CONTINUOUS MONITORING OF BLOOD FLOW

BY
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Abstract: An extensive review of the literature revealed that there are still significant weakness in the available technology for blood flow measurement. This dissertation describes two techniques for blood velocity measurement. The first is an invasive method which uses multimodal optical fibres for light transmission to and from a sensing tip, which attenuates the light depending upon the blood velocity. The design and construction of this flowmeter is presented and bench results shown. The modulated light is transmitted to the detection and processing circuit and provision is made for the transducer to be insensitive to pressure fluctuations and ambient light. The second technique, which is noninvasive, uses a continuous wave Doppler ultrasonic technique; the instrument designed is a portable directional Doppler velocimeter with purpose-built probes intended for monitoring blood flow in femorodistal bypass grafts in ambulatory patients. This portable unit differs from conventional Doppler units in many respect which are described. This unit has been developed in order to understand the behaviour of blood flow in grafts while the patients are persuing everyday tasks. A postoperative study of successful in situ vein grafts from 8 patients has been undertaken to determine the feasibility of the technique. This pilot study shows that posture can have an effect on blood flow in grafts, and also shows that it is possible to monitor blood velocity with Doppler techniques for a long period of time, without intervention of an operator.
This dissertation is dedicated to my family and my best friends.
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Continuous Monitoring of blood flow
by
Naim Dahnoun

Statement of Originality
A thesis submitted in fulfilment of the requirements for the
degree of Doctor of Philosophy in the Department of
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recorded in this thesis is original unless otherwise
acknowledged in the text or by references. No part of this
thesis has been submitted for another degree in this or any
other University.

Naim DAHNOUN
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CHAPTER 1

CURRENT TECHNIQUES FOR THE DETECTION, DIAGNOSIS AND TREATMENT OF ARTERIAL DISEASE

1. 1 INTRODUCTION

Blood vessel disease results in the reduction of the inner diameter and compliance of the arteries. This will have an effect on the haemodynamics so as to increase pressure drops, increase local velocity and reduce flow. In the case of advanced disease the reduction of flow can lead to ischaemia or infarction. To treat these diseases an efficient technique for assessing their severity is required.

This chapter gives an overview of the nature of the disease and the current techniques for the treatment, detection, and evaluation of arterial disease.
1. 2 PERIPHERAL ARTERIAL DISEASE

Peripheral arterial insufficiency can be caused by different factors such as arteriosclerosis obliterans, thrombo-angitis obliterans, arterial thrombosis and embolism, traumatic arterial occlusion and Raynauld's disease. The most frequent disease within the western world is atherosclerosis which normally affects the large and medium-sized arteries such as the aorta, iliac, cerebral, coronary vessels and arteries of the lower extremities. Some 95% of the arterial occlusions that occur in patients over 50 years are caused by atherosclerosis (Hansteen et al, 1974). Atherosclerosis is a generalized disease and is a local symptom, the disease is assumed to be relatively benign (Tillgren et al, 1967). Epidemiological surveys divide the risk factors into major factors and minor factors. The major factors are hypertension, hypercholesterolemia and cigarette smoking and the minor factors are obesity, diabetes mellitus, hypertriglyceridemia, sedentary living, stress, and family history (James, 1984).

1. 2. 1 PATHOGENESIS OF ATHEROSCLEROSIS

There are two principal theories to explain the mechanisms involved, which are the imbibition and incrustation theories.

A. THE IMBIBITION THEORY

Lipid from plasma passes into the arterial wall, possibly at sites of endothelial damage (injured by mechanical damage such as from balloon angioplasty, immunological mechanism, viruses, certain products of tobacco, or haemodynamic forces). The exact mechanism of penetration of lipids into the arterial wall is not precisely known (Stender, 1982). The lipid is then deposited in the intima where it induces smooth muscle proliferation.

B. THE INCRUSTATION OR THROMBOGENIC THEORY

This theory suggests that at the sites of endothelial damage, small thrombi of platelets, leucocytes and fibrin form, which become covered by newly formed endothelium. Organisation (replacement by fibrous tissue) of the thrombi leads to the
formation of plaque. The lipid found in the plaque was presumed to be derived from the degeneration of constituents of the thrombi. Further damage to the newly formed endothelium leads to more thrombi formation and the process is again repeated.

1. 3 TREATMENTS FOR PERIPHERAL ARTERIAL DISEASE
Although different treatments exist as shown below, they are not without shortcomings. Simple measures such as stopping smoking and exercise programmes are valuable for less severe disease, and these patients may not need further treatment.

1. 3. 1 DRUG THERAPY
Since atherosclerosis is a complex disease which is caused by the interaction of many factors, a wide range of drugs with different actions are commercially available, and all are supposed to interact with a specific factor(s) and reverse the process or at least improve the disease. Different actions with corresponding drug(s) have been found, however it is unfortunate that none of these drugs have been proven to reduce the progress of the disease (Boobis et al, 1984); at present the only alternative left is the prevention of the disease (Kramsch et al, 1981). Antiplatelet agents however have been found to be more effective in prevention or reduction of proliferation. The effect of antiplatelet drugs is to prevent adhesion of platelets to the endothelium, which can cause damage. These drugs are currently administered to patients with prosthetic arterial grafts.

1. 3. 2 SURGERY
Direct reconstruction is not without risks, indeed the rate of failure is high, and even difficult to assess with any accuracy because the criteria are not well defined. The probability of success is dependent on many factors including the state of the arterial tree before surgery, speed of progress of the disease, age, presence of diabetes and the surgical procedure used, such as in-situ and reverse veins or prosthetic grafts. Patients with claudication alone have a low risks of limb loss, however patients with severe
claudication and rest pain, are at high risk of amputation. Intensive work is being carried out to find an objective patient selection procedure to improve the rate of graft success.

1. 3. 3 PERCUTANEOUS TRANSLUMINAL ANGIOPLASTY (PTA)

In 1964 Dotter treated stenosis by dilatation of the arteries (Dotter et al, 1964, 65, 68). The technique was improved by the introduction of a double lumen catheter (Gruentzig et al., 1974) and by the improvement of the material of the balloon which allowed higher inflation pressures and different catheter shapes to facilitate catheter manipulations (Levin et al., 1982). PTA is highly dependent upon precise angiographic location of the site of obstruction or blockage and also the balloon. The smaller the size of the catheter the better the results. Doppler ankle pressures help to select patients in which PTA is likely to give good results and also can be used as a means of follow-up (Johnston et al., 1982). When a patient is a candidate for PTA an arteriogram is necessary. The mechanism of action of PTA is not completely understood, this is probably because more than one mechanism is involved (Wolf et al., 1984); and different explanations are given:

(a) The atherosclerotic plaques are compressed against the outer wall of the artery by the pressure exerted by the balloon (Dotter et al., 1964).

(b) The cracking and tearing of the plaque results in the enlargement of the lumen, and explains the small dissections seen by angiography after performing a PTA. Controversially Isner (1983) showed plaque rupture and intimal tearing in patients with coronary atherosclerosis who had never undergone PTA. This, however does not exclude the latter mechanism because the cracking and tearing can exist in both patients who have and who have not undergone PTA since the causes can be different.

PTA has become an effective method of restoring blood flow and relieving symptoms. It has the advantages of reducing pain, morbidity and cost compared with surgery. However its
efficacy is not the same for different parts of the circulation; for instance the efficacy of PTA of the iliac artery is comparable with that of surgery but this is not so for the femoral artery.

While angioplasty is relatively simple to perform, its long term results are not very promising and the only alternative to relieve symptoms in patients with rest pain or critical ischaemic limb is to perform a surgical operation.

1.3.4 LUMBAR SYMPATHECTOMY

Blood flow changes, not only in response to cardiac function, proximal occlusive disease and arteriosclerotic changes in the vessel wall, but it also changes in response to alteration of vasomotor tone. Sympathectomy provokes vascular motor denervation and spontaneously modifies the vasomotor equilibrium. Historical reviews which are beyond the scope of this study can be found in Rutherford, 1984; and Imparto, 1979. It was first performed on patients with thromboangiitis obliterans (Diez, 1925), and later for Raynauld's disease of the lower limb (Adson et al, 1925). Finally the technique was performed on patients with arteriosclerosis obliterans. The technique was found to be more successful for the lower than the upper limbs. It can be performed either by surgery or by injection (phenol lumbar sympathectomy). The effectiveness of both procedures was found to be similar (Ramos et al 1983). Although the blood flow in the arteries and in the skin were measured and found to be increased (Collins, 1978), there is still no clear evidence that all patients will benefit from it. It is very difficult to evaluate the technique in the general case, however when the operation is not possible, the only alternative left is to perform a sympathectomy. More than 50 % of those patients have relief of rest pain, healing of ischaemic ulcers and improved ability to walk (Haimovici et al 1964, Show et al 1964, Szilagyi et al 1967, Kim et al 1976, Shanic et al 1976). The most controversial question is whether sympathectomy will improve intermittent claudication (Mackenzie et al 1962, Strandness et al 1965). Sympathectomy
was also considered as adjunction to vascular surgery (Shanic, 1976). This method is associated with many complications such as loss of temperature control, postsympathectomy neuralgia, and derangement of sexual function in male, depending on the level where the sympathectomy is performed (Rutherford, 1984). Since the blood flow increases mainly to skin but not to muscle (Masuokas, 1978), the technique is used in the case of skin change and as the last resort if surgical reconstruction is not possible.

1.4 METHODS OF DETECTING AND QUANTIFYING PERIPHERAL ARTERIAL DISEASE
The state of the arteries should be studied and the ideal method would be to map the vascular bed under investigation by giving its exact geometrical representation, and the pressure and flow distribution at each point inside the lumen of the artery.

1.4.1 IMAGING
The major roles of imaging techniques are diagnosis, screening, quantification and tissue characterization if possible. The most commonly used methods for mapping the vascular bed are described below.

A. ARTERIOGRAPHY
This technique is the most commonly used, and it is simple in principle. It consists of injection of a radio-opaque dye into the blood stream and subsequently taking X-ray photographs. This technique introduces some risks because the artery is punctured, a foreign body is injected into the blood stream, the artery and surrounding tissues are exposed to X-ray radiation, and there is a possibility of catheter breakage. These factors can cause haemorrhage, local or distal thrombosis (with potential limb loss or renal or cerebral damage), haematoma, extravasation of dye, allergic reaction, embolization, vascular mural disruption, or death.

A mono-planar picture can be misleading and consequently a stenosis or narrowing can be underestimated. This technique
can also miss a patent vessel which the contrast medium could not reach because of an occlusion. With the introduction of intra-arterial digital subtraction arteriography (DSA), a considerable improvement has occurred in the quality of the pictures. DSA consists of taking a plain image before the injection of contrast material and subtracting this from the contrast films in order to obtain an isolated image of the dye-filled arterial bed.

Arteriography and DSA are considered as the "gold standard" for visualization of the arterial bed, however they are invasive and the effects of X-ray radiation are cumulative and present some risks. It is necessary to justify carefully the use of this technique and therefore patient selection is indispensable before proceeding to an arteriogram. The technique has the advantage that it can be used in any part of the vascular tree and it can also be performed during surgery. However the main limitation is a lack of functional information.

B. ULTRASOUND IMAGING

The vascular bed can be visualized by means of ultrasound (Howry et al, 1952). Ultrasonic techniques have proved themselves to be extremely valuable in diagnostic medicine. This can be seen from the extensive scientific literature (Wells, 1969 and 1977. McDicken, 1981; Atkinson et al 1982; Evans el al 1989a) and the development of a wide diversity of equipment including many different transducers, real time and duplex scanners, available in most vascular laboratories, and sophisticated electronics and monitoring techniques.

The objective of pulse echo imaging is to produce, as accurately as possible, the geometry of the vessel wall. This is possible because of the difference between the acoustic impedance of two media which forms an interface. Tissue structure and tissue movements can be visualized. Blood flow can be measured and quantified by the Doppler ultrasound technique. A few tenths of a millimetre in dimension can be resolved.
B. 1 DOPPLER ULTRASOUND IMAGING
This method is based on the detection of the intensity or frequency of the Doppler shift signal obtained by a transducer which has a narrow sensitive area. These techniques can be divided into two groups, continuous or pulsed wave imaging techniques, however pulsed wave systems are more attractive since they have the ability to provide 2 dimensional pictures in an arbitrary plan.

B. 2 DOPPLER COLOUR FLOW IMAGING
Doppler colour flow imaging is a relatively new method for the noninvasive imaging of blood flow. It displays flow information in a two-dimensional colour picture. In general this technique assigns red colour for flow toward the transducer and blue colour for flow away from the transducer. Not only the direction is colour coded as above, but also the velocity information is colour coded. As the velocity increases the hue of red or blue changes (the faster the velocity, the brighter the hue and the slower the velocity, the darker the hue)
Doppler colour flow imaging technique is based upon pulsed Doppler technique with a multi-gated system. So aliasing problems are still present and high velocity information cannot be accurately determined. Although colour flow imaging is not completely quantitative, the time required to learn the technique and the time required for scanning are shorter than for the continuous or pulsed wave techniques. Doppler colour flow imaging has made a great impact in different areas of medicine including cardiovascular, vascular studies, oncology and obstetrics. Although this technique adds spatial information to the picture and saves time, it lacks high velocity information and losses temporal resolution.

B. 3 ULTRASOUND PULSE ECHO IMAGING
This type of imaging technique is based on the detection of echo amplitude. The echo signals detected by the transducer are first subjected to amplification then to time dependent gain to compensate for the attenuation of the pulse as it
travels deeper into the tissue. The compensated signal is
demodulated in order to extract the magnitude of the echo
signal (amplitude demodulation), and finally the signal is
amplified (McDicken, 1981).

The Doppler imaging technique is of great value for long-term
follow-up programmes because it is noninvasive, it has no
cumulative effects and it provides anatomical and functional
information which are of considerable potential diagnostic
value.

C. RADIOISOTOPE TECHNIQUE

Nuclear medical imaging studies involve the intravenous
injection of a bolus of a pharmaceutical labelled with a
radionuclide that emits high energy photons (gamma rays). The
patient is positioned under the camera with the area of
interest centred in the field of view and the
radiopharmaceutical distribution in this area is then
visualized, Fig. 1.1.

When gamma rays leave the patient, they are received through
a collimator which rejects gamma rays coming from non-perpendicular angles. Each hole in the collimator corresponds
to a point on the patient. The beamwidths of the collimator
holes are about 1 to 2 mm in diameter, therefore the
resolution will not be better than this. The collimator is in
contact with a single crystal [NaI(Tl)] (scintillator) which
converts most of the gamma rays to light photons. These are
detected by photomultiplier tubes (PM) which produce a pulse
proportional to the photon rate (light intensity). Since the
unwanted scattered-rays can be received by the crystal, they
may produce visible photons which will be detected by the
photomultiplier, thereby producing a signal. In order to
avoid this, the pulses from all PM are summed, producing an
energy pulse which may be compared using pulse height
analyzer. If the pulse falls within the set energy window a Z
pulse is sent to the computer to validate the data, Fig. 1.1.

The time taken for a valid Z to occur and display the results
will be a few microseconds, and to process images will takes
a few seconds.
At present the most sophisticated gamma-cameras have an anatomical resolution of 5 to 10 mm which is poorer than the conventional contrast arteriography resolution. The method is not strictly non-invasive since an intravenous injection is required but the technique offers certain advantages over invasive methods: it can be repeated frequently; the radiation dose to the patient is much lower than in contrast angiography; and it is better tolerated by patients.

By comparing the activity per unit area of lesion with that of the neighbouring area it is possible to predict whether a lesion is likely to heal (Segel, 1975). This technique has also been used to evaluate the patency of a bypass graft, a comparison has been made with conventional radiography and a 90% correlation has been found (Moss, 1976).

Fig. 1.1 Block diagram illustrating a gamma-camara system.
D. NUCLEAR MAGNETIC RESONANCE

Nuclear magnetic resonance (NMR) in medicine is relatively new. The phenomenon was first discovered in 1946 by the Nobel laureates Bloch and Purcell. The technique was originally used for non-destructive analysis of small samples. It took more than 20 years to extract image information from NMR signals. Nowadays the technique is proliferating because of the high image resolution obtained, and accessibility to parts of the body were other methods have failed.

The NMR technique is based on the principle that a proton with a magnetic moment \( \mu \) will experience a torque which tends to align it with the magnetic field, \( B_0 \), in which it is immersed, see Fig. 1.2.

\[ \mu = \gamma \hbar I \] 

Fig. 1.2 (a) random position of magnetic moment; (b) the magnetic moment is aligned in the direction of external magnetic field \( B_0 \).

Nuclei within the sample under analyses are randomly oriented initially so that the net magnetic moment from all nuclei is nil. For a nucleus that possesses a magnetic moment, \( \mu \), and an angular momentum, \( \hbar I \), the two quantities are interrelated as follow:

\[ \mu = \gamma \hbar I \] 

(1-1)
Where $\Gamma$ is the magnetogyric ratio (which is constant for a particular nucleus), $h$ is the Plank's constant, and $I$ is the nuclear angular momentum measured in units of $h$.

If a second magnetic field, $B_2$, with angular frequency, $W_q$, is applied to the system (see Fig. 1.3), this will produce a precession or resonance, and $w_0$ is called the Larmor frequency determined from the following equation:

$$h w_0 = 2 \mu B_0$$  \hspace{1cm} (1-2)

For the hydrogen atom which is a constituent element of all body matter, $I = \frac{1}{2}$ and equation, 1-2 becomes:

$$w_0 = \Gamma B_0$$  \hspace{1cm} (1-3)

Since $\Gamma$ is a constant, then for each applied magnetic field, $B_0$, there will be only one frequency $w_0$ which causes resonance. Equation 1-3 is the basic equation for NMR and the frequency of oscillation ($2\pi/w_0$) is in the range of radio-frequency ($\approx$ MHz).

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{fig1.3.png}
\caption{The magnetic moment $\mu_y$ oscillating at the Larmor frequency.}
\end{figure}

Prior to an NMR experiment, $\mu$ starts in the direction of $B_0$, i.e. in the vertical axis (its projection on the $z$-axis is a
maximum and on the y-axis is nil). The system is then excited by another magnetic field $B_2$ at the Larmor precession frequency. The angle, $\alpha$ between the z-axis and the magnetic moment gradually increases until $\alpha = 90^\circ$ or $180^\circ$ depending on which method is used. When $B_2$ is turned off, the magnetic moment precesses back to its original equilibrium position at the Larmor frequency and this gives the NMR signal. This signal which is characterized by its frequency, amplitude, phase, and duration of the oscillation, gives information on the density of the protons in the sample. The interaction between different magnetic moments gives rise to a local magnetic field, $B_0 + \omega B$ which causes the local Larmor frequency to change. In this case the magnetic moments are no longer in phase and the phase error increases with the time until the net magnetic moment in x-y plane is zero. The decay of $\mu_y$ is an exponential, with its time constant referred to as the $T_2$. $\mu_z$ also returns to equilibrium in an exponential manner, but with a decay constant referred to as the $T_1$ relaxation time which is longer than $T_2$. (see Fig. 1.4).

![Diagram](image_url)

Fig. 1.4 relationship between $T_1$ and $T_2$. 

\[ \mu_y \propto \exp \left( -\frac{t}{T_2} \right) \]

\[ \mu_z \propto \exp \left( \frac{t}{T_1} \right) \]
In practice, the pulse sequence is repeated before $\mu_Z$ returns to its maximum so that the magnitude of the NMR signal also contains information about $T_1$ and $T_2$. This provides information on the chemical and physical environment since this affects the ease with which nuclei can change their alignments.

In 1973 Lauterbur proposed the field gradient technique to discriminate NMR signals according to their position; this was the start of NMR imaging (MRI). The idea was to superimpose on $B_0$ a gradient field $B_x$, depending on the x-coordinate as shown below (Fig. 1.5). Therefore at each position, $x$, there will be a different resonance frequency $\omega_{0x} = \Gamma(B_0 + B_x)$

![Fig. 1.5 distribution of the field gradient in the x-axis.](image)

The same method is applied for 2 or 3 dimensions scanning by using other field gradients $(B_0 + B_y)$ and $(B_0 + B_z)$. 
E. COMPARISON OF TECHNIQUES

<table>
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<tr>
<th>Method</th>
<th>Invasive</th>
<th>Risk</th>
<th>Relative cost</th>
<th>Time taken for display</th>
<th>Anatomical information</th>
<th>Functional information</th>
<th>Resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arteriography</td>
<td>Yes</td>
<td>High</td>
<td>High</td>
<td>Off line</td>
<td>Yes</td>
<td>No</td>
<td>High</td>
</tr>
<tr>
<td>Doppler Ultrasound</td>
<td>No</td>
<td>No side-effects reported</td>
<td>Very low</td>
<td>Real time</td>
<td>Yes</td>
<td>Yes</td>
<td>1.5 - 0.1 mm</td>
</tr>
<tr>
<td>Radioisotope</td>
<td>Slightly</td>
<td>Low</td>
<td>High</td>
<td>Real time</td>
<td>No</td>
<td>Yes</td>
<td>5 - 1 cm</td>
</tr>
<tr>
<td>MRI</td>
<td>No</td>
<td>No side-effects reported</td>
<td>Extremely high</td>
<td>Real Time</td>
<td>Yes</td>
<td>Yes</td>
<td>*</td>
</tr>
</tbody>
</table>

1. 4. 2 FLOW AND PRESSURE MEASUREMENTS

A. FLOW AND FLOW VELOCITY MEASUREMENTS

There is evidence that the presence of disease in the arterial system can cause fluid-mechanical disorders, including reduction of pressure and volumetric flow rate, and an increase in local flow velocity. For instance the presence of a stenosis in an artery results in the local acceleration of blood flow, and a pressure drop across it, with the flow downstream becoming turbulent (appendix 1). Measurements of flow can thus be a means of assessing arterial disease or can confirm that the symptoms a patient is suffering from are indeed due to a vascular disorder. It is important at this stage to differentiate between flow velocity, which is the velocity of a unit volume and flow which refers to the volume displaced through some arbitrary plane per unit of time.

Flow measurement at rest has not been very successful in assessing or locating arterial disease (Sumner et al, 1969; Alpert et al, 1968). One of the major reasons for this was the lack of an accurate technique for measuring flow directly and noninvasively. However since flow velocity can be measured accurately and noninvasively by means of ultrasound (Ch2 and Ch 3), the haemodynamic effects produced by arterial

* The resolution depends on the signal-to-noise ratio and on the sample size.
disease can be recorded and analyzed. The measurement of peak forward and reverse velocities, mean velocity, acceleration and deceleration can be correlated to diseased arteries and the severity of the disease (Dilley et al, 1979). Different techniques of Doppler waveform analyses are presented in Ch. 5.

B. INDIRECT PRESSURE MEASUREMENTS

Ischaemic rest pain and intermittent claudication are the results of arterial insufficiency. Arterial disease causes a loss of compliance and narrowing of the peripheral arteries, which causes an increase in velocity at the narrowing or stenosis. As a result of this increase in velocity, the kinetic energy of the flow is increased at the stenosis, and subsequently subtracted from the pressure energy and finally the pressure drops across the stenosis, this is known as the Bernouilli effect (see appendix 1).

Distal to the stenosis the pressure increases but does not reach the value it had before entering it, this will result in a pressure gradient opposing the flow (appendix 1) and consequently there is a possible formation of turbulence which again results in the loss of energy which is subtracted from the pressure energy. In conclusion a narrowing or stenosis causes a drop of pressure across and distal to it with the possible introduction of disturbance.

Since the brachial and femoral systolic pressures are approximately the same in a normal subject (Hocken, 1967) and since the brachial vessels are less affected by disease than the lower limb vessels, then a comparison of the brachial and femoral pressures often gives an indication of disease in the lower limb (Yao, 1969; Needham et al, 1972). The Pressure Index (PI) (which is the Ankle / brachial systolic pressure ratio) can be used to quantify the disease. A PI>1 suggests a healthy limb and PI<1 suggests a diseased limb. Some workers (Carter, 1985; Gosling, 1974) tried to correlate the value of the PI to the 'number of sites' of arterial diseases. For instance a PI>0.5 corresponds to a single diseased segment and PI<0.5 correspond to a multisegmental disease, However
this is not always the case because the PI is also related to severity of the disease.

Resting blood flow to the limb can be normal even in the presence of severe arterial disease, however during moderate exercise the flow increases; and in the presence of stenosis the pressure will drop significantly (appendix 1) and therefore mild or asymptomatic disease can be detected (Sumner et al, 1969; Carter, 1972).

The pressure measurement technique consists of obstructing the vessel by applying a sphygmomanometer cuff around the limb and inflating it above the systolic blood pressure. Afterwards the pressure cuff is released gradually and when the pressure in the artery is slightly greater than the pressure in the cuff the blood flow returns. For many years the return of blood flow was detected by stethoscope, but in peripheral arteries this is an insensitive method, especially when the arteries are diseased. In 1950 Windsor, by using the pneumoplethysmographic technique, was able to measure systolic blood pressure (SBP) in the lower limbs and correlate the pressure gradient to the peripheral arterial disease. Later Strandness and Bell (1965) used a Mercury Gauge plethysmograph instead of a stethoscope to measure the SBP for the evaluation of peripheral vascular disease; the technique was considerably improved by the use of ultrasonic velocimeter since the latter is very sensitive to blood flow (Yao et al, 1969; Allan et al, 1969; Carter, 1969). These pressure measurements indicated the presence of arterial disease.

In order to locate the site of disease in the limb, segmental limb blood pressure has been attempted (Raines, 1984; Marsha et al, 1988). Different pneumatic cuffs are placed at high-thigh (HT), above-knee (AK), below-knee (BN) and ankle (A) levels. An abnormal pressure gradient between AK and BK results from distal superficial femoral (Marsha et al, 1988, Osmundson et al, 1985) or popliteal disease or both (Osmundson et al, 1985), and an abnormal gradient between BK
and A reflects tibioperoneal disease (Marsha et al, 1988; Osmundson et al, 1985). The normal pressure gradient between two adjacent cuffs is less than 20 mmHg. In practice multisegmental pressures are not used, because the occlusive cuff inevitably alters the flow, and can make the pressure measurements meaningless in terms of measuring disease severity (Evans et al 1980).

In conclusion, the standard exercise procedure for assessment of peripheral arterial disease in patients with rest pain or intermittent claudication is non-invasive, easy to use, reproducible (Osmundson et al 1985) and sensitive enough to reveal arterial disease. Not all patients however may be able to exercise, they may suffer from angina, or exercise itself may produce other complications.

C. DIRECT PRESSURE MEASUREMENT

Harvey, the founder of physiological measurements of the cardiovascular system, measured the blood pressure of a horse by inserting a tube in an artery and observing the blood level in the tube. Pressure and blood-flow waveforms in the vascular system are complicated because of the nature of the pulsating flow (McDonald, 1974), the opening and closing of valves, the geometry of the vessels varying with time throughout the cardiac cycle, branches coming out of the arteries and pressure wave reflection. Because of the difficulty of interpreting the shape of the pressure waveform, clinicians are forced to rely on the mean or systolic pressure value. In any fluid flow there are basically three causes of pressure drop in addition to the weight effect which we assume to be negligible in humans. These are changes in kinetic energy, energy necessary to accelerate the fluid, and the viscous resistance. These are related by Bernoulli's and Poisseuille's equations (Appendix 1).

Pressure measurements can be used to quantify arterial disease in cases were the arteriography method has failed. The pressure drop across mild stenoses or narrowing can be
enhanced by the injection of a vasodilator such as papavarine (Quin et al 1975), distal to the segment under investigation. The pressure drop increases as a result of the increase in flow caused by the vasodilator. In order to standardize the test, one intra-arterial pressure measurement is taken distal to the stenosis and another pressure is taken from the radial artery.

D. RESISTANCE MEASUREMENTS
Pressure measurements are more often performed simultaneously with flow measurements in order to derive the resistance which is the ratio of the mean (or systolic) pressure to the mean (or systolic) flow (Mundth, 1969; Weale, 1969; Mills et al 1970; Bliss, 1971; Albrechtsen, 1976; Baker et al, 1987; Parvin et al, 1985) or the magnitude of the impedance as described below. Resistance varies considerably from patient to patient but will increase considerably if arterial blockages are severe.

E. HYDRAULIC IMPEDANCE
Hydraulic impedance (HI) is a complex function, its amplitude is defined as the magnitude of the ratio of pressure to flow at a given frequency and its phase the difference between the pressure-phase and the flow-phase. Its measurement has attracted many workers interested in the understanding the behaviour of the arterial system (Randall et al 1956; Gabe et al 1964; Patel et al 1965; O'Rourke et al, 1966 and 1967; Taylor, 1966). HI was first measured in the femoral artery of the dog (Randall et al, 1956), one year later Womersely (1957) introduced the concept of impedance for the study of flow and pressure. In 1976 Cave measured the HI and attempted to correlate the results with the outcome of the operation in order to predict failure or success of surgery. The measurements were performed in patients with femoropopliteal vein bypass grafts, but no correlation was found between successful or failed results. At about the same time Butler (1977) who made measurements of HI on patients undergoing reconstruction surgery of the aorto-iliac and femoropopliteal segment, concluded: "... a 'good' Impedance is
characterized by a modulus curve which falls rapidly and smoothly to a broad minimum and shows little variation between 3 and 14 Hz. The phase curve is negative at low frequencies and tends to become more positive with increasing frequency". In the failure group the phase was found to remain or became more negative. Although the number of cases studied was low, promising results were obtained. However they did not correlate for individuals but only for groups. In 1983 the technique was reevaluated (Law, 1983) by taking a higher number of patients, and performing the measurements in aorto-iliac, femoro-popliteal, axillo-femoral bypass and deep femoral angioplasty. The only 'successful' result was obtained with the aorto-iliac case, which are similar to Butler findings, however the criteria could still not be applied to individual cases. HI is an exciting research subject which may need an accurate means of measuring the pressure and flow to minimize the errors which may be the cause of the inconclusive results obtained so far. Further work is needed on modelling the circulatory system to improve the understanding of the technique (Law, 1983).

F. PULSE GENERATED RUN OFF (PGR)
In the case of severe stenosis or blockage in the lower limbs, one or all of the pulses, which may be detected by ultrasound velocimeter, may be absent. This does not necessarily mean that the vessel is blocked completely since only one small segment can obstruct the flow to all the rest of the artery. If a cuff is put around the limb below the blockage and used to generate pulses pneumatically, a Doppler signal may be generated, indicating a patent vessel with very low flow at rest. It is also possible to determine whether or not there are patent vessels in the foot. This information is of value in deciding whether surgical intervention is likely to be successful (Beard, 1987).

The PGR system is a noninvasive and purely qualitative method. It is an adjunct to arteriography in the case of severe ischaemia where it is possible that the contrast medium does not reach the distal vessels in sufficient
quantity so that judgment on the quality of the distal vessel is not possible.

1.5 CONCLUSION
This chapter has described arterial disease and how it can be diagnosed and monitored by imaging, pressure and flow measurements. These techniques need to be sensitive enough for the detection of mild disease if they are to prove that symptoms that the patient are suffering from are indeed due to vascular disorder, and also to follow the progress of the disease. They should also quantify the disease so that patient selection can be standardized. The main purpose of this work was to develop techniques for the continuous monitoring of peripheral vessels and grafts. Imaging techniques are very expensive and are not practical for continuous measurements. Flow measurements may be made noninvasively and are therefore likely to be more useful for diagnosis and research than pressure measurement.
CHAPTER 2
REVIEW OF BLOOD FLOWMETERS AND BLOOD VELOCIMETERS IN
BIOMEDICAL USE.

2. 1. INTRODUCTION
'It is a source of regret that the measurement of flow is so much more difficult than the measurement of pressure. This has led to an undue interest in the blood pressure manometer. Most organs, however, require flow rather than pressure...'

(Jarisch, 1928)

Since 1928 the number of techniques for blood flow measurement have considerably increased. Some important ones are reviewed in this chapter. However, the challenge to develop a technique for measuring blood flow non-invasively, instantaneously, and accurately in vessels such as coronary arteries and small vessels, without disturbing the system, still exists.

It is not the aim of this chapter to prove the importance of blood flow measurement. It is however important to note that it gives information on the adequacy of the vascular bed (specifically on a segment of the vascular bed) and it helps in diagnosing defects in blood vessels and in assessing the results of vascular reconstructive surgery.
2.2 ELECTROMAGNETIC FLOWMETER

The principle of the electromagnetic flowmeter is based on Michael Faraday's discovery, that a conductor moving with a velocity $v(t)$ through a magnetic field $B(t)$ generates, an electromagnetic force (emf) (or voltage), see fig. 2.1. In fact Faraday himself in 1832, tried to measure the flow of the river Thames at Waterloo bridge but it was not until 100 years later that Fabré (1932) designed the first successful electromagnetic flowmeter.

In theory the voltage generated by the movement of blood in a magnetic field is linearly dependant on the velocity of the moving blood. The equation relating the output voltage ($V$) to the mean velocity of blood ($v$), when the vector $B$ and the vector $v$ are perpendicular, is given by

$$V = B \cdot d \cdot v \quad (2-1)$$

$B$: magnetic flux density.
$d$: diameter of the vessel.

The magnetic flux can be generated by magnetic coils which can be energized in different ways, as described below.

A. DC EXCITATION

A constant continuous current can be used to magnetize the coils (Jochim, 1948). In this mode of excitation the electrodes tend to polarize and limit the sensitivity to continuous flow. In practice it is not convenient to use DC amplifiers because of the problem of voltage offset drift and also because the electrode potentials are very large compared to the flow signal level.

B. ALTERNATING EXCITATION

In order to overcome the problems cited above an alternating current is used to excite the coils. Drift and polarization of the electrodes are avoided, and it also considerably reduces the problem of interference introduced by other medical equipment, if an appropriate frequency of excitation is used. Different modes of alternating excitation exist;
sine, pulse and trapezoidal excitations have been used and all have their advantages and disadvantages. When an alternating excitation is used, the system (blood, electrodes and wires) act as a transformer and produce an unwanted signal known as the 'transformer signal'. Therefore equation 2-1 will be transformed to equation 2-2.

\[ V = B(t) \cdot d \cdot v(t) + k \frac{dB}{dt} \]  \hspace{1cm} (2-2)

Where \( k \) is a constant dependant on the measuring conditions and the probe itself. The factor \( k \frac{dB}{dt} \) is referred to as the transformer voltage.

Other factors can also be added in equation (2-2) in order to get a closer approximation to the real situation (Wyatt, 1984)

C. FACTORS AFFECTING THE SENSITIVITY

C. 1 EFFECT OF VELOCITY PROFILE

Velocity across an artery is not uniform, normally it is greater at the centre of the artery and lower close to the vessel wall. This generates circulating electric current across the vessel as shown in Fig. 2.2a (Shercliff, 1962). These circulating currents cause potential drops in the fluid. Whilst the velocity close to the vessel wall, is lower (a lower voltage is produced) the sensitivity at that point

![Electrode and Vessel Wall Diagram](image)

Fig. 2.1 . Three-dimensional view showing the principle of operation of the electromagnetic flowmeter.
is greater. To account for this effect a weighting function, $w$ (see equation 2-3) has been introduced (Shercliff, 1962; Bevir, 1970; Wyatt, 1972).

$$V = \int w \cdot v \, dr$$ \hspace{1cm} (2-3)

Where

$w$: Weight vector or Weight function

$dr$: Elemental volume

$v$: Velocity of the fluid in the element volume

$\tau$: Volume which contributes to the signal

$V$: Voltage given by the flowmeter.

In the case where the blood velocity is not uniform but rotational symmetry exists, equation (2-1) still holds (averaging the cross-section area). It can be seen from Fig. 2.2 which shows the distribution of the weight function, $w$, that in the case of a velocity profile which is not axi-symmetrical, the flow will in general be overestimated or underestimated depending on the velocity distribution within the weight function.

Fig. 2.2 (a) Induced circulating currents due to non-uniform flow distribution; (b) Distribution of the weight function, $w$. (from Shercliff, 1962).

**C. 2 OTHER EFFECTS**

It was also assumed that the magnetic field was uniform, the contact of the electrodes with the vessel wall was perfect.
and the conductivities of blood, vessel wall, and tissue fluid do not change during the cardiac cycle. In practice none of these assumptions are valid and calibration may be required.

In conclusion, although the electromagnetic flowmeter is invasive, expensive, difficult to use, needs to be calibrated before use, and a good contact with the vessel is required, it is still in use. Furthermore, it is also haematocrit sensitive (Roberts, 1969; Dennis, 1969), and flow-profile dependent when the flow is not axi-symmetrical. A very well designed and calibrated instrument can reach a frequency response of 50 Hz but it is very rare that a commercial flowmeter will reach this value. The phase response may also be frequency dependent.

2.3 ELECTRICAL IMPEDANCE TECHNIQUE
This technique is based on the electrical properties of blood and its surrounding tissues. Blood volume change manifests itself by an impedance change. In the case of a steady flow no impedance changes will occur, however blood in the vascular system is pulsatile in nature (McDonald, 1974) and changes in impedance always occur. This technique is commonly known as plethysmography (Nyboer, 1950). It is non-invasive and has been used for assessing disease in the lower limbs (Allison, 1967; Wheeler et al, 1970; Van De water et al, 1971; Person, 1979).

METHOD
A high-frequency signal (25-100kHz) with a constant current (less than 4 mA r.m.s) is applied between two outer electrodes placed around a selected part of the body under investigation. This part of the body contains a volume of blood during diastole which corresponds to a basal impedance $Z_1$. During systole, this impedance decreases and will have a value, say $Z_2$

In order to simplify the problem, consider the blood volume change in the lower limb (Fig. 2.3).
Fig. 2.3 Arrangement of the electrodes for measurement of the electrical resistance.

The volume \( V \) between the electrodes is

\[ V = Ls = L \pi \frac{d^2}{4} \]

or

\[ Z_1 = \rho L/S_1 \]

and

\[ Z_2 = \rho L/S_2 \]

\( \rho \) : Resistivity

L : Distance between the electrodes

d_1 : Diameter of the limb during diastole.

S_1 : Cross-section of the limb during diastole

S_2 : Cross-section of the limb during systole.

Z_1, Z_2 : Resistance between the electrodes E_1 and E_2, when the cross section area of the limb is respectively S_1 and S_2.

Assuming that the length (L) and the resistivity (\( \rho \)) remain constant, then

\[ S_1 = \frac{\rho L}{Z_1} \text{ and } S_2 = \frac{\rho L}{Z_2}. \]

Thus the change in the volume \( dV \) is:

\[ L(S_2 - S_1) = \rho L^2 \left( \frac{1}{Z_2} + \frac{1}{Z_1} \right) \]

or

\[ Z_1 \cdot Z_2 \approx Z_1^2 \]

and

\[ Z_1 - Z_2 = dZ. \]

\[ dV = \rho L^2 \frac{dZ}{Z_1^2} \]

or

\[ \rho = Z_1 S_1 / L \]

Thus

\[ dV = (Z_1 S_1 / L) \cdot (L^2 \frac{dZ}{Z_1^2}) = (S_1 L / Z_1) dZ \]

\[ = (\pi \frac{d_1^2}{4}) L \frac{dZ}{Z_1} \]

Since \( d_1 = d_m + \delta d \) and \( \delta d \ll d_m \) where \( d_m \) is the mean diameter.

Then \( d_1^2 \approx d_m^2 \)

Or the mean tissue volume \( (v_m) \) is

\[ v_m = \pi d_m^2 L / 4 \]
Or the mean tissue volume \((v_m)\) is
\[ v_m = \pi d_m^2 L/4 \]
Finally

\[ dv = v_m \frac{dZ}{Z_1} \quad (2-4) \]

So the variation of volume will fluctuate according to \(dZ\).
It is important to notice that \(dV\) is a volume change and not a volume flow. If the venous system should become occluded then the volume changes will be equal to the volume flow and this is known as occlusion plethysmography. Occlusion of the venous system may be achieved by placing a cuff around the limb proximal to the segment of the limb under study and inflating it to approximately 50 mmHg. This technique can also be applied to monitor stroke volume, it is non-invasive and yields quantitative results.

2.4 VOLUMETRIC PLETHYSMOGRAPHY
This technique is based on the fact that when venous occlusion is applied, arterial inflow causes an increase in limb volume. This volume change is then measured by a water or air-filled cuff that surrounds the segment of the limb of interest. When blood flow is to be measured it is necessary to occlude the distal artery. Since the change in volume is very small (less than or equal to 1%), it is very important to avoid leakage of the water or air from the cuff, and in the case of an air-filled cuff the temperature should be controlled because of its high thermal expansion coefficient. With air, the frequency response of the volumetric plethysmograph is much higher than when water is used, and the static pressure exerted on the limb by the water-filled cuff is avoided.

2.5 STRAIN-GAUGE PLETHYSMOGRAPHY
This technique consists of measuring changes in the circumference of the limb. This is achieved by encircling the limb with a small bore rubber tube filled with mercury, Fig. 2.4.
THEORY

The resistance $R$, between the electrodes of the rubber filled with mercury, is given by

$$R = \rho \frac{l}{s}$$

Where $\rho$, $s$ and $l$ are respectively the resistivity, cross-section area and the average length of the rubber filled with mercury.

The average length, $l$ and the volume, $v$ of the rubber bore are given by,

$$l = 2\pi r$$

and

$$v = sl = \text{constant}$$

Where $r$ is the radius as defined in fig. 2.4a

Thus $R = \rho \frac{l^2}{v} = k l^2$, where $k = \rho/v = \text{constant}$

Therefore $dR/R = 2 dl/l$

or the cross-section of the limb, $S$ is given by,

$$S = \frac{l^2}{4\pi}$$

Therefore $dS/S = 2 dl/l$

And $dS/S = dV/V$ (if there is no longitudinal variation).

Where $V$ is the volume of the segment of the limb.

Or $S = \pi r^2$ therefore $dS/S = 2dr/r$.

Finally,
The resistance, \( R \) of the mercury can be determined by Ohm's law since the current and the voltage are known.

In this application, the patient attachment is reduced. As we can see from above \( \frac{\delta V}{V} = \frac{\delta S}{S} \), the length of the limb is supposed invariable during the whole cardiac cycle, and the limb is supposed to be circular.

### 2.6 THERMAL METHOD

This technique consists of heating the fluid (blood) to a few degrees above body temperature and recording the amount of heat carried away by the fluid. The technique is extensively used in aeronautics to measure air-flow. In 1928 Rein measured flow in an intact vessel by heating a region of the vessel. The device was sensitive to the mean blood flow but was not suitable for pulsatile flow, and the frequency response was very poor. Most of the present thermal methods for measuring blood flow are designed to be inserted as catheters. The heating elements are kept at a constant temperature less than 5 degrees above body temperature by a servo-mechanism; the amount of current supplied to the heating element is proportional to the velocity of the fluid (Ling et al, 1966; Belhouse et al, 1967-1968; Schultz et al, 1969; Seed et al, 1970). The first flow measurement based on the above principle was introduced by Mellander in 1960, but its frequency response was very poor (less than 5 Hz). The frequency response was increased to 50 Hz by Kuether (1966), and further increased to 150 Hz by Grahn (1969). As a heating element, a wire was used; this was later replaced by a thin film of gold or platinum.

In spite of the good frequency response achieved, many factors such as linearity of response, sterilization, positioning of the probe, change of the blood temperature, and probe-vessel contact pressure, limit its use. It is important to note that the method is invasive and indirect.
2. 7 LASER DOPPLER FLOWMETER (LDF)
The laser device was originally proposed by Schawlow and Townes in 1958 and put into practice by Maiman in 1960. In 1964 Yeh and Cummins were the first to demonstrate the feasibility of optical velocimetry, using a helium-neon laser to measure the velocities of polystyrene spheres in a solution. Blood flow measurement using the laser Doppler technique was first attempted by Riva et al in 1972, when they studied blood flow in retinal vessels in rabbits. As with ultrasonic Doppler velocimeters, the LDFs work only when scattering particles are present.

METHOD
A narrow beam of monochromatic light generated by a laser, is guided by an optical fibre to the region of tissue to be studied. When the light is scattered from the tissue, a proportion of the back scattered light will have undergone multiple Doppler frequency shifts. Since the frequency of the light is very high, it is not possible even with the most sophisticated spectroscope to detect the very small shift in frequency. However to measure the Doppler shift frequency, the same technique used in telecommunication and ultrasound equipment for demodulation is applied.

The frequency shifted light is mixed with unshifted light scattered from static skin structures to produce the demodulated signal. In the case of ultrasound the signals mixed are electrical in nature, whereas in laser Doppler velocimetry the signals are composed of light radiation. The mixing is done simply by photodetectors which are square law devices since they are sensitive to the intensity of light which is the square of its amplitude.

The basic block diagram of the LDF is shown in Fig. 2.5 and it can be divided into four elements
- Light source (Laser).
- Light guide (Fibre optic).
2.8 DYE DILUTION TECHNIQUE.

2.8.1 FICK PRINCIPLE

Adolph Fick (1870) stated that the up-take (z) of a metabolically inert substance by tissue in a unit of time, is equal to the quantity brought by the arterial blood (y) minus...
that present in the venous blood (x). The principle is illustrated in Fig. 2.6. In fact this principle is a restatement of the law of mass conservation.

![Diagram of the Fick principle]

Fig. 2.6 Illustration of the Fick principle.

2.8.2 INDICATOR DILUTION

Only indicator dilution methods measure the cardiac output directly. These methods are based on the principle that an indicator is mixed with a unit volume of blood so that it can be distinguished from all other units. Hearing in 1829 was the first to use this method for determining the circulation time. Potassium ferrocyanide was injected intravenously in order to determine the circulation time by measuring the difference between the time (t₀) at which the indicator was injected and the time (t₁) at which the first indicator was detected. However there are many paths the blood can take, so (t₁ - t₀) represents the circulation time of the blood which has taken the quickest pathway.

2.8.3 INTRODUCTORY CONCEPTS

Dye dilution is based on Fick's principle, the dye concentration (c) is defined by the mass of indicator (m) divided by the volume (v) of the solution to which the indicator has been added.
ASSUMPTIONS
- It is assumed that the variation of volume due to the indicator introduced is negligible.
- It is supposed that the indicator and the solution are thoroughly mixed so that the particles of the indicator are homogeneously distributed.
- It is assumed that no loss of the indicator occurs.
Note that the indicator can take different forms including mass, heat and radiation. If the indicator is mass or heat or radiation, then the technique is known as Dye dilution or Thermodilution or Radiocardiography respectively.

2. 8. 4 PRINCIPLE OF THE METHOD
An indicator is injected into the right ventricle and is mixed with blood in the heart and the lungs. The mixture passes into the circulation. At a sampling point of the cardiovascular system, a catheter is placed to evaluate the concentration of the substance injected. It is also possible to sample blood by using a cuvette oxymetry to evaluate the concentration. Knowing the concentration at different times and knowing the quantity of indicator injected, the volume of blood per unit of time or the cardiac output can be determined as shown below (Fig. 2.7)
There are two methods of injecting an indicator.

![Fig. 2.7 Mixing chamber (schematic representation of the principle).]
2.8.5 CONSTANT RATE INJECTION

This method assumes that the blood flows at constant volumetric flow rate $Q(1/s)$ into and out of the heart, in which complete mixing occurs. To this flow is added an indicator at constant rate $M \text{ (g/s)}$. As a result, the concentration $c(\text{g/l})$ at the output increases from zero to a maximum concentration ($c_{\text{max}}$) see Fig. 2.8. When this occurs, the rate at which the indicator leaves the system must be equal to the rate at which it is introduced so that $c_{\text{m}} = M/Q$ (assuming no indicator loss).

If the fluid already contains a constant concentration ($c_{\text{o}}$) of the same indicator, the maximum concentration will be: $c_{\text{M}} = c_{\text{max}} + c_{\text{o}}$. (see Fig. 2.9).

By measuring the concentration $c_{\text{M}}$ and knowing the rate $M$ at which the indicator was injected, the flow $Q$ can be determined.
Fig. 2.9 Constant rate of injection; (a) Dye dilution curve, with an initial concentration $C_0$; (b) rate of injection curve.

2. 8. 6 SUDDEN INJECTION

In this case an amount ($N$) of the indicator is suddenly injected, the concentration at the output will increase from zero to maximum and will decline to a minimum after a few oscillations due to the recirculation of the dye (Fig. 2.10).

If we consider that the flow $Q(t)$ is constant during a short period of time ($d(t)$) the volume $dv$ displaced is

$$dv = Q \, dt$$

This small volume $dv$ will have a concentration $c(t)$, and will contain a fraction ($N_i$) of the total mass ($N$) injected, so

$$dv = Q \, dt = N_i/c(t)$$

Hence

$$N_i = Q \, c(t) \, dt$$
Fig. 2.10 Sudden rate of injection; (a) Dye dilution curves after sudden injection; (b) rate of injection curve.

\[
\sum_{i=0}^{n} N_i = \int_{0}^{t} Q(c(t)) \, dt
\]

\[
\sum_{i=0}^{n} N_i = M : \text{total mass of the dye injected.}
\]

Q is assumed to be constant during the time t.

And finally

\[
Q = \frac{M}{\int_{0}^{t} c(t) \, dt}. \quad (2.6)
\]
Different methods exist for the calculation of the area \( \int_{0}^{t} c(t) \, dt \) under the curve \( c(t) \), by considering the modulation due to the recirculation (Stow, 1954). One method is shown below.

### 2.8.7 HAMILTON REPLOT METHOD

Kinsman et al (1929) fitted an empirical formula for \( c(t) \) when it started to decline. They assumed that the curve \( c(t) \) declines in an exponential way, and by taking the logarithm of the function \( c(t) \), obtained a straight line. All points which did not fit this straight line were assumed to be due to recirculation and were rejected, (fig. 2.11). The theoretical explanation is as follows.

\[
\begin{align*}
    c(t) &= c_{\text{max}} e^{-kt} \\
    c(t_1) &= c_{\text{max}} e^{-k t_1} \\
    c(t_2) &= c_{\text{max}} e^{-k t_2} \\
    \frac{c(t_1)}{c(t_2)} &= e^{-(t_1 - t_2)k}.
\end{align*}
\]

Therefore \( \ln \frac{c(t_1)}{c(t_2)} = -k(t_1 - t_2) \)

And

\[
k = \frac{\ln \frac{c(t_1)}{c(t_2)}}{t_1 - t_2}
\]

![Fig. 2.11 (a) Dye dilution curve with oscillation due to the recirculation; (b) Replot of the dye dilution curve.](image)
Once k has been deduced using this method, the curve without the effect of recirculation can be plotted and the area evaluated.

2. 9 NUCLEAR MAGNETIC RESONANCE TECHNIQUE

It has been known for many years that blood flow influences the NMR signal, and this was considered as an artifact. In 1978, Singer exploited this 'artifact' and used the time-of-flight technique which is described below, to quantify the flow velocity.

The flow artifact obtained in NMR images was due to the fact that flow causes the excited proton to leave the signal detection area and be replaced by other non-excited protons. Two principal techniques are used for detection of velocity. These are known as time-of-flight and velocity-phase-map techniques.

2. 9. 1 TIME-OF-FLIGHT TECHNIQUE

This technique consists of exciting the protons at a specific point and detecting the NMR signal at a different point. Knowing times of excitation and detection, velocity can be determined (Singer 1959, 1978, 1981). This technique is limited to slow flowing blood and also it assumes that the velocity remains constant between the excitation and detection location. This technique has been modified for use with higher velocities (Singer et al 1983).

2. 9. 2 VELOCITY-PHASE-MAP TECHNIQUE

The extraction of velocity information from the phase of the NMR signal (Bryant et al 1984; Moran et al, 1984; Redpath et al 1984; Van Dijk, 1984; Wedeen et al 1985) is the most commonly used and is more versatile than the time-of-flight technique.

A. THEORETICAL CONSIDERATION

Consider that the precession is at the frequency, \( w_0 \), then the phase shift, \( \delta \phi \), after a certain time, \( \delta t \) will be,
\[ \delta \phi = \omega_0 \delta t = \Gamma B \delta t \]  
(see chapter 1)  
(2.7)

Where

\( \omega_0 \): Angular frequency

\( \Gamma \): Magnetogyric ratio

\( B \): Magnetic field applied.

In the \( x \) direction \( B \) may be chosen to be of the following form, \( B = x B_0 \), then the equation, 2.7 will be

\[ \delta \phi = \Gamma x B_0 \delta t \]  
(2.8)

If the protons are moving with a constant velocity \( (v) \), then their positions will be a function of time, \( x = x_0 + vt \), and equation 2.8 may be rewritten as

\[ \delta \phi = \Gamma (x_0 + vt) B_0 \delta t \]  
(2.9)

By integrating equation, 2.9 then

\[ \int \delta \phi = \int_{T1}^{T2} \Gamma (x_0 + vt) B_0 \delta t \]  
(2.10)

\[ \phi = \Gamma x_0 B_0 (T_2 - T_1) + 0.5 \Gamma v B_0 (T_2^2 - T_1^2) \]  
(2.11)

Where the term, \( (\Gamma B_0 (T_2^2 - T_1^2) v) \) is due to the velocity. From equation 2.11 it can be seen that the phase shift also depends on the stationary protons, this can cause a problem since the magnetic field is not uniform and also there is an additional minute magnetic field produced by the magnetic moments of the other surrounding protons, thus there will be a variation of the Larmor frequency at each point. Finally there will be a loss of phase coherence, which causes a weak NMR signal.

To reduce the phase error described above and to improve the signal-to-noise ratio, a spin echo technique was used, this consists of flipping the magnetic moments by 180° around the \( y \)-axis (see fig. 2.12). This manoeuvre reverses the relative position of the faster and slower magnetic moments. After a certain time the magnetic moments come into phase and produce an echo signal (see Fig. 2.11 and 2.12).
During phase echos which occur at time $\tau$ after the $180^\circ$ pulse is applied, the phases due to the stationary spins become nil, however the phase shift due to the moving spins will still be proportional to the velocity, so equation, 2-11, will be:

$$\phi = \Gamma x_0 B_0 v (T_2^2 - T_1^2) + \Gamma v (-B_0) (T_4^2 - T_3^2)$$

$$\phi = \Gamma x_0 B_0 (T_2^2 - T_3^2 - T_1^2 - T_4^2)v$$  \hspace{1cm} (2.12)

Fig. 2.12 Representation of the magnetic moment at different position.
Fig. 2.13 (a) pulse sequence (b) NMR signal.

B. PHASE DETERMINATION

By measuring the magnitudes of the vector $M$ projected onto the $y$ and $x$-axes (see Fig. 2.13), the modulus and the phase images can be reconstructed (this is normally done using a two-dimensional Fourier transform, 2D-FT).

\[ |M| = \sqrt{R^2 + I^2} \quad \text{: Modulus Image.} \quad (2.13) \]

\[ \phi = \arctan \frac{I}{R} \quad \text{: Phase Image.} \quad (2.14) \]
Finally by combining equation 2.11 and 2.14, the velocity equation is:

\[
V = \frac{2 \arctan I/R}{\Gamma x_0 B_0 (T_2^2 + T_3^2 - T_1^2 - T_4^2)}
\]

Blood flow measurement using the NMR technique is becoming well established; quantitative and in-vivo flow measurements have been achieved. In the case of turbulent flow, dephasing and signal loss occur, this is due to the averaging which occurs in each pixel and therefore bifurcations and turbulent locations should be avoided. This technique has the advantage of being noninvasive, not being restricted by the location or the orientation of the vessel, and can produce a three-dimensional vector velocity. This technique could be unrivalled for the study of intracranial vessels. Motion artifact is normally present with this technique because of the long scan times (up to 30 min), a recent development however (Firmin et al 1989) allows the acquisition of a vessel phase map in less than 80 ms (Smith, 1990).
3.1 INTRODUCTION
The requirements of the ideal blood velocity transducer which can be used as a catheter (i.e., high frequency response, sensitive, miniature, and safe) can be more or less satisfied by the use of optical fibres when the vessel is inaccessible non-invasively. Optical fibres have the advantage of being electrically isolating and can be made very small (few tens of microns in diameter). For these reasons a prototype transducer using optical fibres and associated electronics are described. The technique described in this chapter is based on the detection by optical means of the position of a membrane, which is affected by the drag force of the fluid flow.
3. 2 SYSTEM DESCRIPTION

The prototype instrument which has been developed consists of a sensor, an optical fibre lead, a signal detector and a processing unit, the block diagram being shown in Fig. 3.1. The sensor is composed of an elastic membrane which faces the flow on one side and an optical fibre on the other side. Both sides are submitted to the same static pressure, see Fig. 3.2. The principle of operation is based on the detection of the displacement of the membrane due to the drag force, $F$, resulting from the movement of the fluid, where

$$F = \frac{C \cdot A \cdot \sigma \cdot V^2}{2} \quad (3.1)$$

Where
- $C$ : Drag coefficient,
- $A$ : Cross-section area,
- $\sigma$ : Fluid mass density,
- $V$ : Fluid velocity.

Flow

Transducer

Optical waveguide

Detecting & processing unit

Display

Fig. 3.1 Optical velocimeter system.

The velocity $V$ can be determined from the force $F$ since the diameter and the fluid mass density are known and the drag coefficient can be determined by experiment. This force is applied to the circular diaphragm with a uniform distributed load (assuming laminar flow) and edges held. The deflection, $Y$, at a distance $r$ from the centre is given by equation 3.2 (Roark, 1954).
Fig. 3.2 Schematic representation of the sensor tip.

\[
Y \approx 0.662 \ a \left( \frac{w \cdot a}{E \cdot t} \right)^{1/3} (1 - 0.9 \frac{r^2}{a^2} - 0.1 \frac{r^5}{a^5}) \tag{3.2}
\]

Where

- \( w \): Pressure.
- \( t \): Thickness of the diaphragm.
- \( E \): Modulus of elasticity.
- \( r \): Distance from the centre of the circular diaphragm.
- \( a \): Radius of the diaphragm.

Or

\[
F = w \pi a^2. \tag{3.3}
\]

Therefore the combination of equations (3.2) and (3.3) gives

\[
Y \approx 0.662 \ a \left( \frac{F}{E \pi a t} \right)^{1/3} (1 - 0.9 \frac{r^2}{a^2} - 0.1 \frac{r^5}{a^5}) \tag{3.4}
\]

When equation 3.1 is substituted in equation (3.4) the relationship between the deflection, \( Y \), of the membrane and the velocity, \( V \), of the flow is obtained.

\[
Y \approx 0.662 \ a \left( \frac{C a \sigma V^2}{2 E t} \right)^{1/3} (1 - 0.9 \frac{r^2}{a^2} - 0.1 \frac{r^5}{a^5}) \tag{3.5}
\]
It can be seen from equation (3.6) that in order to achieve a high displacement, the membrane has to have a thickness and a modulus of elasticity as small as possible, and the radius $a$ as large as possible. However in practice these parameters are limited, for instance the radius of the membrane is constrained by the size of the artery and the need to be small in order not to disturb the flow. It is also important to position the optical fibre at the centre of the membrane since its deflection is maximal at that point (see equation 3.2 to 3.6).

3.3 DESIGN AND CONSTRUCTION

The technique of measuring the displacement of the membrane consists of sending the light through a fibre to the membrane and measuring the proportion of light reflected back. The block diagram is shown in Fig. 3.3. The light sent to the membrane is transmitted through fibre optics and is reflected back to the photodiode. The signal is then processed through a band pass filter, an r.m.s. converter, and an off-set to produce the final output.
target is amplitude modulated to overcome the effects of ambient light. The light was pulsed at 10 kHz this being well above mains power frequency.

The light reflected back from the membrane which modulates the light according to the drag force was guided to a photodiode where the light signal was converted to an electric signal. If the membrane is not severely distorted by the drag force then the amplitude of the received signal may be assumed to be approximately inversely proportional to the distance between the tip of the fibre and the membrane. The signal was amplified and filtered by a band-pass filter whose central frequency was set to the transmitting frequency. The signal was then full-wave rectified and the amplitude extracted.

3.3.1 TRANSMITTER AND RECEIVER
A high radiance GaAlAs L.E.D with a peak spectra output at 820 nm was chosen because it could be coupled directly to a fibre optic, its output provided an adequate power (400 µW), it had a fast rise time which allowed the light to be pulsed and was of low cost.

3.3.2 PHOTODETECTION AND SIGNAL PROCESSING
The conversion from light to the electrical signal was achieved using a pin photodiode receiver to match 660 - 830 nm emitters. In order to have a linear response, the load resistance seen by the photodiode (Fig. 3.4) is made to approach zero Ohms. In this case the photodiode is working into a virtual earth point, so that the output current is linear with illumination or radiation. A fast response amplifier is required to reduce the ringing of the photodiode at high frequency (> 10 kHz). The output signal (S) of the photodiode is described by the following equation.

\[ S = A_{dc} \cos \omega_{dc} t + A_n \cos \omega_n t + B \cos \omega_c t \]

Where:
$A_{dc}$ and $w_{dc}$: are the amplitude and angular frequency of the signal due to the natural light, 
$A_n$ and $w_n$ are the amplitude and the angular frequency of the signal due to the artificial light, and
$B$ and $w_c$ are the amplitude and angular frequency of the carrier signal.

Fig. 3.4 (a) voltage-current characteristic of the photodiode; (b) configuration of the photodiode-amplifier used and its equivalent circuit.

Since light from any source can only impinge on the photodiode through the tip of the fibre where the sensor modulates the light, then the output signal modulated will be
\[ S = A_{dc} (A_1 \cos wt)(\cos w_{dc}t) \]
\[ + A_n (A_2 \cos wt)(\cos w_{nt}t) \]
\[ + B (A \cos wt)(\cos w_{ct}t) \]
\[ = 0.5 \left( A_{dc} A_1 \cos(w +w_{dc})t + A_{dc} A_1 \cos(w -w_{dc})t \right) \]
\[ + A_n A_2 \cos(w +w_{nt})t + A_n A_2 \cos(w -w_{nt})t \]
\[ + A B \cos(w +w_{ct})t + A B \cos(w -w_{ct})t \] (3.7)

In practice we have
\[ 0 < w_{dc} < (2 \pi 50 \text{ Hz}) \]
and \[ (2 \pi 50 \text{ Hz}) < w_{nt} < (2 \pi 200 \text{ Hz}) \]
and \[ (0 < w < (2 \pi 100 \text{ Hz}) \]

The angular frequency, \( w \), was defined by the frequency response of the system to be designed, and the angular frequency \( w_{ct} \) was chosen to be \( 10 \cdot 2\pi \text{ kHz} \).

Therefore
\[ 0 \text{ Hz} < w +w_{dc} < 2 \pi 150 \text{ Hz} \]
\[ 0 \text{ Hz} < w -w_{dc} < 2 \pi 50 \text{ Hz} \]
\[ 2 \pi 50 \text{ Hz} < w +w_{nt} < 2 \pi 300 \text{ Hz} \]
\[ 2 \pi 50 \text{ Hz} < w -w_{nt} < 2 \pi 100 \text{ Hz} \]
\[ 2 \pi 10 \text{ kHz} < w +w_{ct} < 2 \pi (10\text{kHz} + 100 \text{ Hz}) \]
and \[ 2 \pi (10\text{kHz} - 100 \text{ Hz}) < w -w_{ct} < 2 \pi 10 \text{ kHz} \]

So after filtering, equation 3.7 becomes
\[ S_1 = 0.5 AB \left( \cos(w_{ct} + w)t + \cos(w_{ct} - w)t \right) \] (3.8)

The signal \( S_1 \) is then full-wave rectified and low pass filtered in order to extract the r.m.s. signal, (see fig. 3.5).

The full-wave rectifier and the low-pass filtering are modelled by the following equations
\[ S_2 = 0.5 AB \left| \cos(w_{ct} + w)t + \cos(w_{ct} - w)t \right| \]
\[ = 0.5 AB \left( \cos^2(w_{ct} + w)t + \cos^2(w_{ct} - w)t \right) \]
\[ + 2 \cos(w_{ct} + w)t \cdot \cos(w_{ct} - w)t \right)^{1/2} \]
\[ = 0.5 AB \left( \cos^2(w_{ct} + w)t + \cos^2(w_{ct} - w)t \right) \]
\[ + \cos 2w_{ct}t + \cos 2wt \]
\[ = 0.5 AB \left( \cos^2(w_{ct} + w)t + \cos^2(w_{ct} - w)t \right) \]
\[ + \cos^2w_{ct}t + \cos^2wt + 1 \right)^{1/2} \]
Fig. 3.5 Complete circuit diagram of the processing unit, (a) transmitter; (b) receiver.
After filtering the high frequency

\[
S_3 = 0.5 \ AB \ (\cos^2\omega t + 1)^{1/2} \\
= 0.5 \ AB \ (2(\cos 2\omega t + 1)/2)^{1/2} \\
= 0.5 \ AB \sqrt{2} \ \cos \omega t \tag{3.9}
\]

Finally equation, 3.9 gives the modulating signal multiplied by a coefficient 0.5 \sqrt{2} \ B. Therefore in order to get a signal as high as possible it is necessary to have a high energy light emitting diode.

Losses of light in the fibre optic itself or in the coupling devices are not easily quantifiable. When movement or bending of the fibre optic does not introduce a significant dynamic loss, a constant factor \( R \) of attenuation which varies from zero to one, depending on the coupling quality may be introduced, so equation (3.9) becomes.

\[
S_3 = 0.5 \ A \ B \ R \sqrt{2} \ \cos \omega t \\
= B \ R \ (A / \sqrt{2}) \ \cos \omega t \tag{3.10}
\]

Equation 3.10 shows that the root-mean-square of the signal multiplied by a coefficient is extracted by this technique. Root mean square to dc converters are commercially available in single chip packages but their frequency response is insufficient. So a purpose built root-mean-square to dc converter has been used to increase the frequency response (see fig. 3.5b).

It has been assumed previously that the system is linear.

### 3.3 SENSOR TIP

The membrane has two critical features. Firstly it has to be very flexible in order to detect small flow rates (see equation 3.6), and secondly it has to be of high reflectivity. Many materials has been tried including gold foil, other metals, latex, and liquid latex. Liquid latex had a low modulus of elasticity, however it has a lower reflectivity than metal materials, and the thickness of the membrane cannot be made reproducible. In order to overcome these problems different techniques have been used to coat the membrane with a material of high reflectivity such as
gold and silver, and without altering the elasticity. Thermal evaporation (under vacuum) has been used but the results were not successful. When gold foil was pressed carefully against the latex material (at room temperature), this gave a good reflecting membrane and the reflective surface was mechanically resistant. Another possibility was to trap an air bubble between the membrane and the fibre so that the membrane acts as an interface between the liquid and the gas (see Fig. 3.6). This technique has the advantage of having a high coefficient of reflection, however it was very difficult to keep the air bubble in place.

Fig. 3.6 Sensor tip with air trapped between the membrane and the optical fibre tip.

3.3.4 FIBRE OPTIC AND OPTICAL COUPLING SYSTEM
Two types of fibre optic configuration have been used. The first, which is relatively simple, consists of two separate fibres, one guiding the light to the membrane and the other returning the light to the detector (Fig. 3.7a). The second used two fibre bundles, one guided the light from the transmitter to a coupler, the second from the coupler to the receiver and a single fibre was used to both guide the light from the coupler to the membrane and from the membrane to the coupler, Fig. 3.7b. This had the advantage of reducing the cost and size of the fibre to be inserted as a catheter and also had the advantage of eliminating the non-linearity introduced by two fibres (see below), but has the disadvantage of increasing the light loss at the coupler.
Fig. 3.7 Fibre optics configuration used; (a) one fibre used to transmit and the other to receive the light; (b) single fibre used to transmit and receive the light at the same time.

A. POWER COUPLED FROM ONE FIBRE TO THE OTHER, WHEN FLAT MIRROR IS USED

A. 1 CASE OF TWO FIBRES
For the sake of simplicity the membrane will be considered to be a flat mirror. One fibre is used to transmit and the other used to receive the light. The fibres are separated from each other by a distance, 2a, and their tips are of distance, d, from the moving reflector (see Fig. 3.8a). This figure shows
that the minimum distance $d_m$ between fibre and membrane at which light coupling is achieved.

$$d_m = \frac{a}{\tan \alpha_c} \quad (3.11)$$

Where $\alpha_c$ is the acceptance cone angle of the fibre, depending on the characteristic of the fibre. In practice it is preferable to keep the distance $2a$ between the two fibres as small as possible in order to obtain the maximum sensitivity.

The estimation of the power coupled between the fibres is

---

Fig. 3.8 Displacement detection (a) using two fibres; (b) using a single fibre; (c) configuration used to calculate the amount of light reflected back for a single fibre.
given in the next paragraph. It is important to note that when the mirror (membrane) is situated between 0 and \(d_m\) from the ends of the fibres the power coupled is practically nil.

### A. 2 CASE OF SINGLE FIBRE

To calculate the ratio of power received to the power transmitted, the situation shown in fig. 3.8b is equivalent to that of fig. 3.8c.

The area over which light from the transmitter is spread by the time it reaches the received area, \(A\), is given by:

\[
A = \frac{2 \alpha_c}{2 \pi} \cdot 4 \pi (2d + x)^2 \tag{3.12}
\]

Where

- \(\alpha_c\) : is the acceptance angle.
- \(2 \pi\): represent the solid angle of the sphere.
- \(4 \pi (2d + x)\): represent the area of the sphere.

\[x = \frac{r}{\tan \alpha_c} \quad \text{(see Fig. 3.8c)}\]

then:

\[
A = 4 \alpha_c \left[ 4d^2 + \frac{4d \cdot r}{\tan \alpha_c} + \left(\frac{r}{\tan \alpha_c}\right)^2 \right] \tag{3.13}
\]

Since the received area is \(\pi r^2\), the maximum coupled power (ideal coupling) is given by:

\[
\frac{P_r}{P_o} = \frac{\pi r^2}{A} = \frac{\pi r^2}{4 \alpha_c \left[ 4d^2 + \frac{4d \cdot r}{\tan \alpha_c} + \left(\frac{r}{\tan \alpha_c}\right)^2 \right]} \tag{3.14}
\]

Where:

- \(P_r\) : Power received by the optical fibre.
- \(P_o\) : Power transmitted through the optical fibre.

If \(r\) is small and tangent \(\alpha_c\) big, which is the case in practical situation (\(r = \) few tens of microns and \(\alpha_c \approx 60^\circ\)) then:
Equation 3.15 has been verified experimentally by plotting the power reflected back, against the distance between the optical fibre and a flat reflective mirror, on log-log scales (fig. 3.9). A straight line with a slope equal to -2.2 was

\[
\frac{P_r}{P_o} = \frac{\pi r^2}{16 \alpha_c d^2} \alpha \frac{1}{d^2}
\]  

(3.15)

Fig. 3.9 Static calibration curve of the optical fibre displacement transducer.
obtained showing that the function was close to the inverse square law expected. However this equation cannot be applied when the flat reflector is replaced by a reflecting membrane, because rays reflected by the membrane diverge in directions other than the fibre-axis direction, and furthermore the direction depend on the curvature of the membrane which changes with the pressure applied to it. In order to find the relationship between the velocity and power reflected back, when the membrane is used a program simulation has been developed (see below).

3.4 COMPUTER SIMULATION
The flow-transducer characteristics described above, involve several independent variables such as the diameter of the fibre, \( a \), the diameter of the membrane, \( D \), the distance from the membrane to the tip of the fibre optic, \( L \), and the velocity of the fluid, \( V \). A simulation program was written to help understand the behaviour of the transducer by changing only one parameter at a time (see appendix 2). The program calculated the amount of light returning to the optical fibre after being reflected by the membrane. The membrane took the shape described by equation 3.2, and the program found the critical angle \( \theta_c \) beyond which the reflected rays were not received by the fibre optic. Knowledge of this angle enabled the integration limits to be defined for the calculation of the ratio of light received and the light transmitted by the fibre optic according to the following equation:

\[
I = \frac{\int_{0}^{\theta_c} 2\pi I(\theta) \sin \theta \, d\theta}{\int_{0}^{\pi/2} 2\pi I(\theta) \sin \theta \, d\theta} \tag{3.17}
\]

Where

\[\int_{0}^{\theta_c} 2\pi I(\theta) \sin \theta \, d\theta\] represents half the flux in a cone of half angle \( \theta_c \).
3-16

\[ \int_{0}^{\pi/2} 2\pi I(\theta) \sin \theta \, d\theta \] represent half the flux in a cone of half angle \( \pi/2 \).

And

\[ I(\theta) = I_o I_r \]

\( I_o \) is the axial intensity and \( I_r \) is the relative intensity. For the special case where the transmitting fibre optic behaves like a Lambertian Radiator, \( I(\theta) = I_o \cos \theta \).

In order to be able to change the characteristics of the transmitting diode, the program calculates the ratio of light from equation (3.17). The critical angle \( \theta_c \) was determined by: first finding the point of intersection, \( P \), of the rays emitted with the curve which represents the membrane; secondly finding the gradient of the tangent to the reflector at the point of intersection, \( P \), determined from the equation of the reflected ray; and finally finding the intersection point \( (Q) \) of the reflected ray with the plane of the tip of the optical fibre which is represented by equation \( X = -L \). If the point \( Q \) has the coordinates \((-L,Y)\) with \( Y = A/2 \), then the angle chosen corresponds to the critical angle (see Fig. 3.10).

Fig. 3.11 shows the outputs of the program for various distances \( L \), and various velocities. The diameters of the fibre optic and the membrane are kept constant. Different positions of the membrane relative to the tip of the fibre have been chosen. It can be seen from Fig. 3.11a to g, that the ratio of light is inversely proportional to the distance, \( L \), when considering the velocity, \( v \) equal to zero (which means a flat reflector). However when the membrane starts to bend (because velocity increases), then the ratio of light transmitted to light reflected depends on the relative position of the tip of the fibre and the membrane.

Figures 3.11a, b and c show that the ratio is proportional to the velocity, this situation is similar to that of flat mirror.
Figure 3.11d, and 3.11e show that the ratio is inversely proportional to the velocity for a certain range of velocities and becomes proportional to the velocity above a certain velocity. This can be explained in the same manner as for Figure 3.11a, b and c for one range, and when the ratio starts to be proportional to the velocity this can be explained by saying that when the membrane reaches a certain distance from the fibre, the deflection of light from the

![Diagram of optical fibre, membrane, and angle θ.](image)

Fig. 3.10 Configuration used to calculate the critical angle and the amount of light reflected back by the membrane.
axis of the fibre has a larger effect than the axial displacement.

Figures 3.11f. and 3.11g show that the deflection of light (due to the curvature of the membrane) from the optical fibre
Fig. 3.11 Simulation program output for different position of the membrane relative to the tip of the optical fibre; (a), (b) and (c) show that the ratio of light reflected back is proportional to the velocity; (d) and (e) show that the ratio is proportional to the velocity for one range and inversely proportional for the other range; (f) and (g) show that the ratio is inversely proportional to the velocity.
axis, causes the ratio to be inversely proportional to the velocity. In this case the figures show that the axial displacement have less effect then the deflection effect.

The situation shown in Fig. 3.11c has been found to correspond to the practical characteristic used, and therefore Fig. 3.11c has been replotted on logarithmic-linear scales. The characteristic obtained is shown in figure 3.12, a curve was fitted to the data, and then the function of the curve was extracted (see equation 3.18).

\[ Y = 0.06023 \exp\{1.348(\text{velocity})\} \]  

\((3.18)\)

![Graphical representation](image)

**Fig 3.12** Theoretical characteristics of the optical fibre velocimeter.
3. 5 TESTING

The system described above can be divided into three main blocks: the electronic circuit, the associated fibre optics, and the transducer tip. To gain a clear picture of the system, the three blocks were tested separately.

3. 5. 1 ELECTRONIC RESPONSE OF THE SYSTEM

Fig. 3.13 shows the experimental configuration used to monitor the frequency response of the electronic circuit. White noise of flat amplitude spectrum from 0 to 40 kHz was applied to the test point T1 (Fig. 3.5) and to a spectrum analyzer (ONNO SOKKI CF-10 Dual channel FFT analyser), and the output of the electronic circuit applied to the second channel of the spectrum analyzer. The analyzer then calculated the transfer function, between the two channels and the coherence. The frequency response of the electronic circuit was found to be flat up to 120 Hz.

![Diagram](Image)

Fig. 3.13 Setting used to measure the frequency response of the electronic circuit.

3. 5. 2 DISPLACEMENT RESPONSE

In order to test the electronic system associated with the optics, a mirror was attached to a piston driven by a hydraulic pump controlled by an external generator of white noise. The position of the piston or the mirror was measured by a linear potentiometer attached to the piston. The output of the linear potentiometer served as a reference and was applied to one channel of the spectrum analyzer. The output of the electronic system was connected to the second channel.
of the analyzer Fig. 3.14. The transfer function and the phase difference between the two channels was calculated and displayed by the analyzer Fig. 3.15.

Fig. 3.14 Setting used for the measurement of the displacement response of the system.
Fig. 3.15 Transfer function of the displacement system; (a) magnitude of the transfer function, (b) phase.
3. 5. 3 IN VITRO TESTING

This consisted of a circuit of silicone tubing with an internal diameter of 12 mm, the fluid flow was generated from a water tank, the flow rate was adjusted by a tap. The volume was measured using a timer and a measuring cylinder. Data obtained from the experiment were plotted in a graph with a log-linear scale. The characteristic obtained is shown in Fig. 3.16; a curve was fitted to the data obtained, and then the function of this curve was extracted (see equation 3.16).

Fig. 3.16 Experimental characteristics of the optical fibre velocimeter.
\[ Y = 19.86 \exp \left( 0.049 \text{(velocity)} \right) \] (3.16)

There is an offset at the origin because at no flow, light is still reflected by the membrane.

3.6 DISCUSSION AND CONCLUSION

To compare the experimental and theoretical characteristics of the transducer, it has been found convenient to plot the proportion of light reflected back on a logarithmic scale and the velocity on linear scales for both characteristics, since this gave linear functions which were easy to identify and compare. The equations obtained were similar but with different parameters. This is due to the arbitrary parameter used for the program simulation. In a practical situation the output of the instrument should be calibrated. The sensitivity of the sensor tip could be further increased by depositing the gold coating on both sides of the membrane.

In conclusion a new velocity transducer was designed, constructed, simulated and tested \textit{in vitro}. The experimental and theoretical findings correlated well, however it is important to bear in mind that this instrument measures point velocity, therefore its position and orientation inside the vessel should be known. It has also been shown that when a membrane is used (either for measuring pressure or flow), the response follows an exponential law and depends on different parameters shown above. However when a flat reflector is used the characteristic follows a simple square law.
CHAPTER 4
DOPPLER ULTRASOUND TECHNIQUES

4.1 HISTORY
Ultrasound Doppler equipment was first introduced by Satumura in 1959 when he studied flow patterns in peripheral arteries. Baker, in 1964, performed the first clinical studies using a non-directional Doppler instrument. The design was improved and the Doppler instrument was made directional by McLeod in 1967. Since then Doppler ultrasound equipment has been applied in areas such as cardiology (eg for studying flow through valves), obstetrics (eg for detecting the fetal heart), and in general circulation (eg for looking at movement, flow profiles, volumetric flow). Ultrasonic devices are also used in pressure measurement, to give more accurate readings than the conventional sphygmomanometer cuff and stethoscope, where Korotkoff sounds are too weak to be audible.

This thesis deals with techniques for the investigation of the circulation and specifically of flow velocity measurement using the Doppler ultrasound technique. A comprehensive review of the Doppler method is presented in this chapter.

4.2 PRINCIPLE
A radio-frequency oscillator tuned to the frequency of a probe (typically 1-15MHz), excites a piezoelectric element to generate a plane acoustic wave. This insonates the blood vessel and the back scattered energy is received by the same or another element, which converts the acoustic back-scattered signal to an electric signal, which is then amplified, demodulated and separated into forward and reverse flow components.

4.3 DOPPLER EFFECT
The Doppler effect was first described by The Austrian Christian Johann Doppler in 1842. Doppler postulated that the colour of a luminous body must change by relative motion of the body and the observer. The Doppler effect can be applied
to all waves including sound waves, as mentioned by Doppler himself. The Doppler principle when applied to sound waves, states that if a sound source moves relative to an observer, the observer will detect a signal whose frequency is shifted from that of the source.

4.3.1 PHYSICAL INTERPRETATION
First consider the case where all velocities act along the same axis.
If S (Source) and O (Observer) are not moving, the observer, during a time, t will receive \( \frac{ct}{cT} \) or \( \frac{ct}{S} \) waves, were c is the velocity of sound in the medium, T is the period of the wave and \( \delta \) the wave length of the source.

A. SOURCE IS FIXED AND OBSERVER MOVING
(a) Observer moving towards the source.
When the observer is in motion toward the source, the observer receives \( \frac{ct}{cT} + \frac{voT}{cT} \) waves during a period of time. And since the frequency is the number of wavelengths per unit of time then;

\[
f' = \frac{ct + voT}{(cT)t} = \frac{c + vo}{c} f \quad (4.1)
\]

(b) If the observer is moving in the opposite direction then:

\[
f' = \frac{c - vo}{c} f \quad (4.2)
\]

B. SOURCE MOVING AND THE OBSERVER FIXED
a. Source moving toward the observer.
At each vibration the source has moved by the distance, \( v_s T \) therefore the wavelengths has been shortened by \( \delta_2 = v_s T \) and the resulting wavelengths, \( \delta' \), will be \( \delta' = \delta - \delta_2 = cT - v_s T = cT' \).

NOTE: Although the source moves with the velocity v and is emitting the waves, the wave velocity is still the same.
Finally:

\[ f' = \frac{c}{c - v_s} f. \]  (4.3)

b. Source moving in the opposite direction.
With the same proof as above, if we replace \( v_s \) by \(-v_s\), we will obtain:

\[ f' = \frac{c}{c + v_s} f. \]  (4.4)

In the general case when the vector velocity is considered the Doppler shift frequency is given by:

\[ f' = \frac{c \pm v_0 \cos \theta}{c \pm v_0 \cos \theta} \]  (4.5)

In medical instrumentation the phenomenon is more complicated because the observer is sending the wave, whilst the source is moving and reflecting the wave.
Let us consider the case where one element both emits and receives the wave (fig. 4.1).

Fig. 4.1 Representation of a single transmitter/Receiver element.

In this case the process can be divided into two: first the transmitter (source) transmits the wave and the target
(observer) is moving toward the source. Equation (4.1) can be applied, therefore:

$$f' = \frac{c + v_o}{c} f$$  \hspace{1cm} (4.6)$$

When the observer (target) receives the wave it will reflect it and the source will become the observer and the observer will become the source (equation 3 will be applied) so that the frequency at the receiver will be

$$f'' = \frac{c}{c - v_o} f'$$  \hspace{1cm} (4.7)$$

The combination of equation (4.6) and (4.7) will give the final Doppler shift equation

$$f'' = \frac{c + v_o}{c} \cdot \frac{c}{c - v_o} f$$

$$f'' = \frac{c + v \cos \theta}{c - v \cos \theta} \cdot f$$  \hspace{1cm} (4.8)$$

Since the Doppler shift, df, is represented by the difference between the transmitted and the received frequencies, we have:

$$df = \frac{2 \cdot v \cdot \cos \theta}{c - v \cdot \cos \theta} \cdot f$$  \hspace{1cm} (4.9)$$

In the practical situation the velocity of sound, c, in tissue is much greater than the velocity of the red cell, v, therefore:

$$df \approx \frac{2 \cdot v \cdot \cos \theta}{c} \cdot f$$  \hspace{1cm} (4.10)$$

In the case of two elements transducer as shown in Fig. 4.2. Equation (4.6) will be

$$f' = \frac{c + v \cos \theta_T}{c} f$$
Fig. 4.2 Representation of twin element, one element transmitting and the other receiving.

And Equation (4.7) will be,

\[ f'' = \frac{c}{c - v \cos \theta_R} \cdot f \]

And finally:

\[ f'' = \frac{c + v \cdot \cos \theta_T}{c - v \cdot \cos \theta_R} \cdot f \quad (4.11) \]

\[ df = f'' - f = \frac{c + v \cdot \cos \theta_T}{c - v \cdot \cos \theta_R} - \frac{c - v \cdot \cos \theta_T}{c - v \cdot \cos \theta_R} \]

\[ = \frac{v [\cos \theta_R + \cos \theta_T]}{c} \cdot f \quad (4.12) \]

Or \( \theta_T = \theta + \theta_A \)

And \( \theta_R = \theta - \theta_A \)

\[ \cos \theta_T + \cos \theta_R = \cos (\theta + \theta_A) + \cos (\theta - \theta_A) = 2 \cos \theta \cdot \cos \theta_A \]

Therefore

\[ df = \frac{2 \cdot v \cdot \cos \theta}{c} \cdot \cos \theta_A \cdot f \quad (4.13) \]
Where $\theta_A$ is half the angle between the transmitting and the receiving elements. In the case of small angle $\theta_A$, $\cos \theta_A = 1$.

Then

$$df = \frac{2 v \cdot \cos \theta}{c} \cdot f \quad (4.14)$$

In conclusion if two elements are to be used, the angle $\theta_A$ should be kept very small or a correction should be introduced if quantitative velocity measurements are required.

Blood flow measurement using Doppler ultrasound can be divided into two principal techniques. The continuous (CW) and the pulsed wave (PW) methods. The two techniques will be described and compared and the technique most suitable for continuous and ambulatory uses will be selected.

4.4 CONTINUOUS WAVE TECHNIQUE
The continuous wave technique is very widely used in clinical diagnostics. It involves continuous acoustic waves with frequencies in the range (1-20MHz). This wave can be transmitted transcutaneously to the vessel to be interrogated. The wave is partially reflected back and shifted in frequency because of the Doppler effect. Note that in this application, the spectrum of the signal reflected back is the combination of every signal coming from each individual moving red cell present in the volume delineated by the intersection of the transmitting and the receiving beams. The beams can be focused in order to interrogate flow at a 'specific' region see chapter 6. The maximum measurable velocity when using continuous wave techniques is only limited by the quality factor 'Q' of the transmitting/receiving elements and the electronic circuit. However this is not so when using pulse wave techniques (see section 4.5).
4.5 PULSED WAVE TECHNIQUE
Continuous wave systems detect the Doppler frequency shifts of all the targets moving within the sensitive region, and do not discriminate between targets which are situated at different distances or ranges. Pulsed Doppler systems use the principles of pulsed-echo to select the range or depth and the principle of Doppler to determine the velocity.

The system operates as follows. A master oscillator tuned to the frequency of the probe is combined with a pulsed signal to produce a burst of a few cycles as shown in Fig. 4.3. The return echoes from moving targets are frequency and amplitude modulated. Continuous echoes from different depths are received at different times (time of flight = $T_f$). These echoes are then demodulated in the same manner as for continuous waves. However by taking a sample of the signal demodulated after a time $T_f$ during a certain time $T_r$, it is possible to determine the depth at which the flow is interrogated and the sampling volume which is dependent on $T_r$ (Fig. 4.3).

Before recovering the complete audio or Doppler shift signal many pulses of the pulse repetition frequency (PRF) need to occur. If the PRF is the sampling frequency, the maximum frequency audio signal that can be recovered is one with a frequency less or equal to half the PRF frequency. For instance if the PRF is equal to 25 kHz the maximum Doppler shift detectable is 12.5 kHz.

4.6 DIRECTION DETECTION
The Doppler shift frequency was shown (above) to be:

$$f_d = \frac{2.v.f.\cos \theta}{c}$$

The back scattered ultrasonic wave is of the following form.

$$R = A\cos(wt+\phi) + A_f\cos[(w+w_f)t+\phi_f] + A_r\cos[(w-w_r)t+\phi_r] + A_{fc}\cos[(w+w_{cf})t+\phi_{cf}] + A_{cr}\cos[(w-w_{cr})t+\phi_{cr}] \quad (4.15)$$

In practice $A >> A_f$ and $A_{cf} >> A_f$

$A >> A_r$ and $A_{cr} >> A_r$. 
(a) Transducer
Coupling-gel
Sample volume
Vessel

(b) PRF
Monostable (delay gate)
Transmitter
Master oscillator
Monostable (range gate)
Probe
Demodulator
\(Q(t)\)
Demodulator
\(D(t)\)
Fig. 4.3 Pulsed Doppler system; (a) Probe-vessel configuration; (b) block diagram of the transmitter and receiver; (c) Timing diagram.
Where \( A \): magnitude of the signal due to the echo from the surrounding tissues.

\( A_f \): Magnitude of the signal due to the forward flow.

\( A_r \): " " " " " " reverse "

\( A_{cf} \): " " " " " " forward tissue movement.

\( A_{cr} \): " " " " " " reverse tissue movement.

The spectrum of the reflected signal is shown in Fig. 4.4.

The way in which the Doppler shift frequencies, due to the forward and reverse flow, are separated and discriminated from the other signals, will be described.

### 4.6.1 SINGLE SIDE BAND FILTERING

This process of separating the forward and reverse signal is shown in fig. 4.5. After processing the signal received by the probe (see equation 4.15), the signal at each stage will be as follows.

\[
R_L = A_f \cos[(w-w_r)t + \Phi_r] \quad (4.16a)
\]

\[
F_p = A_r \cos[(w+w_r)t + \Phi_r] \quad (4.16b)
\]

\[
F_1 = A_f \cos[(w+w_f)t + \Phi_f].\cos wt \quad (4.16c)
\]

\[
R_1 = A_r \cos[(w-w_r)t + \Phi_r].\cos wt \quad (4.16d)
\]

\[
F_1 = 0.5 A_f \left[\cos((2w+w_f)t + \Phi_f) + \cos (w_r t+ \Phi_f)\right] \quad (4.16e)
\]

\[
R_1 = 0.5 A_f \cos[(w_f t+\Phi_f) + \cos (w_r t+ \Phi_r) \quad (4.16f)
\]

\[
F = 0.5 A_f \cos[(w_f t+\Phi_f) \quad (4.16g)
\]

\[
R = 0.5 A_r \cos[(w_r t+\Phi_r) \quad (4.16h)
\]

### PRACTICAL CONSIDERATIONS

Since the characteristics of the filters are not usually synchronised to the oscillator, if there is any drift in frequency the carrier and echo signals and the signal due to the movement of tissue will appear as a forward or reverse signal depending on the direction and extent of the shift of the transmitting frequency.

Since the filters are not ideal and since the amplitude of the carrier signal, echo and moving tissue signal are very high, then they will 'leak' into both channels.
Fig. 4.4 Doppler power spectrum; (a) individual power spectrum; (b) composite power spectrum, (The spectrum is not necessary symmetric).
Fig. 4.5 Block diagram showing the single-side band system.

Therefore
(1) The filters should be very stable and have very sharp cut offs (crystal filters are recommended).
(2) The transmitting frequency should be extremely stable.

4.6.2 HETERODYNE DETECTION
The description of the process is given in Fig. 4.6.

Fig. 4.6 Block diagram of the heterodyne detection system.
After mixing the two signals generated by the two oscillators, this leads to:
\[
\cos w_0 t \cos w_n t = 0.5 \cos (w_0 + w_n) t + \cos (w_0 - w_n) t. \tag{4.17}
\]
And after filtering and mixing the previous signal with the received signals, this leads to
\[
R \cos (w_0 - w_n) t =
\begin{align*}
&\{ A \cos (w_0 t + \phi) + A_f \cos [(w_0 + w_f) t + \phi_f] + A_r \cos ((w_0 - w_r) t + \phi_r) + \\
&A_c f \cos [(w_0 + w_c f) t + \phi_c f] + A_c r \cos [(w_0 - w_c r) t + \phi_c r]\}. \cos (w_0 - w_n) t =
\end{align*}
\]
\[
0.5 A \cos [(2w_0 - w_n) t + \phi] + \cos (w_n + \phi) + \\
0.5 A_f \cos [(2w_0 + w_f - w_n) t + \phi_f] + \cos (w_f + w_n) t + \phi_f) + \\
0.5 A_r \cos [(2w_0 + w_r - w_n) t + \phi_r] + \cos (w_n + w_r) t + \phi_r) + \\
0.5 A_c f \cos [(2w_0 + w_c f - w_n) t + \phi_c f] + \cos (w_c f + w_n) t + \phi_c f) + \\
0.5 A_c r \cos [(2w_0 - w_n - w_c r) t + \phi_c r] + \cos (w_n + w_c r) t + \phi_c r]. \tag{4.18}
\]

After filtering the high frequency
\[
S_1 = 0.5 A \cos (w_n + \phi_f) + 0.5 A_f \cos [(w_n + w_f) + \phi_f] + \\
0.5 A_r \cos [(w_n - w_r) + \phi_r] + 0.5 A_c f \cos [(w_n + w_c f) + \phi_c f] + \\
0.5 A_c r \cos [(w_n - w_c r) + \phi_c r]. \tag{4.19}
\]

The whole spectrum has now been shifted from \(w_0\) to \(w_n\) and therefore the forward and reverse signals are situated at both sides of the heterodyne frequency and the echo from the clutter is still present. A bandpass filter is needed to remove the echo and the forward plus the reverse signals due to the movement of tissue. Finally after filtering
\[
S = 0.5 A_f \cos [(w_n + w_f) t + \phi_f] + 0.5 A_r \cos [(w_n - w_r) t + \phi_r] \tag{4.20}
\]

**PRACTICAL CONSIDERATIONS**

A slow drift of the transmitting frequency \(w_0\) is acceptable in this application, but a drift of the heterodyne frequency will cause a leakage of the echo signal and signal of the movement of tissue to the forward or reverse flow signal depending on the direction of the heterodyne frequency shift. Therefore the heterodyne frequency should be very stable. In practice the heterodyne frequency is obtained by division of the transmitting frequency.
4.6. 3 QUADRATURE PHASE DETECTION

The quadrature phase detection separates the real and imaginary parts of the Doppler signal. The imaginary and real parts are obtained by two signals in phase quadrature. If we consider the simplified form of the backscattered wave \( R(A_f(t) = A_f \cos(w_f t + \Phi_f) + A_r \cos((w_f + \Phi_f) t - \Phi_r)) \) at the output of the demodulators (see fig. 4.7), the signals will be:

\[
D(t) = 0.5 \left[ A_f \cos(w_f t + \Phi_f) + A_r \cos(w_f t - \Phi_r) \right] \quad (4.21)
\]
\[
Q(t) = 0.5 \left[ A_f \cos(w_f t + \Phi_f - \pi/2) + A_r \cos(w_f t - \Phi_r + \pi/2) \right] \quad (4.22)
\]

There are three methods of separating the forward and the reverse signals by using the quadrature signal \( D(t) \) and \( Q(t) \). These methods are described below.

A. TIME DOMAIN PROCESSING

The concept of this technique is shown in Fig. 4.7.

[Diagram of phase detector and flow (F and R)]

Fig. 4.7 Separation of the forward and reverse signals in time domain.

In this case the flow is considered as unidirectional which means that the flow can be either forward or reverse but not both at the same time. Flow in an artery can however be bidirectional. The explanation of the system is nevertheless worthwhile. In the case of forward flow the equations (4.21) and (4.22) become:

\[
D(t) = 0.5 A_f \cos(w_f t + \Phi_f) \quad (4.23)
\]
\[
Q(t) = 0.5 A_f \cos(w_f t + \Phi_f - \pi/2) \quad (4.24)
\]
Notice that \( D(t) \) is leading \( Q(t) \) in time. However in the case of reverse flow the equations (4.21) and (4.22) become:

\[
D(t) = 0.5 A_T \cos(w_T t - \phi_T). \\
Q(t) = 0.5 A_T \cos(w_T t + \pi/2 - \phi_T). 
\]

In this case we notice that \( Q(t) \) leads \( D(t) \) in time. By detecting which channel [\( Q(t) \) or \( D(t) \)] leads the other, a decision can be made to switch the outputs of the demodulators to forward or reverse channel; this technique was first developed by McLeod in 1967. As can be seen this technique suffers from the fact that it cannot handle forward and reverse flow simultaneously (which may be of considerable diagnostic importance). Also the switching system may also introduce noise.

**B. FREQUENCY DOMAIN PROCESSING**

The process is shown in Fig. 4.8. after demodulation the quadrature signals, leads to

\[
D_1(t) = 0.5[A_f \cos(w_f t + \phi_f) + A_r \cos(w_T t - \phi_T)].\sin w_2 t \\
= 0.5[A_f \cos(w_f t + \phi_f).\sin w_2 t + A_r \cos(w_T t - \phi_T).\sin w_2 t] 
\]

**Fig. 4.8 Separation of the forward and reverse signals in frequency domain.**
\[ Q_1(t) = 0.5[A_f \sin(w_2 t + \phi_f) - A_r \sin(w_r t - \phi_r)] \cdot \cos w_2 t. \]
\[ = 0.5[A_f \sin(w_2 t + \phi_f) \cdot \cos w_2 t - A_r \sin(w_r t - \phi_r) \cdot \cos w_2 t]. \]

\[ D_1(t) + Q_1(t) = 0.5 A_f [\cos(w_2 t + \phi_f) \cdot \sin w_2 t + \sin(w_2 t + \phi_f) \cdot \cos w_2 t] + 0.5 A_r [\cos(w_r t - \phi_r) \sin w_2 t - A_r \sin(w_r t - \phi_r) \cos w_2 t] \]

Or \[
\cos a \cos b + \sin a \cos b = \sin(a+b) \]
And \[
\cos b - \sin a \cos b = \sin(b-a)\]

Therefore

\[
S(t) = D_1(t) + Q_1(t) = 0.5 A_f \sin[(w_2 + w_f) t + \phi_f)] + 0.5 A_r \sin[(w_2 - w_r) t + \phi_r)] \quad (4.25)
\]

Notice that the resultant signal \( S(t) \) is similar to that obtained by Heterodyne detection system.

In this case, where 4 quadrature signals were used (\( \sin w_0 t, \cos w_0 t \) and (\( \sin w_2 t, \cos w_2 t \)), the error or cross-talk can cause a problem. Quadrature square waves however can minimize the cross talk. Frequency domain or heterodyne detection systems share the same advantage since only a single channel spectrum analyzer is needed, however the signal has to be further processed if it has to be applied to loudspeakers.

C. PHASE DOMAIN PROCESSING

The system is shown in Fig. 4.9.

From equation (1) and (2) we derive \( Q_{\pi/2}(t) \) and \( D_{\pi/2}(t) \), where \( Q_{\pi/2}(t) = Q(wt + \pi/2) \) and \( D_{\pi/2}(t) = D(wt + \pi/2) \).

Therefore:

\[
D(t) + Q_{\pi/2}(t) = \cos(w_f t + \phi_f) = F \quad (4.26)
\]
\[
Q(t) + D_{\pi/2}(t) = \cos(w_r t - \phi_r + \pi/2)] = R \quad (4.27)
\]

4. 7. CHOICE OF THE PHASE SHIFTER

If the phase-domain processing technique is to be used, it is very important to decide which phase shifter is going to be used; factors such as bandwidth, precision (phase), noise level, and in the case of ambulatory monitoring the power consumption, size, simplicity and cost factors have to be considered.
A review of different SSB phasing techniques has been published (Harrison, 1978). Considering noise level, power consumption and size it is preferable to use a passive phase shifter. If a large bandwidth is necessary it is not possible to use this passive device because the bandwidth are limited at about few kHz. Active phase shifters can accomplish a phase shift of $90^\circ \pm 2^\circ$ over 12 kHz or more if needed (see chapter 6).
CHAPTER 5
DOPPLER SIGNAL PROCESSING

5.1 INTRODUCTION
The Doppler signal obtained from blood flow results from the superposition of ultrasound scattered from each red cell. The Doppler signal can be represented by three variables \( f, A(f), t \) where \( f \) is the Doppler shift frequency which gives the information on velocity, \( A(f) \) is the magnitude of each frequency, which gives information on the relative number of blood cells moving at that specific velocity, and \( t \) represents time. The best way of analyzing the Doppler signal is by performing spectral analysis, in fact it is well established that the use of spectral analysis improves the effectiveness in detecting abnormalities in blood flow (Macpherson et al, 1981; Wells and Skidmore, 1985; Evans et al, 1989a). Blood flow in the carotid and peripheral arteries has been extensively investigated by this technique because of their accessibility. It is important to note that the display of the spectra in real time is very important since (i) it allows the operator to interact with the measuring system and the analyzer (fig. 5.1) and (ii) makes it easier to recognise abnormalities in the flow. However it is worthwhile to summarize the different techniques available for processing the Doppler signals and extracting clinical data.

A good spectrum analyser should have the following attributes:

(1) Visual feed-back should be available instantaneously to the operator.
(2) The system must be capable of handling the whole range of frequencies present in the Doppler signal (50 Hz to 20kHz).
(3) The system has to have a wider dynamic range than that of the Doppler signal which is normally around 40 dB.
5. 2 SPECTRAL ANALYSIS TECHNIQUE
In order to satisfy the above requirements, various techniques have been developed and these can be divided into three main categories which are described below.

5. 2. 1 ANALOGUE TECHNIQUE
This technique uses several narrow band-pass filters, with different centre frequencies. The filter inputs are connected together to receive the Doppler signal (fig. 5.2). The output of each filter is sampled in turn and fed to an electronic device which indicates the r.m.s value. In practice, 80 bandpass filters have been used to achieve an 'acceptable' frequency resolution. Various spectrum analyzers were based on this principle and have been used for various clinical applications (Johnston et al, 1978; Yao et al, 1970). This is not the most appropriate technique with current electronic technology.

5. 2. 2 ANALOGUE/DIGITAL TECHNIQUE
In this technique, only one filter is used instead of many band pass filters as shown above. The Doppler signal is shifted in frequency before being filtered by the single band-pass filter. Since the frequency shift is known, the power spectrum of the signal can be found. This is known as the heterodyne technique. An alternative method is to sample, digitize and store the Doppler signal and circulate it
through a shift register at a higher frequency than that at which it was sampled. This method is known as the time compression technique (Coghlan et al., 1974, Stevens et al., 1976).

![Block diagram of an analogue spectrum analyzer.](image)

**Fig. 5.2** Block diagram of an analogue spectrum analyzer.

### 5.2.3 Digital Technique

The Doppler signal is first sampled at an appropriate rate and a Fourier transform performed, either by the discrete Fourier transform (DFT) or by the fast Fourier transform (FFT) algorithm. To date, personal computers are not fast enough to perform the FFT in real time. The introduction of digital signal processors (DSP) however makes the system run faster by carrying out the multiplication in hardware, for instance the TMS320C25 is able to compute a 32-bit product in one machine cycle (40ns).

### A. Design Criteria for FFT Analyzer

The relation between the maximum velocity and the maximum Doppler shift frequency is given by equation 5.1.
\[ dF_{\text{max}} = \frac{2fo \cdot \cos \theta \cdot v_{\text{max}}}{c} \]  

The symbols carry forward their meaning from chapter 4. Therefore \( v_{\text{max}}, f_0, \cos \theta, c \) will determine the sampling frequency \( (f_{\text{sa}}) \), \( (f_{\text{sa}} \geq 2dF_{\text{max}}) \).

Most commercial and research machines use 256 samples to produce one spectrum, so that the frame length \( (TA) \) is given by \( 256/f_{\text{sa}} \). Therefore in the case where the maximum Doppler signal frequency is 20 kHz, the frame length will be 6.4 ms \( (256/40.10^3) \).

The available 256 samples will produce a spectrum with 128 components (see fig. 5.3).

![Diagram](image)

Fig. 5.3 From a frame of 256 points of Doppler signal, the FFT produces 128 frequency components (Bins).

**B. REPRESENTATION OF DOPPLER SIGNAL**

In the digital technique which was used for the study described in this thesis, each spectrum was displayed on a vertical colour coded line and contained 128 points representing 128 different frequencies. The colour display represented the power of each frequency and the x-axis time. The resolution in the time-axis is determined by the rate of generating new independent spectra. Fig. 5.4 shows an example
of a Doppler sonogram recorded with a directional flowmeter from a femoral by-pass graft, and processed by a purpose built spectrum analyser system described by Schlindwein et al 1988.

Fig. 5.4 Sonogram of Doppler signal (x-axis represents time, y-axis represents frequency and the magnitude of the signal is color-coded.

5. 3 ANALOGUE SIGNAL PROCESSORS
In order to simplify the analysis of the Doppler signal, different single-frequency estimators, including root mean square, maximum and mean frequency estimators have been developed. These techniques are described below.

5. 3. 1 ZERO CROSSING TECHNIQUE
A zero crossing detector gives an output pulse every time a signal crosses its time averaged mean value. Rice (1944)
theoretically analysed and predicted the number \((N)\) of zero crossings from the spectral content of a signal.

\[
N^2 = \frac{\int_{f_1}^{f_2} f^2 \cdot P(f) \, df}{\int_{f_1}^{f_2} P(f) \, df}
\]

(5.2)

Where

\(P(f)\) : Power spectrum.

Equation 5.2 is valid only if there is no constant phase relationship between different frequencies (Rice 1944). It has always been accepted that this condition is valid in the case of Doppler signals.

The output, \(N\) of the zero crossing technique is proportional to the root mean square frequency of the input signal. Kato et al (1965) were the first to apply this theory to Doppler signals and following this the technique became popular because of its simplicity. It is becoming less popular now in the medical field as the digital technique gives much more information.

A. ERROR AND ARTIFACT OF THE ZERO CROSSING TECHNIQUE

The term \(\int P(f) \, df\) represents the area under the curve \(P(f)\) which depends on the profile of the flow. Consequently if the profile changes, the term \(\int f^2 P(f) \, df\) changes and \(N\) changes, for instance when a flat profile is present, the zero crossing frequency \(f_{zc}\) is equal to the mean frequency \(f\) and with a parabolic profile \(f_{zc}\) is equal to 1.16 \(f\) (Woodcock et al 1972; Evans et al, 1982, 1989b), so that an error of 16 per cent is introduced when flow profile changes from flat to parabolic. The flow profile is likely to vary from place to place. The flow profile in a large artery may be flat (Schultz et al, 1969), in a bifurcation the flow profile is asymmetric and in the case of disease or stenosis the flow may become disturbed. Since the profile is not generally known the zero crossing technique is unreliable for
quantifying flow. It is also important to note that the filters for removing the clutter and the carrier signal, will distort $P(f)$, this has the same consequences as changing the flow profile.

**B. NOISE EFFECTS**

Noise level in the Doppler signal may be quite significant and cause the zero crossing detector to trigger, especially during diastole when the Doppler signal is close to zero. The triggering from noise can be reduced by an offset trigger level and most of the commercial zero crossing detectors use the set-reset system (Schmitt trigger). However the theory given above is strictly only applicable to the true zero crossing and not to the crossings at any other level. It has also been shown that the amplitude of the signal at a given trigger level has to be considered if quantitative results are expected (Flax et al, 1973).

Other factors such as uneven insonation and filter roll off affect the power spectrum and consequently the output of the zero crossing circuit. Gerard (1988) still believes that the digital zero crossing technique gives better results than FFT when signal to noise ratio is low, and gives comparable results when the signal to noise ratio is high.

**5. 3. 2 FIRST MOMENT FOLLOWER**

The first moment, $M_f$, of the Doppler power spectrum is defined as $\int f \cdot P(f) \cdot df$. It has been theoretically shown (Saini et al, 1983) that the first moment of the Doppler frequency spectrum is a more linear measure of volume flow rate than peak or mean flow velocity. However it does not give a quantified value of flow. Uneven insonation and filters can affect the linearity.

**5. 3. 3 MAXIMUM FREQUENCY FOLLOWER**

The instantaneous maximum frequency can be extracted from the Doppler signal using either analogue or digital methods. If only the maximum frequency is of interest, the former method may be more appropriate if the cost is an important consideration.
It has been shown by Evans (1989b) that the maximum frequency is less affected by noise than the intensity weighted mean frequency, and when it is affected by noise, it is easy recognized as such on the sonogram, however the relationship between the maximum and the mean frequency depends on the flow profile. For instance in the case of plug flow (flat profile) the mean and the maximum frequency are almost equal, however in the case of parabolic flow, the maximum frequency is twice the mean frequency. One technique for tracking the maximum frequency has been described by Skidmore et al (1978). It consists of changing the cut-off frequency of a high-pass filter whose output is rectified, filtered and fed back. The technique suffers from the fact that it is sensitive to the signal level, because of the design of the voltage-control low-pass filter. Another technique is based on the phase locked loop (PLL) (Sainz et al, 1976). It has been difficult to find any precise analysis of the mechanism involved for the extraction of the maximum frequency by using PLL. Another maximum frequency tracker has been described by Nowicki (1984).

5. 3. 4 MEAN FREQUENCY FOLLOWER
The mean blood flow velocity has been found useful in the assessment of vascular disease because it is linearly dependent on the volume flow, \( Q \).

\[
Q(t) = \dot{v}(t) \cdot S(t) \tag{5.3}
\]

\( Q(t) = \) Instantaneous volume flow
\( \dot{v}(t) = \) Instantaneous spatial mean velocity
\( S(t) = \) Instantaneous cross section area of the artery

The mean frequency of the Doppler signal can be calculated after performing a frequency analysis.
\[
- \omega = \frac{\int w \cdot P(w) \cdot dw}{\int P(w) \cdot dw}
\] (5.4)

Where \( P(w) \) represents the Doppler power spectrum.

An analogue mean velocity processor developed by Arts and Roevros, (1972) is shown in fig. 5.5, and is based on differentiation, multiplication and summation of signals.

\[
D(t) = A_f \cos(w_ft + \phi_f) + A_r \cos(w_rt - \phi_r)
\]

\[
Q(t) = A_f \cos(w_ft + \phi_f + \pi/2) + A_r \cos(w_rt - \phi_r + \pi/2)
\]

\[
Q(t) = -A_f \sin(w_ft + \phi_f) - A_r \sin(w_rt - \phi_r)
\]

Fig. 5.5 Block diagram for calculating the intensity weighted mean frequency (From Arts and Roevro, 1972).

The system description is as follow:
\[
\frac{dQ(t)}{dt} \cdot D(t) = -\left[ w_f A_f \cos(w_f t + \phi_f) + A_r w_r \cos(w_r t - \phi_r) \right] \\
\quad \left[ A_f \cos(w_f t + \phi_f) + A_r \cos(w_r t - \phi_r) \right]
\]

\[
= -\left[ w_f A_f^2 \cos^2(w_f t + \phi_f) + w_r A_r^2 \cos^2(w_r t - \phi_r) \right] \\
\quad + (w_f + w_r) [A_r A_f \cos(w_f t + \phi_f) \cos(w_r t - \phi_r)]
\]

Since

\[ w_f A_f \cos(w_f t + \phi_f) \] and \[ w_r A_r \cos(w_r t - \phi_r) \] are
uncorrelated and their average is equal to zero then the term
\[ Z = (w_f + w_r) [A_r A_f \cos(w_f t + \phi_f) \cos(w_r t - \phi_r)] \]
will average out to zero.

Therefore

\[ Q(t)/dt \cdot D(t) = X(t) = -\left[ A_f w_f \cos^2(w_f t + \phi_f) \right] \\
\quad + A_r w_r \cos^2(w_r t - \phi_r) \]

Since \( \cos^2 \theta = (1 + \cos 2\theta)/2 \)

Therefore

\[ X(t) = -0.5 \left[ A_f^2 w_f + A_r^2 w_r + A_f^2 w_f \cos 2(w_f t + \phi_f) \right] \\
\quad + A_r^2 w_r \cos 2(w_r t - \phi_r) \]

After filtering the AC signal

\[ X(t) = -0.5 \left( A_f^2 w_f + A_r^2 w_r \right) \]

\[ D^2(t) = [A_f \cos(w_f t + \phi_f) + A_r \cos(w_r t - \phi_r)]^2 \]

\[ = A_f^2 \cos^2(w_f t + \phi_f) + A_r^2 \cos^2(w_r t - \phi_r) \]

\[ + 2 A_f A_r \cos(w_f t + \phi_f) \cdot \cos(w_r t - \phi_r) \]

For the same reason cited above the term
\( 2 A_f A_r \cos(w_f t + \phi_f) \cdot \cos(w_r t - \phi_r) \) will average out.

Finally,

\[ D^2(t) = Y = A_f^2 \cos^2(w_f t + \phi_f) + A_r^2 \cos^2(w_r t - \phi_r) \]

After filtering

\[ Y = 0.5 \left( A_f^2 + A_r^2 \right) \]

Finally,

\[ \frac{X}{Y} = -\frac{A_f^2 w_f + A_r^2 w_r}{A_f^2 + A_r^2} \quad (5.5) \]

Which is the intensity weighted mean frequency.
Note in the previous equation that the reverse signal is seen by this signal processor as a forward signal and therefore in the case of simultaneous forward and reverse flow the processor will give a wrong reading. The factor $Z$ can introduce noise to the system because an averaging is needed and since DC signals are involved drift will occur.
CHAPTER 6
A PORTABLE DIRECTIONAL UNIT

6.1 INTRODUCTION
The Design of a portable directional Doppler velocimeter and purpose-built probes intended for monitoring blood flow in femorodistal bypass grafts in ambulatory patients is described. The design is based on a continuous-wave Doppler technique with quadrature demodulation. The Doppler unit was interfaced to a "personal-stereo" recorder to store the Doppler signals. Details on signal processing and the method of improving the signal-to-noise ratio, reducing the power consumption and the size are described.
6. 2 FACTORS TO BE CONSIDERED FOR AMBULATORY MONITORING

6. 2. 1 POWER CONSUMPTION
Pulsed wave systems consume a considerable amount of power since the transmitting element needs a voltage of up to 100 Volts peak to peak in order to achieve an acceptable sensitivity. An eight pole low pass filter to remove the PRF signal, and the control logic also draw a large amount of current. It is important to consider the power consumption factor if the system has to be implemented for ambulatory monitoring.

6. 2. 2 MOVEMENT ARTIFACTS
The pulsed wave technique with short gates is very sensitive to movement since it interrogates flow at a specific range, therefore the relative movement of the probe and the vessel will result in deterioration of the Doppler and the echo signals.

6. 2. 3 FREQUENCY BANDWIDTH
With ambulatory patients the flow velocity is much greater than with resting patients, therefore the unit has to be of very large bandwidth. This has to be considered in the case of pulsed wave technique since the maximum Doppler shift frequency is limited by the PRF frequency. For instance if we want to interrogate flow at distance D equal to 5 centimetres deep vessel, the first echo will be received after \(0.666 \mu s\) \((2D/C = 2 \times 5 \times 10^{-2}/1500)\) which correspond to a frequency of 15 kHz therefore the PRF frequency should be lower than 15 kHz, otherwise the next echo will be superimposed with the previous one. Finally the maximum Doppler shift frequency should not exceed 7 kHz in this case.

6. 2. 4 TIME OF INSONATION
Although in most Doppler units, only the average power is given, it is important for safety reasons to consider the instantaneous radiation administered to the patient. In the case of the pulsed wave technique the instantaneous value is very high. For continuous monitoring this factor should not be ignored.
6. 3 DESIGN AND CONSTRUCTION

The system comprises three separate parts: a continuous wave Doppler unit (with an integral timing circuit), an audio cassette recorder (Sony Walkman), and a Doppler probe. Although a number of small lightweight Doppler units are commercially available, they are, in general, non-directional and have limited bandwidth, and so it was necessary to design and construct a special unit. Commercially available Doppler probes are also unsuitable for ambulatory monitoring, and so again a special probe had to be designed and constructed. A block diagram of the complete recording system is shown in fig. 6.1.

![Block diagram of the complete recording system](image)

Fig. 6.1 Block diagram of the complete recording system

6. 4 THE DOPPLER UNIT

A block diagram of the Doppler unit is shown in fig. 6.2. The direction sensing circuit is based on quadrature phase demodulation. A 3 MHz crystal was chosen in order to permit the interrogation of deep vessels. The Doppler unit can be divided into 3 blocks: the transmitter, the receiver and the timer.
The transmitter was controlled by a standard Colpitts oscillator (Eimbinder, 1970). A field effect transistor (FET type TIS88) was used instead of the usual bipolar transistor because it had the advantage that the internal capacitances were less sensitive to the operating current bias and so it had a lower frequency drift with temperature. It also had a
good high-frequency performance and low noise characteristics. The inductor of the oscillator was adjustable so that it could be tuned in the range 2 to 4 MHz to accommodate different frequency probes. Other types of oscillators using logic devices (TTL and CMOS) were tried but rejected because of their higher power consumptions, larger sizes, and because they were not tuneable over a sufficiently wide range of frequencies.

The crystal was driven by a transformer-coupled transistor amplifier with a variable gain. Usually such amplifiers are tuned to the crystal frequency to reject harmonics; this was not necessary in this case since a pure (single frequency) sine wave (no harmonics) was produced by the Colpitts oscillator.

The transmitter also included two phase shifters which provided two sine waves in quadrature used by the demodulator for the separation of forward and reverse flow. These phase shifters consisted of a high-pass and a low-pass filter made using RC networks whose cut-off frequencies were identical and set to that of the oscillator, (see fig. 6.3a).

\[
V_R = \frac{j RC \omega}{1 + j RC \omega} \frac{V_{in}}{V_{in}}
\]

\[
V_C = \frac{1}{1 + j RC \omega} \frac{1}{V_{in}}
\]

At \( RC = 1/\omega \), \( V_R \) and \( V_C \) will have the same amplitude and will be shifted by 90°.

However in practice the two RC networks cannot be accommodated with the same RCs, therefore the two capacitors are chosen to be as close as possible to each other and the resistors can be chosen to be variable.

The introduction of a phase error \( \Delta \phi \) in both quadrature signals does not affect the output of the demodulators since
Fig. 6.3 Circuit diagram of the Doppler unit including the timing circuit.

(a) transmitter: T1, T2 and T3- TIS88A. T4 and T5- BC109. L- 40 turns 20 S.W.G close wound on 1.5 mm core (of a subminiature I.F transformer). Tr- 20:40 turns 35/36 SWG close wound (Mullard adjustable Vinkor LA 1375). (b) receiver: R=4.7kΩ. C=4.7nF. C1=2.2nF; all operational amplifiers are 1/2 TL072 surface mount. The second phase shifter has not been drawn for simplicity. (c) timer: IC1 and IC2- d.i.l reed relays, SJP1 and SJP2- 3 pole chassis sockets, SJP3- 2 pole chassis socket. (d) New circuit version.
Àcocoswt is of the same form as Acos(wt+θ).cos(wt+Δφ).

Because if wt is replaced by wt−Δφ,

Then

Acosh0. cos(wt−Δφ+θ)coswt
= Acos(wt+δ)coswt  where δ = θ−Δφ

The same can be applied to Acos(wt+δ).sinwt.

The quadrature signals were therefore ±45 degrees out of phase with the signal across the transmitting crystal itself, and avoided phase errors introduced by the crystal driving circuit. So that the output of the power amplifier driving the crystal could be adjusted without affecting the level of the quadrature signals, an FET follower was inserted between the oscillator and the power amplifier and another between the oscillator and the amplifier feeding the phase shifters. These two prevented the phase shifter interacting with the oscillator see fig. 6.3a.

6. 4. 2 RECEIVER

The Doppler-shifted received signal was applied directly to a broadband demodulator (type MC 1496) which had an excellent sensitivity (few femtovolts) and a good dynamic range (90 dB). The integrated circuit was a surface mount device (SMD).

The output of each filter was followed by a first-order bandpass π-network filter to eliminate the low-frequency signals caused by wall movements and the high-frequency signal of the carrier. The two audio signals were amplified and sent to a 90 degree phase shifter. Surprisingly perhaps, there is not a simple electronic device or a single channel device which gives a phase shift of 90 degrees with a constant gain over a range of frequencies. However a constant gain and a variable phase can be achieved using a single operational amplifier. With two such stages in parallel a constant gain over a wide range of frequencies can be obtained and by cascading several such stages, a constant 90 degree phase shift can be produced (Bedrosian, 1960; Lloyd, 1976). An eight-stage network was constructed to give a constant phase shift of 90 degrees over the frequency range 80 Hz to 14 kHz. Whilst the circuit is tedious to tune, since
it requires a total of sixteen stages (eight for each channel), it works well. The gain response of each phase shifter was flat (0 ± 0.1 dB) over a frequency range of 30 Hz to 20 kHz and the phase response is shown in fig. 6.4. It is important to note that, not only should the two outputs from each phase shifter be separated by 90 degrees, but that the corresponding outputs from the two phase shifters should also be in true quadrature; see fig. 6.2. In practice it is not possible to achieve a 90 degree phase shift for each network without introducing an error $\Delta \theta$ between the outputs of the two networks. For this reason a small phase correction $-\Delta \phi$ was introduced into one of the quadrature signals. This manoeuvre also introduced a gain error, but this was easily overcome by adjustment of the gain of the appropriate channel.

![Diagram of phase shifter](image)

Fig. 6.4 Phase responses of the two phase shifters.
An improved version of the phase shifter described above has been developed see fig. 6.3. d. this has the advantage of reducing the size and power consumption of the phase shifter by a factor of 2 and of eliminating completely the error 'ξ' see fig. 6.2 and therefore no measures have to be taken for correction. Fig. 6.3. d shows the complete modified version of the receiver.

In order to reduce the power requirements and size, the input matching transformer and the radio-frequency pre-amplifier were omitted and the probe matched directly to the demodulators. The high-order bandpass filters which attenuate the signals from the wall movement and the carrier frequency respectively, were each replaced by two first-order filters. The size was further reduced by using surface mount technology, as shown in fig. 6.5b.

6. 4. 3 TIMER

The purpose of the timer was to conserve both battery life and tape by activating the Doppler unit and the recorder for only 20 seconds at a time providing sufficient data to be meaningfully analyzed. This also reduced the time exposure of the patient to ultrasonic energy. The measurements could be automatically repeated every 2, 4, 8, 32, 64, or 128 minutes as selected. To achieve this a signal from a clock with a 1 minute period was divided progressively by a counter according to the interval selected. The output signal from the counter triggered a 20 second monostable which controlled a switching circuit. A push button was available to override the clock and manually start a 20s recording period. A two-position switch to bypass the timer and make the Doppler unit operate continuously was also incorporated. Because of the low-power available and because of the high transient current consumption of the timer-counter chip (4066), careful decoupling of the timer circuit was required to prevent harmonics of the clock signal appearing on the Doppler signal.
Fig. 6.5 Photograph showing the portable unit, (a) is the complete Doppler unit and tape recorder, (b) is the electronic circuit.
6.5 Doppler Signal Recorder

Various techniques for gathering the Doppler signal, such as radio telemetry were considered, but the method finally chosen was to record the (stereo) Doppler signal on a 'personal stereo'. This method offered flexibility and portability.

The most critical feature of the recorder was the frequency response. In order to determine the frequency response of the recorder the set-up shown in fig. 6.6 was used. A flat spectrum of white noise was injected into both inputs of the recorder. The recording used Dolby B-type noise reduction and with a high quality tape lasted 3 to 5 minutes in order to average the signal. Later the tape was played back to the spectrum analyzer (ONO-SOKKI CF-10 Dual channel FFT analyser) to extract the transfer function between the two channels and the power spectrum of each recorded signal.

Several recorders were tested but only one gave a sufficiently flat response between 60 Hz and 14 kHz. A 'Sony DC 3 Walkman' was selected which had the frequency response shown in fig. 6.6. This had a power consumption of 200 mA when recording, so that alkaline batteries lasted more than 5 hours and ensured that continuous recording was possible for at least one complete tape. This recorder offered an excellent signal-to-noise ratio (62dB) with Dolby B-type noise reduction. A further requirement was that the recorder remained in a recording mode when power was disconnected and subsequently restored by the timer circuit. The recorder chosen offered all these facilities and also incorporated a useful tape counter. We originally attempted to power the Walkman from the same batteries as the Doppler unit but noise, introduced by the brushes of the motor, corrupted the Doppler signals. The battery compartment of the recorder therefore had to be slightly modified so that it could be switched on and off by the timing circuit, this has also the advantage of saving space.

It was not possible to record the quadrature signals directly because of recording head skew ('static skew') and dynamic
interchannel displacement errors (Smallwood, 1985), and also because of slight gain differences between the two channels. It was therefore decided to further process the Doppler signals and record the true forward and the reverse flow Doppler signals on the two channels. Consideration was given to recording the heterodyne signal instead, but this was rejected for two reasons: firstly it halves the bandwidth available, and secondly dynamic errors still corrupt the heterodyne signal which would be required for the separation of the two channels.

6.6 PROBE DESIGN
The design of the probe was extremely important since optimum electrical and ultrasonic impedance matching was essential to make best use of the limited available power. Most of the design concentrated on the holder of the crystals. The probe described here, shown in fig. 6.7, was designed for the measurement of blood flow in femorodistal bypass grafts. It operated at 3 MHz with two crystals, one emitting and one receiving. The ultrasonic beam to 'vessel' angle was set to
45 degrees. The beams crossed at about 3 cm from the crystals for interrogating flow in grafts which are approximately 2 cm deep. The complete probe was flat and easily fixed to the skin above the graft and did not cause discomfort to the patient. To ensure the small crystals had the same resonant frequency, a single piezoelectric transducer (PZT) element was split into two.

The following manufacturing process was developed. The front faces of the two crystals were positioned by double-sided tape on a former made of Bakelite. Connections were soldered to the back of the crystals. Thereafter the assembly was placed in a mould and coated with silicone oil to promote mould release. Resin (low viscosity hardener and plasticised liquid epoxy resin supplied by CIBA-GEIGY) was then cast into the mould. When the resin was set, the mould and the former were removed and connections were made to the front faces. Finally a thin layer of air bubble free resin was deposited on the front of the crystals in order to improve the acoustic matching and to provide electrical insulation and mechanical protection.
Fig. 6.7 (a) A schematic diagram and (b) a photograph of the flat purpose built Doppler probe.
The beam profile (see fig. 6.8) and the power output of each probe have been obtained using a PVDF membrane hydrophone (produced by Marconi) to detect the ultrasound energy. The output of the hydrophone is plotted for different location across the beam and at a different distances from the probe.

The power output of each probe was determined before use. The probe/transmitter finally chosen for clinical use had an intensity output of 20 mWcm\(^{-2}\) SPTA.
6. 7 MEASUREMENT OF THE PHASE CHARACTERISTIC

Since the phase between the two channels is 90° ± δ, and δ is frequency dependant, this will have an effect on the cross talk level between the two channels. Fig. 6.9 illustrates the way in which the cross talk level versus frequency could be plotted. The transmitting oscillator is replaced by an external oscillator (oscillator 1) to produce the quadrature signals and at the same time to be used for demodulating the signal which simulates the Doppler shift produced by the oscillator 2. If the oscillator 2 has a frequency of 3 MHz + df (df > 0) this df will appear in the forward channel. However the output of the mixer will be |df| for 3 MHz ± df. The output signal from the mixer is then applied to a

![Functional diagram showing how the characteristic (cross-talk level, frequency) could be obtained.](image)

frequency to voltage converter in order to represent the frequency on the y-input of the plotter. The forward and the reverse channel outputs from the Doppler unit are fed to a
divider and finally the 20 \log(V_{FM}/V_{RM}) is calculated and applied to the x-input of the plotter.

6. 8 CALIBRATION OF THE DOPPLER UNIT
There are different methods for testing Doppler units, these are:
Electronic simulation method, flow rig method and flow phantom method.

6. 8. 1 ELECTRONIC SIMULATION
The testing by electronic simulation consists of applying to the receiver a signal of known frequency; knowing this frequency and transmitting frequency (which is used to demodulate the signal), then the difference between the two frequencies which correspond to the Doppler shift can be compared to the output of the Doppler unit. With this method the difficulties in reading the angle, and the velocity of sound, will be eliminated. Since only one frequency is present at a time (received frequency), it is very convenient to observe the output of the forward and the reverse channels and tune the processing circuit to minimize the cross-talk between the two channels.

6. 8. 2 FLOW RIG TECHNIQUE
This method, in which a moving liquid in a tube simulates the artery and blood flow, presents several problems. A whole spectrum of Doppler shift frequency is present due to the velocity profile inside the tube. It is difficult to know accurately the angle of insonation, and finally noise and vibration due to the hydraulic pump are picked-up by the Doppler flowmeter. A further major problem with this technique is that there is no other type of velocimeter which can easily be used as a reference.

6. 8. 3 FLOW PHANTOM TECHNIQUES
Flow phantom simulators do not necessary generate a flow of fluid, but can simulate a moving target. Flow phantoms have the advantage of simulating plug flow, that is a single velocity, which can be controlled precisely, but the angle
problem as well as the vibrations still persists in this case. Testing with this method does not require knowledge about the signal processing involved. In conclusion this technique is suitable for testing but not for calibration. Different techniques exist such as rotating disc described by McDicken (1983), oscillating pistons (Reid, 1983) and moving string described by Walker (1982).

In order to test the Doppler unit built, a moving string described by Walker) has been redesigned and constructed see Fig. 6.10. A circular belt with a known length, was driven by a DC motor. Each turn of the belt was detected by a fibre optic sensor (see Ch 3) immersed in water with the string which was marked by reflecting paint, in order to give a pulse at each turn. The time elapsed between two successive pulses given by the fibre optic sensor were measured by an
Fig. 6.10 (a) Moving string devices; (b) 'Velocity of the string measured by the ultrasonic Doppler unit- Motor input voltage' characteristic of the system.

oscilloscope. Fig. 6.10 b shows the relationship between the velocity of the belt and the velocity measured by the ultrasonic device described in this chapter.

6.  9 CONCLUSION
A compact battery powered portable bidirectional Doppler unit has been developed for use in ambulatory blood flow monitoring. Various 'conventional' elements of similar circuits have been omitted including the input transformer, the radio-frequency pre-amplifier and all high-order filters. As well as reducing the size, power consumption and cost, this also reduced some signal processing errors such as non-linearity, phase errors between the two channels and noise introduced by the active filters. The use of a small flat Doppler probe allows monitoring to be carried out for long periods of time without discomfort to the patient. Due to the shape of the probe, the vessel to probe angle is kept constant and therefore the measurements are reproducible.
CHAPTER 7

FIELD TRIAL

7.1 INTRODUCTION

In this chapter the clinical data obtained with the purpose built Doppler unit described in chapter 6 are presented. The data are divided into two groups, one group of data are taken from patients whilst they exercise for a few minutes and the second group of data are taken from patients during a period of approximately 24 hours postoperatively. This pilot study comprises eight patients, four from the first group and four from the second group. Although the population studied is very limited, the data obtained demonstrated the feasibility of the technique and gives a clear picture of the problems encountered. Solutions to these problems are proposed so that a study of a large population could be carried out.
7. 1. 2 ARTERIAL GRAFTS

For various reasons, especially because of arteriosclerotic disease, the arteries of the lower limb may become narrow and consequently the flow supplying the foot may be reduced. This will manifest itself by causing intermittent claudication, ischaemic rest pain or gangrene. In order to restore flow to the leg, the diseased segment of the artery may be bypassed either with a prosthetic graft or an autogenous vein. The latter material first used by Kunlin in 1949 is very popular. There are two ways of using a saphenous vein as a graft. The first method is to excise the vein, reverse it and suture it at both ends. This procedure has two limitations: firstly there is a size mismatch between the graft (ie the vein which is reversed) and the artery, and secondly the endothelium may be severely damaged. The second method is to use the vein in situ after disrupting the valve leaflets by the passage of a valvulometer. If this procedure is not carried out properly the valve leaflets can cause a problem. In the in situ case the endothelium is theoretically less disturbed, and the matching in size and shape will be more appropriate. This does not mean the artery and the graft are 'completely' matched since their anatomical structure (eg wall compliance) is not the same and also because of the effect of the suture at the anastomosis.

7. 1. 3 CAUSES OF GRAFT FAILURE

Graft failure is classified according to the time from operation. Three categories are recognised. Early failure generally happens during the 30 days following surgery, intermediate failure happens between 1 and 12 months and late failure occurs after 12 months. Early failure is generally due to technical errors (such as persistent valve leaflets, vein twist, or anastomotic problems), use of a poor quality vein, or poor runoff. Intermediate failure is largely due to the development of fibrous stenoses within the graft. Late failure is largely attributed to a progression of the atherosclerosis.
7. 1. 4 CAN GRAFT FAILURE BE PREDICTED?
The following methods of prediction of graft failure have been described by Beard 1987.

DIRECT

PREBYPASS
- Arteriography
- Doppler ultrasound
- Infusion resistance

POSTBYPASS
a. Haemodynamic
   - Blood flow
   - Peripheral resistance
   - Vascular modelling
b. Imaging
   - Arteriography
   - Ultrasound
   - Endoscopy

INDIRECT
- Doppler pressures
- Plethysmography
- Transcutaneous oxygen tension.

These methods can either provide information about the present status of the arterial segment or graft, or they may be used to assess the probability or the degree of success of surgery. In the case of abnormality at the time of measurement they may be used to predict that the graft is at risk of failure. But in the normal case these methods cannot predict success, because of other problems that the graft may experience, such as progression of the disease for instance. A successful bypass must be followed up closely so that any changes due to the development of stenosis can be detected and treated before the graft occludes.
7.2 CLINICAL TESTING

The patient measurements may be divided into two groups:
(i) Continuous measurements (Group 1. Case 1, 2, 3 and 4)
(ii) Periodic measurements (Group 2. Case 5, 6, 7 and 8)

7.2.1 METHOD

For both groups of measurements, a purpose built probe was taped over the patient's graft at an appropriate level. The tape recorder's recording level was adjusted so that even for large increases in velocity during activity, the Doppler signals would not saturate the recording circuits. The timer was set either to operate the system continuously (Continuous measurement), or for approximately 30 seconds every 30 min (Periodic measurement). At the beginning of the measurements data were taken with the patient at rest and in the supine position. The counter of the recorder was set to zero, so that the time at which the measurements had been taken could be determined retrospectively.
7.3 CASE STUDY

A. CONTINUOUS MEASUREMENTS (GROUP 1)

To investigate the effect of exercise and position on flow velocity in arterial grafts, recordings were made whilst the patients were lying down, sitting, standing and exercising on a treadmill at a speed of 2 miles per hour, with an incline of 10 per cent, for 2 min. Patients who were unable to use the treadmill were asked to walk around the laboratory.
7. 3. 1 CASE 1 (GROUP 1, CONTINUOUS MONITORING)
Age: 55
Sex: Male
Diabetic ?: Yes
Smoker ?: yes
Pre-operative pressures: Not recorded (patient claudicant)
Pre-operative Ankle Brachial Index (ABI): Not recorded
Post-operative pressure : PT 170  BP 155
Post-operative Ankle Brachial Index (PABI): > 1
Where BP (brachial pressure) and PT (posterior tibial pressure) are the best pressures recorded from the arm and the leg respectively.
Indication for surgery: Femoro-popliteal occlusion.
Type of graft: in situ vein graft below the knee.
The data obtained with this patient are shown in the graphs below.

Effect of exercise

![Graph showing effect of exercise](image_url)

(a)
Effect of Exercise

Fig. 7.1 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; extracted from the Doppler signal obtained with the patient standing at rest, and then following exercise.
7. 3. 2 CASE 2 (GROUP 1, CONTINUOUS MONITORING)
(This case shows the effect of posture)
Age: 85
Sex: Male
Diabetic ?: No
Smoker ?: No
Pre-operative pressures: PT 35 BP 160
Pre-operative Ankle Brachial Index (ABI): 0.22
Post-operative pressure : PT 170 BP 170
Post-operative Ankle Brachial Index (PABI): = 1
Indication for surgery: Rest pain and ulcer on toes.
Type of graft: *in situ* vein graft from the common femoral artery to the tibial artery.

Effect of posture

![Graph showing the effect of posture on MEAN IWMV and MEAN MAX VEL](image_url)
Fig. 7.2 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity extracted from the Doppler signal obtained with the patient standing at rest, and lying in the supine position.
Fig. 7.3 Shows the flow pattern in an abnormal case with a femorodistal \textit{in situ} vein graft tunnelled behind the tibia and anastomosed onto the anterior tibial artery. The measurements were taken 3 months after reconstructive surgery was performed. The graph shows a series of recordings from the grafts:

(a) With the patient at rest in a supine position, where the flow appeared to be hyperaemic. (b) After exercise with the patient in a standing position; the flow dropped dramatically from a mean velocity of approximately 7 cm/s to 2 cm/s. (c) Flow after 5 min of rest following exercise was still low. (d) The recording immediately after exercise with the patient in a supine position. The flow in this case returned to normal very quickly. Similar flow patterns were later observed using a duplex scanner.
Fig. 7.3 Sonograms recorded from an abnormal case, (a) at rest, (b) immediately after exercise, (c) 5 minutes after exercise and standing still, (d) immediately after exercise with the patient in the supine position.
7.3.3 CASE 3 (GROUP 1, CONTINUOUS MONITORING)

Age: 74
Sex: Male
Diabetic ?: No
Smoker ?: yes

Pre-operative pressures: Claudicant (pressure not recorded).
Pre-operative Ankle Brachial Index (ABI):
Post-operative pressure: PT 190 BP 180.
Post-operative Ankle Brachial Index (PABI): > 1.

Indication for surgery: Claudication and rest pain.
Type of graft: in situ vein graft below the knee popliteal.

---

![Graph showing MEAN IWMV and MEAN MAX VEL over time (min).](a)
Effect of Exercise

Fig. 7.4 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; extracted from the Doppler signal obtained with the patient standing at rest, and following exercise.
7.3.4 CASE 4 (GROUP 1, CONTINUOUS MONITORING)

Age: 64
Sex: Male
Diabetic ?: No
Smoker ?: Yes and still smoking
Pre-operative pressures: PT 70 BP 185
Pre-operative Ankle Brachial Index (ABI): <1
Post-operative pressure : PT 150 BP 144
Post-operative Ankle Brachial Index (PABI): >1
Indication for surgery: Rest pain.
Type of graft: in situ vein graft below knee popliteal.

Effect of Exercise

<table>
<thead>
<tr>
<th>Rest</th>
<th>After Exercise</th>
</tr>
</thead>
<tbody>
<tr>
<td>MEAN IWMV (cm/s)</td>
<td>MEAN MAX Vel (cm/s)</td>
</tr>
<tr>
<td>30</td>
<td>10</td>
</tr>
<tr>
<td>20</td>
<td>10</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
</tr>
</tbody>
</table>

Time (min)

(a)
Fig. 7.5 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; extracted from the Doppler signal obtained with the patient standing at rest, and following exercise.
COMMENTS

From case studies 1, 3, and 4 it can see that intensity weighed mean velocity, the maximum velocity, and heart rate increase with exercise; but the pulsatility index decreases, which is consistent with a hyperaemic response, then return to normal after a few minutes. However in case study 2, the intensity weighed mean velocity, and the maximum velocity increase when the patient is at rest in supine position; and the pulsatility index decreases.
B. PERIODIC MEASUREMENTS (GROUP 2)

With this method, it is hoped to investigate the feasibility of recording the blood velocity in grafts during an extended period of time (24 hours), using the system. With a 60 min tape, and each recording lasting 30s, we had the possibility of recording 120 events. If we wished these 120 events to be spread over 24 hours, the timer had to be set to 32 minutes. When data are lost, because of patient movement or for other reasons, the time axis will be labelled by the letter 'L' on the following figures. An example of sonogram obtained is shown in case study, 7.
7. 3. 5 CASE 5 (GROUP 2, PERIODIC MONITORING)
Age: 67
Sex: Male
Diabetic ?: Yes
Smoker ?: Yes
Pre-operative pressures: PT 90 BP 125
Pre-operative Ankle Brachial Index (ABI): 0.72
Post-operative pressure : PT 150 BP 130
Post-operative Ankle Brachial Index (PABI): > 1
Indication for surgery: Femoro-popliteal occlusion
Type of graft: in situ vein graft from femoral artery to below knee popliteal.
Recording date: Recordings were made at about more than three months after surgery.

'24 hour' post-operative recording

![Graph showing MEAN IWMV (cm/s) and MEAN MAX VEL (cm/s) over time (hours)]
Fig. 7.6 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; plotted against time in a patient who was able to move.
7. 3. 6 CASE 6 (GROUP 2, PERIODIC MONITORING)

Age: 81
Sex: Male
Diabetic ?: No
Smoker ?: Yes and still smoking

Pre-operative pressures: PT 60  BP 170
Pre-operative Ankle Brachial Index (ABI): 0.67
Post-operative pressure : PT 160  BP 140
Post-operative Ankle Brachial Index (PABI): > 1
Indication for surgery: Peroneal and Femoro-popliteal occlusion.

Type of graft: in situ vein graft.
Recording date: Recording were made at about three months after surgery.

(a)
'24 Hour' Post-operative recording

![Graph showing heart rate and pulsatility index over time.](Image)

- Pulsatility index
- Heart Rate

(b)

![Graph showing maximum intensity weighted mean velocity and maximum velocity over time.](Image)

- MAX MAX VEL
- MAX IWMV

(c)

Fig. 7.7 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; plotted against time in a patient who were able to move.
7. 3. 7 CASE 7 (GROUP 2, PERIODIC MONITORING)

Age: 87
Sex: Male
Diabetic ?: No
Smoker ?: No

Pre-operative pressures: PT 55  BP 210
Pre-operative Ankle Brachial Index (ABI): 0.26
Post-operative pressure: PT 175  BP 160
Post-operative Ankle Brachial Index (PABI): >1

Indication for surgery: Rest pain and gangrene.
Type of graft: Reversed vein.
Patient died of renal failure and the graft was patent at that time.
Recording date: The recording were made seven days after surgery.

"24 Hour" Post-operative recording

![Graph showing '24 Hour' Post-operative recording with MEAN IWMV and MEAN MAX VEL over time.](a)
Fig. 7.8 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; plotted against time in a patient who were confine to bed because of surgery.
Fig. 7.9 Sonograms recorded automatically from a femorodistal bypass graft at (a) the beginning and (b) the end of a 25 hour recording period.

(a)

(b)

Fig. 7.9 Shows sonograms extracted from the data; (a) at the beginning and (b) at the end of the 25 hours recording period, without readjusting the position of the probe.
7. 3. 8 CASE (GROUP 2, PERIODIC MONITORING)

Age: 70
Sex: Female
Diabetic ?: Yes
Smoker ?: Yes
Pre-operative pressures: Not recorded
Pre-operative Ankle Brachial Index (ABI):
Post-operative pressure: Not recorded
Post-operative Ankle Brachial Index (PABI):
Indication for surgery: Mixed vein-arterial disease with none healing ulcers.
Type of graft: in situ vein graft.
Recording date: The recording of data were made 24 hours after surgery.

'24 Hour' Post-operative recording

![Graph of MEAN IWMV (cm/s) and MEAN MAX VEL (cm/s) over time (hours)](a)
'24 Hour' Post-operative recording

Fig. 7.10 Shows (a) the mean intensity weighted mean velocity and the mean of the maximum velocity; (b) heart rate and pulsatility index; (c) the maximum of the intensity weighed mean velocity and the maximum of the maximum velocity; plotted against time in a patient who were confine to bed because of surgery.
COMMENTS
In studies 5 and 7 the data were recorded from patients who were confined to bed, and no data were lost during the monitoring period, however in study 6 and 8 the data were recorded from patients who were able to move and data were lost during periods of movement. There is no obvious correlation between the heart rate and the pulsatility index, but changes in the intensity weighed mean velocity and the maximum velocity follow the same pattern.
The table 7.1 During the early post-operative period, the variation of the IWMV, and the maximum velocity were higher than during the late post-operative period. The data were consistent with this expectation.

Table 7.1

<table>
<thead>
<tr>
<th>Case 8</th>
<th>Case 7</th>
<th>Case 6</th>
<th>Case 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Recording time following surgery</td>
<td>= 24 hours</td>
<td>= 7 days</td>
<td>= 3 months</td>
</tr>
<tr>
<td>Range of the maximum maximum velocity (cm/s)</td>
<td>31</td>
<td>51</td>
<td>15</td>
</tr>
<tr>
<td>Range of the mean maximum IWMV (cm/s)</td>
<td>21</td>
<td>32</td>
<td>6</td>
</tr>
<tr>
<td>Range of the maximum IWMV (cm/s)</td>
<td>26.4</td>
<td>28.2</td>
<td>8.9</td>
</tr>
<tr>
<td>Range of the mean IWMV (cm/s)</td>
<td>15.1</td>
<td>20.9</td>
<td>2.1</td>
</tr>
<tr>
<td>Range of the pulsatility index</td>
<td>6.02</td>
<td>4.95</td>
<td>5.38</td>
</tr>
</tbody>
</table>
7.4 DISCUSSION

It has been shown in this chapter that the Doppler unit described in chapter 6 could be used in a clinical environment. It was apparent from the sonograms obtained from the recorded signals, that the data collected from patients were of good quality (good signal-to-noise ratio) and reproducible, (only limited sonograms are shown). Intermittent recording over a long period of time can be made automatically without causing discomfort to the patients or confining them to bed. It has also been shown that blood velocity can be measured immediately after exercise, so that abnormalities such as that shown in case study 2 (the flow dropped significantly when the patient stood up) could be detected and measures taken when possible.

Data recorded in ambulatory patients and in different postures or in different physiological situations may also improve the understanding of blood flow in grafts. Although the population studied was very small it has been shown that posture may affect the blood flow in grafts. It has therefore been shown that the instrument and the technique are useful for studying and monitoring blood flow in grafts.
CHAPTER 8
SUMMARY AND CONCLUSIONS

The need to improve the understanding of the cardiovascular system has focussed attention on blood-flow measurements. Whilst techniques for such measurements have been investigated for over a century the recent introduction of ultrasound techniques has given an added impetus to this field as they are noninvasive, accurate and relatively simple to use. However in certain parts of the body ultrasound waves are highly attenuated by the presence of air (for example in the lung, or bones in the head and movement of the heart) makes blood flow measurement in the coronaries very difficult. Ultrasonic probes which can be inserted in the form of catheters overcome these problems but their minimum size is limited to few millimetres in diameter. Optical fibres however have the advantage of having very small diameters (few tens of microns), and are electrical insulators. These two advantages stimulated the present investigation of a new flow transducer using fibre optics, which has been described in chapter 3. The device has been designed, constructed and bench results have been given. The device is insensitive to fluctuations in pressure or ambient light. The electronic processing circuit with the associated fibre optics has an excellent frequency response for this application. The technical details are shown in chapter 3 and have been submitted for publication to the British Journal of Anaesthesia (Dahnoun et al 1990). The frequency response of the sensor which is located at the tip of the fibre is restricted because of the elasticity of the membrane. The sensitivity of the system was considerably improved by increasing the reflectivity of the membrane (this was achieved by stamping gold leaf onto the membrane at room temperature). The use of the current device is limited to bench tests or big vessels because of its size.

The second part of the work was based on noninvasive blood flow measurement in femorodistal bypass grafts; for this a
purpose-built device described in chapter 6 has been designed and constructed and the technical aspects of it have been published in Medical & Biological Engineering & Computing (Dahnoun et al 1990). This device differs from commercially available units in several ways: it has a low power consumption, and is compact and portable. Also various stages of the conventional processing circuit have been proved to be redundant and their elimination has had the advantage of diminishing phase errors, improving the signal to noise ratio and reducing the cost. The portable Doppler unit has also been interfaced to a standard personal stereo recorder for collecting the data from patients for continuous monitoring and retrospective analysis. Probes were purpose built, they maintain a constant angle between the Doppler beam and blood vessel, and the manufacturing process allowed control over the depth of the sensitive region. This facilitated measurements on grafts situated at different depths under the skin. These probes, with the portable directional unit gave reproducible data over a long period of time (see chapter 7).

The instrument was designed to make ambulatory measurements. Several patients have been monitored and the technique has proven successful in collecting data for long periods of time \((\approx 24\) hours). The data recorded from different patients show how blood velocity changes with activity. During exercise tissue movements which are picked-up by the transducer corrupted the Doppler signal arising from blood flow, but blood velocity can be measured immediately after cessation of exercise or movement. This pilot study on graft monitoring during activity suggests that there is information to be gained. For instance, in one case abnormal flow was seen in the graft after exercise and when standing but appeared normal when the patient lay down immediately after exercise or movement. Unfortunately the follow-up of this patient was impossible.

This study proves the feasibility of recording flow velocity for long periods of time, however it is difficult at this stage to extract any information on whether the graft is
going to fail or not. More patients will have to be monitored so that a meaningful conclusion can be reached.

FURTHER WORK
A great deal of work must be done before the fibre optic transducer can be of clinical use. Improvement of the sensing tip (such as increasing the elasticity and the reflectivity of the membrane) is essential since this will increase the sensitivity and reduce the diameter of the sensing tip. By using laser interferometry for detection of position of the membrane, the system could be more sensitive and dimensions could be considerably reduced. Solutions for the positioning of the sensor inside the vessel should also be considered.

At present the Doppler signals recorded from patients are analysed off-line. A bedside, portable spectrum analyser (based on a DSP chip and LCD display) incorporating the Doppler unit described could be of great value since the Doppler signal would be analysed on-line, and continuous data available to the surgeon.

To reduce the weight of the system, the portable recorder could be replaced by a telemetry system.

Movement artifacts which are of low frequency and large magnitude may be reduced by inserting high-order, high-pass filters between the probe and the demodulators. So the filters should be of crystal type and operating at very high frequency (the same as the frequency of the transmitter). Since the crystal filters are not tuneable, the Doppler unit would only be able to operate at a single frequency.
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Appendix 1

Relationship between pressure and flow in a rigid tube
(Basic considerations)
APPENDIX 1

The French physician J.L.M. Poiseuille (1799-1869) discovered that the relation between volume flow, \( Q \), and the drop in pressure \( \delta P \) under steady state conditions along a cylindrical tube of regular diameter, \( D \), and length, \( L \) (see fig. A1) has the form:

\[
K = \frac{\pi}{128\eta}
\]

And
\( \eta = \text{Viscosity of the fluid.} \)

A more realistic approach is that shown by Lamb, 1932, who derived the following equation, by taking into account the velocity profile.

\[
\frac{\delta^2 v}{\delta r^2} + \frac{1}{r} \cdot \frac{\delta v}{\delta r} + \frac{\delta P}{\eta L} = 0
\]  

Where
\( v: \text{velocity} \)
\( r: \text{radius} \)

Equation A. 2 has been used by Womersley (1955, 1957b).
Poiseuille's equation applies only when the flow is laminar and the fluid is Newtonian. This is a reasonable approximation in major arteries whose diameters are greater than 0.5mm (Whitmore, 1968; Johnson, 1969), the velocity close to the wall is nil, there is steady flow and the cross-section of the vessel is circular and the vessel walls are parallel and rigid.

Bernoulli's equation is based on the conservation of hydraulic energy in frictionless and incompressible fluid. The total energy per unit volume \((wT/V)\) at the site 1 is as follow:

\[
\frac{wT}{V} = P_1 + \frac{1}{2} \rho v_1^2 + \rho g h \tag{A.3}
\]

\(P_1\): Pressure
\(V\): Volume of the fluid involve
\(v_1\): Velocity of the fluid
\(\rho\): Density of the fluid
\(g\): Gravitational acceleration constant
\(h\): Height

The total energy per unit volume \((wT/V)\) at a site 2 will be of the same form as equation 3, and of the same magnitude, therefore

\[
P_1 - P_2 = \frac{1}{2} (v_2^2 - v_1^2) \rho \tag{A.4}
\]

Kaufmann, 1963 derived the 'general form of Bernoulli's equation, when acceleration of the flow take place (see fig. A.2)

\[
\frac{1}{2} v^2 + \frac{P}{\rho} + \int_0^S (dv/\delta t) \delta s = \text{Constant} \tag{A.5}
\]

\(\int_0^S (dv/\delta t) \delta s\): represents the pressure drop due to the acceleration.

Note that equation A.5, does not take into consideration the pressure gradient due to the viscosity.
If in one cardiac cycle it is assumed that $\int_0^2 (\delta v/\delta t) \cdot \delta s = 0$

Then $v_1^2 - v_2^2 = 2(P_2 - P_1) = 2\delta p$.

The volume flow across $S_1$ is equal to the volume flow across $S_2$ and $S_3$ therefore:

$S_1V_1 = S_2V_2 = S_3V_3$

therefore:

$V_1 < V_2$

And $V_3 < V_2$

then $P_2 < P_1$ and $P_3 > P_2$

This shows that the pressure drops across the narrowing, and two pressure gradients opposing each other. So in some circumstances disturbance is created.
Appendix 2

Program simulation of the fibre optic flowmeter
PROGRAM MAIN

C
C THIS PROGRAM CALCULATES THE AMOUNT OF LIGHT RETURNING TO A
C FIBRE OPTIC OF DIAMETER A. THE LIGHT SOURCE IS AT THE POINT
C (-L,0), AND IS SUPPOSED TO MAKE AN ANGLE THETA WITH THE X-AXIS
C ON ITS 'OUTWARD JOURNEY'. A REFLECTOR IS POSITIONED SO THAT
C ITS SURFACE PASSES THROUGH THE FIXED POINTS (0,-D/2) AND
C (0,D/2) AND ITS SHAPE IS GIVEN BY X=S(Y). THE
C CRITICAL VALUE OF THETA, THETA C, THAT FOR WHICH THE BEAM OF
C LIGHT RETURNS EXACTLY TO THE TOPMOST POINT OF THE FIBRE
C (-L,A/2), IS FOUND, AND THEN 2 INTEGRALS ARE CALCULATED TO
C MEASURE THE AMOUNT OF LIGHT RETURNING TO THE FIBRE AND THE
C TOTAL AMOUNT OF LIGHT EMANATING FROM THE FIBRE.
C
C THE CALLING TREE FOR THIS PROGRAM IS AS FOLLOWS -
C
C .FINDTH ...RETJOU ...S
C .
C MAIN ...DGAUSS ...GRAND ...TENSTY
C .
C .GRAPH ...GHOST
C
IMPLICIT DOUBLE PRECISION(A-H,O-Z)
COMMON D,RL,A,V,THICKN,PI,PIO2
EXTERNAL GRAND
PI=4.0*ATAN(1.0D0)
PIO2=PI*0.5
EPS=1.0D-4
WRITE(6,308)
308 FORMAT(  '  THIS VERSION USES THE MORE COMPLICATED EXPRESSION FOR',
' 1 ' DISPLACEMENT OF MEMBRANE'///' PLEASE TYPE THE VALUES OF D ',
' 2 '(DIAMETER OF MEMBRANE), ',
' 1 'L (DISTANCE'///' BETWEEN THE FIBRE AND THE MEMBRANE) AND A ',
' 2 '(DIAMETER OF FIBRE) -'
READ(5,*) D,RL,A
WRITE(6,300) D,RL,A
300 FORMAT('1INPUT VALUES ARE AS FOLLOWS -'///' D IS',T60,1PG14.6/
' 1 ' L IS',T60,G14.6/' A IS',T60,G14.6)
OPEN(7,FILE ='OUT.DAT',STATUS='NEW')
VBEG=0.0
VEND=1.0
VSTEP=0.01
WRITE(6,309) VBEG,VEND,VSTEP
309 FORMAT(' V CHANGING FROM',F7.3,' TO',F7.3,' IN STEPS OF',F7.3)
DO 55 V=VBEG,VEND,VSTEP
CALL FINDTH(THETAC)
WRITE(6,301) THETAC
301 FORMAT( 'CALCULATED VALUE OF CRITICAL ANGLE THETA C IS',T60,
' 1 1PG14.6)
RET=DGAUSS(GRAND,0.0D0,THETAC,EPS)
ALL=DGAUSS(GRAND,0.0D0,PIO2,EPS)
RATIO=RET/ALL
WRITE(6,302) RET,ALL,RATIO
302 FORMAT( '/ RETURNED LIGHT IS',T60,1PG14.6/' ALL AVAILABLE ',
' 1 'LIGHT IS',T60,G14.6/' RATIO OF RETURNED LIGHT TO ALL ',
' 2 'AVAILABLE LIGHT IS',T60,G14.6)
WRITE(7,*) V,RATIO
55 CONTINUE
CALL GRAPH(THETAC)
STOP
END

DOUBLE PRECISION FUNCTION S(Y)
IMPLICIT DOUBLE PRECISION(A-H,O-Z)
COMMON D,RL,A,V,THICKN,PI,PIO2

C THIS FUNCTION RETURNS S, WHICH DEFINES THE CURVED SURFACE
C AS A FUNCTION OF HEIGHT, Y, THICKNESS, THICKN, AND VELOCITY, V.
C DISPLACEMENT OF MEMBRANE SATISFIES S = 0 AT Y = D/2 AND Y = -D/2.
C V = 0 FOR A FLAT SURFACE.

X1=Y/D
X2=2.0/3.0
X3=4.0/3.0
IF(ABS(X1) .LE. 0.5) THEN
  S=-.25*((D/2)**X3)*(1-3.6*(X1**2)-3.2*(ABS(X1)**5))*(V**X2)
ELSE
  S=0.0
END IF
RETURN
END

SUBROUTINE RETJOU(THETA, YR,XI,YI)
IMPLICIT DOUBLE PRECISION(A-H,O-Z)
COMMON D,RL,A,V,THICKN,PI,PIO2

C THIS SUBROUTINE RETURNS IN YR THE HEIGHT OF THE BEAM ON ITS
C RETURN JOURNEY FROM THE REFLECTING SURFACE AT X = -L.
C THE BEAM STARTS ITS JOURNEY AT AN ANGLE OF THETA WITH THE
C X-AXIS.
C BEAM ON OUTWARD JOURNEY STARTS AT THE POINT X = -L, Y = 0 AT
C AN ANGLE THETA WITH THE X-AXIS.
C IT MEETS THE REFLECTOR WHEN THE 2 EQUATIONS
C Y = TAN(THETA) * (X + L), X = S(Y)
C ARE SATISFIED. THIS INTERSECTION POINT IS RETURNED AS (XI,YI).
C SOLVE THESE EQUATIONS BY NEWTON'S METHOD -
C SOLVE FOR Y. INITIAL GUESS IS Y = 0.

DELTAY=0.00001*D
T=TAN(THETA)
Y=0.0
DO 20 I=1,20
SY=S(Y)
ERROR=Y-(SY+RL)*T
WRITE(6,303) ERROR
303 FORMAT(' ERROR IN NEWTON SECTION IN YR',1PG14.6)
IF(ABS(ERROR) .LT. DELTAY) GO TO 30

DELTA Y = -(Y - (S(Y) + L) * TAN(THETA)) / (1 -
DR/DY*TAN(THETA)).

APPROXIMATE DERIVATIVE DS/DY BY DIFFERENCE.

DIFF=(S(Y+DELTAY)-S(Y))/DELTAY
DY=ERROR/(1.0-DIFF*T)
Y=Y+DY
20 CONTINUE
WRITE(6,300)
300 FORMAT( 'FAILED TO CONVERGE IN NEWTON SECTION OF YR.' )
STOP
30 YI=Y
XI=S(Y)

INTERSECTION IS AT THE POINT X = XI, Y = YI. FIND GRADIENT OF TANGENT TO REFLECTOR AT THIS POINT, BY MIDDLE DIFFERENCING.

TANGENT MAKES AN ANGLE OF PHI WITH X-AXIS.

DSDY=(S(Y+DELTAY)-S(Y-DELTAY))*0.5/DELTAY
IF(DSDY .EQ. 0.0) THEN

TANGENT IS PERPENDICULAR TO X-AXIS. CANNOT USE ATAN.

PHI=PI/2
ELSE

PHI=ATAN(1.0/DSDY)
END IF

BEAM MAKES AN ANGLE OF (PI - 2*PHI - THETA) WITH THE A-AXIS ON ITS RETURN JOURNEY. EQUATION OF REFLECTED BEAM IS

(Y - YI) = G * (X - XI)

SO BEAM PASSES THE PLANE X = -L WHEN

Y = YI - G * (L + XI).

G=TAN(2.0*PHI-PI-THETA)
YR=YI-G*(RL+XI)
WRITE(6,301) THETA,XI,YI,PHI,G,YR
301 FORMAT( 'LEAVING YR FOR THETA=',1PG14.6/' PT OF INTERSECTION IS',
1 '2G14.6/' PHI ='',G14.6,' G='',G14.6,' YR='',G14.6)
RETURN
END

SUBROUTINE FINDTH(THETAC)
IMPLICIT DOUBLE PRECISION(A-H,O-Z)
COMMON D,RL,A,V,THICKN,PI,PI/2


YR(THETA) - A/2 = 0.

SOLVE BY NEWTON'S METHOD. INITIAL GUESS IS THETA = 0.
DELTAT=0.0001
THETA=0.0
DO 20 I=1,20
CALL RETJOU(THETA, YRCURR, XI, YI)
ERROR=YRCURR-0.5*A
C WRITE(6,301) ERROR
301 FORMAT(' ERROR IN NEWTON SECTION IN FINDTH IS',1PG14.6)
IF(ABS(ERROR) .LT. DELTAT) GO TO 30
CALL RETJOU(THETA+DELTAT, YRPLUS, XI, YI)
DYRDTH=(YRPLUS-YRCURR)/DELTAT
DTHETA=ERROR/DYRDTH
THETA=THETA+DTHETA
20 CONTINUE
WRITE(6,300)
300 FORMAT(/' FAILED TO CONVERGE IN NEWTON SECTION IN FINDTH')
STOP
30 THETAC=THETA
RETURN
END
DOUBLE PRECISION FUNCTION TENSTY(THETA)
IMPLICIT DOUBLE PRECISION(A-H,O-Z)
COMMON D,RL,A,V,THICKN,PI,PI02
C
C THIS FUNCTION RETURNS THE VALUE OF LIGHT INTENSITY AS A FUNCTION
C OF ANGLE WITH THE X-AXIS, THETA.
C
C MAKE INTENSITY EQUAL I*COS[THETA]).
C AND I=1
I=1
TENSTY=I*COS(THETA)
RETURN
END
SUBROUTINE GRAPH(THETAC)
DOUBLE PRECISION THETAC,THETA,YR,XI,YI,D,RL,A,P,THICKN,PI,PI02
COMMON D,RL,A,V,THICKN,PI,PI02
C
C DRAW A PICTURE WITH SOME SAMPLE RAYS OF LIGHT BEING REFLECTED
C BACK. CHARACTERISTIC LENGTH IS THE BIGGEST OF L, A AND D.
C PLEASE NOTE THAT THE ONLY DOUBLE PRECISION VARIABLES HERE ARE
C THOSE PASSED THROUGH COMMON OR AS PARAMETERS, OR THOSE USED
C AS PARAMETERS IN CALLS TO OTHER ROUTINES. THE REST, USED IN
C CALLS TO GHOST ROUTINES, ARE SINGLE PRECISION.
C=MAX(RL,A,D)*1.1
CALL PAPER(1)
CALL PS空间(0.05,0.95,0.05,0.95)
C
C SET SINGLE PRECISION QUANTITIES ...
C
CO2=0.5*C
AO2=A*0.5
RLSNGL=RL
DO2=D*0.5
CALL MAP(-C,0.0,-C*0.5,C*0.5)
C
C DRAW AXES ...
C
CALL POSITN(-C,0.0)
CALL JOIN(0.0,0.0)
CALL POSITN(0.0,-CO2)
CALL JOIN(0.0,CO2)

C DRAW END OF LIGHT SOURCE ...
C
CALL POSITN(-C,AO2)
CALL JOIN(-RLSNGL,AO2)
CALL JOIN(-RLSNGL,-AO2)
CALL JOIN(-C,-AO2)
C
DRAW REFLECTOR ...
C
CALL POSITN(0.0,-DO2)
DO 10 I=-19,20
Y=D*I/40.0
X=S(Y)
CALL JOIN(X,Y)
10 CONTINUE
C
NOW DRAWN SAMPLE RAYS OF LIGHT FOR VARIOUS ANGLES DEPENDING
ON THETA C ...
C
DO 20 I=1,6
THETA=(-2.0+I*0.6)*THETAC
CALL RETJOU(THETA, YR,XI,YI)
YRSNGL=YR
XISNGL=XI
YISNGL=YI
CALL POSITN(-RLSNGL,0.0)
CALL JOIN(XISNGL,YISNGL)
CALL JOIN(-RL,YRSNGL)
20 CONTINUE
CALL GREND
RETURN
END
DOUBLE PRECISION FUNCTION GRAND(THETA)
IMPLICIT DOUBLE PRECISION(A-H,0-Z)
C
THIS FUNCTION RETURNS THE INTEGRAND OF THE REQUIRED INTEGRAL
TO MEASURE THE LIGHT FLUX.
C
COMMON D,RL,A,V,THICKN,PI,PIO2
GRAND=2.0*PI*SIN(THETA)*TENSTY(THETA)
RETURN
END
DOUBLE PRECISION FUNCTION DGAUSS(F,A,B,EPS)
IMPLICIT DOUBLE PRECISION(A-H,0-Z)
13 FEB 80.
C
INTEGRATION ALONG A LINE SEGMENT
C
DIMENSION W(12),X(12)
C
DATA CONST /1.0E-24/
C
DATAW(1),W(2),W(3),W(4),W(5),W(6),W(7),W(8),W(9),W(10),W(11),W(12)
1/0.10122 85362 90376, 0.22238 10344 53374, 0.31370 66458 77887,
2 0.36268 37833 78362, 0.02715 24594 11754, 0.06225 35239 38648,
3 0.09515 85116 82493, 0.12462 89712 55534, 0.14959 59888 16577,
4 0.16915 65193 95003, 0.18260 34150 44924, 0.18945 06104 55069/

C

DATA(X(1),X(2),X(3),X(4),X(5),X(6),X(7),X(8),X(9),X(10),X(11),X(12)
1/0.96028 98564 97536, 0.79666 64774 13627, 0.52553 24099 16329,
2 0.18343 46424 95650, 0.98940 09349 91650, 0.94457 50230 73233,
3 0.86563 12023 87832, 0.75540 44083 55003, 0.61787 62444 02644,
4 0.45801 67776 57227, 0.28160 35507 79259, 0.09501 25098 37637/

C

Y=B-A
DELS=CONSt*Y*Y
EPSS=EPS**2
DGAUSS=0.0
AA=A
5
Y=B-AA
IF(Y*Y .LE. DELS) RETURN
2 BB=AA+Y
C1=0.5*(AA+BB)
C2=C1-AA
S8=0.0
S16=0.0
DO 1 I = 1,4
U=X(I)*C2
1 S8=S8+W(I)*(F(Cl+U)+F(Cl-U))
DO 3 I = 5,12
U=X(I)*C2
3 S16=S16+W(I)*(F(Cl+U)+F(Cl-U))
S8=S8*C2
S16=S16*C2
S8=S16-S8
IF(S8*S8 .GT. EPSS*(1.0+S16*S16)) GO TO 4
DGAUSS=DGAUSS+S16
AA=BB
GO TO 5
4 Y=0.5*Y
IF(Y*Y .GT. DELS) GO TO 2
C
WRITE(6,7)
DGAUSS=0.0
RETURN
7 FORMAT(1X,36HGAUSS ... TOO HIGH ACCURACY REQUIRED)

C

END
Appendix. 3
List of Publications and Conferences.
REFEREED PAPERS AND CONFERENCE ABSTRACTS

A PORTABLE DIRECTIONAL ULTRASONIC DOPPLER BLOOD VELOCIMETER FOR AMBULATORY USE,
NAIM DAHNOUN, A. J. THRUSH, J. C. FOTHERGILL, D. H. EVANS
Medical & Biological Engineering & Computing. 28: 474-482.
(also published in initial form as Department of Engineering Report, 89/10)

AN AMBULATORY BLOOD VELOCITY MONITOR
NAIM DAHNOUN, A. J. THRUSH, J. C. FOTHERGILL, D. H. EVANS
Institute of Physical Sciences in Medicine.
'Trends in Physiological Ambulatory Monitoring.'
Royal Devon and Exeter Hospital, 18th October 1989.

AUTOMATIC INTERMITTENT DOPPLER RECORDING OF BLOOD FLOW IN FEMORODISTAL BYPASS GRAFTS.
NAIM DAHNOUN, A. J. THRUSH, J. C. FOTHERGILL, D. H. EVANS
Biological Engineering Society.
'Symposium on Occlusive Vascular Disease.'

A PORTABLE DOPPLER DIRECTIONAL UNIT
NAIM DAHNOUN, A. J. THRUSH, J. C. FOTHERGILL, D. H. EVANS
'HPA Trends Region Scientific & Business Meeting.'
Hospital Physicist Association.
Leicester, 8th November 1989.

AUTOMATIC INTERMITTENT DOPPLER RECORDING OF BLOOD VELOCITY IN FEMORODISTAL BYPASS GRAFTS
A. J. THRUSH, N DAHNOUN AND D. H. EVANS.
Proceedings of the British Medical Ultrasound Society.

AUTOMATIC DOPPLER RECORDING OF BLOOD VELOCITY CHANGES IN FEMORODISTAL BYPASS GRAFT IN THE EARLY POST-OPERATIVE PERIOD
A. J. TRUSH, N. DAHNOUN, J. BRENNAN AND D. H. EVANS.
Biological Engineering Society
'Blood Flow'90'.
Durham, 20th September 1990.
NEW METHOD FOR THE INTERMITTENT COLLECTION OF DOPPLER SIGNALS FROM ARTERIAL GRAFTS.
N. DAHNOUN AND D. H. EVANS
XII Brazilian Congress on Biomedical Engineering. Brazil 1990.

AN ACCURATE OPTICAL TECHNIQUE FOR THE MEASUREMENT OF THORACIC WALL MOVEMENT DURING HIGH FREQUENCY JET VENTILATION.
N. DAHNOUN, S. MOTTRAM, N. B. JONES, AND G. SMITH.
To be submitted to the: British journal of Anaesthesia.
Portable directional ultrasonic Doppler blood velocimeter for ambulatory use

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Abstract—The design of a portable directional Doppler velocimeter and of purpose-built probes intended for monitoring blood flow in femorodistal bypass grafts in ambulatory patients are described. The design is based on a continuous-wave Doppler technique with quadrature demodulation. The Doppler unit was interfaced to a‘personal-stereo’ recorder to store the Doppler signals. The advantages and the limitations of recording flow in ambulatory patients are discussed. Examples of blood flow data collected from femorodistal bypass grafts are given.

Keywords—Ambulatory, Blood flow, Bypass grafts, Doppler, Instrument, Patient monitoring, Ultrasound, Walkman


Introduction

Past few years have seen a large increase in the number of noninvasive blood-flow measurements made using Doppler ultrasound. Almost all these measurements have been made with the patient lying down, usually after a period of rest, in order to ‘standardise’ the measurements. Walking, for example, increases the blood flow to legs and may well reduce visceral blood flow, thereby giving non-reproducible data. There may, however, be considerable information contained in the way which flow in diseased arteries, and particularly in arterial changes with variables such as exercise, sleep, and eating. It was with this in mind that we set out to design a portable system to enable blood velocity tracings to be made automatically at predetermined times on ambulant patients.

The resulting unit differs from conventional Doppler both in its size and that it does not contain an input transformer, radiofrequency amplifier or high-order active filter. Velocities measured using this system may, if necessary, be converted to approximate volumetric flow rates by using an ultrasonic measurement of the vessel size and assuming that this does not change significantly over period of study.

Design and construction

The system comprises three separate parts: a continuous-wave Doppler unit (with an integral timing and a power supply), an audiocassette recorder (Sony Walkman), and a Doppler probe. Although a number of small lightweight Doppler units are commercially available, they are, in general, nondirectional, and so it was necessary to design and construct a special unit. Commercially available Doppler probes are also unsuitable for ambulatory monitoring, and so again a special probe had to be constructed.

Block diagram of the complete recording system

A block diagram of the complete recording system is shown in Fig. 1.

2.1 Doppler unit

Block diagrams of the Doppler unit and the circuit diagram are shown in Figs. 2 and 3, respectively. The direction-sensing circuit is based on quadrature phase demodulation. A 3 MHz transmission frequency was chosen to permit the interrogation of deep vessels. The Doppler unit can be divided into three blocks: the transmitter, the receiver and the timer.
2.1.1. Transmitter: The transmitter was controlled by a standard Colpitts oscillator (Eimhinder, 1970). A field-effect transistor (FET type TIS88) was used instead of the usual bipolar transistor because it had the advantage that the internal capacitances were less sensitive to the operating current bias and so it had a lower frequency drift with temperature. It also had a good high-frequency performance and low noise characteristics. The inductor of the oscillator was adjustable so that it could be tuned in the range 2-4 MHz to accommodate different frequency probes. Other types of oscillators using logic devices (TTL and CMOS) were tried but rejected because of their higher power consumptions, larger sizes, and because they were not tunable over a sufficiently wide range of frequencies.

The transducer was driven by a transformer-coupled transistor amplifier with a variable gain. The transformer permits impedance matching to improve power transfer to the transducer. Usually such amplifiers are tuned to the probe frequency to reject harmonics; this was not necessary in this case as a pure (single frequency) sine wave was produced by the Colpitts oscillator.

The transmitter also included two phase shifters which provided two sine waves in quadrature used by the demodulator for the separation of forward and reverse flow. These phase shifters consisted of a high-pass and a low-pass filter made using RC networks whose cutoff frequencies were identical and set to that of the oscillator. The quadrature signals were therefore ±45° out of phase with the signal across the transmitting element, and avoided phase errors introduced by the driving circuit. An FET follower was inserted between the oscillator and the power amplifier so that the amplifier output could be adjusted without affecting the level of the quadrature signals. Another follower was inserted between the oscillator and the amplifier feeding the phase shifters, to prevent the phase shifter interacting with the oscillator.

2.1.2. Receiver: The Doppler-shifted received signal was applied directly to a broadband demodulator (type MC 1496) which had an excellent sensitivity (few femtovolts) and a good dynamic range (90 dB). The integrated circuit was a surface-mount device (SMD). The
output of each filter was followed by a first-order bandpass network filter to eliminate the low-frequency signals caused by wall movements and the high-frequency signal from the carrier.

It was not possible to record the quadrature signals directly because of recording head skew ('static skew') and dynamic interchannel displacement errors (Smallwood, 1985), and also because of slight gain differences between the two channels. It was therefore decided to further process the Doppler signals and record the true forward and reverse flow signals on the two channels. Consideration was given to recording the heterodyne signal instead, but this was rejected for two reasons; first it halves the bandwidth available, and secondly dynamic errors still impair the heterodyne signal, which would be required for separation of the two channels.

The two audio signals were amplified and sent to a 90° phase shifter. Surprisingly, perhaps, there is not a simple electronic device which gives a phase shift of 90° with a constant gain over a range of frequencies. However, a constant gain and a variable phase can be achieved using an angle operational amplifier. With two such stages in parallel a constant gain over a wide range of frequencies can be obtained, and by cascading several such stages, a constant 90° phase shift can be produced (Bedrosian, 1960; Lloyd, 1976). An eight-stage network was constructed to give a constant phase shift of 90° over the frequency range 80 Hz to 14 kHz. Although the circuit is tedious to tune, because it requires a total of 16 stages (eight for each channel), it works well. The gain response of each phase shifter was flat (0 ± 0.1 dB) over a frequency range of 30 Hz to 15 kHz. The phase response is shown in Fig. 4.

It is important to note that not only should the two outputs from each phase shifter be separated by 90°, but also the corresponding outputs from the two phase shifters should also be in true quadrature (Fig. 2). In practice it is not possible to achieve a 90° phase shift for each network without introducing an error $\Delta \theta$ between the outputs of the two networks. For this reason a small phase correction - $\Delta \phi$ was introduced into one of the quadrature signals. This manoeuvre also introduced a gain error, but this was easily overcome by adjustment of the gain of the appropriate channel.

To reduce the power requirements and size, the input-matching transformer and the radiofrequency preamplifier were omitted and the probe matched directly to the demodulators by making the input impedance of the demodulators similar to that of the probe. An initial prototype used high-order bandpass filters to attenuate the signal from wall movement and the carrier frequency, but these were later replaced by passive high-pass and low-pass filters. This improved the signal-to-noise ratio and also reduced power consumption and size. The size was further reduced by using surface-mount technology, as shown in Fig. 5b.

Because the resonance frequency of the transducer elements was not exactly 3 MHz, the master oscillator was tuned until the maximum output from the transmitting element was obtained. Thereafter the transmitting element was replaced by a 50Ω resistive load and the receiving
element was replaced by a precision radiofrequency signal generator with the central frequency set to that of the master oscillator of the Doppler unit, by looking at the forward and reverse output of the Doppler unit. When the two frequencies were equal both outputs were nil. The signal generator was then shifted to a higher or a lower frequency to simulate the Doppler shift. The higher frequencies give an output on the forward channel and the lower frequencies give an output on the reverse channel. With the same setting the error $\Delta \theta$ was minimised by correction of the quadrature phase $(90^\circ + \Delta \phi)$ until the crosstalk between the two channels was less than $-30\,\text{dB}$.

2.1.3. Timer: The purpose of the timer was to conserve both battery life and tape by activating the Doppler unit and the recorder for 20 s at a time, providing sufficient data to be meaningfully analysed. This also reduced the time exposure of the patient to ultrasonic energy. The measurements could be automatically repeated every 2, 4, 8, 32, 64 or 128 min as selected. To achieve this a signal from a clock with a 1 min period was divided progressively by a counter according to the interval selected. The output signal from the counter triggered a 20 s monostable which controlled a switching circuit. A pushbutton was available to override the clock and manually start a 20 s recording period. A two-position switch to bypass the timer and make the Doppler unit operate continuously was also incorporated. Because of the low power available and because of the high transient current consumption of the timer-counter chip (4066), careful decoupling of the timer circuit was required to prevent harmonics of the clock signal appearing on the Doppler signal.

2.2. Doppler signal recorder

Various techniques for gathering the Doppler signal, such as radio telemetry, were considered, but the method

![Diagram of phase responses of the two phase shifters](image)

![Photograph showing the portable unit](image)
ally chosen was to record the (stereo) Doppler signal on personal stereo. This method offered flexibility and por-
tability.

The most critical feature of the recorder was the frequency response. Several recorders were tested but only one gave a sufficiently flat response between 60 Hz and 20 kHz. A Sony DC 3 Walkman, which had the frequency response shown in Fig. 6, was selected. It had a power consumption of 200 mA when recording, so that alkaline batteries lasted more than 5 hours and ensured that continuous recording was possible for at least one complete tape. This recorder also offered an excellent signal-to-noise ratio (62 dB) with Dolby B-type noise reduction. A further requirement was that the recorder remained in a recording mode when power was disconnected and subsequently re­stored by the timer circuit. The recorder chosen offered all use facilities and also incorporated a useful tape counter.

I originally attempted to power the Walkman from the batteries as the Doppler unit, but noise, introduced by the brushes of the motor, corrupted the Doppler signals. The battery compartment of the recorder therefore had to be slightly modified so that it could be switched on and off by the timing circuit; this also has the advantage of saving space.

3 Probe design

The design of the probe was extremely important since optimum electrical and ultrasonic impedance matching is essential to make best use of the limited available power. Most of the design was concentrated on the holder (the probe elements). The probe, shown in Fig. 7, was designed for the measurement of blood flow in femorodistal bypass grafts. It operated at 3 MHz with two elements, one emitting and one receiving. The ultrasonic beam to vessel angle was set to 45°. The beams crossed at about 2 cm from the probe elements for interrogating flow in grafts which are approximately 2 cm deep. The complete probe was flat and easily fixed to the skin above the graft, and did not cause discomfort to the patient. To ensure that all elements had the same resonant frequency, a single piezoelectric transducer (PZT) element was split into two.

The following manufacturing process was developed. The front faces of the two elements were positioned by double-sided tape on a former made of Bakelite. Connections were soldered to the back of the elements. Thereafter the assembly was placed in a mould and coated with silicone oil to promote mould release. Resin (low-viscosity hardener and plasticised liquid epoxy resin supplied by Ciba-Geigy) was then cast into the mould. When the resin was set, the mould and the former were removed and connections were made to the front faces. Finally, a thin flat layer of air-bubble-free resin was painted over the front of the elements to improve the acoustic matching and to provide electrical insulation and mechanical protection. It proved difficult to accurately control the thickness of this layer because of the probe shape. The beam shape and the power output of each probe were determined before use. The probe/transmitter finally chosen for clinical use had an intensity output of 20 mW cm⁻² SPTA as measured by a 0.5 mm diameter membrane hydrophone.

3 Clinical study

3.1 Immediately after exercise

Flow changes after exercise were monitored in 10 patients with femorodistal bypass grafts. The probe was attached over the graft at the mid-thigh level using a double-sided adhesive disk and adhesive tape. Recordings were taken with the patient standing, then, with the probe left in place, the patient was asked to walk on a treadmill at a speed of 2 miles an hour with an incline of 10 per cent, for 2 min. Patients who were unable to use the treadmill were asked to walk around the laboratory. The recording was started during exercise so that information was not lost; however, only the information gained once the patient stopped exercising was useful because of high-level signals from tissue movement during the exercise phase.

3.2 Periodic postoperative monitoring

Four patients with femorodistal bypass grafts were monitored for a period of 24 h within a few days of their oper-
The probe was taped over the graft at mid-thigh level. The unit was switched on for 30 s every 32 min using an automatic timer. As the unit is small it allows the patient to be mobile during the period of monitoring. Monitoring in this manner enabled trends, such as hyperemic behaviour, to be studied, and irregularities to be detected.

**Signal analysis**

The signals recorded using these different methods were analysed using the spectrum-analyser system described by KINDEWEIN et al. (1988). The system produces sonograms in real time showing both forward and reverse flow. The maximum frequency envelope and the intensity-weighted mean frequency are also extracted and exhibited in real time. Signals with spurious noise due to probe movement, and noise introduced by the switching of the recorder, were rejected. The signals were then characterised in terms of their pulsatility index (PI) (GOSLING and NG, 1974), the maximum velocity and the intensity-weighted mean velocity (IWMV).

**Clinical results**

Fig. 8 shows (a) typical sonogram recorded from an in situ vein graft with the patient at rest, (b) a typical sonogram recorded from in situ vein graft with the patient resting immediately after exercise, (c) triphasic flow recorded from an in situ vein graft and showing the directionality of the unit. Fig. 9 shows intensity-weighted mean (IM) velocity, maximum velocity and PI plotted against time in a patient at rest and then after exercise. Ten seconds of data were extracted every 15 s from the continuous recording to monitor the flow changes. In normal grafts a hyperaemic flow pattern was observed after exercise, and the flow quickly reverted to the normal flow pattern within less than 5 min.

**Fig. 9** PI, maximum velocity and IWM velocity extracted from the Doppler signal obtained with the patient standing at rest, and then following exercise O-A resting, A-B exercising, B-standing

Fig. 10 shows an abnormal case with a femorodistal in situ vein graft tunnelled behind the tibia and anastomosed onto the anterior tibial artery. The measurements were taken 3 months after the reconstructive surgery was performed. The graph shows a series of recordings from the graft, (a) with the patient at rest in a supine position, where the flow appeared to be hyperaemic, (b) after exercise with the patient in a standing position, the flow dropped dramatically, from a mean velocity of approximately 7 cm s\(^{-1}\) to 2 cm s\(^{-1}\), (c) flow after 5 min of rest following exercise was still low, (d) the recording obtained immediately after exercise with the patient in the supine position. The flow in this case returned to normal very quickly. Similar flow patterns were later observed using a duplex scanner.

**Fig. 11** shows IWM velocity, maximum velocity and PI plotted against time for a recording taken from a femorodistal in situ vein graft over a 25 h period, 10 days after surgery. This represents 25 min of data in total. The values were obtained by averaging 20 s of data from each 30 s recording, taken each 30 min. The significance of some of the dramatic changes in velocity is not yet understood. Fig. 12 shows sonograms extracted from the data (a) at the beginning and (b) at the end of the 25 h recording period.
Discussion

A compact battery-powered portable bidirectional Doppler unit has been developed for use in ambulatory blood flow monitoring. Various 'conventional' elements of similar circuits have been omitted including the input transformer, the radiofrequency preamplifier and all high-order filters. As well as reducing the size, power consumption and cost, this also reduced some signal processing errors such as nonlinearity, phase errors between the two channels and noise introduced by the operational amplifiers of active filters. The use of a small flat Doppler probe allows monitoring to be carried out for long periods of time without discomfort to the patient. Owing to the shape of the probe, the vessel to probe angle is kept constant and therefore the measurements are reproducible.

It has been shown (Sumner and Strandness, 1969; Yao, 1970) that post-exercise flow measurement can be of value in the assessment of patients with peripheral vascular disease. Using conventional techniques, however, it can be quite difficult to make satisfactory measurements immediately after the cessation of exercise, because of the time needed to get a patient to a couch and locate the arterial signal. It has been suggested (Burnham et al., 1980; McShane et al., 1988) that flexion of the knee may cause a kinking of certain types of graft, that could lead to early failure. Once again it is difficult using standard duplex techniques to monitor the changes in blood flow that this may cause. The Doppler unit described in this paper

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ne that the measurement of velocity (and hence flow) both immediately post-exercise and with the patient in any other posture is a simple matter.

The instrument has been used to make ambulatory measurements, but it has been found not possible to record in the lower limbs actually during exercise because of relative movement of the probe and the artery. Flow measurements can, however, easily be taken during short interruptions to the exercise and in various clinical situations and postures.

Fig. 10 shows how posture can affect blood flow. In this study, an abnormal flow was seen in the graft after exercise when the patient was standing but appeared normal if the patient lay down immediately after exercise. With this normal case it can be seen that more information can be obtained when recordings are made from patients in different postures after exercise.

The device has been used for 24 h blood-flow monitoring in femorodistal bypass grafts after surgery. It may be important to monitor the flow pattern in grafts during the recovery period (the first 30 days after the operation) to assess the high risk of graft failure.

Further work in siture surveillance of grafts is based on clinical observations which are subjective in nature, for example the appearance of the limb, its temperature, absence of graft and pulses, and increase in rest pain. It would be useful to implement a system which could objectively detect the change from normal to abnormal flow in a graft and subsequently trigger an alarm and switch the Doppler unit to continuous recording mode. Further studies are also needed to determine the usefulness of this technique in storing long-term changes in flow in grafts, including effect of exercise.

Conclusions A simple, relatively inexpensive, portable, directional Doppler velocimeter with a small flat probe that can be taped to the skin over the appropriate vessel, has been developed. This portable Doppler velocimeter unit which was faced to a standard portable recorder has been used to monitor blood velocity in situ vein grafts in 14 patients. Although the device may provide useful information for assessing arterial disease or for predicting arterial failure, it has been found difficult to make measurements in the lower limbs of exercising subjects.

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References


Authors' biographies

Naïm Dahnoun was born in Oujda, Morocco, in 1937. He obtained his Ingéniorat d'Etat in Electronic Engineering from the University of Science & Technology of Oran, Algeria, in 1983. From 1983 to 1985 he underwent military service as an electronics designer with the Algerian Air Force. Since December 1986 he has been studying at the University of Leicester, where he is currently finalising his doctoral research in blood flow measurements. His research interests include signal processing and biomedical instrumentation.

Abigail J. Thrush was born in The Hague, The Netherlands in 1961. She obtained a B.Sc. in Physics from Sussex University in 1982 and an M.Sc in Medical Physics from Surrey University in 1983. She worked in the Medical Physics Department in North Wales for three years before moving to Leicester Royal Infirmary where she is currently involved in research work on blood flow measurements using Doppler ultrasound.

John C. Fothergill was born in Malta in 1953. He obtained his B.Sc in Electronic Engineering from the University College of North Wales (Bangor) in 1975, an M.Sc in Electrical Materials & Devices in 1976 and a Ph.D in 1979 for work on the electronic properties of biopolymers, also carried out at UCNW, Bangor. He then investigated electrical breakdown and degradation phenomena in high-voltage power cables at STL Ltd. (now STC Technology), Harlow, Essex. In 1984 he became a lecturer in dielectrics and bioelectronics.

David H. Evans was born in Cheshire, UK, in 1950. He received his B.Sc in Physics from Surrey University in 1972, and his Ph.D from Leicester University in 1979 for his studies on the haemodynamics of peripheral vascular disease. Between 1980 and 1982 he was a Visiting Professor of Biomedical Engineering at the Federal University of Rio de Janeiro, and is currently head of Clinical Measurement in the Leicester District Medical Physics Department. His research interests are varied and include haemodynamics, Doppler ultrasound and manometric techniques. Dr Evans was elected a Fellow of the Institute of Physics in 1984.
NEW METHOD FOR THE INTERMITTANT COLLECTION
OF DOPPLER SIGNALS

by

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ABSTRACT — For various reasons, especially arteriosclerosis, the arteries of the lower limb become narrow and consequently the blood flow supplying the foot may be reduced. This will manifest itself by causing intermittent claudication, ischaemic rest pain or gangrene. In order to restore flow to the leg, the diseased segment of the artery may be bypassed either with a prosthetic graft or an autogenous vein. However these grafts are at high risk of failure, some fail within a few days after surgery and other after an undetermined period of time. In order to attempt to predict graft failure, we present a new method based on the recording of blood velocity postoperatively and during long period of time, by a purpose built Doppler ultrasonic unit and probe, in ambulatory patients.

Key words - Ambulatory, Blood flow, Bypass grafts, instrumentation, Patient monitoring, Ultrasound.

INTRODUCTION

Different methods for the prediction of graft failure have been used these include arteriography, conventional Doppler ultrasound, peripheral resistance and Doppler pressures, but all these methods can either provide information about the present status of the arterial segment or graft, or they may be used to assess the probability or the degree of success of surgery. In the case of abnormality at the time of measurement, they may be used to predict that the graft is at risk of failure. But in the normal case these methods cannot predict success, because of other problems that the graft may experience, such as progression of the disease for

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instance. Therefore a successful bypass must be followed up closely so that any changes due to the development of stenosis can be detected and treated before the graft occludes. There is evidence that the presence of disease in the arterial system can cause fluid-mechanic disorders, including reduction of pressure and volumetric flow rate, and an increase in the local flow velocity. From these different parameters involved in the haemodynamic changes, we have decided to monitor the flow velocity in femorodistal bypass grafts in patients under different physiological conditions such as when exercising or in different position, and during a long period of time so that changes can be detected. To achieve this, we have first to prove the feasibility of the technique, and this constitute the main part of this paper.

**INSTRUMENTATION**

The design and construction of a portable Doppler velocimeter and purpose-built probes intended for monitoring blood flow in femorodistal bypass grafts in ambulatory patients has been found necessary, since there are none commercially available.

**Design criteria**

The Doppler unit has to have:
1. Large bandwidth (70Hz- 12kHz)
2. Directionality
3. Low noise figure
4. Tuneable to probe frequency between 2 and 4 MHz.
5. Compact, Low power consumption, batteries powered and portable.

To achieve all these requirements, the input matching transformer and the radio frequency preamplifier were omitted and the probe matched directly to the demodulators, the high-order bandpass filters which attenuate the signals from the wall movement and the carrier frequency respectively, were each replaced by two first-order filters, and the size was further reduced by using surface mount technology (Dahnoun et al 1990).

A purpose built timer was also incorporated in order to conserve both battery life and tape by activating the Doppler unit and the recorder for only 20 seconds at the time providing sufficient data to meaningfully analysed.

**Doppler signal recording and analysis**

For gathering the Doppler signal, a personal 'stereo' recorder was chosen (Sonny DC 3 Walkman), it has a good frequency response (60Hz to 14kHz) and an excellent signal to noise ratio (62 dB), with Dolby B-Type noise reduction.

The Doppler signal gathered were analysed using the spectrum-analyser system described by Schlindwein et al (1988). the system produces sonograms in real time showing
both forward and reverse flow. The maximum frequency envelope and the intensity-weighted mean frequency are also extracted in real time.

CLINICAL TESTING

After different subjective tests of the portable system, it was decided to proceed with measurements on patients, for this the measurements were divided into two groups:
(i) continuous measurements
(ii) periodic measurements

Method

A purpose built probe, was taped over the patient's graft at an appropriate level, in order to avoid discomfort to the patient. The recording level (on the tape recorder) was then adjusted so that even for a large increase in the velocity during activity, Doppler signals do not saturate the recorder's circuit. The timer was set to operate the system continuously or for approximately 30 seconds every 30 min. At the beginning of the measurements data were taken with the patient at rest and in the supine position. The counter of the recorder was set to zero, so that, the time at which the measurements had been taken could be determined retrospectively.

Continuous measurements — To investigate the effect of exercise and position on flow velocity in arterial grafts, recordings were made whilst the patients were lying down, sitting, standing and exercising on a treadmill at a speed of 2 miles per hour, with an incline of 10 per cent, for 2 min. Patients who were unable to use the treadmill were asked to walk around the laboratory.

Fig. 1 shows the intensity-weighted mean velocity (IWMV), maximum velocity and PI plotted against time in a patient at rest and then immediately after exercise. Ten seconds of data were extracted every 15s from the continuous recording to monitor the flow changes. In normal grafts a hyperaemic flow pattern was observed after exercise, and the flow quickly reverted to a normal flow pattern within less than 5 min.

Fig. 2 shows the flow pattern in an abnormal case with a femorodistal in situ vein graft tunnelled behind the tibia and anastomosed onto the anterior tibial artery. The measurements were taken 3 months after the reconstructive surgery was performed. The graph shows a series of recordings from the grafts:
(a) With the patient at rest in a supine position, where the flow appeared to be hyperaemic. (b) After exercise with the patient in a standing position, the flow dropped dramatically, from a mean velocity of approximately 7 cm/s to 2 cm/s. (c) Flow after 5 min of rest following exercise was
still low. (d) The recording immediately after exercise with patient in supine position. The flow in this case returned to normal very quickly. Similar flow patterns were later observed using a duplex scanner.

**Periodic measurements**

With this method, we hope to investigate the feasibility of recording blood velocity in grafts during a long period of time (24 hours), by using the system. With a tape of 60 min, and each recording lasted 30 s, we have the possibility of recording 120 events, however we also wanted these 120 events to be spread over 24 hours, therefore the timer had to be set at 32 minutes.

**CONCLUSION**

A number of patients have been monitored with this method and we were able to record Doppler signals for periods of 24 hours or more. However it has been found difficult to record Doppler signals from patients actually during activity because of the relative movement of the probe and tissue, but the recording immediately after exercise were possible.

**REFERENCES**


AN ACCURATE OPTICAL TECHNIQUE FOR THE MEASUREMENT OF THORACIC WALL MOVEMENT DURING HIGH FREQUENCY JET VENTILATION.

Running Head: Thoracic wall movement

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SUMMARY

Instrumentation for use in High Frequency Jet Ventilation is not currently capable of providing information suitable for predicting gas exchange with differing jet parameters. Whilst techniques for the accurate measurement of tidal volume during jet ventilation are now available, the relationship between tidal volume and the efficiency of gas exchange varies with frequency and inspiratory/expiratory time (I:E) ratio. We report technical details on a new non-invasive instrument capable of accurately measuring displacement of the thoracic wall at high frequencies. It is hoped that use of this instrument will allow the calculation of an index of efficiency for the transduction of airway pressure changes into peripheral expansion. It is hoped that this index may be of value in predicting the gas exchange response to changes in jet parameters.
High Frequency Jet Ventilation (HFJV) remains a rarely used technique in general clinical practice, despite a number of advantages over conventional ventilation in some patient groups. One reason for the infrequent use of HFJV is the relative difficulty in predicting the effect of any changes in ventilation parameters on gas exchange. This difficulty arises because of a lack of understanding of the basic physical principles by which the technique operates. Measurements of tidal volume ($V_T$) during HFJV have shown a complex relationship with blood gas status [1,2], indicating that $V_T$ may be of relatively little use as a predictive measurement in clinical practice. Studies in the pig by Lin et al [3] have shown that the magnitude of the transfer function between airway pressure and thoracic wall expansion measured with a strain gauge shows a strong correlation with changes in $P_{aO_2}$ and $P_{aCO_2}$. Whilst this approach may be valuable, the use of a strain gauge attached to an elastic belt wrapped around the circumference of the thorax has theoretical and experimental disadvantages.

Because a strain gauge normally measures changes in the force applied to it with minimal variation in dimensions, it must be attached to the chest wall through some elastic medium, so as not to impede inspiration. In order to achieve this, the ends of the strain gauge sensor are attached to an elastic fabric strap which is wrapped around the circumference of the thorax. By Hooke's law, the force acting on the ends of the strap would be proportional to the length of the strap, and hence the circumference of the thorax. Given that the electrical output of the strain gauge is likely to be relatively linearly related to the force acting on the sensor, three major sources of error may be encountered when this technique is used to measure thoracic wall movements during HFJV. Firstly, in using Hooke's law, the assumption is made that the strap does not reach its elastic limit in the range of movement to be measured. This will depend on the resting tension in the strap, which is difficult to control, since it will vary between experimenters and with changes in FRC. Secondly, the linearity of the relationship between force and length depends on the assumption that there are no frictional forces between the elastic strap and the thoracic wall.
These forces are clearly present, varying with the surface properties of the strap and skin, and with the resting tension within the strap. Finally, at frequencies encountered in HFJV, the frequency response characteristics of the strap would be expected to affect the forces applied to the strain gauge. These characteristics would also be expected to be affected by the resting tension in the strap.

The above criticisms are not exclusive to the use of a strain gauge, but also apply to any other technique in which a circumferential belt is used (e.g. Linear voltage transformer or Respitrace). In order to overcome these criticisms, a contact-free transducer system has been developed which uses optical techniques to measure displacement of the chest wall at a single point, and which has a flat frequency response over the range used in existing high frequency ventilation techniques. We present data comparing the frequency response of this system with a standard Lectromed strain gauge and strap combination.

INSTRUMENT AND METHODS

Instrument design

The optical transducer described consists of four main elements; a reflector mounted on the subjects thorax, a transmitter of amplitude modulated infra-red radiation, a bi-directional fibre-optic light guide mounted on a rigid stand a few centimetres above the reflector, and an infra-red detector (fig. 1). In the authors' instrument, the transmitter and receiver electronics were contained within a single pocket sized unit. The fibre optic light guide consisted of a pair of 1mm diameter plastic step-index fibres 2m long, housed within a single silicone-rubber sleeve.

The infra-red transmitter was a Gallium-Aluminium Arsenide (GaAlAs) light emitting diode (LED) producing a peak output of 400 µW at 820nm. The output of this LED was pulsed at 10kHz, close to the optimal frequency response of the system, giving a high efficiency of transmission. The Infra-red output was linked through a "Sweet spot coupling" (RS Components, Corby) to one of the fibres in the
light guide. The fibre tips in each coupling were polished with a slurry of HPLC grade polishing alumina and distilled water.

The receiving system was connected to the second fibre in the light-guide using an identical coupling to that used in the transmitter. The infra-red detector consisted of a silicon photodiode, which produced an electrical output modulated by the intensity of the light reaching the tip of the sensing fibre. The output from the photodiode was amplified and passed through a bandpass filter centred on 10KHz, in order to separate the signal produced by the modulated infra-red from spurious signals produced by artificial lighting. The filtered signal was then demodulated to give a final signal proportional to the received power of the infra-red beam. As will be shown below, under ideal conditions this is inversely proportional to the square of the separation between the reflector and the tip of the fibre optic light guide.

**Principle of operation**

This instrument relies on the fact that light is dispersed from the end of a step-index fibre optic through a fixed solid angle which is dependent only on the properties of the fibre. Figure 2 illustrates the principle of operation. For the sake of simplicity, it is assumed that the separation between the tips of the transmitting and receiving fibres is negligible, so that they can be considered as a single fibre. It is also assumed that the plane of the reflector is perpendicular to the axis of the fibres and that the reflector is optically flat and perfect. Light is emitted from the tip of the transmitting fibre, which has area $A_1$ and is diverged through a solid cone, with half-angle $\Theta$. The cone of light impinges on the surface of the reflector, $R$ at a distance $D$, and is perfectly reflected. Since the reflector is flat and perpendicular to the axis of the fibre, the angle of light reflection is equal to the angle of incidence, and the light returned to the receiving fibre can be considered to be produced by divergence through an angle $2\Theta$ from a virtual point source $2D$ distant. The relationship between power transmitted ($P_1$) and received ($P_2$) can be derived as follows:
If it is assumed that the tip of the transmitting fibre is a point source radiating in all directions, the power density \( P_X \) at a point distance \( x \) from the fibre tip would be given by the total power output divided by the surface area of a sphere radius \( x \) \( (4 \pi x^2) \). However the infra-red beam is only dispersed through a solid angle of \( 2\theta \) degrees, and so the power density in the region \( A_2 \) at the surface of the sphere \( S \) is given by:

\[
P_X = \frac{2\theta}{360} \cdot \frac{P_1}{4\pi x^2}
\]

Therefore, in the case above, with a virtual source \( 2D \) distant from the receiving fibre,

\[
P_X = \frac{2\theta}{360} \cdot \frac{P_1}{16\pi D^2}
\]

If \( A_1 \) is small, so that the curvature of the sphere \( S \) is negligible where it cuts the end of the fibre, the power received at \( A_1 \) can be approximated as

\[
P_2 = A_1 \cdot x \cdot P_X
\]

i.e:

\[
P_2 = \frac{A_1 \cdot \theta \cdot P_1}{360.8\pi D^2}
\]

Assuming that \( P_1, \theta \) and \( A_1 \) are constant,

\[
P_2 \propto \frac{1}{D^2}
\]

The inverse-square law relationship between displacement and voltage output may be corrected electronically, or by the use of a simple software algorithm in a computerised data logging device, to yield a linear measure of displacement. Figure 3 shows the non-linear relationship between displacement and output voltage
obtained in bench calibration procedures. The relationship between voltage output and displacement beyond a distance of around 6mm follows an inverse-square law. The flat region between 1 and 6mm displacement is due to saturation of the instrument amplifiers by the high signal intensity at these small distances. The positive-sloped region between 0 and 1.5 mm displacement is an artifact, due to the breakdown at small displacements of the assumption that the separation of the axes of the transmitting and receiving fibres is negligible.
Comparison with strain guage transducer

In order to investigate the frequency response of the new transducer system and compare this with a conventional strain guage/elastic belt transducer, the following protocol was used.

Materials and methods

A computer-controlled hydraulic ram system used for destructive testing of mechanical components, with a mechanical frequency response flat to 100Hz, was connected to an electronic source of band-limited white noise, contained within a digital frequency spectrum analyser (Onno Sokki F10 dual channel FFT analyser). The system was arranged so that the position of the ram cylinder was proportional to the voltage signal at the input, and therefore the position/versus time plot matched the white noise signal. A strip of reflective adhesive tape was placed on the surface of the ram cylinder, and the measuring head of the optical transducer was fixed to a rigid support 2cm from the mean position of the ram surface, with the axis of the optical fibres perpendicular to the reflector surface. One end of the elastic belt of a strain guage respiration transducer (Lectromed) was attached to the surface of the ram, the other being attached to a rigid support. The mid point of the belt was led round a smooth metal beam to provide tension.

On separate occasions the outputs of the optical and the strain guage transducer amplifiers were connected to the second channel of the digital spectrum analyser. An electronic signal generated by a linear potentiometer connected to the ram cylinder, used internally by the computer control system for feedback control of the ram position, was fed to the first channel. The spectrum analyser was programmed to calculate the average transfer function between the ram position signal and the signal from the transducer under investigation, over a frequency range of 0-50 Hz, using a minimum of sixteen signal samples. The value of the coherence function was also calculated.

Results

The coherence function at all frequencies for both transducers was close to 1.0,
indicating a high degree of linearity over the range measured, and a causal relationship between the input signal and the transducer outputs. The frequency response curves for the two transducers differed markedly. The optical transducer showed a constant magnitude of transfer function over all frequencies (Fig 4), and a linear rising phase shift with frequency (Fig 5), indicating a simple frequency-independent time delay for the measured signal. In contrast the strain gauge transducer showed a complex frequency response (Fig 6), with large variations in response magnitude with frequency. The phase response with frequency showed large non-linear variations (Fig 7). A peak in the response magnitude, accompanied by a phase shift of 180 degrees occurred at a frequency of 23 Hz, indicating the presence of a resonance. An antiresonant feature was also present at approximately 37 Hz.

DISCUSSION

The optical transducer described has a frequency response flat to frequencies well above the highest harmonics of the chest wall displacement waveform obtained during HFJV. Although the linear range of the transducer is limited, the small amplitudes of thoracic wall movement produced during HFJV should be measurable to a high degree of accuracy. In contrast, the strain gauge shows strong frequency dependence, rendering it unsuitable for use in HFJV.

Important differences exist between the measurements obtained with a strain gauge system and with the optical transducer, since the latter measures thoracic wall displacement at a single point. In a patient lying on a fixed flat surface, with the axis of the transducer perpendicular to the reflector, the measurement corresponds to the diameter of the thorax. Since the thorax is not circular in section and the mechanical properties of the local anatomy vary from point to point on the surface, the relationship between the diameter measured and lung volume is unclear. Furthermore, since the phenomenon of pendelluft is known to occur at frequencies used in HFJV, the phase relationship between the signal obtained and the mean lung volume is
unknown. Thus the signal obtained cannot be said to be representative of lung volume at any point in time. This same argument may be applied in criticising the signal obtained from a transducer measuring thoracic circumference, since pendelluft may be occurring between lung lobes within and outside the plane of measurement. However, since it has been observed that tidal volume bears a complex relation to efficiency of gas exchange, the lack of direct correlation between lung volume and surface displacement may not be a disadvantage of either technique.

In conclusion, a transducer has been developed for the remote measurement of chest wall movement during high frequency jet ventilation, which has a flat frequency response. In common with other devices which measure chest wall expansion, calibration of this transducer for lung volume is probably not valid at frequencies in use in HFJV. However, since chest wall expansion may be a more useful predictor of gas exchange than volume change in HFJV, and the device described has technical advantages over existing measurement techniques, further investigation of its application is warranted.
LEGENDS FOR ILLUSTRATIONS

1. Schematic diagram of transducer system. In clinical use the reflector consists of a reflective self-adhesive label.

2. Principle of operation of the fibre-optic displacement transducer. R represents the surface of the reflector. $A_2$ represents a circular region of the surface of the imaginary sphere $S$, in the plane of which the power density of the infra-red beam is constant.

3. Static calibration curve for the transducer.

4. Frequency response characteristics of fibre optic transducer. Y axis is uncalibrated since absolute values depend on amplifier gain.

5. Phase versus frequency plot for fibre-optic transducer.

6. Frequency response of strain gauge transducer.

7. Phase response of strain gauge transducer. The discontinuity at 23Hz represents the occurrence of resonance.
REFERENCES


Metal beam strain gauge

Rigid block

Optical fibre head

Hydraulic ram

While noise signal

Ram position signal

Amplifier

Fibre optic system

Spectrum analyser

Elastic belt

Strain gauge

Hydraulic system

Spectrum analyser

Amplifier

Fibre optic system

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