Pulsed Laser Generation and Optical Fibre Detection of Thermoelastic Waves in Arterial Tissue

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ABSTRACT

This thesis is concerned with the investigation and practical implementation of pulsed photoacoustic spectroscopy as a means of characterising arterial tissues for the purpose of guiding laser angioplasty. This would enable the safety and efficacy of laser angioplasty to be improved by reducing the risk of accidentally perforating the vessel wall.

The first three chapters of this thesis describe the work carried out to demonstrate that pulsed photoacoustic spectroscopy can be used to characterise arterial tissue. Chapter 1 describes the clinical aspects of the project. Chapter 2 discusses the background and theory of pulsed photoacoustic spectroscopy and experiments carried out using a liquid absorber of known and variable optical properties to characterise the technique. Chapter 3 presents the experiments carried out on post mortem arterial tissue that demonstrate atheroma can be identified and the structure of the tissue ascertained from the photoacoustic signature. The remaining chapters are devoted to the in vivo implementation of the technique. This is based upon the generation and detection of photoacoustic signatures at the tip of an optical fibre using an all-optical photoacoustic probe that can be inserted into a blood vessel. The probe consists of a single delivery optical fibre and a transparent Fabry Perot polymer film ultrasound sensor mounted at the distal end of the fibre. Chapter 4 discusses the basic principles of operation and requirements of the photoacoustic probe. Chapters 5 and 6 are concerned with the theory and development of the Fabry Perot polymer film ultrasound sensor. Chapter 7 discusses the evaluation of an experimental photoacoustic probe and the demonstration of a miniature device for intravascular use. Tables of material properties and the optical and acoustic reflection coefficients used throughout are to be found in Appendices 2 and 3 respectively.
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PREFACE

Laser angioplasty is a minimally invasive treatment for reopening narrowed or blocked blood vessels in which an optical fibre is used to deliver laser energy to vaporise or ablate obstructing atheromatous plaque. A problem with the current state of laser angioplasty is that there is a significant risk of incorrectly targeting the laser energy and accidentally perforating the vessel wall. Some means of acquiring information about the composition and structure of the vessel wall is required to guide the procedure so that accidental ablation of normal tissue is avoided. Pulsed photoacoustic spectroscopy, as a means of achieving this, is explored in this thesis. The technique relies upon the generation of ultrasonic thermoelastic waves by the absorption of sub-ablation threshold nanosecond laser pulses in the vessel wall. The amplitude and temporal characteristics of the thermoelastic waves depend upon the optical properties of the target tissue which, at certain wavelengths, are different for normal and atherosclerotic tissue. The thermoelastic wave can therefore represent an acoustic signature characteristic of different tissue types that can be used to differentiate between normal and atheromatous areas. Furthermore, the reflection and generation of subsurface thermoelastic waves can provide spatial information about the structure of the vessel wall. Whilst most of this thesis is concerned with use of pulsed photoacoustic spectroscopy for guiding laser angioplasty, the technique and its method of implementation may also find application as a purely diagnostic tool to provide information about the morphology of the vessel wall. Such information would give valuable insights into the pathology, diagnosis and treatment of atherosclerosis.

To implement photoacoustic spectroscopy as a means of guiding laser angioplasty, an all-optical photoacoustic probe is described. This consists of a single optical fibre with a transparent Fabry-Perot polymer film ultrasound sensor positioned at the distal end of the fibre. The probe is inserted into the vessel. Nanosecond
"diagnostic" laser pulses are transmitted along the fibre, through the sensor producing thermoelastic waves in the vessel wall which are then detected by the sensor. If atherosclerotic plaque is identified from the photoacoustic signature, high energy "therapeutic" pulses can then be delivered along the same fibre to ablate the obstructing plaque. A key part of this system is the Fabry Perot polymer film ultrasound sensor. Accordingly, a significant part of this thesis is devoted to the theoretical and experimental aspects of the sensor.
CHAPTER 1

CLINICAL BACKGROUND

This chapter discusses the clinical/biomedical background to the project. Atherosclerosis is discussed with reference to the anatomy of the normal and atherosclerotic arterial wall. Existing minimally invasive methods of reopening narrowed or occluded blood vessels and diagnostic techniques that can provide information about the characteristics of the arterial wall are reviewed.

1.1 Atherosclerosis

Atherosclerosis is a disease in which the progressive build up of atherosclerotic plaque on the wall of a blood vessel forms a partial or complete occlusion. Blood flow to vital organs is reduced or, in severe cases, ceases altogether resulting in tissue death or organ failure. Among the symptoms of a narrowed or occluded coronary artery are chronic angina or acute myocardial infarction and in the cerebral arteries the result may be a stroke. In all it is estimated that the effects of atherosclerosis accounts for between one third and one half of all deaths in the western world.
1.1.1 The normal arterial wall

The structure of the normal arterial wall is shown in figure 1.1(a). Innermost is a layer of endothelial cells lining the vessel wall which protects the underlying intima from damage. The intima is largely composed of elastin, collagen fibres and smooth muscle cells and is separated from the media by the inner elastic laminar. The main component of the media depends on the type of vessel; in the coronary artery elastin dominates whilst aortic media is composed mainly of smooth muscle cells. The adventitia is the outermost connective tissue layer and is separated from the media by the external elastic laminar. The diameter of the lumen of the human coronary artery lies in the range 2-4mm whilst the diameter of the aorta is approximately 25mm.

1.1.2 The lesions of atherosclerosis

We discuss here only the raised fibro-lipid and calcified plaques and leave aside the fatty streak and gelatinous lesions which do not cause significant narrowing and therefore have no direct clinical implications. They are, however, of interest in the study of atherosclerosis in that they are thought to be the precursor of the fully developed raised atherosclerotic lesion.

The raised fibro-lipid plaques that are characteristic of atherosclerosis are shown in figures 1.1(b) and 1.1(c). The main feature of the plaque is a localised thickening of the intima caused by the accumulation of lipids and the proliferation of intimai smooth muscle cells. Typically the lesion consists of a lipid-rich basal pool covered by a connective tissue cap made up of smooth muscle cells, collagen fibres and elastin. The relative content of lipids and connective tissue in the plaque varies considerably. Some plaques are predominantly composed of lipid material with only a thin connective tissue cap giving it a yellow appearance. Others are almost entirely made up of connective tissue elements with very little basal lipid material in which case the lesion appears white. All gradations between the two extremes exist and the structure of the plaque may be further complicated by factors such as calcification, the presence of intramural thrombi, and ulceration. There is also wide variation between shape, size and extent of lesions adding to the difficulties in classifying atherosclerotic lesions.
1.2 TREATMENT OF ATHEROSCLEROSIS

Traditionally, the bypass operation has been the principal surgical treatment for severe narrowing of the arteries. It involves the grafting of a vessel, synthetic or from another part of the body, on to either side of the occlusion. A number of less invasive treatments for removing arterial obstructions have been developed in recent years which rely on inserting a catheter into the vessel, steering it to the point of obstruction and then removing or compressing the plaque. The three principal minimally invasive techniques are balloon angioplasty, atherectomy and laser angioplasty. These are discussed in the following sections.

1.2.1 Conventional minimally invasive treatments

Balloon angioplasty. The first transluminal coronary balloon angioplasty procedure was performed in a human patient by Gruntzig in 1977. In balloon angioplasty a radio opaque guidewire is inserted into the vessel and guided by angiography to and beyond the narrowed region. A catheter with a balloon at its distal end is then advanced over the guide wire. At the correct position the balloon is inflated, compressing and displacing atheromatous material, thereby dilating the vessel. Restenosis, the re-narrowing of the vessel, occurs within three to six months in about 30% of all cases. The process of restenosis is complex and incompletely understood. The subsequent progression of atherosclerosis or elastic recoil of the vessel wall may cause narrowing of the vessel. In the majority of cases, however, it appears that the trauma to the vessel wall imposed by balloon angioplasty leads to the proliferation of intimal smooth muscle cells and a consequent thickening of the vessel wall. It is the most significant disadvantage of balloon angioplasty and has led to the introduction of other minimally invasive techniques such as atherectomy, insertion of endoluminal stents and laser angioplasty in the hope that restenosis rates can be reduced.
Atherectomy

This is a procedure in which a high speed rotating cutting device is advanced over a guide wire and atheromatous material skimmed from the vessel wall and collected in the device housing. It has the advantage that plaque is actually removed from the vessel wall in contrast to the plaque compression and displacement action of balloon angioplasty and enables collection of vascular tissue for histological examination. It requires a fairly large bore catheter making it unsuitable for narrow or tortuous vessels. Restenosis rates are similar to those of balloon angioplasty.

1.2.2 Laser Angioplasty

Laser angioplasty uses the ablating action of absorbed laser energy to remove occlusive material. An optical fibre is guided to the occluded region using angiographic techniques. Laser light is launched into the fibre and emerges at the other end to ablate the targeted tissue. The advantages of laser angioplasty are well cited: recanalisation by the vaporisation of occlusive material is considered more satisfactory than simply attempting to force it out of the way as in balloon angioplasty. Complete
occlusions that cannot be crossed by balloon catheters can be reopened and it has been suggested that various types of diffuse lesions are more suitably treated by laser angioplasty. The great hope (not yet realised) with laser angioplasty was that by avoiding mechanically inflicted trauma of the vessel wall, as in balloon angioplasty and atherectomy, restenosis rates would fall.

Laser angioplasty was first carried out on a human in 1983 by Ginsberg et al. who recanalised a blocked femoral artery. The first procedure to be carried out on the coronary artery was performed in 1983 by Choy et al. Since then there has been considerable interest in the use of laser angioplasty particularly for the treatment of coronary occlusions. Initial work made use of the CW Argon, Nd:YAG and Holmium lasers. Immediate "technical success", in the sense that completely occluded arteries were reopened, was reported although follow-up reports indicated poor restenosis rates. It was suggested that a significant factor in this was the thermal injury caused by the CW laser energy. In an effort to avoid the problems of thermal damage attention was turned to the use of pulsed lasers. The pulsed Nd:YAG laser and the pulsed dye laser have been experimented with and some work continues in the evaluation of the Holmium laser but most efforts are now concentrated on the excimer laser operating in the UV with sub microsecond pulses. The excimer laser achieves ablation with minimal thermal damage. The exact mechanism is unclear: direct breakage of molecular bonds and fragmentation by shock wave generation have been suggested although even with the predominantly non thermal ablative mechanisms of the excimer laser, restenosis rates have not fallen below those of balloon angioplasty and in some cases they are higher.

Achieving satisfactory ablation of calcified plaques has also been a problem throughout the history of laser angioplasty. It is a significant limitation as calcification is a common feature in cases of advanced atherosclerosis. The predominantly thermal mechanisms of ablation of the CW lasers proved incapable of ablating calcified plaque although some success has been achieved in vitro with pulsed laser energy. The satisfactory removal of calcified plaques in vivo has yet to be demonstrated with any laser system.

Apart from restenosis and the difficulty in ablating calcified plaques, excimer laser angioplasty limitations appear to lie in the catheter design. This makes use of multiple fibres in order to ablate large areas and still be flexible. The "dead space" between fibres
and around the catheter may cause a high rate of dissection (15% in coronaries). The latter can probably be dealt with by improved catheter design although there is suggestion that the dissection is caused by shock waves and cavitation. The high cost of excimer lasers, in excess of $200,000, is also a major disadvantage for a procedure which, if the problem of restenosis is not solved, may only have a niche application as an adjunct to balloon angioplasty for reopening complete occlusions. According to the literature, perforation would appear to be less of a problem with the excimer laser than the Nd:YAG laser. This may be due to the shorter penetration depth of the excimer wavelength (308nm) compared to that of the Nd:YAG (1.06µm) although improved catheter design and careful patient selection are significant factors. Litvack\textsuperscript{28} studied the results of excimer laser angioplasty in over 2000 patients. Multiple fibre catheters passed over a guidewire were used in 90% of cases. In the last consecutive 1000 patients the perforation rate was 0.4% and in the others it was 1.2%. The low perforation rates were due to careful catheter sizing (ie using a catheter that fills the lumen) and avoiding so-called high risk lesions. Despite these results, if the range of lesions that can be treated with the excimer laser is to be extended the limitations imposed by the risk of perforation remain to be addressed. Certainly it remains a serious limitation with the (much cheaper) Nd:YAG laser.

Attempts to reduce the risk of perforation have been made by selecting a wavelength that is preferentially absorbed in plaque thereby reducing the threshold of ablation compared to that of normal tissue\textsuperscript{29,30}. The same thing can be achieved by injecting agents into the bloodstream such as beta carotene which are selectively uptaken by atherosclerotic plaque\textsuperscript{31}. These absorb strongly at the wavelength that is used for ablation so that the plaque is selectively ablated. Much more work is required to identify suitable agents before it becomes an effective treatment. Other means of avoiding perforation have focused on the use of optical and ultrasound diagnostic techniques to describe the composition and structure of the vessel wall in front of the fibre tip. If the normal arterial tissue is identified, the fibre catheter can be repositioned until a region of plaque is identified for ablation. Various so-called guided laser angioplasty systems based on a number of diagnostic techniques have been proposed and are described in the next section.

It would appear that as yet laser angioplasty has failed to fulfill its original expectations of becoming a routine and widely used treatment for reopening narrowed
blood vessels. The most likely reason for this is that it has not been clearly demonstrated to provide a superior and cost effective alternative to other techniques, particularly balloon angioplasty. It has been suggested that it is most likely to find useful application in recanalising complete occlusions prior to balloon angioplasty or for clearing long arterial segments of diffusely distributed plaque in preparation for endoluminal stent placement.

1.3 GUIDED LASER ANGIOPLASTY

The requirements for a guided laser angioplasty system are that the composition of the tissue in front of the fibre tip can be identified and that information about the structure and thickness of the vessel wall can be obtained. A number of techniques, discussed below, have been suggested for this purpose, some of which have actually been used as a means of guiding laser angioplasty whilst others remain, at present, in vitro characterisation tools.

1.3.1 Review of existing techniques

Some of the techniques described below have been suggested to have a somewhat broader role than simply as an identifier of atherosclerotic plaque during laser angioplasty. The optical techniques in particular have been suggested as a means of monitoring the evolution of atherosclerosis by performing a number of intraluminal investigations over a period of time to decide at which point intervention is required.

Angiography X ray imaging can be used to guide a radio opaque catheter to the point of narrowing or occlusion by injecting a contrast agent into the bloodstream. The technique provides a map of the arteries under examination and the location of narrowing. It has been shown to have severe limitations in providing information about the severity of atherosclerotic lesions and does not yield any information about the composition or structure of the vessel wall.
Angioscopy This is direct viewing of the vessel wall using what is essentially a thin fibre optic endoscope. Although it provides an output about the surface of the vessel wall in an easily interpretable form, it does not give information about the thickness or composition of underlying layers.

**Fluorescence spectroscopy** The tissue is excited to fluoresce using low power laser light coupled into the delivery fibre. The fluorescence is collected by the delivery fibre and analysed at the other end. Various lasers have been used including the argon-ion (476.5nm), the XeCl (308nm), XeF (351nm), He-Cd (325nm) and the Nitrogen laser at 337nm as the excitation source. It has been shown that the fluorescence spectra of atheromatous plaques is different to that of normal vascular tissue although the large background fluorescence of the normal tissue can make it difficult to pick out individual spectral features. Although it cannot give spatial information, using a wavelength that penetrates the full thickness of the vessel wall does enable the spectral features from subsurface layers to be identified. *In vitro* results look promising but when implemented *in vivo* the presence of blood and changes in the fluorescence spectra of the vessel wall induced by the ablative pulses made it difficult to ascertain when plaque removal was complete. As a consequence the system did not prevent perforations.

**Raman spectroscopy** The analysis of Raman scattered light from the surface of vascular tissue was first performed by Clarke in 1987. Calcified lesions were studied using 514nm Argon-ion laser excitation. A dominant 960cm\(^{-1}\) shift was reported which was not present in areas adjacent to the calcified plaques. This work was continued with a report by the same author on the investigation of soft fatty plaques. A number of shifts which were not observed in normal tissue were clearly identified. NIR FT Raman spectroscopy has shown that a wide range of vascular components can be identified and it is suggested that it may have a role as a means of performing *in vivo* histological examination.

**Optical coherence tomography (OCT)** The use of OCT for investigation of the arterial wall is a recent development. The technique is based upon the localised interference of subsurface reflections within the tissue obtained using a low coherence source and a Micheleleson interferometer. Spatial information is obtained by matching the optical
pathlength in the reference arm of the interferometer with the optical pathlength taken by the light reflected from a particular point at depth in the tissue. Thus the technique relies upon variations in optical reflectance of the different components of the vessel wall to construct an image. As yet its ability to identify atheromatous components is unclear but it has been shown in vitro that the technique can provide an image of the full thickness of the normal vessel wall.

**Intravascular ultrasound (IVUS).** This is a modality which can provide a cross-sectional image of the vessel wall by the use an ultrasonic transducer operating in pulse-echo mode at the distal end of a catheter\textsuperscript{45}. The origins of IVUS date back to the very beginnings of diagnostic ultrasound when Cieszynski\textsuperscript{46} built an ultrasound probe for intracardial measurements. It is only recently, however, that technological developments have enabled the idea to evolve into a clinically useful technique\textsuperscript{47,48}. There are two principal transducer types for 2-D imaging: (i) the single element transducer in which the ultrasonic beam is rotated around the axis of the catheter either by mechanically rotating the transducer or using a fixed transducer in conjunction with a rotating acoustic mirror. The transducer operates at frequencies in the range 20-35 MHz. (ii) The multielement circular array enables circumferential scanning by electronically addressing each element. In theory the multielement array possesses a number of advantages but the disadvantages imposed by the large number of elements and hence connecting leads to the array has resulted in the use of the mechanically rotated single element device in most in vivo clinical work. Combined IVUS-balloon catheters\textsuperscript{49} have been used to monitor balloon angioplasty and an IVUS guided laser angioplasty system has been reported\textsuperscript{50}. IVUS relies primarily on the acoustic impedance differences of subsurface features to reveal the structure of the vessel wall\textsuperscript{51,52}. Differences in the acoustic impedance of the constituents of non-calcified arterial tissues are generally small and, for certain tissue structures, this can result in poor contrast, ambiguities and images that are difficult to interpret. Furthermore, since ultrasound characterises in terms of structure rather than composition, it can lead to a situation in which two soft atherosclerotic lesions of the same structure but very different composition may appear identical under IVUS investigation\textsuperscript{53}. The expense of IVUS probes can also be prohibitive as, to avoid cross infection, they are single-use devices.
Photoacoustic spectroscopy

Photoacoustic spectroscopy as a means of identifying areas of atheromatous plaque during laser angioplasty was first proposed by Mills\textsuperscript{34} in 1988. The principal advantage of pulsed photoacoustic spectroscopy is that information about both the optical and acoustic properties of a target can be obtained by examining the amplitude and temporal characteristics of the thermoelastic waves that are generated. It can also provide a means of describing the underlying structure of a target by the measurement of the time of arrival of thermoelastic waves that are reflected or generated at subsurface acoustic or optical interfaces. As such it has the potential to overcome the limitations of the investigative techniques described above.
CHAPTER 2

PULSED PHOTOACOUSTIC SPECTROSCOPY

This chapter provides the theoretical and experimental background to the use of pulsed photoacoustic spectroscopy as a means of characterising arterial tissue. The theoretical aspects of the technique are discussed using a semi-qualitative treatment with reference to the extensive literature published on the subject. The aim of the experimental work was to acquire enough knowledge about the photoacoustic process in order to be able to interpret the photoacoustic signatures generated in complex media such as tissue. An experimental technique for the accurate detection of laser generated thermoelastic waves was therefore developed and characterised. A non-scattering liquid absorber whose characteristics could be carefully controlled was used as a substitute target to enable the effects of pulse energy, absorption coefficient and boundary conditions on the photoacoustic signal to be studied.
2.1 INTRODUCTION

The photoacoustic effect, the generation of sound from light, was first demonstrated by Alexander Graham Bell in 1880 with the aim of using it as a communication tool. Although the "Photophone", as Bell called it, was never to find widespread use, the photoacoustic effect did go on later to be studied as an analytical tool. It was not, however, until the invention of the laser in 1960 as a source of concentrated optical energy, that the photoacoustic effect was able to shed its curiosity status and develop its potential as a practical investigative technique.

There are two types of photoacoustic techniques. The first is the gas phase type of photoacoustic spectroscopy. In this technique, a microphone is used to sense the pressure variations in an enclosed gas cell produced by the heating and cooling of the gas layer in thermal contact with a target that is being irradiated by a modulated laser beam. It is principally the opto-thermal characteristics of the target that provide the basis of the analytical abilities of the technique. Since the acoustic signal is not actually generated in the target but in the adjacent gas layer it is termed an indirect photoacoustic technique. The second photoacoustic technique is the direct detection of the stress or thermoelastic wave that is generated in a target as a result of the rapid thermal expansion caused by the absorption of a short laser pulse. The amplitude and temporal characteristics of the thermoelastic wave are dependent upon the optical and acoustic properties of the target and can be used to characterise the target. In addition, the reflection and generation of subsurface thermoelastic waves can provide information relating to the structure and thickness of the target. It is therefore a dual sensing technique combining the spectroscopic nature of purely optical techniques with the ability of ultrasound techniques to obtain spatial information. It is this type of direct time-resolved or pulsed photoacoustic spectroscopy that is of interest in this work. The indirect gas phase type measurements are not considered further.

The direct detection of pulsed laser generated acoustic energy was first demonstrated in 1962 by White using a pulsed Ruby laser. Photoacoustic studies by Carome et al. and Gournay were carried out soon after setting the theoretical and
experimental foundations of modern pulsed laser photoacoustic techniques. Subsequently a wide and varied literature on the subject has been published, summaries of which can be found in a number of review articles.\textsuperscript{59,60,61,62,63,64,65} Applications of photoacoustic techniques have encompassed industrial, scientific and medical fields. Industrial applications often use the technique simply as a non-contact form of ultrasound generation for flaw detection and characterisation of materials and structures. The non contact aspect of photoacoustic generation is particularly attractive when investigating substances such as powders, corrosive liquids or materials at high temperatures. Applications as diverse as defect visualisation in composites\textsuperscript{66}, characterisation of adhesive bonds\textsuperscript{67} and the detection of oil contamination in water for pollution monitoring\textsuperscript{68} have been reported. The spectroscopic nature of the technique makes it a useful tool for measuring the optical absorption and scattering properties of various media particularly those with high optical attenuation that make conventional optical transmission measurements difficult\textsuperscript{69,70}. Indeed, it is the dependence of the photoacoustic signal upon the optical properties of a target that the majority of biomedical applications of the technique exploit. The photoacoustic characterisation of arterial tissue\textsuperscript{71,72} for the identification of atheromatous plaque, first proposed by Mills in 1988\textsuperscript{54}, is based upon the exploitation of preferential optical absorption in atheroma. Other photoacoustic studies of vascular tissue have been carried out to determine its optical properties and ablation threshold\textsuperscript{73,74,75,76}. The use of laser generated stress waves also provides a means of directly visualising axial light distributions in tissues using a laser wavelength that penetrates deep into the tissue. This has been suggested as a means of monitoring heat deposition in tissue during pulsed laser treatments such as port wine stain removal\textsuperscript{77} in order to determine the optimum "dose" of laser light. This mode of use also provides the potential for imaging soft tissues, such as the breast\textsuperscript{78}, based upon spatial variations in the optical properties of the tissue. The dual spatial-spectroscopic basis of photoacoustic imaging is an exciting one and is currently attracting considerable interest\textsuperscript{70,77,78,79,80,81,82}. Development of photoacoustic imaging techniques, is however at an early stage and most of these reports (with the exception of the paper by Kruger \textit{et al.}\textsuperscript{79}) are somewhat speculative reporting at best a point measurement in a tissue phantom rather than a complete \textit{in situ} image. The non-invasive monitoring of blood glucose levels, commonly regarded as the holy grail of physiological monitoring, is currently being evaluated using photoacoustic spectroscopy\textsuperscript{83,84}. A stage removed from
clinical diagnosis and treatment but nevertheless of vital importance, given the increasing use of optical techniques in medicine, is the modelling of light transport in tissue. This requires accurate measurement of the optical properties of tissue and photoacoustic spectroscopy can provide a useful complement to existing optical techniques, particularly for \textit{in vivo} measurements.

It would be overstating the case to suggest that photoacoustic techniques are ever likely to find the very wide range of applications of other diagnostic techniques such as ionising radiation, ultrasound, MRI (magnetic resonance imaging) and near infrared spectroscopy but there is strong potential for its successful use in a few important niche applications such as those described above.

\section*{2.2 LASER GENERATED THERMOELASTIC WAVES}

This section discusses the theoretical aspects of photoacoustic generation and propagation to provide a basis for the interpretation of photoacoustic signatures generated in arterial tissues. Rigorous theoretical treatments of photoacoustic generation have been well documented in the literature. Emphasis is therefore placed, in the first instance, on a semi-qualitative treatment using a simple theory based upon thermoelastic expansion in an optically non-scattering liquid to bring out the key physical aspects of the process. This theory is then modified to take into account the effects of scattering that are found tissues. The effects of diffraction and acoustic attenuation as the thermoelastic wave propagates through water are also discussed.

\subsection*{2.2.1 Thermoelastic wave generation in a non-scattering absorber}

Laser generated acoustic energy can be produced by mechanisms such as electrostriction, radiation pressure, Brillouin scattering and thermoelastic expansion. It is the last of these that is the dominant mechanism at low optical power densities and forms the basis of the discussion in this section.

Consider a pulsed laser beam incident on a homogeneous, non-scattering liquid
absorber as shown in figure 2.1. The irradiance decreases exponentially along the axis of the laser beam in accordance with Lambert’s law

\[ I(z) = I(0)e^{-\mu_{a}z} \]  

(2.1)

where \( I(z) \) is the irradiance at depth \( z \), \( I(0) \) is the irradiance at the surface of the absorber and \( \mu_{a} \) is the optical absorption coefficient. The energy of the optical pulse is absorbed producing a thermal gradient in the liquid. Rapid thermal expansion occurs leading to acoustic energy propagating away from the source. The geometry of the acoustic source approximates to that of a cylinder of radius \( R \) and length \( 1/\mu_{a} \) where \( R \) is the radius of the incident laser beam and \( 1/\mu_{a} \) is the optical penetration depth. When \( R \) is large compared to the penetration depth (the 1-dimensional case), we need consider only the acoustic waves propagating above and below the source. For efficient acoustic generation, the duration of the optical pulse should be sufficiently short that both thermal and stress confinement take place.

**Thermal confinement** To fulfil the condition of thermal confinement, the optical pulse duration should be short compared to the thermal relaxation time \( \tau_{t} \) of the heated volume so that there is no significant heat loss by thermal diffusion during the heating process. \( \tau_{t} \) is usually based (assuming 1-dimensional heat flow) upon the distance \( z \) that heat travels in time \( t \) in a material of thermal diffusivity \( D \)

\[ z = \sqrt{2Dt} \]  

(2.2)

A corresponding thermal relaxation time can be defined by rearranging this equation and substituting for \( z \) a linear dimension that is characteristic of the heated volume. For a non scattering absorber, the optical penetration depth \( 1/\mu_{a} \) is an appropriate substitution for \( z \). This gives the thermal relaxation time \( \tau_{t} \) as

\[ \tau_{t} = \frac{1}{2\mu_{a}^{2}D} \]  

(2.3)

**Stress confinement** For stress confinement, the optical pulse duration should be sufficiently short that the process of heat deposition is faster than that of thermal expansion. This requires a pulse duration that is short compared to the stress relaxation
Figure 2.1 Generation of thermoelastic waves propagating above (A) and below source (B) due to the absorption of short laser pulses in a non-scattering liquid absorber.

This is generally taken as the time $\tau_s$ taken for the thermoelastic wave to propagate a distance equal to the optical penetration depth $1/\mu_s$ where

$$\tau_s = \frac{1}{\mu_s c}$$  \hspace{1cm} (2.4)

and $c$ is the velocity of sound in the medium$^{59}$. Physically, under the conditions of stress confinement, one can therefore visualise the photoacoustic effect as being the process of the near-instantaneous deposition of energy producing a correspondingly instantaneous stress profile in the absorbing volume which then propagates away in the form of an acoustic wave.

Assuming thermal and stress confinement, a simplified model of 1-dimensional thermal expansion in a liquid$^{65,88}$ can be used to examine the factors that affect the characteristics of the thermoelastic wave. In general, the pressure increase $P$ due to a thermally-induced volume change $\Delta v$ in a liquid of volume $v$ is given by
\[ P = \frac{B \Delta v}{v} \quad (2.5) \]

where \( B \) is the bulk modulus of the liquid. The change in volume due to a temperature change \( \theta \) is

\[ \Delta v = \beta v \theta \quad (2.6) \]

where \( \beta \) is the coefficient of volume thermal expansion. Substituting equation 2.6 into equation 2.5 the pressure increase becomes

\[ P = B \beta \theta \quad (2.7) \]

Now consider the absorption of a short optical pulse of duration \( t_p \) that fulfils the thermal and stress confinement conditions described above. We assume that the temporal and spatial characteristics of the optical pulse are that of a top-hat profile. Assuming the geometry shown in figure 2.1, there is a distribution of optical energy \( E(z) \) along the \( z \) axis of the absorbing volume which is absorbed and converted to heat producing a corresponding axial temperature distribution \( \theta(z) \). To calculate \( \theta(z) \) we consider the temperature rise produced by the optical energy absorbed in a thin elemental slice of thickness \( \delta z \) at a depth \( z \). Assuming that all of the absorbed optical energy in the thin slice is converted to heat, then the temperature rise at depth \( z \) is

\[
\theta(z) = \frac{E(z) - E(z + \delta z)}{\rho \pi R^2 \delta z C_p}
\]

which can be written

\[
\theta(z) = \left( \frac{dE(z)}{dz} \right) \delta z \left( \rho \pi R^2 \delta z C_p \right)^{-1}
\]

\[ E(z) = I(0) e^{-\mu s z} \pi R^2 t_p \quad (2.10) \]

Differentiating equation 2.10 with respect to \( z \) and inserting into equation 2.9 gives
\[ \theta(z) = \frac{f_p \mu_a I(0) e^{-\mu_a z}}{\rho C_p} \quad (2.11) \]

where \( C_p \) is the specific heat at constant pressure. Equation 2.11 is in agreement with Gournay.\(^8\) The pressure distribution \( P(z) \) along the axis of the absorbing volume can now be written using equations 2.7 and 2.11

\[ P(z) = \frac{B \beta f_p \mu_a I(0) e^{-\mu_a z}}{\rho C_p} \quad (2.12) \]

Since \( I(0) \) is the irradiance at the surface it can be expressed in terms of the total optical pulse energy \( E \) as

\[ I(0) = \frac{E}{\pi R^2 t_p} \quad (2.13) \]

Substituting this into equation 2.12 gives

\[ P(z) = \frac{B \beta E \mu_a e^{-\mu_a z}}{\rho C_p \pi R^2} \quad (2.14) \]

Inspection of equation 2.14 shows that the amplitude of the thermoelastic wave is proportional to the optical absorption coefficient, the optical pulse energy and the bulk modulus of the liquid absorber. Although equation 2.14 does not provide a complete description of the temporal characteristics of the thermoelastic wave (solving the 1 dimensional wave equation is required for this\(^8,\)\(^9\)), it does enable a useful insight into the shape of the wave to be obtained simply by converting the axial distance \( z \) to time \( t \) using the speed of sound \( c \) (i.e. \( z = ct \)). The time history of the thermoelastic wave \( P(t) \) therefore mirrors the initial spatial pressure profile \( P(z) \) which in turn corresponds to the axial irradiance distribution \( I(z) \). When measuring from above the source (A in figure 2.1) it can be expected that the thermoelastic wave will rise very rapidly (instantaneously if we assume perfect stress confinement) to a maximum that corresponds to the acoustic contribution from \( z=0 \). Thereafter, the acoustic pressure decreases exponentially with exponential constant \( \mu_a c \). Conversely, when detecting the thermoelastic wave propagating below the source (B in figure 2.1), the acoustic pressure will increase exponentially (with exponential constant \( \mu_a c \)) to a maximum that
corresponds to the contribution from $z=0$ and then fall very rapidly to zero. By fitting an exponential to the appropriate part of the thermoelastic wave (above or below source) and with knowledge of the speed of sound, a value of $\mu_s$ can therefore be obtained. A useful advantage of analysing the temporal characteristics (as opposed to amplitude characteristics) of the thermoelastic wave is that the value of $\mu_s$ is independent of variations in incident laser pulse energy, detector sensitivity and source-detector alignment.

Whilst the above has assumed stress confinement according to equation 2.4 to predict the shape of the thermoelastic wave, in practice the non-zero duration of the laser pulse means that the instantaneous edge of the wave corresponding to $z=0$ is modified. For the above source wave for example, the step increase in pressure will actually be approximated to by an increase of finite rise time that is related to the laser pulse duration and shape. Thus, the stress confinement condition does not really apply to the instantaneous edge of the pressure distribution when predicting the shape of the thermoelastic waveform. For this part of the wave, the shape is defined by the temporal characteristics of the laser pulse.

2.2.2 Thermoelastic wave generation in a scattering absorber.

For a target in which scattering dominates (as is the case with many biological tissues\textsuperscript{89}) the axial light distribution in a semi-infinite slab illuminated by a large area collimated beam can be modelled using diffusion theory\textsuperscript{90}. At a certain depth beneath the surface, the light becomes diffuse and the irradiance decreases exponentially with exponential constant $\mu_{\text{eff}}$ such that

$$I(z) = I(0) ke^{-\mu_{\text{eff}}z}$$ \hspace{1cm} (2.15)

where $k$ is a constant. $\mu_{\text{eff}}$ is the effective attenuation coefficient and is a function of $\mu_s$ and the scattering coefficient $\mu_t$ and is given by

$$\mu_{\text{eff}} = \sqrt{3\mu_t \mu_s + (1-g)\mu_s}$$ \hspace{1cm} (2.16)

where $g$ is the mean cosine of the scattering angle. Since the axial light distribution is exponential, the expression for the acoustic pressure at each point in the absorbing volume below the depth at which the light becomes diffuse is of a similar form as
equation 2.14 with \( \mu \) replaced by \( \mu_{\text{eff}} \). The amplitude of the thermoelastic wave is therefore proportional to \( \mu_{\text{eff}} \). We can also, using the same arguments as in section 2.2.1, expect an exponentially decreasing component (when measuring from above the target) in the part of the thermoelastic wave that corresponds to depths beyond which the light has become diffuse. The exponential constant will be \( \mu_{\text{eff}}c \) and so by fitting an exponential to this part of the wave, a value for \( \mu_{\text{eff}} \) can be obtained. It has been suggested that the depth beyond which the light can be considered to be diffuse in predominantly scattering media is somewhere between 1 and 10 transport mean free paths\(^{90,91,92} \). The transport mean free path \( l_t \) is given by \( 1/(\mu_t+(1-g)\mu_L) \). In the region where the light is not diffuse (i.e. close to the surface), the light distribution can be complex and may even exceed the surface irradiance. Such behaviour will manifest itself in a complex manner in the temporal characteristics of the initial part of the thermoelastic wave. The actual penetration depth of the light in strongly scattering media is given by \( (nl_t+\mu_{\text{eff}}^{-1}) \) where \( n \) is an integer (usually between 1 and 10) depending upon the criteria used to define the depth at which the light becomes diffuse.

### 2.2.3 Surface boundary conditions

The discussion in section 2.2.1 assumed that the boundary shown in figure 2.1 is purely an optical one and the acoustic impedances of the media on both sides of the boundary are the same. Under these circumstances, the source can be considered to be acoustically unbounded. In practice, the boundary conditions are often such that there is a significant acoustic impedance mismatch between the target absorber and the medium above it. This can strongly affect the shape of the thermoelastic wave propagating below the source. When measuring below the source, the resultant thermoelastic wave is the superposition of the time-delayed reflection of the above source wave from the boundary and the below source wave. If the acoustic impedance of the medium at the boundary is much larger (e.g. glass) than that of the liquid absorber, the so-called rigid boundary condition, the reflection from the boundary is large and positive. The net downward travelling thermoelastic wave is therefore increased in amplitude and slightly broadened, but otherwise retains the shape of the acoustically unbounded case. If the acoustic impedance of the medium at the boundary is much less (e.g. air) than that of the liquid absorber, the free boundary condition, the
reflection from the boundary is large and negative. Since it is time-delayed the net downward travelling thermoelastic wave acquires a trailing tensile component and becomes bipolar shape in shape.

2.2.4 Propagation

In most practical set-ups, the thermoelastic wave propagates a distance in water or some water-like medium before arriving at a detector. Acoustic propagation in liquids from a source to a detector will involve attenuation and diffraction effects which can have a significant influence on the amplitude and shape of the thermoelastic wave.

2.2.4.1 Attenuation

The acoustic amplitude attenuation coefficient of plane sound waves $\alpha$ in nepers/m is

$$\alpha = \frac{1}{d} \ln \left( \frac{A_i}{A_o} \right)$$

(2.17)

where $A_i$ is the initial pressure amplitude and $A_o$ is the amplitude after propagating a distance $d$ in the medium. In liquids attenuation is frequency dependent, classically varying as the square of frequency. The acoustic attenuation coefficient in water$^{93}$ is $26 \times 10^{-15}$ nepers/m/Hz$^2$. At frequencies up to approximately 20 MHz and for path lengths of a few cm acoustic attenuation is therefore small and can usually be neglected. In soft water-based biological tissues, there is considerable variation in the attenuation coefficients of different tissue-types$^{94}$ but as an approximate guide it can be taken as 1dB/cm/MHz. At 10 MHz the pressure amplitude would therefore have decreased by an order of magnitude after propagating only 1cm.

2.2.4.2 Diffraction

So far, it has been assumed that the shape of the thermoelastic wave propagating away from the heated volume effectively retains that of the initial stress distribution. In practice, diffraction effects alter the shape of the wave as it propagates away from the
source. Since the majority of experimental arrangements for measuring thermoelastic waves involve a finite separation between source and detector, diffraction effects in practice must nearly always be considered.

Assuming the 1-dimensional case and uniform illumination across the diameter of the laser beam we can make use of, in the first instance, the diffraction analysis of a plane circular acoustic radiator. Diffraction effects apply in the region close to the source resulting in complex spatial-temporal characteristics of the acoustic field. This region is termed the near field and, for a plane circular source, the demarcation between this and the far field beyond, is given by

\[ z_d = \frac{R^2}{\lambda_a} \]  

(2.18)

where \( z_d \) represents the axial distance from the source, \( \lambda_a \) is the the characteristic acoustic wavelength and \( R \) is the radius of the source. Some authors\(^{61,63} \) take the acoustic wavelength as being equal to the optical penetration depth: - i.e \( \lambda_a = \mu_a^{-1} \) for a non scattering absorber, for scattering media the definition of penetration depth given in section 2.2.2 can be used to obtain \( \lambda_a \). Others equate the optical penetration depth to one half cycle of a periodic acoustic disturbance\(^{59,74} \) - i.e. \( \lambda_a = 2\mu_a^{-1} \) for a non scattering absorber. Taking the latter case and from equation 2.18

\[ z_d = \frac{\mu_a R^2}{2} \]  

(2.19)

Equation 2.18 is derived by considering the resultant amplitude from contributions over the entire surface of a source radiating continuously at a single frequency. In the case of transient generation, it has been suggested that a more accurate model of diffraction is that based upon consideration of the sum of a contribution from the centre of the source and contributions from the edges\(^{66} \). Although this analysis results in the same expression for the far field zone as equation 2.18 it does raise the question as to what the characteristic wavelength should be for a transient which contains many frequency components. Kramer et al.\(^{66} \) demonstrates that there is considerable variation in \( z_d \) depending upon the criteria used to choose the frequency component from the acoustic frequency spectrum to represent the dominant contribution to amplitude variations of the transient. This aspect does not always appear to have been fully considered in work related to photoacoustics and it is suggested that equation 2.19 should be considered
Figure 2.2 Transformations of thermoelastic waveforms due to propagation from near field to far field.

only as a rough approximation.

The effect of diffraction on the shape of a laser generated thermoelastic wave has been investigated theoretically and experimentally\cite{59,61,97,98,99,100}. The general rule that emerges from these publications, most succinctly put by Sigrist\cite{59}, is that the thermoelastic wave in the far field is the time derivative of the near field waveform. We can apply this rule to the near-field thermoelastic wave profiles discussed in section 2.2.1 for a non scattering absorber. In the near field, when measuring above the source, the thermoelastic wave is a compressive monopolar wave consisting of a rapidly increasing positive edge followed by an exponential decrease. The time derivative of this is a positive delta function followed by an exponentially decaying negative component (figure 2.2). Thus the effect of diffraction is to convert the exponentially decaying compressional part of the near field waveform into an exponentially decaying rarefaction in the far field. When measuring below the source, the situation is effectively reversed so that, assuming there is no acoustic impedance mismatch at the boundary, the far field thermoelastic wave consists of an exponentially rising compressional component.
followed by a negative (rarefaction) delta function. Thus, to obtain \( \mu_s \) assuming far field propagation, the exponential should be fitted to the trailing edge of the signal when measuring above the source and to the initial part of the signal when measuring below the source. The same far field analysis applies to obtaining \( \mu_{\text{eff}} \) in strongly scattering media providing the exponential fit is applied to the part of the waveform that corresponds to the depths beyond which the light has become diffuse. The most obvious effect of diffraction is that the near field monopolar waveform is transformed to a bipolar wave in the far field - the delta functions being approximated to, in practice, by a short pulse as a result of the laser pulse duration-stress confinement considerations discussed in section 2.2.1.

The time derivative rule discussed above emerges from a somewhat complex analytical approach and provides little insight into the underlying physical processes. The treatment of Pauksztas et al., similar to the central-source-edge-effects theory of the generalised circular radiator for transient generation, provides a more intuitive account of diffraction phenomenon. This is discussed in more detail when interpreting the experimental results in section 2.4.4.

2.3 Practical requirements

We now consider the laser requirements for efficient pulsed laser generation of thermoelastic waves and the requirements of the detector in order to obtain an accurate representation of the thermoelastic wave.

2.3.1 Generation

The requirements of the laser system are that it can produce optical pulses of sufficient energy and that the duration of the pulses are short enough to fulfil the thermal and stress confinement conditions discussed in section 2.2.1. The nanosecond millijoule pulses that are typically emitted by Q switched Nd:YAG lasers are commonly used to generate thermoelastic waves. Although pulse energies of a few millijoules are
usually adequate, the limits for photoacoustic generation may begin to be approached when using pulse durations of the order of several nanoseconds to generate thermoelastic waves in a strong absorber. This aspect is examined below assuming a strongly absorbing liquid with the thermal and acoustic properties of water.

For thermal confinement, the optical pulse duration should be small compared to the thermal relaxation time $\tau_t$ of the absorbing volume as defined by equation 2.3. Taking $D=1.4 \times 10^3 \text{ cm}^2/\text{s}$ and $\mu_s=500\text{cm}^{-1}$, equation 2.3 gives $\tau_t=1.4\text{ms}$. Pulses on a nanosecond timescale are therefore well within this limit.

For stress confinement, the optical pulse duration should be short compared to the acoustic transit time $\tau_s$ of the thermoelastic wave across the source as defined by equation 2.4. Taking $c=1500\text{m/s}$ and $\mu_s=500\text{cm}^{-1}$, equation 2.4 gives $\tau_s=13\text{ns}$. Thus for strongly absorbing targets, laser pulses of nanosecond duration only just fulfil this condition making the stress confinement condition the dominant criterion when considering pulse duration.

### 2.3.2 Detection

The accurate measurement of acoustic waves at ultrasonic frequencies presents a number of difficulties which also apply to the detection of laser generated thermoelastic waves. The principal requirements of the detector are that it has adequate sensitivity, a wideband uniform frequency response and a small active diameter. PZT, LiNO$_3$, and PVDF piezoelectric ultrasound transducers have been shown to have the required sensitivity for the detection of laser generated thermoelastic waves in various media. The frequency response and active diameter requirements, applicable to many areas of ultrasound measurement, are more difficult to fulfil using piezoelectric devices.$^{101,102}$

For an accurate representation of the thermoelastic wave, the detector bandwidth should extend to the maximum frequency component of the thermoelastic wave and the response should be uniform to this frequency. For the strong absorber example where $\mu_s=500\text{cm}^{-1}$, the acoustic wavelength is $40\mu\text{m}$. In water this corresponds to an acoustic frequency of $37.5\text{MHz}$. To achieve this bandwidth is in itself non-trivial. However, given that the predicted thermoelastic wave profiles discussed in sections 2.2.1 and 2.2.4.2 contain a near-step change in acoustic pressure (depending on pulse duration), and hence even higher frequency components, it might be suggested that a detector with
a response several times this frequency (e.g. 100MHz) should be specified. Uniformity of frequency response is difficult to achieve with piezoceramic devices due to their large acoustic impedance mismatch with water. PVDF devices have a much lower acoustic impedance mismatch to water offering better uniformity of response. Whatever transducer is used, it cannot be assumed to have a flat frequency response - even PVDF devices may suffer from non-uniformities in frequency response due to the physical transducer design and cable resonance and bandlimiting effects. The transducer should always be calibrated against a known reference traceable to primary standards - an area not always considered in previous photoacoustic studies.

The active diameter is an important consideration to avoid spatial integration of the acoustic field especially when making near field measurements. It should ideally be small compared to the acoustic wavelength which at 40 MHz is less than 40μm in water. It is very difficult to fabricate such a small active diameter piezoelectric transducer whilst retaining acceptable sensitivity.

At present it does not appear that any commercially available piezoelectric transducer can fulfil all the requirements of sensitivity, bandwidth and active diameter in their entirety. The PVDF membrane type of hydrophone\textsuperscript{103,104} probably represents the best compromise. This type of hydrophone currently represents the gold standard in ultrasound measurement on account of its wideband uniform frequency response, linearity, stability and minimum perturbation to the acoustic field under measurement. Membrane hydrophones with a bandwidth to 90MHz\textsuperscript{103} are available although the thinness of the membrane (9μm) required to achieve this bandwidth results in low sensitivity. The smallest active diameter for a commercially available membrane hydrophone is currently 0.5mm, much larger than the tens of micron wavelengths associated with frequencies in the tens of MHz. PVDF needle hydrophones of less than 100μm active diameter are available, although they have poor frequency response characteristics in comparison with membrane hydrophones.

Since the detector requirements are so difficult to fulfil, it is worth considering ways in which useful measurements can be made within the confines of existing technology. The detector bandwidth requirements can be relaxed considerably if the aim of the measurement is to determine only the μ-dependent exponential part of a thermoelastic wave and an accurate representation of the fast edges of the waveform is not required. Furthermore, it can be shown that the time constant of the exponential part
of the thermoelastic wave is not significantly modified by, for example, a PVDF transducer operating in thickness mode that is acoustically thick (i.e. comparable to or thicker than $\lambda_a$). The reason for this is that when such a detector is acoustically thick, the output is proportional to the stress integrated across the depth that the acoustic wave has penetrated into the detector. In the case of an exponentially increasing acoustic wave propagating into the detector (i.e. the initial part of the below-source wave), the output, for high $\mu_a$, is proportional to an expression in which an exponential of the same time constant as the acoustic signal dominates. The same applies to an exponentially decreasing acoustic wave (as is the above-source wave) propagating out of the detector. Values of $\mu_a$ that correspond to an acoustic wavelength comparable to or smaller than the thickness of the detector can therefore still be measured. With regard to the active diameter requirements, these become less critical if measurements are made well into the far field.

2.4 PHOTOACoustIC MEASUREMENTS ON A NON-SCATTERING LIQUID ABSORBER

The experimental work described in this section was carried out in preparation for the tissue experiments described in chapter 3. Measurements were carried out at 532nm using a non-scattering liquid absorber of known optical properties. These measurements enabled the experimental method to be characterised and the photoacoustic process to be studied as a function of pulse energy, optical absorption and surface boundary conditions. The liquid absorber used was a readily available writing ink ("washable" blue Solv-X Quink ink) which was diluted with an acidic buffer (Tris pH3) to obtain a range of stable solutions of different values of $\mu_a$. Optical transmission measurements at 532nm were used to confirm that these solutions were non-scattering and obeyed the Beer-Lambert law. Undiluted, the optical absorption of the ink is 700cm$^{-1}$. The speed of sound and the density of the ink were measured and found to be almost identical to that of water.
2.4.1 Experimental arrangement

The experimental arrangement shown in figure 2.3 was used for most of the photoacoustic studies described in this thesis. This arrangement was used to study the effect of pulse energy and $\mu_s$ on the shape of thermoelastic waves generated above and below source. A different set-up was used to investigate surface boundary effects. This is described in section 2.4.5.

The sample container was filled with the ink/Tris solution of the required concentration. The container was placed on a tilt platform attached to a modified scissor jack and raised until the ink was in contact with the Perspex window (2mm thickness). Nanosecond laser pulses at 532nm were obtained using a frequency doubled, Q switched Nd:YAG laser. Typical pulse durations, depending upon flashlamp voltage were measured to be in the range 12-20ns. The diameter of the laser beam was 3mm. The output beam from the laser system was directed on to the ink absorber producing a thermoelastic wave that propagated vertically upwards (i.e. above source) through the Perspex window. A second thermoelastic wave was also generated simultaneously that propagated vertically downwards (i.e. below source) through approximately a 5mm thickness of ink and was subsequently reflected from the steel block at the bottom of
the container to follow precisely the same path as the upwards travelling wave. The thermoelastic waves arrived at a PVDF membrane hydrophone after reflection from a glass block angled at 45° acting as an acoustic reflector. The total acoustic path length from source to hydrophone was approximately 6cm and was essentially the same for both above and below source thermoelastic waves. The hydrophone was a calibrated 25MHz 50μm bilaminar PVDF membrane hydrophone (Marconi Y-33-7611) of active diameter 1mm and sensitivity 100mv/MPa. The output of the hydrophone was displayed and averaged over 30 shots using a 500MHz digitising oscilloscope.

Alignment of the hydrophone with the incident thermoelastic waves was carried out optically with the sample container empty. The low power frequency doubled fixed Q output of the Nd:YAG laser was used as an alignment beam to mimic the path of the thermoelastic waves. The position of the hydrophone was adjusted until the Fresnel reflection from the Perspex window was visible on its active area. This ensured that the initial upwards travelling wave was correctly aligned. To ensure that the downwards travelling wave, after reflection from the steel block, was coincident with the upwards travelling thermoelastic wave, a plane parallel mirror was placed on the top surface of the block. The tilt platform was then adjusted so that the reflection from the mirror of the 532nm Nd:YAG output was visible on the active region of the hydrophone.

Possible influences, of the experimental arrangement, on the shape and amplitude of the thermoelastic waves are considered in the following section.

2.4.2 Characterisation of experimental set-up

To examine the effect of the hydrophone frequency response and the frequency dependent attenuation of the water on the shape of the thermoelastic waves obtained using the experimental arrangement of figure 2.3, analysis in the frequency domain was carried out. Figure 2.4 (a) shows a typical thermoelastic wave generated in concentrated ink using the experimental arrangement of figure 2.3. The FFT of this signal was taken in order to obtain the acoustic frequency spectrum and is shown in figure 2.4(b). The frequency response of the PVDF hydrophone $BH(f)$, calibrated by the National Physical Laboratory UK (NPL), is shown in figure 2.5. The fit to these data points was achieved using the acoustic interaction model described later in chapter 5 for a 50μm thick PVDF film. The frequency dependent attenuation of water $BW(f)$, using equation 2.17 for a 6cm
path length, is also shown. Multiplying $B_A(f)$ by $B_w(f)$ gives the combined effect of the water attenuation and the hydrophone response and is denoted $C_{wa}(f)$. The combination of the attenuation and hydrophone characteristics somewhat fortuitously results in $C_{wa}(f)$.

Figure 2.4 (a) Thermoelastic wave generated in ink absorber and (b) its acoustic frequency spectrum.
being flat to approximately 22MHz (-3dB point), thereafter falling off rapidly. Since the dominant frequency components of the acoustic frequency spectrum are below 25MHz (figure 2.4(a)), it can be expected that the shape of the thermoelastic wave will not be significantly modified by the effect of the hydrophone characteristics and the water attenuation.

![Graph showing PVDF hydrophone frequency response $B_h(f)$, frequency dependent attenuation of water for 6cm path length $B_w(f)$, and $C_{wh}(f)=B_h(f)B_w(f)$.](image)

**Figure 2.5** Graph showing PVDF hydrophone frequency response $B_h(f)$, frequency dependent attenuation of water for 6cm path length $B_w(f)$, and $C_{wh}(f)=B_h(f)B_w(f)$.

To verify this assertion, the frequency spectrum of the thermoelastic wave shown in figure 2.4(b) was corrected for by dividing by $C_{wh}(f)$. The inverse FFT was then taken to obtain the corrected waveform in the time domain. Above 25MHz, the frequency components of the acoustic spectrum are close to the noise floor and since the scaling factor required for the correction at these frequencies is large, the main effect of applying the correction is to increase the total integrated noise. Taking the inverse FFT therefore resulted in a signal dominated by noise. For this reason, the combined correction was applied to frequencies below 25MHz only. This is reasonable since the dominant frequency spectrum components of the signal arriving at the hydrophone are below 25MHz. The inverse FFT of the corrected spectrum giving the corrected time domain signal is shown in figure 2.6 along with the uncorrected signal. As expected
there is little difference between the shapes of the two signals. Most importantly, the measurement of $\mu_a$ obtained from the thermoelastic wave would be unaffected. It is possible that there are significant frequencies above 25MHz that arrive at the hydrophone but cannot be detected due to the limited bandwidth of the hydrophone. This however is unlikely as above say 20MHz, even taking into account the relatively low sensitivity of the hydrophone, there is no evidence of an increasing trend in the acoustic frequency spectrum shown in figure 2.4(b). Thus even with a wider bandwidth hydrophone it is unlikely that the shape of the thermoelastic would be significantly different. Very high frequency components above 30MHz would anyway be effectively filtered out by the attenuation in the water.

Other possible influences on the thermoelastic waves are attenuation by the 2mm thick Perspex window and the reflection from the 45° glass block acting as an acoustic reflector. Acoustic attenuation in polymers is usually frequency dependent. For a pathlength of a few mm, however, the dependence on frequency can be neglected. The pressure amplitude reflection coefficient at an interface is independent of frequency to a first approximation. These effects can therefore be considered to have an insignificant influence on the shape of the signal.

![Thermoelastic wave corrected for hydrophone frequency response and frequency dependent attenuation of water.](image)

**Figure 2.6** Thermoelastic wave corrected for hydrophone frequency response and frequency dependent attenuation of water.

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Since the frequency dependent nature of the acoustic losses in the experimental set-up can be ignored, a simple amplitude scaling factor correction can be applied to the measured thermoelastic waveforms to estimate the acoustic pressure at the surface of the target. This takes into account the acoustic losses arising from the attenuation by the water, the Perspex window and the reflection from the glass block. For the frequency dependent attenuation of water, the centre frequency of 10MHz (figure 2.4(b)) was used. From the surface of the ink to the hydrophone, a total acoustic loss factor of 3.8 was calculated. The hydrophone output was therefore multiplied by this factor to estimate the acoustic pressures generated at source. The correction was checked experimentally and found to give a good estimate of the acoustic losses in the experimental arrangement. Where the vertical scale of the thermoelastic waveforms shown in this and the following chapter are in units of acoustic pressure, it should be taken that this correction has been applied.

2.4.3 Ablation threshold

An essential requirement is that measurements are made in the thermoelastic

![Figure 2.7](image)

**Figure 2.7** Effect of increasing laser pulse energy on shape of thermoelastic wave generated in concentrated ink absorber. Approximate ablation threshold = 0.6mJ/mm².
expansion regime. If this requirement is not fulfilled the shape of the wave may be
distorted thus affecting the measurement of $\mu_a$. Ablation of the absorber by melting,
vaporisation or optical breakdown is also clearly to be avoided if the technique is to be
used as an investigative technique particularly for medical applications. Experiments
were therefore carried out to determine the onset of ablation processes by studying the
shape of the acoustic wave as the laser pulse energy was increased. Concentrated ink
was used as the absorber as this represents the case in which the ablation threshold is
lowest. The experimental arrangement in figure 2.3 was used. As the fluence incident
on the ink increased beyond 0.6mJ/mm\(^2\) the rarefaction part of the waveform became
distorted (figure 2.7). The absence of this distortion was taken to be indicative that
thermoelastic expansion was the dominant mechanism in generating an acoustic signal.

2.4.4 Absorber concentration

Thermoelastic waves for three different ink/Tris concentrations are shown in figure
2.8 for measurements above and below the source. Propagation in the far field is
assumed. Hence the far field analysis described in section 2.2.4.2 is used to obtain
values of $\mu_a$. When measuring above the source (figure 2.8(a)), the characteristic shape
of the wave is that of an initial compression followed by an absorption-dependent
exponentially decaying rarefaction. By fitting an exponential to the rarefaction part of
the waveform and using a value of 1500m/s for the speed of sound, a value for $\mu_a$ could
be obtained. When measuring below the source (figure 2.8(b)) it is the initial,
exponentially increasing compression part of the wave that is dependent upon absorption
and it is to this part of the wave that an exponential was fitted to in order to obtain $\mu_a$.
For both sets of measurements, values of $\mu_a$ for nine different ink/Tris concentrations
were determined and are shown in figure 2.9. Figure 2.9 also shows a straight line
representing the known values of $\mu_a$ of the ink/Tris solutions as a function of
concentration. Both sets of measurements show a linear relation with concentration, as
expected. The values of $\mu_a$ calculated from the downwards travelling (i.e below source)
thermoelastic waves are in close agreement with the known values of $\mu_a$. The values of
$\mu_a$ calculated from the upwards (i.e. above source) travelling waves however are a factor
of 2.8 lower than the known values. The amplitudes of both above and below source
measurements are shown in figure 2.10 and show a linearly increasing dependence on
Figure 2.8 Thermoelastic waves generated (a) above and (b) below source in three different ink/Tris concentrations. Inset diagrams show the part of the waveform (between A and B) to which an exponential is fitted in order to obtain $\mu_e$. Incident fluence=0.085 mJ/mm², pulse duration=14 ns, acoustic path length=6 cm.
Figure 2.9 Values of absorption coefficient $\mu_a$ calculated by fitting an exponential to thermoelastic waves generated above and below source in different ink/Tris concentrations. The dashed line represents the known value of $\mu_a$ as a function of ink concentration.

Concentration, as expected. The shape of the waves shown in figure 2.8 with reference to diffraction effects and the discrepancy between the above and below source measurements are discussed below.

Thermoelastic waveforms Assuming that detection is taking place in the far field, the signal should be the time derivative of the near field pressure distribution. As discussed in section 2.2.4.2, when measuring above the source the time derivative of the near field thermoelastic wave is a positive delta function followed by an exponentially decaying negative component. The delta function is modified by the effect of the non-zero laser pulse duration (section 2.2.1) and the limited bandwidth of the hydrophone to give an initial short compressional pulse which is then followed by the $\mu_a$-dependent exponentially decaying rarefaction as shown in figure 2.8(a). When measuring below the
source, the situation is effectively reversed so that the far field thermoelastic wave consists of a $\mu_a$-dependent exponentially rising compressional component followed by a short rarefaction pulse (approximating to the negative delta function) as shown in figure 2.8(b). Assuming that these measurements are in the far field, the shape of the waveforms are more or less as expected. The validity of the far field assumption is marginal for the high $\mu_a$ data according to equation 2.19. However, as discussed in section 2.2.4.2, this equation is somewhat approximate due to the uncertainty in defining the characteristic acoustic wavelength of the thermoelastic wave.

Above/below source discrepancy The reason for the discrepancy between the above and below source values of $\mu_a$ is unclear. It is unlikely to be due to a difference in the experimental set-up for the two cases as this was the same for both types of measurements. In particular, the acoustic path length (~6cm) travelled by each wave was very nearly the same. The discrepancy may be due to a combination of diffraction effects and the spatial integration of the acoustic field by the hydrophone due to its relatively large active area as discussed by Paltauf et al.\textsuperscript{100} Paltauf et al., suggests that the tensile component of the thermoelastic wave arises from the edge of the acoustic source. For a detector placed on the axis of the source, an initial compressional

Figure 2.10 Amplitude of thermoelastic waves generated above and below source in different ink/Tris concentrations.
contribution will be obtained from the centre of the source. A short time later, depending on the diameter of the source, the tensile component will arrive from the edge of the source. Providing the detector is not too close to the source, the tensile component will arrive immediately after the compressional pulse thus giving the classic far-field bipolar shape. When measuring the below-source wave we are mainly interested in the initial \( \mu_a \)-dependent exponentially increasing part of the wave. This is unaffected by the tensile components travelling from the edge of the source which arrive later in time. Hence, providing one fits to the initial part of the increasing exponential part of the thermoelastic wave, the effects of diffraction will have a minimal effect on the \( \mu_a \) measurements. Thus the below source measurements agree with the known \( \mu_a \) data. When measuring above the source, however, the time-delayed tensile components from the edge may coincide with the exponentially decaying part of the wave that carries the temporal information relating to \( \mu_a \). It is therefore possible that this part of the wave is somehow influenced by contributions from the edge which arrive in the same time period and modify the decay in such a way so as to produce a lower value of \( \mu_a \). The active diameter has a significant role to play in these types of measurements. The tensile component reaches a maximum on the source axis because the contributions from the edges of the source will have then travelled the same path length and therefore superpose to form a maximum. Off axis, this superposition does not occur so the tensile component is small. The output of a large active area detector (\( \sim 1 \)mm) effectively gives the spatial integral of the acoustic field intercepted by the detector. Therefore, the detector output may be dominated by the off axis contribution which contains only a small tensile component. Thus the magnitude of the tensile component of the hydrophone output is reduced compared to that which would be observed using a very small diameter active area detector on axis. This explanation may well apply to the experimental measurements since a relatively large active diameter (1mm) hydrophone was used and would account for the low \( \mu_a \) measurements obtained when measuring above source.

In many practical medical implementations of photoacoustic spectroscopy, measurements are made on the same side of the target (i.e. above source). If absolute values of \( \mu_a \) or \( \mu_{\text{eff}} \) are required, it would therefore be necessary to take into consideration the effects described above by calibrating the system using a phantom target of variable optical properties and applying an appropriate correction. Ideally, a
scheme in which a small active area detector can be placed in front of and immediately adjacent to the source should be used so that diffraction effects can be neglected. This would require a transparent detector through which the excitation laser pulses can be transmitted.

2.4.5 Surface boundary conditions

When measuring below the source, the surface boundary conditions can play an important role in defining the shape of the thermoelastic wave. To investigate this, the experimental arrangement shown in figure 2.11 was used. Concentrated ink was used as the liquid absorber. A glass plate and air were used to simulate rigid and free boundary conditions respectively. The far field thermoelastic waveforms produced by these two boundary conditions are shown in figure 2.12. In each case the acoustic impedance mismatches at the boundary are large so the resultant thermoelastic wave detected by the hydrophone is the superposition of the wave travelling downwards and the reflection at the boundary of the wave travelling upwards. For the rigid boundary, the thermoelastic wave is the time derivative of the superposition of the downwards travelling monopolar wave and the reflected upwards travelling monopolar wave. Both waves are of the same sign so the resultant is a bipolar wave. For the free boundary, the reflected upwards travelling wave is of opposite sign and therefore adds a tensile

Figure 2.11 Experimental set-up for investigating surface boundary effects. A thick glass plate and air are used to approximate to rigid and free boundaries respectively.
Figure 2.12 Effect of boundary conditions on shape of thermoelastic wave. (a) Rigid boundary (glass) and (b) free boundary (air).
component producing a resultant bipolar waveform in the near field. The far field time derivative transform therefore gives the tripolar shape shown in figure 2.12(b). To examine the case in which the absorbing volume is effectively acoustically unbounded, a thin layer of transparent castor oil was placed on the surface of the ink. Castor oil has a very similar acoustic impedance to ink so in this configuration there is no acoustic impedance mismatch at the ink boundary. The results of this experiment are shown in figure 2.13. The first signal is the wave travelling downwards, the second signal is the inverted reflection at the surface (air) of the castor oil of the upwards travelling wave. The time delay between the two signals corresponds to the thickness of the castor oil layer. This confirms the superposition theory used to describe the shape of the thermoelastic waves shown in figure 2.12.

For the experimental arrangement of figure 2.3, the acoustic impedance of the Perspex window is fairly close to that of water (Appendix B) resulting in a reduced rigid

![Graph](image)

**Figure 2.13** Separation of thermoelastic waves using a layer of castor oil on ink surface. (i.e. acoustically unbounded case) A - initial downwards travelling wave, B - inverted reflection from surface of castor oil layer

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boundary effect. In practice, this is unlikely to have a significant effect on the characteristics of the below source thermoelastic waves.

2.5 CHAPTER SUMMARY

This chapter has discussed the key principles of photoacoustic spectroscopy. In particular, it has shown theoretically and experimentally how the optical absorption coefficient or effective attenuation coefficient of a target can be obtained from a laser generated thermoelastic wave. This is a key feature as it forms the basis of the photoacoustic characterisation of arterial tissues.

A calibrated experimental set-up for carrying out photoacoustic studies has been developed and characterised. It has been demonstrated that this set-up provides an accurate representation of the thermoelastic waves generated. Using this set-up and a non scattering liquid absorber of known and adjustable optical properties, the effect on the amplitude and temporal characteristics of thermoelastic waves of varying the optical absorption coefficient of the absorber has been observed. Qualitatively the results are in agreement with theory - that is the shape and amplitude of the waves are dependent upon optical absorption. However, when measuring the above-source waves it was found that the optical absorption coefficient obtained from the thermoelastic wave was a factor of 2.8 lower than that obtained from the corresponding below source measurements. The below source measurements agreed closely with the known values. It appears that this discrepancy is related to diffraction effects and the spatial averaging effect of the hydrophone. It is an important finding as it shows that when making above source measurements, as is the case with many medical applications of the technique, there are likely to be substantial errors in the measurement of optical properties if the particular source-detector system is not characterised and corrected for.
CHAPTER 3

PHOTOACOUSTIC CHARACTERISATION OF ARTERIAL TISSUE

This chapter is concerned with the generation and detection of photoacoustic signatures in arterial tissues. Previous photoacoustic studies are reviewed and the optical and acoustic properties of arterial tissue discussed. The photoacoustic characterisation of arterial tissue is then considered in terms of the detection of atheroma and acquiring spatial information relating to the structure of the tissue. Experiments on normal and atheromatous post mortem human aorta, in which photoacoustic signatures at three wavelengths namely, 436nm, 461nm and 532nm were generated, are reported.
3.1 INTRODUCTION

It has been shown in the previous chapter that the amplitude and temporal characteristics of laser generated thermoelastic waves depend upon the optical properties of the target. Given that it has been demonstrated by others that there is preferential optical absorption in the range 420nm-530nm in atheroma\textsuperscript{106}, the photoacoustic signature can therefore be used to discriminate between areas of normal arterial tissue and atheromatous plaque\textsuperscript{54,107}. In addition, the reflection and generation of subsurface thermoelastic waves can provide information about the structure and thickness of the vessel wall. Pulsed photoacoustic spectroscopy can therefore be a useful tool for the characterisation of arterial tissue. \textit{In vitro} time-resolved photoacoustic measurements of the type discussed in this thesis have been reported by others. Al Dhahir \textit{et al.}\textsuperscript{73} measured optical attenuation via the photoacoustic signature using a pulsed dye laser and confirmed the preferential attenuation in atheroma over the range 440-500nm. Crazzolara \textit{et al.}\textsuperscript{76} also showed that the photoacoustic signature of calcified plaques, using an excimer laser at 308nm, is different to that of normal arterial tissue. Such measurements provide preliminary experimental evidence that photoacoustic spectroscopy can be used to characterise arterial tissue. Other photoacoustic investigations on vascular tissue have tended to be in relation to ablation studies for laser angioplasty rather than for the purpose of characterisation\textsuperscript{74,75}.

The aim of the experimental work described in this chapter was principally to exploit the increased optical absorption in atheroma between 420nm and 530nm for discrimination purposes. For this reason, laser pulses at 436nm and 461nm were used to generate photoacoustic signatures in arterial tissue. This approach is somewhat non-specific as it signifies the presence of atheroma but provides little further information about the tissue. An additional aim, therefore, was to obtain thickness information using 532nm laser pulses. At this wavelength the light can penetrate the full thickness of the tissue producing a corresponding photoacoustic contribution that extends from the surface of the tissue (intima) to the lowermost layer (adventitia) enabling spatial
information to be obtained. As an introduction to the experimental work, sections 3.2 and 3.3 discuss the optical and acoustic properties of arterial tissue and the use of the photoacoustic signature for characterising arterial tissue.

3.2 OPTICAL AND ACOUSTIC PROPERTIES OF ARTERIAL TISSUE

3.2.1 Optical properties

The optical properties of arterial tissues have been measured \textit{in vitro} by a number of workers. Van Gemert \textit{et al.}\textsuperscript{108} obtained the Kubelka-Munk absorption and scattering coefficients in normal and atheromatous plaque at three commonly available laser wavelengths; 514.5nm, 633nm and 1.06\,\mu m. It was noted that only at 514.5nm was there higher absorption in the plaque specimen. The first comprehensive measurements of the optical properties of normal and atheromatous aortic tissue, also in terms of the Kubelka-Munk coefficients, were made by Prince \textit{et al.}\textsuperscript{106} over the wavelength range 250nm-1.25\,\mu m. It was noted that in soft yellow fibro-fatty plaques there was preferential absorption in the spectral range 420-530nm, peaking at about 470nm when absorption was at least a factor of two higher than in normal areas of tissue (see figure 3.1). Carotenoids within the plaque mass were identified as the chromophore responsible for the increased absorption in the atheroma. Others have also reported preferential attenuation in atheroma around 470nm using a variety of techniques although the effect has not always been specifically attributed to preferential absorption\textsuperscript{109,73,110}. In general, scattering in soft tissues varies little with wavelength so it is likely that any preferential attenuation observed is in fact due to increased absorption. Although the most comprehensive data on both normal and atheromatous arterial tissue is that obtained by Prince \textit{et al.}\textsuperscript{106} it has the disadvantage that the data was analyzed using Kubelka-Munk theory\textsuperscript{111}. This theory is no longer considered to provide an accurate description of light propagation in biological tissues, particularly in terms of providing absolute values of
optical coefficients\(^9\). Despite this limitation, two conclusions can still be drawn from

![Figure 3.1 Preferential absorption in atheroma. (Prince et al.\(^{106}\))](image)

the work of Prince et al.\(^{106}\): firstly, that there is an increase in optical attenuation in arterial tissue in the spectral range 420nm-530nm, and secondly, that this is due to increased absorption rather than scattering.

In terms of the absolute optical absorption and scattering coefficients of arterial tissue the work of Keijzer et al.\(^{112}\) is more useful. Integrating sphere measurements of the intima, media and adventitia of normal aortic samples were made separately and the transport coefficients (as opposed to Kubelka-Munk coefficients) were obtained using diffusion theory. Optical coefficients at the three wavelengths of interest in this study are shown in table 3.1 where \(\mu_a\) and \(\mu_s\) are the absorption and scattering coefficients respectively and \(g\) is the mean cosine of scattering angle. The data in table 3.1 shows that there are significant differences in the values of \(\mu_s\) for the different tissue layers. It also shows, in common with most soft tissues, that scattering dominates. Since the study did not include atheromatous samples, the presence of preferential absorption observed by Prince et al.\(^{106}\) was not confirmed.
Table 3.1 Optical coefficients of normal arterial tissue (Keijzer et al.\textsuperscript{112})

<table>
<thead>
<tr>
<th>λ (nm)</th>
<th>Tissue layer</th>
<th>μ\textsubscript{s} (cm\textsuperscript{-1})</th>
<th>μ\textsubscript{a} (cm\textsuperscript{-1})</th>
<th>g</th>
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<tr>
<td>436</td>
<td>intima</td>
<td>36</td>
<td>320</td>
<td>0.85</td>
</tr>
<tr>
<td></td>
<td>media</td>
<td>12</td>
<td>505</td>
<td>0.91</td>
</tr>
<tr>
<td></td>
<td>adventitia</td>
<td>42</td>
<td>330</td>
<td>0.78</td>
</tr>
<tr>
<td>461</td>
<td>intima</td>
<td>16</td>
<td>300</td>
<td>0.85</td>
</tr>
<tr>
<td></td>
<td>media</td>
<td>8</td>
<td>490</td>
<td>0.91</td>
</tr>
<tr>
<td></td>
<td>adventitia</td>
<td>21</td>
<td>305</td>
<td>0.78</td>
</tr>
<tr>
<td>532</td>
<td>intima</td>
<td>10</td>
<td>220</td>
<td>0.85</td>
</tr>
<tr>
<td></td>
<td>media</td>
<td>5</td>
<td>400</td>
<td>0.90</td>
</tr>
<tr>
<td></td>
<td>adventitia</td>
<td>12</td>
<td>270</td>
<td>0.91</td>
</tr>
</tbody>
</table>

Gas phase photoacoustic spectroscopy was used by Singleton \textit{et al.}\textsuperscript{113} to examine relative attenuation coefficients over the excimer laser wavelength range 200nm-350nm. No difference in attenuation between normal and atheromatous areas of arterial tissues was observed over this wavelength range.

A difficulty with measurements of the optical properties of layered biological media, such as arterial tissue, is that they are often found to differ widely making comparisons between different workers difficult. This is due to the use of different measurement techniques and theoretical analyses and the heterogeneous nature of the tissue. Atheroma is particularly complex as the composition of a raised fibrous-fatty plaque of a particular thickness can vary between the extremes of either predominantly all-lipid or all-fibrous material. Additional complicating factors include calcification, the presence of intramural thrombi and vascularisation. These factors are of particular significance when attempting to exploit preferential absorption in atheroma for discrimination purposes as the presence of preferentially absorbing chromophores within the plaque mass can vary greatly.
3.2.2 Acoustic properties

The acoustic properties of non-calcified arterial tissue, with the exception of attenuation, are similar to those of water\textsuperscript{114}. The velocity of sound in aortic tissue has been measured \textit{in vitro} with values ranging from 1492 m/s to 1650 m/s\textsuperscript{115,116}. The variation was mainly attributed to differences in collagen content. As is to be expected, calcified lesions have significantly higher velocities, in the range 1900-2000 m/s. The acoustic impedance of normal artery wall and fibrous-fatty plaques has been measured as being in the range 1.53-1.86 \times 10^6 Kg/m^2s\textsuperscript{115}. Acoustic impedance differences at the boundaries of the different layers in uncalcified tissue are of the order of only a few percent. Attenuation in non-calcified tissue shows more variation than the other acoustic properties with values of around 6 dB/cm at 10MHz for the normal artery wall and 5.6-28.9 dB/cm for fibrous/fatty lesions at the same frequency\textsuperscript{115}. Apart from attenuation, the differences in acoustic properties between normal and atheromatous tissue are small compared to the differences in optical properties.

3.3 Thermoelastic wave generation in arterial tissue

The basic principles of thermoelastic wave generation and propagation in a non-scattering liquid absorber with the physical properties of water (discussed in the previous chapter) are generally applicable to photoacoustic studies in arterial tissue. The main difference to consider is the axial light distribution arising from the highly scattering nature of arterial tissue. As table 3.1 shows, scattering dominates light transport in arterial tissue. The diffusion theory approximation\textsuperscript{90} discussed in section 2.2.2 can therefore be invoked. This means that the axial irradiance decreases exponentially with exponential constant $\mu_{\text{eff}}$ beyond the depth at which the light can be considered to have become diffuse. Thus by fitting an exponential to the appropriate part of the thermoelastic waveform $\mu_{\text{eff}}$ can be determined. Based upon the transport mean free path criteria discussed in section 2.2.2 and the data in table 3.1, a typical value for the depth beyond which the light has become diffuse is approximately 200\mu m for arterial tissue.
Thus the exponential fit should be applied to photoacoustic contributions that correspond to depths greater than 200μm. In addition arterial tissue is multilayered, each layer having different optical properties as shown in table 3.1. If light penetrates deep into the tissue, a complex photoacoustic signal can be expected reflecting the changes in optical attenuation at boundaries. Consideration is given in following two sub-sections to using the photoacoustic signature to discriminate between atheroma and normal tissue and obtain information relating to the thickness of the tissue.

3.3.1 Detection of atheroma

As described in section 2.2.2, the amplitude and temporal characteristics of the thermoelastic wave depend upon the effective attenuation coefficient $\mu_{\text{eff}}$ of the tissue which in turn is a function of the absorption coefficient $\mu_a$. Since the ratio of optical absorption in atheroma to normal tissue at 470nm is approximately a factor of two, there will be significant differences between the photoacoustic signatures of the two tissue types thus providing a means of discriminating between normal and atheromatous tissue. Although section 3.2.2 indicated that there are differences in the acoustic properties of non-calcified arterial tissue, these are very small compared to the differences in optical properties of the tissue and are not expected to contribute significantly to differences in the photoacoustic signature. Calcified plaques are not specifically considered in this paper although they should be straightforward to detect as both optical and acoustic characteristics are very different from normal tissue resulting in a highly individual photoacoustic signature.

3.3.2 Thickness measurement

Measurements of the thickness of the tissue can be made in two ways. Firstly, by measuring the time of arrival of acoustic reflections from subsurface layers at which an acoustic impedance mismatch occurs. Since acoustic impedance mismatches between the layers of non-calcified arterial tissue are very small, high detection sensitivities and high optical pulse energies (possibly close to the damage threshold of the tissue) would be required to achieve this. An alternative method of acquiring spatial information is to choose an optical wavelength such as 532nm that penetrates the full thickness of the tissue.
tissue. This enables subsurface thermoelastic waves to be generated at the boundaries of tissue layers where there is a change in optical properties. Measurement of the time of arrival of these waves allows the thickness of the layers to be measured. Evidence of the feasibility of this concept is provided by the measurements of Keijzer et al.\textsuperscript{112} (table 3.1) which showed that the optical properties of the intima, media and adventitia of normal aortic tissue are significantly different.

3.3.3 Pulse energy

The fluence incident on the tissue must be below the threshold at which irreversible changes occur yet sufficient to produce thermoelastic waves that can be readily detected. The ablation threshold of normal arterial tissue\textsuperscript{117,75} for a 10ns pulse at 532nm is in excess of 10mJ/mm\textsuperscript{2}. Irreversible changes in tissue may occur at fluences somewhat below the ablation threshold and consideration should also be given to the possibility of damage to tissue by the mechanical disruption of tissue layers as a result of the generation and propagation of the thermoelastic wave. For these reasons it would be sensible to limit the incident fluence to around 1mJ/mm\textsuperscript{2}.

3.4 PHOTOACOUSTIC MEASUREMENTS IN \textit{POST MORTEM} ARTERIAL TISSUE

To study the photoacoustic signals generated in arterial tissue, measurements on 15 \textit{post mortem} human aortas at 3 wavelengths: 436nm, 461nm and 532nm were made. Each specimen was kept refrigerated at 4°C and used within six hours of excision. Areas of atheroma were identified by macroscopic inspection and only soft yellow, predominantly lipid plaques were selected as atheromatous samples. In each case, the fat layer adjacent to the adventitia was removed and the aorta specimens stored in saline until use.
3.4.1 Experimental arrangement and procedure

The experimental set-up shown in figure 3.2 was used for the tissue experiments. It is identical to the arrangement described in chapter 2 except for the laser system described below.

Laser system Optical pulses at 436nm, 461nm and 532nm were obtained using a laser system that consisted of a frequency doubled, Q switched, Nd:YAG laser and a gas filled Raman cell. 532nm optical pulses of nanosecond duration were obtained using the direct output of the frequency doubled, Nd:YAG laser. The diameter of the beam was 3mm. To obtain optical pulses at 436nm and 461nm, the Raman cell was used to shift the 532nm output of the Nd:YAG. When the Raman cell was filled with hydrogen it provided shifts in wavenumber of 4155cm\(^{-1}\) enabling an output at 436nm to be obtained by selecting the first anti-Stokes shift. Similarly, when the Raman cell was filled with methane, the wavenumber shift of 2914cm\(^{-1}\) enabled 461nm to be obtained also by selection of the first anti-Stokes shift.

Procedure The alignment procedure of the experimental set-up described in section 2.4.1 was used. The aorta specimen under investigation was then immersed in saline in the

Figure 3.2 Experimental arrangement for the generation and detection of thermoelastic waves in tissue.
sample container. The scissor jack was raised until the top surface of the tissue was in light uniform contact with the Perspex window. To verify that the sensitivity of the experimental set-up remained constant, ink was used as a reference target absorber in place of the tissue sample before and after each set of tissue measurements were taken.

3.4.2 Results

The photoacoustic signatures generated in normal and atheromatous post mortem aorta at 436nm, 461nm and 532nm are discussed in the following subsections. The amplitude correction factor described in section 2.4.2 was applied to obtain acoustic pressures at source. The effective attenuation coefficients were obtained by fitting an exponential decay to the part of the waveform that corresponds to a depth at which the light could be assumed to have become diffuse. In these experiments this depth was assumed to be at least 400μm - approximately 2 transport mean free paths. Only the thermoelastic waves generated above the source were studied as it is these waves that would be detected in any practical in vivo system. The 2.8 discrepancy factor discussed in section 2.4.4 was therefore used to obtain the correct value of μ_eff. The speed of sound used throughout was 1570m/s. Although a large number of measurements of μ_eff were made, it was considered that due to the large variation in the morphology of the different specimens, a statistical analysis would be inappropriate. Selected waveforms of the photoacoustic signals generated at a number of positions on a single aorta specimen are therefore shown at each wavelength. These were chosen to be as representative as possible of the different specimens with variations from these discussed in detail.

3.4.2.1 λ=436nm

Waveforms generated at four positions on a sample of normal aortic tissue are shown in figure 3.3(a). The decaying rarefaction part of the waves (between A and B) corresponds spatially to the photoacoustic contribution from the media. By fitting an exponential to this part of the wave a value of μ_eff=123cm⁻¹ was obtained.
Figure 3.3 Thermoelastic waves generated in post mortem human aorta at 436nm. (a) Normal aorta (4 positions) and (b) atheromatous aorta (3 positions). \( \mu_{eff} \) is calculated by fitting an exponential between A and B. Incident fluence=0.17mJ/mm\(^2\), pulse duration=13ns, acoustic path length=6cm.
This is somewhat higher than the value of $\mu_{\text{eff}} = 45\text{cm}^{-1}$ obtained by inserting the values $\mu_s = 12\text{cm}^{-1}$, $\mu_a = 505\text{cm}^{-1}$, $\gamma = 0.91$ for the media from table 3.1 into equation 2.16. Figure 3.3(a) also shows that there is significant penetration of the light with a detectable photoacoustic contribution existing more than $0.4\mu s$ after the onset of the rising edge of the initial compressional part of the thermoelastic wave. This corresponds to a depth of 0.63mm. Values of $\mu_{\text{eff}}$ were found to be consistent for the majority of the 9 specimens tested although two showed increased attenuation with $\mu_{\text{eff}}$ approaching 196cm$^{-1}$. The results in figure 3.3(a) show good repeatability in terms of the shape and amplitude of the waves between different points on the same aorta specimen. Between different specimens, however, the shape of the initial part of the waves was found to be variable with the existence of a second peak in some cases. This is most likely due to the generation of a second thermoelastic wave at the intima/media boundary. The reason why this second peak did not appear in every aorta specimen tested is that it is likely that in some cases the thickness of the intima was very small. This results in the superposition of the two thermoelastic waves so that they appear as a single wave. It is also possible that the tissue was compressed when raising the jack to the extent that the two waves were superposed. This may also account in part for variations in the measured effective attenuation coefficient.

Measurements at three points on a single large atheromatous lesion are shown in figure 3.3(b). The waveforms are more complex than for normal tissue with the presence of a second peak. The first peak is due to the thermoelastic wave generated in the intima and is therefore of a similar amplitude to that produced in the normal tissue. The second peak is generated in the upper surface of the plaque mass. The shifts in the positions of the secondary peaks are due to variations in thickness of the intimal layer over the area of the lesion. Since there are no large discontinuities in the exponentially decaying rarefaction part of the wave, it is assumed that the light penetration does not extend beyond the thickness of the plaque mass into the media. Therefore, this part of the thermoelastic wave corresponds to the photoacoustic contribution from the plaque mass and, using an exponential fit, a value of $\mu_{\text{eff}} = 239\text{cm}^{-1}$ is obtained for atheroma. For other atheromatous samples there was wide variation in the shape of the wave and the values of $\mu_{\text{eff}}$ calculated. This is due to variations in lesion thickness, composition and complexity. It was noted that $\mu_{\text{eff}}$ fell to approximately $112\text{cm}^{-1}$ (close to the value of normal tissue) in several cases. This is because the plaque mass was relatively thin and
the decaying rarefaction part of the wave to which the exponential was fitted was associated with the photoacoustic contribution from the media rather than the atheroma. To overcome this problem and obtain a clearly interpretable waveform for thin plaque layers, higher temporal resolution is required. This could be achieved by using shorter optical pulses to generate the thermoelastic waves and increasing the bandwidth of the detector.

The waveforms in figure 3.3 show that there is a clear difference in the characteristics of the thermoelastic waves generated in normal and atheromatous tissue. The preferential absorption in atheroma is demonstrated by the increased amplitude and effective attenuation coefficient. There was however considerable variation in the photoacoustic signatures between different specimens and discrimination between the two tissue types was not always obvious.

3.4.2.2 \( \lambda=461\text{nm} \)

At this wavelength, the efficiency of the Raman cell is very low. Consequently, the amplitudes of the thermoelastic waves are very close to the detection limit of the experimental arrangement. To obtain \( \mu_{\text{eff}} \), the individual waveforms shown in figures 3.4(a) and (b) were averaged and an exponential fitted to the averaged waveforms between points A and B.

Typical waveforms generated in normal tissue are shown in figure 3.4(a). Peak pressures using an incident fluence of 0.013mJ/mm\(^2\) are of the order of 2KPa. A value of \( \mu_{\text{eff}}=97\text{cm}^{-1} \) was obtained which is considerably higher than the value of \( \mu_{\text{eff}}=35.4 \) calculated from the data in table 3.1. Repeatability was found to be excellent both for different positions on the same specimen and between the 5 specimens studied.

The waveforms generated in atheroma are shown in figure 3.4(b). The amplitude of the thermoelastic waves generated are about a factor of two higher than the corresponding waveforms for the normal tissue. A value of \( \mu_{\text{eff}}=187\text{cm}^{-1} \) was obtained demonstrating the strong preferential optical attenuation at this wavelength. The characteristics of the photoacoustic signatures were found to be highly reproducible between different specimens. In each case the waveforms produced in atheroma were a simple bipolar shape indicating that the thermoelastic contribution arises essentially
Figure 3.4 Thermoelastic waves generated in post mortem human aorta at 461 nm. (a) Normal aorta (3 positions) and (b) atheromatous aorta (4 positions). $\mu_{eff}$ is calculated by fitting an exponential between A and B. Incident fluence=0.013 mJ/mm², pulse duration=13 ns, acoustic path length=6 cm.
from the plaque mass without the presence of secondary waves generated at boundaries. At this wavelength, it is concluded that a clearly identifiable and reproducible photoacoustic signature indicating the presence of atheromatous plaque can be obtained.

3.4.2.3 $\lambda=532\text{nm}$

The waveforms shown in figure 3.5 were obtained using 532nm and the same aorta specimens used to obtain the results at 436nm. In normal tissue (figure 3.5(a)), the shape of the thermoelastic waves are similar to that shown in figure 3.3(a) at 436nm. The amplitude at 532nm, however, is much smaller and the optical penetration greater due to lower absorption. This is demonstrated by the appearance of a second thermoelastic wave (S) occurring at about 0.74$\mu$s after the initial signal. This is due to the light penetrating the full thickness of the tissue and generating a thermoelastic wave in the steel block beneath the sample. $\mu_{\text{eff}}$ was calculated to be approximately 92cm$^{-1}$ compared to $\mu_{\text{eff}}=26\text{cm}^{-1}$ obtained using the data in table 3.1.

In atheromatous areas, the peak pressure amplitudes were similar to those of the normal tissue. The shape of the waves, however, is quite different with a triple peak structure due to the generation of subsurface waves in the upper and lower boundaries of the plaque mass. A thermoelastic wave is not generated in the steel block beneath the sample indicating that the light does not penetrate the full thickness of the tissue. There are two reasons for this. Firstly, the atheromatous tissue is thicker than the normal tissue and secondly, the increased amplitudes and reduced decay times of thermoelastic waves suggest significantly higher attenuation. In these particular measurements on atheromatous tissue, $\mu_{\text{eff}}$ was calculated at 168$\text{cm}^{-1}$. This is in contrast to the data of Prince et al.$^{106}$ which suggests that the difference in the optical properties of normal and atheromatous tissue at 532nm is small. When considering all of the measurements made at 532nm, a significant number did not show this increased absorption. In almost all cases, light penetrated the full thickness of a sample of normal tissue. In some cases, light penetrated atheromatous tissue although this was strongly dependent upon the thickness of the lesion.

The penetration of 532nm light through the tissue suggests that spatial information describing the structure and thickness of the tissue can be obtained.
Figure 3.5 Thermoelastic waves generated in post mortem human aorta at 532nm. (a) Normal aorta (4 positions) - S is the thermoelastic wave generated by light penetrating through full thickness of tissue to steel block beneath tissue sample. (b) Atheromatous aorta (2 positions). $\mu_{ff}$ is calculated by fitting an exponential between A and B. Incident fluence=0.17mJ/mm$^2$, pulse duration=15ns, acoustic path length=6cm.
To investigate this, the experiment was repeated using a sample in which the fat layer was not removed. A Perspex block was placed underneath the sample in place of the steel block and the jack raised until the surface of the tissue made light contact with the Perspex window. Care was taken to ensure that the tissue was compressed as little as possible so that the fat layer was at least 1cm thick. This ensured that the arrival of any acoustic reflections from the surface of the Perspex block were sufficiently delayed so that they did not coincide with any part of the initial thermoelastic wave. The results are shown in figure 3.6. A small signal (X) is generated at the adventitia/fat boundary appearing around 1µs after the initial signal (thick line). This corresponds to a depth of 1.4mm which is about the thickness of human aorta. To confirm this (and ensure that this signal is not an acoustic reflection artefact arising from the experimental set-up) the

![Figure 3.6](image)

**Figure 3.6** Thickness measurement of normal post mortem human aorta by generation of subsurface thermoelastic waves. Wave generated in adventitia at X (thick line) moves to Y (thin line) when tissue sample is compressed. R - reflection within Perspex window of experimental arrangement.
tissue was compressed a little by raising the jack slightly. The signal generated in the adventitia moved to the left (Y) as did the latter part of the initial thermoelastic wave (thin line). R is a reflection within the Perspex window and so its position remains constant. Thus, by using a wavelength that penetrates deep into the tissue, it is possible to obtain spatial information.

3.5 CONCLUSIONS

From these experiments, it is concluded that time-resolved photoacoustic spectroscopy can be used to characterise arterial tissue for the purpose of discriminating between normal and atheromatous tissue. By suitable choice of wavelength, information about the composition, structure and thickness of arterial tissues can be obtained. At 436nm, the wide variation in the photoacoustic signature makes it unsuitable for exploiting preferential absorption. In addition its relatively high absorption in both tissue types makes it unsuitable for thickness measurements by generation of subsurface thermoelastic waves. At 461nm, the photoacoustic signature of soft yellow lipid plaques was significantly different to that of normal tissue as a result of the strong preferential absorption in atheroma at this wavelength. This effect was found to be highly reproducible making photoacoustic generation at 461nm a reliable indicator of the presence of this type of plaque. At 532nm, the lower absorption enabled the light to penetrate the full thickness of the tissue producing a photoacoustic contribution from each point in the tissue. This provided information about the structure and enabled the thickness of the tissue to be determined. Subsurface reflections arising from acoustic impedance mismatches at boundaries within the tissue were not observed in any of the experiments at any wavelength in this study. Significantly higher detection sensitivities are required for this.

Although the experimental technique was calibrated with a liquid absorber of known optical properties, the absolute values of the effective attenuation coefficients obtained in normal tissue were found to be significantly higher than the values calculated from the data of Keijzer et al.112. Since reported measurements of optical
coefficients in soft tissue differ widely, it is difficult to say whether this apparent
disagreement is due to the experimental technique and analysis used or simply due to
the different morphology of the samples used in this particular study. The apparent lack
of reproducibility in the photoacoustic signatures of different specimens can also be
attributed to the wide variations in tissue morphology, particularly when studying
atheroma. It could therefore be considered that the variation in photoacoustic signatures
is an expression of the range and diversity of atherosclerotic lesions and demonstrates
the applicability of the technique to characterising complex multilayered tissues.

It is of particular interest to use a wavelength such as 532nm which penetrates the
tissue but is sufficiently strongly absorbed to generate subsurface thermoelastic waves.
Since the photoacoustic signature at this wavelength provides spatial information about
the structure of an atheromatous lesion as well as the total thickness of the tissue, it is
possible that characterisation might be carried out in this way as an alternative to
exploiting the preferential absorption characteristics in atheroma. Furthermore, since a
significant number of atherosclerotic lesions may contain little or no carotenoids,
characterisation by analysis of thickness information obtained using 532nm may prove
to be more useful when studying a wide range of lesion types. It is also considerably
easier to obtain 532nm using a frequency doubled, Q switched, Nd:YAG laser than
obtain a laser wavelengths of around 470nm.

Only soft yellow plaques have been considered in this study. The full range of
fibrous/fatty and calcified lesions require investigation. In addition, higher temporal
resolution would be useful to avoid the superposition problem in which some of the
temporal characteristics relating to the plaque mass are lost.

It has been demonstrated that photoacoustic spectroscopy can be used to
characterise arterial tissue and identify atheromatous plaque. It now remains to
implement the technique \textit{in vivo}. It is to this that the remaining chapters of this thesis
are devoted to.
The previous chapter showed that pulsed photoacoustic spectroscopy can be used to characterise arterial tissue. This chapter introduces the practical in vivo implementation of the technique for guiding laser angioplasty. It is proposed to deliver short "diagnostic" laser pulses to the vessel wall via an optical fibre and detect the resulting photoacoustic signal using an ultrasound sensor mounted at the distal end of the fibre. If the photoacoustic signature indicates the presence of atheroma, higher energy "therapeutic" pulses can then be delivered, along the same fibre, to ablate the tissue immediately in front of the fibre. The basic principles and the requirements of this concept are discussed. A key part in the development of the system is the ultrasound sensor at the end of the fibre. Whilst it is proposed to use a Fabry Perot polymer film sensor for this purpose, the current state of the art in piezoelectric and optical fibre ultrasound sensing techniques is reviewed for the purpose of comparison.
4.1 PRINCIPLES OF OPERATION

The basis of the proposed system, shown in figure 4.1, is an all-optical photoacoustic probe consisting of a delivery optical fibre and a transparent Fabry Perot polymer film ultrasound sensor mounted at its distal end. The probe is inserted into the artery and guided to the stenosed region. Low energy nanosecond diagnostic laser pulses are launched into the fibre and emerge at the other end where they are transmitted through the sensor and absorbed in the vessel wall. Thermoelastic waves are generated immediately adjacent to the tip of the fibre and detected by the Fabry Perot ultrasound sensor. The characteristics of the thermoelastic waves are then used to determine whether high energy therapeutic laser pulses to ablate the obstructing material should be delivered along the delivery fibre. The sensor itself comprises a thin (~ 50μm) transparent polymer film acting as a low finesse Fabry Perot interferometer which is

![Proposed scheme for photoacoustic guided laser angioplasty.](image)

*Figure 4.1 Proposed scheme for photoacoustic guided laser angioplasty.*
illuminated with light coupled into the delivery fibre from a CW wavelength tunable low power laser diode. The stress due to the incident thermoelastic wave modulates the thickness of the film and hence the optical phase difference between the Fresnel reflections from either side of the film. This produces a corresponding intensity modulation in the light reflected from the sensing film that is transmitted back along the fibre for detection at a photodiode. For optimum sensisitivity and linearity, the wavelength of the laser diode is adjusted so that the interferometer is at quadrature.

This scheme was originally envisaged by Mills in 1988, and it was proposed then that the laser system would consist of a frequency doubled Nd:YAG laser in conjunction with a Raman shifter. With the Nd:YAG laser operating in Q switched mode, this system would emit nanosecond diagnostic laser pulses in the waveband of preferential absorption in atheroma. The same laser could also be used to provide the high energy therapeutic laser pulses by operating it in fixed Q mode (at the fundamental wavelength of 1.06µm) in order to ablate the obstructing material. Advances in laser angioplasty and the development of new laser technology have meant that a different laser system would probably be employed if the scheme were implemented today. The ablative therapeutic pulses would most likely be provided by an excimer laser. The diagnostic pulses could still be obtained using a frequency doubled Q switched Nd:YAG-Raman shifter system or, alternatively, a tunable source such as a dye laser or an OPO (optical parametric oscillator) might be used. The basic principles of the system described above would still apply, the only difference is that the same laser would not be used to provide both diagnostic and therapeutic optical pulses.

It is necessary to deliver the diagnostic and therapeutic laser pulses to the obstruction or vessel wall without being attenuated by the surrounding blood. The distal end of the probe could be placed in contact with the tissue so that the surrounding blood is pushed away from the targeted area. Alternatively, the blood could be flushed away by perfusing the vessel with optically clear saline to allow delivery of the laser pulses.

Figure 4.1 shows a forward looking probe negotiating a complete occlusion. Where it is necessary to obtain information orthogonal to the fibre axis, a sideways looking probe could be configured by using a minature prism at the tip of the fibre to deflect the diagnostic and therapeutic laser pulses through 90°.
4.2 REQUIREMENTS

The requirements of the photoacoustic probe are divided into those of the delivery optical fibre and those of the ultrasound sensor at the distal end of the fibre.

4.2.1 Delivery optical fibre

The delivery optical fibre should be capable of transmitting nanosecond laser pulses of the required energy without sustaining optical damage. The damage threshold of a fused silica optical fibre for a pulse duration of approximately 5ns lies in the range 30mJ/mm² at 392nm to 100mJ/mm² at 532nm\textsuperscript{118,119}. A 10mJ pulse at 532nm of 10ns duration, such as that produced by a Q switched frequency doubled Nd:YAG laser, can therefore be safely launched into and transmitted along a 400\textmu m core all-silica optical fibre. These damage thresholds are significantly above the damage threshold of tissue (section 3.3.3 in chapter 3). Delivering the therapeutic laser pulses should not present a problem as 400\textmu m core all-silica optical fibres are commonly used for laser angioplasty. The outer diameter of a 400\textmu m core optical fibre is typically 1mm. It is therefore small enough for insertion into the coronary artery.

4.2.2 Sensor

**Sensitivity** From the measurements in chapter 3 it is apparent that, to detect thermoelastic waves generated in arterial tissue using sub-damage threshold laser pulses, the sensor should have an acoustic noise floor of the order of 5KPa in a measurement bandwidth of 25MHz. For adequate signal to noise ratios the system sensitivity should be of the order of 50mv/MPa.

**Frequency response** A uniform frequency response is important because the photoacoustic characterisation of arterial tissue is based upon the analysis of the...
temporal characteristics of the thermoelastic wave. Any distortion of the photoacoustic signal that is introduced by non-uniform frequency characteristics of the sensor will affect the temporal characteristics of the thermoelastic wave and hence the tissue discrimination capabilities of the technique. Ideally, the sensor should have a flat frequency response which, from the frequency domain measurements described in section 2.4.2, should be to at least 30 MHz.

**Linearity** Very wide dynamic range is not required. The most stringent linearity requirements arise when detecting subsurface generated thermoelastic waves to obtain spatial information relating to the structure of the tissue. This is because the thermoelastic waves generated beneath the surface of the tissue are at least an order of magnitude lower than the thermoelastic wave generated at the surface. Of the order of 20dB of linearity is therefore required.

**Physical requirements** Since the proposed photoacoustic probe relies on the sensor being mounted directly over the tip of the distal end of the fibre it is essential that the sensor is transparent to the diagnostic and therapeutic laser pulses and can transmit these pulses without sustaining damage. A miniature flexible configuration is required for intravascular use - the outer diameter of the sensor head at the end of the fibre should be less than 1mm for insertion into the coronary artery. All components in the sensor head should be inert and capable of being sterilised.

### 4.3 REVIEW OF EXISTING ULTRASOUND DETECTION TECHNOLOGY

This section reviews and compares existing piezoelectric and optical fibre methods of ultrasound detection and discusses their suitability for use in a probe type configuration. Discussion of the non-optical fibre techniques is limited to piezoelectric devices. Alternative ultrasound sensing methods such as EMATs (electromagnetic acoustic transducers), capacitance transducers and various optical techniques are not discussed.\textsuperscript{120,121}
4.3.1 Piezoelectric ultrasound detection

The majority of piezoelectric ultrasound transducers are fabricated from piezoceramic materials such as lead zirconate titanate (PZT) or lithium niobate (LiNbO₃) or piezoelectric polymers such as polyvinylidene difluoride (PVDF). A conservative estimate for the minimum detectable acoustic pressure of a PVDF ultrasound transducer is about 1KPa (over a measurement bandwidth of 25MHz). The detection sensitivity, for a given acoustic frequency, of an optimised piezoceramic transducer such as PZT is between a factor of 5 and 10 higher than that of a correspondingly optimised PVDF transducer\(^\text{122}\). The high sensitivity of a PZT transducer arises, in part, from its highly resonant behaviour due to its high acoustic impedance compared to that of water. This large acoustic impedance mismatch between the transducer and water can, however, result in very non-uniform frequency response characteristics giving a poor representation of the acoustic signal. In contrast to ceramics, polymers tend to have acoustic impedances that are close to that of water. For this reason PVDF transducers exhibit a much more uniform frequency response. Common to transducers fabricated from both types of material are problems associated with the electrical loading imposed by the connecting wires. These include reduced sensitivity and signal distortion which become increasingly prevalent as the active area of the transducer is reduced. Furthermore, since the transducer cannot drive a coaxial cable terminated in 50 Ω, cable resonances can occur due to reflected electrical power at the cable termination resulting in signal distortion. Susceptibility to electromagnetic interference, cost and fragility are further disadvantages of piezoelectric devices. For in vivo use, the safety implications of electrical wires and connections within the body must also be considered.

Despite these limitations, the high sensitivity and wide bandwidth of piezoelectric ultrasound transducers have led to their use in various biomedical photoacoustic techniques\(^\text{83,85,73}\). Furthermore, the development of intravascular ultrasound devices for imaging the vessel wall has shown that the small transducer size required for insertion into a blood vessel is achievable. It would therefore seem that a piezoelectric transducer could be mounted at the end of an optical fibre to form a piezoelectric equivalent of the photoacoustic guided laser angioplasty scheme outlined in section 4.1. There is however a fundamental limitation when using a piezoelectric transducer in this way. This arises from the fact that the transducer at the tip of the fibre must be laterally offset from the
fibre axis to avoid obscuring the diagnostic laser pulses. The problem is illustrated by
the photoacoustic probe reported by Chen et al. In this scheme, the delivery optical
fibre passes through the centre of a PVDF disc transducer situated a few mm from the
end of the fibre. The end of the fibre has to be withdrawn a short distance from the
target to ensure that the divergent beam emerging from the end of the fibre produces a
photoacoustic source that is of larger diameter than the fibre. If the fibre is not
withdrawn, the tip of the fibre will be positioned directly over the source, blocking the
thermoelastic wave thus preventing any part of the photoacoustic signal being detected.
When the fibre is withdrawn from the target, only the contribution originating from the
periphery of the circular photoacoustic source is received. The contribution from the
central region (of diameter equal to that of the fibre) is obscured by the fibre. It is,
however, from this obscured central region that information about the tissue is
principally required because, for a small source-detector separation (as would be the case
in the restricted environment of a blood vessel), most of the ablative laser energy is
concentrated on this area. A further disadvantage arising from the separation of the
source and detector is that a true near field acoustic measurement cannot be made. This
can result in distortion to the thermoelastic wave due to diffraction effects.

4.3.2 Optical fibre ultrasound detection

The optical fibre ultrasound sensors discussed in the following sections are divided
into intrinsic and extrinsic classifications. Intrinsic sensors are those which rely upon
acoustically-induced strains in the fibre producing a change in the phase, intensity or
polarisation of the light guided by the fibre - i.e. the optical fibre itself acts as the
transducer. Extrinsic sensors are those which use the fibre simply to deliver light to and
from an optical sensor which is then acted upon by the acoustic field to produce the
change in phase, intensity, or polarisation in the light. Although this section is
specifically concerned with optical fibre ultrasound sensors, the general principles of
optical fibre sensing are applicable in many cases. These can be found in a number of
reviews in the literature.

A difficulty that has been encountered in evaluating the performance of the optical
fibre ultrasound sensors described in the literature is that many are "proof of principle"
systems. Typically, these may show that a particular system can detect ultrasound but
often a calibrated reference detector has not been used as a comparison. Acoustic noise floors, sensitivity and bandwidth data are therefore often absent making it difficult to make meaningful comparisons between different types of optical fibre sensors and piezoelectric transducers.

**Intrinsic sensors.** Since the first demonstration of an optical fibre acoustic sensor in 1977 by Bucaro et al. it has been substantial development of intrinsic optical fibre acoustic sensors. Most are polarimetric or interferometric devices. Polarimetric sensors rely on acoustically-induced birefringence to modify the state of polarisation of the light guided within the fibre. Interferometric sensors exploit acoustically-induced strains that change the length and refractive index of the fibre and hence the phase of light guided within the fibre. Most of the work on optical fibre acoustic sensing has concentrated on the low frequency end of the acoustic spectrum (< 50 KHz) with the main user being the defence industry for use in sonar applications. Considerably less work has been carried out on the development of sensors for ultrasonic acoustic detection. Early fibre optic ultrasound sensors tended to make use of the principles established for low frequency optical fibre acoustic sensing. Low frequency acoustic sensors often employ a coil of fibre of many turns so that very long interaction lengths (100's of metres) can be obtained resulting in high sensitivity. Applying such a strategy to acoustic detection at ultrasonic frequencies is less successful because it becomes difficult to reduce the spatial size of the sensor coil so that it is small compared to the wavelength of the incident acoustic wave. To keep the dimensions of the sensing element small, only a single length of fibre is usually placed in the ultrasound field. The interaction length is therefore that of the acoustic beam width, typically no more than a few centimetres. Such a small interaction length removes an important attraction of the intrinsic optical fibre acoustic sensor: that of high sensitivity. Whilst it is possible to detect ultrasound with intrinsic optical fibre sensors, the sensitivity rarely approaches that of piezoelectric devices. This tends to limit their advantages to immunity to EMI and the rugged, inexpensive nature of the sensing element rather than acoustic performance.

The first optical fibre ultrasound sensor was an intrinsic polarimetric device reported by De Paula et al. of the Naval Research Laboratories (NRL), Washington D.C. in 1982. This was followed in 1983 by a system from the same group in which acoustically induced phase shifts were detected using a Mach Zender interferometer.

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In this scheme an interaction length of 1cm of single mode fibre was used to sense between 0.5 MHz and 50 MHz with a sensitivity of the order of 1 rad/MPa. The phase noise floor is not given but a typical value for a Mach Zender interferometer\textsuperscript{124} can be taken as 5\(\mu\)rad/\sqrt{Hz}. Over a measurement bandwidth of 50 MHz this corresponds to an acoustic noise floor of approximately 35KPa. The work carried out by NRL on optical fibre ultrasound sensing continued with an investigation into the ultrasonic sensitivity of optical fibres\textsuperscript{135,136,137}. The use of an embedded fibre optic ultrasound sensor for the non destructive testing of composite materials has also been investigated. Liu et al.\textsuperscript{138} describe the implementation of a Michelson fibre interferometer for this application whilst Alcoz et al.\textsuperscript{139} describe a low finesse Fabry Perot configuration formed by the fabrication of internal dielectric mirrors in the fibre. Detection between 100KHz and 5MHz is reported in the latter case with an optimum acoustic sensitivity of 3.3 rad/MPa for an 11mm interaction length at 200KHz. This is in approximate agreement with the acoustic sensitivity reported by De Paula et al.\textsuperscript{134}. An almost identical scheme\textsuperscript{140} and its active stabilisation technique\textsuperscript{141} have also been described although no useful information regarding performance was given. A novel approach to intrinsic ultrasound sensing has been demonstrated in the form of a highly sensitive ultrasonic hydrophone based upon a Sagnac interferometer\textsuperscript{142}. A very low acoustic noise floor of 0.1 Pa (over a 1MHz bandwidth) on account of its long fibre interaction length (42m) was reported. The relatively large size (2.6mm x 32mm) of this device however precludes its use for frequencies much above 1 MHz. The interferometric sensors described above can offer useful sensitivity and some are probably capable of detecting down to a few tens of KPa with a frequency response in the region of tens of MHz. In general however, the sensitivity is at least an order of magnitude lower than that of piezoelectric transducers. Intrinsic optical fibre sensors can also require somewhat complex phase and polarisation control systems to maintain the optimum working point of the sensor\textsuperscript{143,144}. Polarimetric sensors\textsuperscript{145,146,147} offer an alternative that is simpler in implementation. In one system, the output of a high power hyperthermia transducer\textsuperscript{148} was measured and a system sensitivity of 12mv/MPa per mm of interaction length at 1MHz was obtained. No figure for the noise floor was given. The best guide to the sensitivity of polarimetric sensors is a theoretical acoustic noise floor prediction of 0.5Pa/\sqrt{Hz} corresponding to 3.5KPa over 50MHz measurement bandwidth given by De Paula's original paper\textsuperscript{133}. A clear experimental verification of this low acoustic noise floor does not appear in the
literature. Cursory inspection of the waveforms produced by a PVDF membrane hydrophone and the polarimetric sensor of Chan et al\textsuperscript{48} does, however, suggest that this figure may be feasible. This would make the polarimetric sensor more sensitive than interferometric types.

Whilst it is possible that some of the intrinsic sensors described above could have the required sensitivity for this application there are a number of limitations. These are associated with both the acoustic performance and the practicalities of configuring the fibre in a probe type configuration. Uniformity of frequency response when using the fibre itself as the sensing element is generally poor due to the large acoustic impedance mismatch between the fused silica of the fibre and water. If measurements at a point are required, a further limitation arises from the fact that the active length of the fibre responds to the line integral of the acoustic field. If the phase of the ultrasound field over the region intercepted by the fibre is not the same at all points along the fibre, inaccuracies in the measurement of the acoustic field will be introduced. To overcome this problem it would be necessary to localise the active part of the fibre in some way so that its dimensions were significantly less than the acoustic wavelength. For acoustic frequencies of the order of several tens of MHz this would require restricting the interaction length to a few hundred microns. This would be difficult to achieve in systems where the active part of the fibre is non-localised and defined by the acoustic beam width. It would also reduce the acoustic sensitivity by about two orders of magnitude. This last factor would rule out the use of even the localised intrinsic Fabry Perot interferometer type sensors\textsuperscript{138,139,140} at the end of the fibre. Furthermore, many of the intrinsic sensors also require orthogonal orientation to the incident acoustic wave. For all of the schemes described above, single mode fibre was used. Single mode fibre could not be used to transmit the diagnostic and therapeutic laser pulses without damage. Multimode fibres that could deliver the laser energy cannot be used in place of the single mode fibres because of the problems associated with decoding polarisation and phase information from the complex and randomly varying multimode radiation pattern that emerges from the fibre\textsuperscript{149,150}.

A review of optical fibre sensors would be incomplete without commenting upon the rapid development of Bragg grating sensors in recent years. These have yet to make an impact in ultrasound detection although two groups are currently investigating acoustic measurements with these devices\textsuperscript{151,152}. They could be used to replicate the
Fabry Perot sensors by writing a pair of gratings into the fibre. Alternatively, the grating itself could be used as the active element by observing the light reflected from the grating as a function of acoustic strain. An exciting possibility is the prospect of multiplexing a number of sensors along a single optical fibre downlead - a long sought after goal in many areas of optical fibre sensing.

Extrinsic sensors. For this application the concept of an extrinsic sensor is more suitable. Since the optical fibre simply transmits light to and from the sensor at the end of the fibre, the difficulties associated with localising the sensing element that affect many of the intrinsic devices do not apply. Furthermore, the type of fibre (single/multimode) used is relatively unimportant because it does not play a direct role in the sensing mechanism. This allows the possibility of using large diameter multimode fibres suitable for transmitting the diagnostic and therapeutic laser pulses. Furthermore, in contrast with the intrinsic sensors described above, the requirement that the sensor is orientated perpendicularly to the acoustic field does not necessarily apply. Few extrinsic ultrasound sensors appear in the literature although a number of Fabry-Perot pressure sensors for measuring low frequency pressure variations have been reported\textsuperscript{153,154,155}. In the medical field, for example, they have been used for the measurement of intra-arterial blood pressure. Their construction generally utilises a reflective pressure sensitive diaphragm to form one side of a Fabry-Perot cavity and a reflective parallel surface separated by a short distance to form the other. The cavity is attached to the distal end of the fibre. As the diaphragm moves in response to pressure the reflected interference pattern varies. The concept has been used for acoustic detection although has not been demonstrated above a few hundred kilohertz\textsuperscript{156}. Other Fabry-Perot sensor designs, most notably for temperature sensing, have made use of a solid étalon to form the cavity\textsuperscript{157} in which the variation in the thickness of the étalon produces the change in the interference pattern. A solid Fabry Perot cavity lends itself more readily to high frequency acoustic sensing as the inertial limitations of the former approach are avoided. Despite this, no descriptions of acoustic sensors of this type (apart from those arising as a result of the work described in this thesis) appear to have been reported in the literature. The use of pressure-induced birefringence of photoelastic materials for acoustic sensing in the sub 20 KHz range has been reported\textsuperscript{158,159,160}. These sensors make use of an input and output fibre and require orthogonal orientation to the acoustic field.
The use of Raman-Nath diffraction\textsuperscript{161} for ultrasound detection has been described but this scheme also requires a transmitting and receiving fibre, is of low sensitivity and has limited application. An ultrasound sensor based upon pressure-induced changes in the Fresnel reflection coefficient at the tip of a plastic or fused silica fibre for shock wave measurements during lithotripsy has been reported\textsuperscript{162,163,164}. From a source-detector and simplicity of configuration point of view, this concept would be ideal for the proposed application. It also has potentially extremely wide bandwidth. Unfortunately its low sensitivity (acoustic noise floor $\sim 1\text{MPa}$ over a 20MHz measurement bandwidth) precludes it from most applications other than high amplitude shockwave detection.

More relevant are systems that are based upon the interferometric measurement of acoustically induced displacements of a thin pellicle. One of the first systems based upon this idea was implemented by the National Physical Laboratory (NPL) using a bulk Michelson interferometer for the primary calibration of ultrasonic hydrophones\textsuperscript{165}. An optical fibre Michelson interferometer version of this has been described for measuring laser generated thermoelastic waves\textsuperscript{166,167}. An acoustic noise floor of 0.5KPa over a 20MHz bandwidth was reported showing that the system is capable of high sensitivity. A similar method of optical fibre ultrasound detection employing a pellicle is also under development by NPL for measuring acoustic particle velocity. Its acoustic noise floor is much higher at approximately 250KPa\textsuperscript{168} and is intended for high amplitude applications such as lithotripsy measurements\textsuperscript{169}. A disadvantage of these techniques is that both arms of the interferometer are sensitive to ambient environmental fluctuations. Fairly complex active phase bias and polarisation control techniques are therefore required for stable operation increasing the complexity and expense of the systems.

Possibly the most relevant system reported in the literature is the low finesse Fabry-Perot sensor for surface acoustic wave detection as described by Tran \textit{et al.}\textsuperscript{170}. The cavity is formed by the separation of two fibres in a connecting silica tube. The Fresnel reflections at the air/glass interface of the cleaved ends of the fibre form the mirrors of the cavity. The sensor is therefore effectively transparent and could be configured as a photoacoustic probe by mounting it on the end of a multimode optical fibre along which the diagnostic and therapeutic laser pulses could be delivered. Acoustic strains in the silica tube alter the separation between the two fibres producing an intensity change in the light reflected from the sensor. It was demonstrated at a spot frequency of 1MHz only and the sensitivity is quoted as $6.98 \times 10^{-2} \text{ rad/\micro strain/cm.}$
Signal-to-noise ratios of 38dB are reported although these are quoted without giving the pressure or intensity of the ultrasound wave making it difficult to estimate the true detection limit. The Fabry-Perot cavity length is 3μm and the length of the tube in which the fibres are housed is 1cm in length. The theoretical analysis described uses the cavity depth as the acoustically sensitive interaction length of the sensor. In other words the strain is integrated over the cavity depth rather than the full length of the sensor head. One would therefore expect very low acoustic sensitivity although the results show a good S/N ratio when measuring surface acoustic waves generated by a commercially available PZT transducer. It is possible that using the cavity depth as the interaction length is incorrect and the strain should in fact be integrated over the full length of the sensor head - this would be the case if the construction, perhaps unintentionally, was such that the fibres were firmly bonded only at the ends of the silica connecting tube. Further evidence for integrating over the entire sensor head is provided when temperature sensitivity is considered. The temperature sensitivity of glass is approximately 200rad/ °C/m (see section 5.7 in the next chapter). If only 3μm of the sensor were responsible for thermal strains then to produce a phase shift of π/2 a very large temperature rise of 2617 °C would be required. If the interaction length were 1cm however the same phase shift would be produced by 0.8 °C. The authors of the paper show the sensor cycling in and out of quadrature over temperature changes of less than 10°C and in fact it was deemed necessary to implement a dual wavelength scheme of interrogation to provide some compensation for thermal effects. This would certainly have been unnecessary if the interaction length was 3μm. If the entire length of the sensor head is indeed acoustically active then, to measure thermoelastic waves with frequency components in the tens of MHz range, the sensor would have to be reduced in length by two orders of magnitude. This would reduce the sensitivity by the same amount and make fabrication very difficult. A further disadvantage relates to the "acoustic complexity" of the sensor head. There are several air/glass interfaces (not to mention glass/adhesive interfaces) presenting the possibility of axial and lateral resonances in the the silica tube that connects the two fibres. These would result in poor frequency response characteristics.

It would appear that, conceptually at least, extrinsic ultrasound sensors are more suitable for incorporation into a photoacoustic probe than intrinsic sensors. Of existing designs however, none appear to fulfill the requirements of the single delivery fibre
photoacoustic probe discussed in section 4.2. It is considered that the concept of the Fabry Perot polymer film ultrasound sensor is the best option because it can provide a transparent localised sensor element that can be mounted directly over the end of the fibre. Furthermore the use of a polymer provides a sensing element that has an acoustic impedance close to that of water offering the prospect of uniform frequency response characteristics. The sensing element can be of thickness significantly less than the acoustic wavelength of the thermoelastic wave and because the Young's modulus of polymers is low, sensitivity will be high compared to the optical fibre sensors that use glass or fused silica as the sensing element.

4.4 CONCLUSION

This section has outlined the basic principles of operation of the proposed optical fibre photoacoustic-guided laser angioplasty scheme and discussed various options for the detection of the photoacoustic signal at the tip of the fibre. Whilst piezoelectric transducers can fulfill to some degree the performance criteria in terms of sensitivity and bandwidth and the required degree of miniaturisation, the source-detector geometry of such transducers would impose a fundamental limitation. Of the optical fibre ultrasound sensors that have been reviewed none appear to fulfill simultaneously the performance and source-detector geometry requirements. For this reason it is considered that the proposed transparent Fabry Perot polymer film sensor is the most suitable option for this application.
CHAPTER 5

THE FABRY PEROT POLYMER FILM ULTRASOUND SENSOR - THEORY

In this chapter, the basic principles of operation and the theoretical aspects of the low finesse Fabry Perot polymer film sensor are described. The optical output of a low finesse Fabry Perot interferometer is examined in order to derive the interferometric phase sensitivity. The acoustic sensitivity and frequency response are modelled by considering the interaction of the sensing film with an acoustic field. The interferometric and acoustic parameters are then brought together to give the overall system sensitivity. This allows the output signal of the sensor for a given acoustic wave to be predicted. To obtain a measure of the acoustic noise floor and typical signal to noise ratios, an analysis of photodiode and laser noise is presented. Finally, the temperature sensitivity and phase bias control requirements for maintaining quadrature are discussed.
5.1 PRINCIPLES OF OPERATION

Figure 5.1 Scheme for the optical fibre detection of ultrasound using a Fabry Perot polymer film sensor.

A schematic of the system for the optical fibre detection of ultrasound is depicted in figure 5.1. Light from a wavelength tunable laser diode is launched into a multimode optical fibre downlead. The sensing element, mounted at the end of the fibre, comprises a thin (typically a few tens of microns thick) transparent polymer film acting as a low finesse Fabry Perot interferometer. The optical reflection coefficients of the "mirrors" of the interferometer are defined by the Fresnel reflection coefficients arising from the refractive index mismatches between the two sides of the film and the surrounding media. The stress due to an incident acoustic wave modulates the thickness of the film and hence the optical phase difference between the two Fresnel reflections. A corresponding intensity modulation is produced in the light reflected from the film which is then transmitted back along the fibre for detection at a photodiode.
Figure 5.2 Intensity output $I_o$ of a low finesse Fabry Perot interferometer as a function of phase difference $\phi$. Diagram illustrates operation at quadrature for the linear detection of an acoustic signal. $dI_o$ - signal intensity modulation resulting from a small acoustically-induced phase modulation $d\phi$.

For optimum sensitivity and linear operation, it is necessary to adjust the phase bias of the interferometer so that it is operated at the so-called quadrature point as depicted in figure 5.2. In practice, this can be achieved by thermally tuning the wavelength of the laser diode.

5.2 LOW FINESSE FABRY PEROT INTERFEROMETER OUTPUT

In this section, the optical output of a transparent polymer sensing film acting as a low finesse Fabry Perot interferometer is considered. A key parameter in describing the performance of the sensor is the interferometric phase sensitivity. This gives a measure of the intensity modulation produced by an acoustically-induced phase modulation. In addition to the signal intensity modulated output, there is also a dc
component which plays a critical role in the optimisation of the sensor performance. With the aim of obtaining the maximum signal intensity modulation for a given phase modulation, the factors that affect the phase sensitivity and dc level and their interrelation are examined.

In the first instance, a sensing film illuminated by coherent light at normal incidence is considered. Figure 5.3 shows a polymer sensing film of refractive index \(n\) and thickness \(l\) in contact with two different media: on one side medium 1 of refractive index \(n_1\) and on the other, medium 2 of refractive index \(n_2\). The Fresnel reflection coefficients arising from the refractive index mismatches at the boundaries of the sensing film are assumed to be sufficiently low (<0.05) to permit the contribution of multiple reflections within the interferometer to be neglected.

Figure 5.3 Polymer sensing film acting as a low finesse Fabry Perot interferometer.

The analysis is therefore that of a two beam interferometer. The resultant intensity \(I_o\) reflected from the interferometer can be found by considering the superposition of the reflections \(I_1\) and \(I_2\) from the two sides of the film. For light at normal incidence on the sensing film, \(I_o\) is given by
where $\Phi$ is the total phase difference arising from the optical path length difference between the two reflections. Appendix A gives a full derivation of equation 5.1 based upon the treatment by Hecht and Zajac\textsuperscript{171}.

### 5.2.1 Phase sensitivity and dc level

When an acoustic wave is incident on the sensing film, $\Phi$ in equation 5.1 consists of two components: the unsignalled phase bias term $\phi$ that defines the working point of the interferometer and a time varying signal term $d\phi$ that arises from the acoustically-induced change in thickness of the sensing film.

$$\Phi = \phi + d\phi$$ (5.2)

Inserting equation 5.2 into equation 5.1 and expanding gives

$$I_o = I_1 + I_2 + 2\sqrt{I_1 I_2} \cos \Phi \cos \phi - \sin \phi \sin d\phi$$ (5.3)

For optimum sensitivity and linearity it is necessary to set the phase bias term $\phi$ at quadrature where $\phi = (2m+1)\pi/2$ (for integer $m$) by tuning the wavelength of the laser source. At the first quadrature point $\phi = \pi/2$ and, for small $d\phi$, equation 5.3 reduces to

$$I_o = I_1 + I_2 - 2\sqrt{I_1 I_2} d\phi$$ (5.4)

Under these conditions, the output of the interferometer that is detected by the photodiode consists of an acoustically-induced time varying intensity modulated term $dI_o$ that is linearly dependent upon $d\phi$ and a dc component $I_{dc}$

$$dI_o = -2\sqrt{I_1 I_2} d\phi$$ (5.5)

$$I_{dc} = I_1 + I_2$$ (5.6)

$I_1$ and $I_2$ can be written in terms of the incident intensity $I$ and the Fresnel reflection coefficients $r_1$ and $r_2$ defined by the refractive index mismatches at the boundaries of the sensing film where
\[ I_1 = I r_1 \quad I_2 = I (1 - r_1)^2 r_2 \]  \hspace{1cm} (5.7)

and

\[ r_1 = \left( \frac{n - n_1}{n + n_1} \right)^2 \quad r_2 = \left( \frac{n - n_2}{n + n_2} \right)^2 \]  \hspace{1cm} (5.8)

From equations 5.5 and 5.6, the phase sensitivity of the interferometer, defined as the magnitude of the intensity modulation per unit phase shift \( dI_1/d\phi \), and the dc level \( I_{dc} \), can now be written

\[ \frac{dI}{d\phi} = 2I(1 - r_1)\sqrt{r_1 r_2} \]  \hspace{1cm} (5.9)

\[ I_{dc} = I(r_1^2 + (1 - r_1)^2 r_2) \]  \hspace{1cm} (5.10)

It is desirable to maximise the phase sensitivity by suitable choice of \( I \), \( r_1 \) and \( r_2 \) but it is important to recognise that these parameters also affect the dc component. This is an undesirable component of the interferometer output as it produces a large photocurrent with attendant shot noise that can dominate the noise characteristics of the photodiode. More importantly, if the dc level is too high it will saturate the photodiode limiting the maximum phase sensitivity that can be achieved by increasing \( I \). The choice of the optimum values of \( r_1 \) and \( r_2 \) are discussed in the next section.

5.2.2 Optimisation of reflection coefficients

Different criteria may be used to optimise the reflection coefficients \( r_1 \) and \( r_2 \) in order to obtain the maximum signal intensity modulation. First, consider optimising the reflection coefficients solely on the basis of obtaining the maximum phase sensitivity. By differentiating equation 5.9 it can be shown that the maximum normalised value of \( dI_1/d\phi \), for any given value of \( r_2 \), occurs when \( r_1 = 1/3 \). This can be seen in figure 5.4(a). The maximum phase sensitivity is, however, in many cases not the only consideration when selecting \( r_1 \) and \( r_2 \). The dc component of the interferometer output \( I_{dc} \) is also detected along with the signal intensity modulation \( dI_1 \). If \( I_{dc} \) is too high, it can saturate
the photodiode completely. It is therefore useful to plot the normalised dc level as a
function of \( r_j \) and \( r_2 \) as shown in figure 5.4(b) and use this in conjunction with the
normalised phase sensitivity obtained from figure 5.4(a) to select values of \( r_j \) and \( r_2 \).
This provides a means of obtaining the highest phase sensitivity whilst ensuring that the
dc level is below the saturation threshold of the photodiode. This procedure is useful if,
for example, there is limited laser power available or there are limitations on the choice
of \( r_j \) and \( r_2 \).

If there is the freedom to choose any value of \( r_j \) and \( r_2 \) and there is adequate laser
power available, the optimum compromise is represented by the highest phase sensitivity
that can be achieved with the lowest possible dc level. This can be expressed by
defining a figure of merit \( M \) as the ratio of the phase sensitivity to the dc level

\[
M = \frac{dI_\phi/d\phi}{I_{dc}} \tag{5.11}
\]

Substituting from equations 5.9 and 5.10

\[
M = \frac{2(1-r_j)\sqrt{r_1}r_2}{r_1+(1-r_j)^2r_2} \tag{5.12}
\]

At the maximum of each curve in figure 5.5, the reflection coefficients are those that
yield the optimum figure of merit \( (M=1) \) in terms of sensitivity and dc level. If \( I_j \) and
\( I_2 \) are now substituted back in to equation 5.12, it can be seen that the figure of merit
\( M \) is in fact recognisable as the fringe visibility.

\[
M = \frac{2\sqrt{I_1I_2}}{I_1+I_2} = \frac{I_{max} - I_{min}}{I_{max} + I_{min}} \tag{5.13}
\]

where \( I_{max} \) and \( I_{min} \) are the maxima and minima intensities of the interferometer output
defined by equation 5.1 and shown in figure 5.2. If \( r_j \) and \( r_2 \) are such that \( M=1 \), the
incident laser intensity can be increased to its maximum level before the dc level
saturates the photodiode thus enabling the highest possible phase sensitivity to be
achieved.
Figure 5.4 (a) Normalised phase sensitivity $dl_o/d\phi$ and (b) normalised dc level $I_{dc}/I$ for different reflection coefficients $r_1$ and $r_2$. 
The question of which criteria should be used to select $r_1$ and $r_2$ is considered in the following numerical examples. Suppose the rear face has a completely reflective coating applied to it so that $r_2=1$. For maximum sensitivity figure 5.4(a) shows that $r_1=0.33$ giving $dI/d\phi=0.778I$. The corresponding dc level from fig 5.4(b) for these values of $r_1$ and $r_2$ is $0.778I$. If we now choose $r_1$ by looking for the maximum fringe visibility in figure 5.5 we get $r_1=0.38$, and by substituting into equations 5.9 and 5.10, $dI/d\phi=0.764I$ and $I_d=0.764I$. In this particular case it makes little difference as to which method is chosen to select $r_1$. For lower values of $r_2$, for example $r_2=0.1$, using Fig 5.4, $r_1=0.33, dI/d\phi=0.243I$, and $I_d=0.379I$. Based on fringe visibility, for the same value of $r_2$, the optimum value is now $r_1=0.075$. This gives $dI/d\phi=0.160I$ and a dc level also of $0.160I$. Thus $dI/d\phi$ is down by about 40% when using fringe visibility as the basis for optimisation and the dc level is reduced by over 60%. In this case, if only limited laser power was available or the detector had a particularly high saturation threshold (so that

Figure 5.5 Fringe visibility $M$ for different reflection coefficients $r_1$ and $r_2$. 

![Graph showing fringe visibility M for different reflection coefficients r1 and r2.](image)
there was no prospect of saturating the photodiode), the effects of the dc level could be
neglected. Simply selecting \( r_j \) and \( r_z \) to give \( M=1 \) would not result in the maximum
signal and it would be advantageous to choose \( r_j \) and \( r_z \) on the basis of phase sensitivity.
If however there is sufficient laser power, it would be advantageous for \( r_j \) and \( r_z \) to be
such that the fringe visibility is equal to 1. The laser power could then be increased to
just below the photodiode saturation threshold thus yielding the the maximum possible
phase sensitivity. For the intended application, the sensing film has to be effectively
transparent to allow transmission of the diagnostic and therapeutic laser pulses. The
reflection coefficients will therefore be defined by the small Fresnel reflection
coefficients \(<0.05\) due to the refractive index mismatches at the two faces of the
interferometer. Given that several tens of mW of laser power are readily available from
laser diode sources, and the saturation threshold of a typical photodiode-amplifier
combination is of the order of 100\( \mu \)W, it is likely that the detection saturation threshold
will be the limiting factor. Hence optimum performance will be achieved when \( M=1 \).
For small \( r_j \) and \( r_z \), figure 5.5 shows that for \( M=1 \), \( r_j \) and \( r_z \) should be approximately
equal.

5.2.3 Phase sensitivity of different polymer sensing films

The previous section established that, for the proposed application, \( r_j \) and \( r_z \),
should be chosen to give a fringe visibility of 1 for optimum phase sensitivity. A
convenient means of achieving this is to arrange for the illuminated side of the sensing
film (the left hand side in figure 5.3) to be in contact with water. Since the sensor is to
be used for measuring ultrasound in water, both sides will then have the same (small)
Fresnel reflection coefficients \( (r_j=r_z) \) which, as discussed in section 5.2.2, result in a
fringe visibility of 1. This is the basis on which the phase sensitivities for different
sensing film materials in table 5.1 are calculated for an incident intensity \( I=1 \)mW.

Table 5.1 shows that there is considerable variation in the phase sensitivities of the
different materials despite relatively small differences in refractive index. Equation 5.8
shows that the Fresnel reflection coefficient is related to the square of the refractive
index mismatch between the sensing film and water. When \( r_j=r_z \), the phase sensitivity
(equation 5.9) is in turn a quadratic function of the Fresnel reflection coefficient. Hence
the phase sensitivity varies very rapidly with refractive index mismatch.

<table>
<thead>
<tr>
<th>Material</th>
<th>( n )</th>
<th>( \frac{dI}{d\phi} (I=1\text{mW}) )</th>
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<tr>
<td></td>
<td></td>
<td>( \text{\mu W/rad} )</td>
</tr>
<tr>
<td>(^1\text{PMMA} )</td>
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<td>6.4</td>
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</tr>
<tr>
<td>Glass (crown)</td>
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<td>7.2</td>
</tr>
</tbody>
</table>

Table 5.1 Phase sensitivities of different sensing film materials in contact with water on both sides. \( I=1\text{mW} \). \(^1\text{Polymethylmethacrylate, }\) \(^2\text{polyethylene terephthalate, }\) \(^3\text{low density polyethylene, }\) \(^4\text{polyvinylidene fluoride, }\) \(^5\text{polyether sulphone.} \)

Since the data in table 5.1 is calculated on the basis of a fringe visibility of 1, it would ultimately be possible to get the same phase sensitivity from each material by increasing the incident intensity until the dc level had reached its maximum sub-photodiode saturation threshold. If however, the maximum power emitted by the laser source is limited, it is advantageous to choose a polymer film that gives the highest phase sensitivity per unit incident intensity - i.e a material with a large refractive index mismatch with water such as PET. Assuming that a typical silicon photodiode can detect down to 0.1μW (over a 25 MHz measurement bandwidth), a PET sensing film surrounded by water would allow phase shifts of the order of 5mrad to be detected using a modest laser power of 1mW. \( I_{dc} \) (equation 5.10) would be 20.4μW which is also well below the saturation threshold of such a photodiode.
5.2.4 Optical fibre illumination of the sensing film

The analysis of the interferometer output has considered the case in which the light is normally incident on the sensing film. We now consider the effect on the phase sensitivity and the dc level of illuminating the interferometer with the divergent output of an optical fibre.

Consider the case in which a collimated beam of light is normally incident upon the sensing film. All of the light that is transmitted into and reflected back from the film travels the same optical path. Each point across the reflected optical field that is due to the interference between the reflections from the two sides of the sensing film is therefore associated with the same phase bias. Thus all of the reflected signal intensity-modulated light $dI$, arriving at the photodiode is in phase. This is in contrast to the situation in which the sensor is illuminated with the divergent light emerging from the end of a large diameter multimode optical fibre. The optical path length and hence the phase bias associated with the reflected light is dependent upon the angle of incidence and this can lead to conditions in which partial or complete cancellation of the signal occurs. Consider the case in which the light striking the film at normal incidence travels an optical path length $2l$ and assume that this path length corresponds to the first quadrature point. A ray at an angle of incidence $\theta$ takes an extra path length $\Delta l$

$$\Delta l = 2l \left( \frac{1}{\cos \theta} - 1 \right) \quad (5.14)$$

The corresponding difference in the phase bias $\phi_d$ as a result of this extra path length is

$$\phi_d = \frac{4\pi nl}{\lambda} \left( \frac{1}{\cos \theta} - 1 \right) \quad (5.15)$$

where $\lambda$ is the wavelength of the laser source. As the angle of incidence is increased the phase bias difference $\phi_d$ also increases until $\phi_d = \pi$. At this point the total optical path length corresponds to the next quadrature point and so the intensity modulated signal is now in antiphase with the signal arising from the normally incident light. When imaged on to the same detector the two signals cancel. This signal cancelling effect not only reduces the phase sensitivity but also adds to the dc level. The solution is to ensure that
the thickness of the interferometer and the maximum angle of divergence are such that the phase bias difference \( \phi_d \) due to the extra path length \( \Delta l \) is less (ideally significantly less) than \( \pi \) radians. To meet this condition using a PET film \((n=1.64)\), for example, and 850nm light emerging from an optical fibre with a maximum half-angle of divergence of \(5^\circ\), the thickness of the film from equation 5.15 should be less than 35\(\mu m\).

5.2.5 Section summary

To get maximum phase sensitivity, it is necessary to increase the intensity of the light incident on the sensing film as much possible without saturating the photodiode. To do this optimally, the reflection coefficients should be such that the fringe visibility is 1. If the interferometer is illuminated with the divergent output of an optical fibre, rather than with a collimated beam, the phase sensitivity is degraded and the dc level increased. To reduce this effect, the angle of divergence and the thickness of the interferometer should be kept to a minimum.
5.3 SENSOR-ACOUSTIC FIELD INTERACTION - ACOUSTIC SENSITIVITY AND FREQUENCY RESPONSE

This section is concerned with the interaction of the sensing film with an acoustic field and considers the phase modulation produced by an incident acoustic wave. The acoustic sensitivity, as a function of frequency, is considered by examining the thickness resonance modes arising from reflections of a normally incident acoustic wave within the sensing film.

The phase shift between the two interfering Fresnel reflections shown in figure 5.3 is

\[ \phi = \frac{4\pi nl}{\lambda} \]  \hspace{1cm} (5.16)

where \( n \) is the refractive index of the sensing film, \( l \) the thickness of the film and \( \lambda \) the optical wavelength. The strain due to an acoustic signal alters the optical path difference between the two reflections by producing a change in the thickness \( dl \) and a change in the refractive index \( dn \) each of which contribute to the net phase modulation \( d\phi \). By differentiating equation 5.16

\[ d\phi = \frac{4\pi}{\lambda} (ndl + ldn) \]  \hspace{1cm} (5.17)

The individual contributions to the net phase modulation due to \( dl \) and \( dn \) are considered individually in the next two sub-sections

5.3.1 Phase modulation due to the change in thickness

The strain due to a normally incident acoustic wave produces a change in the thickness of the polymer film \( dl \) giving rise to a phase shift \( d\phi_l \), which from equation 5.17 is
\[ d\phi_i = \frac{4\pi n dl}{\lambda} \]  

(5.18)

where \( dl \) is given by

\[ dl = - \int_0^l \frac{P_r(x,t)}{E} dx \]  

(5.19)

where \( E \) is the Young’s modulus of the polymer film. \( P_r(x,t) \) represents the spatial distribution of pressure across the thickness of the sensing film and is the sum of the

\[ P_1, P_2, P_3, \ldots P_n \]

component of the incident acoustic wave that is transmitted into the sensor film \( P_1 \) and subsequent acoustic reflections \( P_2, P_3, \ldots P_n \) at the boundaries of the film. These reflections, shown schematically in figure 5.6, arise as a result of the differences in acoustic impedance between the polymer film and the surrounding media.

In general, for a sinusoidally varying incident acoustic wave of amplitude \( P_0 \) and angular acoustic frequency \( \omega \) travelling in the negative \( x \) direction, the stress distribution across the film due to the superposition of \( P_1 \) and an odd number of subsequent reflections \( N \),

---

**Figure 5.6** Acoustic reflections within the sensing film.
neglecting acoustic attenuation, is

\[ P_T = P_0 T \sum_{i=0}^{(N-1)2} \left( R_1^i R_2^i \sin(\omega t - k (2li - x)) + R_1^{i+1} R_2^{i+1} \sin(\omega t - k (2li + x)) \right) \] (5.20)

T is the pressure amplitude transmission coefficient due to the acoustic impedance mismatch between the sensing film and surrounding media and \( R_i \) and \( R_2 \) are the pressure amplitude reflection coefficients at the two surfaces of the film where

\[ T = \frac{2Z}{Z_1 + Z_2}, \quad R_i = \frac{Z_1 - Z}{Z_1 + Z_1}, \quad R_2 = \frac{Z_2 - Z}{Z_2 + Z_2} \] (5.21)

where \( Z \) is the acoustic impedance of the polymer sensing film and \( Z_1 \) and \( Z_2 \) are the acoustic impedances of the media on either side of the film. Substituting \( P_T \) into equation 5.19 and evaluating the integral gives an expression of the form

\[ dl = -\frac{P_0 T}{E_k} \sum_{i=1}^{N+2} \psi_i \cos(\omega t + \xi_i) \] (5.22)

This is the sum of \( N+2 \) sinusoids of the same frequency but different amplitudes and phases and can be written

\[ dl = -\frac{P_0 T}{E_k} \psi \cos(\omega t + \xi) \] (5.23)

where the amplitude \( \psi \) is

\[ \psi = \sqrt{\left( \sum_{i=1}^{N+2} \psi_i \cos \xi_i \right)^2 + \left( \sum_{i=1}^{N+2} \psi_i \sin \xi_i \right)^2} \] (5.24)

The thickness of the sensor film is therefore modulated at a frequency \( \omega \) and is of amplitude

\[ dl_0 = \frac{P_0 T}{E_k} \psi \] (5.25)

When detecting ultrasound, the side of the sensing film that the acoustic wave is incident upon will always be in contact with water. The pressure amplitude reflection coefficient due to the acoustic impedance mismatch between water (acoustic
impedance=$1.5 \times 10^6$ Kg/m$^2$s) and a polymer film such as PET (acoustic impedance=$3.1 \times 10^6$ Kg/m$^2$s) is small at -0.35. Thus, for the purpose of modelling the frequency response of the sensor film it is sufficient to consider 5 reflections. After this number of reflections, the amplitude is reduced by more than two orders of magnitude if water is in contact with both sides of the sensing film (water backed) ie. $R_i=R_r=-0.35$. Even if one side of the film is in contact with a medium of very high acoustic impedance (rigid backed) so that $R_i=1$, the amplitude would still be down by almost an order of magnitude after 5 reflections. Evaluating equation 5.20 for $N=5$ and the integral in equation 5.19, equation 5.24 gives

$$\Psi_5 \approx [(1-R_1 + (R_1-R_2R_1-1) \cos kl + (R_2R_1-R_2R_1^2) \cos 2kl + (R_2^2R_1^2-R_2^2R_1^2) \cos 3kl + (R_2^2R_1^2-R_2^2R_1^2) \cos 4kl + (R_2^2R_1^2-R_2^2R_1^2) \cos 5kl)^2 + [(R_2R_1-R_1-1) \sin kl - (R_2R_1-R_2R_1^2) \sin 2kl - (R_2R_1-R_2R_1^2) \sin 3kl - (R_2^2R_1^2-R_2^2R_1^2) \sin 4kl - (R_2^2R_1^2-R_2^2R_1^2) \sin 5kl)^2]$$

(5.26)

The corresponding phase modulation amplitude $d\phi_{oi}$ from equations 5.25 and 5.18 due to the change in thickness for $N=5$ is given by

$$d\phi_{oi} = \frac{4\pi n}{\lambda} \frac{P_o T}{Ek} \Psi_5$$

(5.27)

5.3.2 Phase modulation due to the change in refractive index

The phase modulation due to the change in refractive index alone from equation 5.17 is

$$d\phi_n = \frac{4\pi ldn}{\lambda}$$

(5.28)

If a stress applied to a normally isotropic material induces birefringence, the material takes on the properties of a uniaxial crystal with the optic axis in the direction of stress. For light that is polarised along the direction of stress the change in refractive index is given by$^{72}$

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\[ dn = -\frac{n^3 p}{2} \left( \frac{dl}{l} \right) \]  

(5.29)

where \( p \) is the photoelastic constant of the material and \( dl/l \) is the strain due to the applied stress. Section 5.3.1 derived the acoustically-induced strain parallel to the direction of light propagating to and from the sensor. Clearly there is no light polarised in the direction of the longitudinal strain and so there is no change in refractive index due to this component of strain. The only change in the refractive index arises from the strain perpendicular to the direction of optical propagation and this is obtained by multiplying \( dl/l \) by Poisson’s ratio, \( \sigma \). Hence the change in refractive index due to the lateral strain is given by

\[ dn = \frac{n^3 p}{2} \sigma \left( \frac{dl}{l} \right) \]  

(5.30)

Note that the sign has changed - a longitudinal compression produces a lateral extension. Substituting equation 5.30 into equation 5.28 gives the phase modulation due to \( dn \)

\[ d\phi_n = \frac{4\pi n dl}{\lambda} \frac{n^3 p \sigma}{2} \]  

(5.31)

Equation 5.31 shows that \( d\phi_n \) is simply \( d\phi_t \) (equation 5.18) multiplied by a scale factor equal to \( n^2 p \sigma/2 \). Furthermore, it is of the same sign as \( d\phi_t \) and therefore adds to the phase modulation due to the change in thickness. The significance of \( d\phi_n \) when using a polymer sensing film can be difficult to determine as the photoelastic constants for many polymers do not appear in the literature. Photoelastic constants for polymethylmethacrylate \( (p=0.32) \) and polystyrene \( (p=0.31) \) are among the few that are quoted\(^1\)\(^2\). Using these values of \( p \) and the data in Appendix B (table of material properties) for \( n \) and \( \sigma \), gives approximately 0.13 for the \( (n^2 p \sigma/2) \) factor for both materials. \( d\phi_n \) can therefore be neglected. This is not the case, however, for certain materials such as photoelastic PSM-1 which have a photoelastic constant\(^1\)\(^3\) an order of magnitude higher than materials such as polystyrene, polymethylmethacrylate and glass.
5.3.3 Acoustic sensitivity

The acoustic sensitivity of the sensing film is defined as the magnitude of the net phase modulation per unit acoustic pressure $d\phi_d/P_o$. Since photoelastic data for most polymers is not readily available, and in the two examples given above (polystyrene and polymethylmethacrylate) $d\phi_d$ can be neglected, only the phase modulation due to the acoustically induced change in thickness given by equation 5.27 is considered. Thus the acoustic sensitivity as a function of frequency is

$$\frac{d\phi_d}{P_o} = \frac{4\pi n}{\lambda} \cdot \frac{T}{Ek} \psi_{[5]}$$

(5.32)

Equation 5.32 can be used to predict the sensitivity and frequency response for different media in contact with the two sides of the film and various sensor-film materials.

5.3.4 Effect of different backing configurations

A key factor that affects the bandwidth and uniformity of frequency response is the backing medium in contact with the sensing film. The acoustic sensitivity as a function of frequency response is considered below for five different acoustic loading configurations of practical importance using equation 5.32. The acoustic sensitivity as a function of acoustic frequency for each case is shown in figure 5.7 for a 50μm thick PET sensing film using the values $E=4.4$GPa, $c=2200$m/s and $n=1.6$, $Z=3.1 \times 10^6$Kg/m²s, $\lambda=850$nm.

1. Matched load

This is the ideal case in which the polymer sensing film is surrounded by media of the same acoustic impedance ($Z=Z_i=Z_o$) so there are no acoustic reflections at the boundaries. Under these conditions $R_i=R_o=0$ and $T=1$ and equation 5.26 reduces to

$$\psi_{[5]} = 2\sin\left(\frac{kl}{2}\right)$$

(5.33)

inserting this into equation 5.32 and using the relation $k=2\pi/\lambda_a$ where $\lambda_a$ is the acoustic
wavelength, the acoustic sensitivity is given by

\[ \frac{d \Phi_o}{P_o} = \frac{4n}{\lambda} \frac{\lambda_a}{E} \sin\left(\frac{\pi l}{\lambda_a}\right) \]  

(5.34)

There are two frequency regimes of interest. Firstly, the case in which the acoustic wavelength \( \lambda_a \) is much greater than the thickness \( l \) of the film and secondly when \( \lambda_a \) approaches \( l \) and the sensor can no longer respond dynamically to the acoustic wave. If \( \lambda_a \) is large compared to \( l \), the pressure gradient across the sensor is effectively zero and equation 5.34 reduces to

\[ \text{for } \lambda_a >> l \quad \frac{d \Phi_o}{P_o} = \frac{4\pi n}{\lambda} \frac{l}{E} \]  

(5.35)

This shows that at low frequencies, for a given laser wavelength and refractive index \( n \), \( d \Phi_o/P_o \) is dependent upon only the thickness and the Young’s modulus of the sensor material. As the acoustic wavelength is reduced and approaches \( l \) the spatial variation of the pressure field across the sensor becomes significant and \( d \Phi_o/P_o \) decreases becoming zero when \( \lambda_a = l \). The first frequency at which \( d \Phi_o/P_o \) is zero, the cut-off frequency, is given by

\[ f_c = \frac{c}{l} \]  

(5.36)

where \( c \) is the speed of sound in the sensor material. Beyond \( f_c \) the value of \( d \Phi_o/P_o \) varies rapidly with increasing frequency becoming zero for wavelengths equal to multiples of \( l \). In the example shown in figure 5.7 this behaviour is shown with \( f_c \) occurring at 44MHz. This simple analysis gives an insight into the required acoustic properties of the sensing film. For maximum acoustic sensitivity and frequency response a material with both a low Young’s modulus and a high speed of sound are required. A material with a very low Young’s modulus will exhibit high acoustic sensitivity for \( \lambda_a >> l \). If it also has a very low speed of sound then equation 5.36 shows that for high frequency response \( l \) will have to be small. This in turn reduces acoustic sensitivity.

2. Rigid backed

The sensing film is backed with a medium that has a very high acoustic impedance
and \( R_j = 1 \). For algebraic simplicity the front face reflection is neglected by setting \( R_2 = 0 \) and \( T = 1 \) and from equation 5.26, \( \psi_{[5]} \) becomes

\[
\psi_{[5]} = 2 \sin kl
\]

(5.37)

and the acoustic sensitivity

\[
\frac{d\phi_o}{P_o} = \frac{4n}{\lambda} \frac{\lambda_a}{E} \sin \left( \frac{2\pi l}{\lambda_a} \right)
\]

(5.38)

In the low frequency limit this reduces to

\[
\text{for } \lambda_a \gg l \quad \frac{d\phi_o}{P_o} = \frac{8\pi n}{\lambda} \frac{l}{E}
\]

(5.39)

This is a factor of two higher than the acoustic sensitivity of the matched load configuration discussed in (11) although the cut off frequency \( f_c \) has decreased by a factor of two now occurring at

\[
f_c = \frac{c}{2l}
\]

(5.40)

This is shown in figure 5.7 occurring at 22MHz

3. Air backed

In this case the film is backed by air \((Z_i \ll Z)\) so that \( R_i = -1 \). To simplify equation 5.26 the reflection from the front face of the sensing film is again neglected by setting \( R_2 = 0 \) and \( T = 1 \). Equation 5.26 becomes

\[
\psi_{[5]} = 4 \sin^2 \left( \frac{kl}{2} \right)
\]

(5.41)

and from equation 5.32 the acoustic sensitivity is

\[
\frac{d\phi_o}{P_o} = \frac{8n}{\lambda} \frac{\lambda_a}{E} \sin^2 \left( \frac{\pi l}{\lambda_a} \right)
\]

(5.42)

The non uniform frequency response characteristic of this configuration is shown clearly in figure 5.7 with the acoustic sensitivity rising from zero to a maximum. Such a configuration would give a poor representation of an acoustic transient with a broadband
frequency content.

4. Water/PET/water

The closest practical configuration to the matched load ideal of (1) is to arrange for water to be in contact with both sides of the PET sensing film. In this case $Z_1 = Z_2 = 1.5 \times 10^6$ Kg/m$^2$s so $R_1 = R_2 = -0.35$. The acoustic sensitivity is modelled using the full form of equation 5.26. The prominent features of the frequency response curve shown in figure 5.7 are the $\lambda/2$ resonance at approximately 20MHz and the cut off frequency $f_c$ at 44MHz. In practice it can be expected that the effects of attenuation within the film (not considered in this model) would act to reduce the magnitude of the $\lambda/2$ resonance. In the low frequency regime where $\lambda >> l$ and when $R_1 = R_2$, equation 5.26 reduces to

$$\psi_{[5]} = k l \sum_{i=0}^{5} R_i^i = \frac{kl(1-R_1^6)}{1-R_1}$$

(5.43)

which for small $R_i$ further reduces to

$$\psi_{[5]} = \frac{kl}{1-R_1} = \frac{kl}{T}$$

(5.44)

Substituting equation 5.44 into equation 5.32 gives

$$\frac{d\phi_0}{P_o} = \frac{4\pi n l}{\lambda E}$$

(5.45)

which is the same expression for the low frequency acoustic sensitivity for the matched load case given in equation 5.35. This behaviour is shown clearly in figure 5.7 with the departure from the matched load case beginning at approximately 2MHz. The low frequency acoustic sensitivity of the water/PET/water configuration is 0.28 rad/MPa.

5. Glass/PET/water

A useful configuration, in practical terms, is that of the sensing film attached directly to the tip of an all-silica optical fibre. The acoustic sensitivity of this configuration is modelled using the full form of equation 5.26 assuming that the acoustic impedance of
fused silica is close to that of glass \((Z=12.7 \times 10^6 \text{ Kg/m}^2\text{s})\). At low frequencies, this configuration approximates to the rigid backed case of (2) showing the increased acoustic sensitivity compared to the water backed sensing case. Characteristic of a

![Graph](image)

**Figure 5.7** Theoretical acoustic sensitivity as a function of frequency for different acoustic loading configurations using a 50\(\mu\)m thick PET sensing film.

practical rigid backed configuration (i.e. \(R_z\neq 0\)) is the \(\lambda_c/4\) resonance occurring at 8.5MHz.

### 5.3.5 Acoustic sensitivity of different polymer sensing films

Table 5.2 shows the acoustic sensitivities and cut-off frequencies of different sensing film materials. The acoustic sensitivities are calculated using the material properties listed in Appendix B and the low frequency matched load configuration described by equation (5.35). This configuration approximates to the water-backed case at low frequencies described in (4) and therefore represents a physically useful case. An indication of frequency response is given by the cut-off frequency \(f_c\) (equation 5.36)
which corresponds to the full wavelength resonance condition. From $f_c$, the frequencies corresponding to the half and quarter wavelength resonances of the water and fused silica backed configurations can be easily estimated to give an indication of the frequency response under different backing configurations. The parameters in table 5.2 are for a sensing film thickness of 50μm. Strictly speaking, to make a true comparison of acoustic sensitivity of different materials, the thickness of each sensing film should be such that they all have the same value of $f_c$. In practice however, most polymers tend to have approximately the same values for the speed of sound and so $f_c$ will be similar.

<table>
<thead>
<tr>
<th>Material</th>
<th>$d\phi/P_o$</th>
<th>$f_c$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>rad/MPa</td>
<td>MHz</td>
</tr>
<tr>
<td>$^1$PMMA</td>
<td>0.44</td>
<td>54</td>
</tr>
<tr>
<td>Polystyrene</td>
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<td>47</td>
</tr>
<tr>
<td>Polycarbonate</td>
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<tr>
<td>$^2$PET</td>
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<td>44</td>
</tr>
<tr>
<td>$^3$LDPE</td>
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</tr>
<tr>
<td>$^4$PVDF</td>
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<td>40</td>
</tr>
<tr>
<td>$^5$PES</td>
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<td>-</td>
</tr>
<tr>
<td>Glass (crown)</td>
<td>0.016</td>
<td>113</td>
</tr>
</tbody>
</table>

Table 5.2 Low frequency ($\lambda_o>>l$) matched-load acoustic sensitivities $d\phi/P_o$ and cut-off frequencies $f_c$ of different sensing film materials. $\lambda_o=850\text{nm}$, $l=50\mu\text{m}$.

$^1$polymethylmethacrylate $^2$polyethylene terephthalate, $^3$low density polyethylene, $^4$polyvinylidene fluoride, $^5$polyether sulphone.

Table 5.2 shows that most of the polymers offer similar acoustic sensitivity and frequency response. An exception is LDPE which, due to its low Young’s modulus, has an acoustic sensitivity more than an order of magnitude higher than the other polymers. The speed of sound of LDPE is similar to the other polymers so it has a similar cut-off frequency. In addition, it has an acoustic impedance closer to that of water than any other polymer (Appendix B). Hence when water-backed it should
offer a very uniform response. Glass has a high Young’s modulus resulting in very low acoustic sensitivity although its high speed of sound gives a very high cut off frequency. To increase sensitivity without the cut-off frequency falling below the 45MHz of the polymer sensing films, the thickness could be increased by a factor of 2.5. Even with the increased thickness, the acoustic sensitivity would still be an order of magnitude lower than the polymers. In addition, when water backed it is likely to display highly resonant behaviour due to its high acoustic impedance mismatch with water.

5.4 **OVERALL SENSITIVITY**

For high phase sensitivity it is desirable that the Fresnel reflection coefficients at the faces of the sensing film are the same in order to obtain a fringe visibility of one. For a flat frequency response it is desirable that the sensing film is backed by a medium of acoustic impedance close to that of the sensing film so that acoustic reflections are minimised. Both requirements can be fulfilled in practice by backing the sensing film with water. To obtain the highest acoustic sensitivity and optimum uniformity of frequency response, the material properties of the polymer should be such that it is of low Young’s modulus and possesses an acoustic impedance close to water. Although most of the requirements for optimum phase and acoustic sensitivity can be considered individually, there is the potential for conflict when considering the thickness of the sensing film. Section 5.2.4 described a signal cancelling effect whereby the phase sensitivity is degraded (and the dc level increased) when the sensing film is illuminated by the divergent output of an optical fibre. This effect increases with the angle of divergence of the illuminating light and the thickness of the interferometer. Thus, the thickness of the sensing element is not determined solely by considerations of acoustic sensitivity and frequency response as might be expected. For example, if a low frequency response is required, the film could be made quite thick for high acoustic sensitivity. This however would increase the signal cancelling problems arising from optical fibre illumination of the sensing film and degrade the overall sensitivity. The trade-off is therefore between the requirements of phase sensitivity, acoustic sensitivity
and frequency response. Where the very highest sensitivity is required for detecting small amplitude signals, it is necessary to strive for both the highest possible acoustic sensitivity and phase sensitivity. If, however, high amplitude signals are to be detected it is important to ensure that the acoustic sensitivity is not too high. This is because if the signal-induced phase modulation becomes too large, the small phase angle approximation cannot be invoked resulting in a non linear response. In such circumstances it is the phase sensitivity that should be optimised for maximum dynamic range. For this reason alone, if a conflict arises between the requirements to optimise phase and acoustic sensitivity the priority, in general, should be to maximise the phase sensitivity.

Multiplying the phase sensitivity (equation 5.9) and acoustic sensitivity (equation 5.32) brings together the interferometric and acoustic aspects of the sensor operation to give an expression in terms of the intensity per unit acoustic pressure. This represents the overall system sensitivity

\[
\frac{dI}{P_o} = \frac{8 \pi n I (1 - r_1) \sqrt{r_1 r_2} T \cdot \Psi_{[5]}}{Ek \lambda}
\]  

(5.46)

Note that this expression does not take into account the phase sensitivity degradation occurring when the sensing film is illuminated by the output of an optical fibre. It also neglects the contribution to the acoustic sensitivity arising from acoustically-induced changes in the refractive index.

In terms of acoustic sensitivity and acoustic impedance considerations, LDPE would make an excellent sensing film. Unfortunately, the phase sensitivity for a given incident laser power is low due to its low refractive index and the surface quality of LDPE films is generally poor. For these reasons we consider the example of a 50μm PET sensing film surrounded by water. This polymer, due to its high refractive index mismatch with water, gives high phase sensitivity and its acoustic impedance is reasonably well matched to that of water. The 50μm thickness provides an acceptable compromise between obtaining an acoustic frequency response of the order of tens of MHz and limiting the phase sensitivity degradation discussed in section 5.2.4. Using the values 21.6μW/rad for the phase sensitivity from table 5.1 and 0.28 rad/MPa for the
acoustic sensitivity from table 5.2, equation 5.46 gives the overall sensitivity as 6μW/MPa. Taking the minimum detectable intensity of a typical silicon photodiode as 0.1μW (over a 25 MHz measurement bandwidth) this would give a minimum detectable acoustic signal of approximately 17kPA.

5.5 NOISE

There are two types of noise to be considered. Noise due to environmental effects and noise associated with the laser and the photodiode detection system. Environmentally based effects such as temperature and pressure fluctuations are low frequency disturbances and unlikely to be of significance in the frequency band of interest which is of the order of tens of MHz. The limiting noise performance of the system will therefore be dominated by the noise characteristics of the photodiode, the photodiode amplifier and the laser source. The aim of this section is to identify the dominant sources of noise and obtain noise floor values that can be expected when using typical laser sources and photodiode-amplifier combinations.

5.5.1 Photodiode noise

Three sources of photodiode noise are considered: the shot noise due to the photocurrent, the shot noise due to the dark current and the thermal noise due to the shunt resistance. 1/f noise predominates at frequencies below 100 Hz and can be neglected. The photodiode is assumed to be operating in a reverse bias configuration.

(i) The rms shot noise current per unit bandwidth due to the photocurrent $I_{ph}$ is given by

$$i_{sp} = \sqrt{2eI_{ph}} \quad (5.47)$$

where $e$ is the electronic charge.
(ii) The shot noise associated with the dark current $I_d$ is

$$i_{sd} = \sqrt{2eI_d}$$  \hspace{1cm} (5.48)

(iii) The rms thermal noise current per unit bandwidth due to the shunt resistance $R_{sh}$ is

$$i_{R_{sh}} = \sqrt{\frac{4KT}{R_{sh}}}$$  \hspace{1cm} (5.49)

where $K$ is Boltzmann’s constant and $T$ is the absolute temperature.

The total rms noise current is the square root of the sum of the individual contributions

$$i_{np} = \sqrt{i_{sp}^2 + i_{sd}^2 + i_{R_{sh}}^2} \quad A/\sqrt{Hz}$$  \hspace{1cm} (5.50)

For a typical high speed silicon PIN photodiode such as the Hamamatsu S-3072: $R_{sh} = 1 \Omega$, $I_d = 10 \text{nA}$ (for a reverse bias voltage of 24V) and the responsivity $S_R$ is 0.5 A/W at 700 nm. The photocurrent consists of contributions arising from both the signal intensity modulation $dI_o$ and the dc level $I_{dc}$. In practice $I_{dc} \gg dI_o$ so the photocurrent due to $dI_o$ can be neglected. For $I_{dc} = 100 \mu\text{W}$ incident on the photodiode, a photocurrent $I_{ph} = 50 \mu\text{A}$ is generated. Using the equations above and the values $K = 1.33 \times 10^{-23} \text{ J/K}$, $T = 293 \text{K}$, $e = 1.6 \times 10^{-19} \text{C}$, the individual contributions to the noise are:

$$i_{sp} = 4 \times 10^{-12} \quad A/\sqrt{Hz}$$
$$i_{sd} = 5.64 \times 10^{-14} \quad A/\sqrt{Hz}$$
$$i_{R_{sh}} = 3.94 \times 10^{-15} \quad A/\sqrt{Hz}$$

Under these conditions, the shot noise $i_{sp}$ due to the photocurrent clearly dominates. As the incident intensity is decreased, the noise falls as the square root of the intensity until the shot noise due to the dark current dominates. Over a measurement bandwidth of 25 MHz, the shot noise due to the dark current will dominate when the incident light level falls below approximately 0.5 nW. This is well below typical values of $I_{dc}$ which will be of the order of several tens of microwatts. Hence for this application the photodiode noise will be dominated by the photocurrent shot noise.
5.5.2 Amplifier noise

A practical photodiode detection system requires some form of amplifier. To determine the overall system noise, the noise characteristics of a transimpedance amplifier are modelled using an equivalent noise circuit.\(^\text{(177)}\)

The four sources of noise shown in the equivalent circuit of figure 5.8 each contribute to the total output voltage noise. Using the assumptions of an ideal op-amp, each noise source is considered separately. The noise currents are expressed as the r.m.s values per unit bandwidth.

\[ Z_f = \frac{R_f}{1 + j \omega C_j R_f} \quad (5.52) \]

\[ v_{op} = i_n Z_f \quad (5.51) \]

---

**Figure 5.8 Equivalent noise model of a transimpedance amplifier**

(i) The photodiode noise current \( i_n \) is multiplied by the impedance of the feedback network \( Z_f \) comprising the feedback resistor \( R_f \) and capacitor \( C_j \) in parallel to give a noise voltage \( v_{op} \) at the output.
(ii) The amplifier input noise current is also multiplied by $Z_f$ to produce a noise voltage $v_{oi}$ at the output

$$v_{oi} = i_n Z_f \tag{5.53}$$

(iii) The thermal noise current $i_{bf}$ due to the feedback resistor produces a voltage $v_{of}$ across $Z_f$ appearing at the output as

$$v_{of} = i_{bf} Z_f \tag{5.54}$$

$$v_{of} = \sqrt{\frac{4kT}{R_f} Z_f} \tag{5.55}$$

(iv) The amplifier input noise voltage $v_n$ appears at the output as a voltage $v_{ov}$. In an ideal op-amp no current flows into the input terminal so

$$\frac{v_{ov} - v_n}{Z_f} = \frac{v_n}{Z_p} \tag{5.56}$$

$$v_{ov} = v_n \left(1 + \frac{Z_f}{Z_p}\right) \tag{5.57}$$

where $Z_p$ is the impedance of the photodiode - the shunt resistor $R_{sh}$ and the junction capacitance $C_j$ in parallel.

$$Z_p = \frac{R_{sh}}{1 + j\omega C_j R_{sh}} \tag{5.58}$$

The total rms noise voltage $v_{on}$ at the output is the square root of the sum of the squares of the individual contributions

$$v_{on} = \sqrt{v_{op}^2 + v_{oi}^2 + v_{of}^2 + v_{ov}^2} \tag{5.59}$$


\[ v_{on} = \sqrt{\frac{R_i^2}{1 + (\omega C_i R_i)^2} \left( i_{np}^2 + i_n^2 + \frac{4KT}{R_i} \right) + v_n^2 \left( 1 + \frac{R_f \sqrt{1 + (\omega C_f R_f)^2}}{R_{sh} \sqrt{1 + (\omega C_f R_f)^2}} \right)^2} \quad V/\sqrt{Hz} \]

(5.60)

Figure 5.9 shows the contribution of each source of noise to the total noise using the following values:

- \( R_{sh} = 1 \ \Omega \)
- \( C_f = 7 \ \text{pF} \)
- \( i_{np} = 4 \times 10^{12} \ \text{A/Hz} \)
- \( S_n = 0.5 \ \text{A/W} \)
- \( i_n = 10 \times 10^{-15} \ \text{A/Hz} \)
- \( v_n = 20 \times 10^{-9} \ \text{V/Hz} \)
- \( R_f = 30 \ \text{K}\Omega \)
- \( C_f = 0.3 \ \text{pF} \)

\[ \begin{array}{l}
\end{array} \]

**Figure 5.9** Contribution of noise voltages to total output noise voltage \( v_{on} \) appearing on the output. \( v_{op} \) - photodiode noise, \( v_{ov} \) - amplifier input noise voltage, \( v_{of} \) - thermal noise current, \( v_{oi} \) - amplifier input noise current.
Figure 5.9 shows that, up to a few MHz, the dominant noise contribution is due to the photodiode noise $v_{op}$ which, as section 5.5.1 showed, is due to the shot noise associated with the photocurrent. Above about 5MHz the amplifier input voltage noise $v_{ov}$ begins to dominate which, at these frequencies is largely dependent upon the photodiode junction capacitance $C_j$. This factor should be borne in mind if a larger area photodiode (consistent with the frequency response requirements) is being considered as a means of obtaining a higher saturation threshold in order to increase the phase sensitivity of the interferometer. Since $C_j$ is proportional to the active area of the photodiode, $v_{ov}$, which dominates the noise characteristics above 5MHz, will increase. One solution would be to bandlimit the amplifier above 5MHz. Below this frequency the noise characteristics are mainly dominated by $v_{op}$. So, for applications of the sensor that require a frequency response below 5MHz, a larger area photodiode could be used without increasing the overall noise. The conditions described for the proposed application are somewhat unusual because the light incident on the photodiode consists of a small signal intensity modulation imposed upon a relatively large dc level. Hence there is always a large photocurrent with accompanying shot noise that provides a significant contribution to the noise characteristics. By contrast, in many applications, the aim is to detect a small time varying optical signal alone. The noise associated with the photocurrent, dark current and shunt resistance are so small that the overall noise at all frequencies is usually dominated by the amplifier noise. Under these conditions, it would then be advantageous to use an avalanche photodiode because the internal gain mechanism of the device increases the amplifier input signal relative to the amplifier noise floor. This would not be the case with the proposed application because the significant source of noise is the shot noise due to the large photocurrent generated by the dc level intensity. There would be no improvement in SNR using the avalanche photodiode, because the shot noise due to the large photocurrent will simply be amplified by the avalanche process along with the signal.

Integrating equation 5.60 over a bandwidth of 25 MHz gives the total noise voltage output

$$v_{ov} = 1.44 \text{mV}$$

The sensitivity $S_c$ of the photodiode-amplifier combination is the photodiode responsivity
multiplied by the impedance of the feedback network \( Z_f \)

\[ S_c = S_R Z_f \]  \hspace{1cm} (5.61)

For \((\omega C_f R_f) \ll 1\), \( Z_f = R_f \)

\[ S_c = S_R R_f \]  \hspace{0.5cm} \frac{\mathrm{V}}{\mathrm{W}} \hspace{1cm} (5.62)

Thus \( S_c = 0.5 \times 3 \times 10^3 = 15 \) mV/\(\mu\)W and converting \( \nu_{m} \) to an equivalent rms intensity noise \( i_{on} \) gives

\[ i_{on} = (1.44/15) \times 10^{-6} = 0.096 \mu\text{W}. \]

This represents the minimum detectable optical signal over a 25 MHz bandwidth using a typical silicon photodiode and transimpedance amplifier.

5.5.3 Laser noise

There are two sources of laser noise that contribute to the overall system noise: intensity noise and phase noise. Typical characteristics of a GaAlAs laser diode (SDL 5412) and the Helium-Neon laser are considered. In the absence of spectral noise density data the rms noise figures are given over a bandwidth of 25 MHz unless otherwise stated.

1. Intensity noise

In addition to the dc output of the sensor, there is a fluctuating intensity due to the laser intensity noise. For Helium Neon lasers this is typically rated at less than 1\% between 30 Hz and 10 MHz with low frequency 1/f noise providing the dominant contribution. For a dc level of 100\(\mu\)W, the associated intensity noise \( i_{nl} \) is therefore

\[ i_{nl} = 0.01 \times 50 = 1 \mu\text{W} \]

This is a very high figure and may be unduly weighted by low frequency fluctuations which, for this application, are outside the 1MHz - 25 MHz band of interest. For a laser diode such as the SDL 5412, \( i_{nl} \) is dominated by the characteristics of the current source driving the laser diode. For a typical laser diode driver, the current noise is less than 2.5
μA over the band 10 Hz to 25 MHz. The slope efficiency of the SDL 5412 is 0.92 μW/μA giving an absolute intensity noise of 0.92 x 2.5 = 2.3μW. This is constant and independent of the laser output power. If the laser output is 2mW then, expressed as a percentage of output power, the noise is 0.12% and the intensity noise, for a dc level of 100μW is

\[ i_n = 0.12 \text{ μW} \]

This could be reduced significantly by operating the laser at a higher power and attenuating the beam.

2. Phase noise

Phase noise due to frequency instabilities and the finite linewidth of the laser are converted to intensity noise by the interferometer. The effect is proportional to the optical path length 2nl and the phase sensitivity dl/dφ. The phase noise dφvn due to a frequency instability/linewidth dv̄n is

\[ d\phi_{vn} = \frac{4\pi nldv_n}{c_o} \]

where \( c_o \) is the velocity of light. The phase noise appears as an intensity noise \( i_{np} \) and is obtained by multiplying \( d\phi_{vn} \) by the phase sensitivity

\[ i_{np} = \frac{dl_o}{d\phi} d\phi_{vn} \]

For a He-Ne, \( dv_n \) is typically around 1MHz over a 1 minute interval with a very much lower "instantaneous" value. Taking the example of a 50μm thick PET sensing film and using the values \( n=1.64, l=50μm, c_o=3 \times 10^8 \text{m/s} \) equation 5.64 gives

\[ d\phi_{vn} = 3.5 \times 10^{-6} \text{ rad} \]

and using \( dl_o/d\phi=21.6\mu\text{W/ rad} \) gives a corresponding intensity noise

\[ i_{np} = 7.4 \times 10^{-5} \text{ μW} \]
For the laser diode the contribution of the current noise from the driver to \( d\nu_n \) is small at 2 MHz/\( \mu \)A and is neglected. The linewidth of the SDL 5412 is 15 MHz so

\[
d\phi_{\nu n} = 52.5 \times 10^6 \text{ rad}
\]

and

\[
i_{\phi} = 111 \times 10^5 \text{ } \mu \text{W}
\]

The phase noise is very low in both cases and, for short path length (tens of \( \mu \)m) interferometers, can be ignored.

The total laser noise \( i_n \) is the square root of the sum of the squares of the intensity noise and the phase noise

\[
i_n = \sqrt{i_{\phi}^2 + i_{\nu n}^2}
\]

The dominant source of laser noise for the examples shown above is therefore intensity noise. The value for the laser diode is considered more appropriate than that for the He-Ne and so

\[
i_n = 0.12 \text{ } \mu \text{W}
\]

**5.5.4 Overall system noise**

The overall system intensity noise \( i_n \) is made up of the photodiode-amplifier noise voltage \( \nu_{on} \) expressed as an equivalent intensity \( i_{on} \) (section 5.5.2) and the laser noise \( i_n \) (section 5.5.3) above

\[
i_{sn} = \sqrt{i_{on}^2 + i_{n}^2}
\]

Using the values in sections 5.5.2 and 5.5.3, \( i_{on} = 0.096 \mu \text{W} \) and \( i_n = 0.12 \mu \text{W} \)

\[
i_{sn} = 0.15 \text{ } \mu \text{W}
\]

This figure represents the minimum detectable signal intensity modulation in a 25MHz measurement bandwidth.
5.6 **ACOUSTIC DETECTION LIMIT**

The phase sensitivity, acoustic sensitivity and noise characteristics are combined to determine the acoustic detection limit using a 50μm PET sensing film illuminated by 1mW laser source at 850nm. The configuration is that used in the phase and acoustic sensitivity calculations in sections 5.2.3 and 5.3.5 whereby the sensing film was in contact with water on both sides. A phase sensitivity of 21.6μW/rad (section 5.2.3), acoustic sensitivity of 0.28rad/MPa (section 5.3.5) and dIo/Po=6μW/MPa (section 5.4) are assumed. Using a typical laser source and photodiode detection system, the minimum detectable intensity modulation is approximately 0.15μW (section 5.5.4). This corresponds to an acoustic noise floor of 25KPa and represents the minimum detectable acoustic signal.

5.7 **TEMPERATURE SENSITIVITY**

Thermal fluctuations do not represent a source of noise because they are low frequency (<100Hz) effects and outside the measurement bandwidth of interest. They can however cause signal fading by shifting the phase bias of the interferometer away from the quadrature point. If the phase bias shifts to the extent that it corresponds to a maxima or minima of the interferometer transfer function (figure 5.2), the signal can disappear completely. Whilst there are techniques for phase bias control (discussed in the following section), they can become difficult to implement if the sensor is highly sensitive to ambient thermal fluctuations. To limit the requirements of any phase bias control system and retain the inherent simplicity of the system, it is desirable that the temperature sensitivity of the sensor is low.

By analogy with the definition for acoustic sensitivity, the temperature sensitivity in rad/°C is the phase shift produced per unit temperature rise and is given by

126
\[
\frac{d\phi_i}{dT} = \frac{4\pi n\beta_i l}{\lambda}
\]  \hspace{1cm} (5.67)

where \(d\phi_i\) is the temperature-induced phase shift, \(dT\) is the temperature change and \(\beta_i\) is the coefficient of linear thermal expansivity. The table below gives the temperature sensitivities calculated for different materials for \(l=50\mu m\) and \(\lambda=850nm\).

<table>
<thead>
<tr>
<th>Material</th>
<th>(\beta_i \times 10^{-3} / ^\circ C)</th>
<th>(d\phi_i / dT \ (l=50\mu m))</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMMA</td>
<td>7.3</td>
<td>0.08</td>
</tr>
<tr>
<td>Polystyrene</td>
<td>3.2</td>
<td>0.038</td>
</tr>
<tr>
<td>Polycarbonate</td>
<td>6.6</td>
<td>0.077</td>
</tr>
<tr>
<td>PET</td>
<td>4.8</td>
<td>0.058</td>
</tr>
<tr>
<td>LDPE</td>
<td>15</td>
<td>0.167</td>
</tr>
<tr>
<td>PVDF</td>
<td>11</td>
<td>0.115</td>
</tr>
<tr>
<td>PES</td>
<td>5.5</td>
<td>0.067</td>
</tr>
<tr>
<td>Glass (crown)</td>
<td>0.9</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Table 5.3 Temperature sensitivities of different materials for \(\lambda=850nm\), \(l=50\mu m\).

\(^1\text{poly(methylmethacrylate) \ ^2\text{polyethylene terephthalate, \ ^3\text{low density polyethylene,}} \ ^4\text{polyvinylidene fluoride, \ ^5\text{polyether sulphone.}}\)

The data in table 5.3 shows that phase shifts produced by environmental thermal fluctuations of a few \(^\circ C\) are small even for a polymer such as LDPE that has a high coefficient of linear thermal expansion. If the interferometer is initially set at quadrature, thermally-induced phase shifts will have a negligible effect on the sensitivity and linearity of the sensor. For a 50\(\mu m\) thick PET sensing film, the phase shift produced by a 1\(^\circ C\) change in temperature is just 0.058\(rad\). To produce a phase shift of \(d\phi_i=\pi/2\), which would cause the signal to disappear completely if the interferometer were initially biased at quadrature, a temperature change of 27\(^\circ C\)
would be required. Thus for most polymers, the sensor should be relatively insensitive to ambient thermal fluctuations.

5.8 QUADRATURE CONTROL

In a practical system, some form of quadrature control is required to prevent signal fading arising from environmentally-induced perturbations (such as ambient temperature fluctuations) which can shift the phase bias of the interferometer from the quadrature point. Even if the sensor is inherently stable to environmental influences, some means of setting the phase bias on "start-up" is required. Conceptually, the simplest method of quadrature control is achieved by adjusting the wavelength of the laser source. A number of other methods have been suggested. Dual-wavelength schemes employing two laser sources of different wavelengths have been reported. If the signal fades at one wavelength, due to the phase bias shifting towards a maxima or minima, the other wavelength moves towards quadrature thus providing a degree of compensation for the faded signal. Such a scheme requires careful choice of laser diode wavelengths for the appropriate phase bias separation between the two wavelengths. Alternatively, some means of actively controlling the Fabry Perot interferometer thickness during fabrication can be used to obtain the required phase bias separation. In both cases very high stability between the two interrogating wavelengths is required. This has prompted an interesting solution whereby two modes of a multimode laser diode were used to provide the two wavelengths. This provides greater stability between two wavelengths than can be achieved using two independent laser sources. A variant on the two-wavelength scheme is the use of two cavities of different thicknesses to obtain the differential phase bias although this has the disadvantage that it requires two sensors and two fibre downleads. A dichroic ratio technique that uses an LED source to interrogate a very small path length Fabry Perot cavity has been described. This system is not strictly a quadrature control scheme but takes the ratio of the output of the long and short wavelength components selected using a dichroic mirror. This technique enables the linear operating range to be increased compared to the single wavelength transfer function. Unfortunately, using a short coherence length source such as an LED means
that the Fabry Perot interferometer thickness has to be of only a few μms.

Whilst a dual wavelength scheme could be implemented in some form for the proposed application, it is considered that, in the first instance, a wavelength tunable laser diode offers a simple and easily implemented solution. Providing sufficient continuously variable tuning range can be achieved, it will always be possible to set the interferometer at quadrature. The phase bias in the low finesse Fabry Perot interferometer is

\[ \phi = \frac{4\pi nl}{\lambda} \]  

(5.68)

Differentiating equation 5.68 and rearranging provides the wavelength shift as a function of phase shift

\[ \frac{d\phi}{d\lambda} = -\frac{4\pi nl}{\lambda^2} \]  

(5.69)

\[ d\lambda = -\frac{\lambda^2}{4\pi nl} d\phi \]  

(5.70)

However much the phase bias drifts, the maximum corrective phase shift required to return to quadrature is \( d\phi = \pi/2 \) rad. The corresponding wavelength change \( d\lambda_{q2} \) required is

\[ d\lambda_{q2} = -\frac{\lambda^2}{8nl} \]  

(5.71)

From equation 5.71, using a 50μm thick PET interferometer (\( n=1.64 \), \( l=50μm \)) and \( \lambda=850nm \), \( d\lambda_{q2}=1.1nm \). This tuning range can be achieved by adjusting the temperature of a single longitudinal mode Fabry Perot or Distributed Bragg Reflector (DBR) laser diode. For a Fabry Perot laser diode, the wavelength varies typically at a rate of 0.3nm°C (SDL 5412) requiring a temperature change of 4°C to obtain \( d\lambda_{q2}=1.1nm \). The maximum tuning range of such a device is approximately 7nm. DBR laser diodes exhibit a reduced thermal wavelength sensitivity typically at 0.07nm°C (SDL 5712) requiring a temperature change of 16°C to obtain \( d\lambda_{q2}=1.1nm \). The tuning range of such
a device is limited to about 2nm. Although the tuning requirements can be fulfilled by both types of laser diode, the DBR diode is inherently more stable being less susceptible to optical feedback and provides smooth wavelength tuning across the wavelength range. The disadvantage of the DBR laser is that if a thinner interferometer than 50µm were required to increase the acoustic frequency response, the limited tuning range might be insufficient to obtain a large enough phase shift to return the interferometer to quadrature. The Fabry Perot type laser diode has a wide tuning range but it is highly susceptible to even very low levels of feedback often requiring an expensive optical isolator to be used. Furthermore, it can exhibit numerous mode hops when being thermally tuned making it difficult to locate the quadrature point with certainty. An alternative to the Fabry Perot and DBR tunable laser diodes is the type of external cavity laser diode used by Dorighi et al.\textsuperscript{182} to stabilise a Fabry Perot interferometer. These lasers can offer narrow linewidth, continuous smooth tuning over 20nm but are very expensive. Whichever laser system is used, the wavelength could be adjusted manually to obtain quadrature on start-up and maintain it thereafter or alternatively, an automatic quadrature control system using standard feedback techniques\textsuperscript{141} could be implemented. A requirement of any quadrature control system is that it should have a response time comparable to the timescale over which environmental influences occur. Thermal tuning of a laser diode is usually associated with a response time of a few seconds and would therefore be adequate for most situations in which ambient thermal fluctuations are the principal source of phase bias variations.

5.9 Concluding remarks

This chapter has examined the feasibility of the Fabry Perot polymer film ultrasound sensor and shown that it has the potential to fulfill the sensitivity and bandwidth requirements to detect thermoelastic waves in arterial tissue. The practical implementation of the system would be straightforward to achieve. The backing configuration for optimal acoustic and phase sensitivity can be simply realised by using a water backed film. Furthermore, no special optoelectonic or optical components are
required since the performance of readily available laser diodes and photodiodes have been shown to be adequate.

It is perhaps of interest to note that the Fabry Perot polymer film sensor can in many ways be considered as an optical analogue to PVDF ultrasound detectors operating in thickness mode; both are based upon the measurement of acoustically-induced changed in thickness in a thin polymer film which is acoustically well matched to water. This means that many of the established principles used in PVDF transducer design could be transferred to the Fabry Perot polymer film sensor. Existing models of acoustic field interactions of PVDF devices could be used to provide, for example, a useful insight into the directional performance of the sensor, an aspect that has not been discussed in this work.
CHAPTER 6

THE FABRY PEROT POLYMER FILM ULTRASOUND SENSOR - EXPERIMENTAL

The aim of the experimental work described in this chapter was to demonstrate the feasibility of the transparent Fabry Perot polymer film ultrasound sensor. Preliminary demonstration of the concept was carried out using air-backed sensing films illuminated firstly with a collimated laser beam and secondly with the divergent output of an optical fibre. Subsequent experiments concentrated on the optimal water-backed configuration discussed in chapter 5. In these experiments, the sensing film was mounted in an experimental sensor head at the end of an optical fibre and the system characterised using a PVDF membrane hydrophone as a reference. Miniature optical fibre probes were then developed to demonstrate that the small size required for intravascular use is feasible.

6.1 PRELIMINARY EXPERIMENTS

This section describes the initial experiments carried out to demonstrate that a polymer film acting as a low finesse Fabry Perot interferometer could be used to detect
acoustic waves at ultrasonic frequencies. Initially a bulk optical arrangement was set-up to illuminate the sensing film with the collimated output of a He-Ne laser. Sensing films of different materials and thicknesses were evaluated. A 400μm optical fibre was then used to illuminate the sensing film and transmit the reflected intensity output to the photodiode.

6.1.1 Collimated beam illumination

6.1.1.1 Experimental set-up

Figure 6.1 shows the experimental arrangement. Thermoelastic waves were generated by the absorption, in washable blue Solv-X Quink ink, of 25ns pulses at 532nm emitted by a frequency doubled, Q switched Nd:YAG laser. The repetition rate of the Nd:YAG laser was 20 Hz and the beam diameter was 3mm. The thermoelastic waves propagated through the ink and were detected by the sensor film mounted on the
opposite side of the ink bath. With this arrangement, a 2mJ pulse produces a
thermoelastic wave of peak acoustic pressure $P$ of approximately 0.3MPa arriving at the
sensing film. This figure was obtained using the calibrated hydrophone arrangement
described in chapter 2 to generate thermoelastic waves in the same ink absorber. An
appropriate correction was then applied to take into account the acoustic losses of the
two experimental set-ups in order to estimate the acoustic pressure incident on the
sensing film. The sensor film was illuminated with a 1mm diameter beam from a 1mW
He-Ne laser at 633nm at normal incidence and the reflected output directed, via a 50/50
beamsplitter, on to a 25MHz silicon PIN photodiode of active diameter 0.8mm
containing an integral transimpedance amplifier. The output of the photodiode was
displayed and signal-averaged over approximately 30 shots using a 500MHz digitising
oscilloscope. Initial experiments were carried out using a 1mm thick PMMA sensor
element. This was "silvered" on both sides by flash evaporation of aluminium to give
optimum reflection coefficients $r_1=0.4$ and $r_2=1$ (section 5.2.2). The Q switched pulse
energy incident on the ink absorber was 3.3mJ. Thereafter, a variety of unsilvered
polymer films of different thicknesses were evaluated using a pulse energy of 2mJ.
Quadrature was achieved in each case by exploiting the fact that a polymer sheet or film
is not perfectly flat. The phase bias therefore varies for different points on the film that
are illuminated. The position of the ink bath, and hence the area of the sensor film that
is illuminated by the He-Ne beam, was adjusted until it was at a point equidistant
between locations corresponding to two consecutive signal minima. At this point it was
assumed that the interferometer was at quadrature.

6.1.1.2 Results

(i) Silvered PMMA sensor $l=1mm$ Although demonstrating good signal-to-noise
ratio, the shape of the signal shown in figure 6.2 is quite different from the bipolar
shape generated in the same absorber but detected using the PVDF membrane
hydrophone shown in figure 2.4(a) in chapter 2. It is the thickness of the sensor
compared with the wavelength of the thermoelastic wave and the acoustic impedance
mismatches at the sensor boundaries that account for the difference.
Consider the sequence of events depicted in figure 6.3. A bipolar thermoelastic wave is incident on say a PMMA sensor element that is of the order of at least a few acoustic wavelengths thick (1). As the wave begins to strain the sensor, the output increases to a maximum until exactly one-half wavelength has penetrated (2). Thereafter the rarefaction part of the wave produces a negative strain reducing the output (3). When the thermoelastic wave has completely penetrated the sensor (4), the strains produced by the compression and rarefaction parts of the thermoelastic wave are equal and opposite resulting in zero output. Thus the first output pulse detected by the photodiode is monopolar and positive. The thermoelastic wave then travels across the sensor. When it arrives at the opposite side it is incident on a PMMA/air boundary. The acoustic impedance of air is very much less than that of PMMA so the wave is reflected with equal amplitude but opposite sign. There is now a net negative strain due to the superposition of the rarefaction part of the incident wave and the reflected wave which is also a rarefaction (5). The peak of the second detected pulse is therefore negative and twice that of the first pulse. The reflected wave travels across the sensor (6) during which time the output is zero. The third pulse begins as the thermoelastic wave exits the sensor (7) leaving the compression part of the wave to strain the sensor. The effect of
these reflections within the sensor is to produce the characteristic "triple peak" output (subsequent pulses are due to multiple reflections within the sensor) that is observed with air/sensor/water boundaries. It is interesting to note that even with a sensor thickness much greater than the acoustic wavelength, the fast rise time of the incident thermoelastic wave can still be detected.

(ii) Transparent polymer film sensors The output shown in figure 6.2 was achieved with the benefit of a very high phase sensitivity resulting from the use of optimum values of $r_1$ and $r_2$. For this application it is intended that the Fresnel reflection coefficients at the two faces of the sensor film will define the values of $r_1$ and $r_2$ so that the sensor is transparent. Various unsilvered polymer films of different thicknesses were
Figure 6.4 Output from transparent polymer film sensors using collimated beam illumination. (a) \( l=1\text{mm} \) - PMMA, (b) \( l=250\mu\text{m} \) - PMMA, (c) \( l=100\mu\text{m} \) - PET. Pulse energy=2mJ, \( P\sim0.3\text{MPa} \).
therefore evaluated. The results using PMMA and PET films are shown in figure 6.4. The triple peak behaviour described above is apparent in each case with the separation of the three peaks decreasing as the thickness is reduced. The amplitudes of the signals are an order of magnitude lower than the those shown in figure 6.2 due to the much lower phase sensitivity of the unsilvered films. The amplitude of the 100μm PET sensor is larger than the other two because of its higher phase sensitivity which is as a result of the higher refractive index of PET (n=1.64) compared with PMMA (n=1.49).

6.1.2 Optical fibre illumination

6.1.2.1 Experimental set-up

For the optical fibre detection of thermoelastic waves, the experimental technique is similar to that described in section 6.1.1.1 with the exception that a 2m length of

![Experimental arrangement for the optical fibre detection of thermoelastic waves.](image)

Figure 6.5 Experimental arrangement for the optical fibre detection of thermoelastic waves.
400μm optical fibre was used to illuminate the sensor and transmit the reflected intensity modulated signal to the photodiode. The arrangement is shown in figure 6.5. To overcome additional optical losses incurred by the optical fibre, the 1mW He-Ne was replaced by a 7mW version. The method and conditions of thermoelastic wave generation were identical to those described in section 6.1.1 as were the data collection and processing methods. For improved frequency response, a 50μm PET film was used as the sensor film.

6.1.2.2 Results

The signal obtained is shown in figure 6.6. Taking into account the losses incurred by coupling light into and out of the fibre, the higher laser power and the reduced acoustic sensitivity of the 50μm film, the amplitude of the signal should be at least a factor of 2 higher than the 100μm PET sensor used in the collimated illumination scheme. The results in figure 6.4(c) and figure 6.6 therefore indicate that there is a substantial signal degradation when the sensor is illuminated with the divergent light emerging from an optical fibre. Despite this, the system sensitivity of this configuration is a respectable 20mv/MPa. The system is limited by the photodiode noise and, with signal averaging (30 averages), this corresponds to a minimum detectable signal of 12.5KPa over a 25MHz bandwidth.

6.1.3 Signal cancellation

To identify the source of the signal degradation when using the optical fibre arrangement of figure 6.5, the signal cancelling effect described in section 5.2.4 was investigated. The experimental arrangement was similar to that shown in figure 6.5 whereby the output of an optical fibre was used to illuminate the sensor film. Instead of transmitting the light reflected from the sensor film back along the fibre however, a beamsplitter was inserted between the distal end of the fibre and the sensor film. Light reflected from the sensor film was then reflected from the beamsplitter, passed through an aperture and imaged, using a lens, on to the photodiode. By imaging the output of the sensor film directly on to the photodiode in this way, any possible deleterious effects associated with the transmission of the signal along the optical fibre could be eliminated.
Since the fibre effectively acts as an extended source, the interference pattern reflected from the film appears as a central maximum or minimum surrounded by a set of concentric circles. The number of rings observed depends upon the extra path length $\Delta l$ (equation 5.14) taken by the ray with the greatest angle of incidence compared to that taken by a ray at normal incidence. The number of rings is therefore a function of the divergence angle of the light emerging from the optical fibre and the thickness of the interferometer. Varying the number of rings that are imaged on to the photodiode by adjusting the aperture size provides a means of studying signal cancellation. For example, if the aperture is made small enough, only the central maximum or minimum is imaged on to the photodiode. All the rays that interfere to produce this part of the interference pattern have a very similar angle of incidence on the sensing film so their optical path lengths and hence associated phase biases are similar. Thus the signal cancelling effect described in section 5.2.4 will be small. When studying this effect, the important measurable quantity is the ratio of the signal $dI_o$ to the total dc level $I'_{dc}$. $I'_{dc}$ is the sum of the reflected interferometer dc level $I_{dc}$ (equation 5.10) and reflections such as those from the launch lens and the two ends of the fibre that are also imaged on to the photodiode. Using a 0.175mm PET sensing film, $dI_o/I'_{dc}$ was found to increase from
1.1%, when the aperture was fully open, to 3% when it was reduced to the minimum size that allowed enough light through for a detectable signal. When using the collimated beam set-up for this sensing film, \( \frac{dI}{I'} \) was 3.8%. Thus, as the part of the interference pattern arising from the divergent or non axial rays is removed, \( \frac{dI}{I'} \) increases and approaches that obtained with a collimated beam. Since aperturing in this way is effectively the same, in terms of \( \frac{dI}{I'} \), as reducing the extra path length \( \Delta l \), this result demonstrates that the signal degradation observed when using an optical fibre is due to the signal cancellation effects described in section 5.2.4.

6.2 DEVELOPMENT AND CHARACTERISATION

The previous section showed that the Fabry Perot polymer film sensor could detect thermoelastic waves with good SNR in an air-backed configuration. The system is, however, far from optimal. When air-backed, the Fresnel reflection coefficients on the two sides of the sensor film can differ by more than an order of magnitude (depending on the refractive index of the film) resulting in low fringe visibility. The non uniform frequency response characteristics of this configuration, due to the low acoustic impedance of air, also means that the sensor gives a poor representation of a thermoelastic wave. The aim of the work described in this section was to overcome these limitations and form the basis of a practical device using a water-backed sensing film incorporated in a sensor head mounted at the end of an optical fibre. A shortcoming of the measurements described in the previous section is that the experimental arrangement did not allow direct comparison with a known measurement reference device for characterisation purposes. In this section, a new experimental set-up is described permitting detailed characterisation of the sensor performance using a calibrated PVDF membrane hydrophone as a reference.
6.2.1 Experimental sensor head design

The sensor head was designed primarily for evaluation purposes so that the sensing film could be easily removed and replaced, the fibre-film separation varied and different areas of the film interrogated for phase bias control. The latter was necessary since, for the purposes of demonstrating the concept, a fixed wavelength He-Ne laser was used rather than a tunable laser diode source. The sensor head, shown in figure 6.7, was constructed from brass and is of overall diameter 12mm. The optical fibre is inserted through an eccentrically located hole in a PTFE insert in the main screw enabling it to be rotated without gripping the fibre. A plastic disc is bonded to the fibre a few mm from its distal end. Turning the screw advances the fibre towards the film and a locking ring enables the screw to be locked at the appropriate fibre-film separation. The fibre is eccentrically located, so rotation of the screw results in the light emerging from the end of the fibre scanning a circle on the polymer film. Since a typical polymer film is not perfectly flat, the phase bias is different for each point on the film that is illuminated. Exploiting this fact provides a means of setting the interferometer at quadrature. Firstly, the main screw was rotated until it had passed through two consecutive points that produced a signal minimum. The screw was then turned back until the illuminated region of the film was at a point equidistant between the two points corresponding to the signal minima. At this point the interferometer was at quadrature and the signal a maximum.

![Illustration of the experimental sensor head](image)

**Figure 6.7** Illustration of the experimental sensor head.
The sensing film used was 50µm thick PET film. This thickness was chosen as an acceptable compromise between the requirements of acoustic sensitivity and limiting the signal cancelling effects described in section 5.2.4. PET has a high refractive index giving high phase sensitivity thus reducing the laser power requirements. PET also has excellent optical properties and is readily available on a commercial basis.

6.2.2 Experimental arrangement and method

The experimental arrangement for characterising the sensor is shown in figure 6.8. Ultrasonic thermoelastic waves were generated by the absorption of 20ns optical pulses at 532nm in Quink Solv-x ink. The optical beam diameter was 4mm. The optical pulses, produced by a frequency doubled Q switched Nd:YAG laser, were directed on to the ink which was placed in contact with the Perspex window in the bottom of the water bath. The resulting thermoelastic waves propagate vertically upwards and are reflected from a glass block angled at 45° acting as an acoustic reflector. A calibrated 25MHz 50µm bilaminar PVDF membrane hydrophone of active diameter 1mm immersed in the water bath was used as a reference hydrophone. A 7mW He-Ne laser operating at 633nm was used to illuminate the sensor film via a 2m length of 400µm all-silica optical fibre. The intensity modulated signal reflected from the sensor film was transmitted back along the fibre and directed on to a 25MHz silicon PIN hybrid photodiode of active diameter

![Figure 6.8 Experimental arrangement for measuring sensor performance.](image-url)
Experiments were carried out to measure the sensitivity, frequency response, linearity and stability of the optical fibre sensor. For the experiments in which linearity and stability were being measured, the sensor head was positioned directly behind the hydrophone in the water tank so that the same region in the acoustic field was measured simultaneously by both devices. For the sensitivity and frequency response measurements, even the relatively small acoustic perturbation produced by the membrane hydrophone was unacceptable. These measurements were therefore obtained by immersing the optical fibre sensor in the tank, taking a measurement and then removing it and replacing it with the hydrophone. To ensure that the same point in the ultrasound field was being measured, alignment with the incident thermoelastic waves was carried out optically using the low power fixed Q output of the Nd:YAG laser as an alignment beam. The position of the hydrophone/sensor was adjusted until the Fresnel reflection from the Perspex window was visible on the active area of the device. Fine adjustment was then carried out by steering the optical beam, whilst thermoelastic waves were being generated in the liquid absorber, until the output of the device under alignment was a maximum. In addition, care was taken to ensure that the acoustic path length (approximately 6cm) travelled by the thermoelastic wave was identical for both measurements. This was achieved to a high degree of accuracy by adjusting the position of each device so that the time from the firing of the Q switched laser pulse to detection of the thermoelastic wave was the same. There is significant variation from pulse to pulse in the output of the Q switched Nd:YAG laser so the output from each device was averaged over 60 shots using the signal averaging function of a 500MHz digitising oscilloscope.

6.2.3 Sensor mode of operation

In principle, it is possible that the signal detected by the sensor is due to acoustically induced bulk movement/displacement of the sensing film. In this case it would be the Fresnel reflections from the cleaved distal end of the fibre and the front face of the film (the side closest to the end of the fibre) that interfere. In order to verify that it is changes in the thickness of the sensing film that are being detected rather than its bulk displacement, the sensor was set up and aligned so that a clearly detectable

0.8mm.
acoustic signal could be observed. The back face of the film (the side furthest from the fibre) was then abraded with sandpaper thus destroying the specular reflection from the back face. The only significant interference is now between the reflections from the end of the optical fibre and the front face of the polymer film. Under these conditions no signal due to an incident acoustic wave could be detected verifying the thickness mode of operation of the sensor.

6.2.4 PVDF membrane hydrophone and sensor comparison

The signals measured by the optical fibre sensor and the hydrophone are shown in figure 6.9. A measurement was made firstly with the optical sensor. This was then removed and replaced with the membrane hydrophone as described in section B so that the same point in the acoustic field was measured by both devices. Signal averaging over 60 pulses was performed in each case. The first signal A is a small amplitude thermoelastic wave generated at the boundary between the upper surface of the of the Perspex window and the surrounding water. The second signal B is the thermoelastic wave generated in the ink. C is the reflection of A within the Perspex window. R is the reflection of the thermoelastic wave B from the tip of the optical fibre (showing that the fibre-film separation is approximately 1.6mm) and therefore appears only on the sensor output. These results indicate that the sensor exhibits comparable signal-noise ratio to the membrane hydrophone. In addition, the shape of the wave measured by the sensor, as shown more clearly in the normalised expanded view in figure 6.10, is in good agreement with that of the hydrophone suggesting comparable frequency response characteristics.

6.2.4.1 Sensitivity

The end-of-cable sensitivity of the membrane hydrophone is 150mv/MPa at 10MHz. By direct comparison of the peak positive pressures registered by each device shown in figure 6.9, the sensitivity of the sensor is 61mv/MPa. The sensitivity of the photodiode/amplifier combination is 10mv/μW so this corresponds to a system sensitivity $dI/P$ of 6.1μW/MPa. Taking into account that the laser power is 7mW, the wavelength is 633nm, $dI/P$ is about a factor of 5 lower than the value predicted for normally incident
Figure 6.9 Comparison of (a) the optical fibre sensor output with (b) the 50μm PVDF membrane hydrophone in response to a thermoelastic wave of amplitude 0.1MPa. A - thermoelastic wave generated in the Perspex window, B - thermoelastic wave generated in the ink absorber, C - reflection of B within the Perspex window, R - reflection of B from tip of optical fibre.
light in section 5.4. The reason for this is most likely due to the signal cancelling effects
due to the divergent output of the optical fibre which reduces and limits the phase
sensitivity. The Fresnel reflections from the front and distal cleaved ends of the fibre
that fall on the photodiode further exacerbate this problem by adding to the dc level.

6.2.4.2 Noise

The noise floor of the membrane hydrophone is very low and in these
measurements is dominated by the noise associated with the oscilloscope. It is measured
in this example over 60 averages at 0.050mv (0.33KPa) in a measurement bandwidth
of 25MHz. The corresponding measurement for the sensor, dominated by the
photodiode/amplifier combination noise, over the same measurement bandwidth is
0.14mv (2.30KPa). Thus in terms of signal to noise ratio the sensor output is around a
factor of 7 lower than the PVDF membrane hydrophone.
The frequency response was obtained by taking the Fourier transform $B_s(f)$ of the normalised signal in figure 6.10 measured by the sensor. The Fourier transform of the corresponding signal measured by the hydrophone $B_h(f)$ was then computed and a correction applied to take into account the frequency response characteristics of the hydrophone giving $C_h(f)$ as shown in figure 6.11. This gives the distribution of frequency components arriving at the sensor head. Dividing $B_s(f)$ by $C_h(f)$ gives the frequency response of the sensor and is shown in figure 6.12.

The theoretical frequency response given in section 5.3.4 for the 50μm PET film surrounded by water is also shown along with the calibrated frequency response of the PVDF membrane hydrophone. At low frequencies the correlation between theory and experiment is good. The half wavelength resonance appears as expected at approximately 20 MHz although its magnitude is a factor of 1.4 larger than expected. In the time domain, this apparently enhanced resonance manifests itself as a slightly

![Figure 6.11](image.png)

**Figure 6.11** Acoustic frequency spectra of the signals measured by the optical fibre sensor and the PVDF membrane hydrophone shown in 6.10.
increased tensile component of the thermoelastic wave (figure 6.10). Although not terminated in 50Ω the length of the coaxial cable (0.6m) connecting the hybrid photodiode to the oscilloscope should not have produced a resonance close to this frequency. Further investigation suggested that it may be a peak in the frequency response of the photodiode/amplifier combination coinciding with the 20 MHz acoustic thickness mode resonance that was responsible. An additional or alternative explanation may lie in the fact that the active diameter of the sensor (~400μm) is smaller than that of the hydrophone (1mm). The spatial integrating effect in the near field (which these measurements may be in), described in section 2.4.4, which has the effect of reducing the measured tensile component of the thermoelastic wave, is therefore more pronounced with the membrane hydrophone. This would produce the difference in the tensile components shown in figure 6.10 and therefore the apparent enhancement of frequency components around 20 MHz observed in the sensor frequency response. If this is the explanation, then the discrepancy with the theory is due to the experimental set-up rather than an inherent characteristic of the sensor. To overcome this problem, care should be taken to ensure that the acoustic path length is long enough to be sure that the measurements are truly made in the far-field.

![Graph](image)

**Figure 6.12** Comparison of the optical fibre sensor frequency response with the theoretical model and the PVDF membrane hydrophone response.
6.2.4.4  Linearity

It is assumed the PVDF membrane hydrophone is linear to well beyond 0.6MPa\(^{184}\). The sensor was placed directly behind the hydrophone and the position of both devices adjusted independently until the maximum signal detected by each was observed. Simultaneous readings from the sensor and the hydrophone were taken. This procedure was repeated over a range of acoustic pressures by varying the energy of the optical pulses incident upon the ink absorber. The results, shown in figure 6.13, show excellent linearity of response up to 0.5MPa limited by the maximum pulse energy of the Q-switched Nd:YAG laser. A dynamic range in excess of 30dB was obtained. It should be noted that the acoustic pressures shown in figure 6.13 are those measured by the hydrophone. The true acoustic pressure arriving at the sensor head is somewhat lower due to the attenuation of the acoustic signal as it passes through the hydrophone.

![Figure 6.13](image)

**Figure 6.13** Optical fibre sensor output as a function of the acoustic pressure as measured by the PVDF membrane hydrophone.
Adequate short and long term stability are essential for a practical measurement device. The usual source of sensitivity fluctuations in homodyne interferometric sensors is due to environmentally induced variations in phase bias particularly in the case of long path interferometers. In this scheme the thinness of the polymer film acting as the interferometer means that it is relatively insensitive to external perturbations that are likely to occur in the short term. The sensitivity over a period of 80 hours is shown in figure 6.14. The experimental method was the same as that used for the linearity measurements. Figure 6.14 shows that the sensor is stable within the uncertainty of the measurement typically over a period of several hours and over 80 hours the output does not fluctuate by more than 10%. This indicates low sensitivity to ambient temperature and pressure fluctuations.

![Figure 6.14 Variation of optical fibre sensor output with time.](image)

Other significant factors that may lead to changes in sensitivity are those influences that act upon the optical fibre downlead. Vibration produced by rapidly and vigorously shaking the fibre had virtually no effect on the signal. The signal was, however observed to decrease when the fibre was bent to a significant degree. Effectively, the the higher order modes (which remain unfilled over a short length of fibre) become filled when the
fibre is bent, increasing the angle of divergence of the light emerging from the end of the fibre and thus bringing into effect the signal cancelling effects described in section 5.2.4. This dependence of the signal on fibre bending was reduced when the modes were scrambled and the input launch conditions adjusted so that all the propagation modes were filled resulting in a constant angle of divergence. This improvement in stability, however, is obtained at the expense of phase sensitivity because, when all the modes are filled, the angle of divergence of the light emerging from the fibre (and therefore signal cancellation) is greatest.

### 6.2.5 Rigid backed sensing film

It would be useful to find some way of removing the reflection of the thermoelastic wave (R in figure 6.9) from the tip of the fibre. This may coincide with the arrival of a signal to be detected and can affect the low (<1MHz) frequency response of the sensor. Rigid backing offers a solution as it gives high acoustic sensitivity and a uniform although reduced frequency response. It could conceivably be achieved by attaching a transparent material of relatively high acoustic impedance, such as glass, to the back of the sensing film. This however would destroy the fringe visibility since the Fresnel reflection coefficient due to the refractive index mismatch between PET and glass is very low. A novel approach has been found to overcome this by ensuring that there is a gap between the film and the backing that, in acoustic terms, is of negligible thickness but large enough to allow a thin layer of water to exist. The water layer maintains the fringe visibility at unity but allows the sensor to be effectively rigid backed. In practice this was achieved by using the cleaved end of the fused silica optical fibre as the backing material. The fibre was moved gradually closer towards the sensing film until the thermoelastic waves reflected from the tip of the fibre were superposed on to the initial wave incident on the sensor film. The effect of this is to increase the acoustic sensitivity by a factor of 2 as shown in the theoretical model shown in figure 5.7. This is verified in figure 6.15 which shows the output of the sensor before and after the fibre was brought close to the sensing film. Some broadening of the second-half of the pulse is apparent in figure 6.15(b) due to the reduced frequency response of the rigid backed configuration.
Figure 6.15 (a) Water-backed configuration showing multiple reflections between the sensing film and the fibre tip and (b) superposition of the same reflections in a rigid-backed configuration.
6.2.6 Miniature optical fibre ultrasound probes

The previous sections have shown that excellent sensitivity and frequency response characteristics can be obtained using a water backed sensing film illuminated by the output of an optical fibre. The experimental sensor head (figure 6.7) used for these experiments is clearly too large to be inserted into a blood vessel. Two probes of a smaller diameter were therefore constructed and tested to demonstrate that a miniature device was feasible. The design of each probe was identical and is shown in figure 6.16. Photographs of 1mm and 2mm outer diameter sensor heads are shown in figure 6.17(a) and figure 6.17(b) respectively.

The sensor heads were assembled under a microscope. A piece of 50µm thick PET film was bonded with epoxy to the end of a PMMA tube.

Figure 6.16 Schematic of miniature sensor head design.

The 400µm all-silica optical fibre was prepared by stripping the last 10mm of the buffer coating and cleaving the end. A drop of water was syringed into the tube, the fibre inserted and bonded into position at the required fibre-film separation. These probes demonstrated comparable sensitivity and frequency response characteristics to those presented in section 6.2.4. Since the He-Ne was used as the laser source, a certain amount of trial and error was required to select a piece of PET film that was of a thickness that corresponded to quadrature.
6.2.7 Quadrature control

It has been shown that it is possible to demonstrate the concept in the laboratory using a fixed wavelength laser such as a He-Ne and operate at quadrature by exploiting the variable thickness of a typical polymer film. In a practical system, a wavelength tunable laser source would be required for remote quadrature control to set the initial phase bias and compensate for environmentally-induced drifts in sensitivity. Two laser diodes have been evaluated for this purpose: a Fabry Perot SDL 5412 laser diode and a distributed Bragg reflector (DBR) SDL 5712 laser diode. Both devices contain an internal Peltier element and thermistor. These allow the temperature of the laser diode to be adjusted and monitored for the purpose of tuning and stabilising the laser wavelength. The laser diodes were evaluated using a collimated beam set-up similar to that shown in figure 6.1. The sensing film used was 50μm thick PET. The drive current to the Peltier element was varied and the dc output intensity reflected from the sensing film was monitored as a function of thermistor resistance - for small temperature changes the thermistor resistance is approximately linear with temperature. The laser diode output was monitored by diverting part of its output to a power meter using the beamsplitter shown in figure 6.1. This enabled changes in the reflected intensity from
the sensing film due to thermally-induced variations in the output power of the laser
diode to be compensated for.

Whilst the Fabry Perot laser diode has the attraction that it can be tuned through
a range of approximately 7nm, it was found to be difficult to use as it was highly
susceptible to optical feedback. Even when feedback was eliminated it exhibited
numerous mode hops throughout the wavelength tuning range making it very difficult
to obtain the quadrature point with any degree of certainty. The DBR laser diode proved
to be more suitable. It was far less susceptible to optical feedback and provided smooth
wavelength tuning without mode hops. This can be seen in figure 6.18 in which several
tuning curves corresponding to different points on the sensing film, and therefore
different initial phase biases, are shown. Figure 6.18 shows that the tuning range is just
sufficient to cycle through a minimum and maximum of the interferometer transfer
function (i.e a phase shift of \(\pi\) radians). A potential limitation with the DBR laser diode
is that if a thinner sensing film were used, to increase the acoustic frequency response,
this tuning range might not be sufficient. It might not always be sufficient to produce
the minimum phase bias shift of \(\pi/2\) radians required to return to quadrature if the phase
bias has drifted to a maximum or minimum on the interferometer transfer function.

![Figure 6.18](image)

**Figure 6.18** Thermal wavelength tuning using SDL 5712 DBR laser diode and 50\(\mu\)m
thick PET sensing film. Graph shows reflected intensity from sensing film as a function
of laser diode temperature (represented by thermistor resistance) for three different
points on film.
6.3 CONCLUSION

An extrinsic optical fibre sensor has been described and demonstrated showing that a thin polymer film acting as a Fabry Perot interferometer can be used for the detection of ultrasound. High sensitivity and a flat wideband frequency response have been demonstrated in good agreement with a theoretical model. The sensor was found to exhibit comparable performance to a PVDF membrane hydrophone making it suitable for the detection of thermoelastic waves in arterial tissue.
CHAPTER 7

THE PHOTOACOUSTIC PROBE

This chapter demonstrates the concept of the all-optical photoacoustic probe described in chapter 4 whereby the generation and detection of thermoelastic waves takes place at the tip of an optical fibre. The system was evaluated using a non-scattering liquid absorber and post mortem arterial tissue.

7.1 EXPERIMENTAL DETAILS

This section describes the experimental implementation of the photoacoustic probe. The system consists of a delivery optical fibre with a transparent Fabry Perot polymer film sensor mounted at its distal end in contact with a target absorber. A Q switched Nd:YAG laser provides nanosecond sub-millijoule optical pulses. These are launched into the fibre, transmitted through the sensor and absorbed in the target producing thermoelastic waves which are then detected by the sensor.
7.1.1 Sensor head

The experimental sensor head described in section 6.2.1 of the previous chapter was used for most of the experiments described in this chapter. This allowed adjustment of the phase bias and easy replacement of the sensing film if damaged by the Q switched laser pulse. 50μm thick PET was used as the sensing film. To demonstrate the feasibility of a miniature photoacoustic probe, a 1mm outer diameter sensor head was also constructed and evaluated.

7.1.2 Experimental set-up

A schematic of the experimental set-up is depicted in figure 7.1. Optical pulses at 532nm and of duration 25ns, emitted by a frequency doubled Q switched Nd:YAG laser operating at a repetition rate of 20Hz, were delivered to the target using a 1m length of 400μm all-silica optical fibre. The sensing film was interrogated by the 633nm CW output of a 7mW He-Ne laser which was combined with the 532nm Q switched optical pulses using a dichroic beamsplitter and launched into the fibre. The light

![Figure 7.1](image)

**Figure 7.1** Schematic of the experimental arrangement for the optical fibre generation and detection of thermoelastic waves. A - 1.06μm colour glass filter, B - 532nm colour glass filter, C - 633nm bandpass filter.
reflected from the sensor film was directed via a 50/50 beamsplitter on to a 25MHz silicon PIN photodiode containing an integral transimpedance amplifier, the output of which was displayed and signal averaged using a 500MHz digitising oscilloscope. In order to avoid scattered light from the launch optics and the front end of the fibre falling on to the photodiode, the fibre was angled relative to the axes of the laser beams and an aperture placed in front of the photodiode to block the path of any stray reflections. Appropriate colour glass filters were used to reduce the reflections of the 532nm pulses from the sensor film on to the photodiode (B) and to attenuate the 1.06μm fundamental component of the Nd:YAG output (A). For additional rejection, a 633nm bandpass filter (C) was also used. Despite the use of these optical filters, it was found that a significant amount of unwanted Q switched light was still incident on the photodiode. In addition, a short pulse of electrical noise from the Q switch of the Nd:YAG was also present. When the sensor is very close to or in direct contact with the target absorber, the photoacoustic signal, as represented by an intensity modulation of the 633nm light reflected from the sensor, coincides with the onset of the much larger Q switched optical and electrical noise signal. This can lead to the photoacoustic signal being completely obscured by the Q switched optical pulse. The two techniques outlined below were used to prevent this. The noise subtraction method described below in section 7.1.3 was sufficient in most cases. The acoustic delay method described in section 7.1.4 was used in addition to that of 7.1.3 for the detection of very low amplitude signals such as those generated in tissue.

7.1.3 Q switched noise subtraction

This method reduced the effect of the Q switched noise signal by subtracting it from the waveform containing the photoacoustic signal. A "reference" waveform containing only the signal-averaged Q switched noise was obtained by switching the He-Ne laser off whilst continuing to operate the Nd:YAG laser. The He-Ne was then switched on so that the sensor was interrogated and the recorded waveform (taken over the same number of averages) contained both the photoacoustic signal and the Q switched noise. The noise reference waveform was then subtracted leaving only the photoacoustic signal. Whilst the Q switched noise signal, which exhibits a high level of repeatability between laser shots, can be removed in this way a disadvantage is that the
uncorrelated photodiode noise (section 5.5.2) increases by a factor of \( \sqrt{2} \).

7.1.4 Acoustic delay

An acoustic delay of a few hundred nanoseconds was used to temporally separate the photoacoustic signal from the Q switched noise signal. This was achieved by placing a transparent 5\( \mu \)m thick PVC (polyvinyl chloride) film on the surface of the target absorber and inserting a thin layer of water between the sensor and the PVC film as shown in figure 7.2. The acoustic transparency of the PVC film was verified by comparing the characteristics of a thermoelastic wave before and after propagation through the film. No significant attenuation or change in temporal characteristics was observed.

![Acoustic delay](image)

Figure 7.2 Acoustic delay.

7.1.5 Target absorbers

Measurements were carried out initially using a non-scattering liquid absorber of known optical properties to characterise the system. The liquid absorber used was "washable" blue Solv-X Quink ink. To demonstrate the feasibility of the system for characterising arterial tissues, experiments were carried out to generate and detect thermoelastic waves in post mortem human aorta. The tissue samples were prepared within 12 hours of excision, rinsed in saline to remove traces of blood and then fully immersed in saline. Care was taken to ensure that the tissue was lying flat and only the more homogeneous non-atheromatous areas were targeted for repeatability.
7.1.6 Sensor film damage threshold

Since the optical pulse has to pass through the sensor film at the end of the fibre it is clear that it must not exceed the damage threshold of the polymer sensing film. For a 50µm thick PET film, with both sides in contact with water, this was measured as approximately $5\text{mJ/mm}^2$ using a 20ns pulse at 532nm. This is the instantaneous damage threshold - i.e. damage produced by a single shot. Cumulative damage was produced at lower levels whereby the sensing film appeared to be unmarked initially but after a number of shots became damaged. A fluence of about half the instantaneous damage threshold was therefore used.

7.2 RESULTS

7.2.1 Non-scattering liquid absorber

Figures 7.3(a) and 7.3(b) show the signals obtained in the ink absorber using the experimental sensor head described in section 6.2.1. In both cases the pulse energy measured at the distal end of the fibre was 0.36mJ which corresponds to a fluence of $2.9\text{mJ/mm}^2$ at the fibre tip. The pulse duration was 25ns. The waveform shown in Figure 7.3(a) was obtained when the sensor head was immersed in the ink absorber ($z=0$) and the subtraction technique of section 7.1.5 used to remove the Q switched noise signal. The time delay of 1.01µs between the photoacoustic signal B and the reflection from the fibre tip R corresponds to a distance between the tip of the fibre and the surface of the ink of 0.76mm. The measured half angle of divergence of the 400µm optical fibre was $6^\circ$ giving a spot size of $0.25\text{mm}^2$ on the ink surface and an incident fluence of $1.44\text{mJ/mm}^2$. Figure 7.3(b) shows the waveform obtained using the acoustic delay arrangement described in section 7.1.4. The first signal represents the optical and
Figure 7.3 Thermoelastic waves generated in ink absorber using experimental sensor head of figure 6.7: (a) with sensing film in contact with ink absorber (z=0) and (b) with acoustic delay (z=0.82mm), B - thermoelastic wave generated in ink, R - fibre-end reflection. Pulse energy=0.36mJ, pulse duration=25ns.
electrical Q switched noise signal. The peak of this signal is cut-off by the digitising oscilloscope - in reality it is at least a factor of two greater than shown. The second signal B is the thermoelastic wave produced by the absorption of the 532nm Q switched pulse in the ink. The time difference between the Q switched noise signal and B is 0.55µs and, using a speed of sound of 1500m/s, represents a spatial separation \( z = 0.82 \text{mm} \) between the surface of the ink and the sensing film. R is the reflection of B from the tip of the fibre and the 1.01µs interval between the two corresponds to a separation of 0.76mm. The distance between the tip of the fibre and the surface of the ink is therefore 0.76+0.82=1.58mm resulting in a spot area of 0.42mm² at the surface of the ink. For a pulse energy of 0.36mJ this gives a fluence incident on the ink surface of 0.86mJ/mm².

Of particular interest, are the shapes of the thermoelastic waves obtained in each case. When the sensing film is in direct contact with the ink absorber (figure 7.3(a)), the shape of the thermoelastic wave is essentially monopolar. In figure 7.3(b), when there is a source-detector separation of 0.82mm the wave appears to have become bipolar. The evolution from monopolar to bipolar is shown more explicitly in figure 7.4 in which thermoelastic waves for three source-detector separations \( z \) are shown. By the time the wave has propagated a distance of 1.5mm the signal has become a symmetrical bipolar

![Figure 7.4](image-url)

**Figure 7.4** Evolution of monopolar to bipolar shape for increasing source-detector separations \( z \). Pulse energy=0.36mJ, pulse duration=25ns.
shape. The monopolar/bipolar aspects of thermoelastic wave propagation are discussed more fully in chapter 2 (sections 2.2.4.2 and 2.4.4) but the transition from monopolar to bipolar is a consequence of diffraction occurring as the thermoelastic wave propagates into the acoustic far field.

To demonstrate that a miniature probe configuration is feasible, the 1mm diameter probe design discussed in section 6.2.6 and shown in figures 6.16 and 6.17(b) was used. Signals generated in ink are shown in figure 7.5. Both an acoustic delay of 0.31μs and the Q switched noise subtraction technique were used. For the same pulse energy, the thermoelastic wave B is considerably smaller compared to those shown in figures 7.3 and 7.4. This is most likely because the thickness of the sensing film did not correspond to quadrature thus reducing the sensitivity of the sensor.

![Figure 7.5](image)

**Figure 7.5** Thermoelastic wave generated in ink using 1mm diameter probe. A - remainder of Q switched noise after subtraction, B - thermoelastic wave generated in ink, R - fibre-end reflection. Pulse energy=0.36mJ, pulse duration=25ns.
7.2.2 *Post mortem* arterial tissue

For these measurements, the photoacoustic signal was much smaller than that obtained using the ink absorber due to the low attenuation of arterial tissue at 532nm. It was therefore necessary to reduce the interference caused by the Q switched noise to a minimum by using both the noise subtraction method of section 7.1.3 and the acoustic delay arrangement of section 7.1.4 to separate the noise signal from the thermoelastic waves. It was also necessary to increase the signal averaging so that it was performed over 80 shots. A typical waveform is shown in figure 7.6 using the experimental sensor head. A is the remainder of the Q switched noise after subtraction and B is the thermoelastic wave generated in the tissue. The much lower absorption at 532nm in aortic tissue means that the amplitude of the photoacoustic signal is around an order of magnitude lower than those observed when using ink as the target absorber. Despite significant optical penetration of light at 532nm, the low signal to noise ratio prevents the detection of a photoacoustic contribution from below the tissue surface.

To increase the signal there are a number of options. Firstly, the fluence delivered to the tissue could be increased subject to the tissue ablation threshold limitations described in section 3.3.3. To achieve this, the pulse energy launched into the fibre could be increased although this would risk damaging the polymer film. Alternatively the fluence delivered to the tissue could be increased by dispensing with the acoustic delay arrangement and placing the tissue directly into contact with the sensing film. This would require a high degree of optical filtering and electrical suppression to reduce the Q switched noise signal to an acceptable level. If a wavelength around the 460nm peak of preferential absorption in atheroma were being used, much higher amplitude thermoelastic waves would be generated due to the increased absorption. There is also scope to increase the sensitivity of the sensor. The sensing film used in these experiments is not necessarily optimal. The phase sensitivity could also be increased by using a larger area photodiode. This would increase the detector saturation threshold thus enabling the interrogating laser power to be increased. It should be possible to obtain an overall increase in the signal by a factor of 5 by a combination of increasing the amplitude of the thermoelastic wave generated and the sensitivity of the sensor.
**Figure 7.6** Thermoelastic wave (B) generated in post mortem human aorta. Acoustic delay=1.02mm, A - remainder of Q switched noise after subtraction, B - thermoelastic wave generated in tissue. Pulse energy=0.36mJ, pulse duration=25ns.

### 7.3 Conclusions

An all-optical photoacoustic probe has been demonstrated. The results show that the probe can generate and detect thermoelastic waves in a liquid absorber with excellent signal to noise ratios. Measurements have also been made in *post mortem* human aorta. The low optical absorption of this tissue at the wavelength used results in relatively small although clearly detectable signals. There is, however, considerable scope to improve both the photoacoustic generating efficiency and the detection sensitivity of the system.

A particular advantage of the acoustic source-detector geometry of the photoacoustic probe is that it enables the detection of the thermoelastic wave in front of and immediately adjacent to the source. Since the measurement is performed in front
of the source there is no acoustic boundary beneath the source to produce reflections that may alter the characteristics of the detected thermoelastic wave. In addition, because the source is immediately adjacent to the detector, the photoacoustic measurement is a true near field measurement thus avoiding diffraction effects which can alter the shape of the signal. Hence a true representation of the thermoelastic wave is obtained. This is extremely difficult to achieve with other methods of detecting photoacoustic signals that require finite source-detector separation.
CONCLUSIONS AND FUTURE WORK

The two principal aims of the project have been fulfilled. Namely, the characterisation of arterial tissue using pulsed photoacoustic spectroscopy and the feasibility of its practical in vivo implementation via the concept of the photoacoustic probe. Specifically, it has been shown that photoacoustic spectroscopy can be used to differentiate between normal and atheromatous tissue using a laser wavelength of 461nm to exploit the preferential optical absorption in atheroma. In addition, it has been shown that spatial information relating to the structure of the tissue can be obtained using a laser wavelength of 532nm which can penetrate the full thickness of the tissue. The Fabry Perot polymer film ultrasound sensor has been shown to have excellent acoustic performance with a sensitivity and bandwidth comparable to that of piezoelectric PVDF devices. The sensor has been incorporated into the photoacoustic probe and used to generate and detect photoacoustic signatures in post mortem arterial tissue. A miniature version of 1mm diameter has been developed demonstrating that the small size required for insertion into a blood vessel is feasible. Further work would involve extending the range of atherosclerotic lesions characterised at different wavelengths to include all variations of fibro-fatty lesions and calcified plaques. Some further work is also required to increase the sensitivity of the photoacoustic probe for practical use. Since the excitation optical pulse energy is limited by the damage threshold of the sensing film, improvements in this respect will most likely be achieved by increasing the sensitivity of the Fabry Perot polymer film sensor at the end of the delivery fibre. The therapeutic aspects of the system, including the ability to withstand the laser ablation pulses, also remain to be developed and evaluated.

It is recognised that laser angioplasty has, in recent years, not fulfilled its original promise. The technique may find certain niche applications but its routine use as an affordable and effective treatment in the immediate future is by no means assured. The
further pursuance of the concept of a combined diagnostic and therapeutic system as discussed in this thesis might therefore be questioned. Dispensing with the therapeutic aspects of the system and developing a purely diagnostic intravascular probe might prove to be a more immediately useful clinical objective. This is because there is a need to have some means of assessing an artery and its atheromatous plaque in vivo, in order to predict its natural history and likely response to intervention. It is not difficult to envisage a sideways-looking intravascular photoacoustic probe based upon the concepts described in this thesis to form a photoacoustic equivalent of intravascular ultrasound (IVUS). In such a scheme, a photoacoustic signature could be obtained at a point on the arterial wall. The probe could then be rotated incrementally to enable an image of the vessel wall to be constructed in a manner notionally similar to mechanically rotated IVUS systems. Since the differences in the optical properties of different arterial tissue compositions and structures are large, compared to the corresponding differences in acoustic properties, it can be expected that the photoacoustic technique will be more sensitive to tissue structure and composition than IVUS thus providing a more powerful investigative tool.

The photoacoustic probe may find use in fundamental studies of laser angioplasty and other pulsed laser ablation procedures. There is a need to gain a greater understanding of the ablative processes particularly with regard to the disruptive effects on the tissue of acoustic transients generated during laser angioplasty. Ultimately it would be desirable to reduce these effects as they can produce damage to the vessel wall beyond the ablation site and, it has been suggested that the resulting increased trauma to the vessel wall may play a role in stimulating restenosis. It would be useful to quantify such photomechanical effects by the direct measurement in vivo of the acoustic pressures and temporal characteristics of the acoustic transients produced by the ablative laser pulse. The photoacoustic probe, suitably calibrated, could provide a means of doing this.

The photoacoustic probe could find application in many areas of photoacoustics which currently use piezoelectric detection methods. This is particularly so where an electrically inert, inexpensive, integrated probe capable of generating and detecting photoacoustic signals on the same side of the target is required. As a research tool for investigating photoacoustic phenomena, particularly diffraction effects, it has excellent potential on account of its unique source-detector geometry that enables near field,
above source measurements to be made.

Perhaps the most exciting development to emerge from this work is the Fabry Perot polymer film ultrasound sensor. It is perhaps the only optical fibre ultrasound sensor that has the potential to rival piezoelectric detectors in terms of acoustic performance, cost and simplicity of configuration. A miniature general purpose ultrasonic hydrophone could be realised by applying optically reflective coatings to the two sides of the sensing film and then bonding the film to the tip of the fibre. In particular, such a hydrophone could provide a solution to the difficulties in obtaining a small active area with piezoelectric devices. A very small active area can be achieved since this is defined by the dimensions of the illuminating beam emerging from the end of the optical fibre. For example, a 50µm core diameter optical fibre could be employed, giving excellent spatial resolution and low directional sensitivity. Furthermore, sensitivity remains constant with active area unlike piezoelectric devices. Another attractive feature of the sensor is that there is a wide range of polymer films available with different acoustic properties and thicknesses. This offers the prospect of tailoring the performance of the sensor for specific applications. Very wide bandwidth is possible since polymer films of the order of 5µm are available giving potential bandwidths of several hundred MHz. The all-optical approach of the system provides immunity from EMI. Furthermore, sensor fabrication is relatively simple and inexpensive offering the prospect of a disposable sensor. Applications that would utilise these advantages are to be found in medical, industrial and ultrasound research fields. In medicine, an increasingly important application is the routine, safety-related characterisation and calibration of diagnostic and therapeutic medical ultrasound transducers. Other medical areas are the measurement of the high amplitude acoustic transients produced by lithotripters and research investigations into acoustic pressures produced in situ by medical ultrasound equipment. The high fields produced by a lithotripter can easily damage a hydrophone and, to avoid cross infection, a hydrophone that is used in situ must be a single use device. The inexpensive nature of the sensor, its physical flexibility and miniature size and potential for high performance make it suitable for all these applications. In industry, hydrophones are required to monitor and characterise transducers used in NDT and industrial processes such as ultrasonic cleaning and mixing. Ultrasonic hydrophones are also required as a specialist research tool in quantifying and mapping the ultrasound fields involved in sonochemistry,
sonoluminescence, cavitation effects and the detection of laser generated acoustic transients. Often these applications require measurements to be made in hostile and electrically noisy environments. The inexpensive nature and EMI immunity that the optical fibre hydrophone offers would present a useful alternative to existing piezoelectric hydrophones.
APPENDIX A : TWO BEAM INTERFEROMETRY

The resultant intensity due to the superposition of two optical fields as in two beam interferometry is derived as follows:

Consider the superposition of two optical fields of the form

\[ E_1 = E_{01} \cos(\omega t + \phi_1) \]  
(A.1)

\[ E_2 = E_{02} \cos(\omega t + \phi_2) \]  
(A.2)

Assuming identical polarisation states, the resultant field \( E_0 \) due to the superposition of \( E_1 \) and \( E_2 \) is

\[ E_0 = E_1 + E_2 \]  
(A.3)

The resultant intensity is given by

\[ I_0 = <E_0^2> \]  
(A.4)

where \(< >\) denotes the time average of a period much greater than \(2\pi/\omega\)

\[ <E_0^2> = <E_1^2> + <E_2^2> + 2<E_1E_2> \]  
(A.5)

and

\[ I_0 = I_1 + I_2 + I_{12} \]  
(A.6)

where

\[ I_1 = <E_1^2> = \frac{E_{01}^2}{2} \]
\[ I_2 = <E_2^2> = \frac{E_{02}^2}{2} \]
\[ I_{12} = 2<E_1E_2> \]  
(A.7)
Evaluating the latter term

\[ 2\langle E_1 E_2 \rangle = 2\langle E_{01} E_{02} \cos(\omega t + \phi_1) \cos(\omega t + \phi_2) \rangle \]

and expanding

\[ 2\langle E_1 E_2 \rangle = 2 E_{01} E_{02} [\langle \cos^2 \omega t \rangle \cos \phi_1 \cos \phi_2 - \langle \cos \omega t \sin \omega t \rangle \cos \phi_1 \sin \phi_2 \\
- \langle \cos \omega t \sin \omega t \rangle \sin \phi_1 \cos \phi_2 + \langle \sin^2 \omega t \rangle \sin \phi_1 \sin \phi_2] \]

evaluating the time averages

\[ 2\langle E_1 E_2 \rangle = E_{01} E_{02} (\cos \phi_1 \cos \phi_2 + \sin \phi_1 \sin \phi_2) \quad \text{(A.10)} \]

\[ 2\langle E_1 E_2 \rangle = E_{01} E_{02} \cos(\phi_1 - \phi_2) \quad \text{(A.11)} \]

From equation A.7

\[ I_{12} = 2\sqrt{I_1 I_2} \cos(\phi_1 - \phi_2) \quad \text{(A.12)} \]

The resultant intensity due to the superposition of \( E_1 \) and \( E_2 \) is therefore

\[ I_0 = I_1 + I_2 + 2\sqrt{I_1 I_2} \cos \phi \quad \text{(A.13)} \]

which describes the familiar "cosine fringes" observed in two beam interferometry where the total phase difference \( \phi \) arising from the optical path difference between the two beams is

\[ \phi = \phi_1 - \phi_2 \quad \text{(A.14)} \]
### Solids

<table>
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<tr>
<th>Material</th>
<th>$\rho$ Kg/m$^3$</th>
<th>$E$ GPa</th>
<th>$c$ m/s</th>
<th>$\sigma$</th>
<th>$\alpha$ neper/m</th>
<th>$Z$ Kg/m$^3$s</th>
<th>$n$</th>
<th>$p$</th>
<th>$\beta_i$ x 10$^5$/°C</th>
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<td>PMMA$^1$</td>
<td>1190</td>
<td>2.5</td>
<td>2700</td>
<td>0.38</td>
<td>57 (2.5)</td>
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<td>0.32</td>
<td>7.3</td>
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<td>2</td>
<td>2000</td>
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<td>110 (1)</td>
<td>3.5 x 10$^6$</td>
<td>1.42</td>
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<td>8-14</td>
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<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
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<td>-</td>
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### Liquids

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<th>$B$ GPa</th>
<th>$c$ m/s</th>
<th>$\alpha$ x 10$^{-15}$ neper/m/Hz$^2$</th>
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<th>$n$</th>
<th>$\beta_i$ x 10$^5$/°C</th>
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<td>-</td>
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<td>1.33</td>
<td>21</td>
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Bracketed figures are frequencies in MHz
$^1$photoelastic constant for flint glass

(see following page for nomenclature)
Abbreviations:
1 polymethylmethacrylate
2 polylethylene terephthalate
3 low density polyethylene
4 polyvinylidene fluoride
5 polyethether sulphone

Nomenclature:
\( p \) - density (Kg/m\(^3\))
\( E \) - Young's modulus (GPa)
\( B \) - Bulk modulus (GPa)
\( c \) - speed of sound (m/s)
\( \sigma \) - Poisson's ratio
\( \alpha \) - acoustic attenuation (neper/m)
\( Z \) - acoustic impedance (Kg/m^2s)
\( n \) - refractive index
\( \beta_1 \) - coefficient of linear thermal expansivity (°C)
\( \beta_v \) - coefficient of volume thermal expansivity (°C)

Sources
Goodfellow Ltd. catalogue, 1995
APPENDIX C: ACOUSTIC AND OPTICAL REFLECTION COEFFICIENTS AT A WATER BOUNDARY

<table>
<thead>
<tr>
<th>Material</th>
<th>$Z$ Kg/m$^2$</th>
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<th>$R$</th>
<th>$n$</th>
<th>$r$</th>
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<td>0.36</td>
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<td>Stainless Steel</td>
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<td>1.94</td>
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* Transmission from water into material
+ reflection at a water boundary

$^1$ polymethylmethacrylate

$^2$ polyethylene terephthalate

$^3$ low density polyethylene

$^4$ polyvinylidene fluoride

$^5$ polyether sulphone

$$
T = \frac{2Z}{Z + Z_w} \quad R = \frac{Z - Z_w}{Z + Z_w} \quad r = \left(\frac{n - n_w}{n + n_w}\right)^2
$$

$Z =$ acoustic impedance of material  
$Z_w =$ acoustic impedance of water  
n =$ refractive index of material  
n$_w =$ refractive index of water
PUBLICATIONS ARISING FROM THIS WORK

Papers:


In addition, aspects of this work have also been presented at a number of meetings and recorded as abstracts.
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