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Ph.D thesis by

R.Beddows B.Sc.(Hons.) M.Sc. Dip.IPEM



**AN ANALYSIS OF FIBRE OPTIC
INTERFEROMETER CONFIGURATIONS FOR
OPTICAL COHERENCE TOMOGRAPHY
APPLICATIONS**

SCHOOL OF ENGINEERING

This thesis is submitted in partial fulfilment of the requirements for the degree
of Doctor of Philosophy of Cranfield University, October 2003

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SCHOOL OF ENGINEERING

Centre for Photonics and Optical Engineering

Ph.D THESIS

Academic year 2003-2004

Ruth Beddows

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CONFIGURATIONS FOR OPTICAL COHERENCE
TOMOGRAPHY APPLICATIONS**

Supervisor: Ralph P Tatam

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Abstract

The Michelson Interferometer has been the principal interferometer configuration for Optical Coherence Tomography (OCT) imaging. For OCT applications, in particular endoscopy, there may be polarisation induced signal fading due to environmental changes within the sample arm. The most significant cause of polarisation induced fading within OCT systems is suggested to be due to changes within the optical fibres. The Fizeau interferometer has been shown to eliminate this polarisation induced fading as the sample and reference arm experience the same environmental effects. However the Signal to Noise Ratio (SNR) of the Fizeau configuration is lower than that of the Michelson because a second processing interferometer is required.

An investigation into suitable interferometer configurations for OCT was carried out in order to increase the performance of the Fizeau interferometer compared with Michelson configuration. It was shown theoretically that by using a circulator and balanced detection, a standard Fizeau interferometer could achieve a signal to noise ratio (SNR) up to 18dB greater than the conventional Michelson interferometer and a comparable SNR to the Michelson interferometer optimised for power conservation.

An analysis into the optical components used in OCT systems was carried out and the optimal Fizeau interferometer design was constructed. The Optimal design was shown to give an increase of 26dB compared to the standard Fizeau interferometer relative to theoretical results using the same optical components. It was experimentally shown that the use of an optical circulator within the Fizeau interferometer improves the SNR compared to the standard configuration.

Acknowledgements

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The author would also like to acknowledge the financial support from the Engineering and Physical Science Research Council and Gloucestershire Royal Hospital under the CASE awards scheme.

Special thanks to my parents John and Rosemary Beddows for all their love, support and encouragement.

Glossary

Adenocarcinoma: carcinoma derived from glandular tissue or in which the tissue cells form recognisable glandular structures.

Barretts esophagus: change in cells within the esophagus due to acidic reflux. Cell changes produce cells similar to those within the stomach and have a greater chance of turning malignant.

Bremstrahlung radiation: radiation produced from the deceleration of electrons, or any charged particles.

Dysplasia: an abnormality of development; in pathology an alteration in size, shape and organisation of adult cells.

Esophagoscopy (Endoscopy): procedure involving a camera being guided down a patients esophagus to observe the surface of the esophagus.

Esophagram: an x-ray image taken of the esophagus following a barium meal consumed by the patient.

Radioactive Tracer (Radiopharmaceutical): compound with a radioactive isotope attached, the compound usually has some biological property that attaches to an organ within the body,

Squamous cell carcinoma: carcinoma derived from the lining of the inner mucosa layer containing the squamous cells.

Abbreviations

BS	Beam Splitter
CCD	Charge Coupled Device
CT	Computed Tomography
CW	Continuos Wave
DC	Directional Coupler
FT	Fourier Transform
FWHM	Full Width at Half Maximum
IML	Index matching Liquid
KLM	Kerr-Lens Mode locked
LCI	Low Coherence Interferometry
MRI	Magnetic Resonance Imaging
NEP	Noise Equivalent Power
NIR	Near Infrared
OCDR	Optical Coherence Domain Reflectometry
OCT	Optical Coherence Tomography
OLCR	Optical Low Coherence reflectometry
OTDR	Optical Time Domain Reflectometry
PET	Positron Emission Tomography
PIF	Polarisation induced fading
PS-OCT	Polarisation Sensitive – Optical Coherence Tomography
PZT	Piezoelectric
QWP	Quarter Wave plate
RPM	Revolutions per minute
SLD	Superluminescence Dide
SNR	Signal to Noise ratio
SOP	State of polarisation
SPECT	Single Positron Emission Computed Tomography

Principle Symbols

k_0	Wave number
δ	Birefringence
λ	Wavelength (nm)
λ_0	Mean wavelength (nm)
ϕ	Phase (rads)
η	Angle between two states of Polarisation (rads)
P_0	Initial Power (mW)
P_r	Power reflected from the reference arm (mW)
P_s	Power reflected from the sample arm coherent with the reference arm (mW)
P_x	Incoherent Power reflected from the sample arm
ρ	Detector responsivity
L_c	Coherence length (nm)
I_{dc}	DC component of the photocurrent
I_s	AC component of the photocurrent
$\Psi(x)$	Wavefunction
$A(x)$	Autocorrelation function
σ_{Tot}	Total noise
σ_{Ex}	Excess noise
σ_{sh}	Shot noise
σ_{Rec}	Receiver noise
σ_{Be}	Beat noise
q	Electronic charge ($1.6 \times 10^{-19} J$)
B	Bandwidth (MHz)
V	Degree of polarisation
c	Speed of light (3×10^8)
Z_R	Rayleigh Length (m)

W	Beam width (m)
f	Focal length
R_r	Reference arm reflectivity
R_s	Sample arm reflectivity coherent with the reference beam
R_x	Sample arm reflectivity incoherent with the reference beam
ΔL	Change in path length between the sample and reference arm in the processing Michelson interferometer
ΔX	Change in path length between the sample and reference arm in the receiving Fizeau interferometer
T_c	Circulator Transmission factor

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Chapter 1

Introduction

1.1 Overview of Medical Imaging and Techniques

Medical imaging is essential for diagnosing illnesses as it provides anatomical and physiological information about the patient and the interpretation of this information is used to prescribe corrective treatment.

There are a number of imaging techniques routinely used which can be used to provide this information, each technique has its own use for the specific diagnosis at hand whether it is static or functional imaging which is required or soft tissue differentiation or bone which is being investigated. Optical Coherence Tomography (OCT) is a high resolution imaging technique, which allows depth images of tissue to be obtained. This imaging technique is

the focus of this project and all various imaging techniques are outlined below discussing their uses and limitations in the medical field and their comparisons to OCT imaging.

1.1.1 X-ray (Transmission)

X-rays are used to produce static images showing different tissue compositions of a patient. X-rays are produced when electrons are accelerated towards a tungsten target and the Brehmstrahlung radiation emitted from the decelerating charge is emitted and used in imaging. These x-rays are directed at the region of interest on the patient and an x-ray film is placed behind the patient to detect the transmitted x-rays.

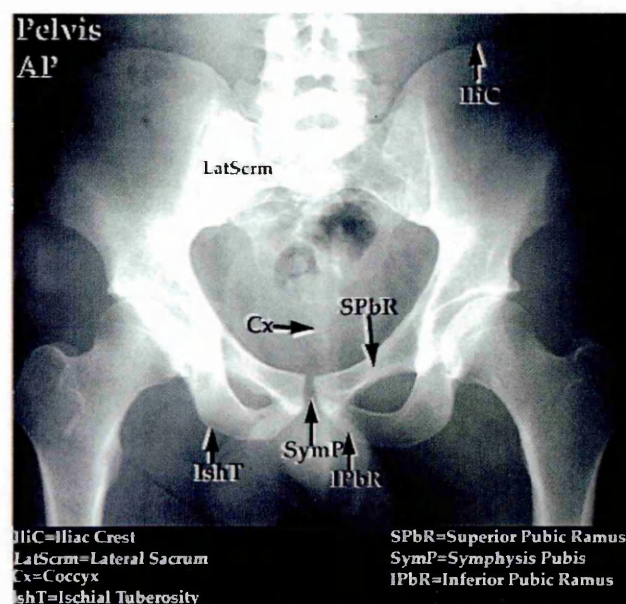


Fig.1.1. X-ray image showing the pelvic region, high density areas such as bone are shown as bright regions in the scan¹

X-rays are deflected and absorbed to different degrees by various tissues and bones in the patient's body. The amount of absorption depends on the tissue composition, for example,

dense bone will absorb more x-rays than softer tissues, such as muscle, fat and blood. The amount of absorption depends on the density of electrons in the tissues. Tissues with high electron densities cause more x-ray scattering than those of lower density. Thus, since less photons reach the x-ray film after interacting with bone or metal rather than tissue, the x-ray film will look brighter.

Because x-rays are usually transmitted through the whole body and condensed on a two dimensional film it is difficult to distinguish anatomical features unless the density is very different, therefore it is a useful technique for imaging bones and dense matter rather than soft tissue as its density is higher. An example of a pelvic x-ray image can be seen in Fig.1.1 showing high-density regions such as bone, as bright regions.

X-rays are quick and easy to take as it just involves switching an x-ray beam on and off, however the information obtained is very limited and not ideal for in depth diagnoses.

1.1.2 Computed Tomography (CT)

Computed Tomography (CT) provides a means of producing transverse slice images of patients. X-rays are used but as a thin 'pencil' beam that rotates on a special gantry around the patient.

Once passed through the patient the transmitted beam is monitored by an array of detectors and the data is processed and reconstructed to produce a Grey-scale image. The spatial resolution of CT is very high clinically of the order of a millimetre and image enhancement is possible, as the information is stored digitally, therefore clear differentiation between soft tissue structures can be achieved². An example of an abdomen CT scan is shown in Fig.1.2

One disadvantage of X-ray and CT imaging is it uses ionising radiation as its source and therefore the patient receives a radiation dose, this dose is higher for CT imaging than x-

ray. Image resolution is dependant on the x-ray tube current and hence patient dose so there has to be a compromise met between image quality and acceptable dose to patients. A typical axial scan takes just 15 seconds to perform and usually up to 30-40 slices are obtained of a region about 5mm apart. Difficulty in imaging the thorax region arises due to lung movement as the patient breathes but fortunately the scans are short enough for patients to hold their breath while each slice is acquired.

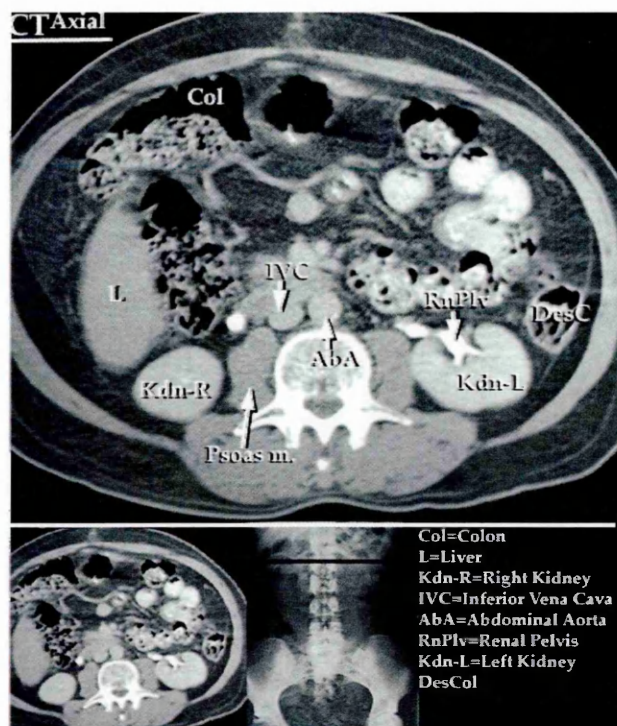


Fig.1.2 CT axial scan of the abdomen showing all internal structures within a single slice¹.

CT imaging has a lot of advantages in medicine and has advanced significantly in the last twenty years. Any part of a patient can be imaged and the internal structures observed. CT imaging is also an essential tool in radiotherapy while planning radiation treatment, the Grey scale image is a measure of the electron density of the tissue it passes through and this is an important parameter for predicting the radiation dose to patients on radiotherapy.

1.1.3 Magnetic Resonance Imaging (MRI)

MRI is an application of Nuclear Magnetic Resonance (NMR) which monitors the distribution of hydrogen atoms within the patient's tissue³.

The spinning protons within hydrogen atoms have magnetic properties and align with an external magnetic field usually 1.5T created by a powerful electromagnet surrounding the patient. Tissue is exposed to short pulses of radio frequency (RF) power which disturbs the alignment of the protons, as the protons re-align they emit weak RF signals which are detected by receiver coils. The strength of the emitted signals determines the proton density. The time the proton takes to re-align is dependant on the bonding within each type of tissue and these recovery times are used to generate a Grey-scale image that can be reconstructed in any plane, showing great soft tissue differentiation.

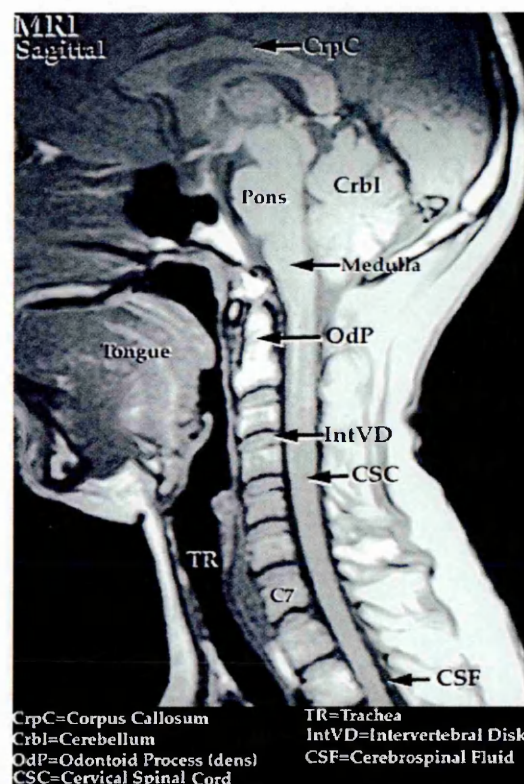


Fig.1.3 Sagittal MRI scan of the head and neck region showing excellent soft tissue differentiation¹.

Scanning times of MRI are longer than CT, usually about 30 seconds per axial slice and patients need at least a 45 minute slot for a typical scan. The cost of running the magnet is high, there are no radiation risks to patients, however the effect of RF signals on the body is still being investigated. The resolution of the MRI scan is dependant on the external magnetic field within the scanner, clinically the resolution is typically of the order of a millimetre, however higher external fields used in research (up to 7T) can achieve resolutions of the order of micrometers. An example of a MRI scan is shown in Fig.1.3.

MRI imaging is very useful for imaging head and neck regions and the brain as there are a lot of critical organs with similar tissue types and with the images being stored digitally image enhancement allows a high definition image to be obtained. It is also an essential imaging technique for identifying tumour volumes as the tumour can be distinguished, more easily than other techniques, from the organ/tissue it is invading.

1.1.4 Ultrasound

High frequency sound waves are transmitted into the patient and the echoes returning from various tissue boundaries are detected². The time interval between transmission and detection of each pulse is measured and since the speed of sound is known, the depth of each tissue interface can be calculated and an image produced. Ultrasound uses non-ionising radiation and does therefore not harm the patient. Scans can be taken in real-time as soon as the transducer is placed onto the patient's surface. The resolution is not as high as other techniques and is the order of centimetres, however it is a quick and safe method for diagnosing certain conditions immediately. An example of an ultrasound scan of the pelvic region is shown in Fig.1.4.

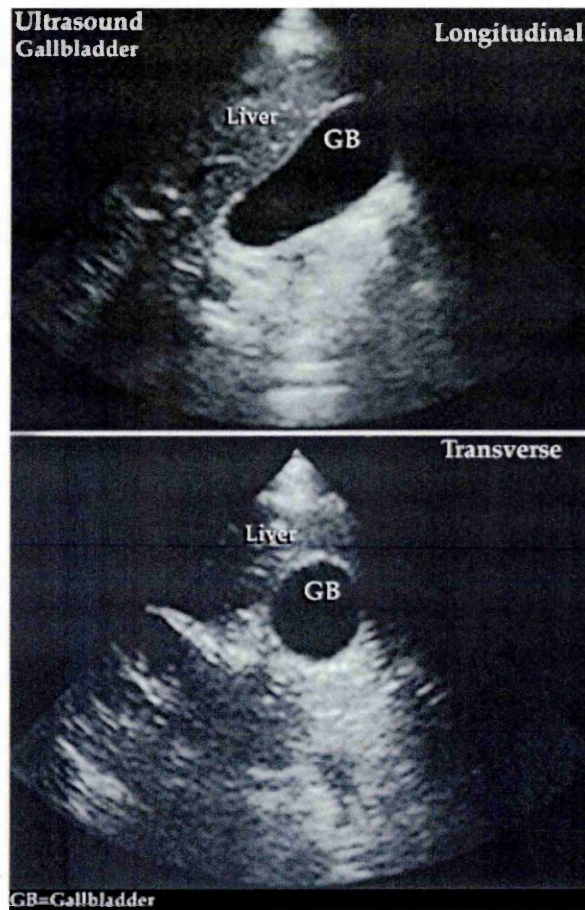


Fig.1.4 Ultrasound scan of the pelvic region showing the gallbladder, both longitudinal and transverse views are shown¹.

Ultrasound can be used for prenatal scanning to make sure the unborn child is in a healthy condition without harming the baby or the parent. It is also a useful technique to detect large objects within a patient such as gallstones or kidney stones. An Ultrasound scan is also a quick way to determine whether a detected lump is a cyst or needs further investigation. There are four types of ultrasound images used:

A-scan (amplitude mode): A one-dimensional scan showing a graphic display of the distances between tissue interfaces.

B-scan (brightness mode): A two-dimensional gray-scale image representing a plane or section through the body, spatial information is determined as well as information relating to the different tissue types.

M-mode (motion mode): Shows information about how the positions of tissue interfaces vary with time. Each interface produces a trace on a screen, stationary interfaces show straight lines whereas moving interfaces show wavy lines.

Doppler Ultrasound: Ultrasound waves reflected from a moving interface suffer a Doppler shift, if the interface is approaching the wave the frequency will be increased and if its moving away the frequency will decrease. Colour maps showing movement towards and away from the transducer are superimposed on B-scans usually representing blood flow.

1.1.5 Single Photon Emission Computerised Tomography (SPECT)

Diagnostic images are produced following the introduction of a radioactive labelled pharmaceutical into the patient, usually Technetium 99m (Tc^{99}). The labelled pharmaceutical attaches to the organ or part of the body it is targeted for and emits gamma rays that can be detected using a gamma camera (a large scintillation crystal, monitored by a larger array of photomultiplier detectors to measure gamma ray emission)². An image can be reconstructed by detecting the light produced by a gamma photon and detecting its origin. The resolution of SPECT images is relatively low, $>1\text{cm}$ due to the difficulty associated with collecting unscattered gamma rays emitted from the source, but due to effective labeling ability information about almost any organ can be obtained that cannot be obtained from any other imaging method. SPECT imaging is also a valuable technique for functional imaging as it is possible to follow the radioactive tracer as it travels through the patient's body.

The scanning time depends on the study required, whole body imaging may take over an hour for a scan whereas a renal study may take 15-30 minutes as it is concentrating on a smaller region. The resolution of a SPECT image is dependant on the time taken over the scan, the longer the scan the more radiation emitted and detected which makes the location of the radioactive pharmaceutical more accurate to predict. However a compromise has to be met over resolution and how long it is reasonable to expect the patient to lie relatively still.

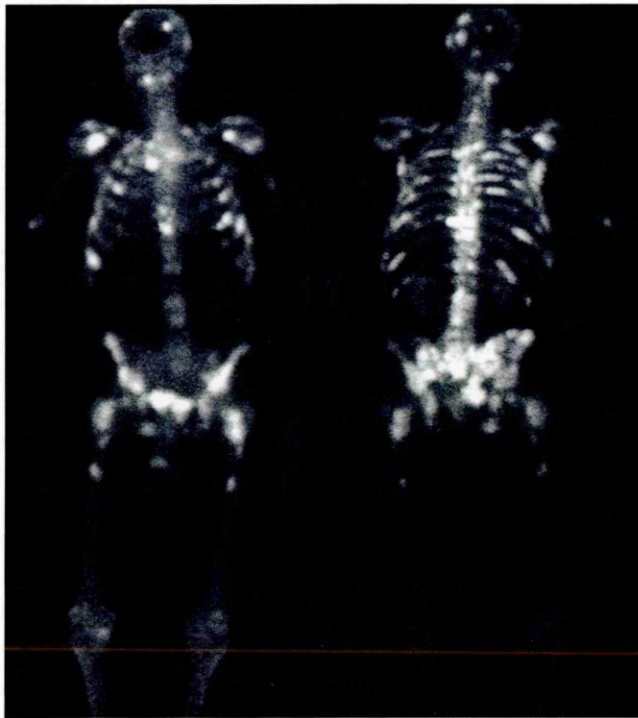


Fig.1.4 Whole body SPECT scan of a patient with widespread metastatic prostate cancer¹.

The radiation dose to patients is relatively low compared to radiographic imaging as the radioactive nucleide is chosen to give minimum dose.

Fig.1.5 shows a whole body SPECT scan, the 'hot' bright regions show areas of uptake of the radio-pharmaceutical, in this case it identifies areas of metabolism and the bright areas show wide spread metastatic tumours all over the body originating from a prostate carcinoma.

1.1.6 Positron Emission Tomography (PET)

A technique very similar to SPECT but uses isotopes that are proton rich² (i.e. too many protons to be stable). These include fluorine 18 (most common isotope used having a half-life of ~110min), oxygen-15 (12mins), Carbon-11 (20mins), Nitrogen-13 (9.9min). A proton in the nucleus decays to produce a neutron, a positron (anti-particle of an electron) and a neutrino. The positron interacts with an electron in the patient and they annihilate due to the fact they are antimatter and matter. This turns their mass into energy and two photons are produced at 180 degrees to each other and an energy of 511keV (corresponding to the electron and positron rest mass). The gamma ray emitted at 180 degrees can be detected accurately which increases the resolution of the images.

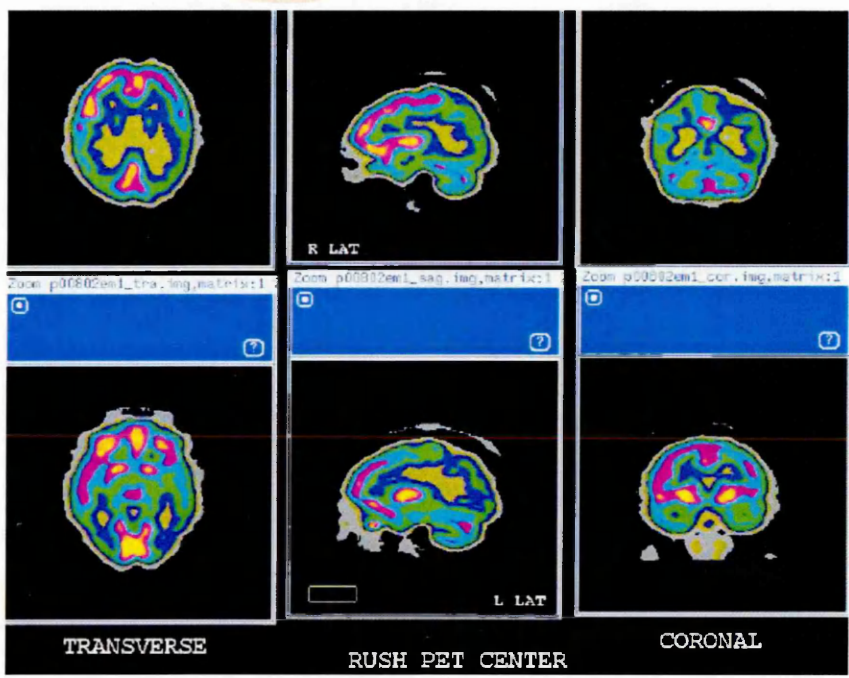


Fig.1.6 PET scans of the brain of a patient with Alzheimer's dementia. Severely decreased metabolism in bilateral cerebral hemispheres, compared to thalamus. More severe in temporal and parietal lobes¹.

Because of the short half-lives of the sources used in PET most hospitals require a cyclotron to produce the sources on site.

PET scans are very useful in determining metabolic activity and have an active roll in functional imaging rather than high resolution static imaging.

1.1.7 Endoscopy

Endoscopy uses a flexible camera that shows images from various cavities within the body². Endoscopes have bundles of optical fibers that either direct light into the body or transmit an image back from the body to the physician. The procedure is usual uncomfortable for the patient but is not harmful as light is used for the imaging source. Only information about the tissue surface can be obtained using Endoscopy. An image from a gastrointestinal endoscopy is shown in Fig.1.7 showing a typical cancer in the esophagus.

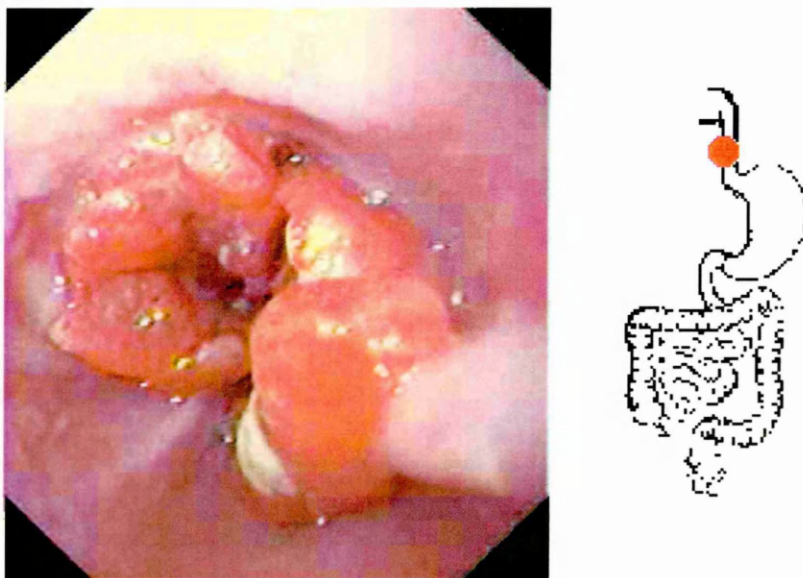


Fig.1.7 Endoscopic image of the esophagus showing a massive obstructive tumour¹.

Endoscopic images show photographic images very clearly but only of the surface of the structure and contains no depth information. Images can be acquired in real time and is an essential tool in surgery where organs and tissue can be magnified on a larger screen.

1.1.8 Optical Coherence Tomography

Optical Coherence Tomography (OCT) is a relatively new imaging technique which uses light, backscattered from tissue, to create cross-sectional images of biological structures³. The two-dimensional images offer extremely high resolution scans, of the order of micrometers, and can have a high Signal-Noise Ratio (SNR) above 100dB. OCT is therefore an attractive technique for studying changes in tissue structure *in vitro* and *in vivo*. OCT uses light as an imaging source, which makes an appealing alternative to other, more hazardous, imaging modalities.

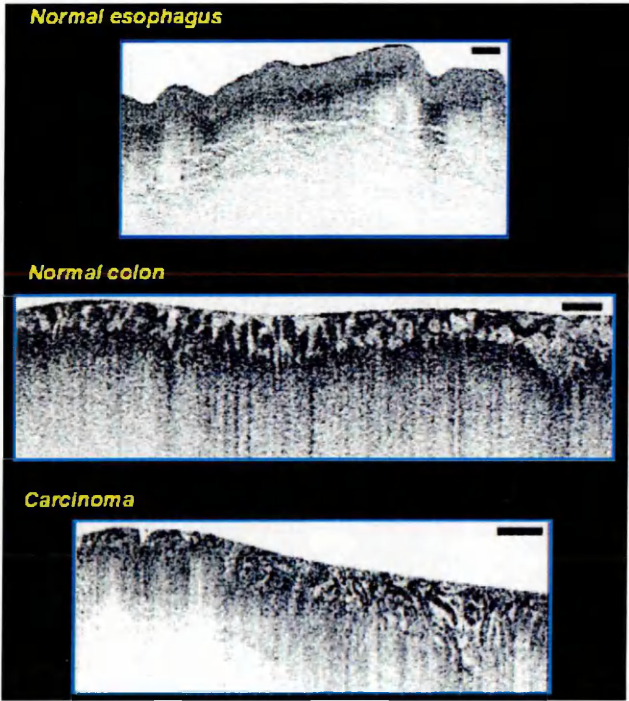


Fig.1.8 OCT images of the epithelial and submucosa layers within a normal esophagus, normal colon and cancerous colon¹.

Fig.1.9 Table showing summary of all imaging modalities.

Imaging Technique	Resolution	Penetration depth	Scanning time	Comments
X-Ray	~mm's	Whole body	<5 seconds per image	Information superimposed so image not clear. Ionising radiation used as imaging source.
CT	<1mm	Whole body	10 seconds per slice. ~30mins per scan	Ionising radiation used as imaging source.
MRI	<1mm clinically ~1 μ m research	Whole body	15-20 seconds per slice. ~45mins per scan	Very good soft tissue differentiation. Long scanning times.
Ultrasound	~0.5-1.0cm	Whole body	Real time	Low resolution. Safe scanning method. Quick and easy scanning.
SPECT	~0.5-1.0cm	Whole body	Depends on resolution and image site. ~hour for whole body imaging.	Functional imaging. Low resolution. Ionising radiation as imaging source.
PET	~0.5-1.0cm 1mm research	Whole body	Depends on resolution and image site. ~hour for whole body imaging.	Functional imaging. Low resolution. Ionising radiation as imaging source.
Endoscopy	Photographic <1mm	Surface	Real time	High resolution. Surface information only.
OCT	<1 μ m	1-2mm	Real time	Very high resolution Depth imaging

Most OCT systems have been used to image the eye as the tissue is transparent and allows more light to be reflected back into the system. Although the depth range of OCT only covers 1-2mm this is an important technique to diagnose microscopic changes such as pre-cancerous cell changes. The images in Fig1.8. show how OCT can be used to look at tissue

layers within organs. Depth images of a normal esophagus is showing giving good definition between tissue layers (epithelial and submucosa) the same can be seen with the image of the normal colon. The image showing carcinoma of the colon shows a breakdown in tissue layer definition due to the cancerous invading cells. Identifying small changes at this stage dramatically increases the patient's chance of survival.

Fig.1.9 summarises all imaging techniques mentioned within this chapter with advantages and limitations of each technique.

1.2 Optical Coherence Tomography

OCT uses Low Coherence Interferometry (LCI)¹ to generate high resolution two-dimensional images³. The most common set-up for an OCT system is the optical-fibre based Michelson Interferometer shown in Fig.1.10. Here light from a low coherent source is split 50/50 by a directional coupler, 50% of the light intensity travels down a reference arm and is reflected back through the fibre from a reference mirror. The other 50% travels down a sample arm and is focussed onto the tissue sample under investigation.

At boundaries of different refractive indices, as with microstructures within the tissue, the light is either scattered or transmitted, the intensity of which is determined by the difference in the refractive index interfaces. The light that is backscattered is coupled back down the fibre and mixes with the light returning from the reference mirror. Only when the path length difference is equal to or within the coherence length of the source interference signals can be detected at the coupler exit arm. The interference signal detected is proportional to the difference in the refractive index interfaces within the tissue.

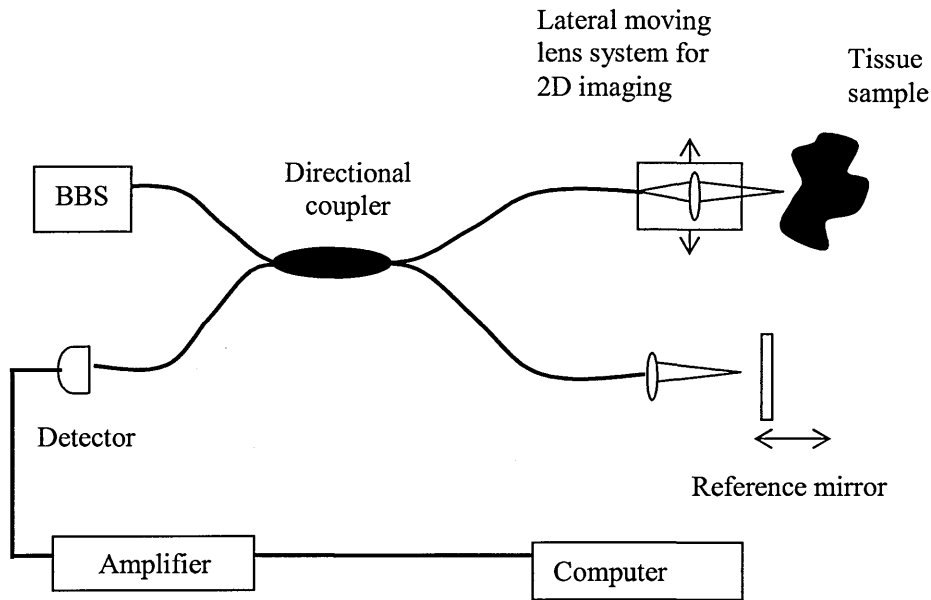


Fig.1.10. A Michelson Interferometer set up for an OCT system, (BBS=Broad band source).

To detect the signals from all the scattering sites within the tissue, the reference mirror is scanned longitudinally in order to change the path length difference between the interferometer arms and hence build up a one-dimensional profile of refractive index interfaces. A typical one-dimensional scan taken through tissue is shown in Fig.1.11.

To obtain two-dimensional images, a number of longitudinal profiles are collected laterally, across the region of interest and compiled together.

The longitudinal resolution of an OCT image is essentially determined by the coherence length of the source, the shorter the coherence length the higher the resolution. The lateral resolution however, is determined by the beam width on the tissue sample. An example of a high-resolution OCT image can be seen in Fig.1.12⁵. The image shows cellular structure of an African frog. The resolution is $1.9\mu\text{m}$ and it is clear to see individual cells and their nuclei.

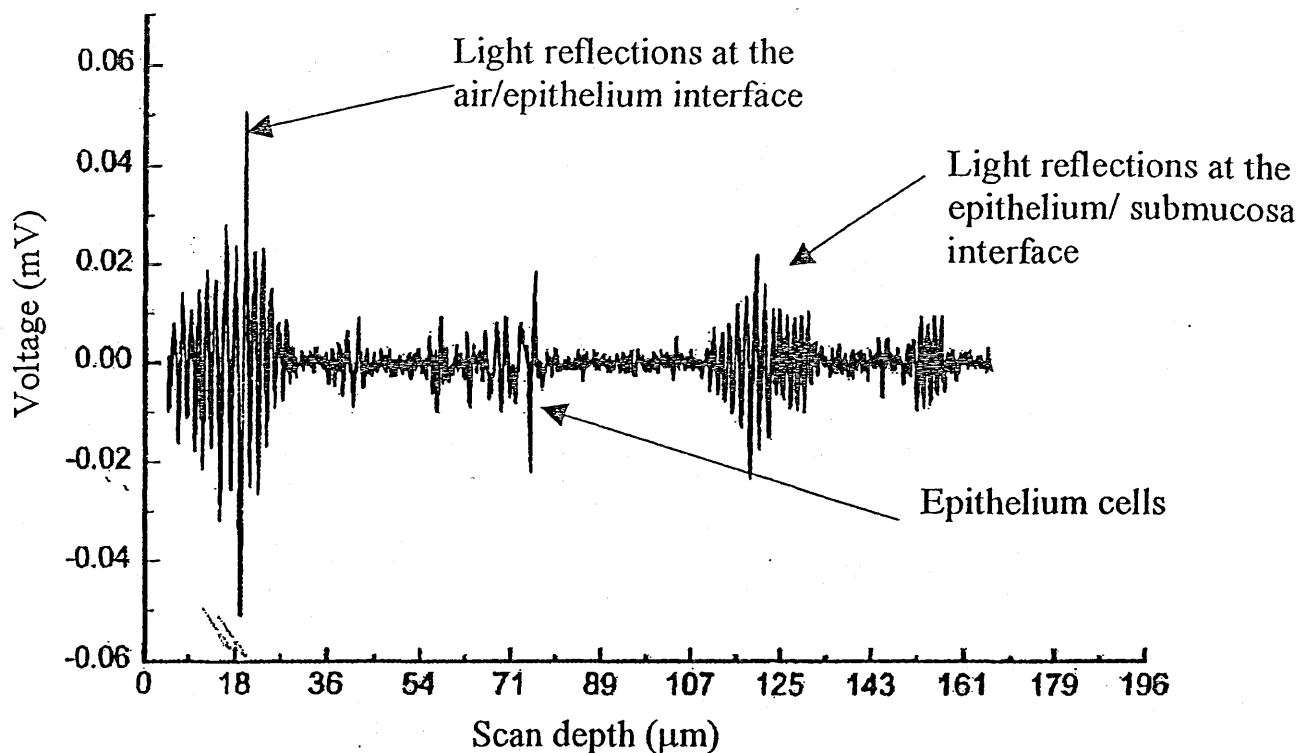


Fig.1.11 Showing a one-dimensional profile of all the interference signals from the various reflecting sites within a tissue sample⁴.

This project focuses on imaging esophagus tissue in order to improve diagnosis of pre-cancerous tissue. OCT is the perfect imaging technique to detect any early cellular changes in tissue, in particular early cancerous changes in esophagus tissue, where it is important to measure thickness changes within the order of micrometers.

The Michelson interferometer is the most popular choice for an OCT system, however it may not be the most suitable for OCT endoscopic imaging which will be required when imaging esophagus tissue, this will be discussed in greater detail throughout this thesis. The aim of this project is to find an alternative interferometer configuration for endoscopic OCT imaging of the esophagus.



Fig.1.12 OCT image of an African Frog showing cellular structure⁵.

1.3 Diagnosing Cancer of the Esophagus

There is evidence to suggest that the number of cases of esophageal cancer is increasing at an alarming rate in Western countries⁶. Cancer of the esophagus accounts for 5% of all gastrointestinal (GI) tumours and is the 7th most common cancer world-wide.

The cell structure showing the different layers within esophagus tissue is shown in Fig.1.13. Cancer of the esophagus falls into two main categories, squamous cell carcinoma and adenocarcinoma⁷, which have approximately an equal number of cases annually. Squamous cell carcinoma affects the upper two-thirds of the esophagus, within the inner mucosa lining containing the squamous cells. Adenocarcinoma occurs in the lower third of

the esophagus within a different type of mucosa containing columnar cells and it is this type of cancer that is on a steady increase.

- E -stratified squamous epithelium
- L -lamina propria
- MM -muscularis mucosae
- S -submucosa
- CM -circular layer of muscularis
- LM -longitudinal layer of muscularis
- A -adventitia

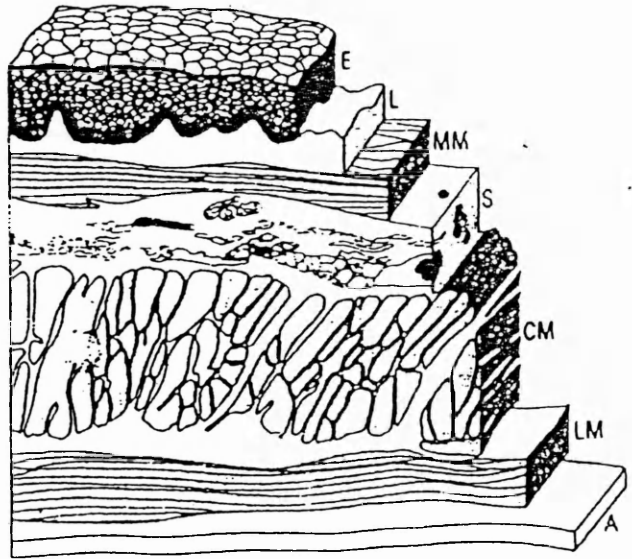


Fig.1.13 Shows the cell structure of esophagus tissue⁸

The causes of cancer are not known, but it is suggested that it is linked to long term irritation; risk factors being regular drinking and smoking. Esophageal cancer is more common in men than women and people over 60 years old have a high risk of developing the disease.

Barrett's esophagus¹⁰, shown in Fig.1.14 is a condition brought on by long term irritation of acidic reflux from the stomach, this may be caused by certain foods or drinks. This acidic environment causes changes within the esophagus cells converting them to a similar cell structure as the cells found in the stomach lining. People with Barrett's esophagus are prone (10-20%) to developing Adenocarcinoma.

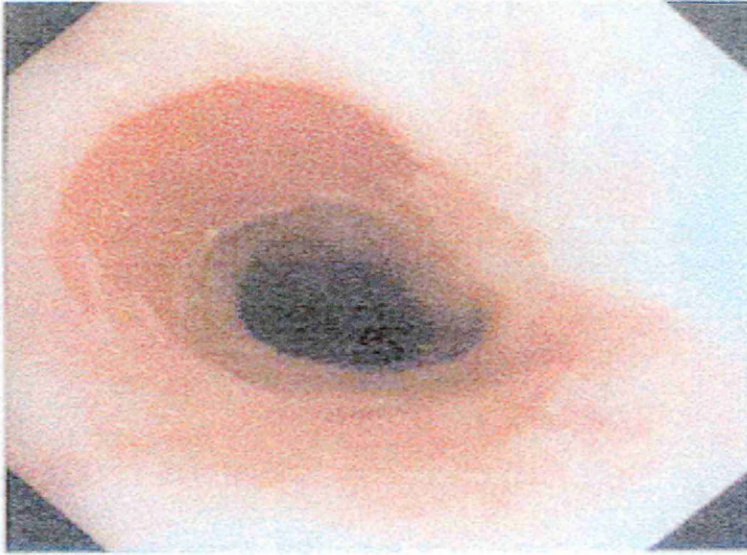


Fig.1.14 Endoscopic view of Barrett's esophagus¹⁰.

Both major types of esophageal cancer have poor prognosis (5% survival over 5 years) when detected in its late stages⁶. However, if detected in the early stages when it is still confined to the mucosa and submucosa layers (Fig1.13), the chances of survival after 5 years increase to 90%. It is therefore vital that the cancer be diagnosed as early as possible.

It is very difficult to predict who will develop esophageal cancer, and it is almost impossible to diagnose in the early stages, as the symptoms are not obvious. Patients with Barrett's esophagus or patients with high risk factors or showing symptoms of cancer such as weight loss, painful swallowing, vomiting, coughing up blood and a medical history of cancer may be tested/screened for the disease every 18 months.

The diagnosis for esophageal cancer usually involves an initial barium x-ray of the esophagus (esophagram) and then an esophagoscopy (endoscopy). During the endoscopy examination, an endoscope is lowered down into the esophagus to view the tissue lining. If

abnormal areas are discovered, cells are collected via the probes in the endoscope and taken for pathological tests, to determine whether they are malignant. If abnormal areas cannot be detected then random biopsies are taken. This procedure involves taking cell samples every 1cm down the esophagus to identify any malignancies or changes in cell structure. Obviously this technique is unreliable, as it is impossible to cover the whole esophageal area and this results in a high percentage of false negative diagnoses using this examination⁸.

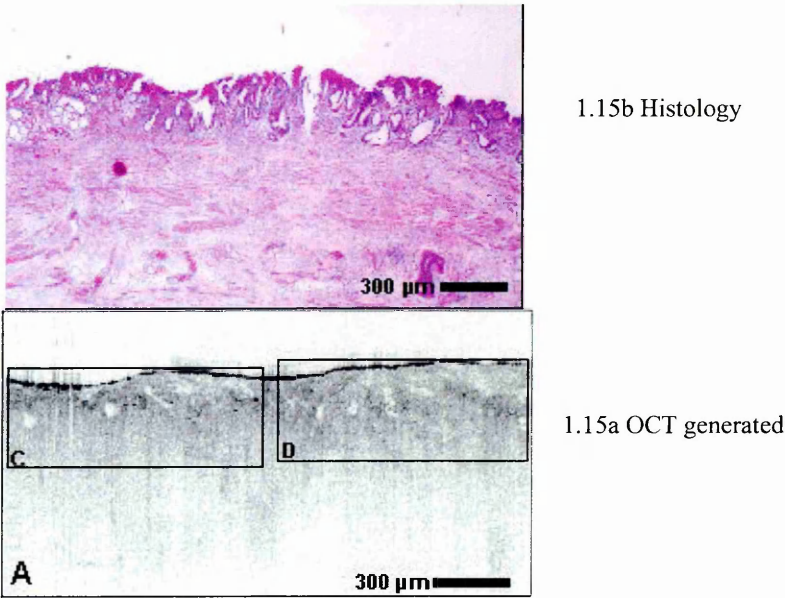


Fig.1.15 Images of Barrets esophagus tissue, top is OCT generated and bottom histology. Barrett's esophagus is a condition caused by excessive and chronic heartburn or acid reflux from the stomach. The cells lining the esophagus undergo changes that lead to a pre-cancerous condition. The use of OCT may help identify patients that are at risk for cancer and help guide physicians to biopsy abnormal-appearing regions of tissue.

By producing depth images, OCT can actually improve this method by identifying areas within the esophagus where the tissue is thickest. It is known that cancerous tissue is thicker than normal tissue and this can be detected using OCT so the appropriate sites for biopsy can be predicted. Alternatively, if the OCT system used for diagnosis has an extremely high resolution, then cell size structures, from underlying tissue, can be clearly seen on the depth images. Any abnormalities from the scans can be identified immediately and the need for biopsy is eliminated, this is demonstrated in Fig.1.15.

1.4 Scope of Thesis

The aim of this project is to analyse and improve the signal of OCT systems suitable for the endoscopic applications which may be used to diagnose pre-cancerous esophagus tissue, when the cell lining starts to thicken.

Presently, most OCT systems use a single-fibre based Michelson interferometer set-up in which lateral scanning is performed by mechanically moving the sample arm across the tissue. This method has proved to be successful for *in vitro* imaging, although it may not be the best configuration for endoscopic applications. During an endoscopy the interferometer sample arm is immersed down the esophagus. This change in environment may cause phase changes between the two interferometer arms and changes in polarisation of the tissue signal that will affect the visibility of the interference signal. It is not clear from the publications, whether other research groups have compensated for this effect by using polarisation-maintaining fibres or other techniques, but it is an important issue.

This project investigates how the polarisation effects within an OCT system (both within the fibre and within tissue) are likely to effect the overall SNR out determining whether these cause significant losses in the signal and how to overcome these losses.

This project aims to overcome or improve the polarisation fading effect in fibres within conventional Michelson fibre optic interferometers by using a Fizeau interferometer as the OCT system (Chapter 3). In this configuration the light in the sample and reference arm travel down the same fibre and so experience the same environmental changes. Therefore upon recombining, as long as there is no change in polarisation as the light travels through the tissue, the polarisation states will be the same and give maximum signal. The Fizeau interferometer as it stands would give a lower Signal to Noise Ratio (SNR) than the Michelson interferometer and therefore perform less effectively. This project will therefore investigate ways to increase the performance of the Fizeau interferometer by looking at different interferometer configurations containing power-conserving components. The Signal to Noise Ratio (SNR) will be calculated for a number of designs and the design shown to be as effective as the Michelson interferometer will be used as the final.

Once the optimal design has been established, it will be set up experimentally using equipment optimised for power conservation. The SNR will be measured experimentally in order to verify the theoretical results for the optimal design.

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Chapter 2

HISTORICAL OVERVIEW

2.1 First Applications of OCT

Optical Coherence Tomography (OCT) has developed from a technique known as Optical Time Domain Reflectometry (OTDR), first developed for imaging of biological tissue in 1986¹. OTDR uses time of flight information from ultra-short laser pulses to determine the location of the reflecting sites. This method is very similar to ultrasound imaging, using light pulses rather than sound. The resolution using this technique was of the order of centimetres, which is incompatible for imaging the microstructures within tissue.

In 1987 an alternative approach was developed by imaging in the frequency domain rather than the time domain² and this achieved millimetre resolution.

With improving technology and the new development of optical components such as broader band sources the technique Optical Coherence Domain Reflectometry (OCDR)³ followed. This technique achieved resolutions of the order of micrometers and a high dynamic range above 100dB. This technique, instead of using pulsed sources and recording the time of flight information, made use of the low coherence of the newly developed, broad band sources. Continuous wave broad band sources were used to detect the location of the scattering microstructure sites by interference with a reference beam. In 1988² OCDR measurements were performed by Fercher *et al* obtaining one-dimensional profiles of the reflecting site within eye tissue. Taking a series of adjacent, longitudinal profiles and combining them in order to create a tomogram produces two-dimensional images of OCDR. This technique is commonly known as Optical Coherence Tomography (OCT). OCT was first demonstrated by Huang *et al* in 1991⁴ using a Michelson Interferometer set-up to obtain two-dimensional, micron-resolution images of the human eye showing detailed structure of the retina, optical disk and images of the coronary artery. Outstanding comparisons were published between the OCT images and the histology specimens under examination⁵ shown in Fig.2.1.

Most early work using OCT was carried out using optical tissue, primarily the eye, for OCT examinations. This was a prime example for presenting the exceptional performance of OCT, since the transparent tissue exhibited little scattering properties, which would degrade the images and prevent measurements to greater depths.

Since the first reports of OCT, the number of medical applications has increased dramatically and OCT can now be used for most tissue types and not just the eye. Improvements in image performance have been made to increase resolution and signal to noise, and faster scanning techniques have also been developed, the progress of which is outlined below, along with other areas of OCT applications.

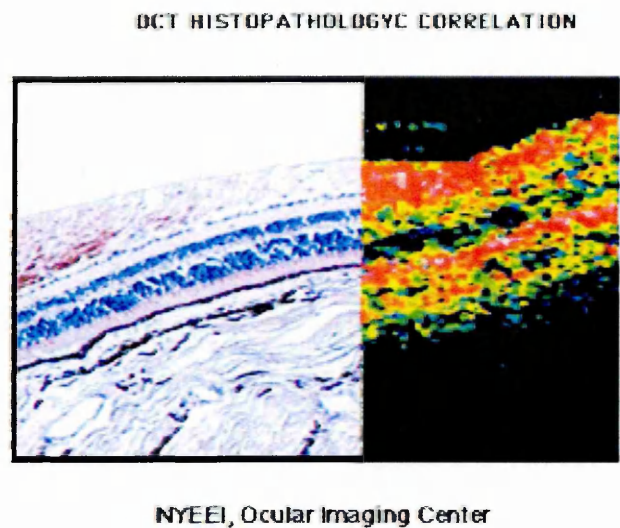


Fig.2.1 showing OCT and histology images of the eye⁵.

2.2 Medical Applications of OCT

As previously stated the first images of OCT⁴ were taken of the retina and the coronary artery. The retina consists of transparent layered tissue with well-defined morphology. The coronary artery consists of highly scattering microstructures. Accurate and high-resolution images were presented of the human retina and optical disk, corresponding very well to the histology samples to depths of 1-2mm. The coronary artery images were of a lower quality with less definition due to the high scattering properties of the tissue. The rapid attenuation achieved depths of only 150-200µm before the signal falls to noise level, both images were scanned using a 830nm SLD.

Due to the success of optical tissue imaging using OCT, most of the early studies used the eye and retina for OCT studies and applications⁶ and OCT offered a ‘safe’ and accurate means of examining the tissue, since light is used as the imaging source. In 1995^{7,8} an OCT system was especially designed for clinical, ocular examinations to be routinely performed.

After optical imaging was well established, emphasis fell on using OCT to image more turbid/scattering tissue such as blood vessels and tissue surface. In 1996⁹ imaging of the coronary artery was re-investigated using a Near infrared (NIR) 1300nm light source to achieve better depth resolution as there is less absorption than the visible region and the probability for scattering is lower. Well-defined high-resolution images of 20µm were produced to a depth of 1mm.

Further interest arose in using OCT to image the respiratory and gastrointestinal (GI) tract when it was suggested that early stage cancer can be identified by observing changes in the squamous cell structure close to the tissue surface. The possibility of using OCT as an alternative for biopsy was also considered. In 1998¹⁰ high resolution, cross-sectional images of the GI tract were presented to investigate the feasibility of endoscopic OCT. Results showed clear definition of the mucosa and submucosa layers within the tissue. A 1270nm SLD was used to obtain 19.2µm resolution longitudinal images to a depth of 1.2mm – blood vessels (0.5mm in diameter), crypts (60-70µm) and other microstructures were clearly identified. During this period, work was being carried out to use OCT to image the skin¹¹. An SLD, 830nm, was used to give 15µm resolution. Accurate images of the superficial layers within the skin were presented. Patients with certain skin conditions volunteered to be scanned and examples displayed showing how OCT can detect eczema tissue, blisters, malignant melanoma and even scabies mites.

With increasing interest in the applications of OCT, Colston *et al*¹² in 1998 were able to image the hard and soft tissue structure of the oral cavity. OCT was now progressing into dentistry. Subsurface imaging is important in this area to identify tooth fractures, demineralised enamel and other indications of tooth decay. A 1310nm SLD was used to image premolar, 5-6 month old pigs teeth using a conventional OCT scanner. The images showed definition of the oral mucosa, the enamel-dentin interface and the tooth-mucosa interface down to a depth of 1mm.

In 1998 Boppart *et al*¹³ assessed the feasibility to identify intra-cortical melanomas where other, lower resolution imaging modalities had been unsuccessful. Images of a specimen from the outer human cerebral cortex were taken which contained an intra-cortical melanoma located near the surface of the cortex. The images showed a definite increase in backscatter within the tumour tissue and even smaller lesions (<500µm) were identified. All applications are summarised in Table 2.1.

Table.2.1 Showing summary of OCT applications in medicine.

Application	Year of application	Centre wavelength of source /nm	Axial resolution /µm	Imaging depth
retina ⁽¹⁾	1991	830	17	1-2mm
coronary artery ⁽¹⁾	1991	830	17	150-200µm
blood vessels ⁽¹⁴⁾	1996	1300	20	1mm
gastrointestinal tissue ⁽¹⁵⁾	1996	1270	19.2	1-2mm
skin ⁽¹⁶⁾	1998	830	15	1mm
dental studies ⁽¹⁷⁾	1998	1310	20	1mm
neurology ⁽¹⁸⁾	1998	1300	23	1.8-2mm

2.3 Improving Resolution

2.3.1 Introduction

The axial resolution of OCT images is primarily dependent on the coherence length of the light source used in the system – the shorter the coherence length, the higher the resolution. The lateral resolution is dependent on the beam diameter of the focussed light on the tissue sample. For accurate, high-resolution imaging it is necessary to use an extremely broad

bandwidth source in order to determine the location of the microstructures within the tissue and gain a higher quality image.

Early studies of OCT made use of the broad bandwidth SuperLuminescent Diodes (SLD) sources, developed in the 1980's, and the first OCT image obtained used an 830nm SLD with a 30nm bandwidth⁴. This allowed an axial resolution of 17 μ m to be achieved along with a beam diameter of 9 μ m for the lateral resolution. Since then, with the development of broader bandwidth sources, the axial resolution has improved considerably.

In 1992 Clivaz *et al*¹⁴ used the fluorescence light of a Ti:Al₂O₃ crystal as a light source to obtain 1.9 μ m resolution measurements with Optical Low Coherence Reflectometry (OLCR). The spectrum from the fluorescent light emitted (after excitation with an argon laser) showed a maximum output at 780nm with an extremely large bandwidth of 180nm. The power of this particular source is low (4.8 μ W) and inadequate for biological imaging.

In 1995 Bouma *et al*¹⁵ published a paper showing 3.7 μ m resolution OCT images of onion cells. The high resolution was achieved using a Kerr-lens mode-locked Ti:Al₂O₃ laser source, generating low coherent light with power levels above 100mW which is more than that required for biological imaging. OCT imaging through non-transparent tissue requires greater light penetration than transparent tissue. This is achieved by using a light source emitting in the near infrared (NIR) region of the spectrum, where absorption is lower than the visible region and the probability of scattering is less.

The first OCT systems were used to examine transparent tissue such as optical tissue, and depth imaging was not a major consideration. The low coherent SLD sources used, centred around 830nm¹ and even with the newly developed, high resolution Ti:Al₃O₂¹⁶ sources the wavelength stayed around 780nm, although these allowed a high power output. Now the interest in OCT imaging has spread to examining non-transparent tissue such as skin¹⁷ or the lining of the gastrointestinal tract¹⁸, and imaging to a greater depth is more relevant. Izatt *et al*¹⁹ used a 1300nm SLD with coherence length 15 μ m to look at gastrointestinal

tissue. Microstructures $<30\mu\text{m}$ were identified to a depth of $600\mu\text{m}$. It is difficult to find a suitable source for all the requirements of OCT imaging – low coherence, high power and NIR wavelength.

A description of some of the main sources used for OCT is given below along with other various techniques, which have been used to obtain suitable sources for OCT imaging.

2.3.2 Super-Luminescent Diodes (SLD's)

Solid state sources similar to laser diodes, however the reflecting ends of the cavity are cleaved in order to prevent lasing. They provide low coherent light with typical wavelengths of 800nm and 1500nm. Their bandwidths range from 20-60nm permitting resolutions of 7-25 μm .

The power output for commercially available SLD's range from a few milliwatts to 20mW amplified devices, with spectral shaping filters.

2.3.3 Kerr-Lens Mode-Locked (KLM) Lasers

Mode locking is amplitude modulation of a laser beam in the optical cavity at the laser resonator mode frequency. In this state each mode contains symmetrical side bands with interact with neighbouring modes^{20,21}. Because of the constant relationship between a mode and its individual side bands this effect produces phase and amplitude locking of the mode resulting in a series of short pulses. The KLM Ti:Al₂O₃ laser has become a reliable source for generating extremely short optical pulses as short as 5.4fs^{22,23,24}.

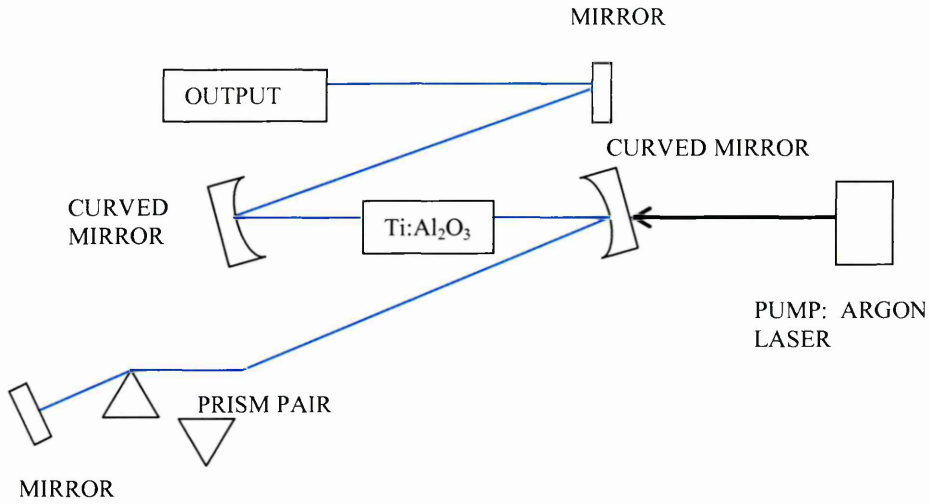


Fig.2.2. KLM Ti:Al₂O₃ laser set-up.

A schematic for a KLM Ti:Al₂O₃ laser is shown in Fig.2.2. The Ti:Al₂O₃ crystal is pumped by a continuous wave (CW) argon laser at 514nm, within a resonator. The prism pair controls dispersion within the laser cavity by allowing longer or shorter path lengths through the dispersive glass.

Powers of 400mW with a centre wavelength of 800nm and bandwidth of 145nm have been achieved allowing axial resolutions for OCT imaging to reach $1.9\mu\text{m}$ ^{20,21}. The KLM laser is very complex and bulky and expensive to produce.

Another crystal has also been investigated as a possible source for OCT imaging, the KLM Cr⁺⁴:forsterite laser^{25,26}. This gives a longer centre wavelength at 1280nm that allows deeper penetration into the tissue and less scattering. The Cr⁺⁴:forsterite oscillator is pumped by a continuous wave Nd:YAG laser. The optical output of this source has reached 300mW with a bandwidth of 75nm.

2.3.4 Frequency Tuneable Sources

Instead of using a broadband source and varying the path length of the reference arm, as is the usual procedure in OCT imaging, a narrow bandwidth source can be swept over a wide optical frequency and the reference arm reflector can stay stationary²⁷. Path length differences that occur between the sample and the reference arm reflections will cause variations of the detected beat frequency and the Fourier transform (FT) of the beat frequency spectrum gives a depth profile of the reflections within the sample.

Fig.2.3 shows a schematic for the frequency tuneable source with a compact external cavity laser using a superluminescence diode at 840nm²⁸. The wave-guide is single mode and angled with respect to the chips cleaved ends. An antireflection coating is used to suppress Fabry-Perot modes. A lens collimates the beam from the source onto the grating (1200 lines/mm) and the light is dispersed. Instead of scanning the grating for frequency sweeping a mirror is used and scanned over a range of angles and allowed a tuning capability of 25nm in 100ms. This frequency tuneable source achieved up to 35mW of power, which is a huge increase, compared to the average 2mW power from a SLD. The bandwidth of this source was equivalent to 20nm, which produced a coherence length of 38 μ m.

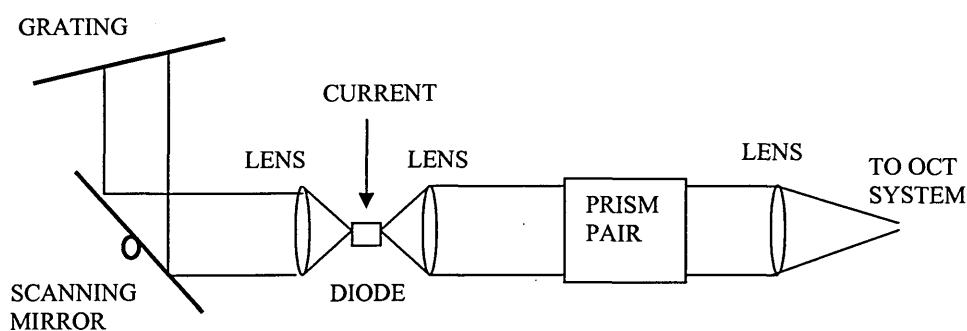


Fig.2.3. Schematic for the frequency tuneable source used for OCT systems²⁹

2.3.5 Fibre Lasers

Rare-earth doped fibres can be diode pumped resulting in fluorescence within the fibre, which is amplified by spontaneous emission along the fibre length. Broad bandwidths are achieved by suppressing lasing by cleaving the fibre ends. These sources have been used with fibres doped with: neodymium^{28,29}, praesodymium³⁰, ytterbium³¹, erbium³² and thulium³³.

Ytterbium gives light with a centre wavelength of 1080nm when pumped with a 2.1W laser diode and a bandwidth of 31nm. The average power supplied from these sources ranges between 40-60mW.

Table 2.2 Summarising sources used for OCT imaging

Source	Centre wavelength /nm	Bandwidth /nm	Power /mw	Comments
SLD	800 - 1500	20 – 60	1 – 2	
High power sld	800 - 1300	Up to 60	Up to 20	
KLM ti:al ₂ o ₃ laser ^{22,23}	800	145	400	Expensive, bulky and complex
KLM cr ⁺⁴ :forsterite laser ^{27,28}	1280	75	300	Expensive, bulky and complex
Frequency tuneable sources ²⁹	840	20	35	
Ytterbium doped fibre laser ³³	1080	31	40 – 60	Expensive
Erbium doped fibre laser ³⁴	1536	45	150 maximum 20 commercial	Expensive
Thulium doped fibre laser ³⁵	1800	80	7	Expensive

Erbium emits light at 1536nm with a bandwidth of 45nm maximum powers for these sources have reached 150mw, however commercial sources give a power output of around 20mW.

Thulium shows to be a promising source for OCT imaging emitting light with a centre wavelength of 1800nm. This wavelength is beneficial as it is above the water absorption value (1600nm) but below the value for increased absorption (2000nm). When pumped with a Ti:Al₂O₃ at 500mW and 785nm, the thulium doped fibre laser will emit 7mW with a bandwidth of 80nm.

A summary of all sources suitable for OCT imaging is shown in Table 2.2.

2.4 Scanning Speed

2.4.1 Introduction

It is important in Optical Coherence Tomography (OCT) that the images are acquired as quickly as possible in order to reduce motion artefacts that blurs images and to retain image quality which is particularly important for micrometer accuracy. It is also important for patient comfort and a faster scanning method means for the patient can be scanned relatively quickly. Increasing axial and lateral scanning speeds are both important, however more A-scans (axial scans) are acquired to improve the accuracy of the images and so the speed of these scans are the limiting factor to the overall acquisition rate. Axial scanning time is limited to the speed of the moving reference arm used to vary the path length in the interferometer. The number of methods used to change the reference arm path length, in order to decrease scanning times, are outlined in this section.

The first OCT system⁴ used a translating reference mirror to vary the path length, moving at a speed of 1.6mms⁻¹. For many years afterwards other researchers took this approach using a translating mirror or piezoelectric (PZT) fibre stretching³⁴. The maximum speed of the translation mirror³⁵ reached 160mm s⁻¹ and the PZT fibre stretching³⁶ 3m s⁻¹.

In 1997 a new method for increasing the scanning speed was presented by using a rotating glass cube to vary the path length³⁷. The cube was placed between the end of the fibre and a fixed reference mirror. The light beam is reflected internally within the cube onto an

external prism. This directs the beam back to the cube and onto the reference mirror where it is reflected back. The cube rotates at 96Hz and a longitudinal scan speed of 21.3m s^{-1} was achieved³⁸. This technique was developed with speeds of 96m s^{-1} for longitudinal scanning by placing the rotating cube in the path of the reference arm and the sample arm. In 1998³⁹ longitudinal scan speeds up to 176m s^{-1} were achieved using a rotating glass cube spinning at 427000 revolutions per minute (RPM) driven by a high speed air turbine. The rotating cube arrangement has proven to be the most successful for attaining fast scanning speeds but the set-up involves a great deal of precision and the system requires high powers to run on.

Although scanning speed is important in clinical imaging, there is a trade off with SNR and detector bandwidth, the SNR is inversely proportional to the bandwidth (B) this is discussed further in Chapter 4. Since the detection bandwidth will need to be much larger to identify all the reflecting positions, a compromise between the two parameters should be met to achieve the best system.

A description of all scanning techniques tried and tested for OCT imaging is given below and summarised in Table.2.3.

2.4.2 Translating Scanning Mirror

The first method and the most simplest method for varying the reference arm path length is using a mirror mounted onto a translation stage⁴ which can be controlled to move axially. Most translating mirrors are computer controlled so that small distances can be programmed to move over accurate distances.

Another method of varying the path length is to sweep the mirror through small angles so that the path length changes with every angle⁴⁰. These techniques have achieved maximum linear scanning speeds of 160mm s^{-1} with a 100Hz A-scan repetition rate.

2.4.3 Piezoelectric (PZT) Fibre Stretching

Most OCT systems are fibre optic based and so extending the fibre in the reference arm, by stretching it, will create a path length change. The first applications wound 10 meters of optical fibre around a PZT cylinder⁴¹. When a voltage is applied to the PZT crystal it expands in proportion to the electrical charge exerted. Although the expansion of the cylinder is small in comparison to the path length change the actual stretch over the whole ten meters of fibre results in a 3mm path length difference which is suitable for OCT scanning. This method allows linear scanning speeds of up to 3ms^{-1} with an A-scan repetition rate of up to 600Hz.

A problem that arises due to the PZT fibre stretching method is that the light suffers from birefringence effects and extra components such as Faraday rotators, are required within the system to compensate for polarisation effects, however this is only limited to polarisation effects in the reference arm. Also the fibre is not temperature stable.

2.4.4 Piezoelectric (PZT) Actuators

For small scanning ranges PZT actuators have been used which provide a fast scanning technique⁴². PZT actuators vary the path length by attaching a mirror to the PZT crystal, when a voltage is applied across the crystal it expands allowing the mirror to move over small axial distances. These PZT actuators are limited to ranges of about $100\mu\text{m}$, which may be insufficient for OCT scanning which requires up to a 3mm scanning depth.

2.4.5 Resonant Actuators

For larger scan ranges and faster scanning speeds other actuators can be used to vary the path length of the reference arm these are resonant actuators. One possible resonant actuator is the double spring moving coil system⁴² shown in Fig.2.4. A mirror is attached to a magnet and an electrical current around the coils move the magnet, and hence the mirror, axially. The resonant frequency varies between 45 and 55Hz, which allows a linear scanning, speed of 0.165m s^{-1} , and can be varied by increasing the spring constant.

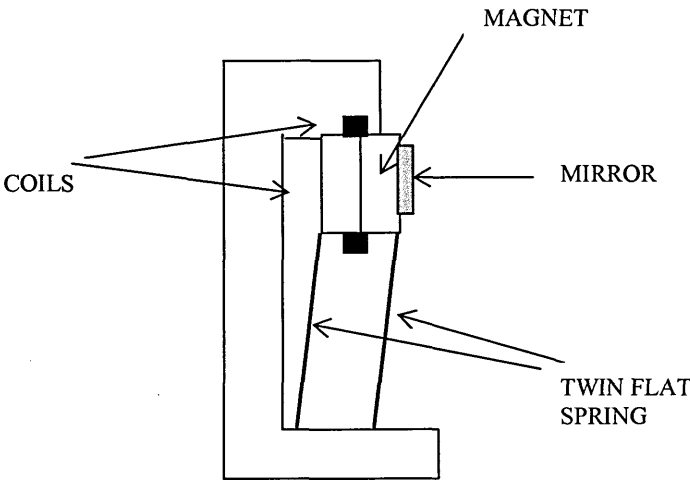


Fig.2.4. Double spring moving coil actuator⁴²

Higher resonant frequencies can be acquired by resonant torsion based scanning devices⁴², such a system is shown in Fig.2.5.

The light beam from the reference arm is deflected by mirror 1 and then hits the mirror of the resonant scanner. The reflected light then passes through the lens and is reflected by mirror 2 and hits the resonant scanner. From here it is reflected onto mirror 3 and then onto the resonant scanner – the light then returns via its original path and is coupled back down the fibre of the reference arm. The beam hits the resonant scanner several times and the path length variation is therefore magnified. Depending on the resonant scanner used in the

arrangement scanning speeds of up to 8kHz can be achieved which is equivalent to 24ms^{-1} linear scanning speed. This scanning method however has a complex set-up and the whole arrangement is not very compact.

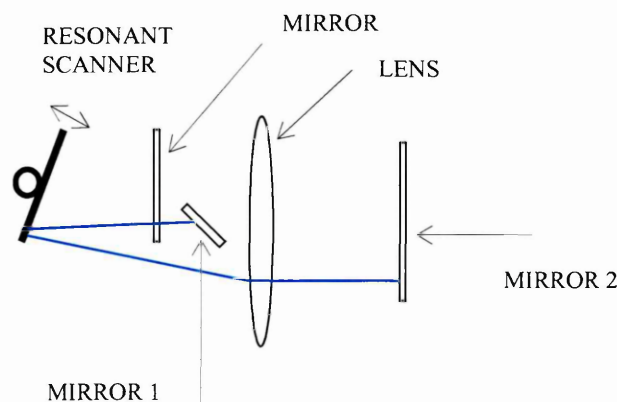


Fig.2.5. Resonant torsion based scanning unit⁴².

2.4.6 Rotating Glass Cubes

Rotating glass cubes have been placed between the end of the fibre and a stationary mirror in the reference arm of the OCT interferometer³⁷. As the cube rotates the light is reflected internally and experiences a change in the path length. The faster the cube rotates the faster the scanning. Electrically driven rotating cubes⁴³ can spin at 1240Hz that allows linear scanning speeds of up to 42ms^{-1} . Some cubes can be driven using high power air turbines⁴⁴ at 427kHz allowing linear speeds of 176m s^{-1} to be achieved. The high-powered air turbines require a high energy input and may become expensive.

The rotating cube has proved to increase scanning speeds but has a very low duty cycle, which will decrease the Signal to Noise Ratio (SNR).

2.4.7 Phase Control Delay Line

An effective method of producing a scanning optical delay in an OCT system uses phase control of the optical field^{45,46,47}. The theory lies in the Fourier Transform (FT) property that a phase ramp in the frequency domain corresponds to a group delay in the time domain:

$$x(t - t_0) \xleftrightarrow{\text{FT}} X(\omega) e^{-j\omega t_0} \quad (1)$$

If the original spectrum $X(\omega)$ is shifted in phase by $-j\omega t_0$ then the function in time will be delayed by t_0 . The high speed scanning delay line consists of a grating-lens pair in a folded, double passed Fourier domain pulse shaping configuration this is show in Fig.2.6.

The light from the reference mirror is diffracted from the grating, which results in a Fourier Transform (FT) of the spectrum. The lens collimates the dispersed spectrum and focuses it onto a scanning mirror. The scanning mirror imposes a linear, wavelength dependent, phase ramp on the spectrum.

A linear phase ramp in the frequency domain results in an optical group delay in the time domain. The light is then redirected back through the lens, which re-collimates the beam and re-converges the spectrum onto the grating where an inverse FT occurs. The beam then diffracts in a reverse manner from the grating and propagates towards the double pass mirror collinear with the incident beam. The double pass mirror returns the light back through an identical path.

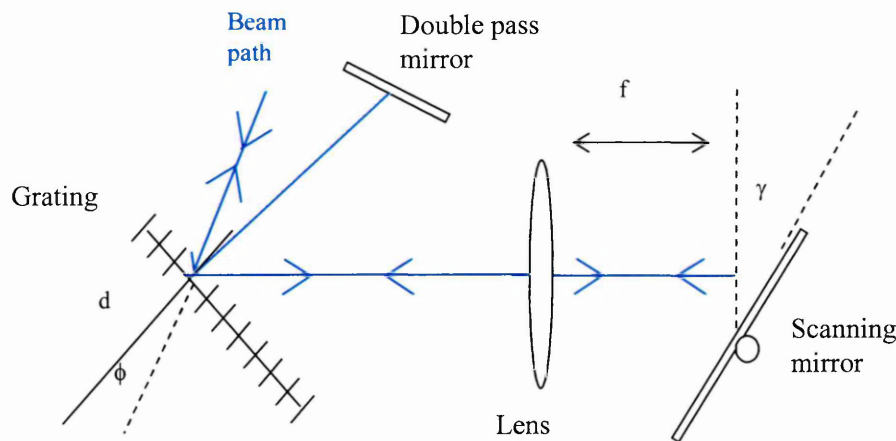


Fig.2.6 Schematic of the Fourier domain optical delay line. ϕ is the angle the light is incident on the grating, d is the groove spacing of the grating, f is the focal length of the lens and γ is the angle tilted by the mirror^{44,45,46}.

This method has achieved linear scanning speeds up to 40ms^{-1} with A-scan repetition rates of 4kHz. Although this technique still uses a scanning mirror, the displacement is angular and only a small angular movement will create a relatively large path length change.

A summary of all scanning methods along with their advantages and limitations is shown in Table.2.3.

Table 2.3 Summarising of all axial scanning techniques

Scanning Technique	Maximum linear speed / m s ⁻¹	Maximum repetition rate / Hz	Typical depth / mm	Comments
Translating Mirror ⁴	<1.0	<10	Up to 20cm or more depending on translating stage length	Bulky for a transportable system using a linear translating mirror.
Sweeping mirror through angles ⁴	1.6	100	>10	
PZT Fibre stretching ⁴¹	3	600	Up to 3mm	Polarisation compensators required and temperature unstable
PZT Actuators ⁴⁴	3.6	1200	Up to 3mm	Limited to depths of 100µm
Resonant Actuator ⁴⁶ (double spring)	0.165	55	>10	
Resonant Actuator ⁴⁶ (torsion based)	24	8000	>10	
Rotating Cube ³⁷	21.3	96	Depending on size of glass cube	Low duty cycle
Rotating Cube with air turbine power ³⁹	176	427000	Depending on size of glass cube	Low duty cycle and expensive.
Phase control delay line ^{44,45,46}	40	4000		

2.5 Extensions of OCT

2.5.1 Introduction

OCT, along with its attractive high resolution imaging of biological tissue, can be used in conjunction with other imaging techniques and also be used for measuring other optical properties, which may enhance contrast or supply additional information about the biological tissue.

2.5.2 Doppler OCT

Ultrasonic Doppler Velocimetry has been used extensively for measuring blood flow by determining the Doppler shift induced in the returning signal after being reflected from flowing blood, either towards or away from the transducer.

In 1997 Chen *et al*⁴⁷ used OCT to look at the velocity of polystyrene beads in a tissue phantom. Izatt *et al*⁴⁸ carried on the idea and later that year showed improvements in accuracy, speed and sensitivity and presented impressive results within rat arteries achieving a resolution of approximately $<50\mu\text{m}$ and a velocity uncertainty of $<0.6\text{mms}^{-1}$. At present the difficulty associated with Doppler OCT is the depth penetration, since the signal fades as the light travels through distances of 2-3mm, therefore the optical signals from only vessels near the surface of the body or *in vivo* measurements are of use when using this technique.

2.5.3 Polarisation Sensitive OCT

A combination of OCT and polarisation sensitive detection was used to produce two dimensional image maps showing the changes in polarisation as a function of depth.

Two-dimensional maps of birefringence of biological tissue are useful in revealing structural information, which may not be displayed using other imaging techniques. It is proposed that Polarisation Sensitive-OCT (PS-OCT) may be used to detect thermal damage by identifying areas of partial loss in birefringence⁴⁹.

De Boer *et al* collected the first results using PS-OCT in 1997⁵⁰ using a modified Michelson Interferometer and an 862nm SLD with bandwidth 21nm.

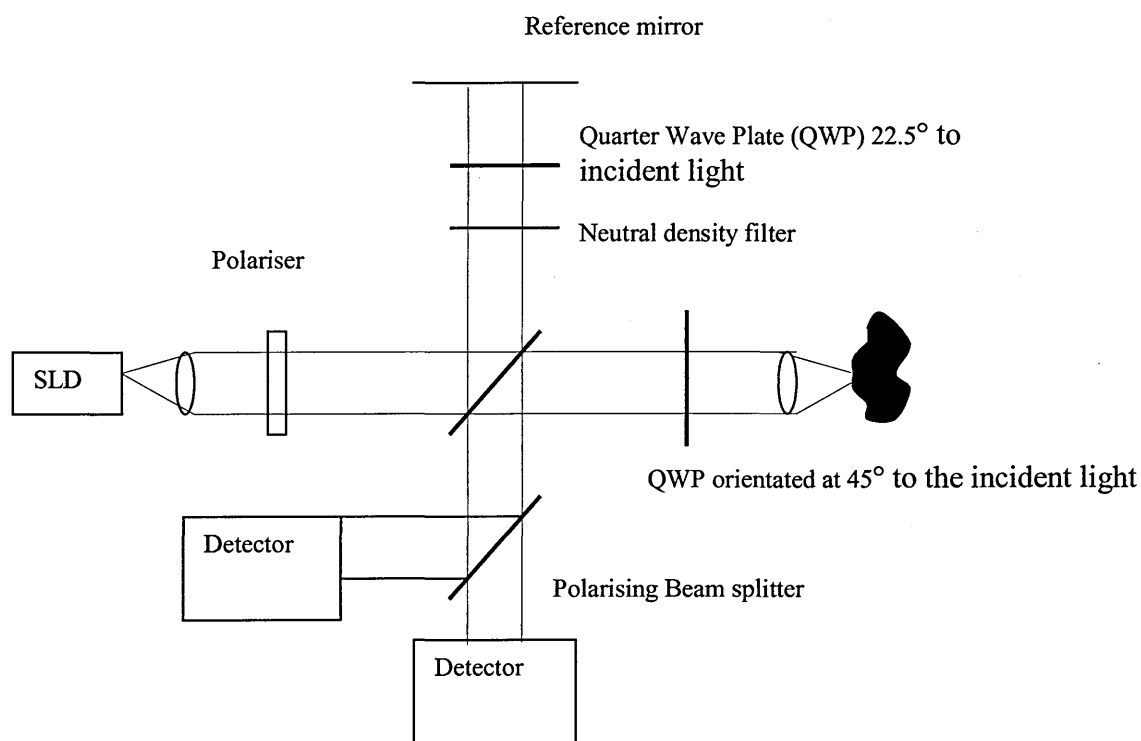


Fig.2.7 Schematic for PS-OCT

Fig.2.7 shows the set-up for PS-OCT. Light from an SLD passes through a polariser, which selects a pure, linear, vertical input state. This is split 50/50 by a polarisation insensitive beam splitter into a reference and sample arm. Light in the reference arm passes through a Quarter Wave Plate (QWP) orientated at 22.5° to the incident polarisation. Following reflection from a reference mirror (mounted on a translation stage for OCT imaging) the light emerges with a linear polarisation state 45° to the incident light.

Light in the sample arm passes through a QWP orientated at 45° to the incident light so that circularly polarised light is incident on the tissue sample. After passing through the lens system and tissue sample the light is in an arbitrary polarisation state depending on the birefringence of the reflection site.

The reference and sample beams recombine and are split into their horizontal (A_H) and vertical (A_V) polarised components (Equation 2 and 3) of the interference intensity, by a polarising beam splitter. The Polarisation state from each reflection is then calculated using the Jones matrix Formalism, discussed in Appendix A.

$$I_{H(z)} = R_{(z)} \sin^2(k_0 z \delta) \quad (2)$$

$$I_{V(z)} = R_{(z)} \cos^2(k_0 z \delta) \quad (3)$$

Total detected signal is given by equation 4

$$\log_{10} I_{s(z)} = \log_{10} (I_{H(z)} + I_{V(z)}) \quad (4)$$

Images of fresh bovine tissue were shown, with astonishing results demonstrating changes in birefringence up to $700\mu\text{m}$ deep. Results were displayed for normal and thermally damaged tissue and there was a clear indication where the tissue was irradiated due to the

loss of birefringence. It has been proposed that this technique would be ideal for optimal dosimetry for patients undergoing laser therapy.

In 1998 Everett *et al*⁵¹ developed a PS-OCT system where the light is coupled from the bulk-optic system to the detectors, the fibres filter all spatially incoherent light improving SNR. The source used was a SLD centred at 1310nm for improved penetration depth. Impressive depth maps of the birefringence of porcine myocardium were published. Evidence showing how PS-OCT could eliminate artefacts found in conventional OCT was displayed by producing an OCT image using only one of the detectors and this was compared to a conventional OCT image. A layer of reduced scattering was displayed on the PS-OCT image approximately 200-450 μ m depth, this was not visible on the conventional OCT scan and is shown to be a birefringence induced artefact eliminated by PS-OCT.

In 1999⁵² OCT was extended to map the birefringence of skeletal muscle, bone, skin and brain tissue.

2.5.4 Spectroscopic OCT

Spectroscopic detection using information from an OCT signal has recently come under investigation due to the availability of sufficiently broad bandwidth sources. This technique measures the spectral content of the back-scattered light over the entire optical bandwidth in order to obtain additional information about the properties of tissue. In January 2000, Morgner *et al*⁵³ reported the highest OCT resolution to date using spectroscopic OCT in conjunction with conventional OCT. Longitudinal resolutions of 1 μ m and lateral resolutions of 3 μ m were achieved using a Ti:Al₂O₃ ultra broad bandwidth femtosecond laser over the range 650-1000nm. In standard OCT only the envelope of the interference signal is detected and the image consists of a two dimensional array representing the backscattered light intensity. Spectral information is obtained by

measuring the full interference signal and gaining the corresponding spectral content. Spectroscopic detection is performed using a Morlet Wavelet transformation (Equation 5) this calculates an entire spectrum $W(\Omega,\tau)$ for each point (x,τ) on the OCT image.

$$W(\Omega,\tau) = \left| \text{FT} \left\{ \int I_d(t+\tau) \exp \left[- \left(\frac{t}{t_0} \right)^2 \right] \right\} \right|^2 \tag{5}$$

Where $\Omega(x,z)$ is the centre of mass of the spectra, FT=Fourier transform, I_d = the resulting interferogram, t = scanning time and τ is the time delay between mirror movements.

Absorption features due to the presence of oxyhemoglobin and deoxyhemoglobin can be identified within the spectral range, and may enable functional imaging or identification of certain tissues. Presently however, spectroscopic OCT has only been used to enhance the resolution and contrast of conventional OCT images.

All the above techniques are summarised in Table 2.4.

Table 2.4 Summary of OCT extensions

Technique	Description	Comments
Conventional oct ⁽¹⁾	Measures change in refractive index as a function of depth.	2-D images of tissue structure.
Doppler oct ^{49,50}	Measures velocity of blood flow.	Limited by depth penetration through vessel walls.
Polarisation sensitive oct ^{51,52}	Measures birefringence of tissue.	Additional image contrast for conventional OCT. Also identifies birefringence properties of tissue
Spectroscopic oct ⁵⁵	Measures spectral information from the interferogram.	Improves resolution of OCT images and identifies spectral properties of tissue

2.6 OCT Imaging Instruments

After the first OCT images of biological structures using a Michelson interferometer⁴, other interferometer configurations were investigated in order to achieve optimal performance.

In 1996 Tearney *et al*⁵⁴ designed a fibre-optic catheter endoscope, 1.1mm in diameter, for *in-vivo* imaging of blood vessels. Instead of the usual lateral scanning that most other authors have demonstrated, Tearney produced a technique where the end of the fibre was allowed to rotate along the fibre axis to create 'ring shaped' tomograms of the vessel surface. This was accomplished by using a right-angled prism at the fibre end to reflect the light 90° onto the tissue sample. The fibre was then rotated to obtain more profiles in order to build up the image.

In 1997 Boppart *et al*⁵⁵ discussed three possible designs for OCT surgical probes. A single fibre was used for the sample arm and in each instance the lateral scanning was performed with piezoelectric cantilevers.

In 1997⁵⁶ the first *in-vivo* OCT images of the human mucosa in the respiratory and gastrointestinal (GI) tract were performed using a single-fibre endoscope 1cm in diameter. Lateral scanning was performed using an electromechanical unit within the sample head to move the fibre known, lateral distances. Methods for lateral scanning can be improved to decrease the scanning speed, but the over all determination of faster scanning is the longitudinal acquisition rate and most endoscopic sample heads are used in conjunction with faster longitudinal scanning methods to improve their performance.

Other improvements with OCT systems have suggested using directional couplers and circulators to improve the power into the system. A number of trials using combinations of these has been carried out^{57,58} and in 1999⁵⁹ a paper was published determining the optimal interferometric design to achieve the highest power-conserving set-up. This configuration consisted of an unbalanced coupler and a circulator, which improved the sensitivity by a 4.1dB advantage over the standard Michelson.

Recently there has been an interest in OCT scanners, which require no moving parts. This is an attractive feature for faster scanning techniques and endoscopic applications. One such system was presented in 1999⁶⁰ using an imaging spectrograph as an alternative to a detector within a Michelson Interferometer (Fig.2.8). The reference mirror and the sample arms are stationary and the interferogram (spectrum of interference output) is analysed instead. The Fourier Transform (FT) of the interferogram contains depth information. The spectrograph images a line along the sample surface onto the entrance slit of the imaging spectrograph. At the output plane of the spectrograph it is possible to obtain depth information (from the spectral content) along each point in that line, which can then be processed into a two-dimensional, depth-by-surface image captured using a CCD camera.

The disadvantage of using this technique is that it is computationally intensive since the FT of each of the acquired spectra must be taken in order to retrieve all the depth-resolved reflectance.

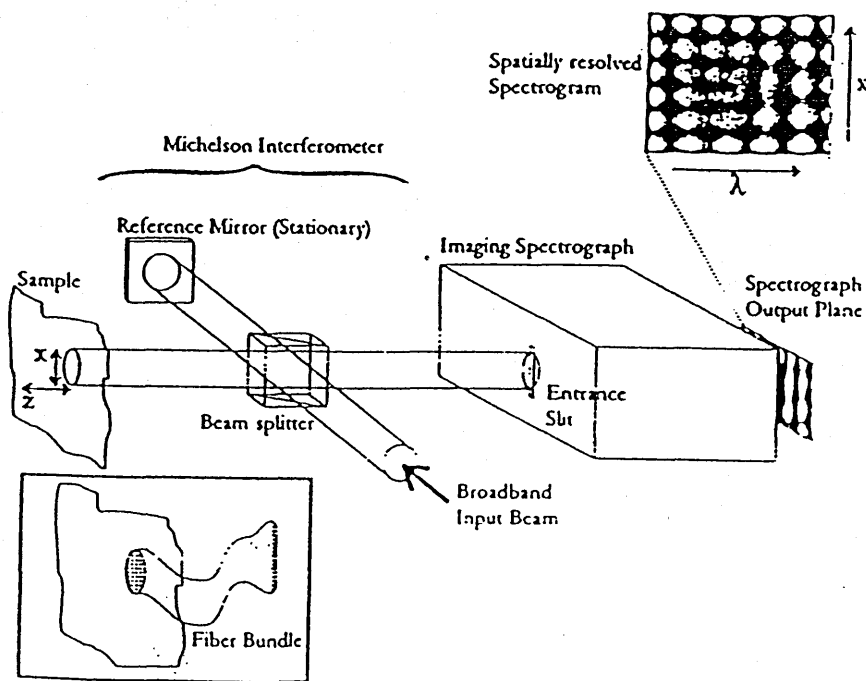


Fig.2.8 showing non-moving parts OCT scanner⁶⁰

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Chapter 3

Polarisation Effects in Interferometric Imaging

3.1 Polarisation Effects of Light Propagating Through Tissue

3.1.1. Polarisation changes due to scattering within tissue

The State of Polarisation (SOP) of light within an OCT system not only changes when non-polarisation-maintaining fibres are used but also when the light propagates through a turbid media such as tissue¹. These polarisation changes decrease the Signal to Noise Ratio (SNR) of an OCT system and need to be compensated for to improve image quality. This section interrogates the effects of depolarisation within tissue by analyzing how the interferometric signal fades as light travels through different tissue types with different birefringence values. This is compared with that occurring within the interferometer to assess the relative magnitude of the effects.

The SOP of light propagating through biological tissue changes due to scattering by particles within the tissue² and birefringence of the tissue³. The most dominant effect is expected to be the birefringence of tissue, as the scattering particles, which are of differing sizes, would change the polarisation in a random manner. However a study has shown that particles that are arranged in a macroscopic order (similarly sized) do exhibit trends in the depolarisation of light as the light is scattered from them.

3.1.1.1. Polarisation changes of light scattering from particles

One study looked at how the polarisation of light changes as it travels through different concentrations of varying sized polystyrene microspheres suspended in solution. It showed that in macroscopic structured particles the polarisation is maintained differently in materials consisting of different concentrations of microspheres^{2,3}. In dilute suspensions of smaller Rayleigh spheres ($d \ll \lambda/10$) circularly polarised light is depolarised more quickly than linear polarised light. As the sphere size increases into the Mie scattering size spheres ($d \geq \lambda/10$) it is found that linearly polarised light is depolarised more quickly than circularly polarised light. In the transition between the two states both SOP are depolarised equally.

For dense suspensions of larger microspheres (9 μm) it was found⁴ that linear polarised light was depolarised more quickly than circular. Circularly polarised light was depolarised similarly or more quickly than linear polarised light in smaller particles (0.5-2 μm) within dense suspensions. Dense suspensions of Rayleigh scatterers were not included.

A similar investigation showed that the SOP is maintained differently in dense concentrations than dilute concentrations. As the scatterer concentration increases up to 3-5%, the degrees of linear and circular polarisation both decrease with linearly polarised light decreasing more quickly in the first 5% concentration. After 5% concentration, however, there is a change and the degrees of circular and linear polarisation starts to increase with scatterer concentration.

3.1.1.2. Polarisation changes within tissue due to scattering

Although the results above were carried out using a suspension of polystyrene microspheres, they can be related to the depolarisation effects in biological tissue with a similar dense suspension². Most tissue is composed of cellular and sub-cellular structures located closely together with sizes of roughly 0.5 - 1 μ m diameter and so it would be expected that Mie scattering would be dominant and circularly polarised light would be depolarised more quickly. Studies carried out with myocardial tissue have shown that circularly polarised light is depolarised more quickly which indicates that the propagation of polarisation within tissue is as expected and more dependant on the density of the scatterers than their size.

3.1.2. Polarisation changes due to tissue birefringence

Certain types of biological tissue have been shown to exhibit birefringence properties⁵. The birefringence is linear and is a result of special proteins and collagen fibres present in certain tissue types, which make up the extracellular matrix. These proteins and fibres are packed in such a way that the index of refraction of their molecular structure is higher along the length of the fibre than along the cross section, leading to linear birefringence.

Linear birefringence in biological tissue can have positive and negative effects for OCT imaging. The birefringence of tissue is characteristic of individual tissue types and by measuring the birefringence value of the tissue the specific tissue type can be identified^{6,7}. Examining the birefringence of tissue can also identify areas of thermally damaged tissue⁸, when the temperature of tissue is raised above a certain 'melting' threshold, the protein and collagen fibres unfold and their structure is lost, this results in a more isotropic configuration and the birefringence properties are lost. This loss in birefringence is a useful tool for dosimetry in laser (or any other thermally damaging) surgery.

However, the changing polarisation of light in OCT systems can degrade the Signal to Noise Ratio (SNR) of the detected signal and also cause artefacts in OCT images. As light passes through tissue with birefringence properties, its polarisation state will change and on reflection from the tissue, will have a different polarisation state than the reference beam. As the polarisation states between the reference and the sample arm become orthogonal the SNR and visibility of the signal will fall to zero. The OCT images that are observed may obtain ‘stripy’ artefacts where there is no signal due to the continuous rotation of polarisation states becoming orthogonal as the light travels deeper in the tissue.

Table 3.1 showing birefringence values of various tissue types at a given centre wavelength

Type of tissue	Tissue birefringence	Centre wavelength / nm
Skeletal Muscle ⁵	0.00220	856
Porcine Myocardium ⁸ ↓	0.00111	1310
Myocardium ⁸ ↓	0.00066	1310
Thermal damaged Myocardium ⁸	0.00018	1310
Rat Skin ⁵	0.00140	850
Rat Muscle ⁵ ↓	0.00220	856
Bovine Tendon ⁵ ↓	0.00370	862

This chapter looks at the negative effects, such as image degradation and signal fading of tissue birefringence and its effects on the signal visibility as light travels through biological tissue and at what depth the polarisation states become orthogonal as this will cause a complete loss of signal.

Data from a number of tissue types with varying degrees of birefringence were used to calculate the phase change between the initial state of polarisation and the polarisation of light at different tissue depths. This was to give an indication of the magnitude of the depth, light penetrating into the sample arm of an OCT interferometer, can travel before decreasing the SNR significantly for OCT imaging.

Table 3.1 shows the birefringence values for a number of various tissue types measured at different wavelengths.

Using Equation 1 the phase change of the sample arm for each tissue type was calculated as a function of depth through tissue⁹.

$$(2\pi)m + \frac{2\pi}{\lambda_0} \delta d = \Delta\phi \quad (1)$$

where m is an integer corresponding to the order, λ_0 is the centre wavelength of the light passing through the tissue, δ is the birefringence of the tissue and $\Delta\phi$ is the phase change of the light in the sample arm.

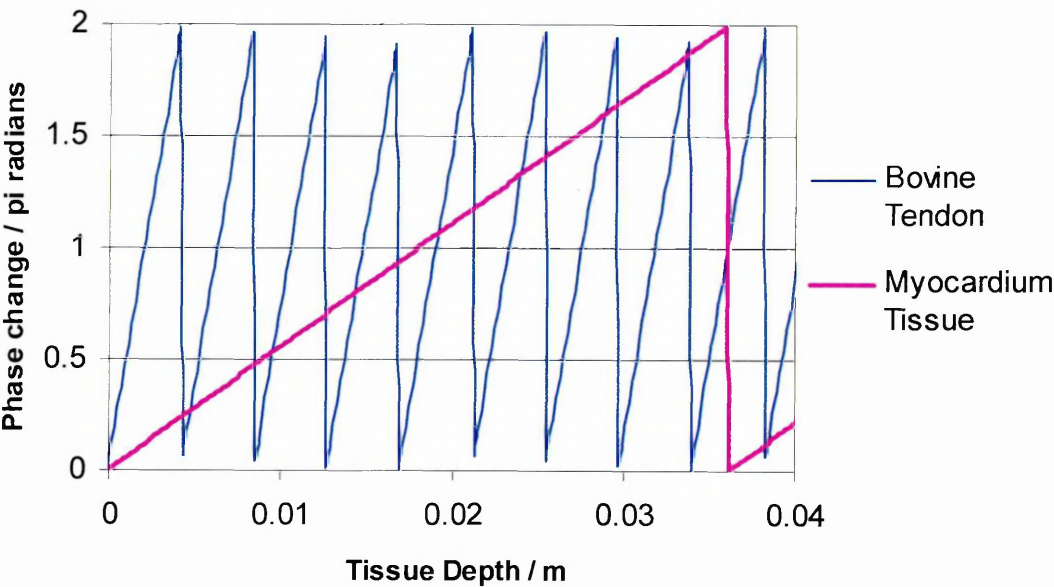


Fig.3.1. Graph showing the Phase change between the initial and final SOP light as it travels through tissue as a function of tissue depth, the wavelength for each tissue type is shown in Table 3.1.

The phase change for the tissues with the highest birefringence (bovine tendon) and lowest birefringence (myocardium) were plotted as a function of depth Fig.3.1 in order to determine the most extreme changes in phase and hence SNR with tissue depth. The graph assumes the initial state of polarisation is linear and the wavelength for each tissue types corresponds to its measured value in Table.3.1.

From Fig.3.1 it is possible to see how the effect of the birefringence of the two tissue types effects the polarisation of the sample arm. The bovine tendon with a relatively high birefringence value results in the light changing phase by 2π radians every 4mm, whereas the myocardium will only result in a phase change of 2π every 37mm.

To determine what effect the phase change of the sample arm has on the detected signal, Fig.3.2. shows how the signal visibility varies with tissue depth for the bovine tendon and myocardium tissue, using the relationship in Equation 2.

$$V = \cos \eta \quad (2)$$

where V is the signal visibility and η is the angle between the two output states of polarisation (sample and reference arm). In this case this will be equal to the phase change of the sample arm, as the polarisation state in the reference arm has not changed.

From Fig.3.2 it can be seen that the signal visibility of the high birefringence bovine tissue will be zero every 3mm where as the lower birefringence myocardium tissue will have zero visibility every 18mm.

A number of different tissue types with birefringence properties were used (from Table 3.1) to plot a graph of tissue depth against phase change using equation 1, this is shown in Fig.3.3. This shows how far light can travel through tissue before experiencing a phase change of 90° which would result in zero visibility.

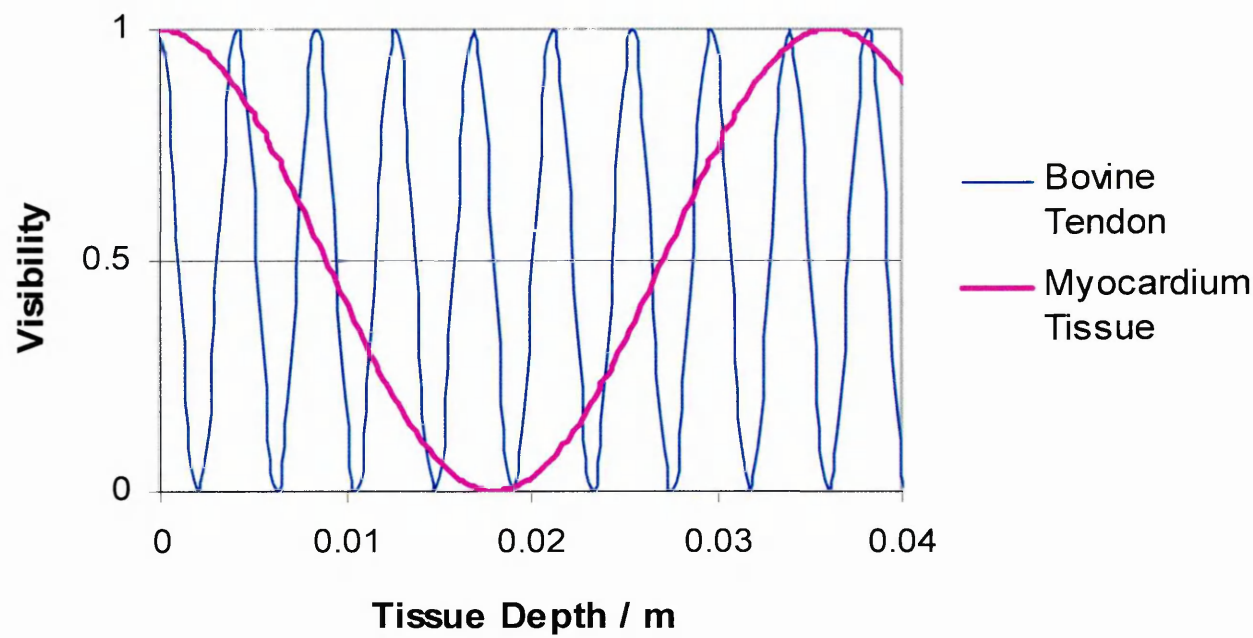


Fig.3.2. Graph showing signal visibility as a function of tissue depth

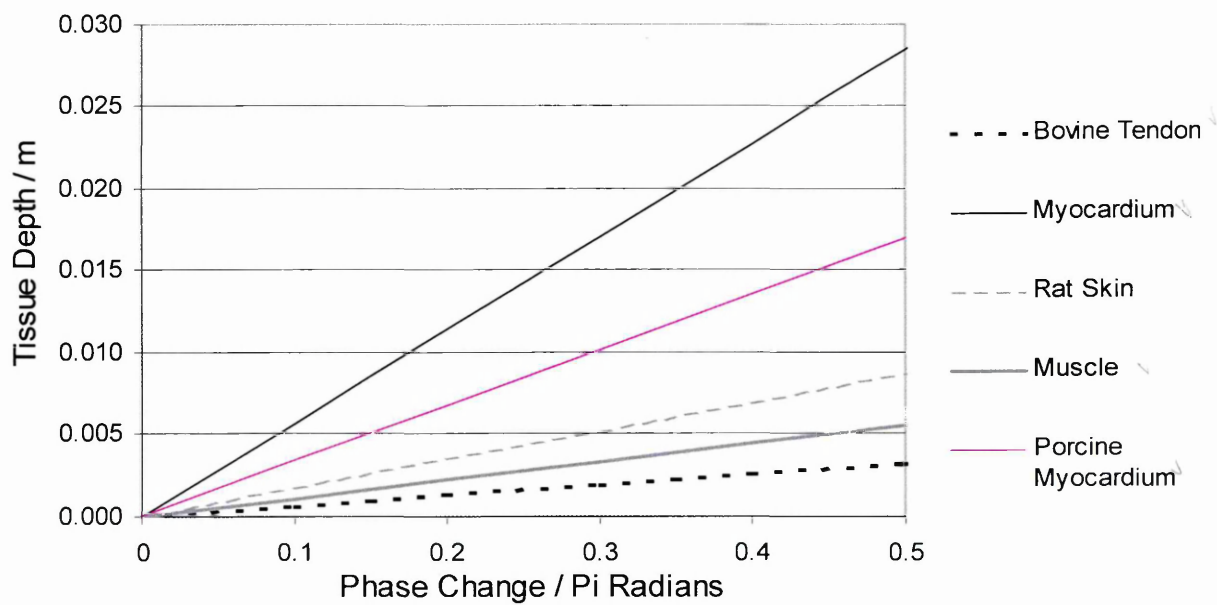


Fig.3.3. Graph showing tissue depth to give 90° phase change of the sample arm light, for various tissue types.

All tissues exhibiting birefringence properties in this study showed that zero visibility for an interferometric signal travelling through the tissue would occur at a minimum distance of 3mm and this is for bovine tendon. Esophageal tissue would be comparable to myocardium or skin rather than bovine tendon (which contains large amounts of collagen fibres) and therefore light would have to travel a minimum of 10mm until the signal visibility was lost due to the tissue birefringence. OCT systems scan tissue thicknesses up to 2-3mm and so the signal loss due to birefringence effects would be negligible except for the particular case of bovine tendon type material.

3.2 Polarisation and Dispersion Effects of Light Propagating Through Optical Fibre

Endoscopic OCT is of great importance in the medical field as it allows high resolution depth images of tissue *in vivo* to be created. Most OCT systems to date use a conventional Michelson interferometer as the imaging system¹¹ as this, compared to other interferometers, is the simplest set-up and gives a high Signal Noise Ratio (SNR). However, this set-up may not be the ideal configuration for endoscopic applications as polarisation and phase changes may cause a decrease in SNR as a result of the sample arm experiencing a change of environment, such as temperature and strain, as it is placed inside the patient^{12,13}.

Most OCT interferometers are constructed using fibre optics as these allow a more flexible, sensitive and stable system⁵. The interference signals from interferometers which use non-polarisation maintaining fibres are subject to polarisation induced fading (PIF) which occurs when the state of polarisation (SOP) of the two interfering beams is different¹⁴, ultimately leading to complete signal failure when the interfering beams are of orthogonal SOP. This is likely to happen when the sample arm of the Michelson interferometer

experiences changes in temperature and bending induced during deployment of the fibre via bends and twists, or the SOP in the two interferometer arms drift randomly in non-polarisation maintaining fibres.

An evaluation of how the SNR is affected by mis-matching polarisation states is carried out along with an assessment to determine how these effects are likely to affect the performance of endoscopic OCT interferometer designs.

The amount of coherent and incoherent power from the sample arm was theoretically varied systematically and the corresponding SNR calculated, shown in Appendix B. The graph showing how the SNR varies with the polarisation mis-match between the two interferometer arms is shown in Fig.3.4. The mixing ratio = 1 when the polarisation between the two arms are equal and 0 when they are orthogonal.

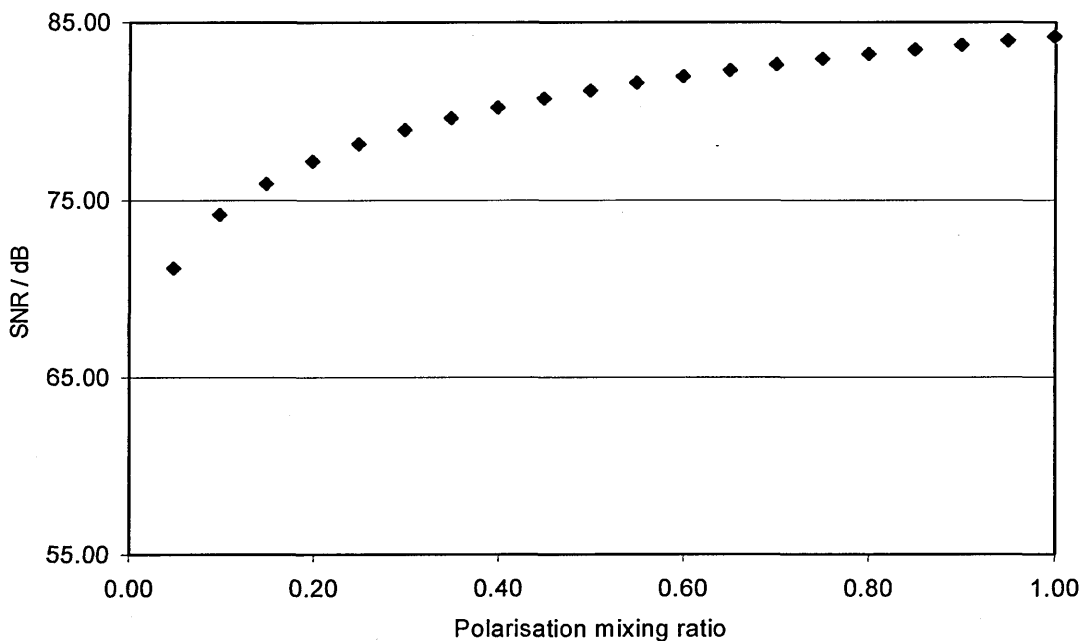


Fig.3.4 Graph Showing SNR as a function of Polarisation mis-match of two Interferometer Arms

As expected the highest SNR is achieved when the polarisation states are equal. There is a sharp fall off in SNR from 80 – 0 between 0.5 and 0 mixing efficiency. The graph shows that small variations between the two polarisation states result in a small change in SNR and this increases with increasing polarisation mis-match. Therefore small changes in the polarisation of the sample arm with respect to the reference arm will have insignificant results on the SNR. However the SNR can decrease to zero when the polarisation states are orthogonal and this is very significant to the image quality.

Polarisation changes in the reference and sample arms can occur if the optical fibres used are non polarisation maintaining and this variation is random within the fibre, but changes with deployment which is a particular problem for endoscopic systems.

3.3 Summary

In conclusion the most important aspect of polarisation induced fading in an endoscopic OCT system occurs within the optical fibres and since we are focussing on esophageal tissue the PIF is insignificant in the sample itself.

An alternative configuration for endoscopic OCT is the Fizeau interferometer¹¹ in which the light in the sample and reference arm travel down the same fibre and therefore experience the same external changes. Consequently any phase and polarisation changes experienced in the fibre are equal in both interferometer beams and any changes in polarisation do not change the SNR of the detected signal upon interference.

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Chapter 4

Initial Experimental Investigation of the Michelson and Fizeau Interferometer

4.1 Initial Experimental Investigation

4.1.1 Introduction

The Fizeau interferometer has been shown theoretically from Chapter 3 to be the most suitable configuration for OCT. One disadvantage of using the Fizeau is that the SNR would be considerably lower than the Michelson interferometer as it requires a second processing interferometer.

The initial investigation involved setting up both the Fizeau and Michelson interferometer to see the type of signals obtained from the experimental set-up using our equipment and how this compared to previous results and how comparable the SNR from both interferometers were to each other.

4.1.2 Interference Signals from Low Coherence Interferometry

An interference signal can be detected when two electromagnetic waves combine that are coherent with one another. The photocurrent for the detected signal is derived in Appendix C for a Michelson Interferometer and Appendix D for a Fizeau Interferometer and is given by Eq 1 and 2 respectively.

$$I_{dc} = \rho [P_r + P_s + P_x + 2\sqrt{P_r P_s} \cos(k_0 \Delta l)] \quad (1)$$

where ρ is the responsivity of the detector, which will be the same for all beams, P_r , P_s and P_x are the reflected powers from the reference arm, sample arm and coherent and sample arm and incoherent, k_0 is the wave number of the centre wavelength of the source and Δl is the path length difference between the sample and reference arm.

$$I = \rho \left\{ P_{11} + P_{12} + P_{21} + P_{22} \left[\frac{2\sqrt{P_{r1}P_{s1}} + P_{r2}P_{s2} \cos ik(X_1) + 2\sqrt{P_{r1}P_{r2}} + P_{s1}P_{s2} \cos ik(X_2) +}{2\sqrt{P_{r1}P_{s2}} + P_{r2}P_{s1} \cos ik(X_1 - X_2)} \right] \right\} \quad (2)$$

P_{r1} , P_{r2} , P_{s1} and P_{s2} are the powers in the reference and sample arms traveling down arm 1 and 2 of the receiving Michelson interferometer. k_0 is the wave number of the centre wavelength of the source and Δl is the path length difference between the sample and reference arm.

The first three terms in Eq.1 and 2 represent a DC component of the signal and the last term represents the AC signal. The AC signal is described by a cosine function with an amplitude dependent on the relative phase between the two combining beams.

OCT uses low coherence interferometry to measure small changes in the distance between the reflected signals. A monochromatic source is a single wavelength source. Low coherence sources, which are used for OCT, are polychromatic. These sources have a spectrum comprising of a large range of wavelengths with a complex waveform $\psi(x)$ ¹ comprising the sum of all monochromatic waves (Eq.3).

$$\psi(x) = \int_0^{\infty} A(\lambda) \cos 2k(\lambda)x \, d\lambda \quad (3)$$

In order to measure the amplitude of the complex wavefunction a Fourier transform is performed on the wave function (Eq.4²).

$$A(\lambda) = \text{FT} [\psi(x)] \quad (4)$$

The Fourier Transform implies that for a low coherence light source the autocorrelation function $A(\lambda)$ will be narrow and sharply peaked as the spread of wavelengths within the wavefunction is broad. Low coherence interferometry can achieve high axial resolution which is necessary for OCT³ as signals can be readily resolved within the coherence length of the source.

The interference signal from a low coherence source is proportional to the interference signal from each individual monochromatic component within the wave function and the autocorrelation function of the low coherence waveform has a Gaussian distribution⁴

Fig.4.1

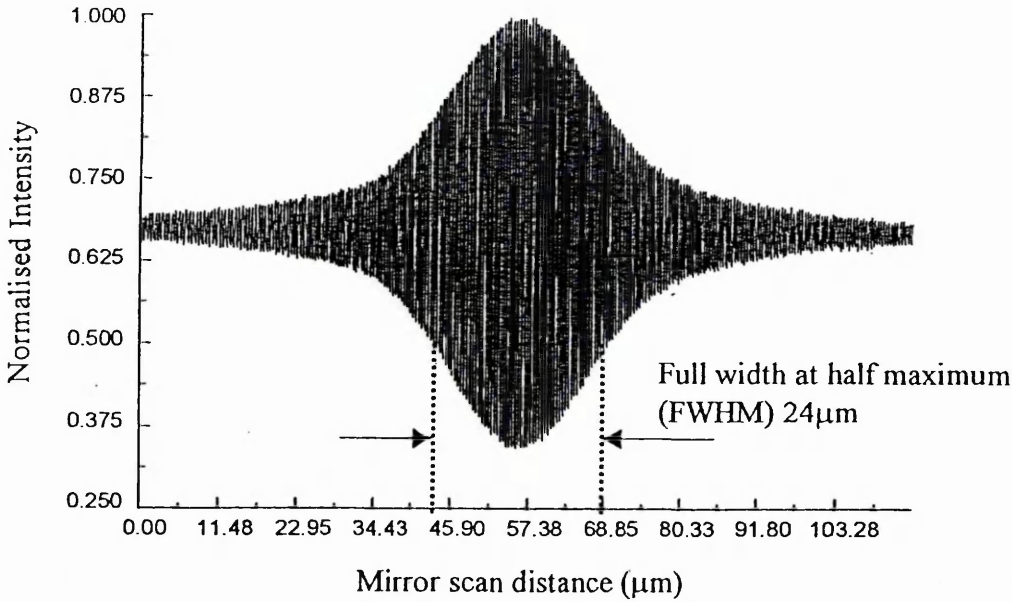


Fig 4.1 Autocorrelation function for a low coherence source (SLD) showing the FWHM corresponding to the coherence length of the source.

The Full Width at Half Maximum (FWHM) of the Gaussian profile is determined by the coherence length of the source⁵ (Eq.5) and is a measure of the axial resolution of the optical signals detected.

$$L_c = \frac{4 \ln 2}{\pi} \frac{\lambda_0^2}{\Delta \lambda_0} \quad (5)$$

where L_c is the coherence length of the source and λ_0 is the mean centre wavelength of the optical source.

In order to distinguish two reflected signals the separation between the reflectors must be greater than half the coherence length of the optical source⁶.

The most common sources for OCT imaging are SLDs (Chapter 2) and these are capable of achieving coherence lengths of ~10s of micrometers, which is an extremely high resolution for medical imaging as you can see down to cell size structure.

4.1.3 Initial Experimental Set-up for the Michelson Interferometer

The Michelson interferometer is the most widely used interferometer for OCT imaging as its set-up is relatively straightforward. A Michelson Interferometer was set-up on an optical bench as shown in Fig.4.2.

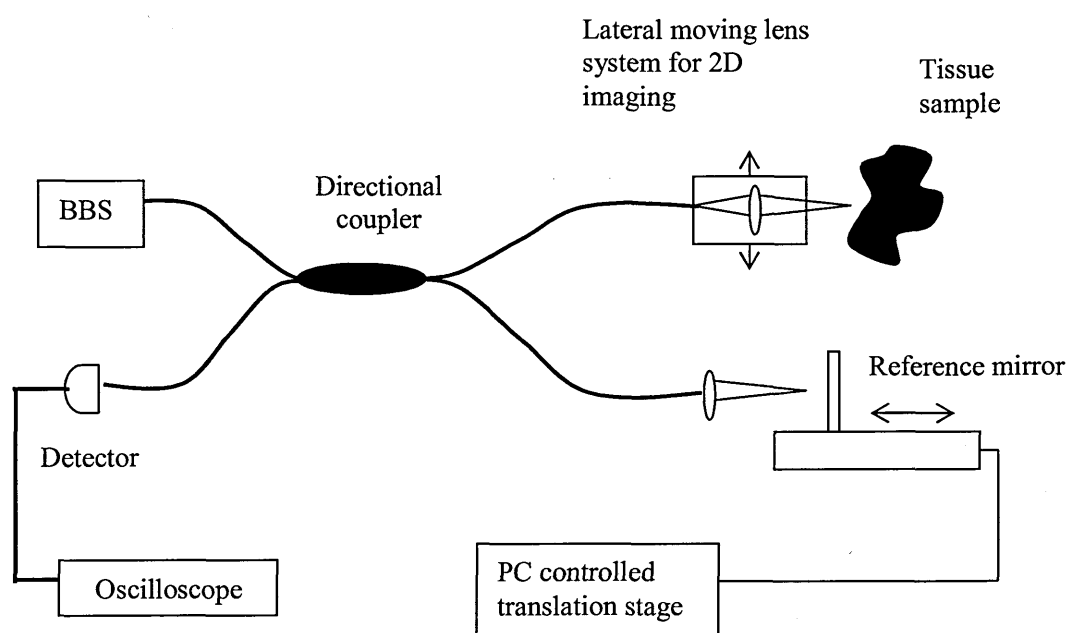


Fig 4.2 Showing the initial set-up for a Michelson interferometer based OCT system

The set up consists of mainly of a directional coupler, a light source and a detection system. The light source used within this system was a SLD (Hamamatsu SD1S501P) of 1mW power and centre wavelength of 840nm. The bandwidth of the source was 30nm corresponding to a coherence length of $24\mu\text{m}$ (Eq.5). The light from the SLD was launched into a single mode optical fibre leading into a 3dB directional coupler with 50/50 splitting ratio. Half of the light through the coupler went down a reference arm where it was reflected back by a 100% reflecting mirror, which was attached to a translation stage. Half of the light was focussed onto another mirror via a lens system, which acted as the reference arm.

The focussing lens system consisted of a 20mm focal length lens which collimated the light beam and a 30mm focal length lens set at the focal length of lens 1 to focus the beam onto the sample, in this case the mirror. For the purpose of this initial set-up the beam waist and the Rayleigh length were not too important as a mirror was being used as a sample which has only a single high reflectivity surface.

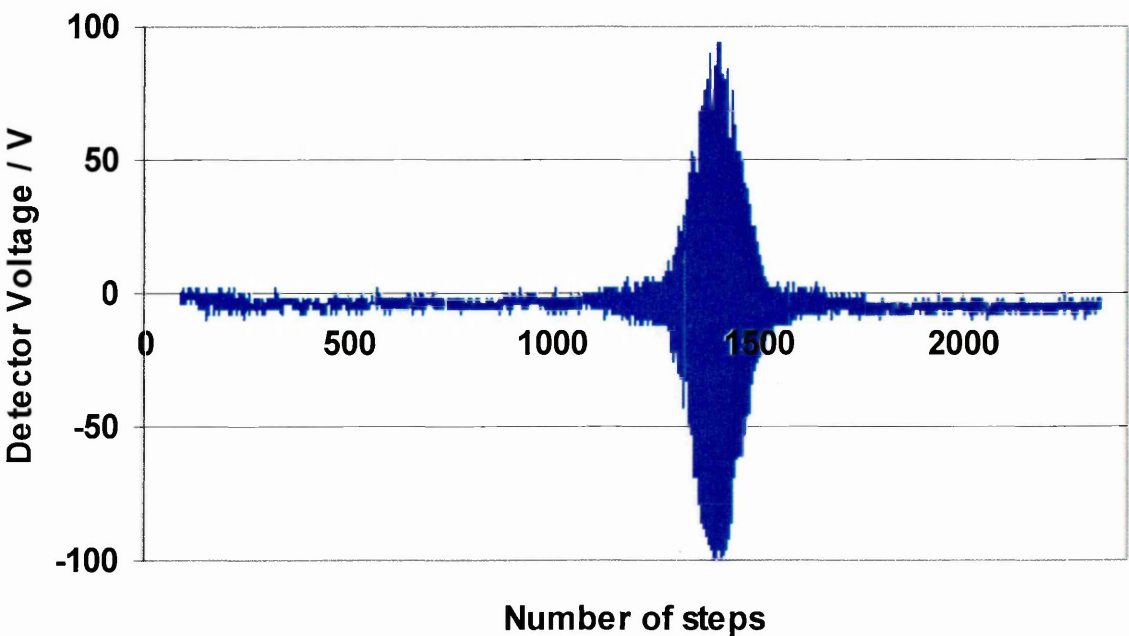


Fig.4.3 Interference signal from a Michelson interferometer. Normalised to 100% and scaled so the baseline=0.

The light reflected back from both interferometer arms re-enters the direction coupler and 50% of the combined light exits through the coupler into a detector. The detector used for this set-up was a PIN photodiode (BPX65) and this was connected to an oscilloscope in order for the signal to be detected (Photodiode specifications are shown in Appendix E)

Once the system was set-up and the optical components aligned, the translation stage was moved backwards and forwards in order to match the path length between the sample and the reference arm. The translation stage was run on a DOS program called PROMOVE and could move in steps of $0.0085\mu\text{m}$ at varying speeds.

An interference signal was observed on the oscilloscope when the path length travelled by the light between the sample and reference arm was equal or to within the coherence length of the source.

The interference signal obtained from the Michelson interferometer is shown in Fig.4.3.

4.1.4 Initial Experimental Set-up for the Fizeau Interferometer

Although the Michelson interferometer is most widely used for OCT imaging it may not be the best configuration for all OCT applications and in particular those using endoscopes as discussed in chapter 3.

The Fizeau interferometer was initially set up for investigation using the same optical components as the previous Michelson set-up.

The Fizeau was set-up on the optical bench as shown in Fig.4.4 The same SLD was used as for the Michelson set-up (Hamamatsu SD1S501P, 1mW, 30nm bandwidth). The light was split 50/50 by a 3dB directional coupler, 50% of light is directed to the Fizeau sensing head and 50% to a fibre arm immersed in index matching liquid (IML) which prevents any reflection back into the interferometer.

Light is reflected at the distal fibre end (4%) forming the reference beam of the Fizeau interferometer. The 96% transmitted light is focussed by the same lens system as the previous set-up onto a stationary mirror where it is reflected back into the system.

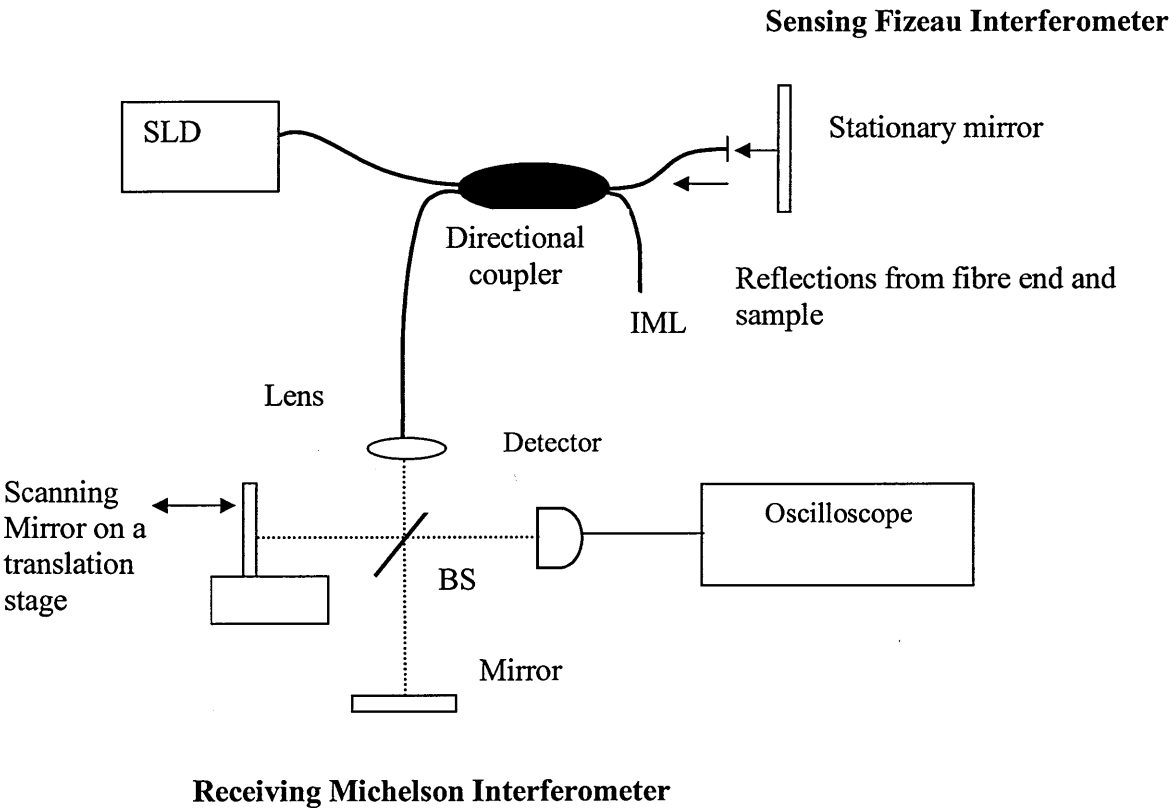


Fig.4.4. Initial set-up for the Fizeau interferometer.

The reference and the sample beam are directed back into the coupler and 50% of the light emerges from each exit arm of the coupler. One arm directs the light back to the source and the other arm into a processing Michelson interferometer. The light from the end of the fibre entering the processing Michelson interferometer is collimated by a 20mm focal length lens into a bulk optic Michelson interferometer. The light is split 50/50 by a beam splitter and the two beams are directed onto 100% reflecting mirrors. One mirror is stationary and reflects light back towards the beam splitter and the second mirror is mounted onto a translation stage where it is scanned axially. The beams recombine at the beam splitter where 50% of the light is directed onto a detector, the same detector as the

previous set-up was used (a PIN photodiode type BPX65). The detector is connected to an oscilloscope to measure the signal from the interferometer.

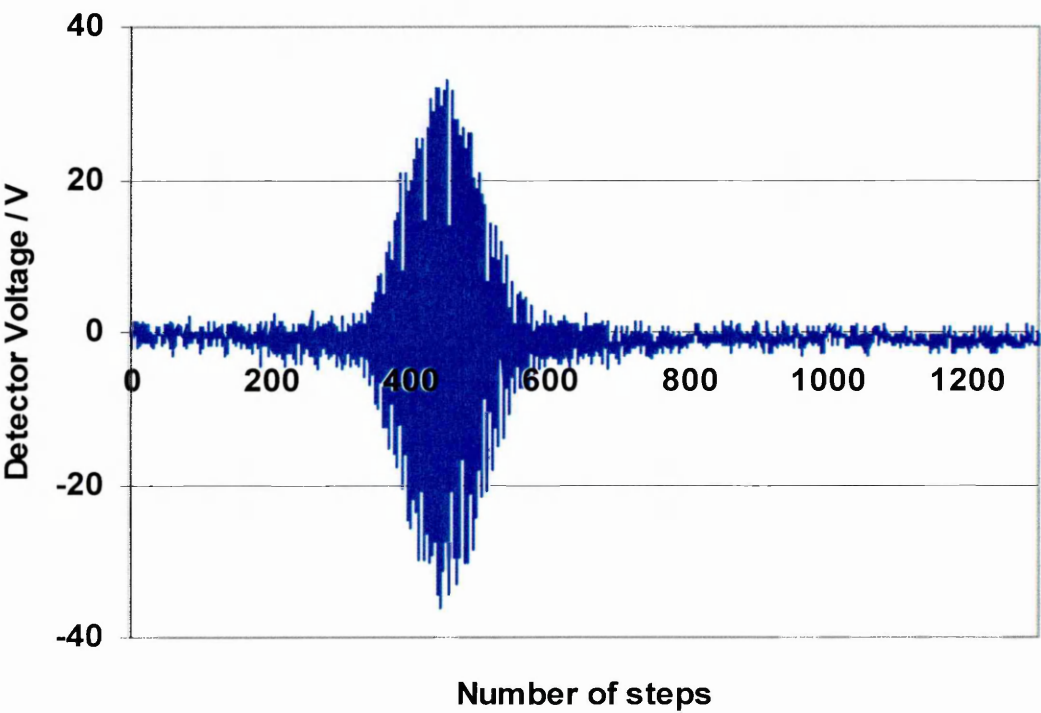


Fig.4.5 Interference signal measured from the Fizeau interferometer set-up. Normalised to 100% and scaled so that the baseline=0

As the mirror is move backwards and forwards it changes the path length of the reference arm and when this is equal to path length of the sample arm (or to within the coherence length of the source) an interference signal can be detected on the oscilloscope. The interference detected from the Fizeau interferometer is shown in Fig.4.5

4.1.5 SNR Measurements for the Fizeau and Michelson Interferometers

Optical Coherence Tomography requires a high Signal to Noise Ratio (SNR) within a system in order to create high quality two-dimensional images of biological tissue. Most OCT systems use SLDs as the light source⁷ which can have relatively low power outputs and the reflectivity of tissue is very low <1% which means that there is very little power in the system to achieve good signals. By carrying out very simple measurements on the Michelson and Fizeau interferometers a comparison between the SNR of the two systems can be made.

A more in-depth analysis of the signal to noise ratio is given in Chapter 5 for both the Michelson and Fizeau configurations.

By using the interference fringes obtained for both interferometers the SNR can be calculated using the relationship⁸ in Eq.6.

$$\text{SNR} = \frac{\text{Signal}}{\text{Noise}} = 20 \log \left(\frac{\text{Fringe}}{\text{Noise}} \right) \qquad \text{(pk volts)} \qquad (6)$$

SNR values were calculated for the Fizeau and the Michelson interferometer from the experimental set-up in section 4.1.3 and 4.1.4 the values are shown in Table 4.1.

Table 4.1 Table showing SNR comparison between the Fizeau and Michelson interferometers

Interferometer	Peak Signal (V)	Peak Noise (V)	SNR / dB
Michelson	98 ± 5	4 ± 1	28 ± 3
Fizeau	33 ± 5	4 ± 1	18 ± 3

From the initial measurements it can be seen that there is a decrease in SNR when using the Fizeau interferometer compared to the Michelson and this is of the order 10dB for the measurements carried out in this chapter using the same source and detector for each set-up.

The signal to noise ratios for these interferometer comparisons appear relatively low for OCT imaging, as the equipment used was not chosen to optimise the SNR for this experiment but just to investigate the relationship between the SNR of a Fizeau and Michelson arrangement.

Although the Fizeau interferometer exhibits a relatively low SNR value compared to the Michelson, this is for measurement conditions performed outside the body where polarisation effects in the fibre (which are likely to decrease the SNR in a Michelson interferometer) are not likely to be as significant.

The initial configuration for the Fizeau interferometer is not ideal for OCT so an investigation on how to improve the SNR is carried out in chapter 5.

4.2 One and Two Dimensional Imaging with the Fizeau Interferometer

Obtaining one-dimensional OCT profiles to look at sub-surface tissue is a good way to detect any pre-cancerous changes occurring in the tissue. It is known that with most cancers and indeed is the case with esophagus cancer where changes occur within the surface tissue layers⁹. The cells within the epithelial layer change, and as they develop start to invade the submucosa layer and there appears to be a thickening of epithelial tissue.

Normal esophagus tissue¹⁰ has an epithelial layer $\sim 50\mu\text{m}$ and as cancer starts to develop this thickness can increase to $\sim 70\mu\text{m}$ during a pre-cancerous condition known as dysplasia

where there is an abnormality in cell development of the size and shape of the normal cells. As these abnormal cells reproduce, cancer is developing and the epithelial layer can increase to $\sim 120\mu\text{m}$. As shown in the beginning of this chapter the signals from OCT scans have an axial resolution capable of measuring these small micrometer changes. Therefore, just by taking single scans from the esophagus tissue, it is in principle possible to determine thickening in the epithelial layer that may point towards cancer developing and highlight areas suitable for biopsy.

While waiting for equipment to arrive a side study was carried out looking into developing a software program in LabView to create two dimensional OCT images.

By combining adjacent one-dimension profiles a two-dimensional map/image of the tissue layers and their thicknesses can be created¹¹ making more information available about changes within the esophagus.

Using the Fizeau interferometer set-up described in section 4.1.3 both the mirrors used for the sample arm within the Fizeau sensing head and the Michelson scanning mirror were mounted on translation stages which were connected and controlled using the LabView platform. By installing the appropriate software these translation stages could be linked to a LabView program and automatically driven.

Software was also installed to enable measurements being read into the LabView program from the PIN photodiode.

A program was written in LabView which recorded and stored a one-dimensional profile from the Fizeau interferometer, each time the mirror in the processing Michelson interferometer moved backwards and forwards in one scan in order to match the path length of the sample mirror.

Once a scan had been performed the mirror within the Fizeau sensing interferometer moved a small distance laterally and then another scan was performed. The program written is shown in Appendix F and allows the user to vary the speed of translation stage motions, incremented lateral motion and number of scans performed.

Once a scan had been performed, the LabView program assigned a grey scale value for the intensity measured by the detector ranging from black for little signal and white for a high signal, therefore any interference fringes would show up as very bright on the image.

This data was then transported to a file written within LabView and accessed using Excel. LabView also had tools to view the data as it was being processed and the Diagram function in LabView (shown in Appendix F) allowed an oscilloscope plot to be shown as the data was being read. It also shows a two-dimensional intensity graph, which plotted out each profile in turn creating a two-dimensional image.

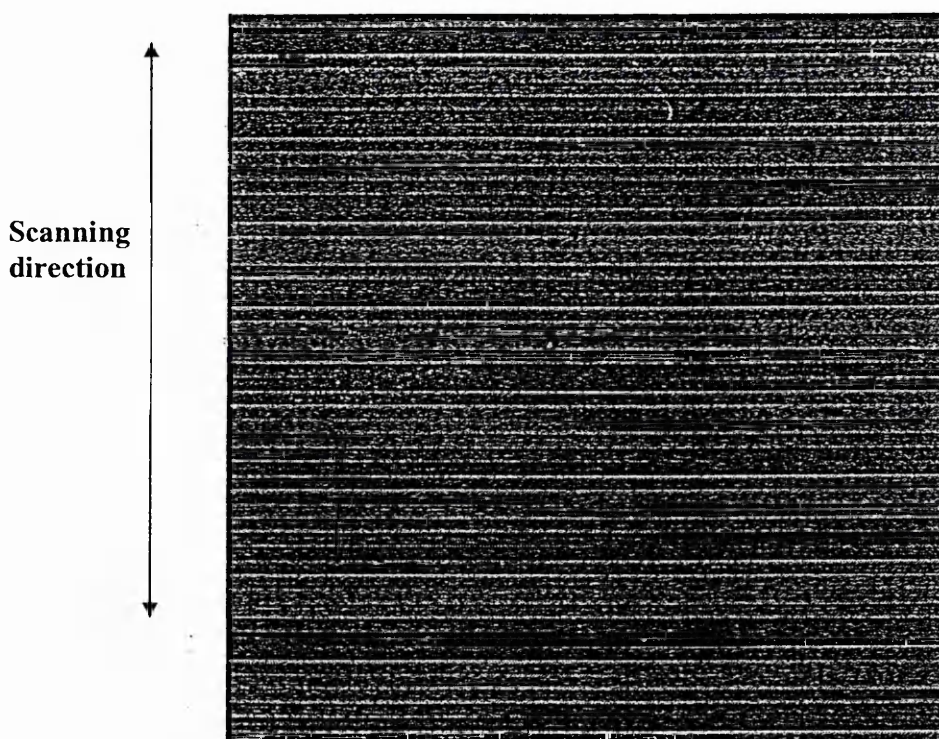


Fig.4.6 Showing a 2-dimensional image of the signal from a HeNe laser

Initial tests of the two-dimensional imaging program were carried out using a HeNe laser in order to test the LabView program and make sure it was working properly, as the coherence

length of the source was relatively long and covered the scanning range of the translation stage.

Fig.4.6 Shows the image obtained using the program and a HeNe laser, it shows a ‘stripy’ effect showing peaks and troughs for the fringes within the interference signal, because the HeNe source has such a long coherence length, the interference signal can be seen over the whole traverse of the translation stage.

An SLD was then measured using this approach and the two-dimensional image can be seen in Fig.4.7. This shows a single high intensity strip corresponding to the interference signal at the detector when the path length between the sample and reference arm is equal or within the coherence length of the source. Because of the short coherence length the signal is displayed as a thin strip of around $\sim\mu\text{ms}$ this is shorter than the difference in length from the different reflecting sites within a tissue sample and so the signals can easily be resolved.

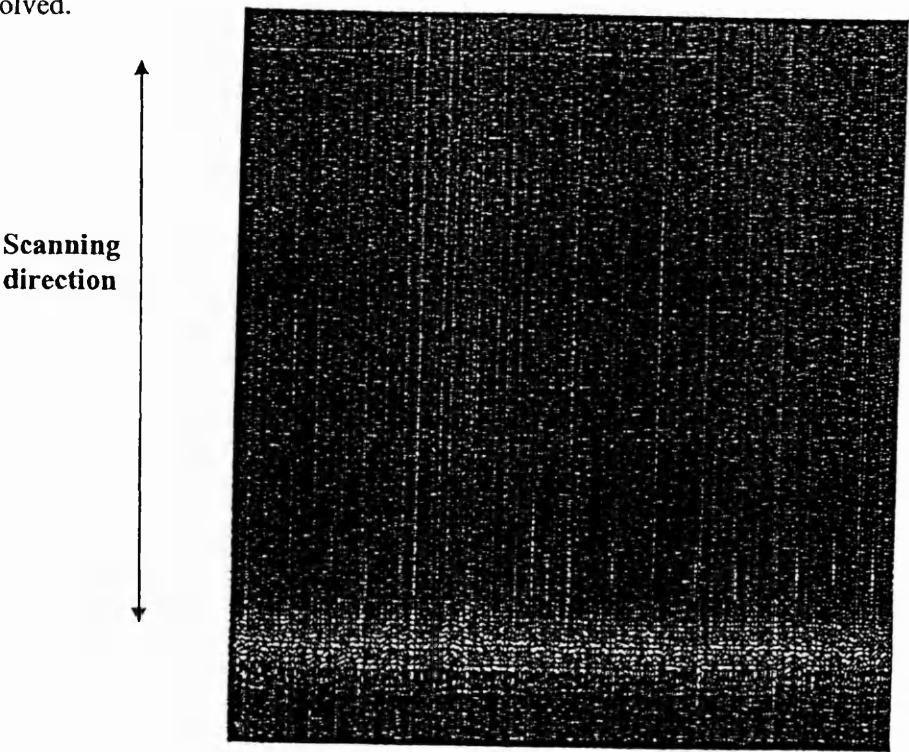


Fig.4.7 Showing a 2-Dimensional image using an SLD

4.3 Summary

This chapter covered the theory of Low Coherence Interferometry (LCI) and has derived the interference signals from both the Michelson and Fizeau interferometers. Theoretically it is clear to see from equations 1 and 2 that when using the same set of equipment the Fizeau interferometer will give an SNR lower than the Michelson.

By setting up both interferometer configurations (Michelson and Fizeau) it was shown experimentally that the Michelson gave a higher SNR of ($\sim 10\text{dB}$) than the Fizeau using the same set of equipment.

As a side study an investigation was carried out into two dimensional OCT imaging, showing how this has its advantages compared to one dimensional profiles. A program was created using LabView which allowed one dimensional scans to be collected and combined to give a two dimensional image. Two dimensional images created from reflections from the surface of a mirror were collected. Results from a HeNe laser and a SLD were compared to show the difference in coherence lengths and to test the ability of the program. Both sources showed the expected interference images and the signals from the SLD source were identified in the same position for all profiles taken.

The next chapter investigates techniques to increase the SNR of the Fizeau interferometer by looking at different configurations and optical components to prevent power loss and improve the overall signal to noise ratio within the system. It also looks at ways to improve the standard Fizeau configuration by changing parameters within the system such as splitting ratios of the directional couplers and the reflectivity of the light reflected back from the fibre end within the sensing interferometer.

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Chapter 5

Optimal Interferometer Designs for Optical Coherence Tomography (OCT)

5.1 Interferometers suitable for OCT Imaging

Optical Coherence Tomography (OCT) is a high resolution imaging technique that uses low coherence interferometry to provide information from below the surface of semi-transparent materials and is thus able to create three-dimensional images of the material structure¹. The technique has been employed to investigate composite materials and in particular biological tissue¹. Interference signals are detected as light backscattered from different refractive index interfaces within the media e.g. tissue samples, are combined with a reference beam (Chapter 1).

Most OCT arrangements use a fibre optic based Michelson interferometer² as the basis of the optical system as shown in Fig.5.1. Here light from a low coherence source is split

SNR is important to achieve high quality images and it is necessary therefore to minimise the loss of optical power within the OCT system.

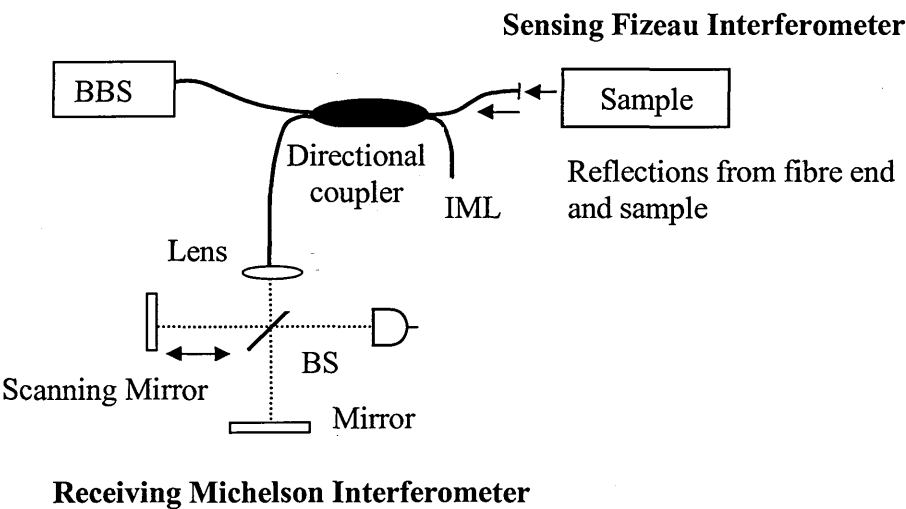


Fig.5.2 Showing a Fizeau Interferometer

Another suitable configuration for an OCT system uses a Fizeau interferometer⁶ shown in Fig.5.2. Here light from a low coherence source is split 50/50 ($D=0.5$) by a DC, 50% propagates down a sample arm to the Fizeau sensing head and 50% travels down the other arm immersed in Index Matching Liquid (IML), which prevents reflections back down the fibre. The light travelling down the sample arm enters the Fizeau sensing head where approximately 4% of the light at the end of the fibre is Fresnel reflected back down to the DC (this 4% is for air-glass interface other reflective values can be produced with additional fibre coating). The light is then guided to a receiving interferometer (in the case of Fig.5.2 a Michelson interferometer) and this acts as the reference beam for the interferometer. The light that is not Fresnel reflected acts as a sample beam and is focussed onto the tissue and back scattered from the different microstructures within the sample and coupled back into the fibre. The light is then guided to the processing interferometer. Interference fringes are observed at the detector when the path length of the Fresnel

reflected light matches that of the tissue reflecting sites to within the coherence length of the source. A reference mirror within the processing interferometer is scanned axially so that all the interference signals, corresponding to the reflections from the tissue, are observed at the detector, where the intensity of the signal is derived in Appendix D.

Two Mach-Zehnder configurations⁷ were also investigated to look for improvements in SNR. The First Mach Zehnder configuration is the Mach Zehnder 'a' which is shown in Fig. 5.3. This configuration uses a broadband source which is split by a directional coupler with one exit arm leading to a circulator, this transmits the light from port one of the circulator onto the sample, which is then reflected by the different refractive index interfaces within the tissue and re-enters the circulator through port two, this then exits through port three of the circulator. The light exiting through the other arm of the directional coupler passes through a delay line, allowing a path length variance between the sample and reference beam. Both arms recombine at a second coupler where interference can be detected on exit from both arms of the coupler when the path length of the sample and reference arm is equal or to within the coherence length of the source. The signals can be detected by two balanced receivers leading to balanced detection.

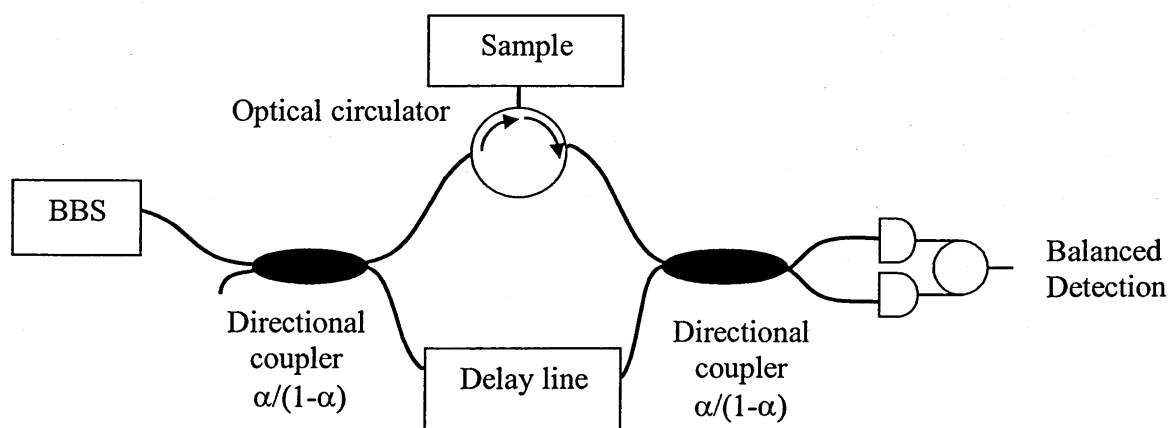


Fig.5.3 Mach Zehnder 'a' configuration with two directional couplers and a circulator.

The second Mach Zehnder configuration, the Mach Zehnder 'b' configuration is shown in Fig.5.4

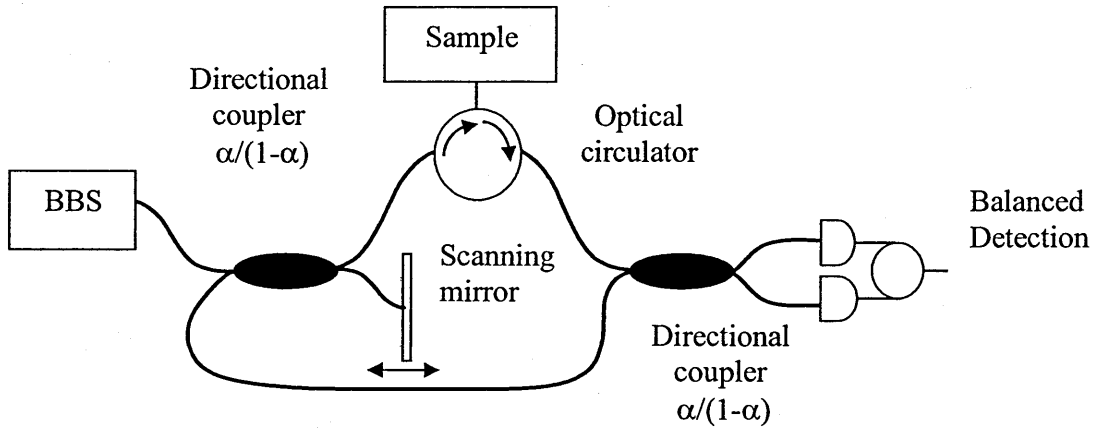


Fig.5.4 Mach Zehnder 'b' configuration using two directional couplers a circulator and balanced detection.

This is the same design as the Mach Zehnder 'a' configuration with a variance in the reference beam path. Light in the sample arm follows the same path as the Mach Zehnder 'a' configuration, however light in the reference arm is reflected by a scanning mirror or other kind of reflecting device (see chapter 2) and re-enters the first coupler before recombining with the sample beam in the second coupler. Interference signals are detected using balanced detection as in the Mach Zehnder 'a' configuration.

All chosen interferometer configurations (Michelson, Fizeau and Mach Zehnder) were analysed using a set of constant conditions to give a good comparison between all systems. The Fizeau and Mach Zehnder configurations showed improved SNR compared with the conventional Michelson interferometer. The Fizeau configuration not only was shown to have comparable SNR to the Michelson and Mach Zehnder with an added advantage of eliminating phase and polarisation effects in the sample and reference arm, leading to a possible decrease in SNR.

5.2 Advantages and Disadvantages of Interferometer Designs

Optical Coherence Tomography is a suitable technique for *in vivo*, endoscopic applications^{8,9} allowing high-resolution, depth images to be produced effectively without any harmful, adverse effects to patients as it uses light as its imaging source.

The Michelson Interferometer has proved to be successful for *in vitro* imaging, although it may not be the best configuration for endoscopic applications. During an endoscopy the interferometer sample arm is inserted down the oesophagus. This change in environment may cause phase changes between the input of the two interferometer arms which in turn may alter the polarisation state of the light and will affect the visibility of the interference signal⁹. The Fizeau interferometer eliminates polarisation and phase changes in the system that may arise due to environmental changes in the interferometer arms, by allowing light in the sample and reference arms to travel down the same optical fibres¹⁰.

Since the sensing interferometer is formed between the distal end of the fibre the tissue sample, effects on the interference signal due to bending of the fibre and temperature changes, which may induce phase and polarisation changes, do not affect the interference pattern observed at the detector. These effects on the signal are likely when performing endoscopic applications due temperature and strain that may affect the input in the fibre, due to the changing environment inside the body, hence this interferometer configuration offers an ideal set-up for endoscopic OCT.

Another advantage of the Fizeau arrangement is that the directional coupler and the processing interferometer can be housed separately from the sensing head when in clinical use.

In the Michelson configuration approximately 75% of the original power is lost, whereas the Fizeau configuration using a receiving Michelson interferometer⁶ almost 94% of the original optical power is lost. This is because a second beam splitter may be used in a receiving interferometer in the Fizeau set-up, and additional optical power is lost.

This chapter shows that by using alternative optical components, power conserving interferometers can be constructed that have such properties that a Fizeau interferometer configuration can achieve a higher SNR than that of a conventional Michelson arrangement and comparable to a Michelson optimised for power conservation.

5.3 Theory for Calculating the Signal to Noise Ratio

Assuming a dispersionless system, the total photocurrent I_{tot} for the Michelson⁷ and Mach-Zehnder interferometers⁷, derived in Appendix C, is given by Eq.1

$$I_{\text{tot}} = \rho [P_r + P_s + 2\sqrt{P_r P_s} \cos(k_0 \Delta l)] \quad (1)$$

For the Fizeau sensing head, the equivalent signal is derived in Appendix D and given by Eq.2

$$I = \rho \left\{ 2P_r + 2P_s + \left[\begin{aligned} & (4\sqrt{P_r P_s}) \cos(k_0 X_1) + (2P_r + 2P_s) \cos(k_0 X_2) + \\ & (2\sqrt{P_r P_s}) \cos(k_0 (X_2 - X_1)) + (2\sqrt{P_r P_s}) \cos(k_0 (X_1 - X_2)) \end{aligned} \right] \right\} \quad (2)$$

where U is the responsivity of the detector, P_r and P_s are the reflected powers from the reference and sample arms which are coherent with the reference beam. P_{r1} , P_{r2} , P_{s1} and P_{s2} are the powers in the reference and sample arms travelling down arm 1 and 2 of the receiving Michelson interferometer. P_s contains a noise component P_x which is the reflected power from the sample arm which is incoherent with the reference beam k_0 is the wave number of the centre wavelength of the source and ' l ' is the path length difference between the sample and reference arm.

For the Michelson and Mach-Zehnder configurations the photocurrent contains a DC background photocurrent I_{dc} given by:

$$I_{dc} = \rho(P_r + P_x + P_s) \quad (3)$$

and an AC interference signal given by:

$$I_s = 2\rho\sqrt{P_r P_s} \cos(k_0 \Delta l) \quad (4)$$

For the Fizeau interferometer, where $P_{r1}=P_{r2}=P_r$ the photocurrent is given by:

$$I_{dc} = \rho(2P_r + 2P_s + 2P_x) \quad (5)$$

and the AC interference signal is given by:

$$I_{AC} = \rho \left[\begin{aligned} & \left(4\sqrt{P_r P_s} \right) \cos(kX_1) + (2P_r + 2P_s) \cos(kX_2) + \\ & \left(2\sqrt{P_r P_s} \right) \cos(k(X_2 - X_1)) + \left(2\sqrt{P_r P_s} \right) \cos(k(X_1 - X_2)) \end{aligned} \right] \quad (6)$$

which includes all interferometric signals within the interferometer, the signal from the reflected light from the tissue sample, used for OCT imaging is given by:

$$I_s = 2\rho\sqrt{P_r P_s} \cos(k\Delta X) \quad (7)$$

The total photocurrent variance (V_{tot}^2) is a summation of the various noise variances within the system, such as due to shot noise (V_{sh}^2), excess noise (V_{ex}^2) and receiver noise (V_{rec}^2).

Shot noise arises due to the random arrival of photons from monochromatic light and creates a photocurrent with variance:

$$\sigma_{\text{Sh}}^2 = 2qI_{\text{dc}}B \quad (8)$$

where q is the electronic charge and B is the electronic bandwidth.

Excess noise occurs with the random arrival of photons from broadband, incoherent light¹¹ and is given by:

$$\sigma_{\text{ex}}^2 = (1 - V^2)I_{\text{dc}}^2 \frac{B}{\Delta} \quad (9)$$

where V is the degree of polarisation and $\Delta\nu$ is the effective linewidth of the source¹¹ given by:

$$\Delta\nu = \left[\frac{\pi}{2\ln(2)} \right]^2 \frac{c\lambda_{\text{FWHM}}}{\lambda_0^2} \quad (10)$$

c is the free space speed of light, λ_{FWHM} is the full width at half-maximum wavelength of the source and λ_0 is the centre wavelength of the source.

Receiver noise occurs due to thermal noise within the detector and is usually specified by the manufacturer for commercial devices.

Balanced detection can be used to increase the SNR of an interference signal. Balanced detection uses two detectors and subtracts the photocurrent in each detector. This cancels out the DC noise components and it increases the AC component by a factor of two.

If balanced detection is used, excess photon noise can be considered negligible. However a component of the excess noise does remain called the beat noise (V_{be})⁷ and is given by:

$$\sigma_{\text{be}}^2 = \frac{(1 + V^2)p^2P_rP_xB}{\Delta\nu} \quad (11)$$

Also, the total signal photocurrent in balanced detection is the sum of the photocurrent in both detectors and is given by Eq.12 For the Michelson, Mach-Zehnder and Fizeau configurations.

$$I_s^2 = 4\rho^2 P_r P_s \quad (12)$$

The SNR is expressed in dB and given by Eq.13

$$\text{SNR} = 10\log\left(\frac{\langle I_{AC}^2 \rangle}{\sigma_{\text{tot}}^2}\right) \quad (13)$$

The SNR for three configurations of interferometers were investigated, the conventional Michelson configuration the Fizeau configuration and the previously reported Mach Zehnder configurations⁷. A number of interferometer configurations were investigated to increase the SNR of the system. This study follows previous work done by Rollins *et al*⁸, who investigated Michelson and Mach-Zehnder configurations. Additional designs of these interferometers will be introduced along with a group of Fizeau interferometers, which have been shown to be an attractive design for OCT imaging.

The parameters throughout used within this study will be identical for all configurations to make a comparison. The chosen parameters were used by looking at typical values expected from an OCT system. Therefore $R_s=0.01$ (typical for 1% reflection from tissue), $R_r = 0.01$ (to match the sample arm reflectivity as this give the best SNR, see Fig.5.5, Fig.5.6, Fig.5.7 and Fig.5.8), $R_x=0.0005$, $U=0.95$ and the receiver noise current = $2\text{pA/Hz}^{1/2}$ these values were used because they are typical of an OCT system⁸. The optical power of the source was assumed to be 20mW with a 1300nm-centre wavelength and a 50nm bandwidth. The source was unpolarised and $B=1\text{MHz}$. Transmission through the circulator, $T_c=0.85$ again these are predicted typical values⁸.

5.4 Theoretical Calculation of Signal to Noise Ratio

5.4.1 Reference arm reflectivity

The SNR of an interference signal was show in section 5.3 to be dependent on the power in the reference and sample arms and therefore be dependent on the reference arm reflectivity. To investigate how important this relationship is a graph was plotted of SNR as a function of reference arm reflectivity for both interferometer designs. The Total SNR was plotted along with the individual noise contributions, Fig.5.5 shows SNR as a function of reference arm reflectivity for the Michelson interferometer, Fig.5.6 shows SNR variation in the Fizeau interferometer and Figs.5.5 and 5.6 the variation in SNR for the two Mach Zehnder configurations, Fig.5.5 the Mach Zehnder ‘a’ and Fig5.6 the mach Zehnder ‘b’ configuration.

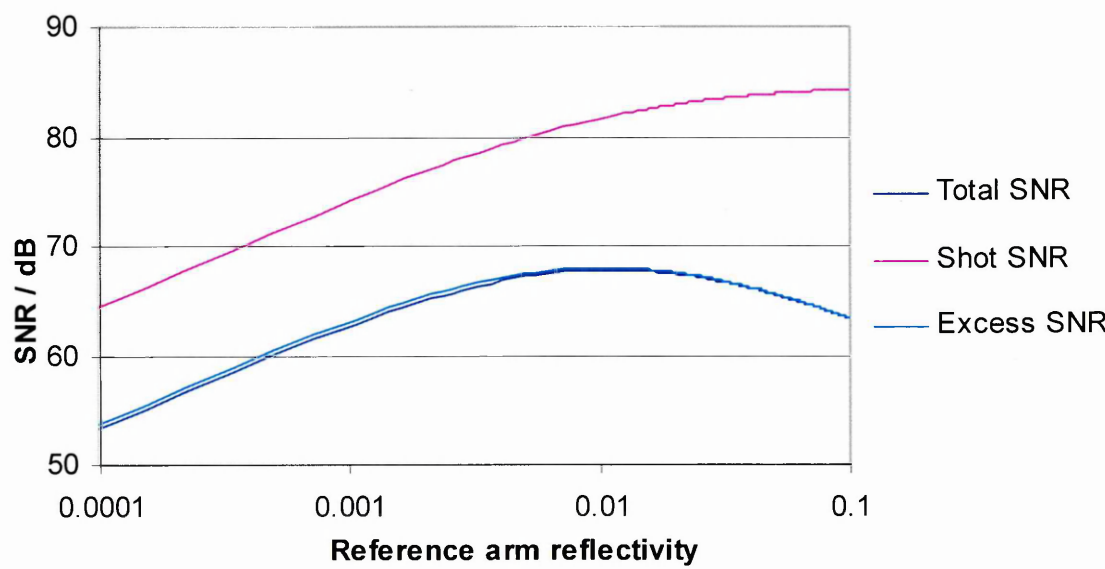


Fig.5.5 Variation of SNR with reference arm reflectivity for the Michelson interferometer.

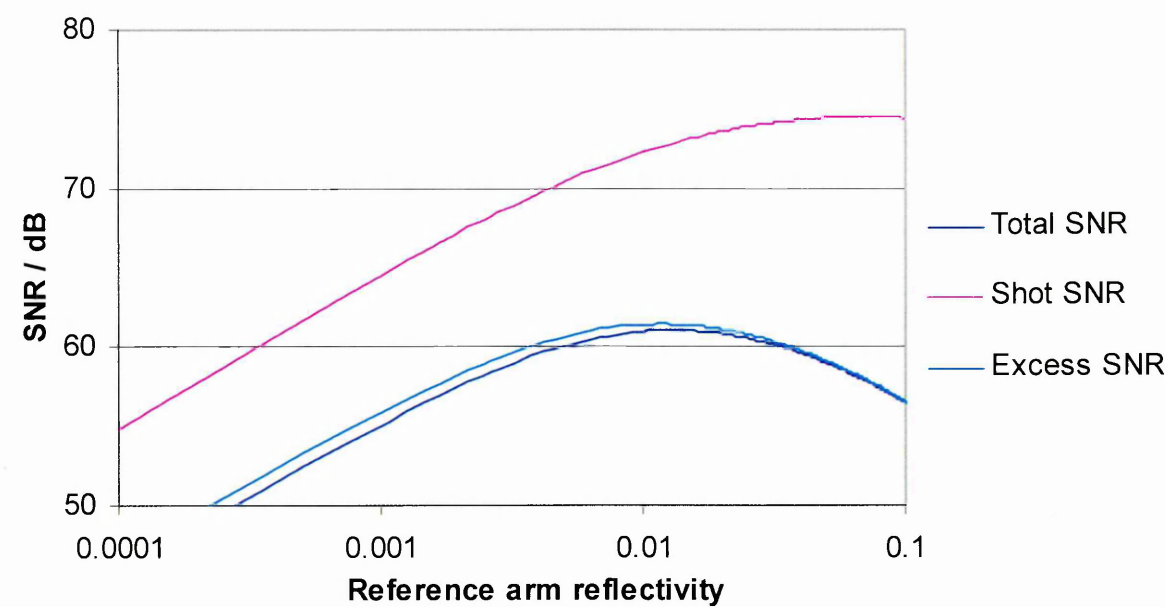


Fig.5.6 Variation of SNR with reference arm reflectivity for the Fizeau interferometer.

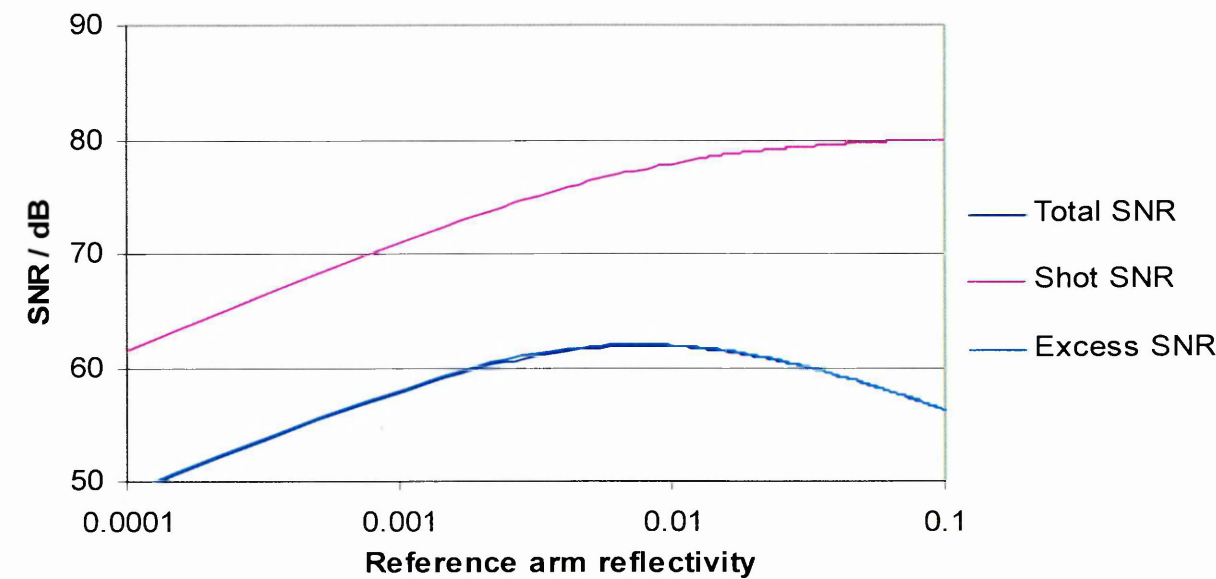


Fig.5.7 Variation of SNR with reference arm reflectivity for the Mach Zehnder ‘a’ interferometer.

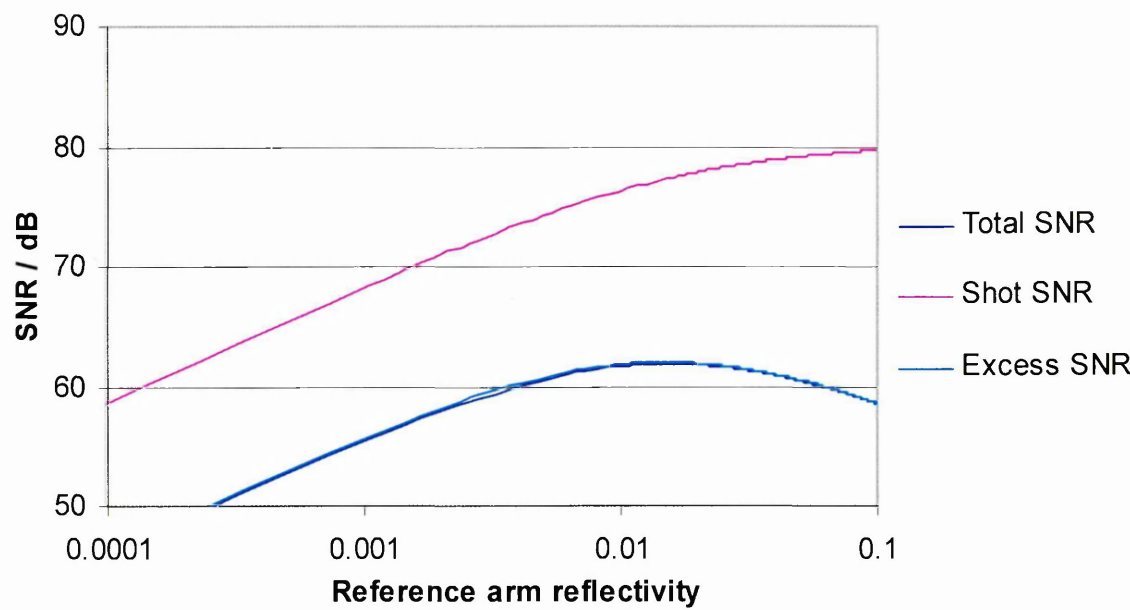


Fig.5.7 Variation of SNR with reference arm reflectivity for the Mach Zehnder ‘b’ interferometer.

The total variation of noise between both interferometers is shown in Fig 5.5 and clearly shows that the Michelson interferometer always gives a higher SNR than the other configurations.

Fig.5.8 also shows that a maximum SNR is achieved at a reflectivity of 0.01 for all the interferometers which matches the 1% reflectivity for the tissue sample. Therefore the value of the reference arm reflectivity will be 0.01 for all calculations, which is comparable to the reflectivity of the sample arm. Therefore this suggests for the maximum SNR, the reference and sample arm reflectivities should be matched.

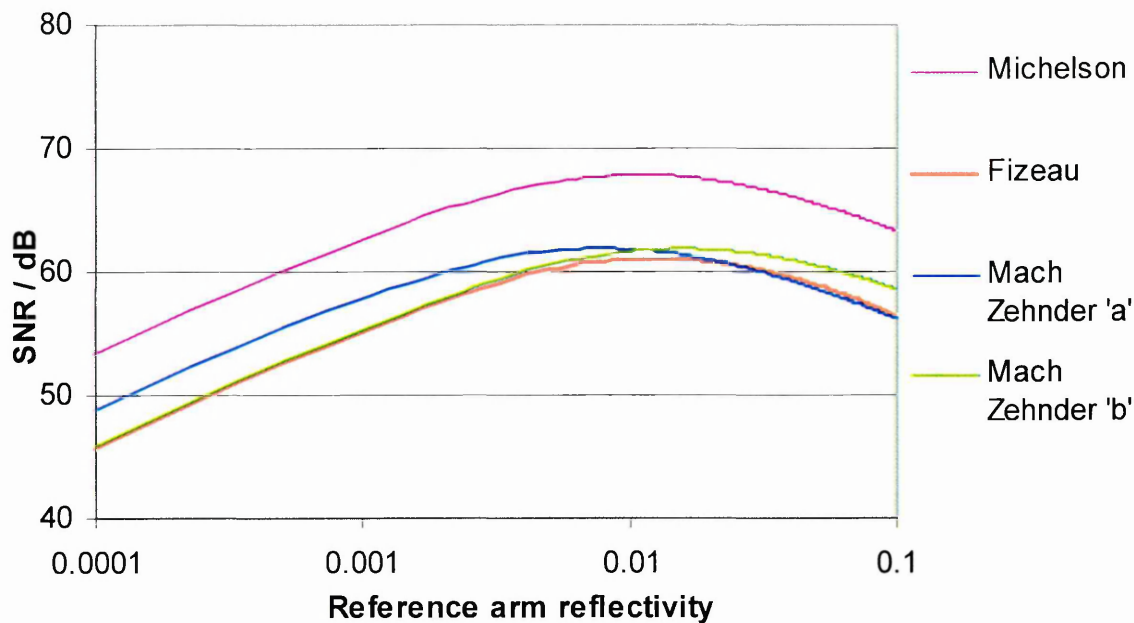


Fig.5.8 Comparison of SNR variation with reference arm reflectivity for the Michelson, Fizeau and two Mach zehnder configurations.

5.5 Optimal Interferometer Configurations for the Michelson interferometer

All Interferometer configurations in this section use the Michelson design with added components which are unbalanced couplers, circulators and balanced detectors, to increase the SNR

The power equations are given by the equations below and show the power of coherent light in the sample arm P_s , the power of light in the reference arm P_r and the power of the incoherent light in the sample arm P_x calculated at the detector:

$$P_r = P_0 R_r \alpha (1 - \alpha), \quad P_s = P_0 R_s (1 - \alpha) \alpha, \quad P_x = P_0 R_x (1 - \alpha) \alpha$$

5.5.1 Michelson interferometer using an unbalanced coupler⁷, Fig.5.9

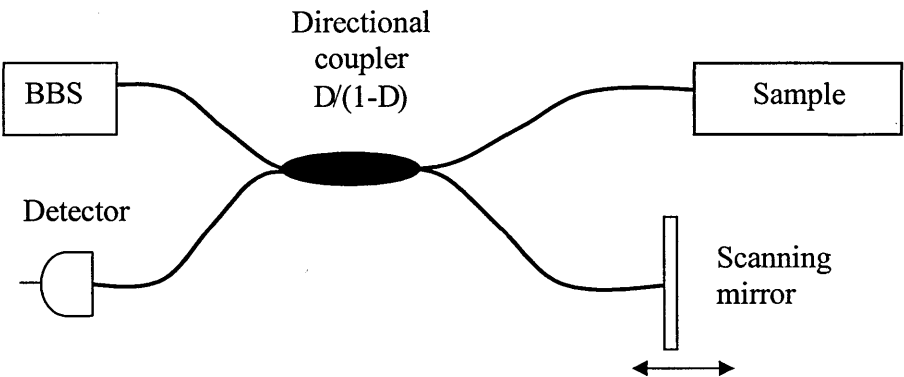


Fig.5.9 Michelson interferometer using an unbalanced coupler. BBS=broadband source, D=coupler splitting ratio, D=0.5 for balanced coupler.

The SNR is a function of coupler splitting ratio and using Equation 14 gives a maximum value of 68dB with a 0.5 splitting ratio.

$$SNR=10\log\left[\frac{2\rho^2P_rP_s}{\sigma_{rec}^2+2qB\rho^2(P_r+P_s+P_x)+\frac{B\rho^2}{\Delta\nu}(1-V^2)(P_r+P_s+P_x)^2}\right] \quad \text{Eq.14}$$

5.5.2 Michelson interferometer using an unbalanced coupler and circulator⁷, Fig.5.10

A circulator is used within this configuration and is a device that allows light to be collected and recirculated within a system, this is especially useful when light is being lost through the exit arm of a directional coupler (Chapter 6). Light through the entrance arm of

the circulator can now be recirculated and collected at the detector in a conventional Michelson interferometer.

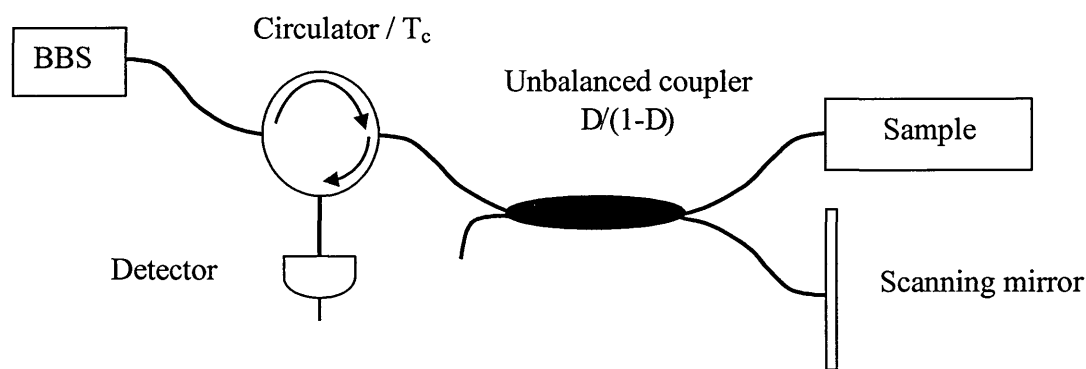


Fig.5.10 Michelson interferometer using an unbalanced coupler and circulator.

BBS=broadband source, D=coupler splitting ratio, D=0.5 for balanced coupler, T_c = circulator transmission factor.

The power equations are given by where P_s, P_r and P_x are defined in 5.5.1 :

$$P_r = P_0 R_r T_c^2 \alpha^2, \quad P_s = P_0 R_s T_c^2 (1 - \alpha)^2, \quad P_x = P_0 R_x T_c^2 (1 - \alpha)^2$$

The maximum SNR is calculated using Equation 14 to be 68dB with splitting ratio 0.5 shown in Fig.5.11

The SNR becomes a function of the splitting ratio for both interferometer designs Fig.5.11 shows this relationship and identifies where the maximum SNR occurs.

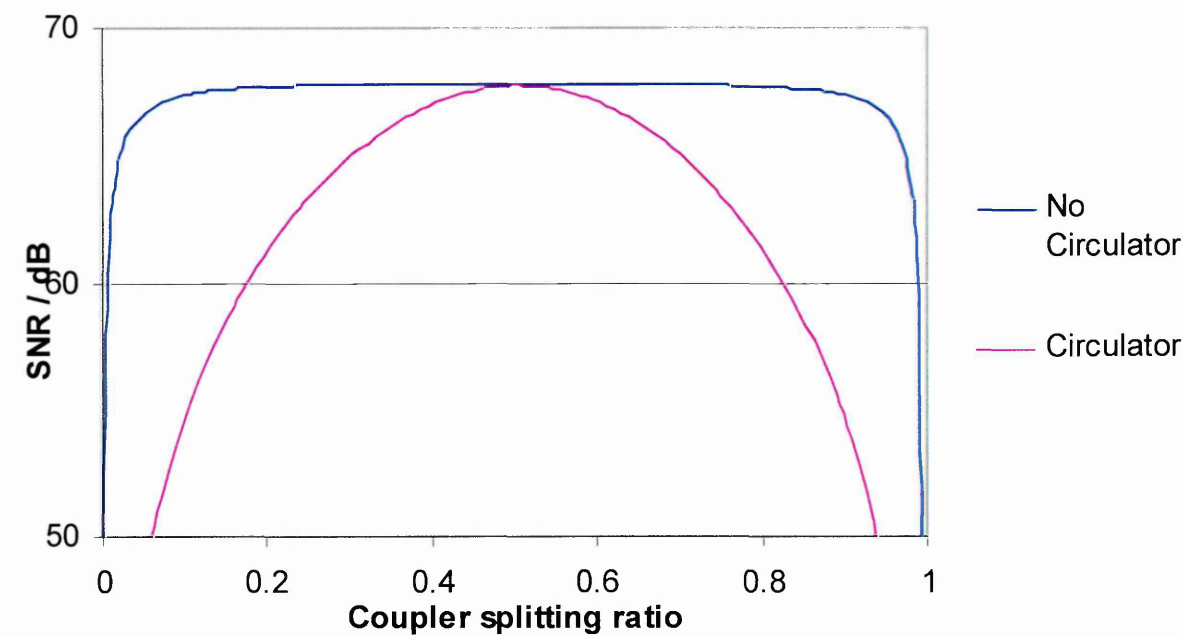


Fig.5.11 Graph showing SNR as a function of splitting ratio for two Michelson interferometer designs.

5.5.3 Michelson interferometer using an unbalanced coupler and circulator with balanced detection⁷ (Fig.5.12)

The Michelson interferometer now uses the light recirculated by the circulator and the coupler to supplement the balanced detection

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = P_0 R_r T_c^2 \alpha^2, \quad P_s = P_0 R_s T_c^2 (1 - \alpha)^2, \quad P_x = P_0 R_x T_c^2 (1 - \alpha)^2$$

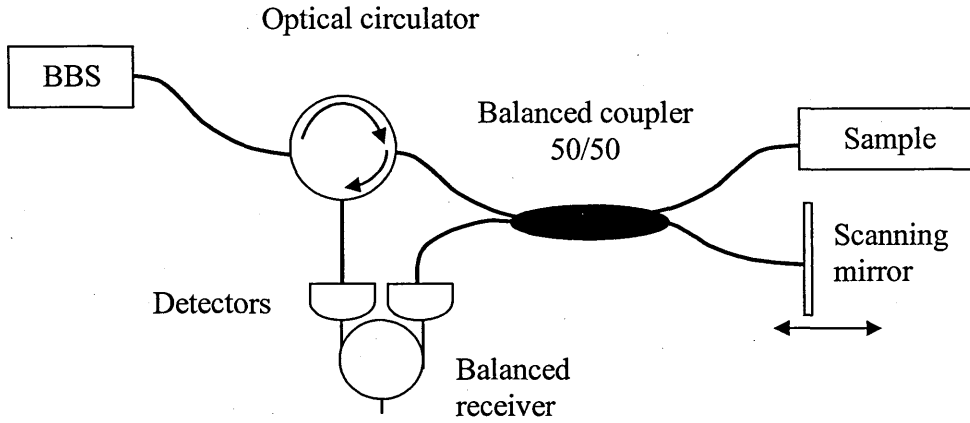


Fig.5.12 Michelson Interferometer using and unbalanced coupler, circulator and balanced detection. BBS=broadband source, D =coupler splitting ratio, $D=0.5$ for balanced coupler, T_c = circulator transmission factor.

The SNR for this configuration is calculated from Equation 15 to be 81dB.

$$\text{SNR} = 10 \log \left[\frac{4\rho^2 P_r P_s}{\sigma_{\text{rec}}^2 + 2qB\rho^2 (P_r + P_s + P_x) + \frac{BP_r P_x \rho^2}{\Delta\nu} (1 - V^2)} \right] \quad \text{Eq.15}$$

The SNR of all the Michelson interferometer designs were calculated using Equations 14 and 15 and with the system parameters defined in section 5.3. The calculated values are shown in Table 5.1

The maximum SNR calculated from the Michelson configurations is 81dB using the configuration with balanced coupler with circulator and balanced detection. This is an increase of 13dB compared to the conventional Michelson design. This configuration ensures that the light which is usually lost as it exits through the coupler is recirculated to the detector using an optical circulator to the balanced detector, and a more power is received from the interfering signals.

Table.5.1 showing SNR for all Michelson Interferometer designs

Michelson Interferometer Design	SNR
Michelson interferometer using an unbalanced coupler	68 dB
Michelson interferometer using an unbalanced coupler and circulator	68 dB
Michelson interferometer using an unbalanced coupler and circulator with balanced detection	81 dB

There is little difference in SNR when using the circulator and unbalanced coupler as the light in the interferometer is split between the two coupler arms and the SNR will be maximum with equal splitting ratio.

5.6 Fizeau Interferometer Configurations

All Interferometer configurations in this section use the Fizeau design with added components (unbalanced coupler,circulator and balanced detectors) to maximise power and increase the SNR

5.6.1 Fizeau interferometer using an unbalanced coupler (Fig.5.13)

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1

$$P_r = \frac{P_0 R_r \alpha (1 - \alpha)}{4}, \quad P_s = \frac{P_0 R_s \alpha (1 - \alpha)(1 - R_r)^2}{4}, \quad P_x = \frac{P_0 R_x \alpha (1 - \alpha)(1 - R_r)^2}{4}$$

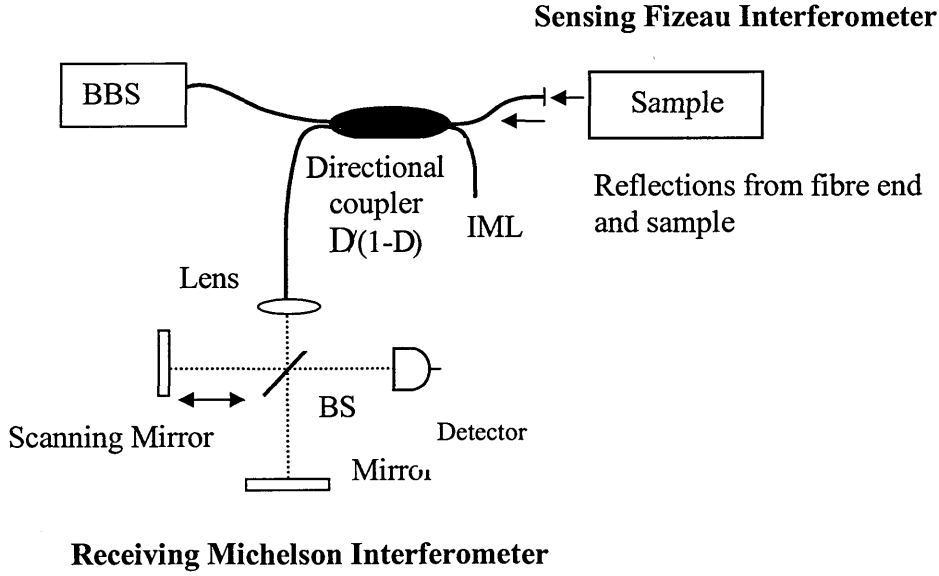


Fig.5.13 Showing a Fizeau interferometer using an unbalanced coupler. BBS=broadband source, D =coupler splitting ratio, $D=0.5$ for balanced coupler, BS=beam splitter, IML=index matching liquid.

The SNR for a the Fizeau interferometer with receiving Michelson interferometer is a function of the coupler splitting ratio, shown in Fig.5.13 and gives a maximum value of 61dB at a 0.5 splitting ratio calculated using Equation 16.

$$\text{SNR} = 10 \log \left[\frac{2\rho^2 P_r P_s}{\sigma_{\text{rec}}^2 + 2qB\rho^2 (2P_r + 2P_s + 2P_x) + \frac{B\rho^2}{\Delta\nu} (1 - V^2) (2P_r + 2P_s + 2P_x)^2} \right] \quad \text{Eq.16}$$

5.6.2 Fizeau interferometer using an unbalanced coupler and circulator(Fig.5.14)

A Fizeau sensing interferometer with an unbalanced coupler and circulator is considered. Light is recirculated in the coupler entrance arm using a circulator, in order to conserve power and increase the SNR. The light then enters a processing Michelson interferometer on exiting from the circulator.

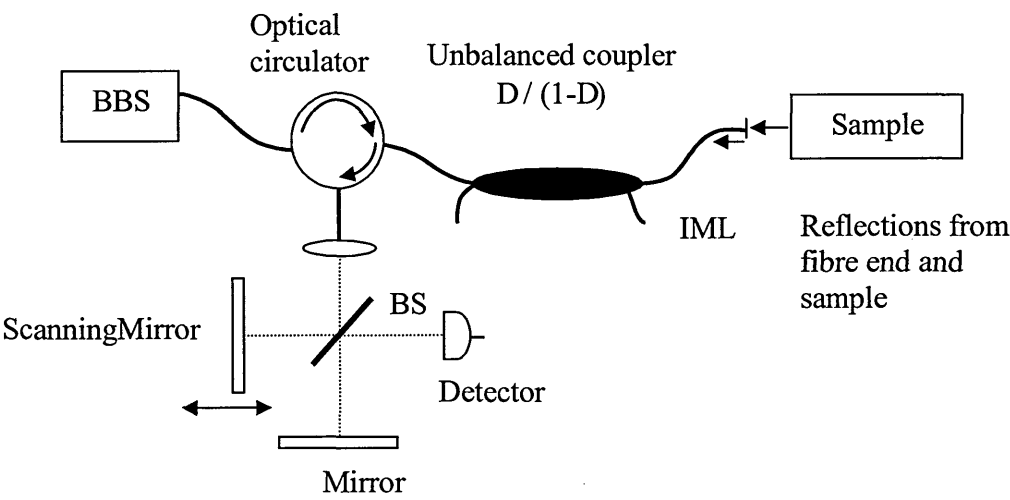


Fig.5.14 Fizeau interferometer using an unbalanced coupler and circulator. BBS=broadband source, D =coupler splitting ratio, $D=0.5$ for balanced coupler, BS=beam splitter, IML=index matching liquid.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r T_c^2 \alpha^2}{4}, \quad P_s = \frac{P_0 R_s T_c^2 \alpha^2 (1 - R_r)^2}{4}, \quad P_x = \frac{P_0 R_x T_c^2 \alpha^2 (1 - R_r)^2}{4}$$

The maximum SNR is 62dB with a splitting ratio approaching 1.0, calculated using equation 16.

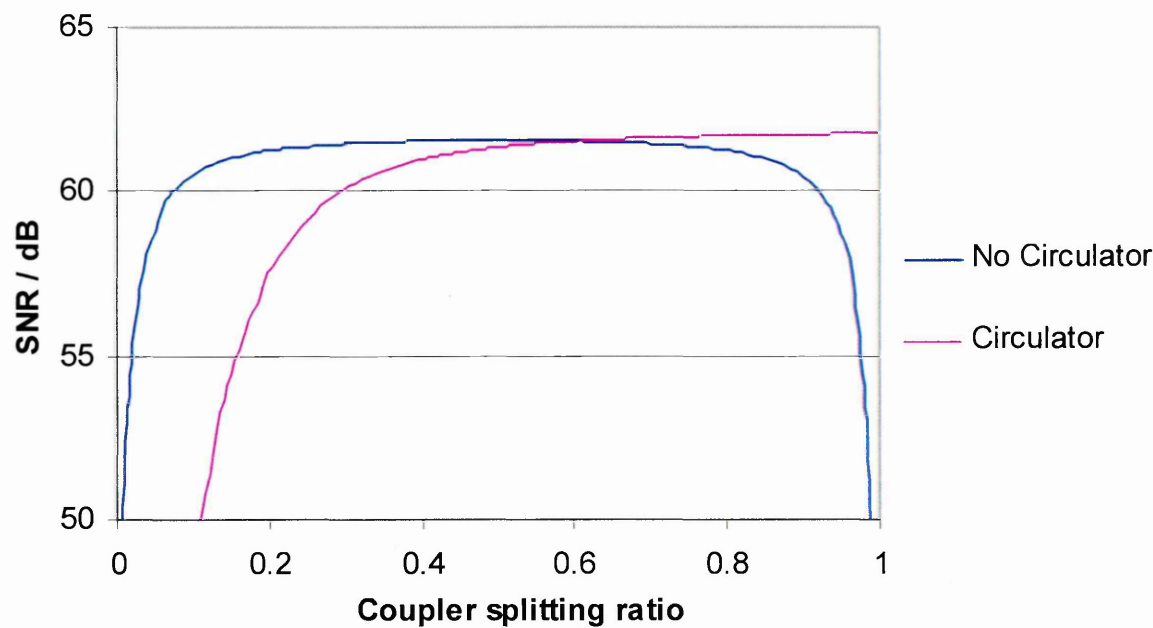


Fig.5.15 Graph showing SNR as a function of coupler slitting ratio for the configurations with and with a circulator.

The SNR becomes a function of the splitting ratio for both interferometer designs Fig.5.15 shows this relationship and identifies where the maximum SNR occurs.

5.5.3 Fizeau interferometer using a 3-port circulator(Fig.5.16)

A Fizeau sensing interferometer in which the coupler has been removed from the set-up and replaced by a single circulator. The light then enters a processing Michelson interferometer on exiting from the circulator.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r T_c^2}{4}, \quad P_s = \frac{P_0 R_s T_c^2 (1 - R_r)^2}{4}, \quad P_x = \frac{P_0 R_x T_c^2 (1 - R_r)^2}{4}$$

The SNR for this configuration is 62dB determine using equation 16.

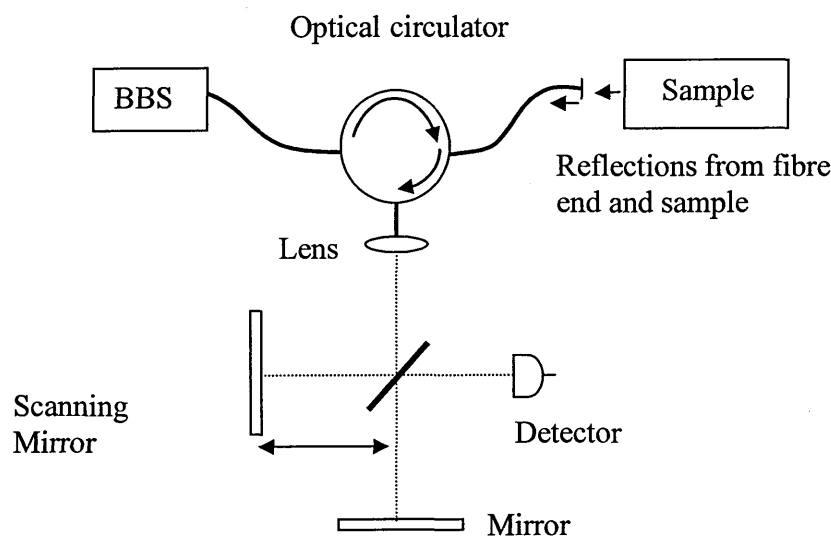


Fig.5.16 Fizeau interferometer using a 4-port circulator. BBS=broadband source, D=coupler splitting ratio, $D=0.5$ for balanced coupler, BS=beam splitter, IML=index matching liquid.

5.5.4 Fizeau interferometer using 2 x 3-port circulators and balanced detection (Fig.5.17)

A Fizeau sensing interferometer with two circulators one used in the Fizeau sensor head and one in the processing Michelson interferometer to supplement balanced detection.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r T_c^4}{4}, \quad P_s = \frac{P_0 R_s T_c^4 (1 - R_r)^2}{4}, \quad P_x = \frac{P_0 R_x T_c^4 (1 - R_r)^2}{4}$$

The SNR for this configuration is 78dB calculated using Equation 17.

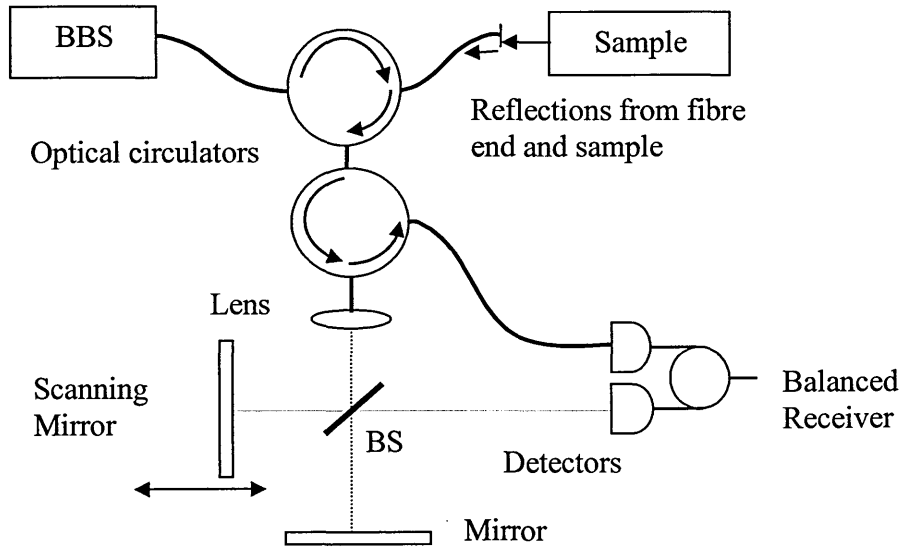


Fig.5.17 Fizeau interferometer using 2 x 3-port circulators and balanced detection

BBS=broadband source, , BS=beam splitter,

$$\text{SNR} = 10 \log \left[\frac{4\rho^2 P_r P_s}{\sigma_{\text{rec}}^2 + 2qB\rho^2 (2P_r + 2P_s + 2P_x) + \frac{B\rho^2 P_r P_x (1 - V^2)}{\Delta\nu}} \right] \quad \text{Eq.17}$$

5.5.5 Fizeau interferometer using 1 4-port circulators and balanced detection (Fig.5.18)

A Fizeau sensing interferometer with a single circulator to supplement balanced detection.

The power equations are given by:

$$P_r = \frac{P_0 R_r T_c^2}{4}, \quad P_s = \frac{P_0 R_s T_c^2 (1 - R_r)^2}{4}, \quad P_x = \frac{P_0 R_x T_c^2 (1 - R_r)^2}{4}$$

The SNR for this configuration is 79dB, calculated using Equation 17.

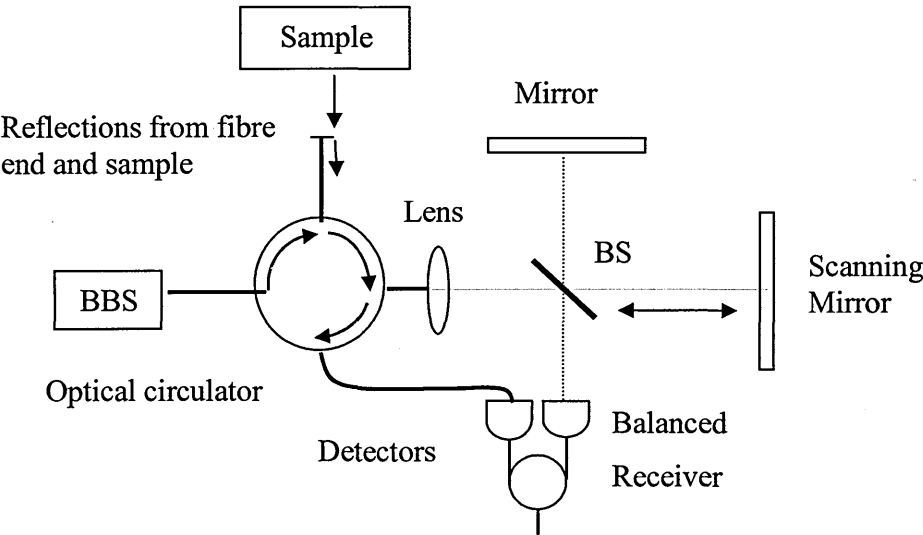


Fig.5.18 BBS=broadband source, D=coupler splitting ratio, D=0.5 for balanced coupler, BS=beam splitter, IML=index matching liquid.

5.5.6 Fizeau sensing interferometer using a receiving Fizeau interferometer using 2 3-port circulators (Fig5.19)

A Fizeau sensing head can be used with a receiving Fizeau interferometer using a 2 x 3-port circulator configuration to eliminate the need for couplers or beam splitters, as a lot of power is wasted on transmission through these devices. Fig.22 shows how the SNR becomes a function of processing interferometer reference arm reflectivity.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = P_0 R_r T_c^4 (1 - r_r)^2, \quad P_s = P_0 R_s r_r T_c^4 (1 - R_r)^2, \quad P_x = P_0 R_x r_r T_c^4 (1 - R_r)^2$$

Where r_r is the end of fibre reflection in the Fizeau processing interferometer

Maximum SNR is 59dB with end of fibre reflection at 0.4 shown in Fig.5.22, this is calculated using Equation 17.

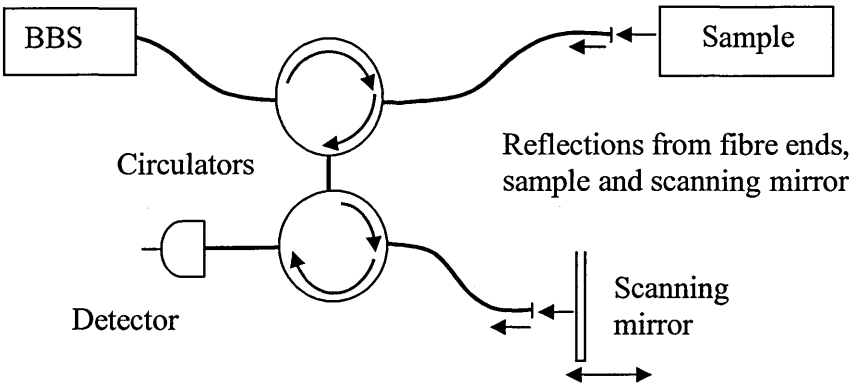


Fig.5.19 Fizeau sensing interferometer using a receiving Fizeau interferometer and 2 x 3-port circulators. BBS=broadband source, BS=beam splitter.

5.5.7 Fizeau sensing interferometer using a receiving Fizeau interferometer using 2 3-port circulators and a 50% transmitting mirror for balanced detection (Fig.5.20)

A Fizeau sensing head can be used with a receiving Fizeau interferometer using a 2 x 3-port circulator configuration to eliminate the need for couplers or beam splitters as a lot of light power is lost on transmission through these devices. A 50% transmitting mirror can also be used in the receiving Fizeau interferometer to supplement balanced detection.

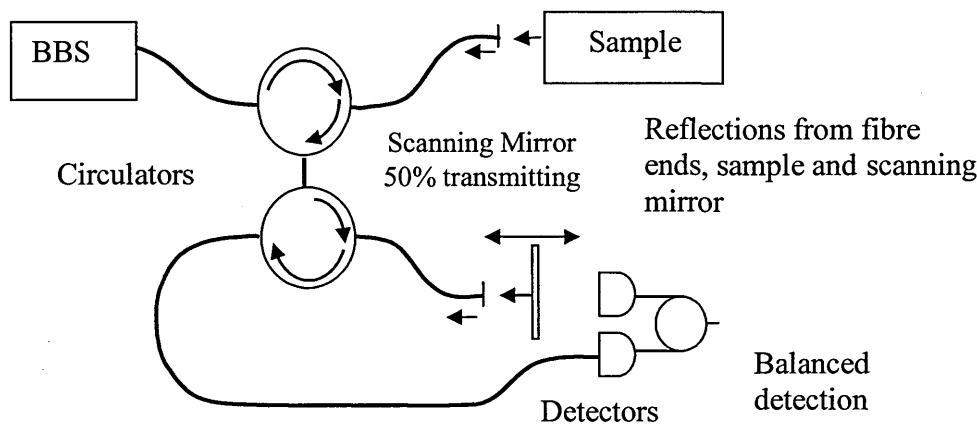


Fig.5.20 Fizeau sensing interferometer using a receiving Fizeau interferometer with 2 x 3-port couplers and a 50% transmitting mirror for balanced detection. BBS=broadband source, , BS=beam splitter.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r T_c^3 (1 - r_r) r_r}{4}, \quad P_s = \frac{P_0 R_s T_c^3 (1 - R_r)^2 r_r}{2}, \quad \frac{P_0 R_x T_c^3 (1 - R_r)^2 r_r}{2}$$

Maximum SNR is 75dB with end of fibre reflection at 0.45 shown in Fig5.22, calculated using Equation 17.

5.5.8 Fizeau sensing interferometer using a receiving Fizeau interferometer using 1 4-port circulator and a 50% transmitting mirror for balanced detection (Fig.5.21)

A Fizeau sensing head can be used with a receiving Fizeau interferometer using a 1 x 4-port circulator configuration to eliminate the need for couplers or beam splitters as a lot of light power is lost through transmission through these devices. A 50% transmitting mirror can also be used in the receiving Fizeau interferometer to supplement balanced detection.

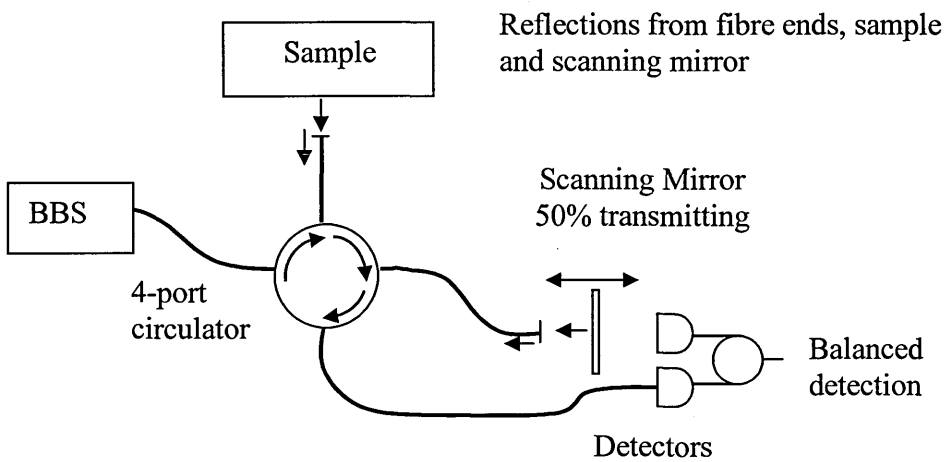


Fig.5.21 Fizeau sensing interferometer using a receiving Fizeau interferometer using 1 4-port circulator and a 50% transmitting mirror for balanced detection. BBS=broadband source, BS=beam splitter

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r T_c^2 (1 - r_r) r_r}{4}, \quad P_s = \frac{P_0 R_s T_c^2 (1 - R_r)^2 r_r}{2}, \quad \frac{P_0 R_x T_c^2 (1 - R_r)^2 r_r}{2}$$

Maximum SNR is 76dB with end of fibre relection at 0.45 shown in Fig5.22 calculated using Equation 17.

The SNR becomes a function of processing interferometer reference arm reflectivity shown in Fig.5.22

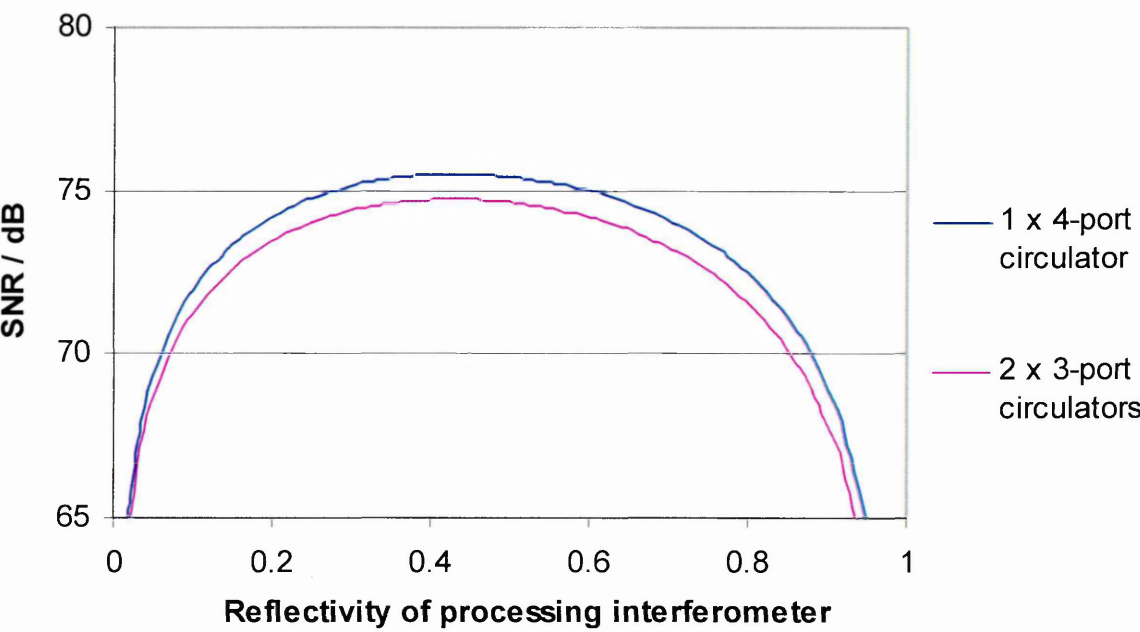


Fig.5.22 SNR as a function of receiving interferometer reference arm reflectivity

The SNR’s of all the Fizeau interferometer designs considered were calculated and are shown in Table.5.2.

The maximum SNR calculated for the Fizeau configuration is 79dB calculated by using a 1x4-port circulator with a Fizeau sensing interferometer and Michelson processing interferometer and using balanced detection. In this configuration all possible light is recirculated and used in detection. The SNR for this configuration is over 18dB greater than the conventional Fizeau design and 11dB greater than that of a conventional

Michelson interferometer (68dB). The maximum SNR calculated for the Fizeau design (79dB) is also comparable to the maximum achievable SNR calculated for the Michelson configurations in this study (81dB).

Fig.5.2 Table showing SNR for all Fizeau Interferometer designs

Fizeau Interferometer Design	Maximum SNR
Fizeau interferometer using an unbalanced coupler	61dB
Fizeau interferometer using an unbalanced coupler and circulator	62dB
Fizeau interferometer with a 3-port circulator	62dB
Fizeau interferometer with a 2, 3-port circulators and balanced detection	78dB
Fizeau interferometer with a 4-port circulator and balanced detection	79dB
Fizeau sensing interferometer with a receiving Fizeau interferometer using 2 3-port circulators	59dB
Fizeau sensing interferometer with a receiving Fizeau interferometer using 2 3-port circulators and a 50% transmitting mirror for balanced detection	75dB
Fizeau sensing interferometer with a receiving Fizeau interferometer using 1 4-port circulator and a 50% transmitting mirror for balanced detection	76dB

The optimal Fizeau design for increased SNR, although only 2dB lower than the optimal Michelson configuration has the advantage that environmental perturbations on the fibre, down lead, causing polarisation and phase fluctuations, have no effect on the received signal and therefore no signal losses due to polarization induced fading will occur. The Michelson interferometer does not have this advantage and any perturbations within the fibre could lead to total signal loss in a worst case scenario.

The optical circulator on its own used in the Fizeau configuration makes a larger contribution to the increased SNR, compared to the Michelson this is because the light travels down the same interferometer arm for the reference and sample beams, therefore the maximum light can be directed into an unbalanced coupler, or with the use of a circulator there is no need for a coupler. The use of a circulator without balanced detection gives an increase of 1dB, however the most significant device for increasing the SNR is the balanced receiver which gives over a 14dB increase in SNR for all configurations where it is viable.

5.6 Mach-Zehnder Interferometer Configurations

All Interferometer configurations in this section use the Mach-Zehnder design with added components (circulators, unbalanced couplers and balanced receivers) to increase the SNR

5.6.1 Mach-Zehnder ‘a’ configuration with one unbalanced coupler⁷ (Fig.5.23)

A Mach-Zehnder ‘a’ configuration in which two couplers are contained is shown in Fig.5.23. This configuration can be used with and without balanced detection and the SNR is a function of coupler splitting ratio. For single detection both couplers are unbalanced and the SNR is a function of the coupler splitting ratio as shown in Fig.5.25. For balanced detection the second coupler is balanced and SNR is a function of splitting ratio for the first coupler shown in Fig.5.29.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r \alpha}{2}, \quad P_s = \frac{P_0 R_s T_c^2 (1 - \alpha)}{2}, \quad P_x = \frac{P_0 R_x T_c^2 (1 - \alpha)}{2}$$

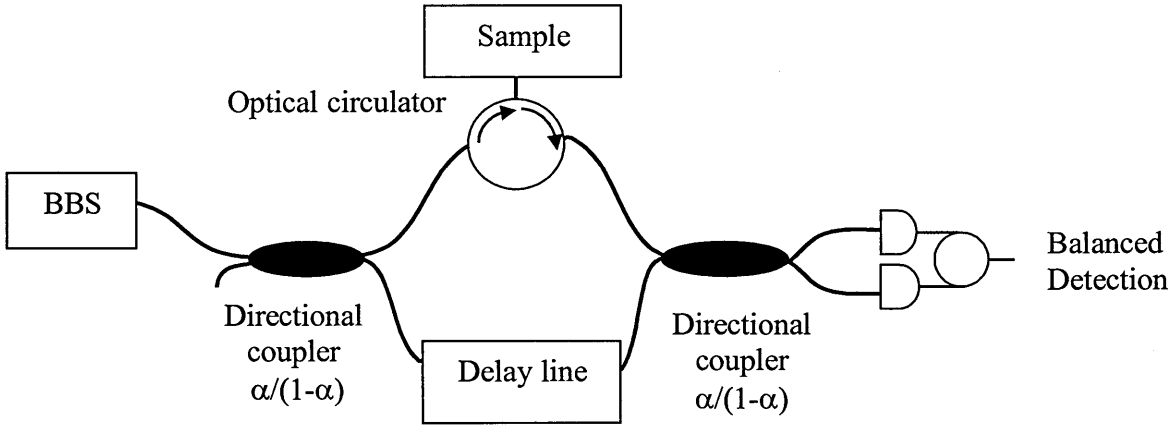


Fig.5.23 Fibre optic based Mach-Zehnder 'a' interferometer OCT system⁷.
 BBS=broadband source, α =coupler splitting ratio, $\alpha=0.5$ for balanced coupler.

5.6.2 Mach-Zehnder 'a' configuration with two unbalanced couplers (Fig.5.24)

A Mach-Zehnder 'a' configuration in which a coupler replaces the circulator, is used. Fig.5.24 shows the SNR as a function of the splitting ratio when the coupler replacing the circulator is unbalanced and this is shown in Fig.5.25.

$$\text{SNR} = 10 \log \left[\frac{4\rho^2 P_r P_s}{\sigma_{\text{rec}}^2 + 2qB\rho^2 (P_r + P_s + P_x) + \frac{B\rho^2 P_r P_x}{\Delta\nu} (1 - V^2)} \right] \quad \text{Eq.18}$$

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_0 = P_0 R_r \alpha^2, \quad P_s = P_0 R_s T_c^2 (1 - \alpha)^2, \quad P_x = P_0 R_x T_c^2 (1 - \alpha)^2$$

Maximum SNR is 68dB and 0.5 splitting ratio by using equation 18.

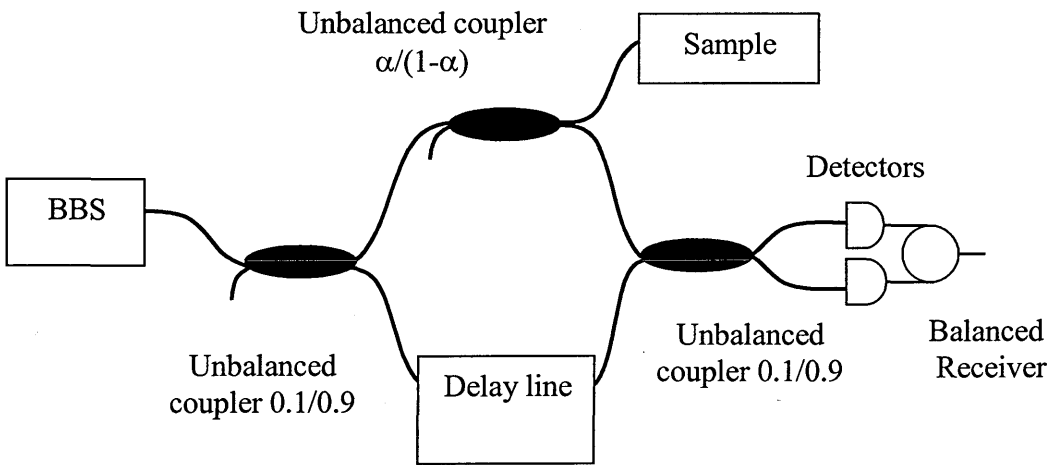


Fig.5.24 Fibre optic based Mach-Zehnder 'a' interferometer OCT system⁷.
BBS=broadband source, α =coupler splitting ratio, $\alpha=0.5$ for balanced coupler.

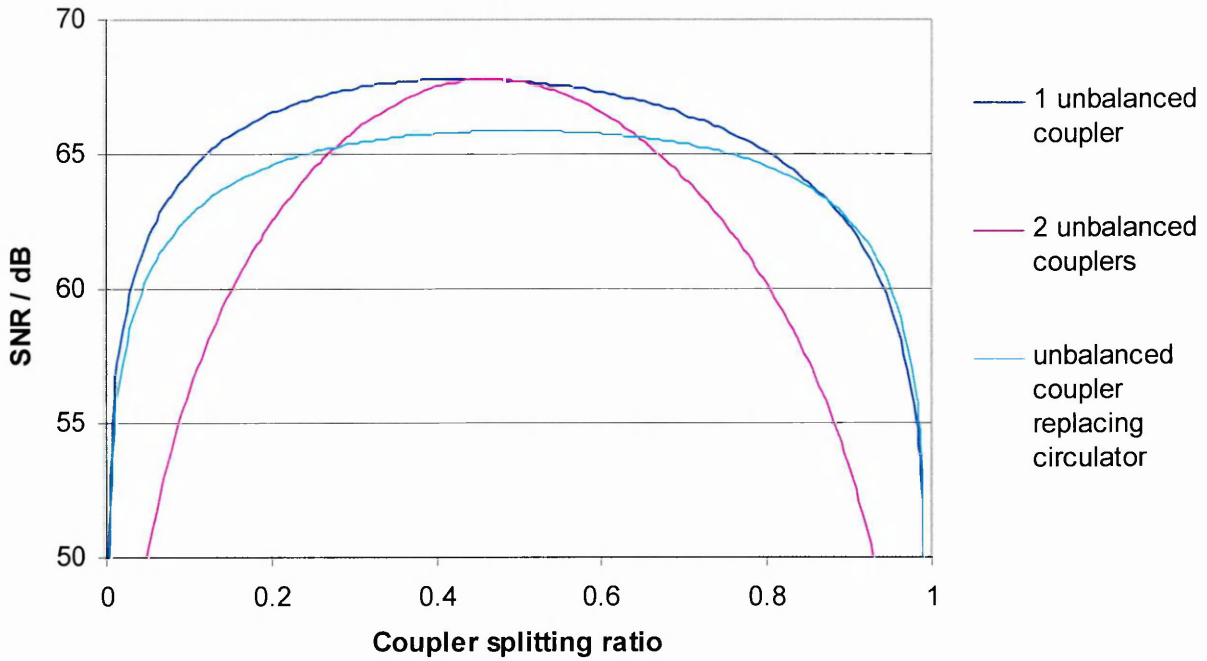


Fig.5.25 Variation in SNR for the Mach Zehnder ‘a’ configurations.

5.6.3 Mach-Zehnder ‘b’ configuration with one unbalanced coupler⁷ (Fig.5.26)

A Mach-Zehnder ‘b’ configuration in which two couplers are used is presented. This configuration can be used with and without balanced detection and the SNR is a function of the coupler splitting ratio. For single detection both couplers are unbalanced and the SNR is a function of the coupler splitting ratio shown in Fig.5.28. For balanced detection the second coupler is balanced and the SNR is a function of the splitting ratio for the first coupler as shown in Fig.5.29.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r \alpha (1 - \alpha)}{2}, \quad P_s = \frac{P_0 R_s T_c^2 (1 - \alpha)}{2}, \quad P_x = \frac{P_0 R_x T_c^2 (1 - \alpha)}{2}$$

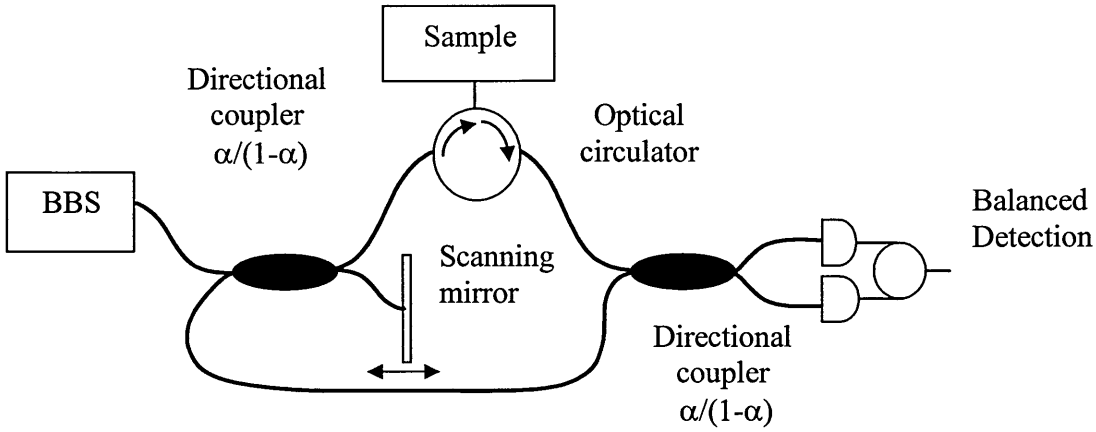


Fig.5.26 Fibre optic based Mach-Zehnder 'b' interferometer OCT system⁷.
BBS=broadband source, α =coupler splitting ratio, $\alpha=0.5$ for balanced coupler.

SNR is a function of the coupler splitting ratio as shown in Fig.5.28, and the maximum SNR is 68dB at 0.5 splitting ratio calculated using Equation 18.

5.6.4 Mach-Zehnder 'b' configuration with two unbalanced couplers (Fig.5.27)

A Mach-Zehnder 'b' configuration in which a coupler replaces the circulator is used this is shown in Fig.5.27. The SNR is a function of splitting ratio when the coupler replacing the circulator is unbalanced, this is shown in Fig.5.29.

The power equations are given by where P_s , P_r and P_x are defined in 5.5.1:

$$P_r = \frac{P_0 R_r}{6}, \quad P_s = \frac{P_0 R_s \alpha (1-\alpha)}{4}, \quad P_x = \frac{P_0 R_0 \alpha (1-\alpha)}{4}$$

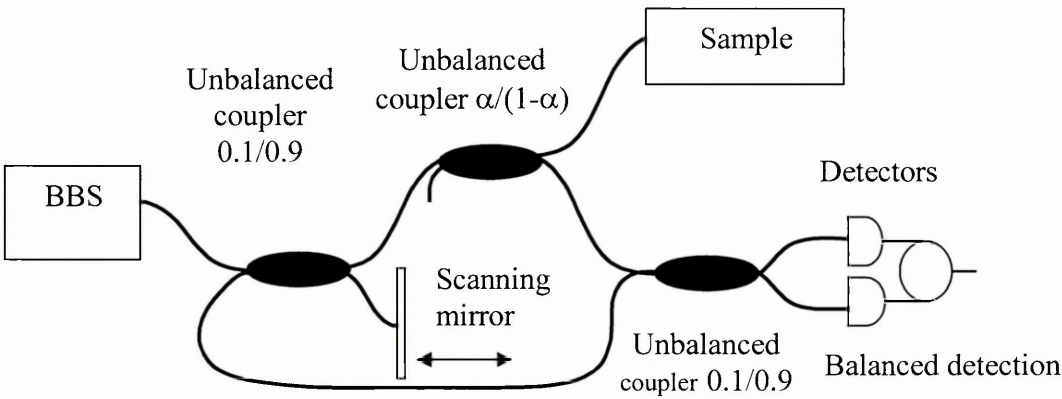


Fig.5.27 Fibre optic based Mach-Zehnder ‘a’ interferometer OCT system⁶.
BBS=broadband source, α =coupler splitting ratio, $\alpha=0.5$ for balanced coupler.

The maximum SNR is 68dB using Equation 18.

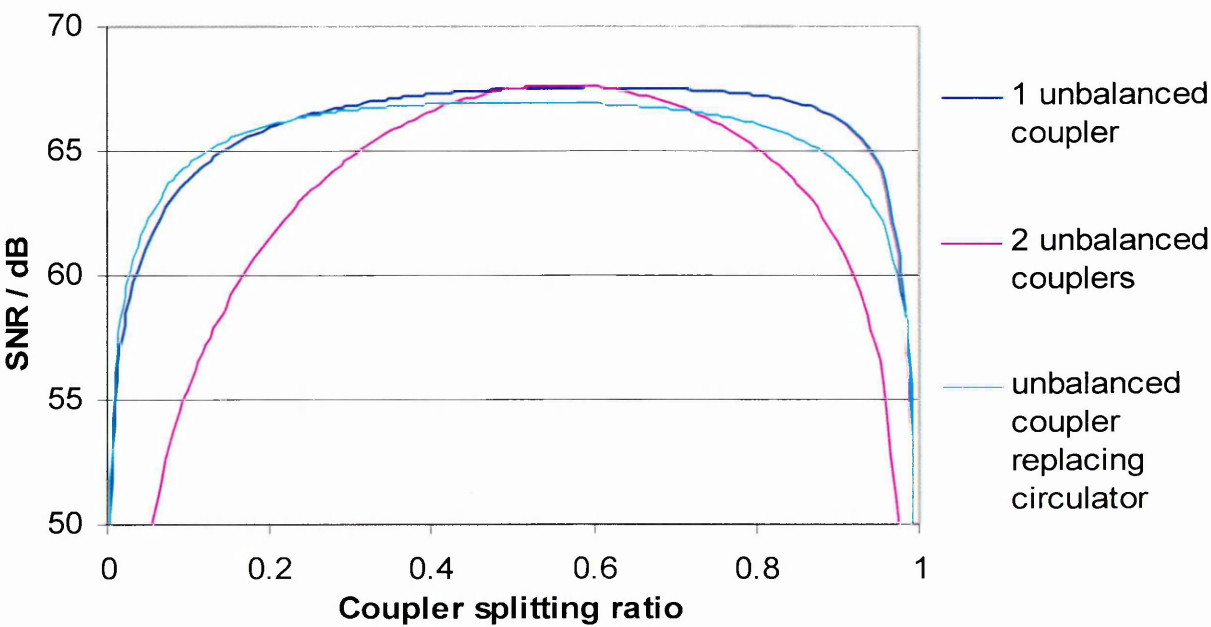


Fig.5.28 Variation in SNR for the Mach Zehnder ‘b’ configurations.

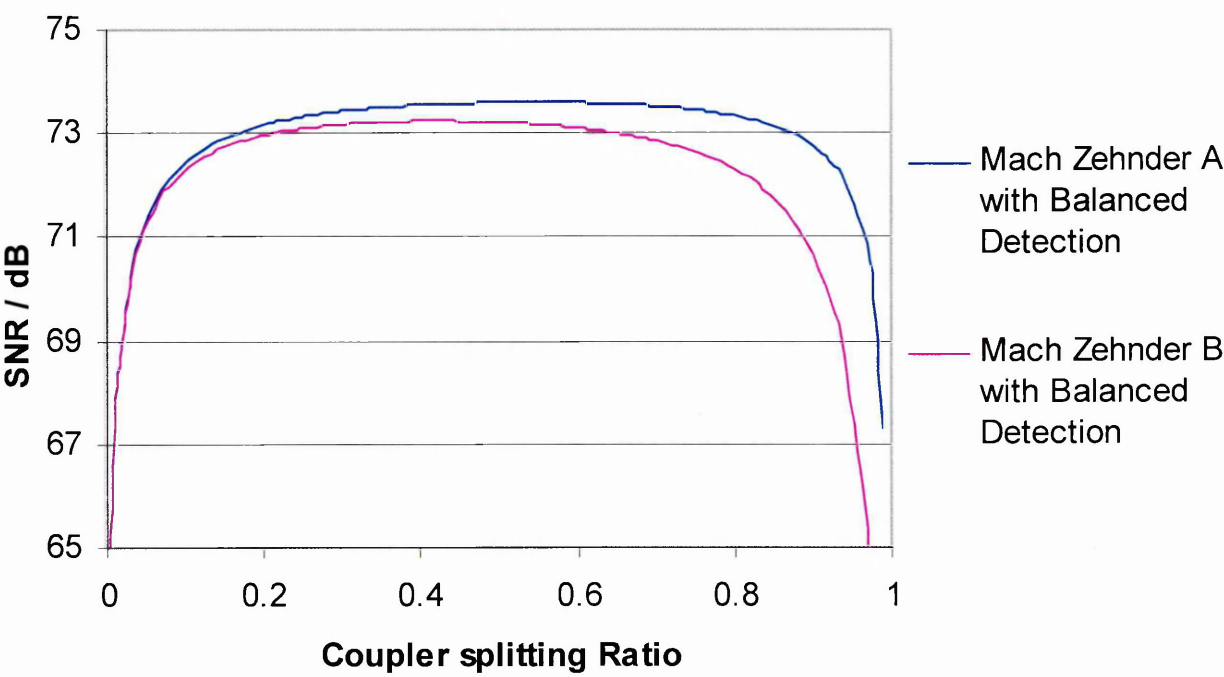


Fig.5.29 SNR as a function of coupler splitting ratio for Mach Zehnder configurations using balanced detection.

The SNR of all the Mach-Zehnder interferometer designs were calculated and are shown in thable.5.3.

The maximum SNR calculated for the Mach-Zehnder ‘a’ configuration is 74dB and uses only one unbalanced coupler and balanced detection.

The maximum SNR calculated for the Mach-Zehnder ‘b’ configuration is 73dB and uses only one unbalanced coupler.

The SNR for the Mach Zehnder configurations can be increased by using power conserving optical components (unbalanced couplers, circulators and balanced detectors) but is lower overall SNR compared to the Michelson and Fizeau configurations. Replacing the

circulator with a coupler decreases the SNR by 1dB but could be used as an alternative configuration for an OCT interferometer, as a less costly option.

Fig.5.3 Table showing SNR of all Mach-Zehnder configurations

Mach-Zehnder Interferometer Design	Maximum SNR
Mach-Zehnder A configuration with one unbalanced coupler	68dB
Mach-Zehnder A configuration with two unbalanced couplers	68dB
Mach-Zehnder A configuration with unbalanced coupler replacing circulator	66dB
Mach-Zehnder A configuration with 1unbalanced coupler, circulator and balanced detection	74dB
Mach-Zehnder B configuration with one unbalanced coupler	68dB
Mach-Zehnder B configuration with two unbalanced couplers	68dB
Mach-Zehnder B configuration with unbalanced coupler replacing circulator	67dB
Mach-Zehnder B configuration with 1unbalanced coupler, circulator and balanced detection	73dB

5.6 Summary

The SNR for a number of OCT configurations has been calculated in order to determine the design with the best performance for endoscopic OCT, ie a configuration that will eliminate polarisation induced fading due to perturbations in the sample arm. These designs have been compared to previous designs studied by Rollins *et al*⁸. The new Fizeau designs have shown to give a comparable SNR values to the previous Michelson designs with the added advantage of eliminating unwanted polarisation effects in the fibres that may decrease the

SNR. The Mach-Zehnder configurations increase the SNR compared to the standard configuration but give an overall lower SNR than the Michelson and Fizeau. Using three couplers gives a relatively high SNR and reduces the cost of the system by avoiding the use of circulators which may be expensive, a few thousand pounds compared to ~£50 for couplers.

Overall all the designs considered in this study significantly improved the SNR compared to their conventional configurations, up to 21dB for the Michelson and 18dB for the Fizeau configurations. The use of a circulator alone within the configuration was negligible for the Michelson but gave a slight improvement for the Fizeau. The use of balanced detection was shown to give the greatest SNR improvement. With the need for faster scanning techniques it is important for a high SNR and these calculations have shown that it is achievable.

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Chapter 6

Experimental Investigation of Fizeau OCT Using Optical Circulator

6.1 Optical components for the system

Optical Coherence Tomography (OCT) is a valuable imaging tool as it provides a higher resolution compared to other imaging techniques¹, allowing individual cells to be resolved. Subcellular depth images can also be obtained with OCT, which has not been achieved with other imaging modalities. OCT is also a valuable imaging technique as it uses low power light sources which are not harmful to patients.

The high resolution of an OCT system is extremely useful for looking for small changes in cells, which would be useful for diagnosing cancerous tissue². One drawback of OCT is that the imaging depth range is ~2-3mm for turbid biological tissue, which is adequate for detecting cell changes in cancer as these occur in the first few hundred micrometers of

scanning. However most tissue surfaces require the OCT probe to focus the light on the surface of the tissue inside the body, as the light will get attenuated travelling through centimetres of tissue, which is why there is a great demand for high power, low coherent sources in *in vivo*, endoscopic OCT³.

The results from Chapter 5 show that the Fizeau Interferometer can be modified, by changing the configuration and using power conserving optical components, to give a higher SNR than the conventional Michelson configuration presently used for *in vitro* applications³ and a comparable SNR to a Michelson optimised for power conservation.

As discussed in Chapter 5, the Fizeau configuration that theoretically gives the highest SNR uses a 4-port circulator and balanced detection, shown in Fig.6.1. In order to set-up a system with the highest SNR all components had to be considered in order to optimise the SNR for OCT imaging of tissue and these system components are discussed in this chapter.

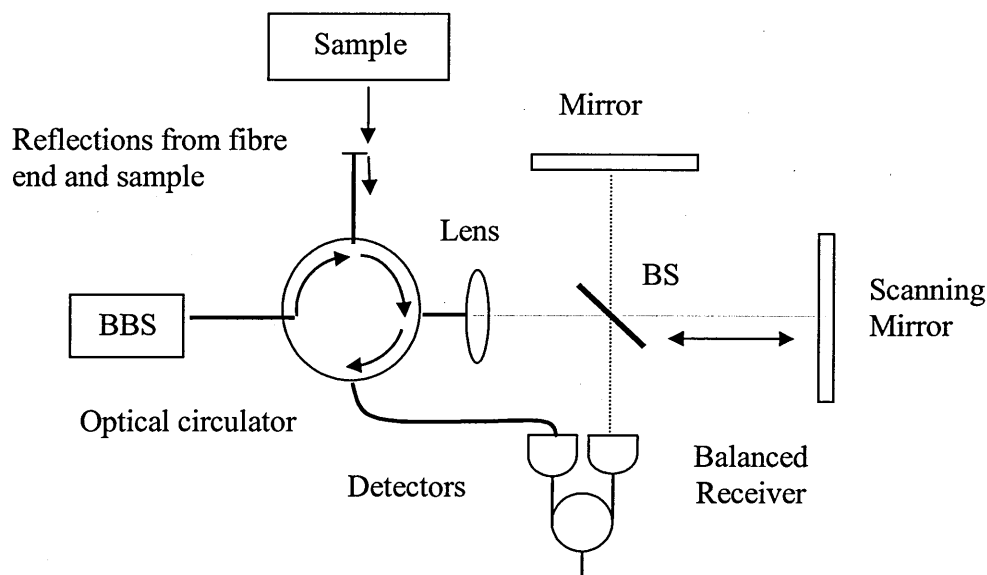


Fig.6.1 Fizeau configuration to give the maximum SNR

6.1.1 Design Specifications

The design requirements for this OCT system is to be able to improve the SNR for the Fizeau interferometer so that it has a higher or comparable performance to the Michelson interferometer so that an interferometer can be setup that eliminates polarisation induced signal fading and ultimately improves endoscopic measurements. The system is to be set-up on an optical bench in the laboratory and reflections taken from a mirror placed in the sample arm. The different components in the OCT system were chosen to optimise the performance of the Fizeau configuration and are discussed below.

6.1.2 Optical Light Source

The importance of the light source in an OCT system is discussed in Chapter 2 and is one of the most important basic components in the OCT system. The spectral characteristics of the source used are crucial as it directly related to the image quality and resolution of imaging⁴.

Since the purpose of the improved interferometer design is to image tissue (of which the esophagus is an example), it is important that the light from the source is able to penetrate deep enough into the tissue, to distinguish tissue layers and to give a high resolution. The depth of penetration is determined by the wavelength of the source⁵, the longer the wavelength the higher the depth resolution. However tissue is mainly made up of water and the wavelength of the source should be chosen so that water absorption is as low as possible. There are three wavelengths, where the absorption losses (due to impurities of iron / water) in optical fibres have a minimum: ~830nm, ~1300nm and 1550nm and this is shown in Fig.6.2⁶.

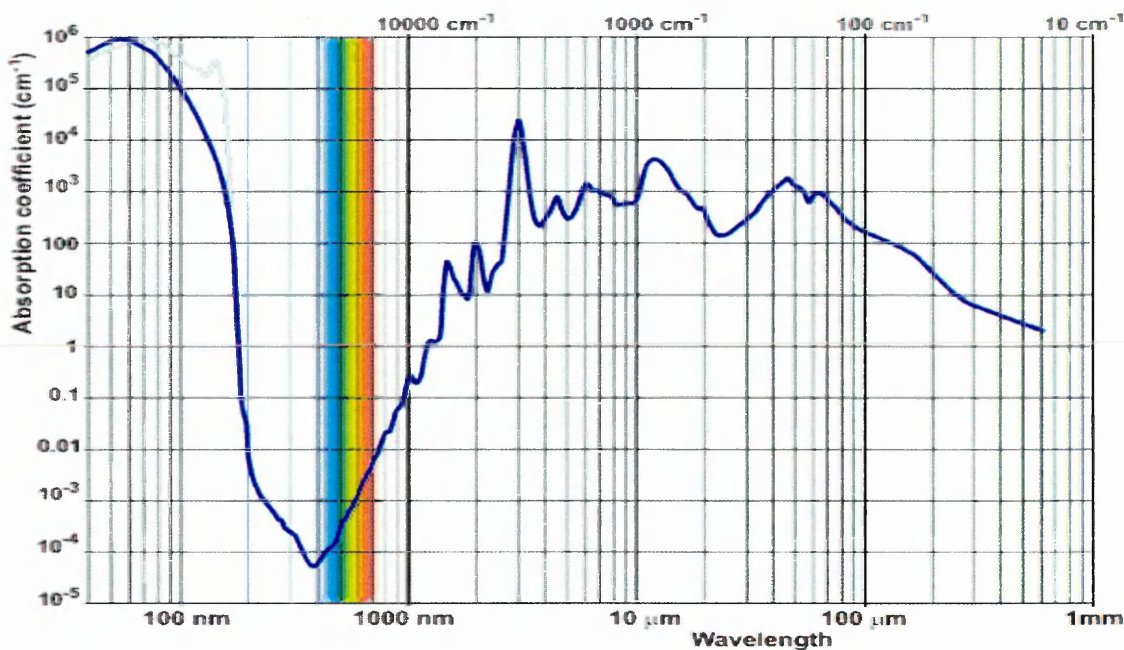


Fig.6.2 Water absorption spectra for the near infra red region showing low absorption at ~800nm, 1310nm and 1550nm⁶

Commercially available sources suitable for setting up the OCT system were investigated to find the most suitable for the project. The table shown below shows the sources readily available with their characteristics

A previous study comparing a 1310nm and a 1550nm source for OCT imaging concluded there was no overall disadvantage to using either wavelength⁷. However 1550nm would give an increased depth resolution which is especially important for imaging turbid tissue such as the esophagus. Another advantage to choosing 1550nm is that it is the optimum wavelength for the optical telecommunications industry and therefore fibre components are more readily available over this wavelength region. Optical circulators are generally only available at 1550nm and this confirmed with the ready availability of other fibre components and sources it was decided that 1550nm would be used in the OCT set-up.

Table.6.1 showing commercially available SLD sources suitable for OCT imaging

CENTRE WAVELENGTH /nm	BANDWIDTH /nm	COHERENCE LENGTH / μ m	OUTPUT POWER	COMPANY	COST on July 2000
800	50	12.8	1mW	Superlum ⁸	£750
940	70	12.6	0.5mW	Superlum ⁸	£1500
980	40	24.0	1-2mW	Profile ⁹	£4000
1300	25	67.6	120 μ W	Profile ⁹	£5000
1300	30	56.3	3mW	Superlum ⁸	£1400
1300	50	33.8	>3mW	Laser 2000 ¹⁰	£1000
1550	25	96.1	75 μ W	Profile ⁹	£2000
1550	25	96.1	350 μ W	Profile ⁹	£2000
1550	60	17.0	2mW	Opto Speed ¹⁰	£1900
1550	25	96.1	0.5mW	Superlum ¹¹	£1850

A previous study comparing a 1310nm and a 1550nm source for OCT imaging concluded there was no overall disadvantage to using either wavelength¹¹. However 1550nm would give an increased depth resolution which is especially important for imaging turbid tissue such as the esophagus. Another advantage to choosing 1550nm is that it is the optimum wavelength for the optical telecommunications industry and therefore fibre components are more readily available over this wavelength region. Optical circulators are generally only available at 1550nm and this confirmed with the ready availability of other fibre components and sources it was decided that 1550nm would be used in the OCT set-up.

The most appropriate source within the budget of this project was a superluminescent diode (SLD) as these are available with large bandwidths. The SLD (Opto Speed SLED1550D) had a centre wavelength of 1550nm with a bandwidth of 60nm, which gave a calculated coherence length and therefore optimum image resolution of 17 μ m which is sufficient for measuring the epithelial layer thickness (~50-120 μ m).

Approximately 1% of light incident on a tissue interface is reflected and used for detection⁵. On this basis, providing the sensitivity of the detectors is sufficient then the reflected signals will be detected adequately with initial optical powers >0.5mW. The power of the SLD (Opto Speed) exiting the fibre is 2mW which is relatively high for

commercially available SLDs and sufficient for this project. The Specifications for the Opto Speed SLED1550D are shown in Appendix G.

6.1.3 Focussing Lens System

In order to produce a high quality 2-dimensional image the resolution in both the lateral and the longitudinal directions is important. The diameter of the beam focussed onto the tissue determines the lateral resolution and so it should preferably be as small as possible. Another advantage of a small beam diameter is that all the optical power can be focussed onto a small area giving a larger signal on reflection. The beam used for OCT imaging should also have a large Rayleigh length¹³ this is so the light is focussed to as small as possible over the scanning length, which may be a relatively long distance.

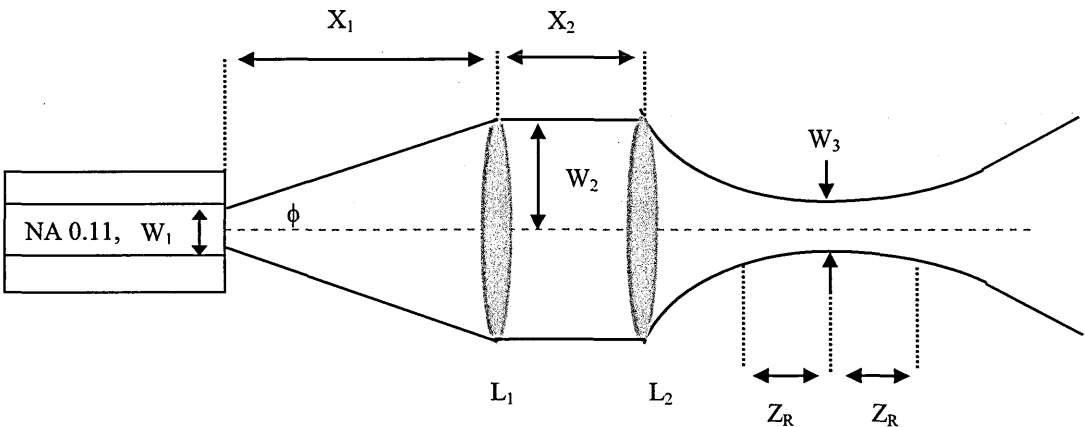


Fig.6.3 Lens system for focussing the light from the fibre onto the sample, Gaussian optics.

As the light emerges from an optical fibre, diffraction effects cause a divergence of the light beam at an angle of ϕ (Fig 6.3) towards the sample. In order to focus the light beam onto the sample with a small beam width and long Rayleigh length a lens focussing system is required, as shown in Fig.6.4. Here a biconvex lens (L_1) is placed at its focal distance (X_1) from the fibre, this collimates the beam after passing through the lens (L_1) and has a

constant beam diameter (W_2). By placing another biconvex lens (L_2) in the collimated beam at a distance (X_2) from the first lens further diffraction occurs and the beam converges to a beam waist (W_3) at its focal length and Rayleigh length of Z_R . Light that is reflected within the Rayleigh range (focal point of the second lens, L_2) is most efficiently launched back into the optical fibre allowing the Fizeau sensing head to obtain a high sensitivity.

Calculating the focal length of the lenses suitable for the Fizeau sensing head is determined using the laws of Gaussian Optics¹⁴.

The beam waist of the light after being transmitted through the first lens can be calculated using trigonometry as shown in Eq.1

$$\tan\phi = \frac{W_2}{X_1} \quad (1)$$

Where X_1 is equal to the focal length of the first lens and W_2 is the beam width of the diverging beam.

The beam width after passing through the second lens is given by Eq.2

$$W_3 = \frac{\lambda f}{\pi W_2} \quad (2)$$

Where $n=1,2,\dots$, λ = wavelength of the light, f = focal length of the lens and W_n is the beam width.

The Rayleigh length, Z_R , is calculated using Eq.3

$$Z_R = \frac{\pi W_3^2}{\lambda} \quad (3)$$

By using these equations and substituting varying focal length values for the two lenses the beam width and the Rayleigh length for the focussed beam can be calculated. Ideally the Rayleigh length should be as long as possible so that the light can be efficiently collected over a long distance or at least over the scanning distance. OCT has been shown to image to depths between 1-2mm¹⁵ but for the purpose of this system which is interested in the changing thickness of epithelial tissue a scanning over 150µm would be sufficient. The beam width should also be as narrow as possible in order to focus most of the power onto the reflecting sites and also to keep the lateral resolution as high as achievable.

In practice it is difficult to achieve a long Rayleigh length with a narrow beam waist as Z_R is proportional to W_3 . The optics commercially available are also limited by their sizes and focal lengths and so a compromise had to be met to obtain the most suitable lens configuration for the system. This was achieved by using a 10mm diameter biconvex lens with a focal length of 15mm (Melles Griot), this collimated the beam to give a beam waist of 1.7mm. A second 10mm diameter lens with a 40mm focal length (Melles Griot) focussed the beam down to 12µm with a Rayleigh length of 300µm. This effectively allows the light to focus over 2 times the Rayleigh length, 600µm this is more than adequate for the OCT set-up. The choice of lenses used in this system was limited by the manufacturers and only lenses of certain focal lengths and diameters were available. It was also limited by the dimensions of the OCT system.

The lens system was made up into a cylindrical probe shown in Fig 6.4 which could be easily moved around and the placed on any object suitable for imaging where The probe was set-up was focussed at end of the probe.

The lenses were mounted on frames that screwed into the probe and could move position by using a special screw key for the set-up. Once the lenses are set-up in the exact position for the required Rayleigh length at the end of the probe, it was straightforward to position the sample at the point of highest light collecting efficiency.

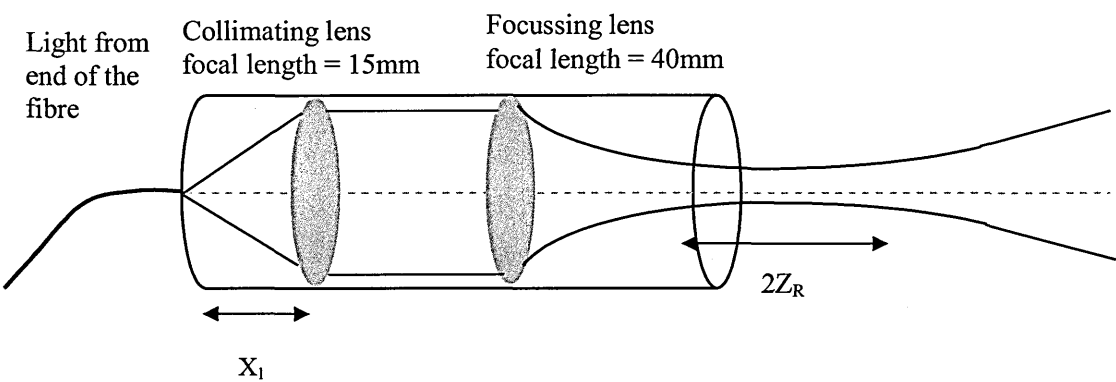


Fig.6.4 Probe containing focussing lens system used in the Fizeau sensing head.

For clinical endoscopic applications it is important that the focussing lens system is as small as possible as it is required to fit down the patients esophagus, therefore there maybe some limitations, from the commercially available lenses, as to the focussing ability of the lens system.

6.1.4 Optical Circulator

An optical circulator (Fig 6.5) allows light to be transmitted through a port and the reflected light is circulated and transmitted through another port. In the case of Fig.6.5, light entering port 1 is transmitted through port 2 and the reflected light entering port 2 is transmitted through port 3.

Most optical circulators have 3 or 4 ports used for circulating light. For the optimal Fizeau interferometer design a 4-port circulator is used. The circulator used in the interferometer works at 1550nm (Oplink, Appendix H) and most circulators are only commercially available at 1550nm which is an advantage of using light at this wavelength for the imaging system. The transmission through each port of the circulator is approximately 0.87.

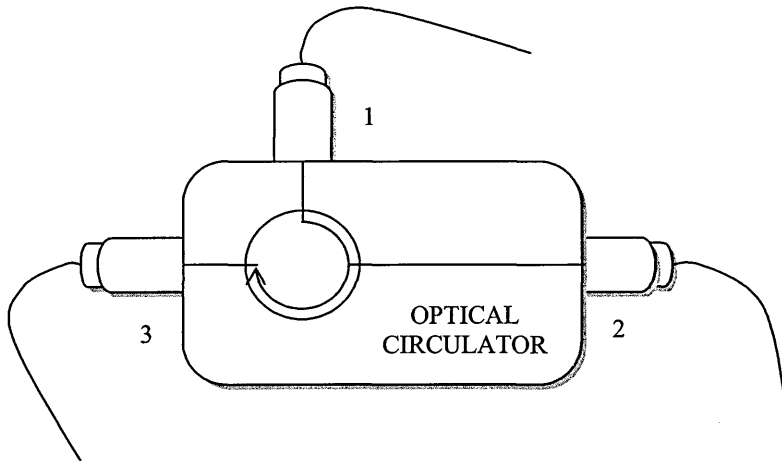


Fig 6.5 A 3-port optical circulator.

6.1.5 Detection System

Balanced detection can be used to reduce noise in an interferometer¹⁶. Common-mode noise that is present on both the reference and signal beams, such as laser-intensity noise, is cancelled out and does not appear as part of the signal. Any imbalance between the photocurrents generated by the reference and signal detectors, however, is amplified. Most balanced detectors work in the 1550nm wavelength band.

The bandwidth of the detector is important in data acquisition as it affects the SNR of the signal, the bandwidth is inversely proportional to the bandwidth so ideally should be kept as low as possible. Estimating the number of data points required for an adequate signal and the speed of the translation stage made an estimation of the bandwidth suitable for the imaging system.

The translation stage was set at a maximum speed of $\sim 20\text{m/s}$ scanning over a distance of $\sim 0.005\text{m}$ per scan. The number of data points required over this range ~ 10000 over estimating for a good signal. The bandwidth of the detector should therefore be $\sim 8000\text{Hz}$.

The detector ordered was a New Focus balanced receiver working at 800-1700nm and with a bandwidth of 125kHz. The receiver could be used with one port for single detection and two ports for balanced detection, each detector consists of a InGaAs PIN photodiode and the signals are subtracted if balanced detection is used. The noise equivalent power (NEP) of the balanced detector is $3\text{pW/Hz}^{-1/2}$ (Appendix I).

Unfortunately the detector ordered had many associated problems on arrival that have taken 15months to correct. The time was taken up re-sending the faulty detector to the USA for repair, this occurred twice. The results which would have been obtained using the balanced receiver are therefore not used for the results presented here, the specifications of the detector ordered is shown in Appendix I.

6.1.6 Scanning Method

For the purpose of this investigation the mirror (selected for use at 1550nm wavelength) was mounted onto a translation stage (National Instruments) which varied the path length between the reference and sample arm was by scanning the mirrors. This was controlled by a LabView program.

The mirrors used for the sample and scanning system were suitable for a broadband 1550nm source coated with protected gold and 100% reflectivity (Melles Griot).

6.2 Calculating the SNR for the Optimal Fizeau Configuration

The signal to noise ratio for the optimal Fizeau configuration was calculated for the equipment used in the experiment, using Equation 4. Only the configuration with a 4-port circulator and a single detection system was used for this investigation.

$$\text{SNR} = 10 \log \frac{\text{Signal}}{\text{Noise}} = \frac{I_s^2}{\sigma_{\text{Tot}}^2} \quad (4)$$

where I_s is the AC component of the interference photocurrent and σ_{Tot} is the sum of all the noise within the system due to receiver noise, shot noise and excess noise shown in Eq.5.

$$\sigma_{\text{Tot}}^2 = \sigma_{\text{Rec}}^2 + \sigma_{\text{Ex}}^2 + \sigma_{\text{Sh}}^2 \quad (5)$$

The AC signal is calculated using Eq.6.

$$I_s^2 = 2\rho^2 P_r P_s \quad (6)$$

For this Fizeau configuration P_r and P_s are given by Eq.7 and 8.

$$P_r = \frac{P_0 R_r T_c^2}{4} \quad (7)$$

$$P_s = \frac{P_0 R_s T_c^2 (1 - R_r)^2}{4} \quad (8)$$

P_0 is the power of the optical source and for the SLD=2mW. However the power emitted from the SLD fibre pigtail was measured to be 1.8mW at 140mA.

T_c is the transmission factor through the circulator ports, the 4-port coupler has an insertion loss of ~0.6dB and therefore the transmission factor = 0.87 through each port.

The detector used instead of the balanced receiver was a InGaAs PIN Photodiode (Hamamatsu, G8370 series) with responsivity 0.95 and NEP $0.1 \text{ pW/Hz}^{1/2}$. Specifications are given in Appendix J.

$R_s=1$ as a 100% reflecting mirror is used as the sample and $R_r=0.04$ from the Fresnel reflected light from the end of the fibre assuming no other losses in the system. The SNR for three Fizeau configurations were calculated using the same method as in Chapter 5. The calculated results are shown in Table 6.2.

The SNR values cannot be directly compared to the theoretical SNR values calculated in chapter 5 because different parameters have been used. However the relative relationship between the experimental configurations can be compared to one another to see if they have the same comparison as the theoretical configurations.

Table 6.2. showing calculated SNR values for the Fizeau interferometer configurations using the specified equipment and the SNR equation in Chapter 5.

Fizeau Configuration	SNR / dB
Fizeau using a 50/50 Directional Coupler with single detection	53
Fizeau with 4-port circulator with single detection	54
Fizeau with 4-port circulator with balanced detection	79

The values calculated have taken into account maximum specification values quoted by the manufacturers. There are likely to be other power losses occurring due to launching light into the optical fibre and general losses throughout the system.

The theoretical results show a significant improvement in SNR using the circulator with balanced detection by up to 26dB, whereas there is only a 1dB improvement by using a circulator in the system without the use of balanced detection.

6.3 Measuring the SNR for the Optimal Fizeau Configuration

6.3.1. Measurements from a single reflection

Two Fizeau interferometer configurations were set-up using the equipment described within this chapter.

The first system used a directional coupler with a 50/50 splitting ratio (Fig.5.2) and the second configuration replaced the coupler with a 4-port circulator (Fig.6.1). Both systems were set up using a single detector as the balanced receiver was faulty. The Interference signal from both configurations is shown in Fig.6.6 and 6.7 using the equipment described in the previous sections.

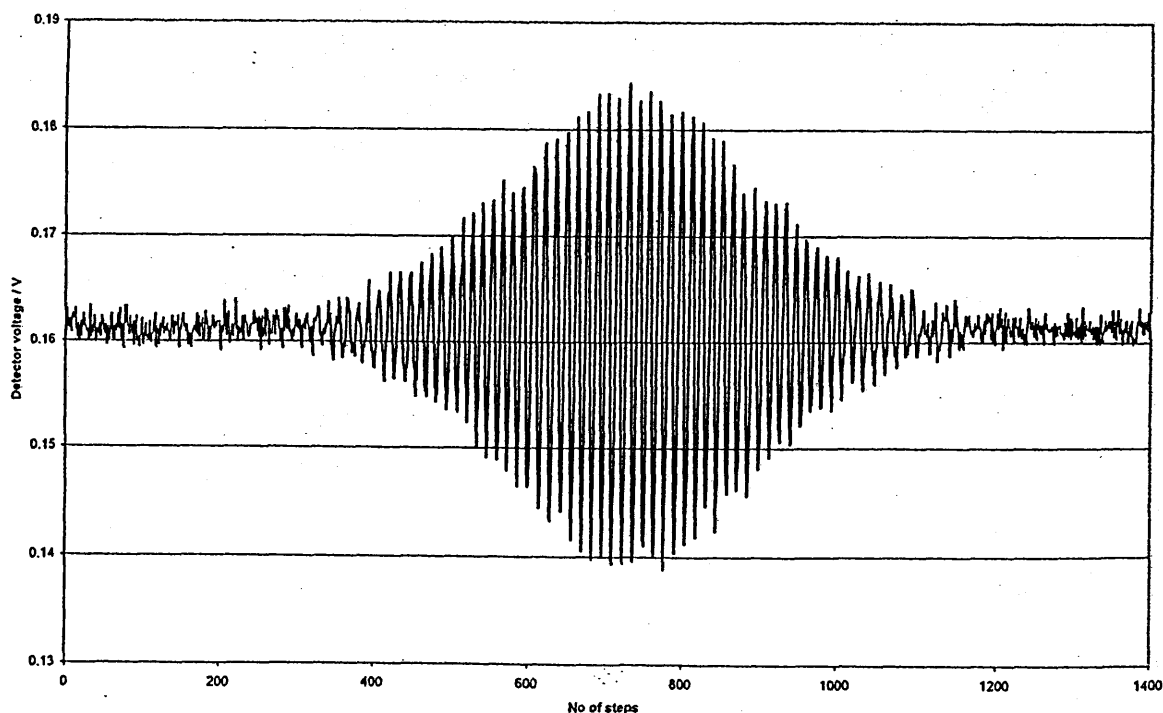


Fig.6.6 Interference signal obtained from the Fizeau set-up using a 50/50 directional coupler.

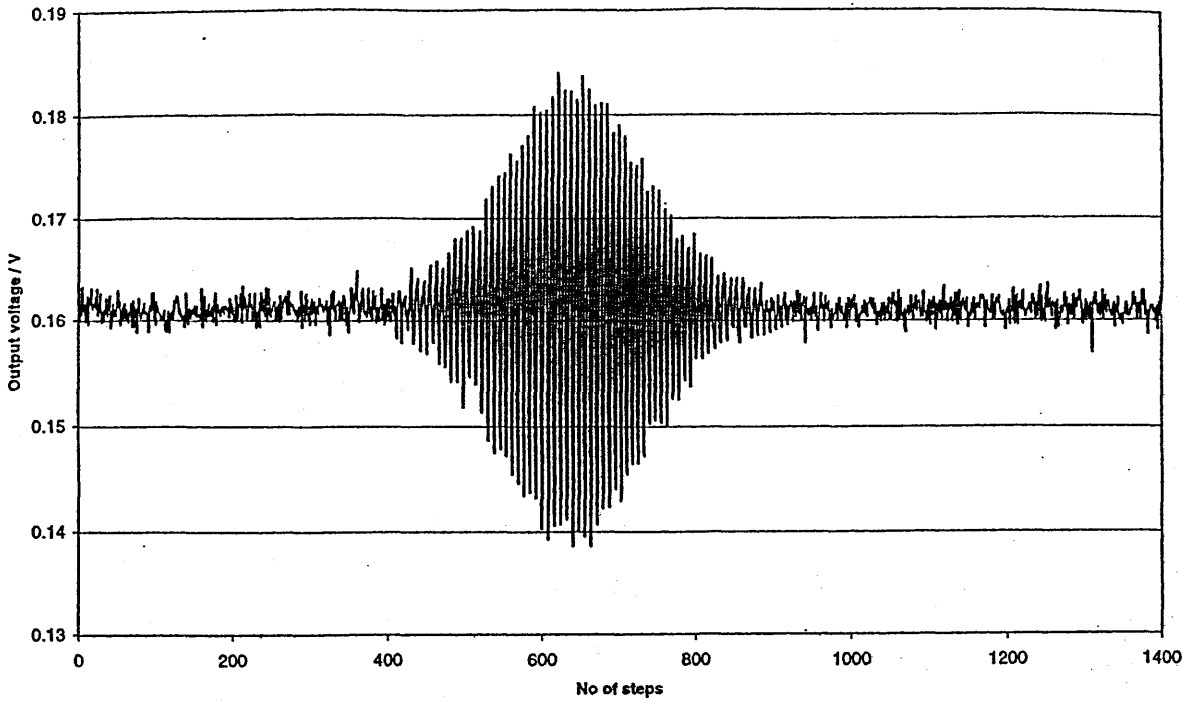


Fig.6.7 Interference signal obtained from the Fizeau set-up using a 4-port circulator.
(Courtesy of Dr. Helen Ford, Cranfield University)

The SNR for each configuration was calculated using the relationship:

$$\text{SNR} = \frac{\text{Signal}}{\text{Noise}} = 20 \log \left(\frac{\text{Fringe}}{\text{Noise}} \right) \quad (\text{pk volts}) \quad (9)$$

The results for the calculated SNR values of the two interference signals, calculated using Equation 9 is shown in Table 6.3

Table 6.3 SNR values from interferometric signals measured from the experimental set-up.

Interferometer configuration	Peak Signal (mV)	Peak Noise (mV)	SNR / dB
Fizeau using a 50/50 coupler	23.0 ± 0.5	3.0 ± 0.5	17.7 ± 1.4
Fizeau using a 4-port circulator	23.5 ± 0.5	2.5 ± 0.5	19.4 ± 1.4

The results show an improvement of just over 1dB between the conventional Fizeau configuration and the circulator only configuration. This corresponds to what the theoretical results in Chapter 4 suggested and the SNR values in section 6.2 predicted. This result suggests that the SNR of the Fizeau interferometer can be increased by the use of power conserving components and in this case a circulator replacing a directional coupler. This would allow the Fizeau interferometer to be used as a suitable OCT system with comparable performance to the Michelson Interferometer and eliminating any polarisation induced signal fading within the OCT system. Measurements using the balanced receiver were unable to be obtained during the scope of this project, as previously discussed, but theoretically this would have predicted a greater increase in SNR than replacing the directional coupler with an optical circulator.

6.3.2 Measurements from reflection from a glass slide

More than one interference signal can be obtained for a single scan if there are more than one reflecting site within the sample arm of the interferometer. This can easily be achieved by placing a glass slide in front of the mirror, in the sample arm. With the glass slide in position there should be three interference signals observed on the scan, the first signal from the reflection from the glass slide the second from the mirror and a third from a reflection from the back of the glass slide onto the mirror. The signal reflected back from

the mirror would give the biggest peak as it is 100% reflecting compared to the glass slide which is ~4% reflecting.

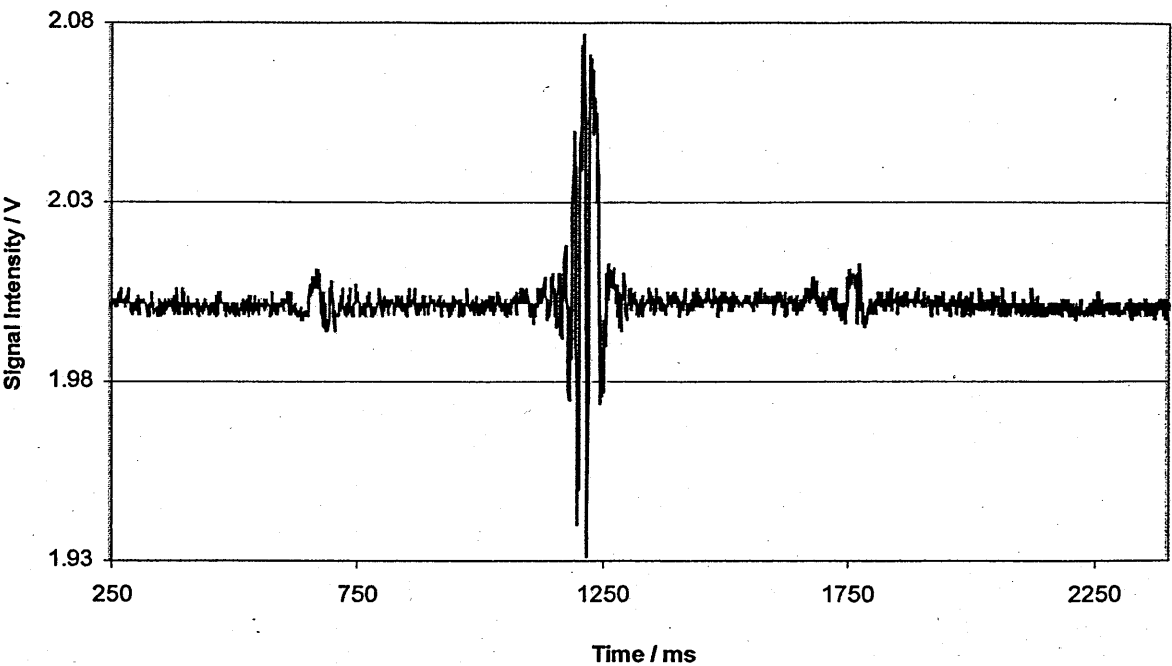


Fig.6.8 Scan obtained from a mirror and glass slide from a Fizeau interferometer using a balanced coupler and PIN photodiode.

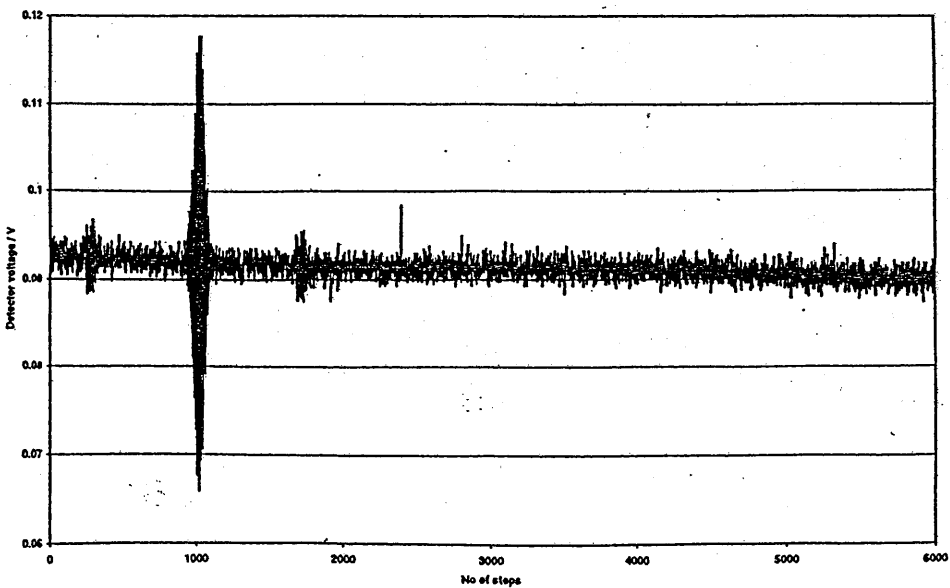


Fig.6.9 Scan obtained from a mirror and glass slide from a Fizeau interferometer using a 4-port circulator and PIN photodiode.

Scans of reflections of glass slides were taken with a Fizeau configuration using a 50/50 directional coupler (840nm with the Fizeau set-up described in chapter 4) and a Fizeau configuration using a 4-port circulator (1550nm). These can be seen in Fig.6.11 and 6.12.

The peak signal and noise were measured from the scans show in Fig.6.9 and the SNR of each peak was calculated as a comparison for interferometer performance and shown and Table 6.4

Table 6.4 Measured signal and noise values from interference peaks taken from a Fizeau interferometer with a glass slide placed in front of the sample mirror.

Signal from Scan	Signal		Noise	
	Coupler	Circulator	Coupler	Circulator
First reflection from glass slide	0.015 ± 0.005	0.004 ± 0.001	0.01 ± 0.005	0.002 ± 0.001
First reflection from mirror	0.075 ± 0.005	0.025 ± 0.001	0.01 ± 0.005	0.002 ± 0.001
Second reflection from mirror on reflection from back of slide	0.015 ± 0.005	0.003 ± 0.001	0.01 ± 0.005	0.002 ± 0.001

Table 6.5 Signal to Noise Ratio calculated from the interference signals from the glass slide.

Peak from scan	SNR for Interferometer Configuration / dB	
	Coupler	Circulator
First reflection from glass slide	3.5 ± 1.7	6.0 ± 1.8
First reflection from mirror	17.5± 2.0	21.9 ± 2.0
Second reflection from mirror on reflection from back of slide	3.5± 1.7	4.9 ± 1.7

The results from the signals from the glass slide show a decrease in signal on reflection from the glass slide due to the lower reflectivity of the glass and less power is being

reflected back into the system. The results also show an overall increase in SNR with the configuration using the circulator reaching up to 4dB. The circulator is therefore shown to be a valuable component in the Fizeau OCT interferometer.

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Chapter 7

Summary and Conclusion

7.1 Summary

This chapter will summarise the work presented throughout this thesis before going on to draw conclusions from the work. It will also discuss future avenues to explore the work further within this area of study.

Chapter 1 discussed techniques for diagnostic imaging and outlined the problem within this area diagnosing in vivo precancerous tissue¹. Optical Coherence Tomography (OCT) was introduced discussing its advantages compared to the other imaging modalities. The aim of this study was established and the work proposed, with a discussion into the applicability.

An investigation into polarisation induced fading within the tissue sample and within the optical fibres of an OCT system would be carried out, identifying the magnitude and relative affects of each on the signal fading.

Different interferometer designs were proposed to overcome or improve this signal fading and an analysis of the SNR of different interferometer configurations would be carried out to determine the optimal design.

The optimal design would be set up experimentally in order to confirm the theoretical calculations.

Chapter 2 discussed the first applications of OCT in medicine and how these developed over time up to present day. It also highlighted the most important aspects to an OCT system such as scanning speed, methods of scanning, sources used in OCT systems, extensions of OCT systems and novel designs for an OCT interferometer. It also discussed improvements that are continually being made within these areas.

Chapter 3 theoretically analysed how the polarisation state of light changes in a Michelson interferometer either with changes due to scattering², birefringence³ or using non-polarisation maintaining optical fibres⁴, and how this affected the signal to noise ratio for Optical Coherence Tomography systems. By analysing previous work and calculating degrees of polarisation within different tissue types it was decided that the most important change in polarisation induced signal fading was due to disturbances within the optical fibre, likely during endoscopic applications. The Fizeau interferometer⁵ was introduced as an alternative interferometer configuration that would eliminate polarisation induced fading within the sample arm of the OCT interferometer.

In Chapter 4 the advantages and disadvantages of the Fizeau and Michelson interferometers were discussed and it was concluded the Fizeau interferometer would be ideal for endoscopic applications of OCT however the Signal to Noise ratio would always be considerable lower than the conventional Michelson interferometer. This was demonstrated experimentally by setting up both systems and calculating the signal to noise

ratio. All system components were discussed along with results of one-dimensional and two-dimensional imaging.

Chapter 5 carried out a theoretical analysis of interferometer configurations for OCT consisting of variations of the Michelson, Fizeau and Mach Zehnder⁶ designs. By varying the configurations and using new optical components it was shown the Fizeau interferometer can achieve a considerable higher signal to noise than the conventional Michelson design. It also gives a comparable signal to noise ratio to the Michelson designed for power conservation and yet eliminated any polarisation induced fading in the sample arm.

Chapter 6 discussed the optimal Fizeau configuration determined in chapter 5 and discussed each component necessary for the system in detail. The optimal configuration was set-up experimentally and the signal to noise ratio calculated theoretically for the equipment used and experimentally with the interference signals obtained within the system. The experimental values were systematically lower than the theoretical values but showed the same trend for an increase in signal to noise.

7.2 Conclusions

At present screening procedures for patients suffering with Barretts esophagus⁷ involve random biopsies taken every six months to determine whether there are any abnormal changes within the cells. Common imaging techniques⁸ such as x-ray, magnetic resonance imaging, computed tomography and ultrasound have insufficient resolution to identify any individual cell changes especially with gastrointestinal tissue. Microscopy has offered a high resolution imaging technique and in endoscopic form can image the GI tract easily with micrometer resolution allowing histopathological visualisation of tissue down to cellular and sub-cellular levels. This imaging technique is limited in depth resolution and is

only able to provide information about the surface layer of tissue, therefore any changes within the sub-surface layer will go undetected ie epithelial layer which is especially important for precancerous changes.

Optical Coherence Tomography (OCT) is a high resolution imaging technique⁹ that can be used to image tissue at depth rather than just the surface. It uses Low Coherence Interferometry to produce images of micron resolution down to depths of 2-3mm. With State of the art OCT imaging equipment individual cells at these depths within the epithelial and sub-mucosa layers can be imaged and any cancerous changes identified without any need for biopsy. Standard OCT systems incapable of achieving sub-cellular resolution maybe used to image changes in thickness between the epithelial and submucosa layers, where cancer begins. The thickness of the epithelial layer can increase from 50-70 μ m for dysplasia and up to 120 μ m for cancerous tissue¹⁰. These changes in thickness can easily be detected by OCT and improve the accuracy for biopsy sites.

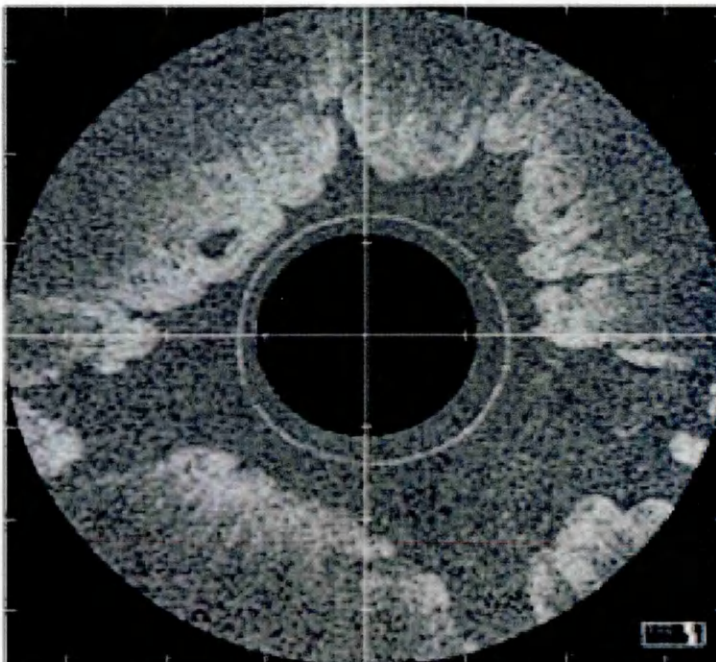


Fig.7.1. Endoscopic image of gastrointestinal tissue

Using Michelson interferometer and due to temperature and phase changes as the optical fibre enters the esophagus the light reflected back may have a different polarisation state than the light within the reference arm of the interferometer. This has shown to occur in a random manner and has shown to cause a possible decrease in signal to noise to zero visibility. There are other factors within the OCT system that may change the state of polarisation of the light within the sample arm such as scattering off tissue and tissue birefringence. The uncertainty in polarisation induced fading due to optical fibres was shown to give the greatest decrease in signal for imaging depths up to 2mm.

The Fizeau interferometer is an alternative configuration for endoscopic OCT applications. Since the reference beam is created by the reflection of light from the end of the fibre within the sample arm then any polarisation changes will be experienced by both reference and sample arms and give maximum signal at the detector.

The Fizeau interferometer requires a second processing interferometer in order to detect the interference fringes and this results in the Fizeau interferometer always having a lower signal to noise ratio than the conventional Michelson interferometer.

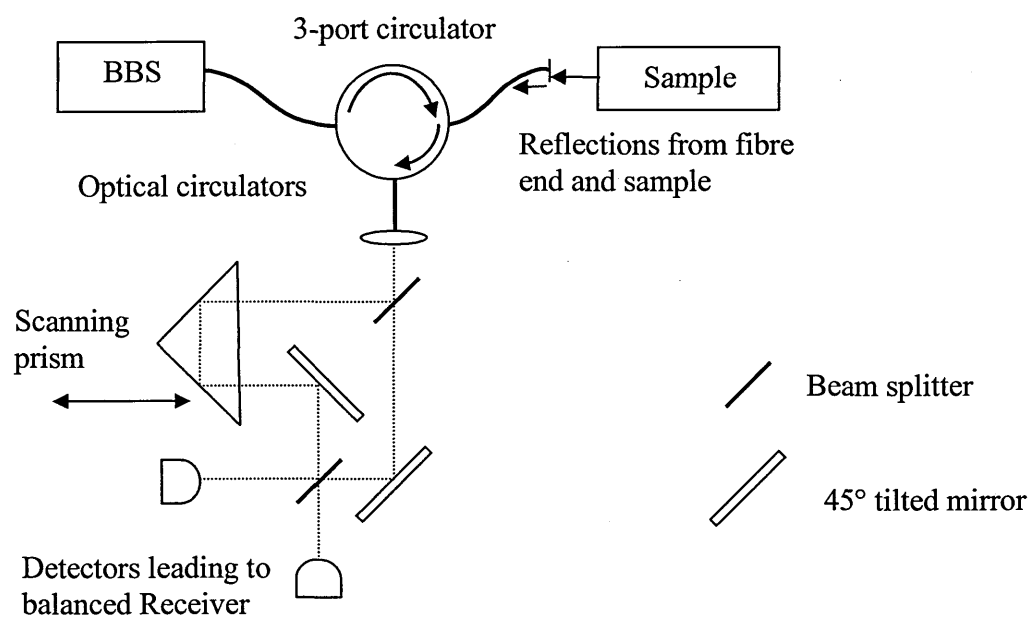
An analysis of a number of Fizeau configurations was carried out using power conserving optical components and a variation of Fizeau interferometer was theoretically shown to give a signal to noise ratio considerably higher than the conventional and comparable to a Michelson designed for power conservation. This consisted of using a 4-port circulator and balanced detection.

The components used within this configuration were discussed and chosen in order to experimentally set-up the optimal design. The signal to noise ratio was calculated theoretically and experimentally for the optimal design using the experimental components. Theoretically the signal to noise was increased using the power conserving optical components, this was also shown experimentally with a systematic reduction in signal to noise compared with the theoretical results, almost certainly due to losses within the set-up and launching light into the optical fibre.

This project has shown that the Fizeau interferometer can improve the performance of endoscopic OCT imaging compared with that of the conventional Michelson interferometer without compromising any decrease in signal to noise due to polarisation fading effects.

7.3 Future Work

7.3.1 Using a different processing interferometer design



Mach Zehnder Receiving interferometer

Fig.7.2 Fizeau set-up using a bulk optic Mach Zehnder receiving interferometer.

The use of a circulator in a balanced detection configuration may cause problems with data collection as the light re-entering the circulator from the Michelson interferometer is

difficult to collect and launch back into the fibre. To overcome this problem a different interferometer configuration could be used instead of the Michelson and an ideal configuration would be a bulk optic Mach Zehnder interferometer as shown in Fig.7.2.

Light exiting through the third port of the circulator enters the Mach Zehnder processing interferometer and is split 50/50 by a beam splitter, 50% of the light is reflected from a mirror onto a second beam splitter and 50% of the light is internally reflected through a glass prism. The prism scans backwards and forwards and is used as the scanning arm to match the path length of the reference and the sample beam. Both beams recombine at a second beam splitter and interference signals can be collected at two detectors allowing balanced detection to be performed.

7.3.2 Fibre Bundle Configuration

In order to improve the OCT design further the possibility of incorporating a fibre bundle into the interferometer reference arm, instead of a single fibre. The advantage of a fibre bundle is that larger area scans can be produced and images obtained faster. Using a fibre bundle less longitudinal scans will need to be performed and lateral scans can be performed simultaneously. Fibre bundles consist of up to twenty thousand optical fibres, each individual fibre acting as an individual sensing arm. The bundle of fibres will be incorporated into the sample arm and for endoscopic applications will be the arm placed down the esophagus to image the tissue.

Each fibre in the bundle will collect individual reflections from different locations within the tissue (as with each fibre in the conventional OCT set-up). The returning light from each fibre will enter an interferometric processing system where the reflections will interfere with a light beam reflected from the reference mirror (when the path lengths are equal or to within the coherence length of the source). The interference signals from each

fibre will be detected using a Charge Coupling Device CCD-camera and processed to create two-dimensional images. Each element in the CCD array would be aligned to each fibre within the bundle, however because of the added dimension of each fibre and its cladding, there will be information lost between the fibres due to lack of data collection. To over come this problem, piezoelectric transducers (PZT) can be fitted to the fibre head in order to initiate small orthogonal movements so that measurements can be taken in the intermediate positions between the fibres (Fig7.4).

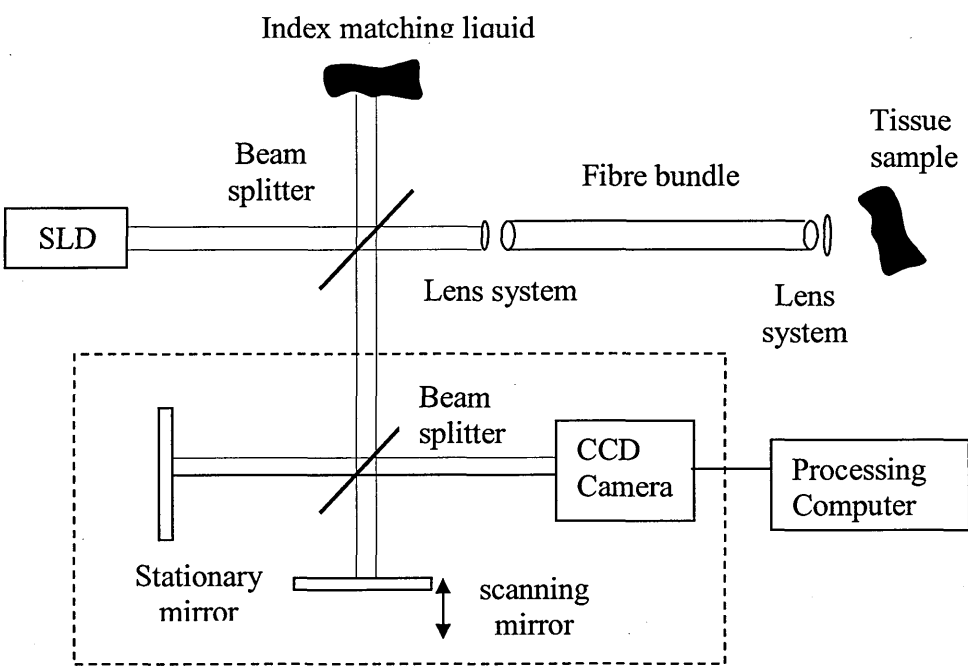


Fig.7.3 Showing the fibre bundle set-up incorporated into the sample arm Fizeau configuration.

The proposed fibre bundle set-up, for the sample arm can be seen below, Fig.7.4: (a) Side view of the endoscope sample arm, (b) Front view of the fibre bundle sample arm

SIDE VIEW

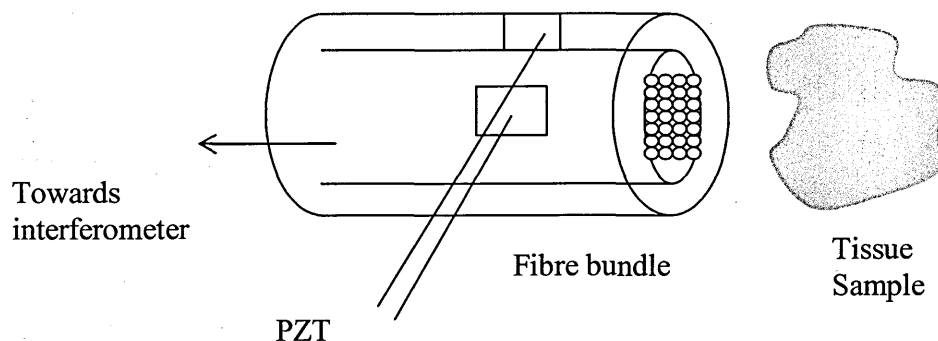
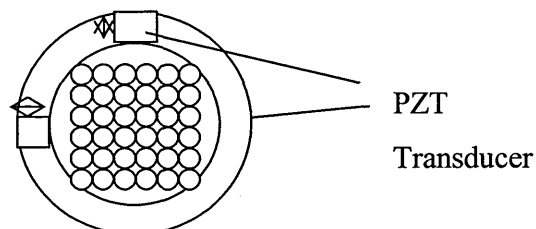


Fig.7.4 showing the fibre bundle set-up incorporated into the sample arm. (a) Side view, showing how each fibre collects reflections from different locations within the tissue.

FRONT VIEW



(b) Front view of the sample arm, showing how the PZT moves the fibre bundle small distances to collect intermediate reflections within the tissue.

One problem with using this technique maybe the difference in relative lengths of the optical fibres within the bundle and one solution to this maybe to carry out a calibration to correct for this. Further investigation would be necessary to see if this affect is negligible.

The scope of this thesis has been somewhat general in interferometry both theory and experimentation with very little medical application. Biomedical optics has a great future in the medical field as it offers a safer alternative to other techniques for example those which use ionising radiation and allows images of miron resolution to be obtained.

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Appendix A

Appendix A contains a detail analysis of Polarisation Sensitive Optical Coherence

Tomography (OCT)

POLARISATION SENSITIVE OCT

The signal from a low coherence interference signal can be split at the detector into its horizontal and vertical components within its polarisation state.

The signal intensity from each polarisation component is given by;

$$\text{Horizontal} = A_H = \sqrt{R_{(z)}} \cos(2k_0\Delta z + 2\alpha) \exp\left(-\left(\frac{\Omega\Delta z}{c}\right)^2\right) \sin(k_0 z \delta) \quad A1$$

$$\text{Vertical} = A_V = \sqrt{R_{(z)}} \cos(2k_0\Delta z) \exp\left(-\left(\frac{\Omega\Delta z}{c}\right)^2\right) \cos(k_0 z \delta) \quad A2$$

Where Δz is the path length difference between the two interferometer arms, z is the depth of the light reflected from the sample, $R_{(z)}$ is the reflectivity at depth z , c is the speed of light in a vacuum, k_0 is the attenuation of the coherent beam by scattering (Equation A3), δ is the tissue birefringence (Equation A4), α is the angle of the fast optical axis measured with respect to the vertical and Ω is given by Equation A5.

$$k_0 = \frac{2\pi}{\lambda_0} \quad A3$$

$$\delta = n_s - n_f \quad A4$$

$$\Omega = \frac{\Delta\lambda c}{\lambda_0^2 \sqrt{\ln 2}} \quad A5$$

Where $\Delta\lambda$ is the bandwidth of the source, λ_0 is the centre wavelength and n_s and n_f re the refractive indices along the fast and slow axes of the tissue respectively.

From the vertical and horizontal components detected the carrier frequency $[\cos(2k_0\Delta z)]$ within the coherence envelope $[\exp(-(\Omega\Delta z/c)^2)]$ is filtered out leaving a signal oscillating as a function of birefringence and depth. Assigning a grey scale value to the common log (I_s) of both polarisation channels (I_H and I_V), Equation.A6, and plotting as a function of depth forms the polarisation sensitive images.

$$\log_{10} I_{s(z)} = \log_{10} (I_{H(z)} + I_{V(z)}) \quad A6$$

Where

$$I_{H(z)} = R_{(z)} \sin^2(k_0 z \delta) \quad A7$$

$$I_{V(z)} = R_{(z)} \cos^2(k_0 z \delta) \quad A8$$

Appendix B

Appendix B contains a details of the Derivation of Polarisation Induced Fading in
Fibres

APPENDIX B

Derivation of Polarisation Induced Fading in Fibres

The photocurrent at a detector is given by B1

$$I_d = \rho \left[P_r + P_s + P_x + 2\sqrt{P_r P_s} \cos(k_0 \Delta l) \right] \quad \text{B1}$$

Where ρ is the detector respositivity, P_r , P_s and P_x are the powers of the reference, sample and incoherent light, k_0 is the centre wave number of the optical source and Δl is the optical path length difference between the reference and the sample arms.

The signal photocurrent is made up of the intereference AC signal and the offset DC signal

$$I_{DC} = \rho (P_r + P_x + P_s) \quad \text{B2}$$

$$I_{AC} = 2\rho\sqrt{P_r P_s} \cos(k_0 \Delta l) \quad \text{B3}$$

When the polarisation states of the two interferometer arms differ there is a decrease in the signal to noise ratio, this is because the DC component is increasing and the AC component decreasing.

In order to theoretically understand how the SNR varies with polarisation mis-match between the two interferometer arms, the polarisation state of one of the interferometer arms must stay constant (reference arm) and the other one is varied (sample arm). The power in the sample arm can be split into coherent $[P_{s(\text{coh})}]$ and incoherent $[P_{s(\text{incoh})}]$ components, the total power in the sample arm is given by the sum of these components.

$$P_{s(\text{total})} = P_{s(\text{coh})} + P_{s(\text{incoh})} \quad \text{B4}$$

As the polarisation in the sample arm varies from the reference arm, the incoherent component of the sample arm power increases.

The AC and DC components now become:

$$I_{DC} = \rho [P_r + P_x + P_{s(\text{coh})} + P_{s(\text{incoh})}] \quad B5$$

$$I_{AC} = 2\rho \sqrt{P_r P_{s(\text{coh})}} \cos(k_0 \Delta l) \quad B6$$

The SNR can be calculated using B7:

$$SNR = 10 \log \left(\frac{I_{AC}}{\text{Noise}} \right) \quad B7$$

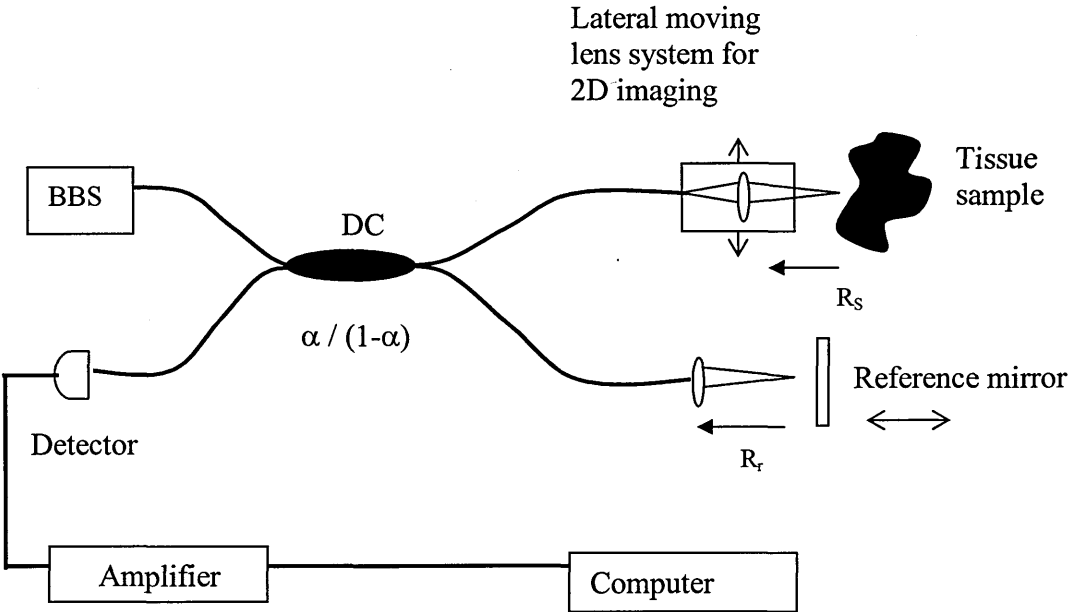
The ammount of coherent and incoherent power from the sample arm was varied systematically and the corresponding SNR calculated.

Appendix C

Appendix C contains a detail analysis of the Signal Intensity from a Michelson
Interferometer and Interference between two beams

Signal Intensity for a Michelson Interferometer

Fig.C1 The Michelson Interferometer



The electric field component of the wave from the SLD is given by E_0

The component of light incident on the detector, returning from the reference arm of length l_r , with mirror reflectivity R_r and coupler efficiency $(1-\alpha)$ to reference arm and α to the detector the electric field is given by B1:

$$E_r = E_0 \alpha (1-\alpha) R_r \exp i(2kl_r) \tag{C1}$$

The component of light incident on the detector returning from the sample arm of length l_s and tissue reflectivity of R_s and coupler efficiency α to sample arm $(1-\alpha)$ to the detector, the electric field is given by:

$$E_s = E_0 (1 - \alpha) \alpha R_s \exp i(2kl_s) \quad C2$$

The total electric field E_T at the detector is given by the sum of the combining fields, E_r and E_s :

$$E_T = E_r + E_s = E_0 (1 - \alpha) \alpha R_r \exp i(2kl_r) + E_0 \alpha (1 - \alpha) R_s \exp i(2kl_s) \quad C3$$

The Intensity at the detector is given by the magnitude squared of the combined fields:

$$I_d = |E_r + E_s|^2 \quad C4$$

$$I_d = E_r^2 + E_s^2 + E_r E_s^* + E_r^* E_s \quad C5$$

$$I_d = E_0^2 \alpha^2 (1 - \alpha)^2 R_r^2 \exp i(4kl_r) + E_0^2 (1 - \alpha)^2 \alpha^2 R_s^2 \exp i(4kl_s) + E_0^2 \alpha^2 (1 - \alpha)^2 R_r R_s \left[\exp i2k(l_r - l_s) + \exp i2k(l_s - l_r) \right] \quad C6$$

Where:

$$E_0^2 \alpha^2 (1 - \alpha)^2 R_r^2 \exp i(4kl_r) = I_r = \rho P_r \quad C7$$

$$E_0^2 \alpha^2 (1 - \alpha)^2 R_s^2 \exp i(4kl_s) = I_s = \rho P_s \quad C8$$

ρ is the responsivity of the detector and P_r and P_s are the powers in the reference and sample arms respectively

Substituting and changing to polar form gives:

$$I_d = I_r + I_s + 2\sqrt{I_r I_s} \cos k(l_r - l_s) = I_r + I_s + 2\sqrt{I_r I_s} \cos k\Delta L \quad C9$$

Where

$$(l_r - l_s) = \Delta l \quad C10$$

The Photocurrent is dependant on the responsitivity ρ of the detector and therefor the total photocurrent becomes:

$$I_d = \rho [P_r + P_s + 2\sqrt{P_r P_s} \cos(k\Delta L)] \quad C11$$

Appendix E

Appendix E contains the data sheets for the PIN photodiode (BPX65)

VII. SPECIFICATIONS

■ MAXIMUM RATINGS

Parameter	Value	Unit
Positive Supply Voltage	+16	V
Negative Supply Voltage	-16	V
Operating Temperature	0 to +60	°C
Storage Temperature	-55 to +100	°C

■ SPECIFICATIONS (Typical at Ta=25°C, Vcc=±12V, unless otherwise specified)

Photoelectric Conversion Section (APD)

Parameter	C5460	C5460-01	Unit
Active Area	φ1.5	φ3.0	mm
Spectral Response	400 to 1000		nm
Peak Wavelength	800		nm
Photo Sensitivity (800nm, gain=1)	0.5		A/W
Temperature Stability of Gain* (25°C±10°C, gain=30)	±2.5 Typ.	±5 Max.	%

* The gain is preset to "30" when shipping.

High-speed Amplifier Section (C5460)

Parameter	Min.	Typ.	Max.	Unit
Cut-off Frequency (-3dB)	High	9	10	MHz
	Low	-	DC	
NEP (800nm)	-	0.2	0.4	pw/Hz ^{1/2} @10MHz
Feedback Resistance	-	10	-	KΩ
Photoelectric Sensitivity* (including APD, 800nm, gain=30)	1.4	1.5	1.6	10 ⁶ V/W
Maximum Input Light Intensity	5.0	6.0	-	μW
Minimum Detection Limit	-	0.8	1.6	nW

* The gain is preset to "30" when shipping.

High-speed Amplifier Section (C5460-01)

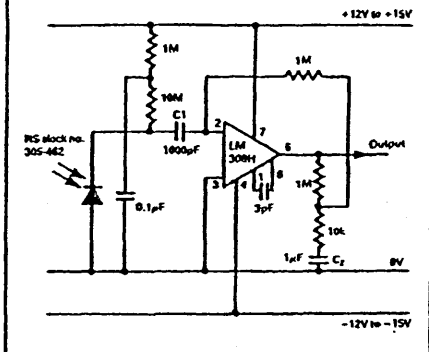
Parameter	Min.	Typ.	Max.	Unit
Cut-off Frequency (-3dB)	High	80	100	kHz
	Low	-	DC	
NEP (800nm)	-	0.02	0.04	pw/Hz ^{1/2} @100kHz
Feedback Resistance	-	10	-	MΩ
Photoelectric Sensitivity* (including APD, 800nm, gain=30)	-1.4	-1.5	-1.6	10 ⁶ V/W
Maximum Input Light Intensity	0.05	0.06	-	μW
Minimum Detection Limit	-	0.005	0.01	nW

* The gain is preset to "30" when shipping.

■ GENERAL RATINGS

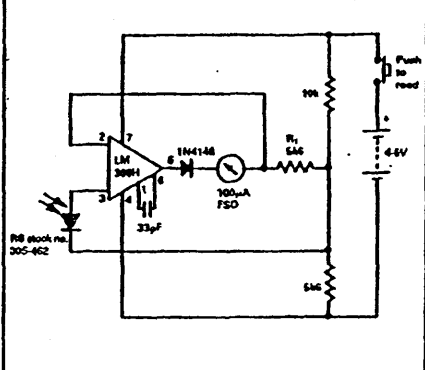
Parameter	C5460			C5460-01			Unit
Supply Voltage (+12V)	+11.4	+12	+12.6	+11.4	+12	+12.6	V
	-11.4	-12	-12.6	-11.4	-12	-12.6	
Current Consumption (+12V)	+30	+45		+35	+45		mA
	-11	-16		-11	-16		
Board Dimensions	80X50X23						mm
Weight	52						g

Figure 15 ac coupled analogue sensor for optical communications links



Logarithmic detector for exposure meter
Feeding the RS stock no. 305-462 into a high impedance gives a logarithmic voltage/illumination response.
The circuit shown is the basis for a simple battery-operated exposure meter. At very low light levels where the amplifier bias current may cause the output to go negative, a diode avoids the spurious state of negative indication.
The movement may be calibrated in photographic scales, one stop being approximately 7µA. Sensitivity can be trimmed by adjusting R1.

Figure 16 Logarithmic detector for exposure meter



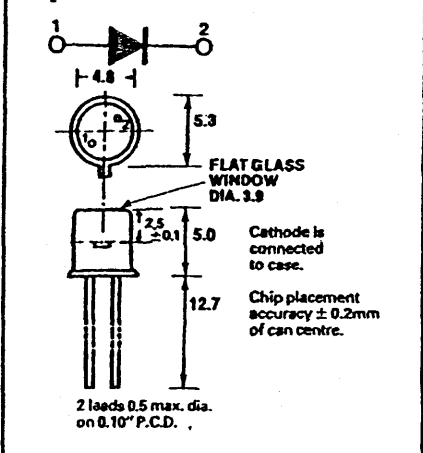
BPX 65 high speed photodiode
(RS stock no. 304-346)

The BPX 65 is a planar silicon PIN photodiode housed in a modified TO-18 case incorporating a plain glass flat window which has no influence on the beam path of optical lens systems. The cathode is electrically connected to the case. Because the BPX 65 is capable of detecting wide bandwidth signals due to its excellent high frequency response, this coupled with its high sensitivity makes the device ideal for signal detection applications. This photodiode is outstanding for low junction capacitance and short switching times.

Absolute maximum ratings

at +25°C (unless stated)
Reverse voltage V_R _____ 50V
Forward current I_F _____ 10mA (200mA pulsed 1µs 1% duty cycle)
Operating temperature range _____ -25°C to +70°C
Storage temperature range _____ -55°C to +125°C
Junction temp. T_J _____ +125°C
Power dissipation P_d _____ 250mW
(derate linearly 2.5mW/°C above +25°C)

Shape and dimensions



Electrical characteristics at +25°C (unless stated)

	Parameter	Test conditions	min.	typ.	max.	units
A	Radiant sensitive area			1		mm ²
λ_s max.	Wavelength of max. sensitivity			850		nm
R_s	Responsivity	$\lambda = 450\text{nm}$		0.2		A/W
		$\lambda = 900\text{nm}$		0.55		A/W
		$\lambda = 1064\text{nm}$		0.15		A/W
t_r	Response time (10-90% levels)	$R_L = 50\Omega$; $V_R = 20\text{V}$; $\lambda = 900\text{nm}$		0.5	1	ns
C_0	Capacitance $V_R = 0\text{V}$			15		pF
C_1	$V_R = 1\text{V}$			12		pF
C_{20}	$V_R = 20\text{V}$			3.5		pF
f_c	Cut-off frequency			500		MHz
I_p	Dark current	$V_R = 20\text{V}$, Dark ($E = 0$)		1	5	nA
S	Spectral sensitivity	$V_R = 20\text{V}$; see Note 1	7	10		nA/Lx
NEP	Noise equivalent power	$V_R = 20\text{V}$		3.6×10^{-11}		W/Hz

Note 1. The illuminance indicated refers to unfiltered radiation of a tungsten filament lamp at a colour temperature of 2856K (standard light A in accordance with DIN 5033 and IEC publ. 306-1).

Typical performance curves

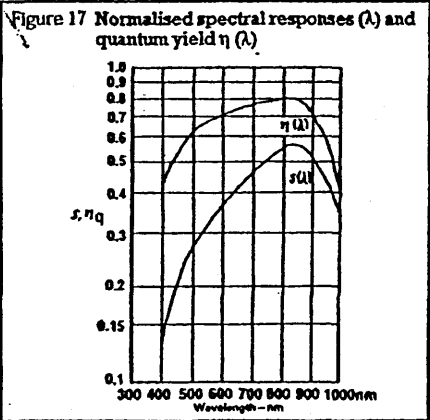
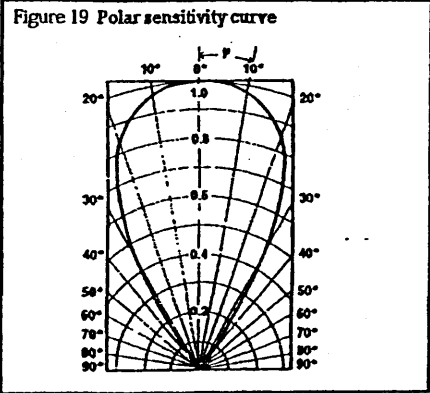
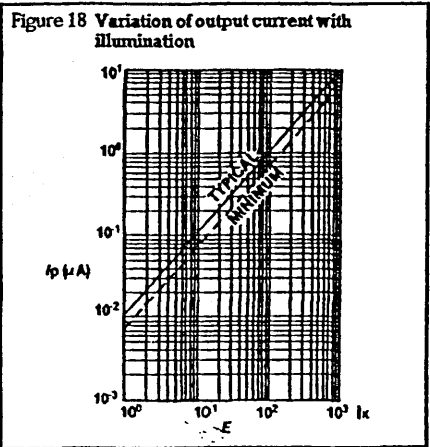
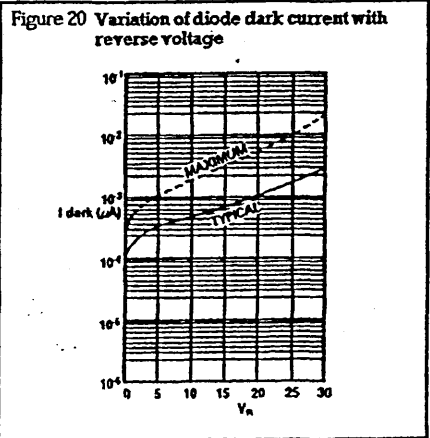


Figure 21 Variation of output current with temperature

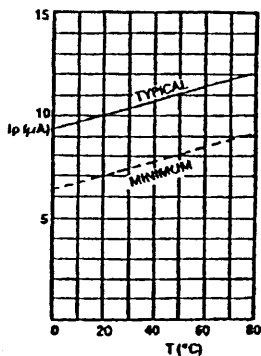


Figure 23 Permissible power dissipation

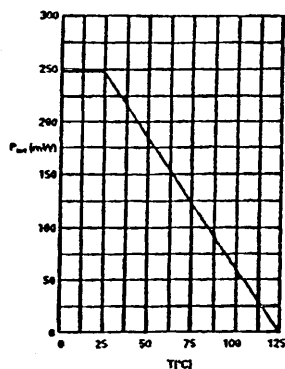


Figure 22 Variation of diode capacitance with reverse voltage

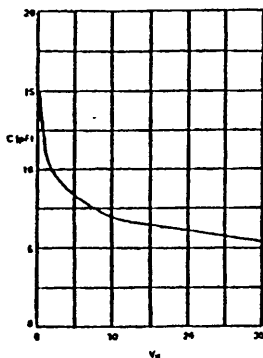
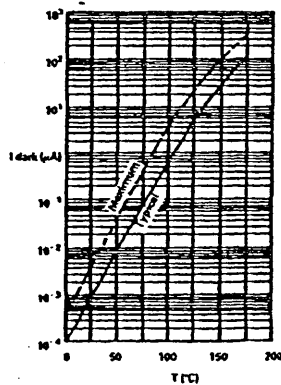


Figure 24 Variation of dark current with diode temperature



BPW21 photodiode

(RS stock no. 303-719)

A silicon photodiode housed in an hermetically sealed case with a flat window incorporating built-in colour correction. Sensitivity approximating the human eye response. Linear current (short circuit) versus illumination. Log. voltage versus illumination. This photodiode is designed for use in the photoamperic mode and is ideally suited for use in light monitoring and control, optical instrumentation and camera control.

Absolute maximum ratings

Ambient temperature range -25°C to +100°C
Reverse voltage, V_R 10V
Open circuit voltage 650mV
Power dissipation (at 25°C), P_d 250mW
Illuminance, E 10⁶ lux

Features

- Response approximating to the human eye
- Photovoltaic cell operation
- Linear output current versus illumination
- Hermetically sealed TO5 case.

Appendix F

Appendix F contains details of the LabView program written for two-dimensional
Imaging

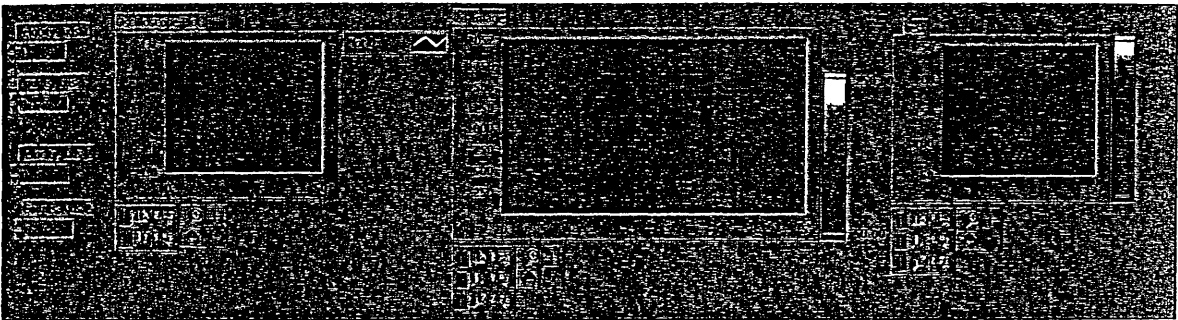
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Printed on 05/09/2003 at 16:48

Connector Pane



2D Image program-print.vi

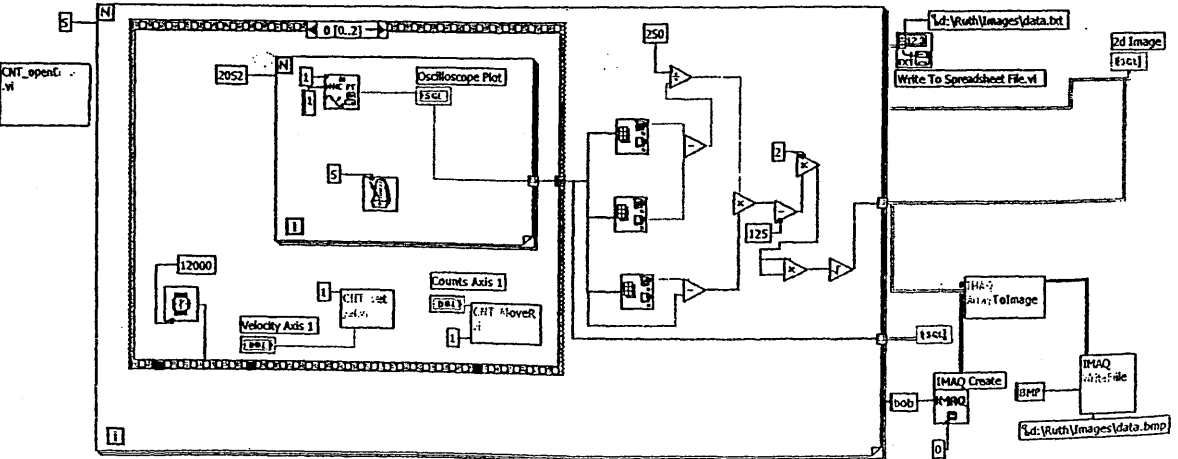
Front Panel



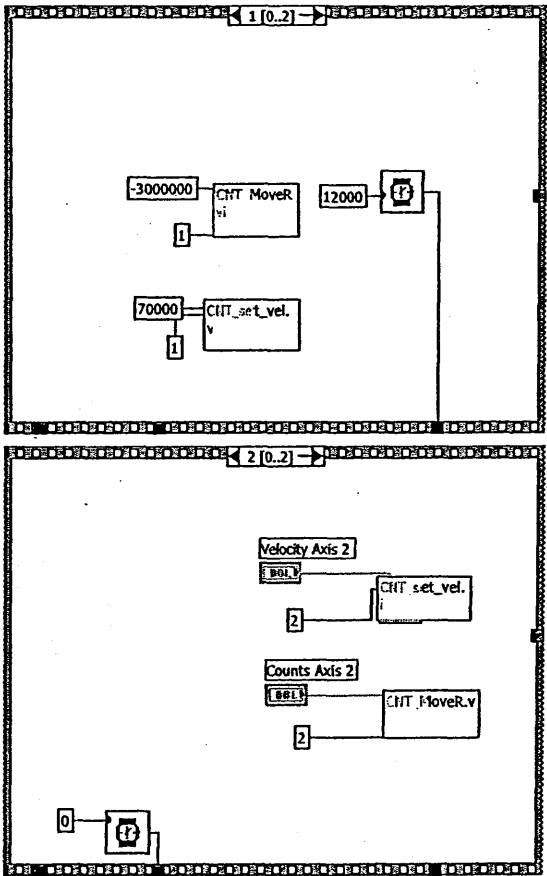
Controls and Indicators

-  Velocity Axis 1
-  Counts Axis 1
-  Velocity Axis 2
-  Counts Axis 2
-  Oscilloscope Plot
-  2d Image
- 

Block Diagram

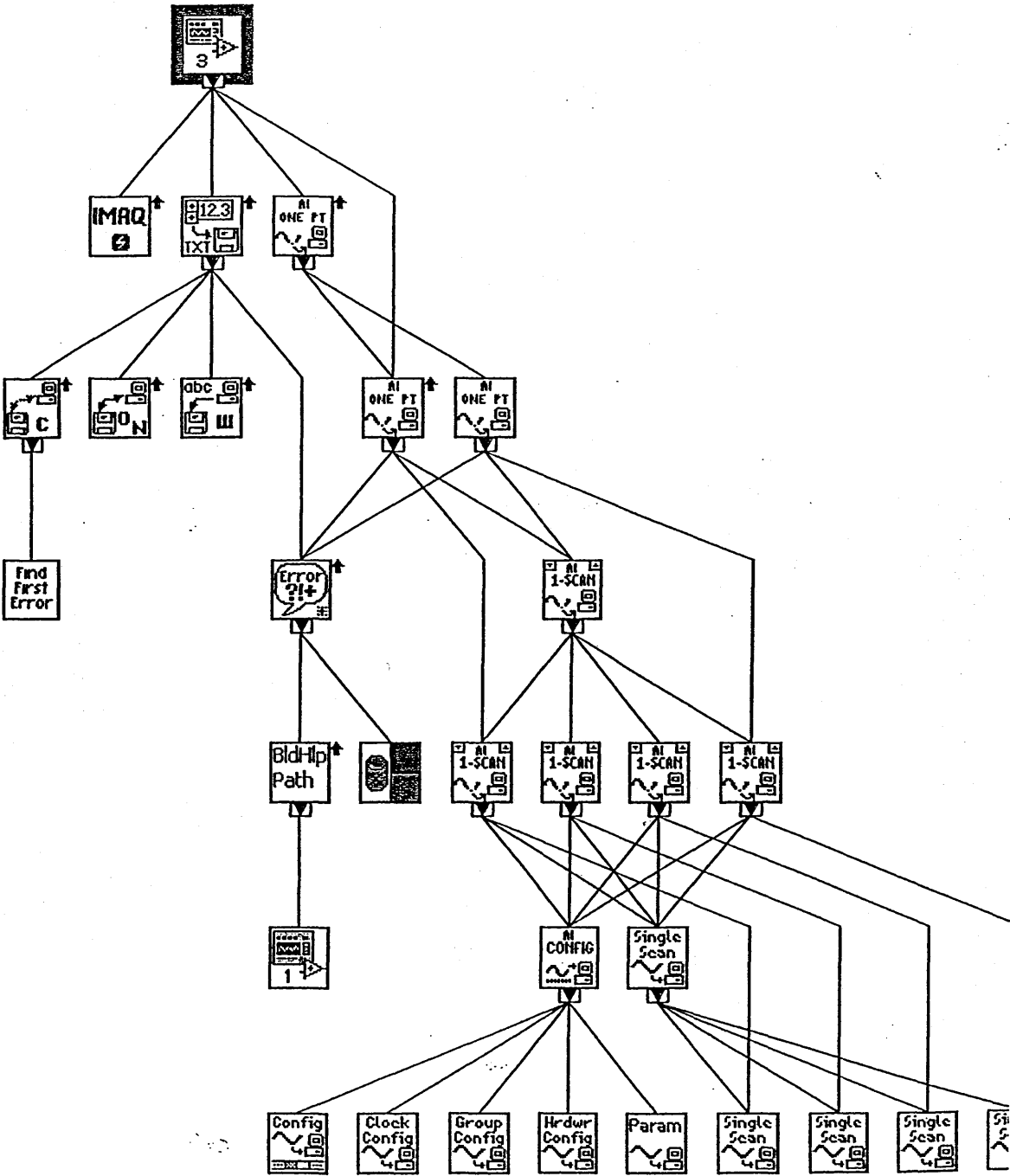


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Position in Hierarchy

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List of SubVIs



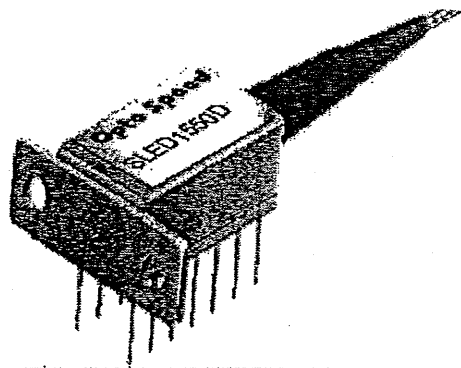
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C:\Program Files\National Instruments\LabVIEW 6.1\vi.lib\Utility\file.lib\Write To Spreadsheet File.vi



IMAQ Create
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Appendix G

Appendix G contains specifications for the SLED 1550D



REV 4.0 / 15.04.02

SLED 1550D

HIGH POWER**BROAD BANDWIDTH****1550 NM****SUPERLUMINESCENT LED****SUMMARY****SPECIFICATIONS**

Note: Please note that the summary specifications give an overview on all the different SLED types per wavelength. The release number and release date of the summary specifications may be different from the specifications of the different SLED types. Please refer to the release number and release date given below when you place an order for a specific SLED type.

SLED1550D1A (Specifications 4.0 / 15.04.02)

SLED1550D5A (Specifications 4.0 / 15.04.02)

SLED1550D10A (Specifications 4.0 / 15.04.02)

SLED1550D20A (Specifications 4.0 / 15.04.02)

Opto Speed™

Opto Speed Zürich AG
Moosstrasse 2
CH-8803 Rüschlikon, Switzerland
Tel. +41 1 704 62 62
Fax +41 1 704 62 64
sales@optospeed.com
www.optospeed.com
VAT 488 639

A. ABSOLUTE MAXIMUM RATINGS

Stresses beyond the absolute maximum ratings may cause permanent damage to the device. Exposure to absolute maximum rating conditions for extended periods may affect device reliability.

Parameter	Symbol	Cond.	Min	Max	Unit
Reverse voltage	V_R			2	V
Forward current	I_F			$I_{S,max} + 50$	mA
Forward voltage	V_F	$I_{S,max}$		2.5	V
Storage temperature (no condensation)	T_{slg}		-10	50	°C
Operating temperature (no condensation)	T_{op}	$I_{S,max}$	0	40	°C
Storage humidity (no condensation)				97	% r.h.
Operating humidity (no condensation)				85	% r.h.
Maximum optical feedback	P_b			1	mW

B. SPECIFICATIONS ($T_{SLED} = 20^{\circ}C$)

MODEL	SLED1550D1A			SLED1550D5A			SLED1550D10A			SLED1550D20A			Unit
	Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	
Supply current IS	0		150	0		180	0		200	0		350	mA
Power in SMF	0.05	0.08		1.5	2.5		5	8		13	18		mW
Bandwith FWHM	90	110		50	60		25	35		25	35		nm
Central wavelength	1480	1500	1520	1530	1550	1570	1530	1550	1570	1530	1550	1570	nm
Spectral ripple		0.1	0.2		0.1	0.2		0.15	0.3		0.3	0.5	dB

C. IMPORTANT NOTES

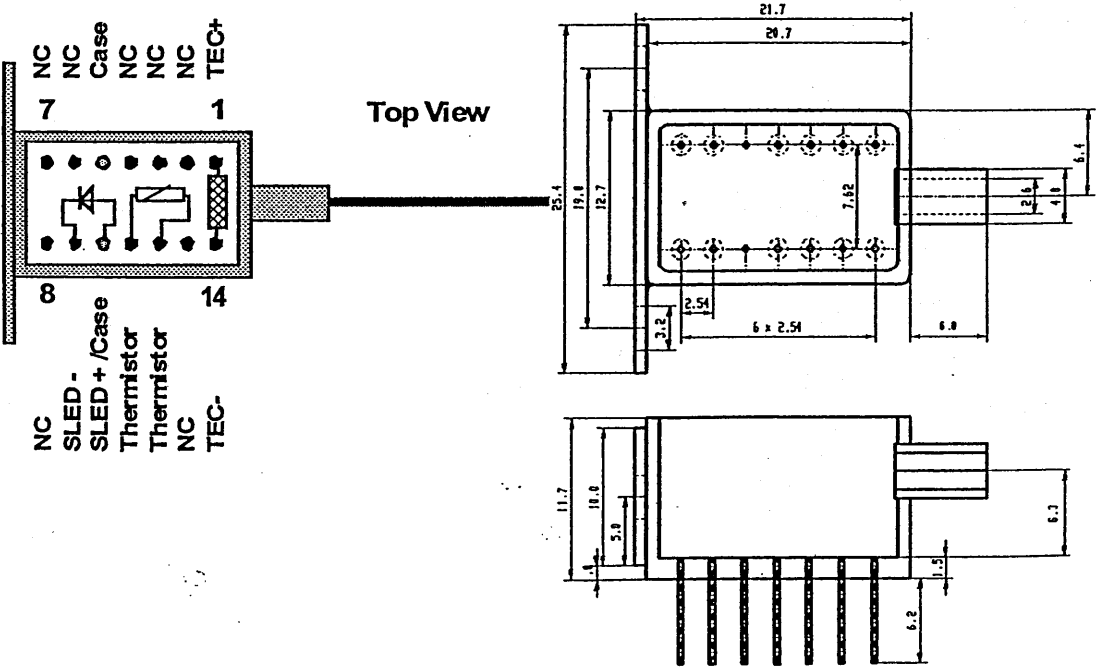
- 1. Avoid electrostatic discharges, which may destroy the SLED.
- 2. The device is intended for laboratory use only.
- 3. The module is not designed to work in aggressive atmosphere.
- 4. The module is not guaranteed for operation under vibration conditions.
- 5. Mechanical shocks may destroy the module.
- 6. Never use the module without heat sinking.
- 7. Adequate eye protection against laser radiation should be used while handling and operating the module.
- 8. The module must not be used in applications where human life may be endangered.
- 9. Opto Speed declines any responsibility if the device is used in applications where human life may be endangered.

D. PACKAGE

Parameter	Symbol	Cond.	Unit	Min	Typ	Max
Termistor resistance (at 25 °C)	R_{th}		kOhm		10	
Thermistor constant (tol. 0.4% in 0-50 °C)	B		K		3892	
Thermistor self heating	H_{th}		°C/mW		1.0	
TEC current	I_{TEC}		A			1.6
TEC voltage	V_{TEC}		V			2.4

Part	Description
Fibre	SMF-28, 9/125/250 μm ¹⁾
Jacket	0.9 mm loose tube
Fiber pigtail length	1 m
Optical connector	Optional ²⁾
Dimensions	See figure

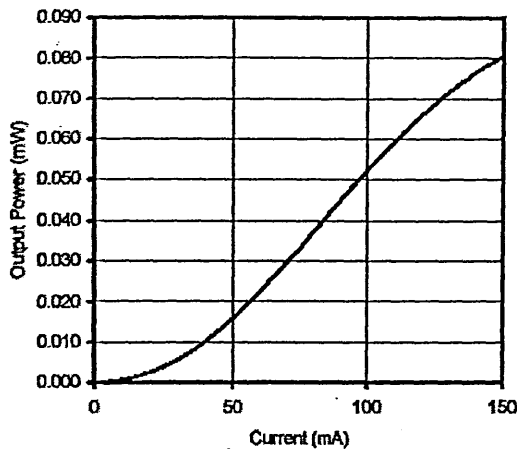
- 1) Other fibres possible upon request
- 2) FC APC, DIN APC, E2000 APC and others possible upon request



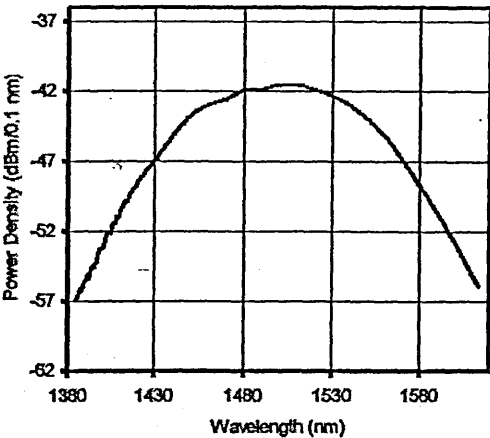
E. TYPICAL CURVES

SLED1550D1A

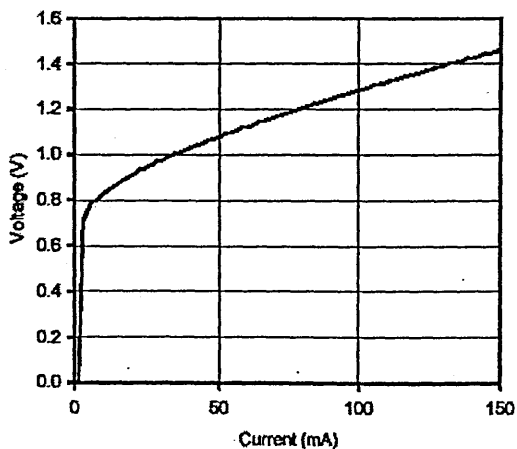
Module output power



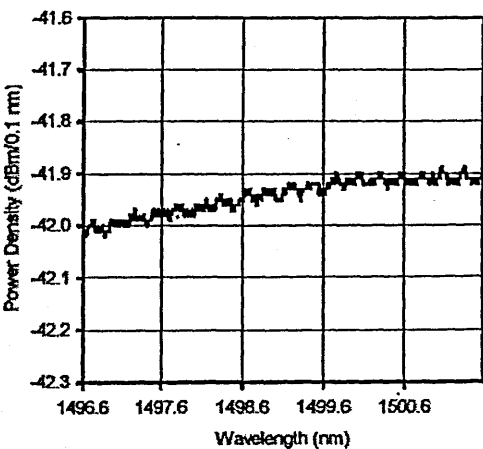
Spectrum at 150 mA



Voltage



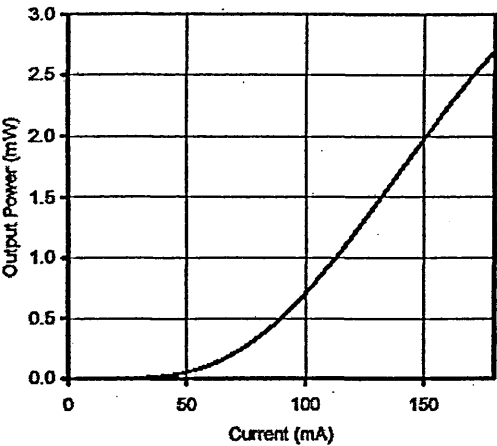
Ripple at 150 mA



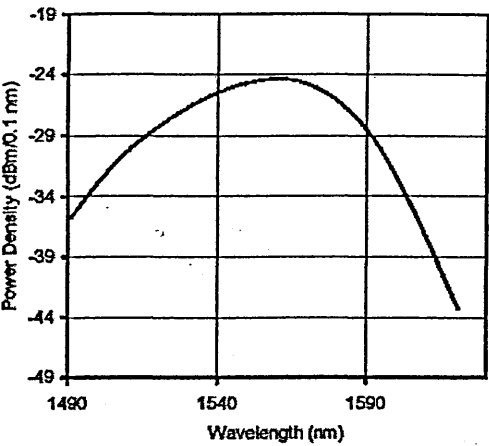
Temperature:	20.00 (°C)
Max Power:	0.08 (mW)
Peak Wavel.:	1505.93 (nm)
Central Wavel.:	1499.12 (nm)
Spectral Width:	109.51 (nm)
Max Ripple:	0.02 (dB)

SLED1550D5A

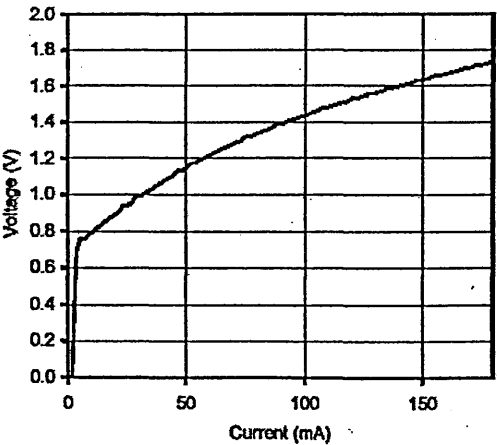
Module output power



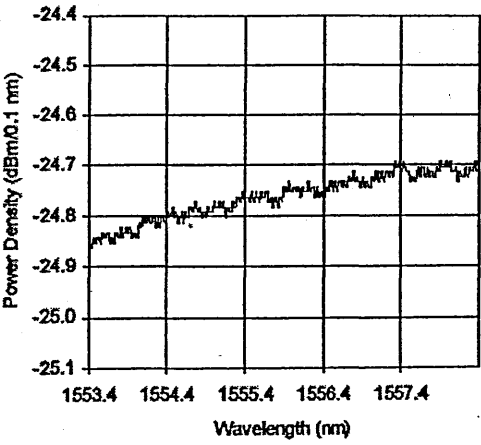
Spectrum at 180 mA



Voltage



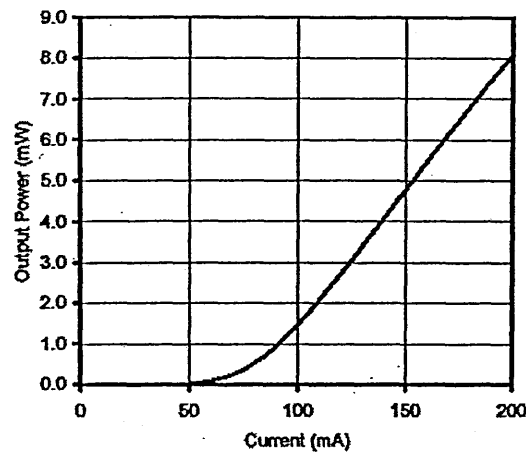
Ripple at 180 mA



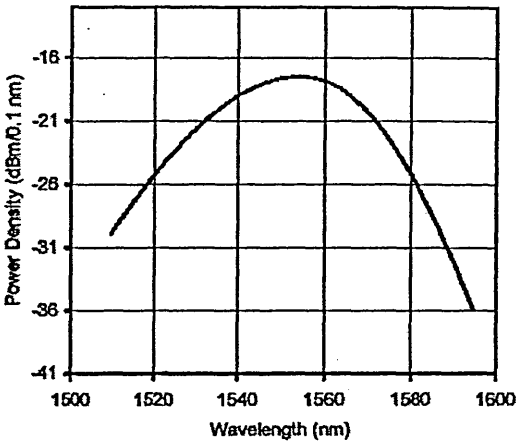
Temperature:	20.00 (°C)
Max Power:	2.7 (mW)
Peak Wavel.:	1554.99 (nm)
Central Wavel.:	1550.85 (nm)
Spectral Width:	59.25 (nm)
Max Ripple:	0.03 (dB)

SLED1550D10A

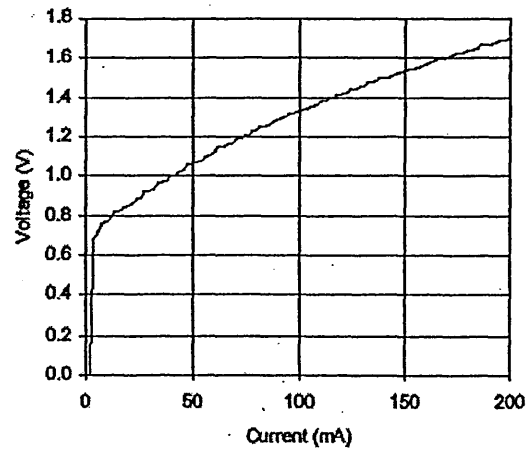
Module output power



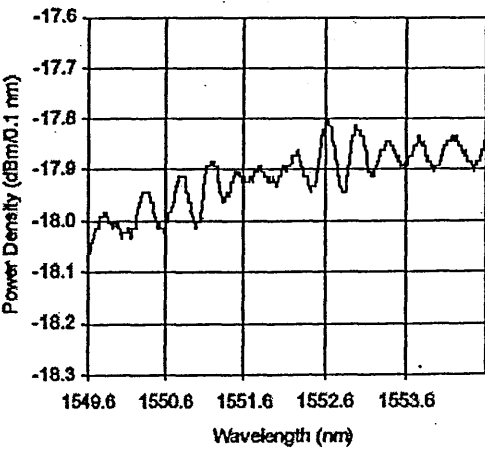
Spectrum at 200 mA



Voltage



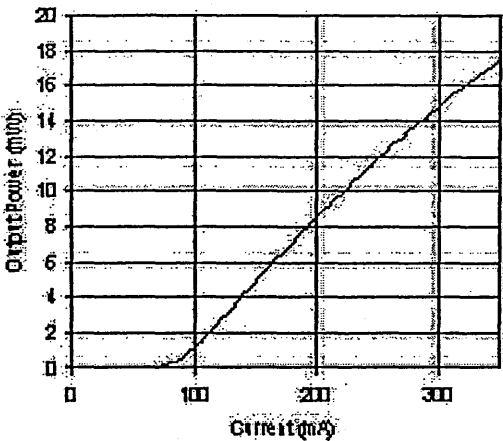
Ripple at 200 mA



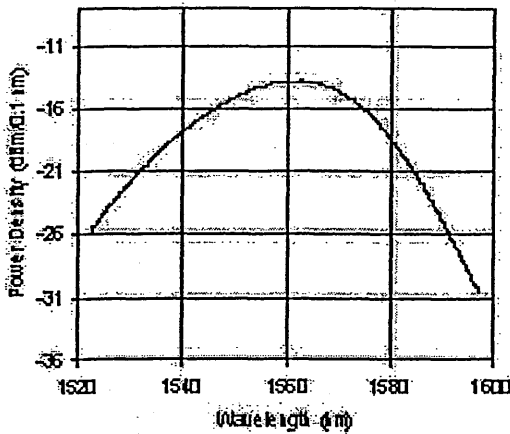
Temperature:	20.00 (°C)
Max Power:	8.1 (mW)
Peak Wavel.:	1554.84 (nm)
Central Wavel.:	1552.13 (nm)
Spectral Width:	36.42 (nm)
Max Ripple:	0.14 (dB)

SLED1550D20A

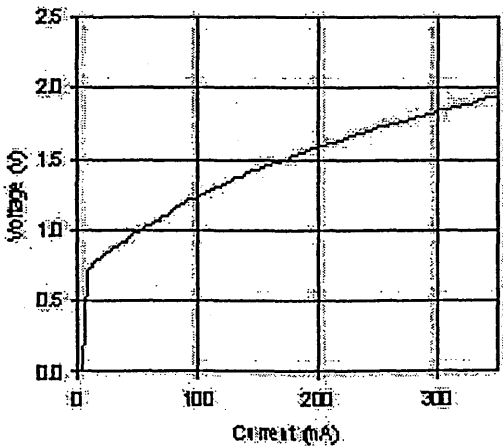
Module output power



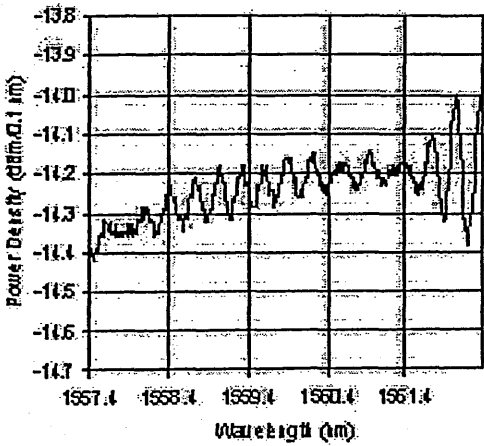
Spectrum at 350 mA



Voltage



Ripple at 350 mA



Temperature: 20.00 (°C)
Max Power: 17.4 (mW)
Peak Wavel.: 1562.31 (nm)
Central Wavel.: 1559.92 (nm)
Spectral Width: 33.23 (nm)
Max Ripple: 0.34 (dB)

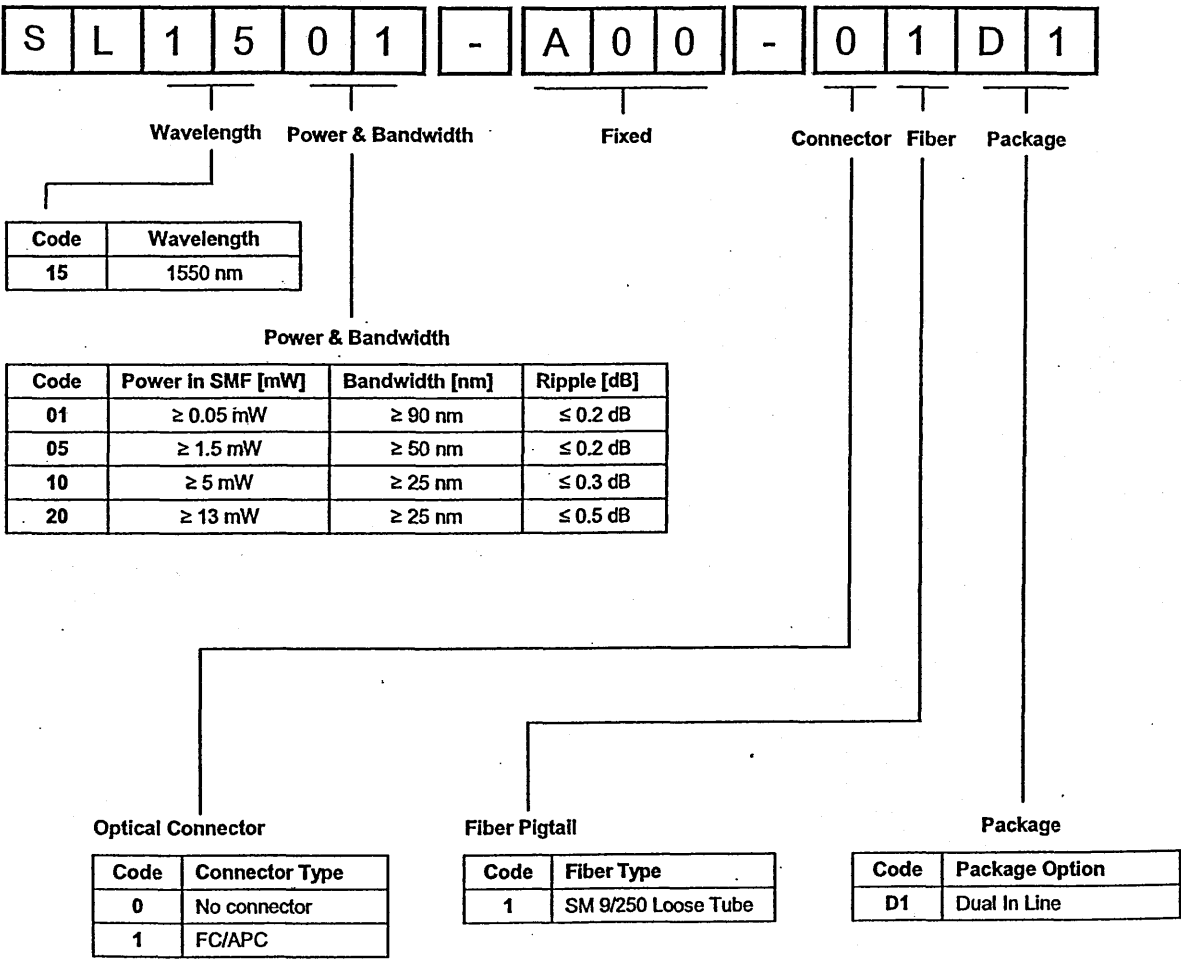
F. APPLICATION HINTS

The Opto Speed superluminescent light emitting diode (SLED) family is based on ultra high power, broadband edge emitting SLED's. For proper operation the rear tab of the DIL package should be mounted on a heatsink with a thermal resistance of 5 K/W or less.
Back reflections may influence the output power and spectral characteristics of the SLED. The use of optical isolators and/or angled connectors is recommended. Back reflection of less than -30dB are suggested.

G. ORDERING INFORMATION

Please use the following code system to order products from Opto Speed:

SLED Summary Di1 1550D Product coding:



-

0

1

D

1

Connector

Fiber

Package

Code	Power in SMF [mW]	Bandwidth [nm]	Ripple [dB]
01	≥ 0.05 mW	≥ 90 nm	≤ 0.2 dB
05	≥ 1.5 mW	≥ 50 nm	≤ 0.2 dB
10	≥ 5 mW	≥ 25 nm	≤ 0.3 dB
20	≥ 13 mW	≥ 25 nm	≤ 0.5 dB

Power & Bandwidth

Code	Connector Type
0	No connector
1	FC/APC

Optical Connector

Code	Fiber Type
1	SM 9/250 Loose Tube

Fiber Pigtail

Code	Package Option
D1	Dual In Line

Package

Appendix H

Appendix H contains data for the 4-port Optical Circulator.

3-PORT AND 4-PORT POLARIZATION INSENSITIVE OPTICAL CIRCULATOR

Product Description

Features

- ◆ Compact Package
- ◆ Low Insertion Loss
- ◆ Low PDL
- ◆ High Channel Isolation
- ◆ Highly Stable & Reliable
- ◆ Low dispersion
- ◆ Epoxy-Free Optical Path

Applications

- ◆ Optical Amplification
- ◆ Metropolitan Area Network
- ◆ Dynamic Gain Equalization
- ◆ Dispersion Compensation
- ◆ Laboratory R&D

Oplink Fiber Optic Product Lines

- ◆ Amplifier Components
- ◆ Amplifier Modules
- ◆ DWDMs
- ◆ Switching/Routing/
Signal Conditioning

3-PORT and 4-PORT POLARIZATION INSENSITIVE OPTICAL CIRCULATOR

Oplink's polarization insensitive miniature circulator features excellent optical characteristic over a wide wavelength range while delivering a compact package size. It possesses low insertion loss, high isolation, and low PDL making it an ideal candidate for EDFA, add/drop, dynamic gain equalization, dispersion compensation and other optical network applications.

Oplink can provide customized designs to meet specialized feature applications. Also, Oplink offers modular assemblies that integrate other components to form a full function module or subsystem.



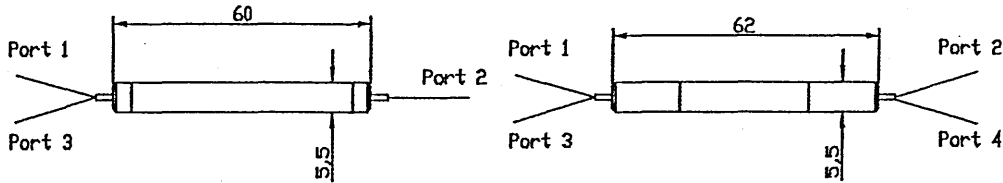
Performance Specifications

Parameter		Specifications			Units	
Operating Wavelength Range (λ_{op})	I310	1290 ~1330			nm	
	C Band	1525 ~1565			nm	
	L Band	1570 ~ 1610			nm	
	C + L	1525 ~1610			nm	
Type		3-Port		4-Port		
		I310	C,L	C+L	I310, C, L	
Insertion Loss' (1→2, 2→3 or 3→4 @ λ_{op} , Top, SOP)	Typical	0.6		0.7	0.6	dB
	Maximum	0.7		0.9	0.8	dB
Channel Isolation (2→1, 3→2 or 4→3)	Peak	≥50				dB
	@ λ_{op} , Top, SOP	≥40		≥32	≥40	dB
Directivity (1→3 or 2→4)		≥50				dB
Polarization Dependent Loss (PDL)		≤0.10			≤0.15	dB
Polarization Mode Dispersion (PMD)		≤0.06				ps
Return Loss'		≥50				dB
Optical Power		<500				mW
Operating Temperature Range (Top)		0 to 70				°C
Storage Temperature		-40 to 85				°C
Fiber Type		Corning SMF-28				
Physical Dimension		φ5.5 × 60.0 (L)		φ5.5 × 62.0 (L) mm		

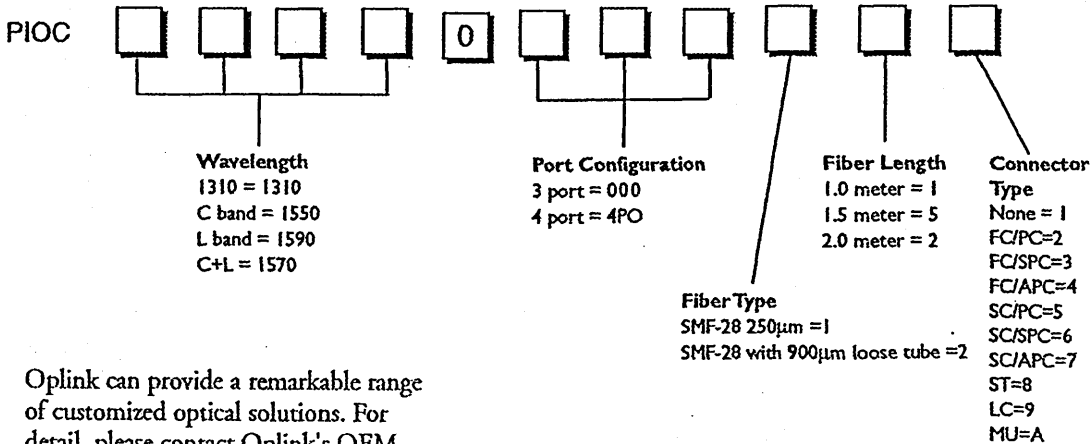
Notes: 1. Excluding connectors.

3-PORT AND 4-PORT POLARIZATION INSENSITIVE OPTICAL CIRCULATOR

Mechanical Footprint **Dimension (unit: mm)**



Ordering Information



Oplink can provide a remarkable range of customized optical solutions. For detail, please contact Oplink's OEM design team or account manager for your requirements and ordering information (408) 433-0606.



Oplink Communications, Inc.
 3469 North First Street San Jose, CA 95134-1803
 Tel: (408) 433-0606 Fax: (408) 433-0608
 Email: Sales@Oplink.com

Appendix I

Appendix I contains data sheets for the New Focus 125-kHz Auto-Balanced Receiver

2017



Products Company Support Co

PRODUCT SEARCH

Have an account? [click here](#)

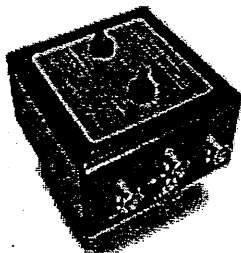
Home: Products > Detectors/Receivers/Transmitters > 125-kHz Auto-Balanced Receivers > Compare Specific

For best results when printing change page layout to landscape mode

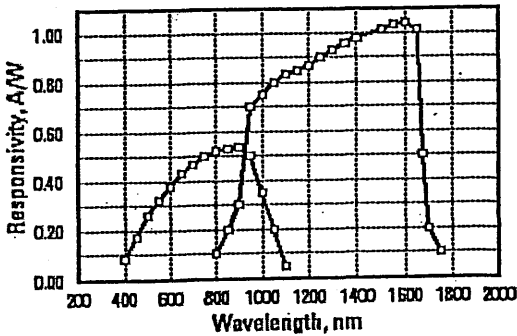
125-kHz Auto-Balanced Receivers

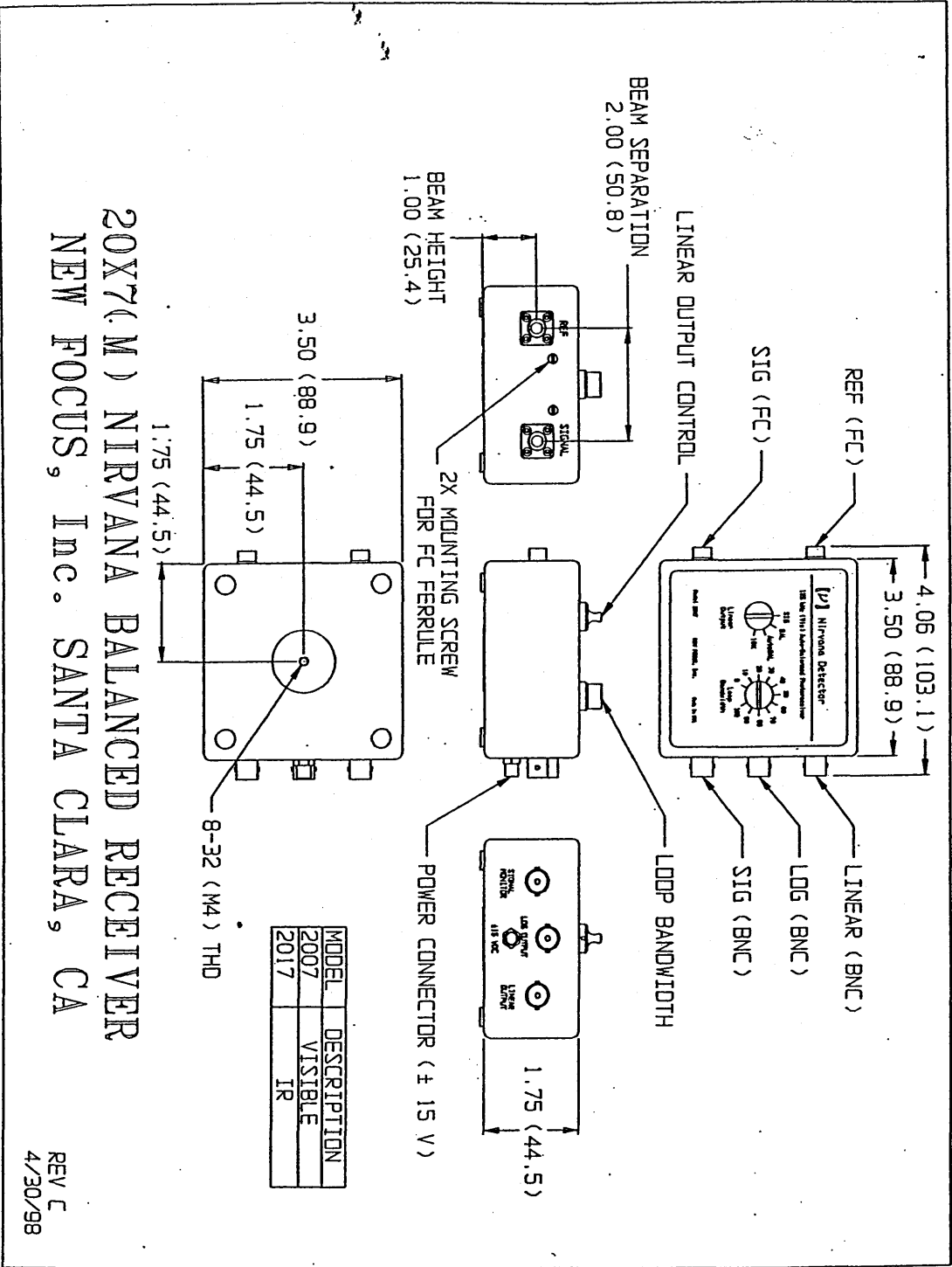
	2007	2007-M	2017
Wavelength Range	400-1070 nm	400-1070 nm	800-1700 nm
Common Mode Rejection Ratio (CMRR)	50 dB	50 dB	50 dB
3-dB Bandwidth	125 kHz	125 kHz	125 kHz
Rise Time	3 μ s	3 μ s	3 μ s
Maximum Conversion Gain	5.2×10^5 V/W	5.2×10^5 V/W	1×10^6 V/W
Typical Maximum Responsivity	0.5 A/W	0.5 A/W	1.0 A/W
Maximum Transimpedance Gain	1×10^6 V/A	1×10^6 V/A	1×10^6 V/A
Output Impedance	100 Ω	100 Ω	100 Ω
Minimum NEP	<3 pW/ $\sqrt{\text{Hz}}$	<3 pW/ $\sqrt{\text{Hz}}$	<3 pW/ $\sqrt{\text{Hz}}$
cw Saturation Power	1 mW	1 mW	0.5 mW
Detector Material/Type	Si/PIN	Si/PIN	InGaAs/PIN
Detector Diameter	2.5 mm	2.5 mm	1.0 mm
Power Requirements	± 15 V DC, <300mA	± 15 V DC, <300mA	± 15 V DC, <300
Optical Input	FC & Free Space	FC & Free Space	FC & Free Space
Electrical Output	BNC	BNC	BNC

Model 2007



Responsivity





20X7(M) NIRVANA BALANCED RECEIVER
NEW FOCUS, INC. SANTA CLARA, CA

REV C
4/30/98

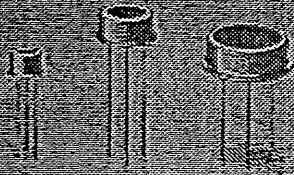
Appendix J

Appendix J contains data sheets for the Hamamatsu InGaAs PIN Photodiode

PHOTODIODE

InGaAs PIN photodiode
G8370 series

Large active areas from $\phi 1$ to $\phi 5$ mm



InGaAs PIN photodiodes have large shunt resistance and feature very low noise. Hamamatsu provides various types of InGaAs PIN photodiodes with active areas from $\phi 1$ to $\phi 5$ mm.

Features

- Low noise, low dark current
- Large active area
- Various active area sizes available

Applications

- Laser monitor
- Optical power meter
- Laser diode life test

■ Specifications / Absolute maximum ratings

Type No.	Dimensional outline/ Window material *1	Package	Active area (mm)	Absolute maximum ratings		
				Reverse voltage VR Max. (V)	Operating temperature Topr (°C)	Storage temperature Tstg (°C)
G8370-01	①/K	TO-18	φ1	10	-40 to +85	-55 to +125
G8370-02	②/K	TO-5	φ2	5		
G8370-03			φ3			
G8370-05			③/K			

■ Electrical and optical characteristics (Typ. Ta=25 °C, unless otherwise noted)

Type No.	Spectral response range (μm)	Peak sensitivity wavelength λ_p (μm)	Photo sensitivity S		Dark current Id VR=1 V		Cut-off frequency fc VR=1 V RL=50 Ω -3 dB (MHz)	Terminal capacitance Ct VR=1 V f=1 MHz (pF)	Shunt resistance Rsh VR=10 mV (M Ω)	D* $\lambda=\lambda_p$ (cm·Hz ^{1/2} /W)	NEP $\lambda=\lambda_p$ (W/Hz ^{1/2})
			1.3 μm (A/W)	$\lambda=\lambda_p$ (A/W)	Typ. (nA)	Max. (nA)					
					1 *2	5 *2					
G8370-01	0.9 to 1.7	1.55	0.9	0.95	1	5	35	90	100	5 × 10 ¹²	2 × 10 ⁻¹⁴
G8370-02					5	25	4	550	25		4 × 10 ⁻¹⁴
G8370-03					15	75	2	1000	10		6 × 10 ⁻¹⁴
G8370-05					25	125	0.6	3500	3		1 × 10 ⁻¹³

*1: Window material K: borosilicate glass with anti-reflective coating (optimized for 1.55 μm peak)

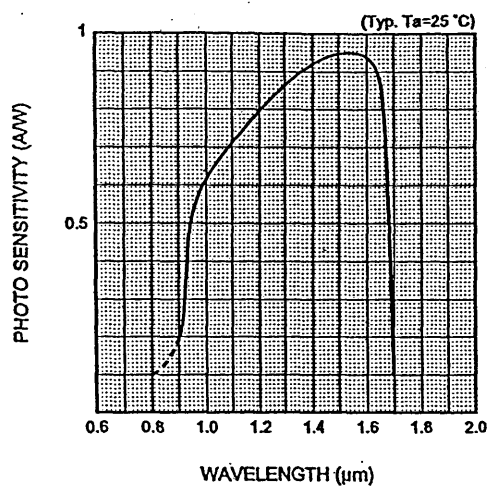
*2: VR=5 V

SOLID
STATE DIVISION

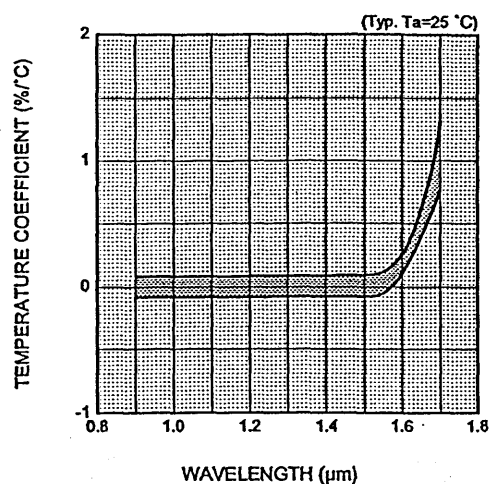
HAMAMATSU

InGaAs PIN photodiode G8370 series

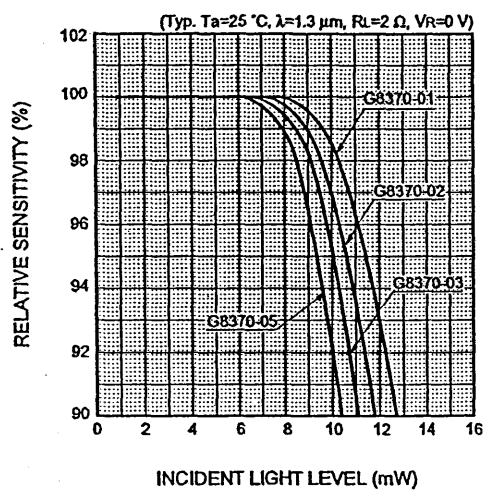
■ Spectral response



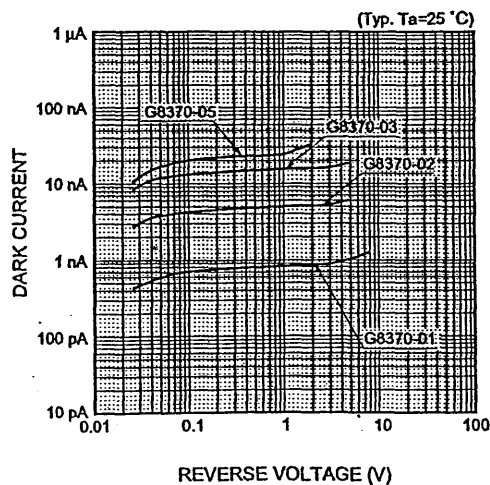
■ Photo sensitivity temperature characteristic



■ Photo sensitivity linearity

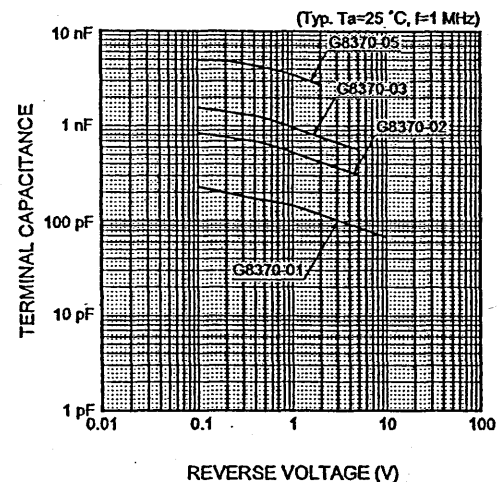


■ Dark current vs. reverse voltage

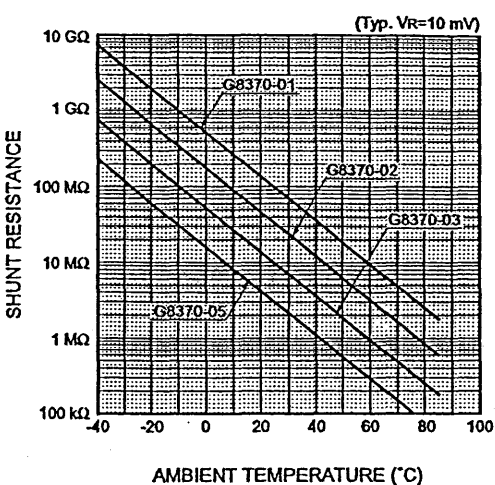


InGaAs PIN photodiode **G8370 series**

■ Terminal capacitance vs. reverse voltage

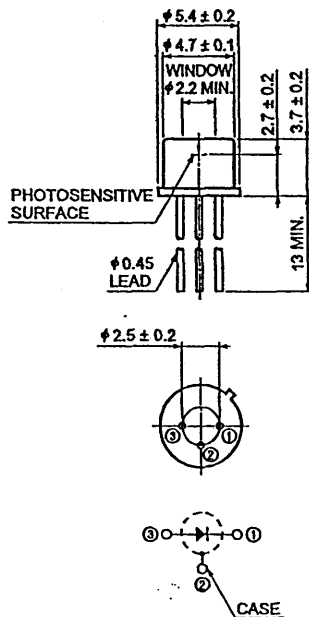


■ Shunt resistance vs. ambient temperature



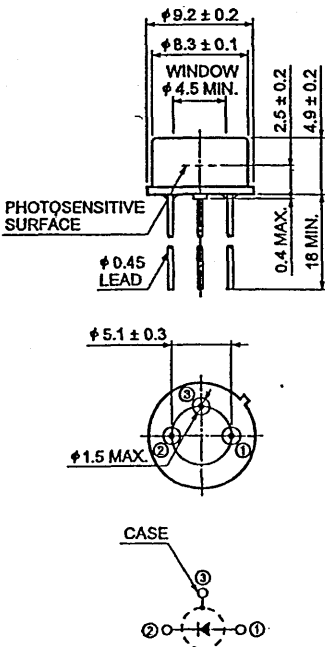
■ Dimensional outlines (unit: mm)

① G8370-01



K9D8C2U7EA

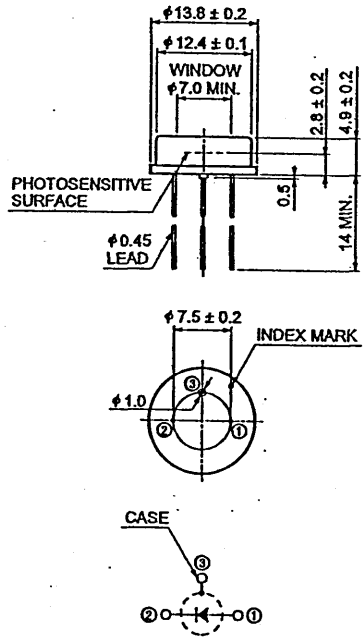
② G8370-02/-03



K9D8C2U7EA

InGaAs PIN photodiode G8370 series

③ G8370-05



REF:G0000000

HAMAMATSU

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HAMAMATSU PHOTONICS K.K., Solid State Division

1126-1 Ichino-cho, Hamamatsu City, 435-8558 Japan, Telephone: (81) 053-434-3311, Fax: (81) 053-434-5184, <http://www.hamamatsu.com>

U.S.A.: Hamamatsu Corporation, 360 Foothill Road, P.O. Box 6910, Bridgewater, N.J. 08607-0910, U.S.A., Telephone: (1) 908-231-0900, Fax: (1) 908-231-1218

Germany: Hamamatsu Photonics Deutschland GmbH, Arzbergstr. 10, D-82211 Herrsching am Ammersee, Germany, Telephone: (49) 06152-3750, Fax: (49) 06152-2658

France: Hamamatsu Photonics France S.A.R.L., 8, Rue du Saule Trappu, Parc du Moulin de Massy, 91882 Massy Cedex, France, Telephone: 33 (1) 69 53 71 00, Fax: 33 (1) 69 53 71 10

United Kingdom: Hamamatsu Photonics UK Limited, 2 Howard Court, 10 Twinn Road, Welwyn Garden City, Hertfordshire AL7 1BW, United Kingdom, Telephone: (44) 1707-294888, Fax: (44) 1707-325777

North Europe: Hamamatsu Photonics Norden AB, Smidesvågen 12, SE-171 41 Solna, Sweden, Telephone: (46) 8-509-031-00, Fax: (46) 8-509-031-01

Italy: Hamamatsu Photonics Italia S.R.L., Strada della Moia, 1/E, 20020 Arese, (Milano), Italy, Telephone: (39) 02-935-81-733, Fax: (39) 02-935-81-741

Cat. No. KIRD105

Feb. 2002 DN

Appendix K

Appendix K contains Publications and Presentations

Estimation and Implecations of Random Errors in Whole-body Dosimetry for Targeted Radionuclide Therapy Glenn D Flux, Matthew J Guy, Ruth Beddows, Matthew Pryor and Maggie A Flower. Physics in Medicine and Biology, 47 2002 pp3211-3223.

Improved Performance Interferometer Designs for OpticalCoherence Tomography

R. Beddows, S. W. James and R. P. Tatam

Presented at the 15th International Conference on Optical Fibre Sensors, (OFS-15), Portland, Oregon, USA, May 2002. IEEE Cat. No: 02EX533, pp 527-530.

Optical Coherence Tomography for the Detection of Pre-cancerous Tissue.

Ruth Beddows, Seminar presented at Kent Oncology Centre, January 2003.

Improvements in CT Virtual Simulation Quality Assurance. Presentation at the British

Institute of Radiology, Improvements in CT Virtual Simulation, Dec.2003.

Assessing Lung Depth Versus Lung Volume – A clinical study. Presentation at the

Exomio User Group Meeting, Dec.2003.