious effect on the performance of the pulsed wave device. Once again product detection proved to be more sensitive than envelope detection.

6.3 B-SCAN OR REAL TIME SCANNER

For this initial trial with the pulsed wave Doppler instrument B-scan imaging has proved most useful. Static B-scanners do have problems however. Their restricted scanning plane makes it difficult to locate and follow a vessel. Also, moving structures, such as heart chambers, appear as blurred images lacking detail.

Real time scanning is more convenient. It offers considerable flexibility in the scanning plane, removes patient movement effects, and enables rapid visualisation and location of foetal anatomical and blood flow sites. Blood vessels can be seen to pulsate and foetal heart chambers can be individually identified.

A real time scanner, although not available initially is now used with the equipment.

CHAPTER 7

A COMBINED REAL-TIME SCANNER AND PULSED WAVE DOPPLER INSTRUMENT FOR OBSTETRICAL BLOOD FLOW STUDIES

There are two types of real time scanner commonly used in obstetrics, namely the linear array and the rotating transducer mechanical scanner. This latter type of scanner was being developed within the department for use in heart and foetal breathing studies (Bow et al., 1979, McDicken et al., 1979). It was decided to combine the pulsed wave Doppler with a real time rotating transducer scanner. The real time scanner is used to locate and measure the blood vessel and to facilitate the placing of the pulsed wave Doppler range gate exactly in the region of interest.

7.1 BASIC BLOOD FLOW SCANNING INSTRUMENT

Figure 7.1 shows the basic blood flow scanning head. It consists of two parts,

(a) the real time horizontal tubular scanner, and

(b) a mechanical framework to which the pulse wave Doppler probe is attached.

Such a combination is both light and compact and, when placed in a special holder, can be belted onto the abdomen.

As with the static B-scan configuration, two modes of operation exist, namely real time visualisation and Doppler flow recording. The real time scanner is used to explore the foetal anatomy and locate the desired blood flow site. The line of the Doppler beam

Real time scanner with Doppler probe holder

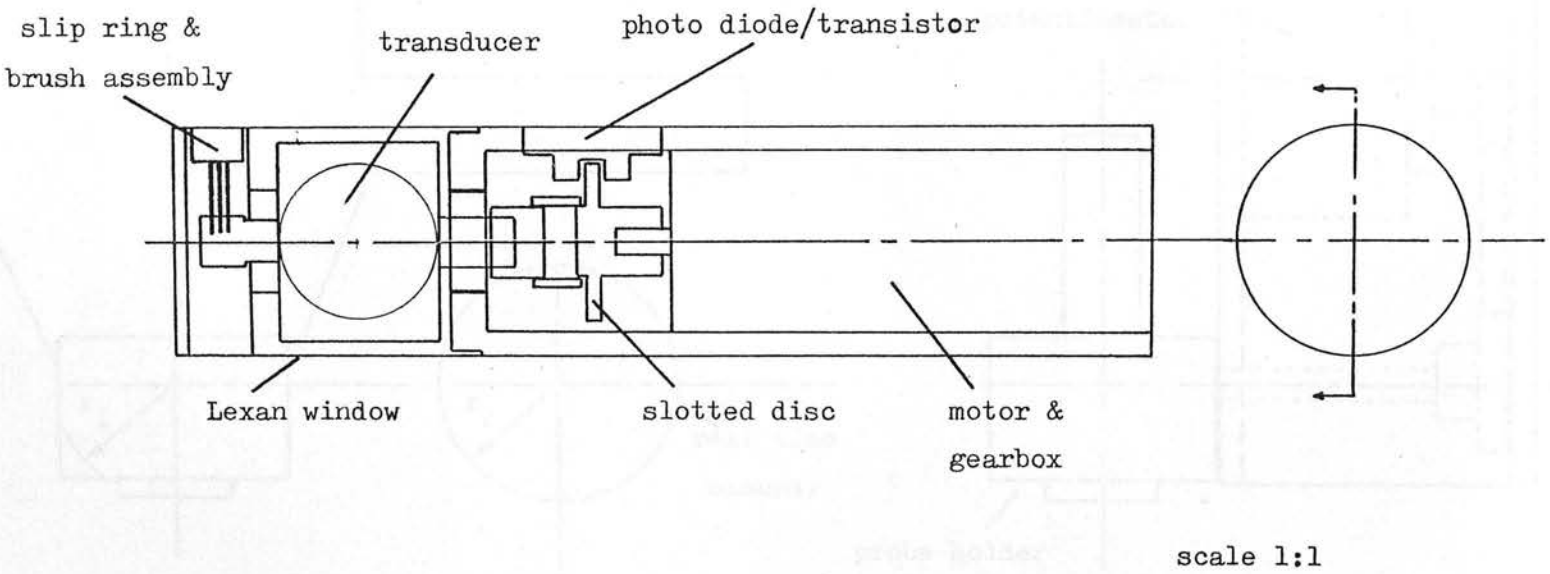
Figure 7.1

is shown as a white line superimposed on the real time image with a bright-up portion to show the exact location of the Doppler range gate. The Doppler beam angle is altered by mechanically rotating the probe in its holder. A photograph is taken of the viewing screen to record vessel diameter and vessel to Doppler beam inclination. The real time scanner is then quickly stopped and pulsed wave Doppler measurements of flow velocity commenced.

7.2 EQUIPMENT DESCRIPTION

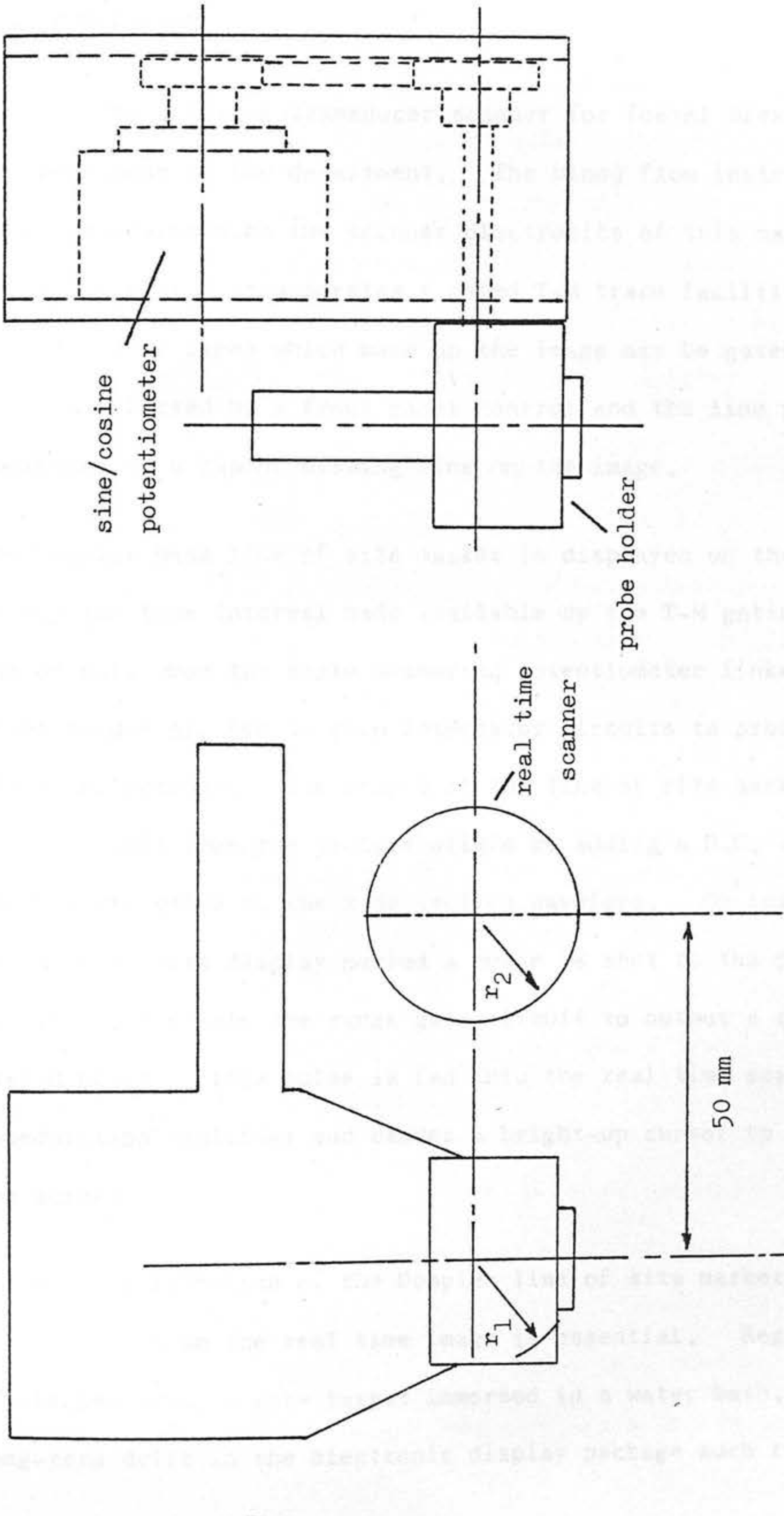
Figure 7.2 shows a diagram of the tubular scanner. The scanning head consists of two standard 15 mm diameter, 2.5 MHz damped transducers mounted 180° to each other in a Perspex wheel. The Perspex wheel rotates inside an oil filled cylindrical cavity. The scanning head is 3 cm in diameter and 2.5 cm in length. A D.C. motor is used to rotate the scanning head. Electrical connection to the transducer is made by a miniature gold brush and slip ring assembly. No angle measuring device is fitted to this basic scanner. A 150 line, 180° image at a frame rate of 22 frames per second is generated.

A drawing of the Doppler probe holder attachment is shown in Figure 7.3. The holder can be rotated through an angle of about 60° and is linked by means of a belt arrangement to a sine/cosine potentiometer. This potentiometer is used to indicate the line of site of the Doppler probe on the real time screen. The central axis of rotation of the real time scanner and attachment is offset by 50 mm in the x-axis. The scanning plane and plane of rotation of the probe are identical.



Tubular real time scanner

Figure 7.2



Doppler probe holder attachment

Figure 7.3

The radii r1 and r2 on figure 7.3 are identical.

7.3 INTERFACING

A real time rotating transducer scanner for foetal breathing studies has been made in the department. The blood flow instrument has been interfaced with the scanner electronics of this machine. The breathing system incorporates a gated T-M trace facility. Any one of the radial lines which make up the image may be gated out. The line is selected by a front panel control and the line position is indicated by a gap or missing line on the image.

The Doppler beam line of site marker is displayed on the image screen during the time interval made available by the T-M gating facility. The outputs from the angle measuring potentiometer linked to the probe holder are fed to ramp integrator circuits to produce the x and y deflections. The origin of the line of site marker is offset in the x-axis from the picture origin by adding a D.C. voltage of appropriate value to the x deflection waveform. On initiation of the line of site display period a pulse is sent to the pulsed wave Doppler to initiate the range gate circuit to output a range gate period pulse. This pulse is fed into the real time scanner z-modulation amplifier and causes a bright-up cursor to appear on the screen.

Accurate registration of the Doppler line of site marker and range gate position on the real time image is essential. Registration is checked using a wire target immersed in a water bath. Due to long-term drift in the electronic display package such registration checks are routinely performed prior to any use of the system.

It was found impossible to operate both the real time scanner and the pulsed wave Doppler instrument simultaneously. Both machines ultrasonically interfere with each other. Pulse repetition frequency interlacing has not been attempted. The real time scanner must be stopped before any Doppler signals are recorded. It takes approximately 3 seconds for the real time scanner to slow down and stop and Doppler recording to commence.

The probe holder may also accept a continuous wave Doppler probe (Sonicaid D205). In situations where the C.W. Doppler beam interrogates a blood flow site and all other structures within the beam are at rest a good C.W. Doppler recording is obtained.

7.4 PULSED WAVE DOPPLER OPERATION USING A TRANSDUCER IN THE REAL TIME SCANNING HEAD

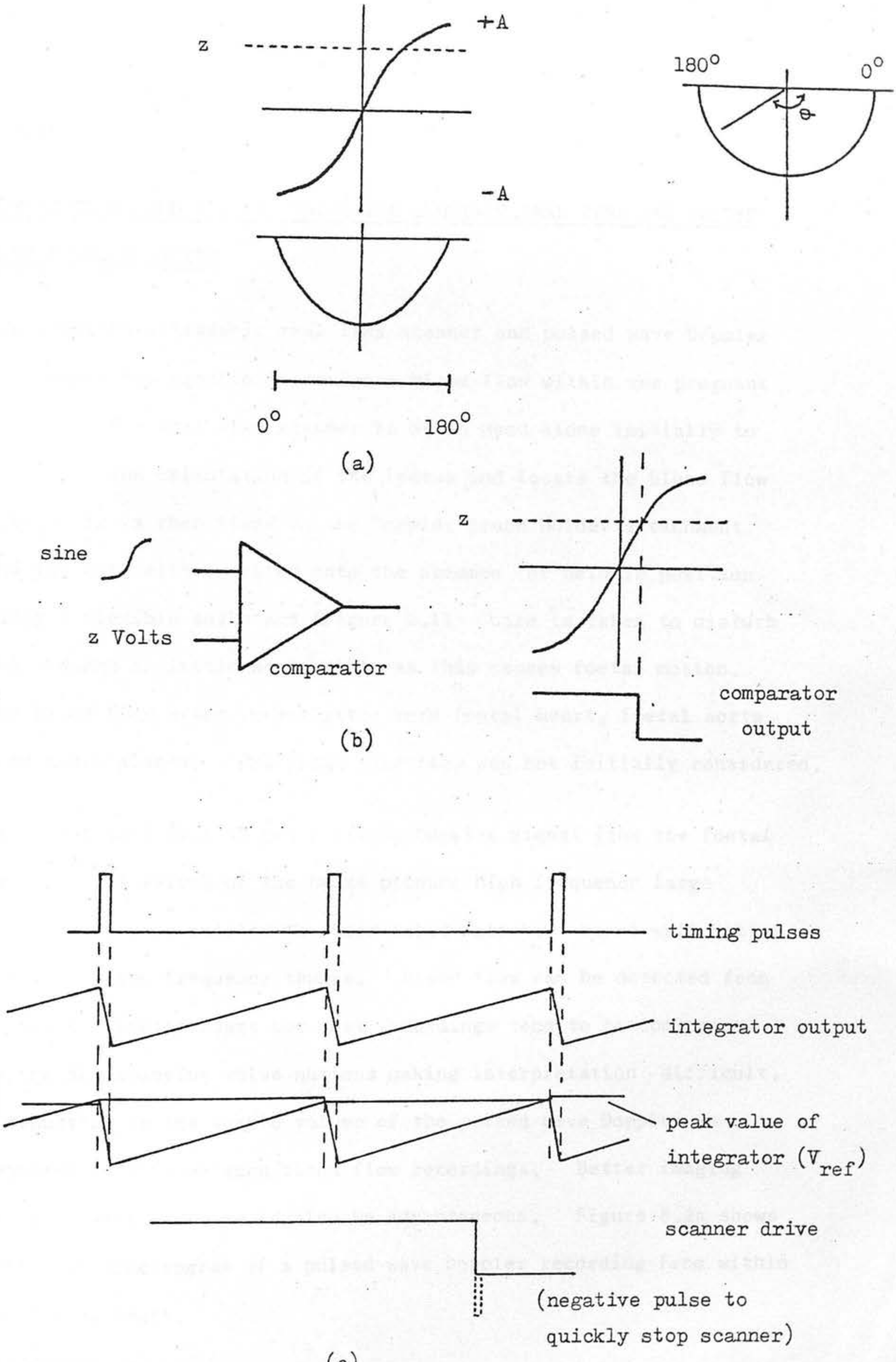
The damped transducers in the real time scanning head are identical in performance to the hand-held pulse echo transducer used by the pulsed wave Doppler. If the real time scanner can be stopped at a selected angle on the real time screen then a transducer in the scanning head can be used as the Doppler transducer.

Figure 7.4a shows the x and y deflection waveforms for a 180° sector. A voltage z on the x-deflection waveform will uniquely define a radial line at an angle θ on the real time screen. If the x-deflection waveform and a D.C. voltage z are compared in a voltage comparator circuit, a timing pulse can be generated which can be used to stop the scanner at the selected angle (see Figure 7.4b).

The basic real time scanner had no angle measuring device. The only angle information came from timing slots on an aluminium disc (see Figure 7.2). These slots occur at the beginning of each 180° sector. By using an integrator circuit an x-deflection type waveform can be generated from these timing slots. The scanner is made to rotate at a slow but constant speed. By comparing the integrator output and a tapped value of Vref, (the peak detected value of the integrator output) the scanner can be stopped at a selected angle (Figure 7.4c). A front panel potentiometer selects the stopping angle and this is displayed on the real time screen.

On changing from real time mode to stopping mode a few seconds must be allowed for the scanner to gently decelerate to a slow running speed before stopping can be implemented. It takes 4 to 5 seconds to stop the scanner. If used over a sector angle of between 45° and 125° of image sweep the stopping error is $\pm 5^{\circ}$.

This technique is not ideal but in the first prototype it proved to be adequate. In a subsequent model an angle measuring device is included in the real time scanner. The probe can now be stopped to an accuracy of $\pm \frac{1}{2}^{\circ}$. Many of the Doppler results presented in this thesis were obtained using the original prototype.



+A
z
-A
180°
0°
θ
0°
180°
(a)
sine
z Volts
comparator
z
comparator
output
(b)
timing pulses
integrator output
peak value of
integrator (V ref)
scanner drive
(negative pulse to
quickly stop scanner)
(c)

Figure 7.4

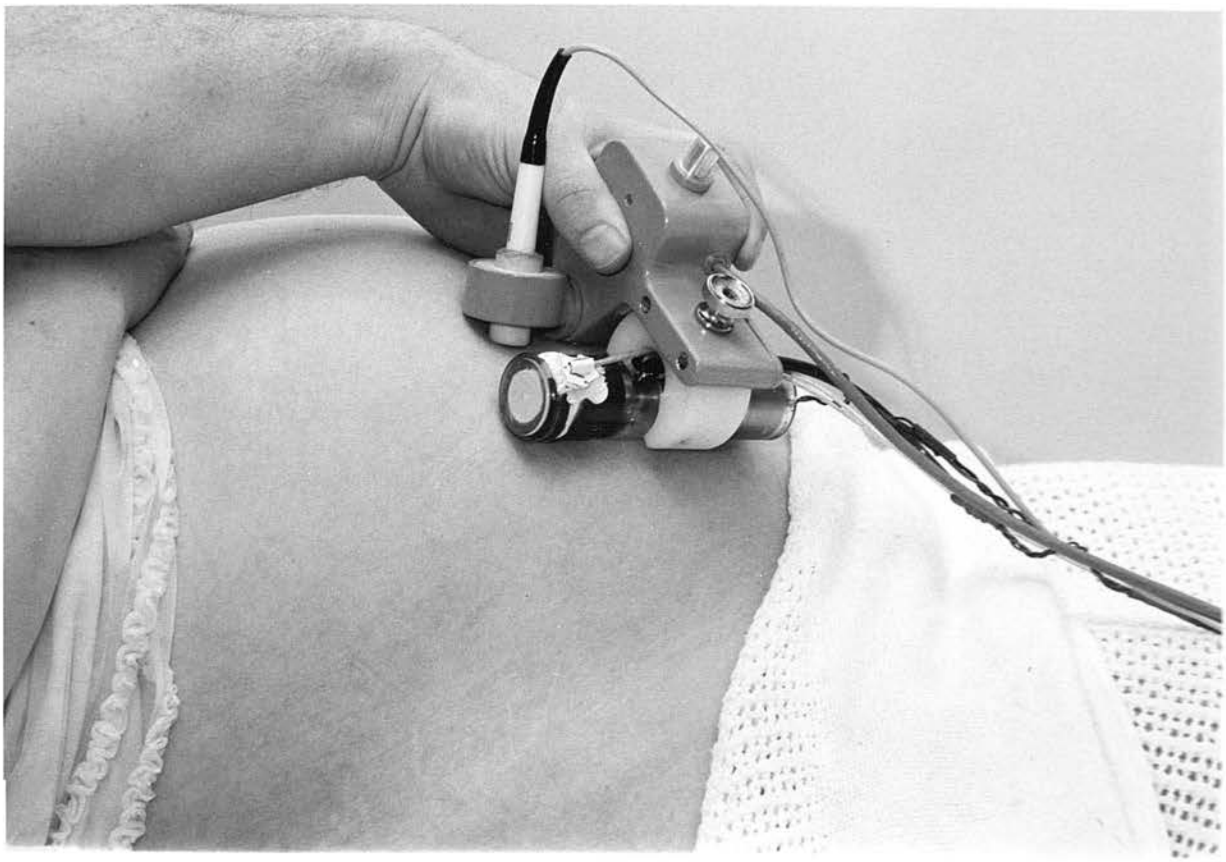
CHAPTER 8

APPLICATIONS AND RESULTS USING THE COMBINED REAL TIME AND PULSED WAVE DOPPLER SYSTEM

The combined ultrasonic real time scanner and pulsed wave Doppler instrument was used to investigate blood flow within the pregnant uterus. The real time scanner is often used alone initially to ascertain the orientation of the foetus and locate the blood flow site. It is then fixed to the Doppler probe holder attachment and the unit either belted onto the abdomen or held in position using a flexible snake arm (Figure 8.1) Care is taken to disturb the abdomen as little as possible as this causes foetal motion. The blood flow sites investigated were foetal heart, foetal aorta, cord and placenta. Umbilical vein flow was not initially considered.

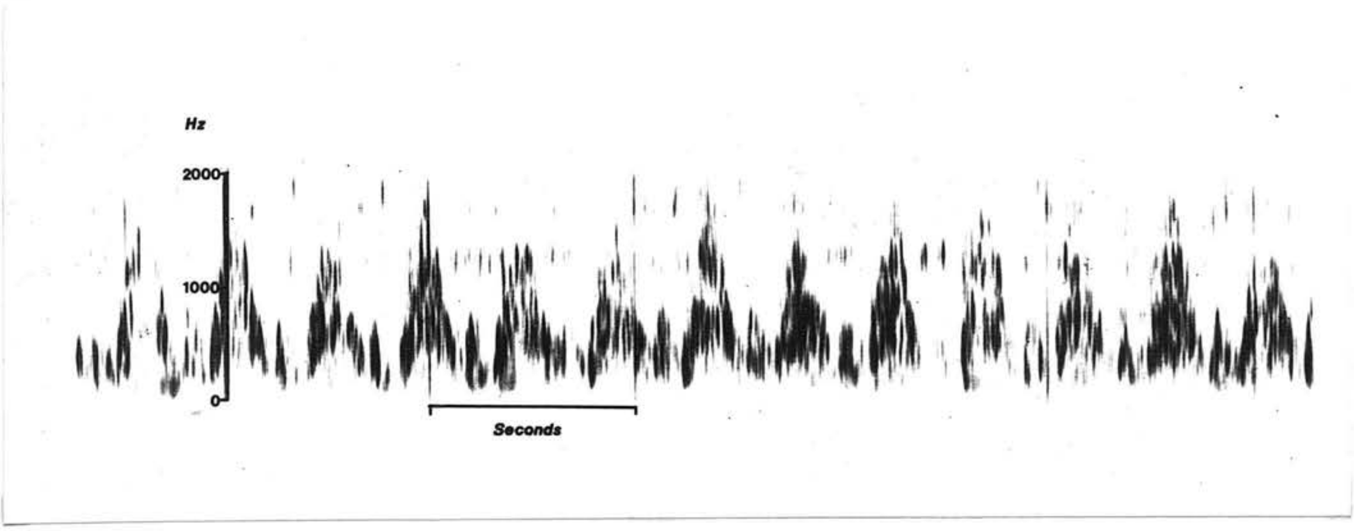
It is not difficult to get a strong Doppler signal from the foetal heart. The valves of the heart produce high frequency large amplitude sounds related to their rapid motion. Heart wall motion results in low frequency thumps. Blood flow can be detected from within the foetal heart but most recordings tend to be dominated by the accompanying valve motions making interpretation difficult. A reduction in the sample volume of the pulsed wave Doppler is required to improve such blood flow recordings. Better imaging of the foetal heart would also be advantageous. Figure 8.2a shows a typical spectrogram of a pulsed wave Doppler recording from within the foetal heart.

A long section of foetal aorta is most easily located by performing

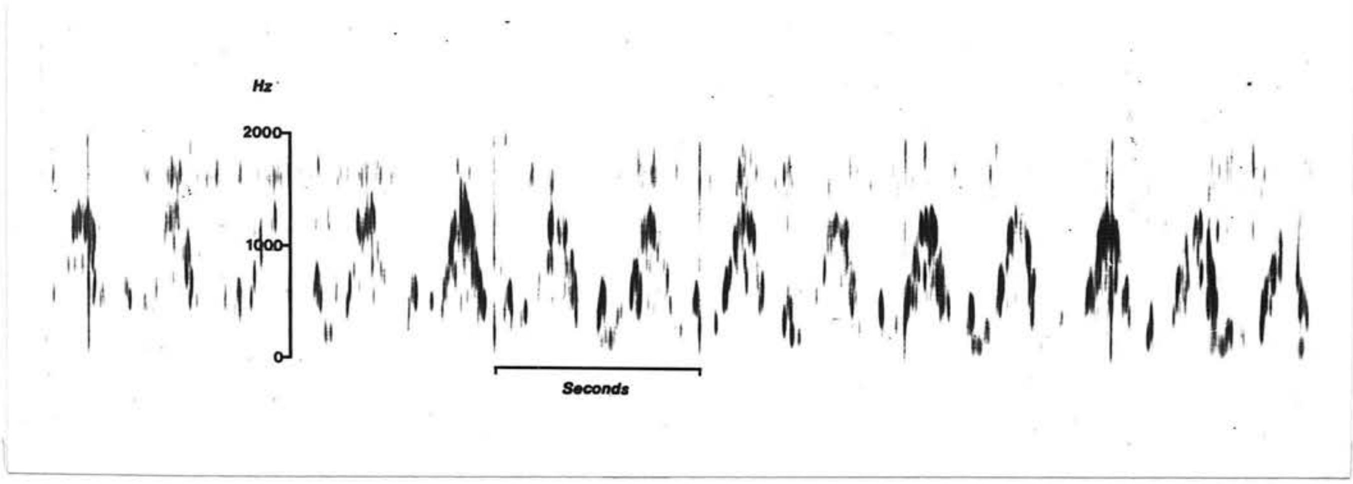


The basic blood flow scanning head

Figure 8.1



(a)



(b)

Pulsed wave Doppler spectrograms from (a) foetal heart and (b) foetal aorta.

Figure 8.2

a scan along the foetal spine on a foetus lying spine uppermost. The foetal aorta is clearly seen to lie just below the spine. The walls of the vessel are seen to pulsate. Aorta can lie at a depth of between 4 and 15 cm from the skin surface. Recordings are taken with a wide transmitter width and range gate straddling of the vessel. Figure 8.2b shows a typical spectrogram from the foetal aorta. A transmitter pulse width and range gate width of 10 μs was used. The p.r.f. was set at 4.0 kHz. The maximum Doppler frequency component set by the low pass filters was 2.0 kHz. In many of the blood flow recordings from the aorta vessel wall effects have obscured the blood flow signal. At large beam depths, typically greater than 12 cm, the signal to noise ratio of the detected signal begins to deteriorate.

Recordings from the foetal cord were attempted by taking transverse sections through large coils of cord. The range gate was set to straddle the coils. Cord motion causes recordings to be discontinuous and results in very large amplitude motion signals. Cord has proved to be too mobile a structure to be considered as a suitable site for investigation with the pulsed wave Doppler.

Within the placenta a number of different sites were investigated with the pulsed wave Doppler. An anterior placenta with a well defined placental line is usually chosen. Despite a concentrated effort this was the least successful of all the flow studies, no meaningful recording being obtained at any site. The continuous wave Doppler instrument does produce a flow record but only when aimed at a region in the placenta very close to the placenta cord junction. The recording of blood flow from within the placenta appears to be a difficult technical problem.

Summary

In all, some 20 normal patients were scanned with this blood flow system. Pregnancies in excess of 32 weeks gestational age were chosen in all cases and one or more blood flow sites investigated in each patient. Typical examination time was about one half hour.

These trials have established that it is possible to obtain blood flow recordings from the foetal aorta and foetal heart. Recordings from the foetal cord can also be made but motion of the cord causes practical difficulties.

In all these recordings, vessel or tissue motion or both, produce very large amplitude low frequency Doppler shifts. Up to the first 400 Hz of Doppler signal many have to be filtered out before the blood flow signal can be audibly discerned. The signal to noise ratio of the Doppler signal deteriorates rapidly at beam depths in excess of 12 cm. Wall masking effects appear to reduce the sensitivity of the Doppler instrument if a short acoustic sample volume is used. Only with a wide transmitter pulse, at least 6 μs wide, is the Doppler signal to noise ratio adequate. A 6 μs pulse is equivalent to an axial resolution of approximately 5 mm. The sample volume is normally arranged to straddle the vessel.

During foetal blood flow studies with this instrument, foetal breathing movements of the chest or abdominal wall have been observed on the real time image. These movements can be recorded using the pulsed wave Doppler instrument. A description of this technique is presented in Appendix 2. The effect of such motions on foetal blood flow velocities has yet to be examined.

CHAPTER 9

THE CONTINUOUS WAVE INTERSECTING ZONE ULTRASONIC DOPPLER INSTRUMENT

The real time scanner/Doppler blood flow system can be used with either pulsed wave Doppler or continuous wave Doppler instruments. In vivo trials with the pulsed wave Doppler have shown the difficulties of obtaining blood flow recordings with good signal to noise ratios in structures such as the placenta. Theoretical maximum velocity-depth restrictions make measurements of large flow velocities at large depths impossible with this instrument. The maximum measurable velocity with a 2.5 MHz Doppler unit for depths of 5, 10, 15 and 20 cm is at most 200, 100, 75 and 50 cm/s respectively.

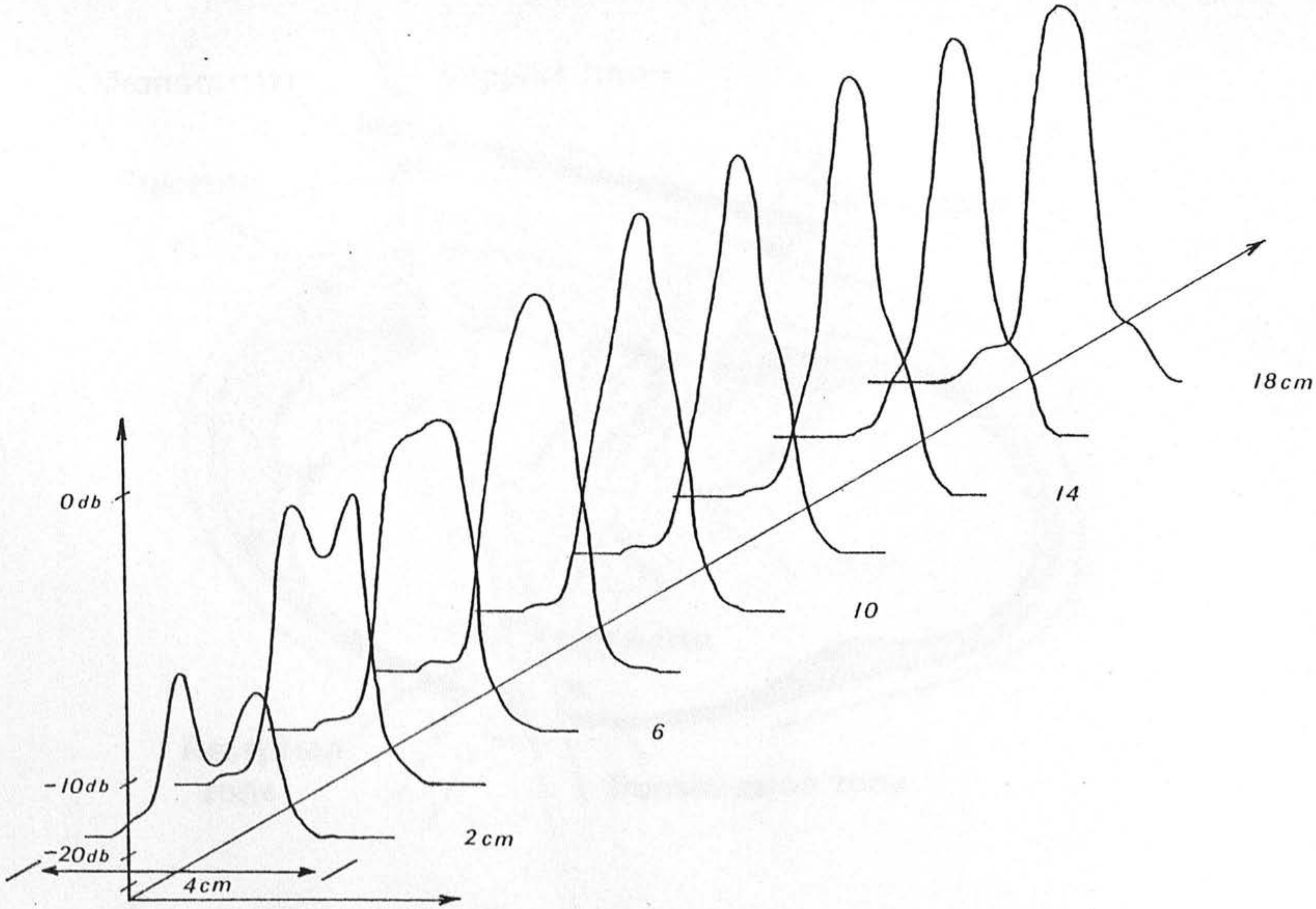
The sensitivity of the standard double crystal Doppler instrument in vivo is superior to the pulsed wave Doppler. Flow recordings may also be made to much greater depths with the continuous wave device. Where the real time scan has shown that only the blood flow site of interest will contribute a Doppler signal in the line of site of the Doppler beam the continuous wave device has been used. In this case good quality flow signals from foetal aorta, cord and umbilical vein have been recorded. There is no practical limit to the maximum velocity of the signal which can be measured with the continuous wave device. It has no depth resolution however.

9.1 THEORY OF OPERATION OF THE C.W. INTERSECTING ZONE DOPPLER

The continuous wave Doppler transducer typically consists of two separate elements spaced apart or inclined to one another.

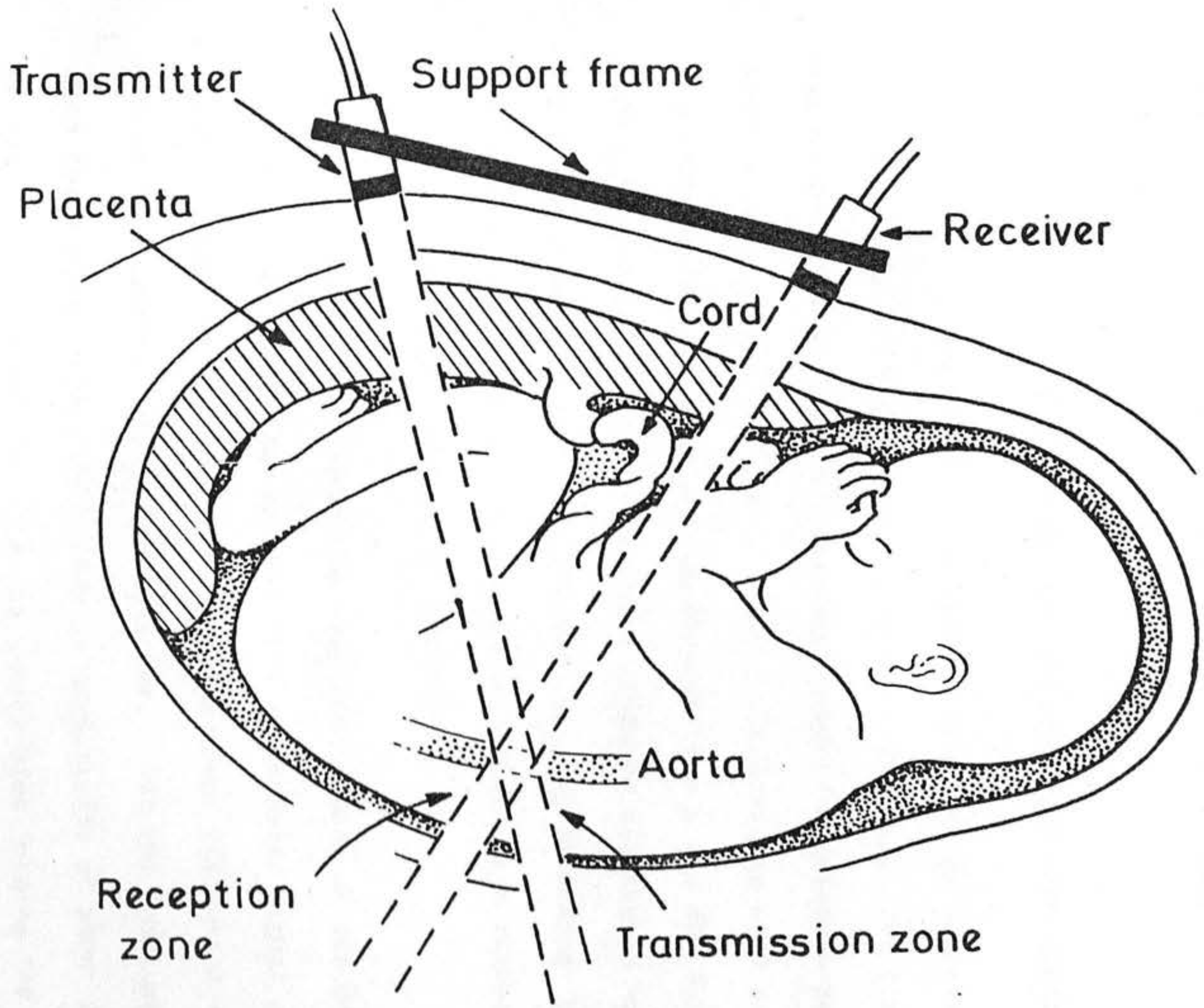
one element is used as the transmitter and the other as the receiver of the continuous wave ultrasound. The effective sensitivity region or sample volume of such an arrangement depends on the transmitting and receiving zones. The sample volume usually extends from close to the transducer to a depth determined primarily by the ultrasonic frequency. Figure 9.1 shows a beam plot obtained for a Sonicaid D205, 2.0 MHz, continuous wave Doppler instrument. This plot was constructed by noting the Doppler signal output level obtained when the transducer was moved transverse to a narrow tube flow phantom. The phantom consisted of a rubber tube, 2 mm inner diameter and 0.5 mm thick, through which a concentrated milk solution flowed. The tube was inclined to the Doppler beam at an angle of 80^{o}. A gravity feed arrangement was used giving a flow rate of approximately 2 ml s^{-1}. This will give a Doppler shift of approximately 850 Hz for the 2.0 MHz Doppler system.

In the C.W. intersecting zone Doppler the transmitting and receiving transducers are placed several centimetres apart. The transmitting and receiving transducer patterns can be altered to intersect in the region of the sample volume as indicated in Figure 9.2. Continuous wave ultrasound is transmitted. As blood is a Rayleigh scatterer of ultrasound it reflects the incident ultrasound wave in all directions (Shung et al., 1976). Part of this reflected energy is returned in the direction of the receiving zone. Only energy from the sample volume is received. The size of the received signal is not dependent on the angle between the transmission and reception zones. The energy from vessel wall echoes will tend to



Beam plot of a 2·0 MHz continuous wave Doppler

Figure 9.1



Intersecting zones

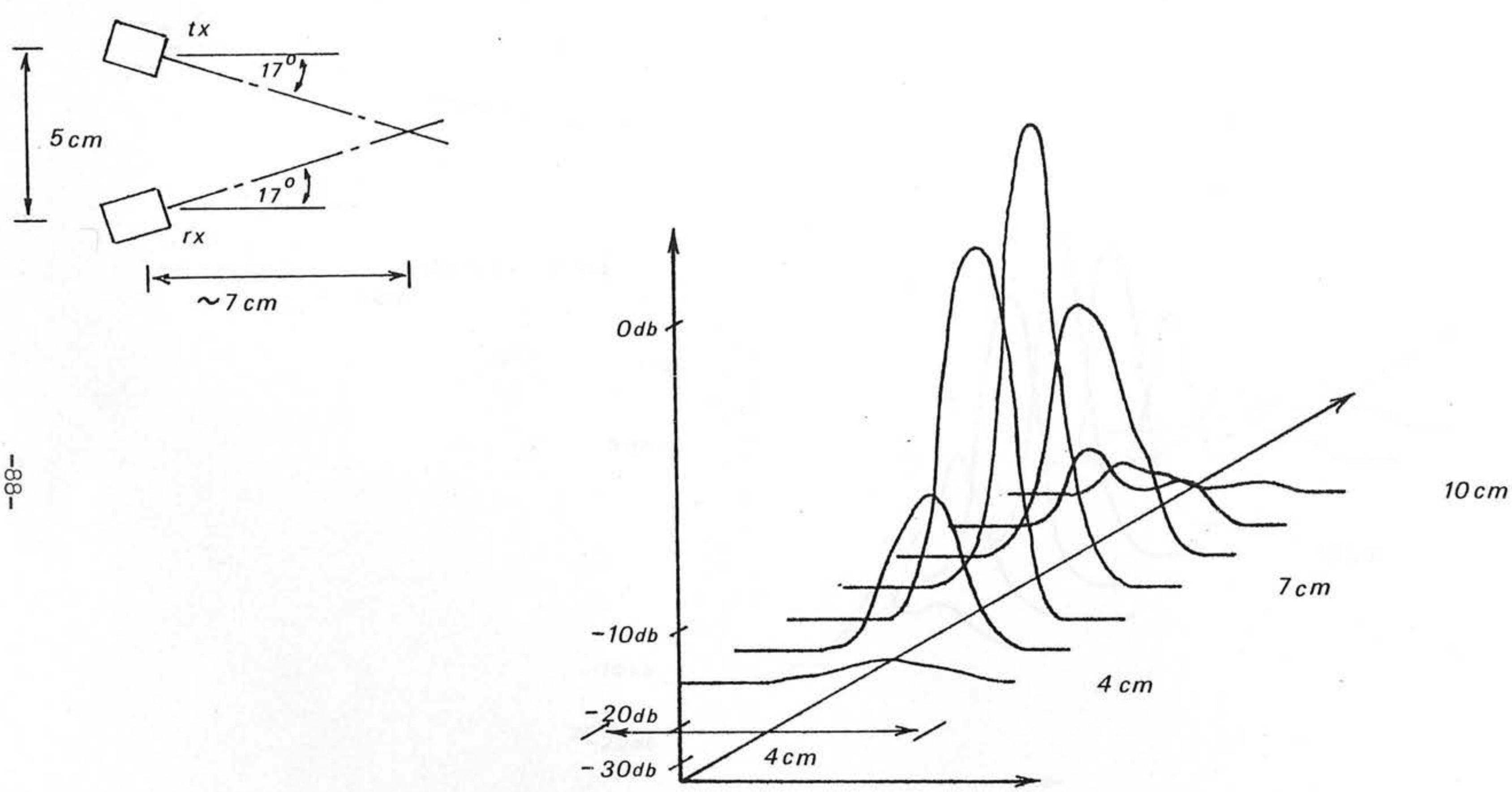
Figure 9.2

be returned in the direction of the transmitting transducer. This should enhance the separation of the vessel wall and blood flow signals. As this is a continuous wave system it is a low noise system which has no practical limit in the upper velocity measurement.

9.2 IN VITRO EVALUATION OF THE C.W. INTERSECTING ZONE DOPPLER

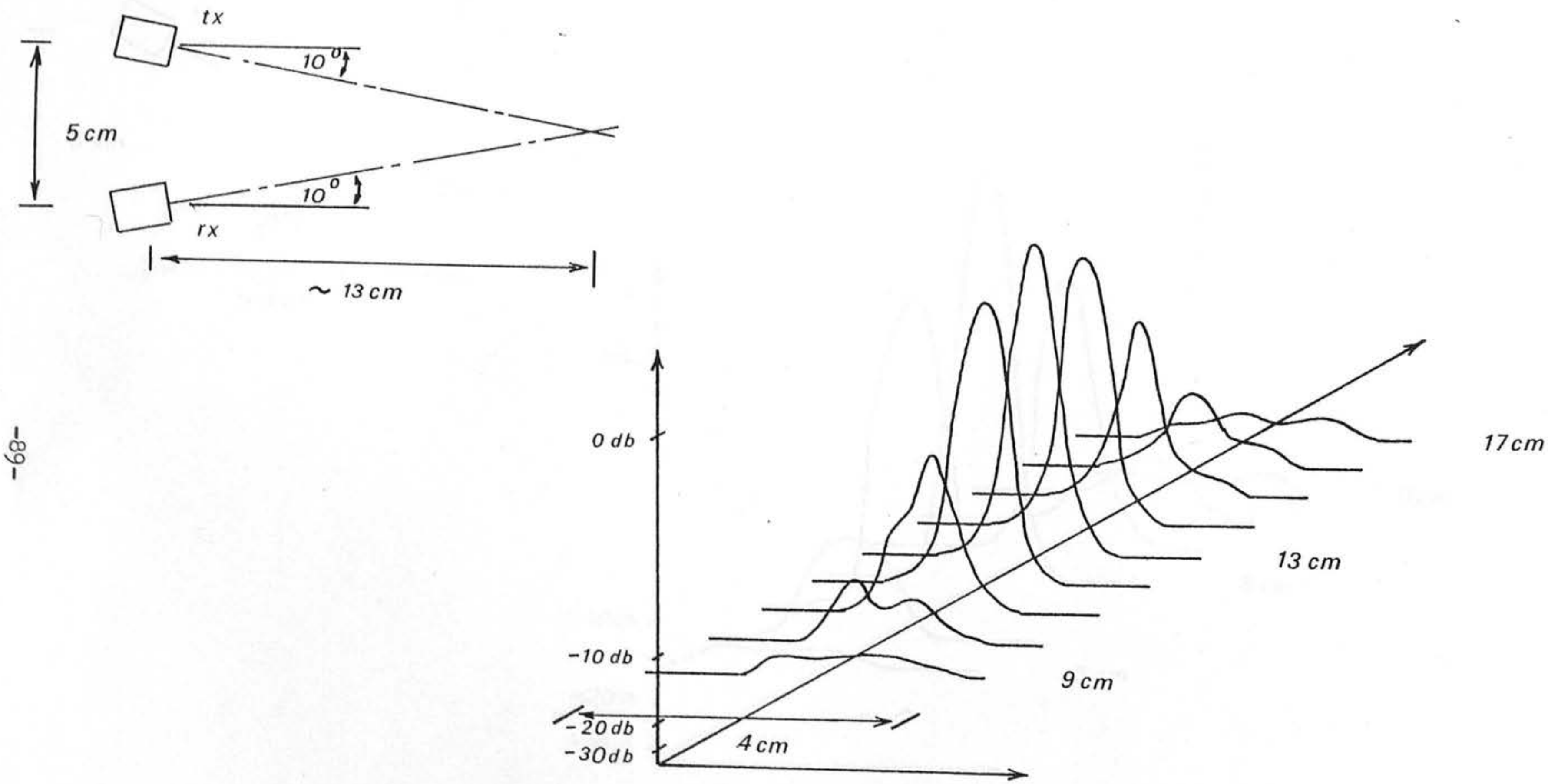
To evaluate the degree of localisation which can be obtained with this beam intersection a test rig was constructed. It consisted of two 15 mm diameter unfocused 2.0 MHz elements mounted in delrin holders and placed 5 cm apart. One transducer was used as the transmitter and the other as the receiver for a 2.0 MHz Sonicaid D205 C.W. Doppler. Each transducer could be rotated to vary the region of intersection of the transmitting and receiving zones. The test rig was immersed in a water bath lined with rubber foam to absorb any internally reflected ultrasound.

Figure 9.3 (a-c) illustrates the sensitivity zone of the Doppler unit. These plots were constructed by noting the Doppler signal output level obtained when the transducer test rig was moved transverse to the narrow tube phantom, as described above. For the instrument settings used these plots show that a zone of sensitivity of about 2 cm x 2 cm occurs at a range of 7 cm from the centre line joining the transducers. At 13 cm range the zone of sensitivity is about 4 cm x 2 cm. The zone size could be improved if focused transducers were used. Increasing the ultrasonic frequency to, say, 3.5 MHz would further improve the localisation.



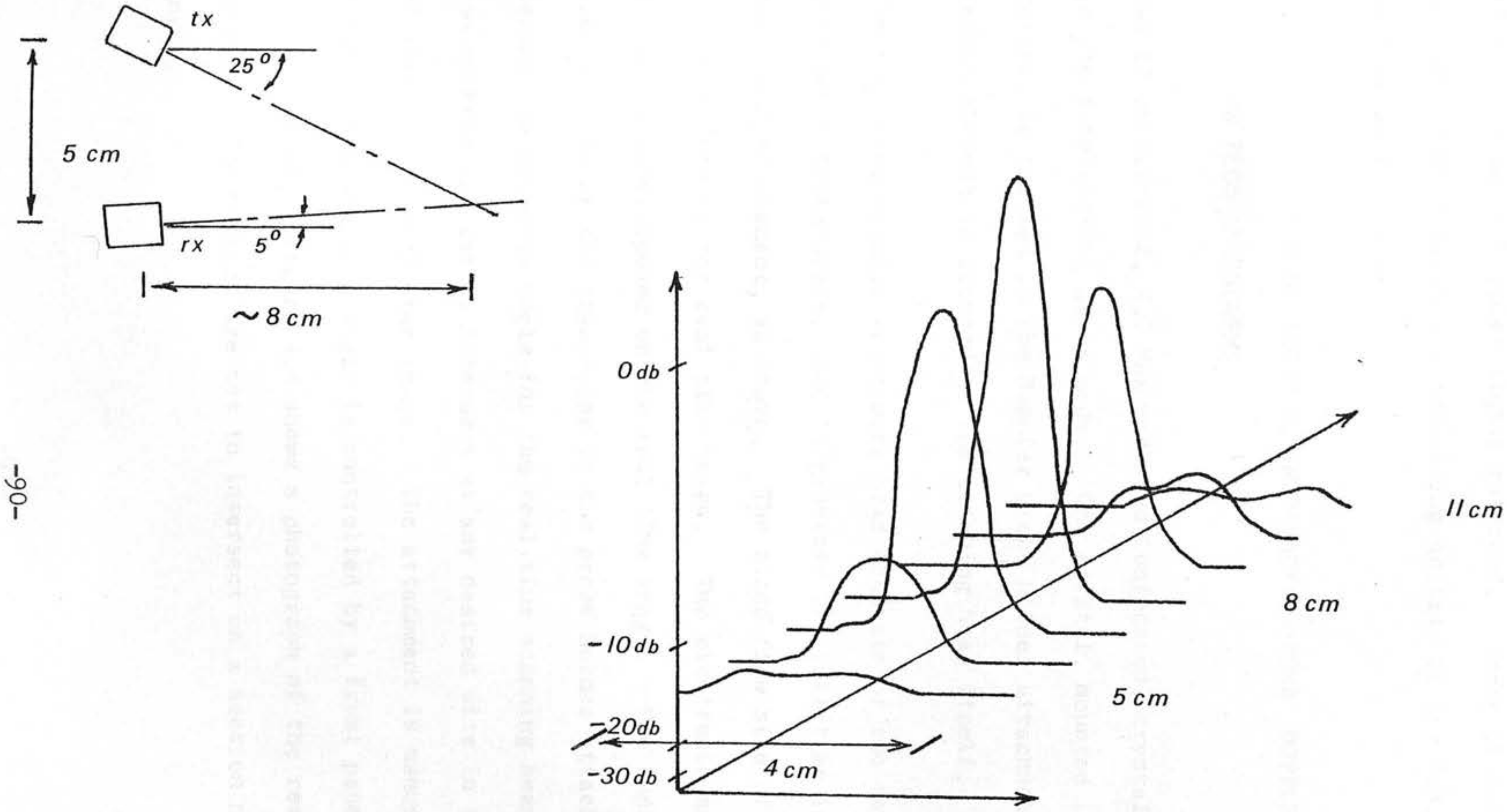
Intersecting zone beam plot

Figure 9.3a



Intersecting zone beam plot

Figure 9.3b



Intersecting zone beam plot

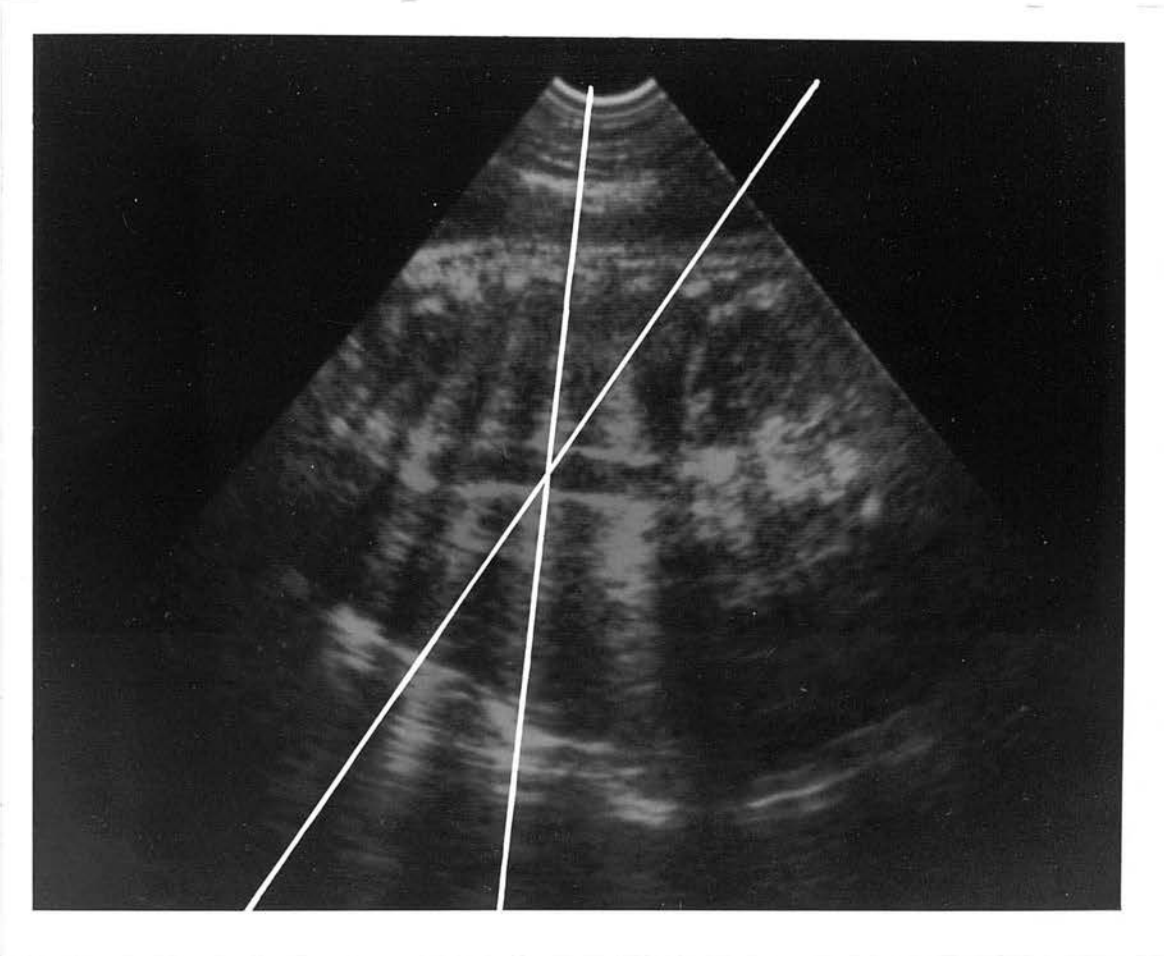
Figure 9.3c

A similar experiment was performed using the large Perspex tube phantom (10 mm inner diameter) as described in Chapter 5. Only when the zone of intersection of the two beams interrogated the fluid flow was a Doppler signal returned. These phantoms demonstrated in vitro the localising ability of the C.W. intersecting zone Doppler.

9.3 IMPLEMENTATION OF THE C.W. INTERSECTING ZONE DOPPLER IN THE BLOOD FLOW INSTRUMENT

Two 15 mm diameter, 2.0 MHz undamped, unfocused, crystals are used in the intersecting zones mode. One crystal, mounted in a delrin holder, is located in the Doppler probe holder attachment. The second element is located in the scanning head itself.

The real time scanner transducer head consists of two damped ultrasonic transducers, used to generate the real time image, and one undamped element, as above. The blood flow site of interest is identified on the real time image. Two electronic markers are displayed superimposed on the real time image. One indicates the line of site of the transducer in the probe holder attachment, the second the stopping angle for the real time scanning head. The two markers are set to intersect at any desired site in the field of view of the real time image. The attachment is manually rotated, the stopping angle is controlled by a front panel potentiometer. Figure 9.4 shows a photograph of the real time screen. The markers are set to intersect on a section of foetal aorta.



Real time image with electronic markers

Figure 9.4

When a blood flow site has been selected, the real time scanning head is quickly brought to a halt with the undamped transducer stopping at the desired angle. Intersecting zone operation is automatically selected. The transducer in the scanning head is normally used as the transmitter and the separate probe as the receiver for the Sonicaid D205 instrument. The roles of the transducers may be reversed.

The effective direction of the ultrasonic beam, the angle of attack, will lie approximately mid-way between the transmitting and receiving beams. Care is taken to ensure that the angle between the blood vessel and this angle of attack does not approach 90^{o}.

9.4 IN VIVO RESULTS USING THE C.W. INTERSECTING ZONE DOPPLER

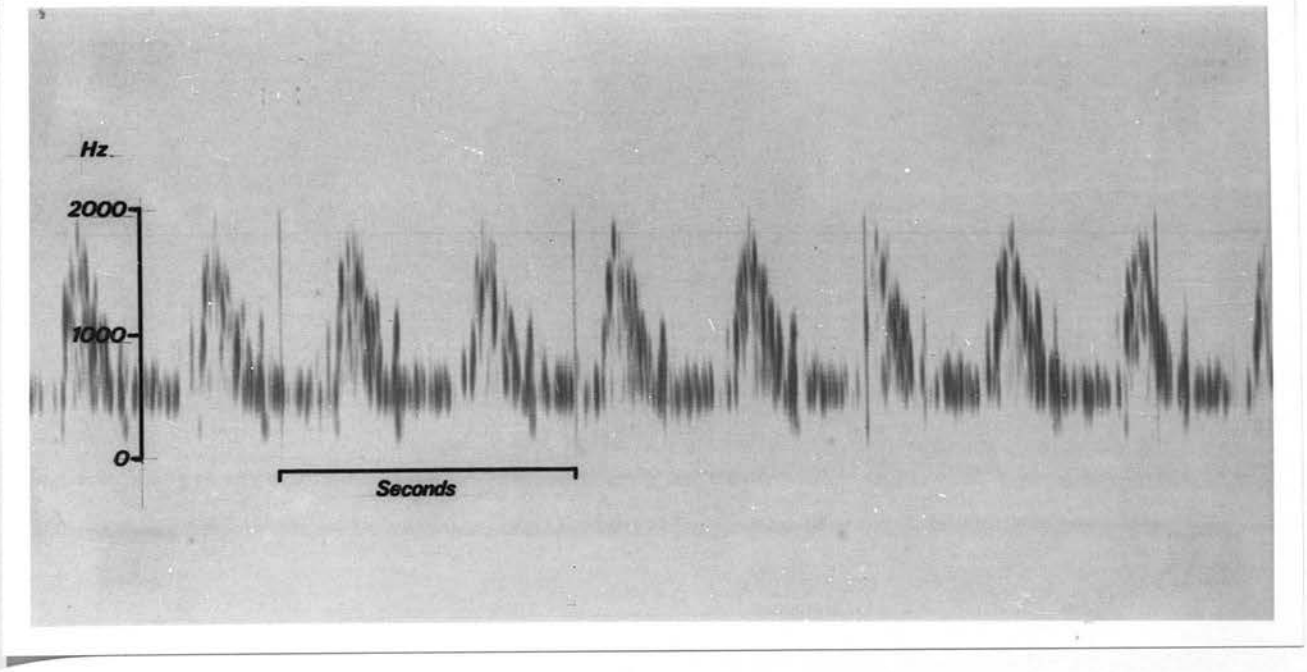
Blood flow signals from foetal aorta, umbilical cord and placenta have been recorded using the intersecting zone Doppler. In order to enhance the signal to noise ratio of such recordings the output power of the Sonicaid D205 Doppler was increased by 6 db. This raises the output ultrasonic power to approximately 20 $mwatts/cm^2$.

Over forty normal pregnancies have been examined with this technique. In each case pregnancies of over 35 weeks were chosen and total examination time was restricted to about one half hour. It was decided to concentrate primarily on recording aortic flow.

The recording of foetal aortic blood flow has proved to be very successful. In over 80% of the patients an acceptable quality of aortic flow signal has been obtained. The flow spectrograms range from the characteristic hollow type to the more broadband, filled in spectrogram. Figure 9.5 shows a typical blood flow spectrogram obtained from the foetal aorta.

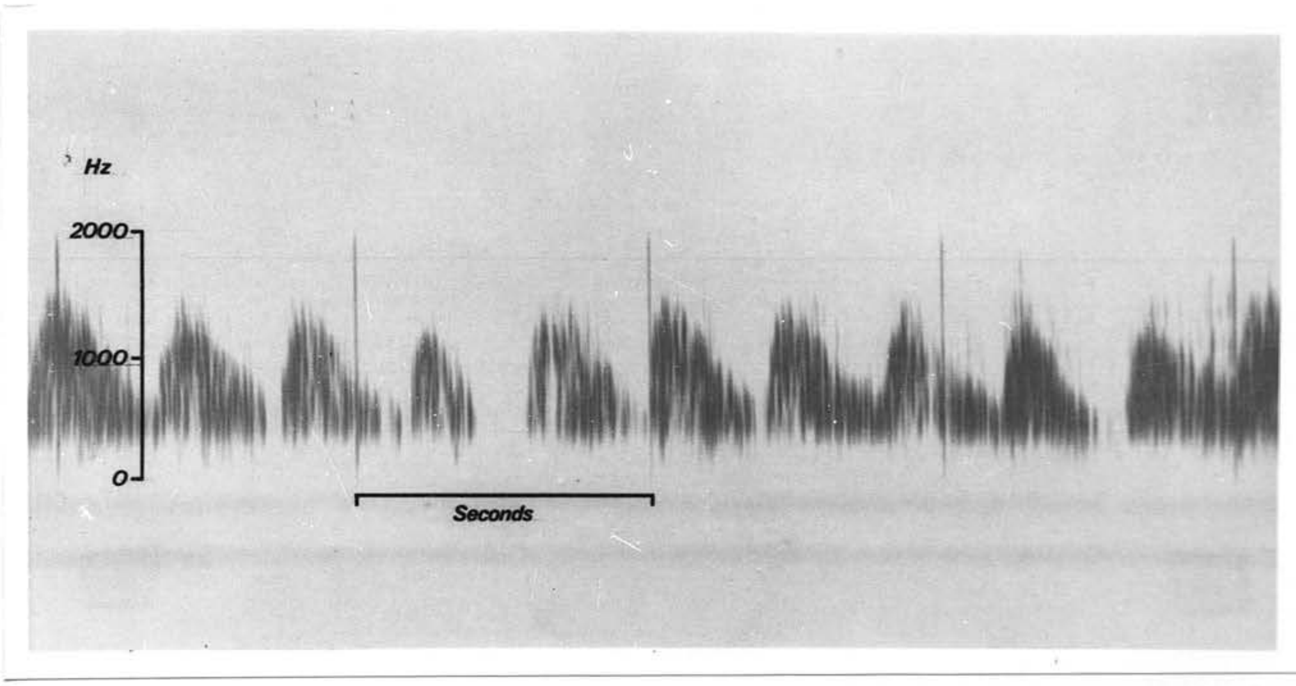
Umbilical cord flow has been recorded by locating the intersecting zones on a large coil of cord. The cord flow is discontinuous due to cord motion in and out of the sample volume. Figure 9.6 shows a typical blood flow spectrogram obtained from cord. In all but a few active foetuses recordings can be obtained for quite long periods, typically 1-2 minutes with little variation in signal amplitude or quality. In foetal aortic flow recordings changes in flow rate have been noted on a number of patients during a recording. Vessel wall motions do not appear to have nearly so much of an effect on the blood flow signals as was experienced when using the pulsed wave Doppler. Flow signals have been recorded from deep within the uterus, up to 15 cm or more.

The intersecting zones technique has been used to explore various regions within the placenta. Low level blood flow signals have been recorded from different sites. At sites in the vicinity of the umbilical cord the flow rates show the influence of cord flow. At sites away from this region the flow signals recorded are unrelated to either foetal or maternal heart rates. Different regions within the placenta produce different flow patterns. Figure 9.7 shows typical blood flow spectrograms obtained at two different sites within the placenta. A zero signal or background noise level, for comparative purposes, is obtained by placing the intersecting zones sample volume in a pool of liquor. Comparison of this zero signal with that obtained from within the placenta is a useful measure of



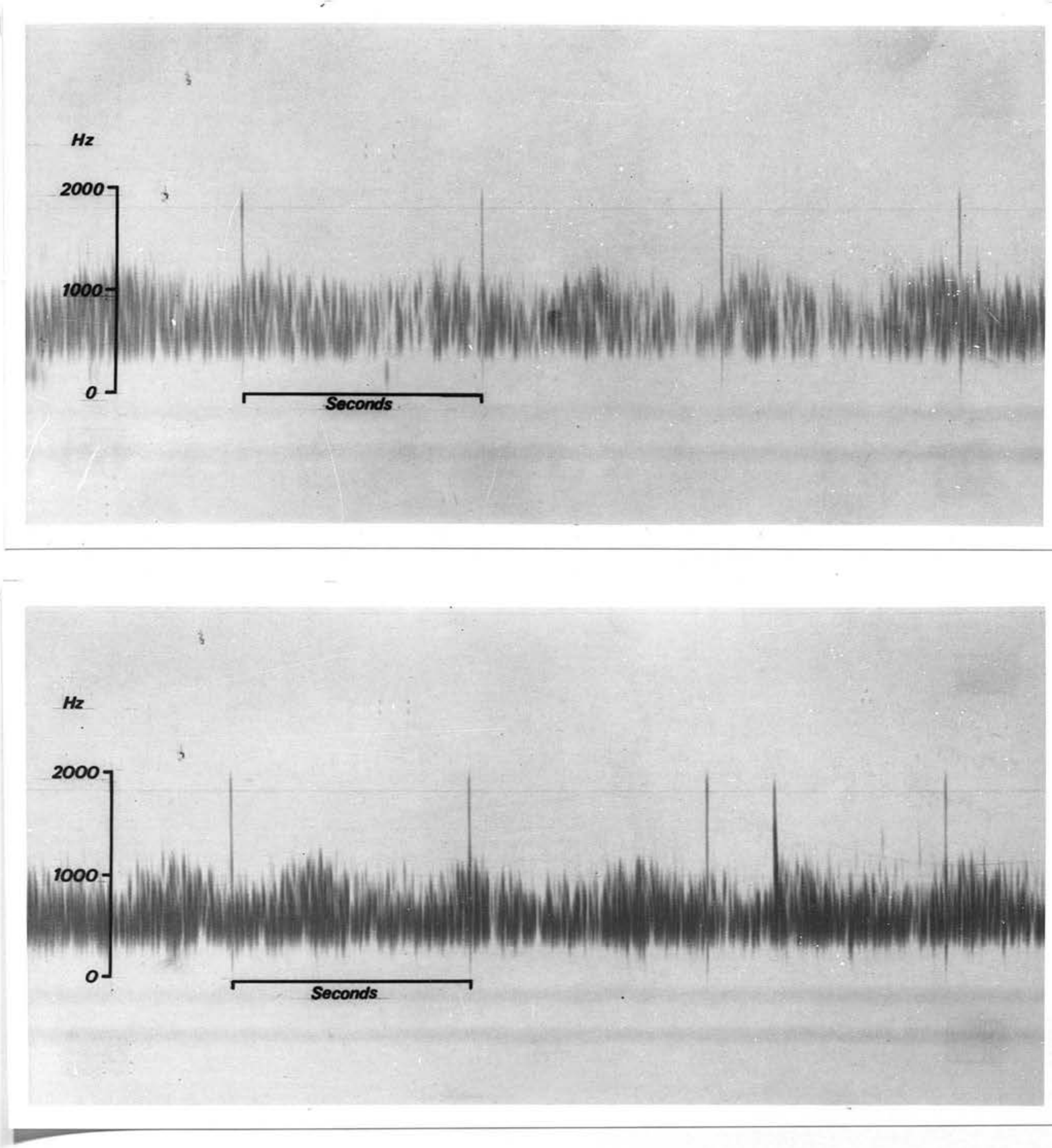
Intersecting zone Doppler spectrogram from foetal aorta

Figure 9.5



Intersecting zone Doppler spectrogram from cord

Figure 9.6



Intersecting zone Doppler spectrograms from two different sites within the placenta.

Figure 9.7

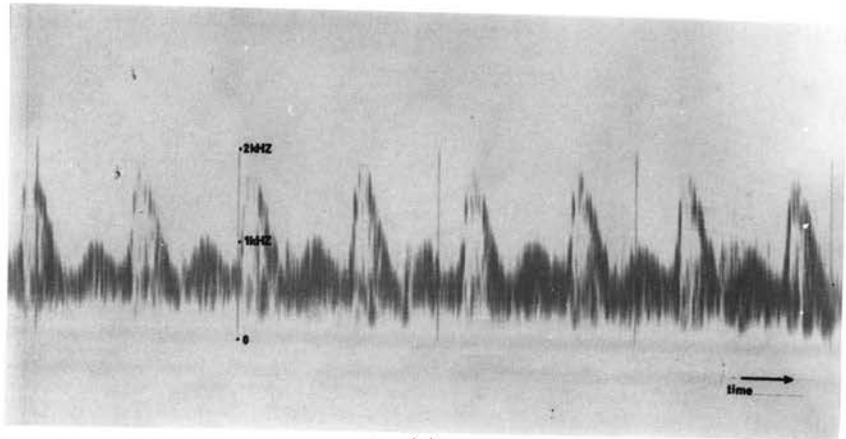
the ability of the equipment to record placental flow. The low frequency cut-off filter inherent in the Sonicaid Doppler has not been found to remove any useful flow data. The recording of flow from within the placenta is extremely encouraging as it represents detection of low level low frequency blood flow in the presence of high level static signals from the vascular structure of the placenta.

Figure 9.8 shows a comparison of blood flow signals recorded by the continuous wave and intersecting zones Doppler. The signals were obtained from the placenta at a region near the chorionic membrane. The rate of pulsation indicated that the dominant vessel was influenced by the foetal circulation.

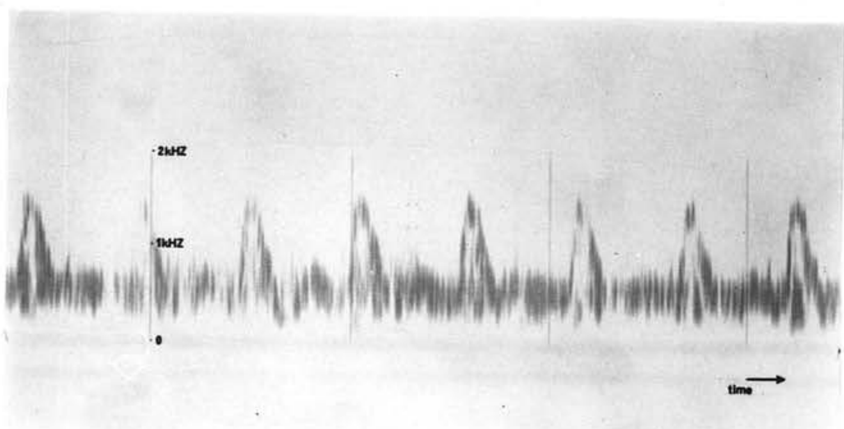
9.5 CALCULATION OF BLOOD FLOW IN THE FOETAL AORTA

Calculations of blood flow in the foetal aorta have been made. Vessel diameter and beam inclination angles α and β are calculated from a polaroid photograph taken from the real time screen just prior to a Doppler recording. The spectrogram obtained with the filter bank analyser is used to compute the peak Doppler frequency component. A detailed description of the method of flow computation is given in Appendix 1.

There are a number of errors inherent in these calculations. The major source of error arises from the calculation of the vessel cross-sectional area. The foetal aorta has been assumed to be a cylindrical vessel of constant diameter. The error in this measurement is currently $\pm$ 1 mm. In practice the foetal aorta is a



(a)



Flow signals from a dominant fetal vessel

(a) C.W. Doppler

(b) Intersecting Zones

A comparison of flow signals recorded by the continuous wave and intersecting zone Doppler techniques. The signals were obtained from the placenta at a region near the chorionic membrane. The rate of pulsation indicated that the dominant vessel detected was influenced by the foetal circulation.

Figure 9.8

pulsating vessel with a continually changing cross-sectional area.

The peak frequency in the Doppler spectrogram can be measured with an accuracy of $\pm$ 100 Hz. An error of $\pm 2^{\circ}$ in the measurement of the beam angle has been assumed.

The shape of the foetal aortic flow spectrogram is indicative of the flow pattern. The hollow type of spectrogram shown in Figure 9.5 suggests a fairly flat velocity profile across the vessel, whereas a more broadband spectrogram is more likely to arise from a parabolic velocity profile. The instantaneous average velocity is equal to the instantaneous peak velocity for plug flow and one half this value for parabolic flow. A decision as to which type of flow occurs is crucial to the final calculation.

Results obtained for foetal aortic blood flow in six normal pregnancies are presented in Table 9.1. The flow rate is given assuming a parabolic velocity profile across the vessel. This assumption is made because the spectrograms used were all broadband in nature. The gestational age data was obtained from patient records.

Clinicians often quote such flow rates in ml $min^{-1}kg^{-1}$. Table 9.2 shows the flow rate presented in this format. The median weight for a foetus of the appropriate gestational age has been used. Eik-nes et al. (1980) quote a mean blood flow in the descending part of the foetal aorta of 191 $\pm$ 12 ml $min^{-1}kg^{-1}$. The authors assumed a flat velocity profile in the vessel.

The results presented indicate values higher than those reported elsewhere. The large errors involved in such calculations must be borne in mind. Also, there are many hidden variables not appearing in the volume flow equation which will influence the final measurement. These variables, such as the interaction of the beam and vessel geometries, still require investigation before any such Doppler techniques can truly be said to produce quantitative data. The current equipment is best suited to monitoring changes in flow pattern rather than to the calculation of absolute values.

Table 9.1

Foetal aortic flow in six normal pregnancies

Patient	Gestational age (weeks)	Angle (degrees) α	β	Average velocity ($cm\ s^{-1}$)	Vessel diameter (mm)	Flow ($ml\ min^{-1}$)
1	37	85	124	32 ± 5.7	5	628 ± 275
2	36	95	133	22.7 ± 1.8	5	445 ± 182
3	37	94	131	25.6 ± 2.6	7	985 ± 299
4	38	81	130	33.4 ± 3.7	5	657 ± 272
5	37	95	132	26.9 ± 2.1	5	528 ± 215
6	37	96	134	24.9 ± 1.8	6	704 ± 240

Table 9.2

Foetal aortic flow expressed in $ml\ min^{-1}\ kg^{-1}$

Patient	Gestational age (weeks)	Average weight (kg)	Flow $ml\ min^{-1}\ kg^{-1}$
1	37	2.8	224 ± 98
2	36	2.6	171 ± 70
3	38	3.0	328 ± 99
4	38	3.0	219 ± 90
5	37	2.8	188 ± 76
6	37	2.8	251 ± 86

CHAPTER 10

A COMPOSITE BLOOD FLOW INSTRUMENT

It has been the aim of this work to investigate the use of ultrasonic Doppler techniques for the study of blood flow at selected sites in the pregnant uterus. A composite blood flow instrument has evolved which combines the use of real time visualisation with three types of Doppler unit. The standard double crystal continuous wave Doppler, the pulsed wave Doppler and the continuous wave intersecting zone Doppler instruments have been evaluated and their relative advantages and disadvantages studied. A clinical evaluation of each of these Doppler techniques has been performed.

10.1 MODES OF OPERATION

The study of blood flow in the pregnant uterus presents some difficult technical problems. The blood flow site of interest may be at a depth of ten or more centimetres from the skin surface. There is a large variation in flow velocities, very low velocity flow in the presence of large static echoes in the placenta, to quite high velocity flow in vessels such as foetal aorta. Foetal motion may also cause problems.

With the blood flow instrument any of the three Doppler techniques may be selected, whichever is best suited to the particular blood flow site. Four basic modes of operation can be selected using the dual probe system. These are:

(a) pulsed wave Doppler operation using a damped ultrasonic transducer in the real time scanner head.

(b) pulsed wave Doppler operation using a pulse echo transducer located in the probe holder attachment.

(c) continuous wave Doppler operation using a C.W. probe located in the probe holder attachment.

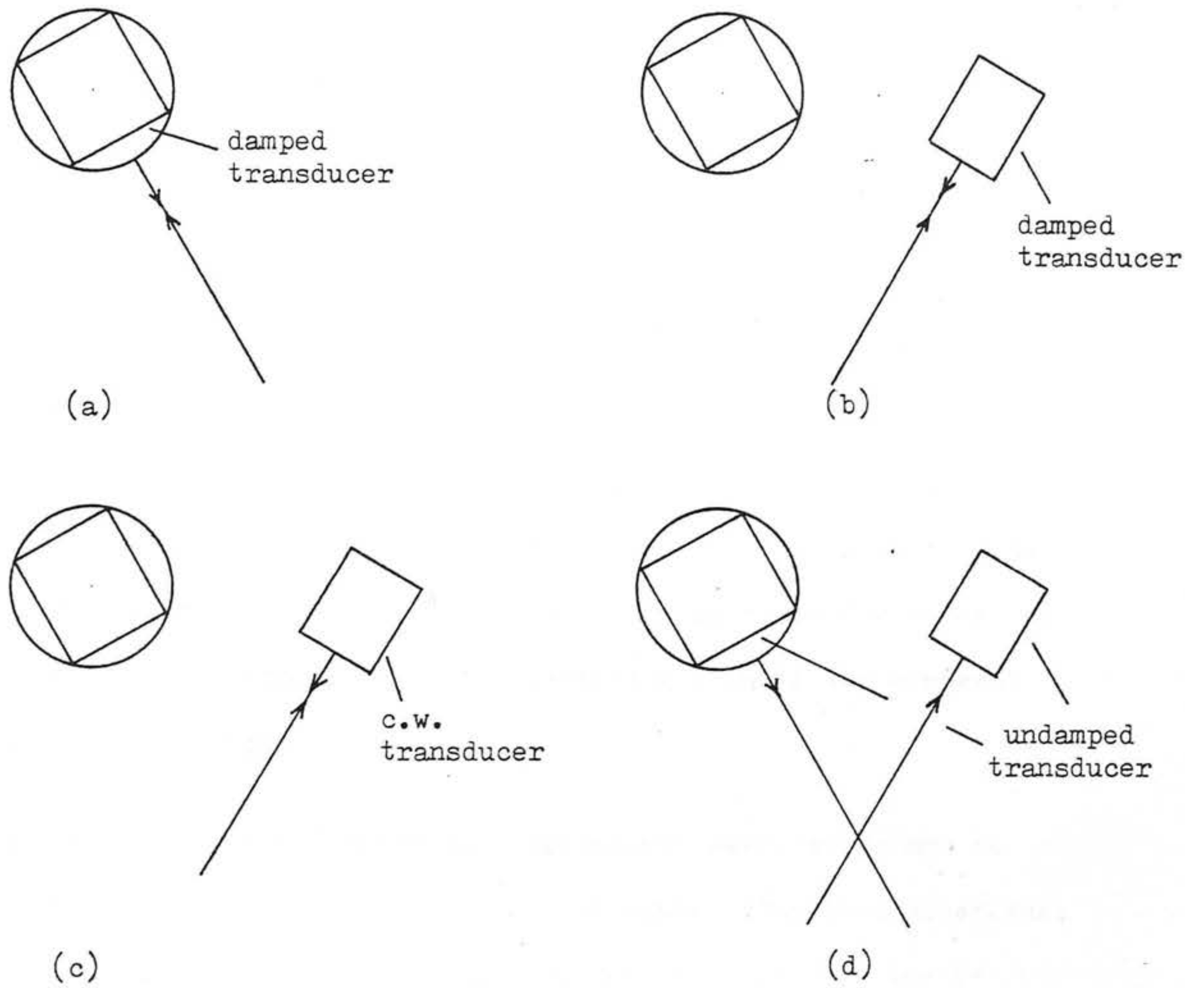
(d) intersecting zone Doppler operation. The undamped transducer in the real time scanning head is used as the transmitter and an undamped transducer in the probe holder as receiver for a continuous wave Doppler instrument.

These modes of operation are illustrated in Figure 10.1.

10.2 THE RELATIVE MERITS OF THE THREE DOPPLER SYSTEMS

The continuous wave Doppler instrument is a low noise system with no practical upper velocity limitation, but no range resolution. In rare situations where the real time scan has shown that only the blood flow site of interest will produce a Doppler shift, good quality blood flow recording from foetal aorta, umbilical vein and cord have been achieved. With the 2.0 MHz Doppler such recordings can be made at depths of up to 15 cm.

Blood flow recordings at specific depths can be achieved with the pulsed wave Doppler instrument. The low ultrasonic frequency and the need for vessel straddling result in poor range resolution. The equipment is usable to a depth of about 12 cm. The 2.5 MHz pulsed wave Doppler has been used to record blood flow from foetal aorta and from within the foetal heart chambers. Vessel wall motions produce large amplitude low frequency Doppler shifts which



Modes of operation of the dual probe system

(a) A damped transducer in the real time scanning head is stopped at the desired angle. The pulsed wave Doppler is used with this transducer.

(b) Real time scanner and a damped transducer. The real time scanner is stopped and the pulsed wave Doppler is used with the separate damped transducer.

(c) Real time scanner and a continuous wave transducer. The real time scanner is stopped and c.w. Doppler operation commenced.

(d) Real time scanner and an undamped transducer. The undamped transducer in the real time scanning head is stopped at the desired angle. This transducer is used as the transmitter and the separate transducer as the receiver for continuous wave intersecting zone Doppler operation.

Figure 10.1

unless removed by filtering will mask the blood flow signal making it inaudible. The sensitivity of the pulsed wave Doppler is less than that of a C.W. device having an equivalent specification. This may be due to the pulsed wave Doppler's broadband characteristics. The pulsed wave Doppler's sensitivity is reduced at sites where blood flow occurs in the presence of large static echoes. This particular instrument appears to be of use in recording flow from large, well defined vessels when used with a wide transmitter pulse and sample volume. It does not appear to be suitable for recording very low velocity blood flow or for detecting flow in the presence of large static echoes.

The intersecting zones Doppler is a continuous wave device and is thus a low noise system with no practical upper velocity limitations. It is as sensitive as the standard C.W. device. In vivo and in vitro trials have shown that the instrument exhibits a fair degree of localisation of the Doppler flow signal. Blood flow recordings from foetal aorta, umbilical vein, cord and placenta have been made. The detection of low velocity flow from within the vascular structure of the placenta is extremely encouraging. It illustrates the high sensitivity and the low noise characteristics of the system. It is a simple system to operate.

10.3 SUMMARY

The pulsed wave Doppler technique seemed appropriate to the study of foetal haemodynamics. Experience with such a system has shown a number of difficulties. It is not as sensitive as an equivalent continuous wave device. It suffers from masking effects when detecting flow in the presence of large static echoes, and has an upper maximum velocity measurement limitation. Clinical experience with such a system has shown its use to be limited to investigating flow in large vessels such as foetal aorta.

The recording of blood flow within the pregnant uterus can be achieved with a continuous wave device (Fitzgerald et al., 1977). All motion within the ultrasound beam is detected. This lack of localisation of the Doppler information severely restricts the use of such a technique.

The continuous wave intersecting zone Doppler technique has proved useful. Clinical trials have shown that a fair degree of localisation can be achieved. Blood flow spectrograms from foetal aorta, cord and from within the placenta have been recorded using this technique.

The composite blood flow instrument is well suited to monitoring blood flow patterns at various sites within the uterus. It is not yet capable of producing quantitative blood flow data.

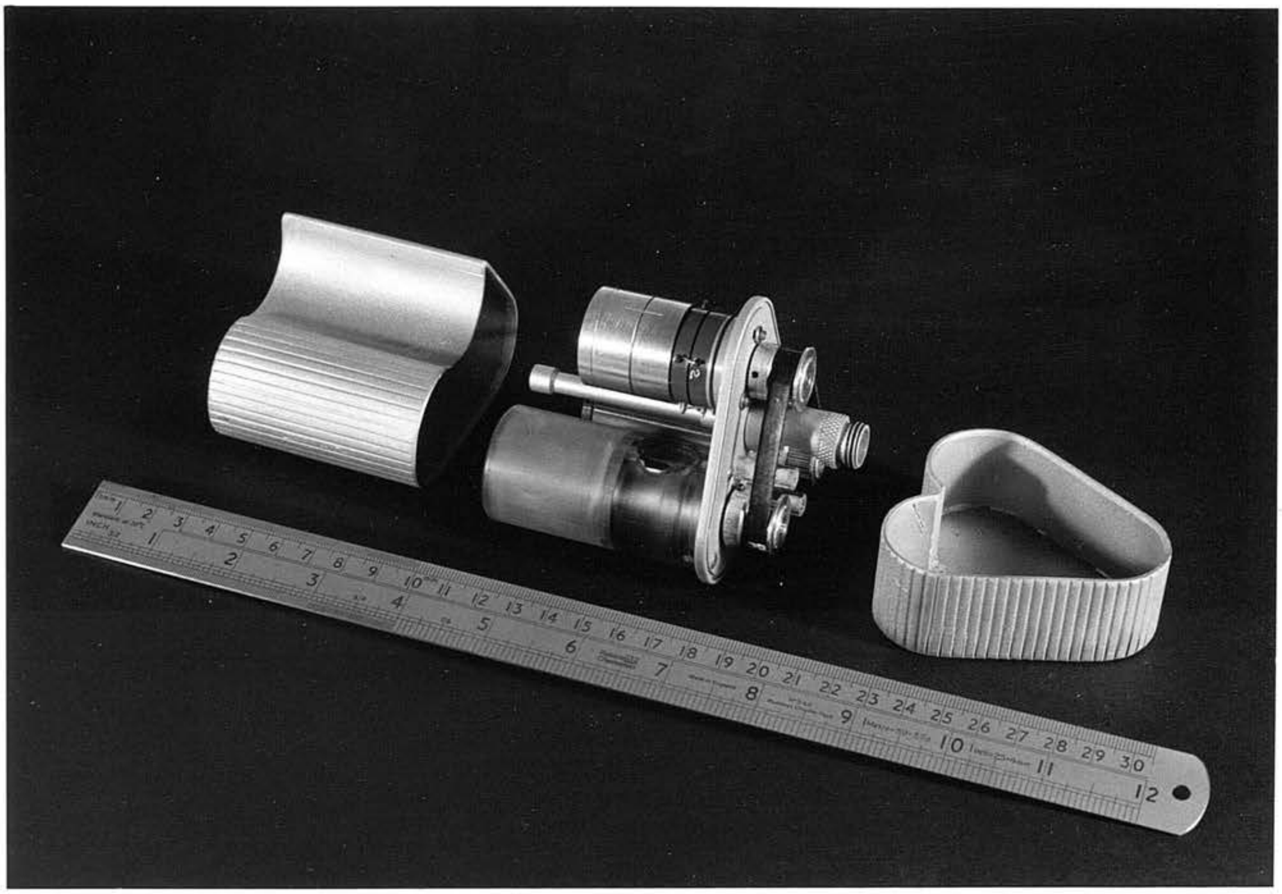
To date, confidence in the equipment and its ability to detect flow is being built up. Clinicial trials have concentrated on studying normal foetuses in late pregnancy. The investigation of foetal blood flow at various gestational ages will be the next application of this equipment.

10.4 FURTHER DEVELOPMENTS

There are a number of improvements to the basic blood flow system described which are either in progress or will be implemented in the future. The modifications attempt to update the basic development system to an instrument suitable for routine clinical use.

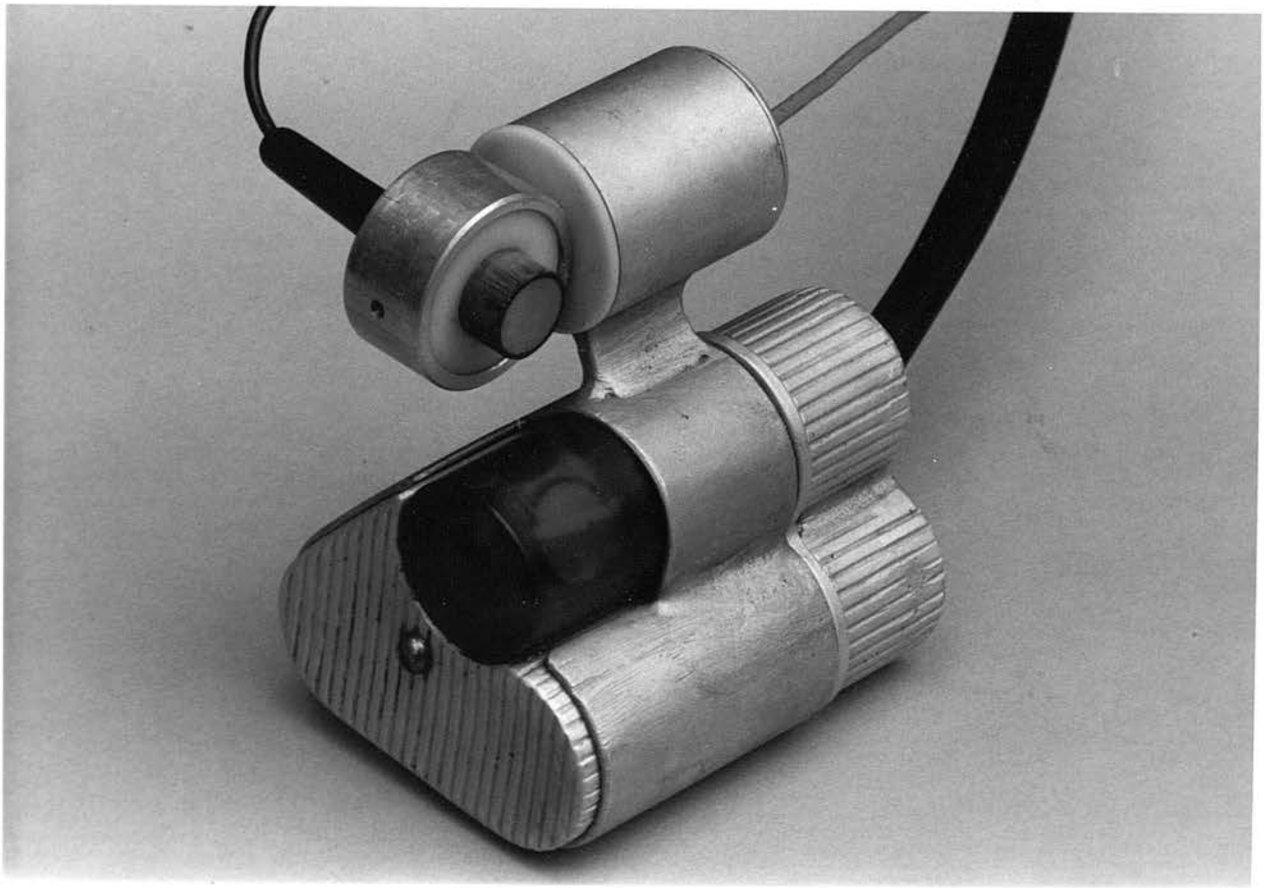
The pulsed wave Doppler used with this equipment is a basic non-directional system operating at 2.5 MHz. I have designed an updated version of this equipment. This design is for a directional instrument capable of operating at ultrasonic frequencies of 2.5 MHz, 3.5 MHz or 5.0 MHz. It uses coherent product phase quadrature detection followed by a phase domain processor using the SQ decoder integrated circuit as used by Coughlan et al. (1978). This equipment is now operational. Work remains to be done on calculating the pulsed wave Doppler sample volume for various equipment settings.

The main difficulty with the basic horizontal barrel real time scanner was the lack of an angle measuring device. This basic scanner has been superseded by a 'tri-axis scanner' which incorporates a sine/cosine potentiometer capable of high speed rotation. A transducer in the scanning head can now be stopped to an accuracy of $\pm \frac{1}{2}^{\circ}$. A photograph of this scanner and the composite scanning head are shown in Figures 10.2 and 10.3. The 2.5 MHz pulse echo transducers used in the scanning head are to be replaced by 3.5 MHz transducers in order to improve picture quality. Electronic calipers will be used to measure vessel diameter.



The 'tri-axis' scanner

Figure 10.2



The 'tri-axis' blood flow scanning head

Figure 10.3

APPENDIX 1

CALCULATION OF BLOOD FLOW USING THE INTERSECTING ZONE DOPPLER

The Doppler effect describes the phenomenon that the observed frequency of a constant frequency source, f, is dependent on the relative motion of the source and the observer. Figure A1.1 illustrates a typical beam and vessel axis configuration in the intersecting zone Doppler. The angle α is the angle between the receiver and the vessel axis and the angle β is the angle between the transmitter and the vessel axis. In this situation the Doppler equation is given by,

$$f_D = \left(\frac{C - v\cos\alpha}{C + v\cos\beta} - 1\right) f \quad \ldots\ldots\ldots\ldots\ldots \text{(a)}$$

where, f_D is the Doppler frequency shift,

v is the velocity of the flow, and

C is the velocity of ultrasound in blood (1570 m/s)

if $C \gg v$ the above equation may be rewritten as

$$f_D = \frac{-fv}{C}(\cos\alpha + \cos\beta) \quad \ldots\ldots\ldots\ldots\ldots \text{(b)}$$

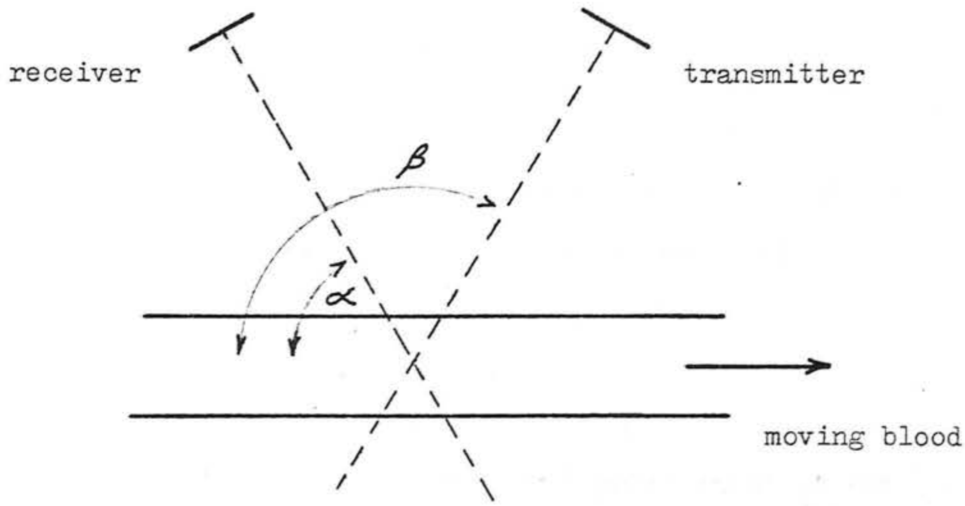
This can be rearranged to give the well known equation,

$$v = \frac{-f_D . C}{2 f \cos\gamma} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \text{(c)}$$

where,

$$\cos\gamma = \cos\frac{(\alpha + \beta)}{2} \cdot \cos\frac{(\alpha - \beta)}{2}$$

and γ is the effective angle of attack.



Typical beam and vessel axis configerations
in the intersecting zones Doppler

Figure A1.1

The calculation of flow through a vessel can be computed from the product of average velocity and vessel cross-section, that is

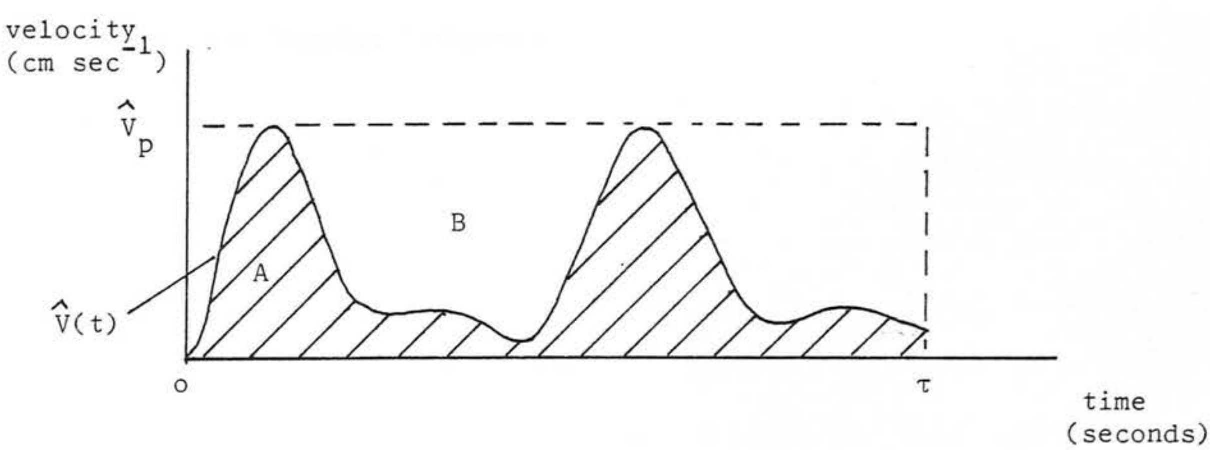
<u>Flow = average velocity X cross-sectional area</u>

Cross-sectional area

The cross-sectional area is computed by taking a measure of the vessel diameter from the real time screen. A cylindrical vessel of constant diameter is assumed. In a pulsating vessel such as the foetal aorta this assumption is incorrect, but is a convenient assumption.

Average velocity

Consider the spectrogram shown below. By the Doppler equation the envelope of the spectrogram is a measure of the instantaneous peak velocity of flow in the vessel. The assumption is made that the velocity flow profile across the vessel is parabolic. For this condition instantaneous average velocity is equal to one half the instantaneous peak velocity.



The area under the spectrogram (the hatched area) is directly proportional to volume flow. Consider the time period τ, greater than one foetal cardiac cycle.

$$\text{Average velocity} = \frac{1}{\tau} \int_{0}^{\tau} \frac{\hat{V}(t)}{2}\, dt$$

$$= \frac{\hat{V}_p}{2} \times \frac{\text{Area (A)}}{\text{Area (A+B)}}$$

The area ratios of a section of spectrogram are estimated by tracing the envelope velocity waveform onto mm graph paper and counting the squares.

In summary

Flow = average velocity x cross-sectional area

$$= \frac{\hat{f}_D}{2} \times \frac{C}{2f \cos\gamma} \times \left[\frac{\text{Area A}}{\text{Area (A+B)}}\right] \times \frac{\pi d^2}{4} \quad m^3\,sec^{-1}$$

$$= \frac{\hat{f}_D}{2} \times \frac{C}{2f \cos\gamma} \times \left[\frac{\text{Area A}}{\text{Area (A+B)}}\right] \times \frac{\pi d^2}{4} \times 10^6 \times 60 \ ml\ min^{-1}$$

where, d = vessel diameter (metres)

f = ultrasonic frequency = 2.0 MHz

C = 1570 m sec^{-1}

$\hat{f}_D$ = peak Doppler frequency (Hz)

APPENDIX 2

THE RECORDING OF HUMAN FOETAL BREATHING MOVEMENTS USING A PULSED WAVE DOPPLER INSTRUMENT

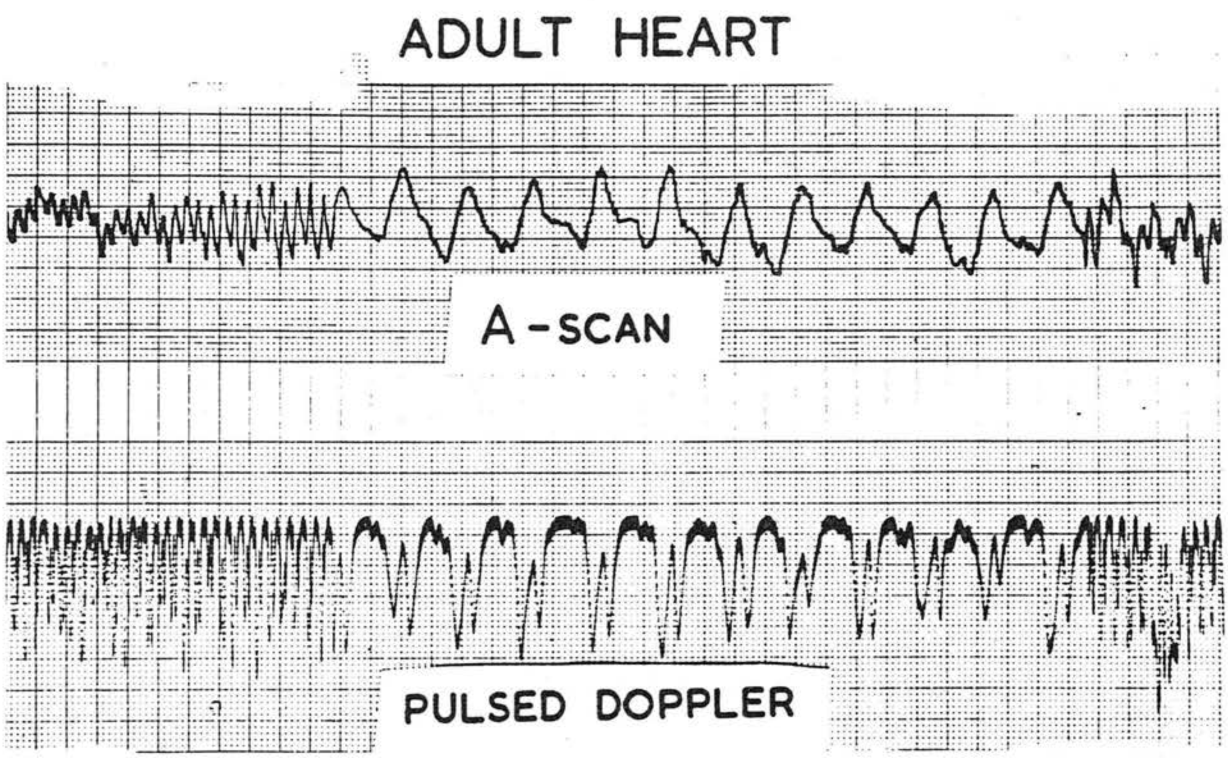
Foetal breathing movements in human pregnancy have been shown to be a valuable indicator of foetal wellbeing (Boddy et al., 1971) and the monitoring and interpretation of such motion is becoming an established clinical research technique. Foetal breathing movements of the chest or abdominal wall can be clearly seen on real time images of the foetus in the third trimester of pregnancy.

An established technique for monitoring such foetal breathing movements involves recording the motion of a distinct chest or abdominal wall echo using an A-scan machine with a gated T-M trace. This technique has disadvantages (Farman et al., 1975). A continuous wave Doppler instrument has been used to monitor foetal breathing activity (Boyce et al.,1976 and Gough et al., 1979). This technique is said to record blood flow in the great veins below the foetal heart.

The pulsed wave Doppler instrument is well suited to recording tissue motion. It was found that by placing the range gate of the Doppler instrument to lie within the foetal chest or abdomen the Doppler output gives a signal which correlates well with the A-scan technique recording of foetal breathing motions. The real time scanner is used for initial location of the sample volume on the chest wall. The Doppler output is fed through a zero crossing detector to give a suitable paper trace recording.

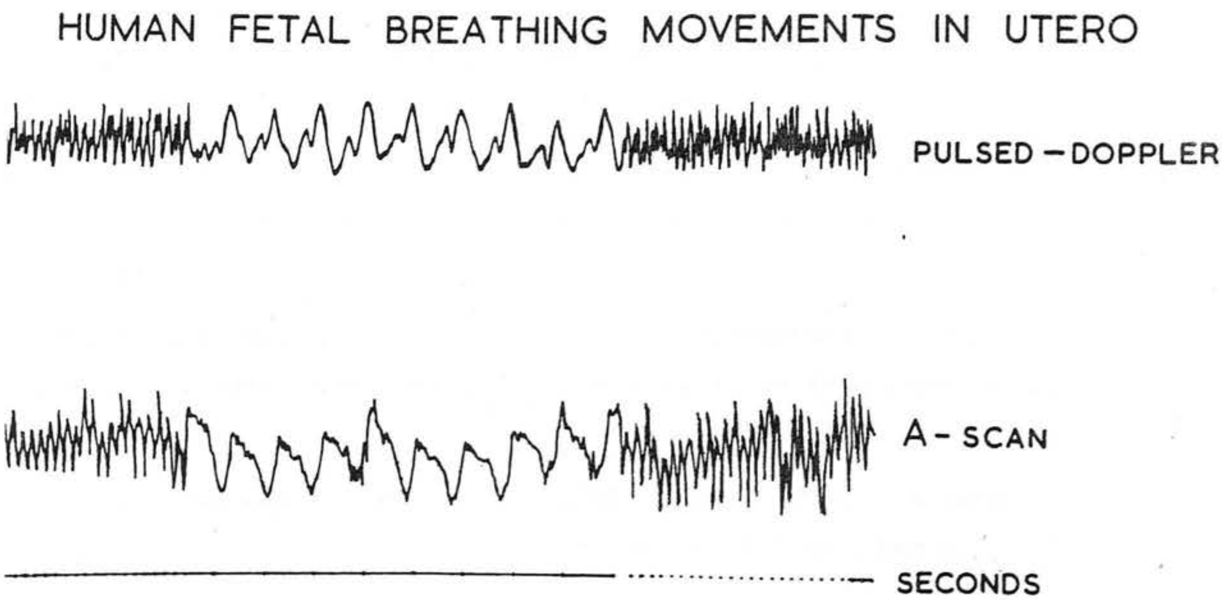
To establish the validity of this pulsed wave Doppler technique the pulsed wave Doppler and an A-scan instrument with gated T-M trace facility were linked together to run simultaneously using the same 2.5 MHz probe. The p.r.f. was 680 Hz. Figure A2.1 shows a section of T-M trace and pulsed wave Doppler recording from an adult heart with the range gate located on the posterior chest wall. It can be seen that there are two peaks of the pulsed wave Doppler trace for every one peak of the T-M trace. This is a result of tissue movement towards the transducer and away from the transducer. Figure A2.2 shows a section of T-M trace and pulsed wave Doppler recording of foetal breathing activity. The T-M trace shows the change of position of anterior chest wall with time. The pulsed wave Doppler range gate was set just inside the anterior chest wall. As with the adult heart recording, there are two peaks of pulsed Doppler trace to each of the T-M trace.

The results of this trial indicated that human foetal breathing movements in utero can be recorded using a pulsed wave Doppler instrument.



A section of an A-scan plus T-M trace and a pulsed wave Doppler recording of posterior chest wall movement from an adult heart. The time base has been expanded at the centre of the trace.

Figure A2.1



A section of an A-scan plus T-M trace and a pulsed wave Doppler recording of foetal breathing activity.
The time base has been expanded at the centre of the trace.

Figure A2.2

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