INVESTIGATION OF NEW ELECTRO - OPTICAL TECHNIQUES
FOR MONITORING PATIENTS WITH COMPROMISED
PERIPHERAL PERFUSION IN ANAESTHESIA

A thesis submitted for the degree of
Doctor of Philosophy in the University of London

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ABSTRACT

Pulse oximeter probes placed peripherally may fail to give accurate values of arterial blood oxygen saturation (SpO₂) when peripheral perfusion is poor. Since central blood flow may be preferentially preserved, the oesophagus was investigated as an alternative monitoring site. Miniature opto-electronic reflectance photoplethysmographic (PPG) probes were developed to allow PPG measurements within the whole depth of the oesophagus. Instrumentation was constructed to detect, process and display the PPG outputs on a laptop computer. Preliminary studies using a prototype system showed that measurable PPG signals with large amplitudes and good signal-to-noise ratio at an infrared wavelength (880 nm) could be detected in the deep oesophagus of a normal healthy volunteer. Further investigations on 20 anaesthetised patients showed that measurable PPG signals could also be obtained at red wavelengths (655 nm) in the mid-third of the oesophagus. A multiplexed PPG system, which operates at two wavelengths was developed to investigate in detail the morphology and quality of PPG signals from the entire length of the oesophagus in 16 anaesthetised patients. Oesophageal PPGs have been obtained with high signal-to-noise ratio at both wavelengths from the entire length of the oesophagus. The maximum PPG amplitudes, and therefore the optimum monitoring depth, appeared to be in the mid-oesophagus. A new three channel (oesophageal, finger PPG and ECG) isolated processing system was developed using printed circuit board technology. Software was implemented in LabVIEW, which allowed the continuous estimation of oesophageal and finger SpO₂. The new system was evaluated and calibrated on clinical measurements on 50 thoracic patients. The oesophageal SpO₂ readings agreed within ±3% with CO-Oximetry results. In five patients peripheral oximetry failed for at least ten minutes while the oesophageal readings remained reliable. The results from the clinical measurements have confirmed that oesophageal pulse oximetry may be used as an alternative for patients with compromised peripheral perfusion.
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# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>ABSTRACT</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACKNOWLEDGEMENTS</td>
<td>3</td>
</tr>
<tr>
<td>LIST OF FIGURES</td>
<td>13</td>
</tr>
<tr>
<td>LIST OF TABLES</td>
<td>22</td>
</tr>
<tr>
<td>CHAPTER 1: INTRODUCTION</td>
<td>23</td>
</tr>
<tr>
<td>CHAPTER 2: BASIC PRINCIPLES AND PHYSIOLOGY OF THE RESPIRATORY SYSTEM AND THE HUMAN OESOPHAGUS</td>
<td>27</td>
</tr>
<tr>
<td>2.1 RESPIRATORY SYSTEM</td>
<td>27</td>
</tr>
<tr>
<td>2.1.1 Elements of Respiratory Physiology</td>
<td>28</td>
</tr>
<tr>
<td>2.1.2 Gas Exchange</td>
<td>30</td>
</tr>
<tr>
<td>2.2 REGULATION OF RESPIRATION</td>
<td>31</td>
</tr>
<tr>
<td>2.3 BLOOD CIRCULATION</td>
<td>32</td>
</tr>
<tr>
<td>2.3.1 Heart</td>
<td>32</td>
</tr>
<tr>
<td>2.3.2 Systemic Circulation</td>
<td>32</td>
</tr>
<tr>
<td>2.3.3 Pulmonary Circulation</td>
<td>32</td>
</tr>
<tr>
<td>2.4 THE HUMAN OESOPHAGUS: BASIC PHYSIOLOGY</td>
<td>33</td>
</tr>
<tr>
<td>2.4.1 Structure of the Oesophagus</td>
<td>34</td>
</tr>
<tr>
<td>2.4.2 Blood Vessels of the Oesophagus</td>
<td>35</td>
</tr>
<tr>
<td>2.4.3 Gastro-Oesophageal Junction</td>
<td>35</td>
</tr>
<tr>
<td>CHAPTER 3: PHOTOPLETHYSMOGRAPHY</td>
<td>36</td>
</tr>
<tr>
<td>3.1 PRINCIPLE OF PHOTOPLETHYSMOGRAPHY</td>
<td>36</td>
</tr>
<tr>
<td>3.2 REVIEW OF THE LITERATURE OF PHOTOPLETHYSMOGRAPHY</td>
<td>39</td>
</tr>
<tr>
<td>3.3 ORIGIN OF THE PHOTOPLETHYSMOGRAPHIC SIGNAL</td>
<td>40</td>
</tr>
<tr>
<td>3.4 PHOTOPLETHYSMOGRAPHY AND PULSE OXIMETRY</td>
<td>42</td>
</tr>
<tr>
<td>CHAPTER 4: REVIEW OF THE HISTORY OF BLOOD OXYGEN SATURATION AND PULSE OXIMETRY</td>
<td>43</td>
</tr>
</tbody>
</table>
6.8 DISPOSABLE PROBES 82

CHAPTER 7: APPLICATIONS OF PULSE OXIMETRY 83

7.1 ANAESTHESIA 83

7.2 EMERGENCY MEDICINE 85
7.2.1 Pre-hospital care: Monitoring on the road and in the air 85
7.2.2 Pulse Oximetry in the Accident and Emergency Department 86

7.3 POSTOPERATIVE RECOVERY 86

7.4 CHILDBIRTH 87

7.5 NEONATAL AND PAEDIATRIC CARE 88

7.6 DENTISTRY AND ORAL SURGERY 89

7.7 SLEEP STUDIES AND EXERCISE 89

7.8 FUTURE APPLICATIONS OF PULSE OXIMETRY 89

CHAPTER 8: LIMITATIONS OF PULSE OXIMETRY 91

8.1 LIMITATIONS OF THE PULSE OXIMETRY DUE TO CALIBRATION ASSUMPTIONS 91

8.2 LIMITATIONS OF PULSE OXIMETERS DUE TO INTERFERENCE 92
8.2.1 Dyshaemoglobinaemias 92
8.2.2 Billirubin and Intravenous Dyes 95

8.3 LIMITATIONS OF PULSE OXIMETERS DUE TO SIGNAL ARTEFACT 96
8.3.1 False Signal 96

8.4 OTHER PHYSIOLOGICAL AND TECHNICAL LIMITATIONS OF PULSE OXIMETERS 99
8.4.1 Anaemia 99
8.4.2 Skin pigmentation and nail polish 99
8.4.3 Light emitting diode centre wavelength variability 100
8.4.4 Effects of temperature 100
8.4.5 Electromagnetic interference 101
8.4.6 Interference due to Electrocautery 101
8.4.7 Interference due to MRI 102
8.5 LIMITATIONS OF PULSE OXIMETERS DUE TO INADEQUATE PULSATILE PERFUSION

CHAPTER 9: DESIGN AND EVALUATION OF REFLECTANCE OESOPHAGEAL AND FINGER PHOTOPLETHYSMOGRAPHIC (PPG) PROBES

9.1 INTRODUCTION

9.2 OESOPHAGEAL PPG PROBE I
  9.2.1 Optical Components
  9.2.2 Mechanical Construction of Oesophageal PPG Probe I
  9.2.3 Performance Evaluation of Oesophageal PPG Probe I
  9.2.4 Limitations of Oesophageal PPG Probe I

9.3 OESOPHAGEAL PPG PROBE II
  9.3.1 Optical Components
  9.3.2 Mechanical Construction of Oesophageal PPG Probe II
  9.3.3 Performance Evaluation of Oesophageal PPG Probe II
  9.3.4 Temperature tests on Oesophageal PPG Probe II

9.4 FINGER REFLECTANCE PPG PROBE
  9.4.1 Optical Components
  9.4.2 Mechanical Construction of Finger PPG Probe
  9.4.3 Performance Evaluation of the Finger PPG Probe

CHAPTER 10: PPG PROCESSING SYSTEM 1

10.1 INTRODUCTION

10.2 INSTRUMENTATION
  10.2.1 IRED driver
  10.2.2 Photodetector I-V Amplifier
  10.2.3 PPG filter & Pre-amplifier
  10.2.4 Variable amplification and DC level shifting stage
  10.2.5 Analogue-to-Digital Converter (ADC)
  10.2.6 Mechanical Construction

10.3 CLINICAL MEASUREMENTS WITH PPG PROCESSING SYSTEM 1
  10.3.1 Healthy Volunteer
  10.3.2 Anaesthetised Patients
10.4 RESULTS
10.4.1 Healthy Volunteer
10.4.2 Anaesthetised Patients

10.5 DISCUSSION AND CONCLUSIONS

CHAPTER 11: PPG PROCESSING SYSTEM 2

11.1 INTRODUCTION

11.2 INSTRUMENTATION
11.2.1 Input Side Circuitry
11.2.2 Isolation and Output Side Circuitry

11.3 MECHANICAL CONSTRUCTION
11.3.1 Electronic Circuits
11.3.2 Power Supply
11.3.3 Enclosure

11.4 PERFORMANCE EVALUATION
11.4.1 PPG Processing System 2 Electrical Safety Test
11.4.2 Results of PPG Processing System 2 Electrical Safety Test

11.5 CLINICAL MEASUREMENTS WITH THE PPG PROCESSING SYSTEM 2

11.6 DATA ANALYSIS AND STATISTICS

11.7 RESULTS

11.8 DISCUSSION AND CONCLUSIONS

CHAPTER 12: PPG PROCESSING SYSTEM 3

12.1 INTRODUCTION

12.2 INPUT SIDE CIRCUITRY OF THE OESOPHAGEAL PPG CHANNEL
12.2.1 Master Clock and Timing Generator Circuit
12.2.2 Emitter Driver
12.2.3 Preamplifier
12.2.4 Demultiplexer
12.2.5 Filtering and Attenuation

12.3 ISOLATION AND OUTPUT SIDE CIRCUITRY OF THE OESOPHAGEAL PPG PROCESSING SYSTEM 3
12.4 INPUT SIDE CIRCUITRY OF THE ECG CHANNEL
12.4.1 Instrumentation Amplifier
12.4.2 Band-pass Filter

12.5 ISOLATION AND OUTPUT SIDE CIRCUITRY OF THE ECG CHANNEL

12.6 SIGNAL DIGITISATION
12.6.1 Introduction
12.6.2 Laptop Personal Computer
12.6.3 Installation and Configuration of the DAQCard-1200

12.7 SIGNAL ACQUISITION USING LABVIEW
12.7.1 Introduction to LabVIEW

12.8 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT 1 (VI-1)
12.8.1 Analogue Input Acquisition
12.8.2 Data Storage

12.9 DEVELOPMENT OF A SIGNAL DISPLAY VIRTUAL INSTRUMENT 2 (VI-2)

12.10 MECHANICAL CONSTRUCTION
12.10.1 Electronic Circuits
12.10.2 Power Supply
12.10.3 Enclosure

12.11 EVALUATION OF THE HARDWARE AND SOFTWARE DEVELOPMENTS OF THE PPG PROCESSING SYSTEM 3
12.11.1 Evaluation of the Hardware and Software Stages
12.11.2 PPG Processing System 3 Electrical Safety Test

12.12 PATIENTS AND MEASUREMENTS
12.12.1 Data Analysis and Statistics

12.13 RESULTS

12.14 DISCUSSION AND CONCLUSIONS

CHAPTER 13: PPG PROCESSING SYSTEM 4
13.1 INTRODUCTION
13.2 PPG PROCESSING SYSTEM 4
13.2.1 Input Side Circuitry of the Oesophageal PPG Processing Channel 226
13.2.2 Isolation and Output Side Circuitry of the Oesophageal PPG Processing Channel 227
13.2.3 Input Side Circuitry of the ECG Channel 232
13.2.4 Isolation and Output Side Circuitry of the ECG Channel 233

13.3 POWER MANAGEMENT OF THE PROCESSING SYSTEM 4 234

13.4 MECHANICAL CONSTRUCTION OF THE PROCESSING SYSTEM 4 236
13.4.1 Electronic Circuits 236
13.4.2 Processing System Enclosure 237
13.4.3 Battery Enclosure 238

13.5 DEVELOPMENT OF A PRINTED CIRCUIT BOARD (PCB) VERSION OF PROCESSING SYSTEM 4 (SYSTEM 4a) 239

13.6 SIGNAL DIGITISATION 242
13.6.1 Introduction 242
13.6.2 Installation and Configuration of the DAQCard-AI-16XE-50 242

13.7 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT 3 (VI-3) 243
13.7.1 Analogue Input Acquisition 244
13.7.2 Data Storage 245
13.7.3 Signal Gain and Offset Software Control 247
13.7.4 Estimation of Oesophageal and Finger SpO₂ 249
13.7.5 Development of a Battery Condition Algorithm 256

13.8 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT VERSION 3a (VI-3a) 258
13.8.1 Introduction 258
13.8.2 Heart Rate Algorithm 259

13.9 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT VERSION 3b (VI-3b) 264
13.9.1 Introduction 264
13.9.2 Theory of Digital Filters 264
13.9.3 Digital Filter Algorithm 265
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>13.10</td>
<td>EVALUATION OF THE HARDWARE AND SOFTWARE DEVELOPMENTS</td>
<td>270</td>
</tr>
<tr>
<td>13.10.1</td>
<td>Evaluation of the Hardware and Software Stages</td>
<td>270</td>
</tr>
<tr>
<td>13.10.2</td>
<td>Electrical Safety Test for PPG Processing Systems 4 and 4a</td>
<td>272</td>
</tr>
<tr>
<td>CHAPTER 14: INVESTIGATION OF OESOPHAGEAL PULSE OXIMETRY USING PROCESSING SYSTEM 4a AND VIRTUAL INSTRUMENT 3b IN ADULT THORACIC SURGERY PATIENTS</td>
<td>274</td>
<td></td>
</tr>
<tr>
<td>14.1</td>
<td>INTRODUCTION</td>
<td>274</td>
</tr>
<tr>
<td>14.2</td>
<td>PATIENTS AND MEASUREMENTS</td>
<td>275</td>
</tr>
<tr>
<td>14.2.1</td>
<td>Data Analysis and Statistics</td>
<td>280</td>
</tr>
<tr>
<td>14.3</td>
<td>RESULTS FROM THE INVESTIGATION OF PPG SIGNALS IN CARDIAC PATIENTS</td>
<td>282</td>
</tr>
<tr>
<td>14.4</td>
<td>RESULTS FROM AN INVESTIGATION TO DETERMINE THE CONTRIBUTION OF AORTIC OR CARDIAC MOVEMENT ARTEFACT TO THE OESOPHAGEAL PPG SIGNALS</td>
<td>287</td>
</tr>
<tr>
<td>14.4.1</td>
<td>Investigation of Oesophageal PPG signals during aortic cross-clamp</td>
<td>287</td>
</tr>
<tr>
<td>14.4.2</td>
<td>Investigation of Oesophageal PPG signals during pulsatile flow bypass</td>
<td>289</td>
</tr>
<tr>
<td>14.4.3</td>
<td>Evidence from the quantitative estimations of Oesophageal SpO₂</td>
<td>290</td>
</tr>
<tr>
<td>14.5</td>
<td>RESULTS FROM THE BLOOD OXYGEN SATURATION MEASUREMENTS</td>
<td>291</td>
</tr>
<tr>
<td>14.5.1</td>
<td>Comparisons of blood oxygen saturation measurements between CO-Oximetry and oesophageal, finger, and commercial finger pulse oximeters</td>
<td>291</td>
</tr>
<tr>
<td>14.5.2</td>
<td>Comparisons of blood oxygen saturation measurements from oesophageal, finger, and commercial finger pulse oximeters and values from blood gas analysis</td>
<td>297</td>
</tr>
<tr>
<td>14.5.3</td>
<td>Comparisons of blood oxygen saturation measurements between oesophageal, finger, and commercial finger pulse oximeters</td>
<td>303</td>
</tr>
<tr>
<td>14.6</td>
<td>PATIENTS IN WHOM PERIPHERAL PULSE OXIMETRY FAILED: RESULTS OF THE OESOPHAGEAL BLOOD OXYGEN SATURATION MEASUREMENTS</td>
<td>307</td>
</tr>
</tbody>
</table>
# LIST OF FIGURES

<table>
<thead>
<tr>
<th>FIGURE</th>
<th>DESCRIPTION</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>Organs of the respiratory system</td>
<td>28</td>
</tr>
<tr>
<td>2.2</td>
<td>Lower respiratory tract</td>
<td>29</td>
</tr>
<tr>
<td>2.3</td>
<td>Diagram showing the course of the human oesophagus in relation to the vertibral column, the trachea, and the aorta</td>
<td>33</td>
</tr>
<tr>
<td>2.4</td>
<td>Transverse section through the wall of the oesophagus</td>
<td>34</td>
</tr>
<tr>
<td>3.1</td>
<td>A PPG probe used in the transmission mode</td>
<td>37</td>
</tr>
<tr>
<td>3.2</td>
<td>A PPG probe used in reflection mode</td>
<td>37</td>
</tr>
<tr>
<td>3.3</td>
<td>Photoplethysmographic waveform as measured by transmission through tissue</td>
<td>38</td>
</tr>
<tr>
<td>3.4</td>
<td>Erythrocytes changing their axis during cardiac cycle</td>
<td>41</td>
</tr>
<tr>
<td>4.1</td>
<td>Ludwig Nicolai’s first apparatus for studying oxygen saturation in tissue</td>
<td>44</td>
</tr>
<tr>
<td>4.2</td>
<td>Kramer's apparatus for optically recording oxygen saturation in unopened arteries of experimental animals</td>
<td>45</td>
</tr>
<tr>
<td>4.3</td>
<td>J.R. Squire's instrument for measuring blood oxygenation in the web of the human hand</td>
<td>46</td>
</tr>
<tr>
<td>4.4</td>
<td>The Millikan oximeter</td>
<td>46</td>
</tr>
<tr>
<td>4.5</td>
<td>Simplified diagram of the Hewlett-Packard ear oximeter</td>
<td>47</td>
</tr>
<tr>
<td>4.6</td>
<td>Simplified diagram of the Minolta pulse oximeter</td>
<td>49</td>
</tr>
<tr>
<td>5.1</td>
<td>Extinction coefficients of the four most common haemoglobin species</td>
<td>56</td>
</tr>
<tr>
<td>5.2</td>
<td>Absorption coefficient versus wavelength of oxyhaemoglobin and deoxyhaemoglobin</td>
<td>57</td>
</tr>
<tr>
<td>5.3</td>
<td>Schematic diagrams for use in considering the Beer-Lambert Law with and without scattering</td>
<td>58</td>
</tr>
<tr>
<td>5.4</td>
<td>Absorbed and transmitted light in living tissue</td>
<td>59</td>
</tr>
<tr>
<td>5.5</td>
<td>Displayed SpO₂ values for a pulse oximeter if calibrated using the simple Beer-Lambert Law</td>
<td>61</td>
</tr>
<tr>
<td>5.6</td>
<td>Empirical relationship between arterial saturation and “red:infrared” ratio</td>
<td>64</td>
</tr>
<tr>
<td>5.7</td>
<td>Block diagram of a commercial pulse oximetry showing the major electronics components</td>
<td>65</td>
</tr>
<tr>
<td>Section</td>
<td>Description</td>
<td></td>
</tr>
<tr>
<td>---------</td>
<td>-------------</td>
<td></td>
</tr>
<tr>
<td>6.1</td>
<td>Absorption spectra of oxygenated and deoxygenated haemoglobin, showing the two most commonly used wavelengths for pulse oximetry</td>
<td></td>
</tr>
<tr>
<td>6.2</td>
<td>Centre wavelength variation of emitters of the same type from the same lot, 660 ± 15 nm</td>
<td></td>
</tr>
<tr>
<td>6.3</td>
<td>Typical photodiode packaging</td>
<td></td>
</tr>
<tr>
<td>6.4</td>
<td>Transmission Pulse Oximetry Probe</td>
<td></td>
</tr>
<tr>
<td>6.5</td>
<td>Reflection Pulse Oximetry Probe</td>
<td></td>
</tr>
<tr>
<td>6.6</td>
<td>MRI pulse oximetry probe</td>
<td></td>
</tr>
<tr>
<td>6.7</td>
<td>Foetal Pulse Oximetry Probe</td>
<td></td>
</tr>
<tr>
<td>6.8</td>
<td>Placement of foetal probe within the uterus</td>
<td></td>
</tr>
<tr>
<td>6.9</td>
<td>Clip type reusable pulse oximetry probe</td>
<td></td>
</tr>
<tr>
<td>6.10</td>
<td>Reusable pulse oximetry probe</td>
<td></td>
</tr>
<tr>
<td>6.11</td>
<td>Reusable reflectance pulse oximetry probe</td>
<td></td>
</tr>
<tr>
<td>6.12</td>
<td>Disposable pulse oximetry probe</td>
<td></td>
</tr>
<tr>
<td>8.1</td>
<td>Relationship of red/infrared ratio to oxygen saturation, as typical pulse oximeter calibration curve</td>
<td></td>
</tr>
<tr>
<td>8.2</td>
<td>Haemoglobin extinction curve, demonstrating relative absorptions of the four major classes of haemoglobin</td>
<td></td>
</tr>
<tr>
<td>8.3</td>
<td>Ambient light interference and optical shunt in pulse oximetry</td>
<td></td>
</tr>
<tr>
<td>8.4</td>
<td>The photoplethysmographic signal of a subject at rest and during exercise</td>
<td></td>
</tr>
<tr>
<td>8.5</td>
<td>Calibration curves derived from the Beer's Law when substituting, the extinction coefficient of Hb and HbO₂ calculated at different wavelengths (660 nm and 950 nm) and temperatures [0°C to 50°C], into the relationship between SpO₂ and R</td>
<td></td>
</tr>
<tr>
<td>9.1</td>
<td>Block diagram of the <em>Oesophageal PPG Probe I</em> connected to the PPG Processing System</td>
<td></td>
</tr>
<tr>
<td>9.2</td>
<td>Top view of the Oesophageal <em>PPG Probe I</em></td>
<td></td>
</tr>
<tr>
<td>9.3</td>
<td>Photograph of the infrared Oesophageal <em>PPG Probe I</em></td>
<td></td>
</tr>
<tr>
<td>9.4</td>
<td>Photograph of the Oesophageal tube</td>
<td></td>
</tr>
<tr>
<td>9.5</td>
<td>Positioning of perspex rod in oesophageal tube</td>
<td></td>
</tr>
<tr>
<td>9.6</td>
<td>Side and cross sectional view of the Oesophageal <em>PPG Probe I</em></td>
<td></td>
</tr>
<tr>
<td>Section</td>
<td>Description</td>
<td></td>
</tr>
<tr>
<td>---------</td>
<td>-------------</td>
<td></td>
</tr>
<tr>
<td>9.7</td>
<td>Finger Infrared AC PPG traces obtained using the oesophageal reflectance PPG Probe I</td>
<td></td>
</tr>
<tr>
<td>9.8</td>
<td>Block diagram of the Oesophageal PPG Probe II connected to the PPG processing System</td>
<td></td>
</tr>
<tr>
<td>9.9</td>
<td>Top view of the Oesophageal PPG Probe II</td>
<td></td>
</tr>
<tr>
<td>9.10</td>
<td>Photograph of Oesophageal PPG Probe II</td>
<td></td>
</tr>
<tr>
<td>9.11</td>
<td>Positioning of rubber stopper in oesophageal tube</td>
<td></td>
</tr>
<tr>
<td>9.12</td>
<td>Side and cross sectional view of the Oesophageal PPG probe II</td>
<td></td>
</tr>
<tr>
<td>9.13</td>
<td>Infrared and red AC PPGs obtained from the finger using Oesophageal PPG Probe II</td>
<td></td>
</tr>
<tr>
<td>9.14</td>
<td>Block diagram of the in vitro temperature test</td>
<td></td>
</tr>
<tr>
<td>9.15</td>
<td>Temperature processing circuit</td>
<td></td>
</tr>
<tr>
<td>9.16</td>
<td>In vitro temperature rise test with the thermocouple located at the outside surface of the plastic oesophageal tube adjacent to an infrared emitter</td>
<td></td>
</tr>
<tr>
<td>9.17</td>
<td>In vitro temperature rise test with the thermocouple located at the outside surface of the plastic oesophageal tube adjacent to a red emitter</td>
<td></td>
</tr>
<tr>
<td>9.18</td>
<td>In vitro temperature rise test with the thermocouple located at the inside of the plastic oesophageal tube adjacent to an infrared emitter</td>
<td></td>
</tr>
<tr>
<td>9.19</td>
<td>In vitro temperature rise test with the thermocouple located at the inside of the plastic oesophageal tube adjacent to a red emitter</td>
<td></td>
</tr>
<tr>
<td>9.20</td>
<td>Block diagram of the in vivo temperature test</td>
<td></td>
</tr>
<tr>
<td>9.21</td>
<td>In vivo temperature rise test (mucosa of the mouth) with the thermocouple located at the outside of the plastic oesophageal tube adjacent to an infrared emitter</td>
<td></td>
</tr>
<tr>
<td>9.22</td>
<td>In vivo temperature rise test (mucosa of the mouth) with the thermocouple located at the outside of the plastic oesophageal tube adjacent to a red emitter</td>
<td></td>
</tr>
<tr>
<td>9.23</td>
<td>Top view of the Finger PPG probe</td>
<td></td>
</tr>
<tr>
<td>9.24</td>
<td>Photograph of the Reflectance Finger Probe</td>
<td></td>
</tr>
<tr>
<td>9.25</td>
<td>Infrared and red AC PPGs obtained from the finger using the reflectance finger PPG probe</td>
<td></td>
</tr>
<tr>
<td>10.1</td>
<td>Basic block diagram of the infrared PPG Processing System 1</td>
<td></td>
</tr>
<tr>
<td>10.2</td>
<td>Detail block diagram of one of the two identical channels of the infrared PPG Processing System 1</td>
<td></td>
</tr>
</tbody>
</table>
10.3 IRED driver circuit

10.4 Transresistance amplifier

10.5 Band-pass and pre-amplification circuit

10.6 Theoretical frequency response of the band-pass filter

10.7 Experimental frequency response of the band-pass filter

10.8 PPG variable amplification and DC level shifting circuit

10.9 Photograph of the analogue-to-digital converter ADC-11

10.10 Photograph of the PPG Processing System 1

10.11 Photograph of the front panel of the PPG Processing System 1

10.12 The oesophageal PPG probe contained within the stomach tube is seen placed in the oesophagus via the mouth

10.13 Oesophageal and Finger infrared AC PPG signals from a healthy oesophagus

10.14 Oesophageal and finger infrared AC PPG signals from an anaesthetised patient

10.15 Infrared AC PPG traces from the deep oesophagus of an anaesthetised patient with the mechanical ventilator switched on

11.1 Basic block diagram of the red and infrared PPG Processing System 2

11.2 Detail block diagram of one of the two identical channels of the red and infrared PPG Processing System 2

11.3 Diagram illustrating the manual switching between the two wavelengths

11.4 Schematic Diagram of the Isolation and Output Side Circuitry of the PPG Processing System 2

11.5 Photograph of the two isolated processing channels

11.6 Photograph of the complete PPG Processing System 2

11.7 Photograph of the front panel of the PPG Processing System 2

11.8 Diagram for the Insulation Resistance Test

11.9 PPG Processing System 2 connected on the Safety Tester for the Insulation Resistance Test

11.10 Diagram for the Mains on Applied Parts Safety Test

11.11 PPG Processing System 2 connected on the Safety Tester for the Mains on Applied Parts Test
11.12 Typical AC PPG traces for the red and infrared wavelengths from the middle third of the oesophagus and the finger of an anaesthetised patient with the mechanical ventilator temporarily switched off.

11.13 AC PPG traces for the red and infrared wavelengths from the middle third of the oesophagus and the finger of an anaesthetised patient with the mechanical ventilator switched on.

11.14 Mean peak-to-peak AC PPG signals (± SD) at red and infrared wavelengths from the mid-third of the oesophagus and the finger.

11.15 Normal probability plot of the residuals.

12.1 Detail block diagram of the PPG Processing System 3.

12.2 Block diagram of the Master Clock and Timing Generator circuit.

12.3 Schematic Diagram of the Master Clock and Timing Generator Circuit. Frequency Divider.

12.4 Timing diagram of the 4-bit ring-shift counter.

12.5 Emitter Driver Circuit.

12.6 Timing Diagram of PPG Processing System 3.

12.7 Preamplifier Circuit.

12.8 Demultiplexing of the mixed red and infrared PPG signals.

12.9 Low-pass filter and attenuation circuit for the infrared channel.

12.10 Isolation and Output Side circuitry of the Oesophageal channel of the PPG Processing System 3.

12.11 Schematic diagram of the 3-lead ECG channel of the PPG Processing System 3.

12.12 Block Diagram of the PPG Signal Acquisition Virtual Instrument 1 (VI-1).

12.13 Front panel of the PPG Signal Acquisition Virtual Instrument 1.

12.14 Block diagram of the Signal Display Virtual Instrument 2 (VI-2).

12.15 Front Panel of the Signal Display Virtual Instrument 2 (VI-2).

12.16 Photograph of the two isolated processing channels (Oes. and ECG).

12.17 Photograph of the front panel of the PPG Processing System 3.

12.18 Testing parts of the electronic circuits of PPG Processing System 3.

12.19 Typical PPG and ECG traces using the PPG Processing System 3 and the VI-1.
12.20 AC and DC PPG signals corresponding to red (R) and infrared (IR) wavelengths from five oesophageal depths and ECG traces with the mechanical ventilator temporarily switched off

12.21 AC and DC PPG signals corresponding to red (R) and infrared (IR) wavelengths from five oesophageal depths and ECG traces with the mechanical ventilator switched on

12.22 Mean (± SE) AC PPG Peak-to-Peak Amplitudes at two wavelengths and five Oesophageal Depths

12.23 Oesophageal pulse transit times ± SD using the ECG “R” wave as reference

13.1 Simplified block diagram of the PPG Processing System 4

13.2 Detailed block diagram of the Processing System 4 showing one of the two identical PPG channels (oesophageal) and the ECG channel

13.3 Detail Schematic Diagram of the Input Side Circuitry of the Processing System 4

13.4 Isolation and Output Side circuitry of the Oesophageal channel of the PPG Processing System 4

13.5 Schematic diagram of the 3-lead ECG channel of the PPG Processing System 4

13.6 Schematic diagram illustrating the power management of the Processing System 4

13.7 Photograph of the isolated processing channels of Processing System 4

13.8 Photograph of the completed Processing System 4

13.9 Photograph of the new PCB developed channels

13.10 Block Diagram of the Virtual Instrument 3 (VI-3)

13.11 Front panel of the PPG Signal Acquisition Virtual instrument 3

13.12 Gain and Offset Control of the Oesophageal and Finger AC PPG Channels and ECG Channel

13.13 Illustration of the algorithm used for calculating the ratio (R) for the oesophageal channel

13.14 Illustration of Shift Registers used in LabVIEW

13.15 Part of the Virtual Instrument 3 illustrating the algorithm for the estimation of Oesophageal SpO$_2$ when the Boolean case structure is true

13.16 Part of the Virtual Instrument 3 illustrating the algorithm for the estimation of Oesophageal SpO$_2$ when the Boolean case structure is false
13.17 Battery condition algorithm

13.18 Block diagram illustrating the algorithm for calculating Heart Rate (HR)

13.19 Front panel of the PPG Signal Acquisition Virtual Instrument 3a

13.20 Part of the Virtual Instrument 3a illustrating the algorithm for the calculation of Heart Rate when the Boolean case structure is False

13.21 Front panel of the Virtual Instrument 3b (VI-3b)

13.22 Block diagram of the digital filter algorithm

13.23 Illustration of the three subcases containing the filter VIs

13.24 Front panel of VI-3 after execution using the Processing System 4

13.25 Section of the front panel (IIR Filters are not shown) of VI-3b after execution using the Processing System 4a

14.1 Photograph of an anaesthetised cardiopulmonary bypass patient in the induction room

14.2 Measuring of the oesophageal monitoring depth using a ruler

14.3 Photograph showing the positioning of the Processing System 4a and the laptop in the operating theatre

14.4 Diagram illustrating the set-up of the heart lung machine

14.5 Oesophageal, Finger and ECG traces obtained from an anaesthetised patients undergoing cardiopulmonary bypass surgery

14.6 Oesophageal and finger PPG signals during aortic cross-clamping

14.7 PPG signals during cardiopulmonary bypass surgery with the heart lung machine switched from continuous to pulsatile flow

14.8 Comparison of SpO₂ measurements obtained from the Oesophageal Probe and the CO-Oximeter

14.9 Comparison of SpO₂ measurements obtained from the Finger Probe and the CO-Oximeter

14.10 Comparison of SpO₂ measurements obtained from the commercial Finger Probe and the CO-Oximeter

14.11 Comparison of the CO-Oximeter and the red/infrared Ratios measured by the oesophageal pulse oximeter

14.12 Difference against mean for SaO₂ data obtained from the CO-Oximeter (CO-OX) and blood gas analyser (BGA)
14.13 Comparison of SpO$_2$ measurements obtained from the oesophageal Probe and the Blood Gas Analyser

14.14 Comparison of SpO$_2$ measurements obtained from the finger Probe and the Blood Gas Analyser

14.15 Comparison of SpO$_2$ measurements obtained from the commercial finger Probe and the Blood Gas Analyser

14.16 Comparison of the Blood Gas Analyser and the red/infrared Ratios measured by the oesophageal pulse oximeter

14.17 Difference against mean for SpO$_2$ data obtained from the oesophageal (OES.) and commercial finger (CF) pulse oximeters

14.18 Difference against mean for SpO$_2$ data obtained from the finger (FING) and commercial finger (CF) pulse oximeters

14.19 Difference against mean for SpO$_2$ data obtained from the finger (FING) and oesophageal (OES) pulse oximeters of the Processing System 4a

14.20 Typical PPG traces obtained from a cardiopulmonary bypass patient during times of finger failure

14.21 Comparison of SpO$_2$ measurements obtained from the Oesophageal Probe and the Blood Gas Analyser in 5 patients in whom peripheral pulse oximetry failed

15.1 Reflectance abdominal organ PPG probe placed on the surface of the human bowel

15.2 AC PPG traces from simultaneous measurements at abdominal organs (bowel, kidney, and liver) and the finger

15.3 AC Peak-to-Peak Amplitudes, Mean (± SD), at two wavelengths from the three abdominal organs and the finger
# LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.1</td>
<td>Chronological evolution of Pulse Oximetry</td>
<td>51</td>
</tr>
<tr>
<td>8.1</td>
<td>Pulse Oximetry Data Failure by Surgery Classification</td>
<td>104</td>
</tr>
<tr>
<td>9.1</td>
<td>Optical, Electrical and Package Specifications of the Infrared Emitter</td>
<td>109</td>
</tr>
<tr>
<td>9.2</td>
<td>Optical, Electrical and Package Specifications of the silicon diode photodetector</td>
<td>109</td>
</tr>
<tr>
<td>9.3</td>
<td>Specifications of Screened Multicore Cable</td>
<td>111</td>
</tr>
<tr>
<td>9.4</td>
<td>Optical, Electrical and Package Specifications of the Red Emitter</td>
<td>118</td>
</tr>
<tr>
<td>9.5</td>
<td>Specifications of the AD59CQ Thermocouple Amplifier</td>
<td>125</td>
</tr>
<tr>
<td>9.6</td>
<td>Results from the <em>in vitro</em> temperature tests for the infrared and red emitters</td>
<td>128</td>
</tr>
<tr>
<td>10.1</td>
<td>Main Electrical and Package Specifications of the TL084 Op-Amp</td>
<td>138</td>
</tr>
<tr>
<td>10.2</td>
<td>VCVS LOW-PASS FILTERS</td>
<td>141</td>
</tr>
<tr>
<td>10.3</td>
<td>Specifications of the analogue-to-digital converter (ADC-11)</td>
<td>145</td>
</tr>
<tr>
<td>11.1</td>
<td>Technical specifications of the Analogue Isolation Amplifier HCPL7820</td>
<td>156</td>
</tr>
<tr>
<td>11.2</td>
<td>Analysis of all gain stages of the <em>PPG Processing System</em> 2</td>
<td>157</td>
</tr>
<tr>
<td>11.3</td>
<td>Patient Details</td>
<td>165</td>
</tr>
<tr>
<td>11.4</td>
<td>Means, standard deviations (SD) and standard errors (SE) of the peak-to-peak amplitudes of the infrared and red AC PPGs for the oesophagus and the finger</td>
<td>168</td>
</tr>
<tr>
<td>11.5</td>
<td>Means, standard deviations and standard error of the ratio of the oesophageal to finger AC PPG amplitudes at infrared and red wavelengths</td>
<td>169</td>
</tr>
<tr>
<td>11.6</td>
<td>Results from the Normality Test (Kolmogorov-Smirnov) performed on the data of Table 11.4</td>
<td>170</td>
</tr>
<tr>
<td>11.7</td>
<td>Results of One Way Analysis of Variance Test between the infrared and red oesophageal at the mid-third of the oesophagus and finger AC PPG amplitudes</td>
<td>172</td>
</tr>
<tr>
<td>12.1</td>
<td>Truth table of the select inputs (A, B, C, D) of the MC14536B Frequency Divider</td>
<td>178</td>
</tr>
<tr>
<td>12.2</td>
<td>Specifications of the MC14503 Analogue Multiplexer</td>
<td>182</td>
</tr>
<tr>
<td>12.3</td>
<td>Specifications of the MC14502 Analogue Demultiplexer</td>
<td>185</td>
</tr>
</tbody>
</table>
12.4 Analysis of all gain stages of the PPG Processing System 3 at different switch positions
12.5 Patient Details
12.6 Mean ± SE of AC peak-to-peak PPG amplitudes (mv) at two wavelengths and five oesophageal depths
12.7 Results of comparisons between the infrared and red oesophageal AC PPG amplitudes at five oesophageal depths showing mean PPG amplitude differences
12.8 Mean ± SD of the pulse transit times (PTTs) to the oesophagus
12.9 Mean ± SD of the difference between the calculated Oesophageal and finger blood oxygen saturation
13.1 Technical specifications of the Analogue Isolation Amplifier ISO122
13.2 Technical specifications of the Yuasa NP Sealed Lead-Acid Batteries
13.3 Specifications of the 722BG Dual Isolated DC/DC Converter
14.1 Patient Details
14.2 Summary of the oesophageal monitoring depth of the fifty patients
14.3 Mean ± SE of AC peak-to-peak PPG amplitudes (mv) at two wavelengths measured in the upper, mid and lower oesophagus
14.4 Calculated values of the relationships between SpO₂ (oesophageal, finger and commercial finger pulse oximeters) and SaO₂ measured by CO-Oximetry
14.5 Calculated values of the relationships between SpO₂ and SaO₂ obtained from BGA for the oesophageal, finger and commercial finger pulse oximeter
14.6 Mean of AC peak-to-peak PPG amplitudes (mv) at two wavelengths measured in the upper and mid oesophagus of five patients in whom peripheral pulse oximetry failed
15.1 Details of the Patients used for the Visceral PPG Study
15.2 Mean (± SD) of AC peak-to-peak PPG amplitudes (v) at two wavelengths at the three abdominal organs and the finger
CHAPTER 1

Introduction

Pulse oximetry has been one of the most significant technological advances in clinical monitoring in the last decade (BOWES et al., 1989). The oximeter is widely used in anaesthesia, intensive care, operating theatres, emergency rooms and general wards.

Pulse oximetry provides continuous non-invasive monitoring of arterial blood oxygen saturation (SpO\textsubscript{2}) by shining light at two different wavelengths, one in the red and the other in the infrared spectrum through a tissue bed such as finger or earlobe. Pulse oximeters estimate arterial oxygen saturation by sensing differences in the light absorption of the oxygenated and deoxygenated haemoglobin.

Although pulse oximeters give reliable readings of blood oxygen saturation for many clinical purposes, there are significant limitations on the accuracy and the availability of pulse oximetry data in some circumstances (FREUND et al., 1991; MOLLER et al., 1993; REICH et al., 1996). Pulse oximetry is a pulse dependent technique and any significant reduction in the amplitude of the pulsatile arterial component, known as the photoplethysmographic (PPG) signal (discussed in detail in chapter 3), can lead to dubious values for blood oxygen saturation (SpO\textsubscript{2}) or complete failure. Hence, pulse oximeters require adequate peripheral perfusion to operate correctly. When peripheral perfusion is poor, as in states of hypovolaemia, hypothermia and vasoconstriction, oxygenation readings become unreliable or cease. Such clinical situations occur, for example, after prolonged operations, especially hypothermic cardiopulmonary bypass surgery, and in patients with extensive burns. The problem arises because conventional sensors must be attached to the most peripheral parts of the body where pulsatile flow is most easily
compromised. Measurements at sites other than the finger or ear, such as the forehead and nose, give no improvement in poorly perfused patients (ROSENBERG and PEDERSEN, 1990; CLAYTON et al., 1991). Thus, SpO₂ readings are often unobtainable at just the time when they would be most valuable.

The hypothesis underlying this project is that a more central site, such as the oesophagus, will remain adequately perfused in these clinical situations, giving the possibility of monitoring SpO₂ at the oesophagus when conventional peripheral oximetry fails. This thesis describes various designs of electro-optical oesophageal photoplethysmographic probes and processing systems, which were developed to investigate the suitability of the oesophagus as an alternative monitoring site for the continuous measurement of SpO₂ in cases of poor peripheral circulation. The details of both the hardware and software required to fabricate the oesophageal pulse oximeter will be the subject of the following chapters. Additionally, this thesis details the test methods at all development stages of the project as well as the resulting accuracy of the clinical methods.

A brief description of the subjects that are covered in the following chapters is presented below.

**Chapter 2** covers the basic physiology of the respiratory system with emphasis on the methods of transport of oxygen to the tissue by ventilation, binding to haemoglobin in the red blood cells, and transport through the blood circulation. Basic anatomy and physiology of the human oesophagus is also discussed since the oesophagus is the main area of investigation for this project.

**Chapter 3** introduces the principles of photoplethysmography and reviews the history of the subject in brief. The origin and description of the photoplethysmographic signal is discussed along with its relation with the estimation of blood oxygen saturation by pulse oximetry.
Chapter 4 reviews the literature of the history of the measurement of blood oxygen saturation and describes the experimental and technological evolution that led to the development of the current pulse oximeter.

Chapter 5 begins with the Lambert and Beer Law for the absorption of light by haemoglobin and oxyhaemoglobin, and develops the equations required for converting the measured light transmission through the tissue to values of the haemoglobin oxygen saturation. The limitations of the Lambert and Beer law in pulse oximetry due to multiple scattering are briefly discussed and modern methods for calibrating pulse oximeters are pointed out. A brief review on the technical developments of commercial pulse oximeters is the subject of the last section of this chapter.

Chapter 6 describes some of the details of the opto-electronic components comprising the pulse oximeter probe such as emitters (red and infrared) and photodiodes. The principles of transmission and reflection probes are explained. Also, this chapter details the design of reusable, disposable probes and other special application probes.

Chapter 7 covers the many applications of pulse oximetry in anaesthesia, intensive care, operating rooms, emergency, patient transport, birth and delivery, neonatal care, sleep laboratories and in dentistry and oral surgery.

Chapter 8 presents the physiological and technological limitations of the commercial pulse oximeter.

Chapter 9 describes in detail the development of the new oesophageal and finger reflectance PPG probes.

Chapter 10 covers the hardware development of the first two channel (oesophageal and finger) PPG Processing System 1. The first infrared AC PPG measurements from the deep oesophagus of a healthy volunteer and anaesthetised patients are presented and discussed.

Chapter 11 presents the hardware and software of a new electrically isolated two channel PPG Processing System 2. Detail investigations of AC PPG signals, at red and infrared wavelengths, in the mid-third of the oesophagus and the finger of anaesthetised patients undergoing low-risk surgery are presented.

Chapter 12 details the hardware and software (implemented in LabVIEW) design of a two channel (oesophageal PPG and ECG) time-multiplexed PPG
*Processing System 3.* Investigations of simultaneous AC and DC PPG signals obtained at both wavelengths on anaesthetised patients undergoing low-risk surgery, to determine an optimal position for reliable SpO₂ monitoring are presented. Pulse transit times (PTTs) to the oesophagus are also measured and discussed.

**Chapter 13** describes the development of a new *Processing System 4*, which comprises three channels (two PPG channels (oesophageal and finger) and an ECG channel). New software implemented in *LabVIEW* is also described including the details of the algorithms for estimating SpO₂ and heart rate. The upgrading of the *Processing System 4* from a veroboard design into a printed circuit board design (PCB) is also described.

**Chapter 14** presents in great detail the results of the clinical study on fifty patients undergoing high-risk surgery (cardiothoracic bypass). The accuracy of the oesophageal, finger and commercial pulse oximeters against the “gold standards” (CO-Oximeter and a blood gas analyser) is presented and discussed.

**Chapter 15** describes the investigation of PPG signals from various abdominal organs using the oesophageal PPG probe.

**Chapter 16** presents the conclusions along with suggestions for future work.
CHAPTER 2

Basic Principles and Physiology of the Respiratory System and the Human Oesophagus

To sustain life the human body must take in oxygen (O₂) which, is utilised on the cellular level with other essential nutrients in the metabolic oxidation process. Oxygen delivery to cells requires the use of the respiratory system as well as the circulatory system. Inspiration and expiration is the initial step, moving air in and out of the lungs where gas exchange occurs. Oxygen is diffused into the blood, while carbon dioxide (CO₂), a byproduct of cellular respiration, diffuses into the lungs. The oxygenated blood circulates around the body until it reaches oxygen depleted areas, where oxygen diffuses into cells, and CO₂ is transferred to the blood returning to the lungs. This ventilatory process is controlled by neurons in the brain stem (MARIEB, 1992).

Pulse oximetry is one of several methods used to analyse oxygen delivery. It is a non-invasive optical technique used extensively in many clinical environments. This thesis discusses a new application of pulse oximetry which estimates arterial blood oxygen saturation in the human oesophagus. Before continuing, it is essential to understand normal oxygen transport and the underlying physiology of the respiratory system and the oesophagus and this is the subject of this chapter.

2.1 RESPIRATORY SYSTEM

The exchange of gases in any biological system is termed respiration. Respiration is divided into external and internal respiration (MARIEB, 1992). External respiration is a process whereby the O₂ in the inhaled air is
delivered to the cells and the CO₂ is removed from the cells via the exhaled air. Internal respiration is a chemical process, referred to as oxidation, which takes place in the cells thereby generating energy. Carbon dioxide (CO₂) and water (H₂O) are end-products of the oxidation process which are worthless from an energy viewpoint (MARIEB, 1992). Therefore, they are released by the cells and carried to the excretory organs via the blood circulation.

Humans need two systems for external respiration:

a) the respiratory system for gas exchange
b) the blood circulation for delivery of oxygen to the tissue cells and removal of excessive carbon dioxide from the tissue cells

2.1.1 Elements of Respiratory Physiology

Air enters the upper respiratory tract (mouth, nose, throat (pharynx), and larynx) via the mouth, nose, or both (Figure 2.1). The nose has the important function of cleaning, warming and humidifying the inhaled air.

![Figure 2.1: Organs of the respiratory system (MARIEB, 1992)](image)

The inside of the nose nearest to the nostrils is provided with hairs, which clean the air of larger particles. In the nasal cavity there are a great number of superficial thin walled blood vessels that radiate heat and thereby warm
the inhaled air. The nasal cavity is kept moist by glandular secretions, which also humidify the air. The inspired air that passes through the nose is fully humidified and has a temperature of 36°C, irrespective of the ambient temperature. Air and food both go through the throat (pharynx). When swallowing, food is prevented from entering the nasal cavity by a closing upward movement of the soft palate in the roof of the mouth. After passing through the nasal cavity and the pharynx the inhaled air reaches the larynx. The larynx is partly covered by the epiglottis which, during swallowing, completely covers the upper opening of the larynx (COMROE, 1966; MARIEB, 1992).

The trachea is located directly behind the larynx and is an organ of the lower respiratory tract together with the bronchial trees, the bronchioles and the alveoli (Figure 2.2). The trachea is a tube approximately 12 cm long made of several chondroid rings. The trachea branches into two tubes (bronchi) approximately at the level of the sixth dorsal vertebra. The two bronchi provide a path to each lung. These tubes referred to as the left and right bronchus, branch into the bronchial trees. Bronchial trees divide into even narrower bronchioles, which are the ramifications of these trees. The alveoli or air cells are the “leaves” of the trees (ROMANES, 1972).

Figure 2.2: Lower respiratory tract (trachea, bronchial tree, bronchioles, and alveoli)
The lungs, as mentioned in the previous paragraph, include the bronchi, the bronchioles, the alveoli (or air cells) and the blood vessels. The primary function of the lungs is to provide adequate oxygen to blood leaving the pulmonary circulation and to remove the appropriate amount of carbon dioxide from blood entering the pulmonary circulation. The alveoli in the lungs consist of fragile and moist elastic double walls. Each individual alveolus has a network of fine blood vessels (capillaries). The diameter of the alveoli during exhalation and inhalation is 0.1 mm to 0.2 mm and 0.3 mm to 0.5 mm respectively. This means, the volume of the alveoli after inhalation is approximately twice as great as after exhalation. The number of alveoli is estimated approximately 600 million; the total surface which is available for gas exchange is approximately 70 to 100 m². The walls of the alveoli are very thin, less than 0.5 μm, thus allowing gas diffusion. Diffusion is the gas exchange between the alveoli and the blood and is described in more detail in the following section (WEBSTER, 1997).

2.1.2 Gas Exchange
The alveoli are elastic structures, however they do not have any muscles and thus cannot take in and release respiratory air on their own. They totally depend on the movements of the respiratory muscles of the chest wall and the diaphragm. During inspiration the chest wall is lifted by the intercostal muscles, at the same time the diaphragm is lowered. The negative pressure, which is built up during this process, causes the lungs to distend thereby allowing the air to enter the lungs via the upper and lower respiratory tract and the oxygen to diffuse through the walls of the alveoli into the blood.

The oxygen is chemically combined with haemoglobin (Hb) into oxyhaemoglobin (HbO₂). This combination however can only take place if the partial pressure of oxygen (PO₂) or the oxygen concentration is sufficiently high. Air containing less than 16% of oxygen (the air that we breath contains approximately 20.96% O₂) will not combine with Hb. As soon as the HbO₂ reaches a region containing little oxygen, this compound will decompose and release the oxygen into the cells (COMROE, 1966).
During expiration, the intercostal muscles will relax and the chest wall will return to its initial position. As a result, the lung volume decreases and the air is expired. The carbon dioxide, which was produced in the cells during the oxidation process, is absorbed by the haemoglobin and to a great extent also by minerals present in the blood. The carbon dioxide diffuses through the alveolar walls and is expired. A normal individual with healthy lungs inspires approximately 10 to 15 times per minute.

2.2 REGULATION OF RESPIRATION

Respiration can only be partly influenced by man. Breath can only be held for a short time. Respiration is controlled by the autonomic nervous system (WEBSTER, 1997). The respiratory system has its own control centre, referred to as the respiratory centre, which is located in the brain stem. Respiratory nerves coming from this centre pass through the spinal cord and reach the various respiratory muscles. The respiratory centre sends electrical impulses to the respiratory muscles thus forcing them to function. It adapts the respiratory mechanism to the individual requirements. Therefore, it must be constantly informed about the individual effort. With increased physical effort, an increase amount of oxygen is required to generate energy. During this process a greater amount of carbon dioxide is produced and absorbed by the blood cells. This increased carbon dioxide content results in an accelerated generation of impulses in the respiratory centre, which sends these impulses to the muscles at a higher frequency. The greater the physical effort, the higher the amount of oxygen required. This is achieved by faster and deeper breaths. However, supply of oxygen to the lungs alone is not sufficient. The oxygen must also be delivered to the cells at a higher rate, i.e. the blood circulation must be increased considerably, which is easily achieved by a higher heart rate and increased blood pressure. There is a useful correlation between respiration and blood circulation, which is created by the chemical and nervous connections between both systems.
2.3 BLOOD CIRCULATION

The function of the blood circulation system is to carry the oxygen to the tissue cells and to remove excessive carbon dioxide from the tissue cells. The blood flow is kept in motion by the heart.

2.3.1 Heart

The heart is the motor of the blood circulation system. Its function is to pump the blood through the body to the most remote capillaries. At rest, the heart rate is approximately 70 beats per minute. The action of the heart is subdivided into two phases, systole and diastole. Systole is the contraction of the heart during which the blood is pumped into the arteries. In clinical practice, the arterial blood pressure is of great importance as it can be determined quickly and easily and to some extent allows evaluation of the current state of blood circulation. In a healthy adult the systolic pressure is approximately 120 mmHg (15.9 kPa). Diastole is the relaxation of the heart between two systoles during which both ventricles are filled with blood. Since the arteries are elastic structures, the pressure during the ventricular diastole will not drop to zero but to approximately 80 mmHg (10.7 kPa) in a healthy adult (MARIEB, 1992).

2.3.2 Systemic Circulation

The systemic circulation starts at the left ventricle. The function of the systemic circulation is to provide oxygenated blood (light red blood) to the organs and tissues via the arteries and to remove deoxygenated blood (dark red blood) saturated with carbon dioxide via the veins. In addition, the systemic circulation controls the body temperature (MARIEB, 1992).

2.3.3 Pulmonary Circulation

The function of the pulmonary circulation is to carry the blood saturated with carbon dioxide to the lungs via the pulmonary artery and blood saturated with oxygen from the lungs to the heart via the pulmonary veins (MARIEB, 1992).
2.4 THE HUMAN OESOPHAGUS: BASIC PHYSIOLOGY

The oesophagus or gullet starts at the continuation of the pharynx at the cricoid cartilage opposite the sixth cervical vertebra, and is from 23 to 25 cm in length (10 cm in newborn). It descends in front of the sixth and seventh cervical vertebrae and the upper ten thoracic vertebrae, then passes through the oesophageal opening in the diaphragm (Figure 2.3). It ends at the cardiac orifice of the stomach 1.5 cm further on, behind the seventh left costal cartilage 2.5 cm from the median plane (ROMANES, 1972).

Figure 2.3: Diagram showing the course of the human oesophagus in relation to the vertebral column, the trachea, and the aorta (ROMANES, 1972)

The oesophagus follows the longitudinal curve of the vertebral column. In the midline at its origin, it curves slightly to the left to return to the midline at the level of the fifth thoracic vertebra. At the seventh thoracic vertebra it again curves to the left and forwards across the descending thoracic aorta to the
diaphragm (Figure 2.3). The structures immediately in front of the oesophagus are the whole length of the trachea, to which it is connected by some loose tissue, the aortic arch to the left, the left bronchus, the right pulmonary artery, the pericardium posterior to the atrium, and the diaphragm.

The oesophagus is said to have four constrictions: (1) at its beginning 15 cm from the incisor teeth; (2) where it is crossed by the arch of the aorta in the superior mediastinum 22.5 cm from the incisor teeth; (3) where it is crossed by the left principal bronchus a little lower down; and (4) where it pierces the diaphragm 37-40 cm from the incisor teeth (18 cm in the new-born) (ROMANES, 1972).

2.4.1 Structure of the Oesophagus

The oesophagus has thick walls made up of an outer fibrous layer with some elastic fibres, a muscular layer, a submucous layer, and a mucous membrane (Figure 2.4). The outer fibrous layer is loosely arranged and permits considerable freedom of movement and variation in diameter.

![Figure 2.4: Transverse section through the wall of the oesophagus (ROMANES, 1972) (Image)](image-url)
The muscular coat consists of outer longitudinal and inner circular muscle fibres. In the human oesophagus, the upper third or so has striated muscle only, the lower third non-striated muscle only, and the middle portion a varying proportion of striated and non-striated muscle. While the cricopharyngeous acts as a sphincter at the commencement of the oesophagus, there is no anatomical sphincter at the gastric end, although radiological examination shows that there is a functional sphincter with some hold-up of the food before it enters the stomach, which may be due to local contraction of the circular muscle encircling the oesophageal opening in the diaphragm (ROMANES, 1972; MARIEB, 1992).

2.4.2 Blood Vessels of the Oesophagus
The cervical oesophagus is supplied mainly by branches of the inferior thyroid artery. The chief blood supply of the thoracic oesophagus is either by branches of the bronchial arteries, or direct from the aorta, and there are branches of the left gastric and inferior phrenic arteries running along the surface of the lower oesophagus for 2 to 3 cm above the diaphragm. There is a submucous plexus of a dozen or more longitudinal venous channels, with branches penetrating the oesophageal wall at intervals to join the inferior thyroid and the vertebral veins above, the azygos and the hemiazygos veins lower down, and the left gastric vein in the abdomen. There are large submucous venous trunks forming anastomotic channels between the left gastric vein and the azygos vein at the lower end of the oesophagus, thus forming a connection between the portal and systemic venous systems (ROMANES, 1972; MARIEB, 1992).

2.4.3 Gastro-Oesophageal Junction
When empty the oesophagus is collapsed. Fluid passes directly, with no hold-up, into the stomach, but with solid food there is a hold-up of about 2 seconds at the gastric end of the oesophagus. Moreover, there is no reflux of gastric contents into the oesophagus (ROMANES, 1972).
CHAPTER 3

Photoplethysmography

Photoplethysmography is a non-invasive optical technique widely used in the study and monitoring of the pulsations associated with changes in blood volume in a peripheral vascular bed (ROBERTS, 1982; DORLAS and NIJBOER, 1985; HIGGINS and FRONEK, 1986; LINDBERG and OBERG, 1991). Whether the term “plethysmography”, is a misnomer is a matter of debate, yet the title has received general consent. An excellent review of photoplethysmography was made by CHALLONER (1979). This chapter presents a brief review of the history of photoplethysmography, the basic principles of the method, and practical applications.

3.1 PRINCIPLE OF PHOTOPLETHYSMOGRAPHY

The human body is normally assumed opaque to light transmission, but most soft tissues will transmit both visible and infrared radiation to some extent (WEBSTER, 1997). For example when a hand is held over the end of a torch in a dark room it can be seen that a small but perceptible amount of light is transmitted through the whole hand. This transmitted light has a definite red colour due to selective absorption by the blood in the hand. Furthermore, on the side where the torch is held on the skin surface there is a “halo” of back scattered light around the body of the torch. As the eye is essentially an integrating instrument, it is not possible to see the minute changes in the intensity of transmitted light which occur due to the blood pulse in the hand. If, instead of looking at this variable light level with the eye, a photoelectric detector were used, then it is comparatively easy to record these “pulsations” (CHALLONER, 1979).

This is the principle on which photoplethysmography is based. The emitted light, which is made to transverse the skin, is reflected, absorbed and scattered in the tissue and blood. The modulated light level, which emerges,
is measured using a suitable photodetector. It is possible for the hand to be directly transilluminated where the light source, usually in the region of 800 nm to 960 nm, is on one side of the skin and the detector on the other side (Figure 3.1). This method, also called *transmission mode*, is limited to areas such as the finger the ear lobe or the toe (NIJBOER et al., 1981; MENDELSON and OCHS, 1988).

![Figure 3.1](image)

**Figure 3.1:** A Photoplethysmography (PPG) probe used in the transmission mode (CHALLONER, 1979)

However, when light is directed down into the skin a proportion of this is backscattered so that it emerges from the skin adjacent to the light source. The light source and the photodetector can be positioned side by side as shown in Figure 3.2. This method, also called the *reflection mode*, allows measurements on virtually any skin area (NIJBOER et al., 1981; MENDELSON and OCHS, 1988).

![Figure 3.2](image)

**Figure 3.2:** A Photoplethysmography (PPG) probe used in reflection mode (CHALLONER, 1979)
The intensity of the reflected and backscattered light which reaches the photodetector in either reflection or transmission mode is measured and the variations in the photodetector current are assumed to be related to blood volume changes underneath the probe (NIJBOER et al., 1981; ROBERTS, 1982). These variations are electronically amplified and recorded as a voltage signal called the photoplethysmograph (PPG).

The photoplethysmographic signal (Figure 3.3) is divided into two components:

i) A DC PPG component, a relatively constant voltage offset of which the magnitude is determined by the nature of the material through which the tissue passes (skin, cartilage, venous blood, etc.). The slowly changing DC component may be extracted using a low pass filter (typical bandwidth DC – 0.5 Hz) (WEBSTER, 1997).

ii) A pulsatile or AC PPG component synchronous with the heart rate is often assumed to be related to the arterial blood volume pulse. The AC PPG pulse shapes are indicative of vessel compliance and cardiac performance. The AC component, usually has an amplitude of 1-2% of the DC value and it may be extracted by a bandpass filter (typical bandwidth 0.5 Hz to 20 Hz) (WEBSTER, 1997).

![Figure 3.3: Photoplethysmographic (PPG) waveform as measured by transmission through tissue](image-url)
3.2 REVIEW OF THE LITERATURE OF PHOTOPLETHYSMOGRAPHY

As discussed in Challoner’s review, a great pioneer, to whom the method of photoplethysmography owes much of its success, was A. B. Hertzman, who published his first paper on the subject of photoplethysmography in 1937. In this paper pointed out by Challoner the method was used to show blood volume changes induced by exercise, cold and the valsala manoeuvre (CHALLONER, 1979; DORLAS and NIJBOER, 1985).

During the next three decades Hertzman and his associates contributed greatly to the understanding of this technique (CHALLONER, 1979). The earlier workers in this field were limited by the size, reproducibility and sensitivity of their photodetectors. However, with the advance of semiconductor technology these detectors have become much smaller and more sensitive and later authors have been able to investigate the technique further.

Photoplethysmography applied to the skin has found many applications. Some of the most important applications as discussed in Challoner’s review are: Cummings in 1969 recorded the vasodilatory effect of drugs, and Ramsay in 1969 followed the changes induced by thymoxamine in patients with peripheral vascular disease. Also, Hertzman and Dillon in 1940 first used photoplethysmography for assessing the completeness of sympathectomy. Measurement of cutaneous blood pulse height was used as an assessment of the effectiveness of vascular surgery by Eldrup-Jorgensen et al. in 1966. One field in which photoplethysmography has proved of great value was in demonstrating viability of tube pedicle flaps raised in plastic surgery techniques by Hayes et al. in 1967 (CHALLONER, 1979).

However, photoplethysmography has not been restricted to the human skin. Many other organs have been studied. FERRARA et al. (1988) has made photoplethysmographic measurements on animal organs such as the bowel, liver, kidney, and brain. Other clinical applications of photoplethysmography
such as measurements of systolic blood pressure, detecting brachial artery compression, locating vessels for cannulation, monitoring vascular integrity after replantation, and early detection of hypovolaemia have been discussed by KELLEHER (1989).

There has been continued development of the technique with several improvements and refinements. Although this technique has been used over the past 65 years, there are still disagreements about how the information collected arises in photoplethysmography and there are some doubts and fundamental misconceptions associated with the method, some discussed below (ROBERTS, 1982; DORLAS and NIJBOER, 1985).

### 3.3 ORIGIN OF THE PHOTOPLETHYSMOGRAPHIC SIGNAL

As discussed in Challoner’s review Hertzman in 1937 first coined the term plethysmograph. In an earlier section of this chapter it was pointed out that there is not total agreement that this is a strictly accurate name. An etymological definition would suggest that a plethysmograph records volume; thus volumetric changes are recorded in the blood vessels of an organ. But whether photoplethysmography measures only blood volume changes is open to question. The origin of the PPG signal has been the subject of continuing debate (CHALLONER, 1979; ROBERTS, 1982).

In the transmission mode it is quite conceivable that the excursions of the PPG signal are caused by variations in absorption caused by the greater volume of blood between the light source and the photodetector. If it is assumed that the change of attenuation of the light signal is due to absorption of the blood, then the more blood in the sector (tissue volume between the light source and the detector) the less light will be received by the detector and the converse situation also applies (CHALLONER, 1979; ROBERTS, 1982). Unfortunately the situation is not this simple. Challoner points out that D’Agrosa and Hertzman in 1967 obtained blood pulse recordings using photoplethysmography from individual vessels of the mesentery (CHALLONER, 1979). They concluded that with their
experimental set-up it was not possible that they could be recording volumetric changes and suggested that the alternating signal was due to a change in the optical transmittance of the blood due to change in orientation of the erythrocytes. The question of the role of erythrocyte orientation in the generation of the PPG signal has still not been resolved. One possible explanation based on the fluid mechanics of blood, shows that the axis of the erythrocytes changes during the cardiac cycle (Figure 3.4). During diastole the erythrocytes tend to align their major diameter parallel to the direction of flow, whereas during systole their major diameter aligns perpendicular to the flow, thus presenting a greater absorption path length (NIJBOER et al., 1981; MOYLE, 1994).

![Figure 3.4: Erythrocytes changing their axis during cardiac cycle (MOYLE, 1994)](image)

A simple experiment was carried out by Challoner in 1971 using whole blood circulated by a peristaltic pump through a system of closed glass tubes to give a pulsatile flow at a pump rate of 1 Hz (CHALLONER, 1979). A transmittance photoplethysmographic probe was placed on the glass tube and the whole apparatus was covered with black cloth so that no ambient light could affect the results. From this arrangement a normal photoplethysmographic recording was easily and clearly obtained. As it was not possible for there to be any volumetric change in the glass tube, the conclusion was that there was a change in transmission of the blood during different parts of the pressure cycle, which is most probably caused, as suggested by D'Agrosa and Hertzman, by the orientation of the erythrocytes during the cardiac cycle (CHALLONER, 1979; NIJBOER et al., 1981; MOYLE, 1994).
It can now be seen that there are at least two factors influencing the amount of light transmitted to the detector from the light source and that these are blood volume and orientation of erythrocytes. Which of these is the more important factor is at present not known, consequently the true mechanism underlying this method is not clearly understood. Nevertheless, arterial blood pulsations are easily obtained by this technique.

3.4 PHOTOPLETHYSMOGRAPHY AND PULSE OXIMETRY

Pulse oximeters, as will be discussed in more detail in the following chapters, estimate arterial oxygen saturation by shining light at two different wavelengths, red and infrared, through vascular tissue. In this method the pulsatile photoplethysmographic (AC PPG) signal associated with cardiac contraction, as described in the previous section, is assumed to be attributable solely to the arterial blood component. The amplitudes of the red and infrared AC PPG signals are sensitive to changes in arterial oxygen saturation because of differences in the light absorption of oxygenated and deoxygenated haemoglobin at these two wavelengths. From the ratios of these amplitudes, and the corresponding DC photoplethysmographic components, arterial blood oxygen saturation (SpO$_2$) is estimated. Hence, the technique of pulse oximetry relies on the presence of adequate peripheral arterial pulsations, which are detected as photoplethysmographic (PPG) signals (MENDELSON and OCHS, 1988).
CHAPTER 4

Review of the History of Blood Oxygen Saturation and Pulse Oximetry

The development of pulse oximetry has been based on more than a hundred years of experimental and engineering antecedents. The discovery that enabled the others to come together in a single instrument was the combination of transmission oximetry and photoplethysmography in 1974 by the Japanese bioengineer Takuo Aoyagi, who developed the first pulse oximeter. SEVERINGHAUS and ASTRUP (1986) have reviewed in detail the development of blood gas analysis. This review includes the fascinating history of pulse oximetry, of which the following is a brief summary.

The relationship between the absorption of light and the amount of absorbant was first described by Johann Heinrich Lambert (1728-77) in Augsberg, Germany, and published by him in 1760. Lambert’s ideas were further investigated by August Beer (1853-1932), who published his findings as the Beer-Lambert law in 1851 (SEVERINGHAUS and ASTRUP, 1986).

In 1864 Stokes reported his discovery that the coloured substance in blood is also its carrier of oxygen. Around the same time, Hoppe-Seyler, who first crystallised this substance and coined for it the term haemoglobin, showed that its pattern of absorption of light from the solar spectrum changed when it was shaken with air. The haemoglobin-oxygen dissociation curve was elucidated, and, in subsequent decades, Hufner studied the optical spectra of haemoglobin and oxyhaemoglobin (KELLEHER, 1989).

The first device to use these spectra to measure oxygen saturation in the human body by transilluminating it with coloured light was built in 1935 by
Matthes. He based his invention on studies by Nicolai, who initiated the quantitative spectrophotometric investigation of light transmitted through human skin in an effort to understand the dynamics of tissue oxygen consumption (SEVERINGHAUS and ASTRUP, 1986). Nicolai devised the apparatus (Figure 4.1) that was to become the oximeter using the blue-green bands of mercury vapor light, a rotating wheel filter, a vacuum photoelectric cell, and a vacuum tube amplifier. By occluding the circulation, he obtained exponential curves for the decay of oxyhaemoglobin and the increase in deoxyhaemoglobin and showed that the logarithm of the light changed linearly with time (SEVERINGHAUS and ASTRUP, 1986; KELLEHER, 1989).

Figure 4.1: Ludwig Nicolai’s first apparatus for studying oxygen saturation in tissue (SEVERINGHAUS and ASTRUP, 1986).

The first precise measurements of oxygen saturation in cuvettes and in flowing blood in vitro were reported in 1934 by Nicolai’s associate Kurt Kramer (1906-1985). Kramer instead of following Nicolai’s lead chose to use an incandescent light, a red filter, and a photocell with a galvanometer, and he demonstrated that saturation could be measured reproducibly. In 1935 Kramer reported a way (Figure 4.2) to record continuously the oxygen saturation of blood flowing through unopened vessels in animals. He slightly compressed the artery between glass walls in a trough to fix the optical path length between the incandescent light and the barrier layer photocell (SEVERINGHAUS, 1993; WAHR et al., 1995).
Matthes and his collaborator Franz Gross were the first to use two spectral regions, one not affected by oxygen, to compensate for changes in tissue thickness, blood content, light intensity, and other variables (SEVERINGHAUS and ASTRUP, 1986). They first used blue-green light, following Nicolai’s lead, but later switched to infrared. In 1939 they described the first red and infrared ear oxygen saturation meter. Their infrared detector was a gas filled phototube covered with a filter transmitting only infrared light. This made it rather large and inconvenient to use.

The problem of excluding venous and capillary blood to measure only oxygen arterial saturation ($\text{SaO}_2$) was addressed in the early 1940s by Goldie and Squire, who zeroed an oximeter by taking “bloodless” readings from the web of the hand. Red and infrared light transmission were compared by changing filters (Figure 4.3). Squire used a method, later used by Wood, of setting the zero of the two optical channels by compressing the tissue to squeeze out the blood (KELLEHER, 1989; SEVERINGHAUS, 1993).
In 1941, Millikan was principally responsible for the development of a lightweight and practical aviation ear oxygen meter for which he coined the term “oximeter” (Figure 4.4). This instrument was designed for the British war effort specifically to address the problem of loss of consciousness by aviators during “dogfights”. Its use was first reported in connection with surgical anaesthesia in 1948 (MENDELSON, 1992).
In its initial clinical development, the ear oximeter had several limitations. It was a delicate instrument that required a technician to operate and maintain. The earpiece was large, difficult to position, and produced enough heat to cause second degree burns on the pinna. Furthermore, it required calibration on each patient prior to use. During the 1950s, Earl Wood devised a modification of the Millikan earpiece, by improving the infrared filter and adding an inflatable balloon with which the ear could be made bloodless for initial zero setting, that was used in many clinical and laboratory investigations. Although the ear oximeter showed promise in some settings, it was still considered a research tool (TREMPER and BARKER, 1989; WAHR et al., 1995).

Around 1964, Robert Shaw, a surgeon and inventor, began the design and construction of an eight-wavelength ear oximeter (SEVERINGHAUS, 1993). His concept was to solve uniquely the simultaneous equations by using one more wavelength than the number of separate forms of haemoglobin needing identification. In the 1970s, Hewlett-Packard marketed the first eight wavelengths ear oximeter (Figure 4.5).

Figure 4.5: Simplified diagram of the Hewlett-Packard ear oximeter (MOYLE, 1994)

Hewlett-Packard’s oximeter also used the method of heating the ear to “arterialised” the capillary blood. This oximeter quickly became a standard clinical and laboratory tool in pulmonary medicine. Although it was demonstrated to be accurate for intraoperative monitoring, its size and
expense, and the cumbersome nature of the ear probe prevented its acceptance as a routine monitor (SEVERINGHAUS, 1993).

Meanwhile, an entirely separate phototechnology had been awaiting application to this problem for years. The recording of a pulse wave in the finger was first reported in 1911, and the technique of photoplethysmography was born in the mid-1930s when changes in the intensity of transmitted light were used to describe this waveform. It had been used mainly to assess the adequacy of perfusion in extremities (SEVERINGHAUS and ASTRUP, 1986).

Aoyagi an engineer working for Nihon Kohden, originated the idea of using photoplethysmography to synchronise spectrophotometric measurements of haemoglobin saturation with the peak and trough of the pulse waveform, thereby isolating and characterising a purely pulsatile component (KELLEHER, 1989). He discovered the principle in the early 1970s while trying to accomplish something else altogether. He was developing a method to estimate cardiac output semi-noninvasively by detecting the washout curve of dye injected into a peripheral vein as it perfused the ear. This washout curve was measured in the ear with a red and infrared light densitometer similar to the Millikan ear oximeter. He notice that his washout curves contained pulsations due to the arterial pulse in the ear (SEVERINGHAUS and ASTRUP, 1986). To more easily analyse the dye washout curve, he subtracted these pulsations from the curve, and in doing so he discovered that the absorbance ratio of the pulsations at the two wavelengths changed with arterial haemoglobin saturation. He soon realised that he could build an ear oximeter that measured arterial haemoglobin saturation without heating the ear by analysing pulsatile light absorbances. This first pulse oximeter, developed by Nihon Kohden, used filtered light sources similar to Millikan’s ear oximeter. The device was evaluated clinically in the mid 1970s and marketed with little success (KELLEHER, 1989; WAHR et al., 1995). A competing pulse oximeter was introduced by the Minolta Co, marketed in the United States as the Oximet MET-1471 in 1977 with a finger tip probe and fibre optic cables form the instrument (Figure 4.6). The device was tested clinically from 1976 to 1980 and was found to provide continuous, non-
invasive monitoring of oxygen saturation (SEVERINGHAUS, 1993; WAHR et al., 1995). However, it overestimated saturation, averaging a reading of 70% at 50% actual saturation. Shimada and colleagues showed that the overestimation was due to multiple scattering in the finger, and that the theoretic equations did not apply (SHIMADA et al., 1984).

Figure 4.6: Simplified diagram of the Minolta pulse oximeter: Light transmitted by the halogen lamp is transferred by the glass fibres and applied to the fingertip. The transmitted light is led through 650 nm and 805 nm filters reaching photocells that convert the light energy into an electrical signal (MOYLE, 1994).

In the late 1970s, Scott Wilber in Boulder, Colorado, developed the first clinically accepted pulse oximeter by making two modifications of the Nihon Kohden method (SEVERINGHAUS and ASTRUP, 1986; TREMPER and BARKER, 1989). First, he produced a lightweight sensor by using light emitting diodes (LEDs) as light sources and photodiodes as photodetectors. Consequently, the instrument was connected to its ear clip sensor only by a small electrical cable. Wilber also improved the saturation estimates by using a digital microprocessor to store a complex calibration algorithm based on human volunteer data. This device was developed by Biox Corporation of Boulder, Colorado, and was successfully marketed to pulmonary function laboratories in the early 1980s (SEVERINGHAUS, 1993).

The clinical utility of the non-invasive oximeter in the operating room was rediscovered in the 1980s by William New, an anaesthesiologist at Stanford University. New, also an electrical engineer, teamed with Jack Lloyd to found Nellcor, Inc.. Nellcor produced a microprocessor-based pulse oximeter, the N100, which was smaller, less expensive, needed no user calibration, and
was accurate enough for clinical use. Nellcor is still one of the market leaders in pulse oximetry. About the same time, Ohmeda came up with a similar device, the BIOX II, which had the same success (KELLEHER, 1989; WAHR et al., 1995). Table 4.1 summarises this chapter with a synoptic chronological history of the evolution of pulse oximetry.

Pulse oximeters now occupy most critical care areas and virtually every operating theatre. They require no calibration by the user or matching to each subject. A major advance in pulse oximetry has been the reduction in the size of the light source and the detectors. Photodiodes and light emitting diodes at red and infrared wavelengths have made pulse oximetry probes inexpensive enough to be considered disposable in some models. They have been found to work on the ear, the bridge of the nose, the nasal septum, the finger, the temple over the temporal artery, and on the foot or palm in infants. The number of articles published in the past decade that describe the use and evaluation of pulse oximetry suggests that explosive growth has already occurred in its use.
<table>
<thead>
<tr>
<th>Year</th>
<th>Event</th>
</tr>
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<tbody>
<tr>
<td>1851</td>
<td>Beer-Lambert law</td>
</tr>
<tr>
<td>1864</td>
<td>Georg Gabriel Stokes discovers a pigment that is the oxygen carrier in blood</td>
</tr>
<tr>
<td>1864</td>
<td>Felix Hoppe-Seyler purifies the pigment and calls it haemoglobin</td>
</tr>
<tr>
<td>1887-90</td>
<td>Carl Gustav Hufner (1840-84) studies absorption spectra of haemoglobin</td>
</tr>
<tr>
<td>1931</td>
<td>Ludwig Nicolai investigates the quantitative spectrophotometry of light transmitted through human tissues</td>
</tr>
<tr>
<td>1934</td>
<td>Kurt Kramer (1906-85) makes precise measurements of the oxygen saturation of blood flowing through cuvettes</td>
</tr>
<tr>
<td>1939-45</td>
<td>World War II: great military interest in oximetry in pilots at high altitude</td>
</tr>
<tr>
<td>1940</td>
<td>JR Squire passes red and infrared light through the finger web for the continuous monitoring of oxygenation; it requires compression of tissues to create a bloodless field for calibration</td>
</tr>
<tr>
<td>1940-42</td>
<td>Glen Millikan coins the term oximeter and develops the Millikan oximeter</td>
</tr>
<tr>
<td>1948-50</td>
<td>Earl Wood develops Wood’s ear oximeter</td>
</tr>
<tr>
<td>1960</td>
<td>Development of the first bench “CO-Oximeter” able to distinguish between haemoglobin, carboxyhaemoglobin, and methaemoglobin</td>
</tr>
<tr>
<td>1964</td>
<td>Robert Shaw develops the eight wavelength ear oximeter</td>
</tr>
<tr>
<td>1970</td>
<td>Hewlett-Packard market the eight wavelength ear oximeter</td>
</tr>
<tr>
<td>1971</td>
<td>Takuo Aoyagi uses the pulsatility of the absorption signal to separate absorption due to the arterial blood from that of other tissues</td>
</tr>
<tr>
<td>1974</td>
<td>Aoyagi develops the prototype pulse oximeter using an incandescent light source, filters, and analogue electronics</td>
</tr>
<tr>
<td>1975</td>
<td>First commercially available pulse oximeter</td>
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</tbody>
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CHAPTER 5
The Physics Physiology and Current Technology of Pulse Oximetry

This chapter describes the theoretical background for the measurement of light absorbance in pulse oximetry as a basis for determining arterial oxygen saturation. The Beer-Lambert law and the derivation of a theoretical calibration curve for measured light absorbances in pulse oximeters is explained. By itself, however this approach is incomplete, as it does not adequately account for the effects of a physical phenomenon called light scattering present within the tissue region under investigation. Biological tissue is a highly light-scattering medium and little information about its optical properties is available (WRAY et al., 1988). In conventional practice the effects of light scattering are accounted for by empirically calibrating the sensor and oximeter, and this appears to work well. Nevertheless, the Beer-Lambert law approach helps to develop an understanding of the absorbance of light as it passes through living tissue and why and how pulse oximetry works. The effects of multiple scattering and mathematical models for light absorbance, which take the scattering theory into consideration, are also discussed. A brief review on the technical developments of commercial pulse oximeters will be the subject of the last section of this chapter.

5.1 BEER-LAMBERT LAW
In the 1930’s, Matthes used spectrophotometry in vitro to determine haemoglobin oxygen saturation. This method is based on the Beer-Lambert law, which describes the attenuation of light travelling through a uniform medium containing an absorbing substance (TREMPER and BARKER, 1989). If monochromatic incident light of intensity $I_\text{in}$ enters the medium, a part of this light is transmitted through the medium while another part is
absorbed. The intensity $I_{out}$ of light travelling through the medium decreases exponentially with distance (TREMPER and BARKER, 1989).

$$I_{out} = I_{in} e^{-c(\lambda)cd} \quad (5.1)$$

Where $c(\lambda)$ is the extinction coefficient or absorptivity of the absorbing substance at a specific wavelength ($\lambda$), $c$ the concentration of the absorbing substance which is constant in the medium, and $d$ the optical path length through the medium. The concentration $c$ is measured in mmol L$^{-1}$ and the extinction coefficient is expressed in L mmol$^{-1}$ cm$^{-1}$.

The transmittance ($T$) of light travelling through a medium with an absorbing substance is defined as the ratio of transmitted light $I_{out}$ to the incident light $I_{in}$ (WEBSTER, 1997).

$$T = \frac{I_{out}}{I_{in}} = e^{-c(\lambda)cd} \quad (5.2)$$

The unscattered absorbance ($A$) of this process also known as the optical density of a medium, is defined as the negative natural logarithm of the transmittance of light (WEBSTER, 1997).

$$A = -\ln T = -\ln \left( \frac{I_{out}}{I_{in}} \right) = c(\lambda)cd \quad (5.3)$$

The properties of Beer’s law are valid if more than one substance absorbs light in the medium. The mathematical representation of this multiple absorber system is a superposition of the individual absorbing processes. The total absorbance $A_t$ of light in a medium with $n$ absorbing substances is the sum of their $n$ independent absorbances

$$A_t = \varepsilon_1(\lambda)c_1d_1 + \varepsilon_2(\lambda)c_2d_2 + \ldots + \varepsilon_n(\lambda)c_nd_n = \sum_{i=1}^{n} \varepsilon_i(\lambda)c_id_i \quad (5.4)$$

where $\varepsilon_i(\lambda)$ and $c_i$ represent the extinction coefficient and concentration of the substance $i$ and $d_i$ represents the optical path length through the absorbing substance.

Beer's law allows the determination of unknown concentrations of $n$ different absorbing substances in a homogeneous medium if the absorbance of light is
measured at \( n \) different wavelengths and the extinction coefficients of the substance are known (TREMPER and BARKER, 1989).

### 5.2 OPTICAL PROPERTIES OF BLOOD

#### 5.2.1 Haemoglobin

Haemoglobin is the main light absorber in human blood at wavelengths used in pulse oximetry (TREMPER and BARKER, 1989; MOYLE, 1994). The absorbing characteristics of haemoglobin change with its chemical binding and the wavelength of the incident light.

Adult blood usually contains four species of haemoglobin:

- **Oxyhaemoglobin (HbO\(_2\))**: haemoglobin fully saturated with oxygen
- **Deoxyhaemoglobin (Hb)**: haemoglobin not fully saturated with oxygen
- **Methaemoglobin (MetHb)**: Methaemoglobin is oxidised haemoglobin and is a result of oxidation of a free heme iron (Fe\(^+\)). MetHb levels may be high congenitally or as a result of exposure to a number of agents, most notably anaesthetics, sulfa drugs, and nitrites (EISENKRAFT, 1988).
- **Carboxyhaemoglobin (COHb)**: Is formed when haemoglobin combines with carbon monoxide (CO). People that are smokers or working in underground garages, mines etc have an increased amount of CO in the blood.

Haemoglobins that are able to bind with molecular oxygen are called functional haemoglobins. **Functional haemoglobins** are HbO\(_2\) and Hb. Most of the haemoglobins in a healthy individual are functional haemoglobins (MOYLE, 1994). Haemoglobins that are unable to bind to oxygen or interfere with the ability of oxyhaemoglobin to release oxygen to the tissue are called **dysfunctional haemoglobins**. The two most common dysfunctional haemoglobins are methaemoglobin (MetHb), and carboxyhaemoglobin (COHb) (MOYLE, 1994).

#### 5.2.2 Functional and Fractional Oxygen Saturation

The **functional oxygen saturation** (functional SO\(_2\)) is measured in percentage and determined by the amount of oxygenated haemoglobin (HbO\(_2\)) as compared to the sum of oxygenated and deoxygenated
haemoglobin (Hb). Another way to define this ratio is to use the concentrations of oxygenated haemoglobin (c_{HbO2}) and deoxyhaemoglobin (c_{Hb}) (TREMPER and BARKER, 1989).

\[
\text{Functional SO}_2 = \frac{HbO_2}{Hb + HbO_2} \times 100\% = \frac{c_{HbO2}}{c_{HbO2} + c_{Hb}} \times 100\%
\] (5.5)

The fractional oxygen saturation is the fraction of oxygenated haemoglobin to the total haemoglobin. It is usually measured in percentage as well and is determined by the ratio of the concentrations of oxygenated haemoglobin and total haemoglobin, where total haemoglobin represents all four different species of haemoglobin present in the blood (TREMPER and BARKER, 1989).

\[
\text{Fractional SO}_2 = \frac{HbO_2}{HbO_2 + Hb + COHb + MetHb} \times 100\% = \frac{c_{HbO2}}{c_{\text{total haemoglobin}}} \times 100\%
\] (5.6)

5.3 HAEMOGLOBIN ABSORBANCE SPECTRA

Figure 5.1 shows how the absorption of oxyhaemoglobin, deoxyhaemoglobin, methaemoglobin, and carboxyhaemoglobin varies in the wavelength range from the visible light (600 nm) end of the spectrum into the near infrared (1000 nm). The absorbance of light in the red region of the spectrum is much higher for deoxyhaemoglobin than for oxyhaemoglobin. The extinction coefficients of both haemoglobin species are equal at a point called the isosbestic point. The isosbestic point is the wavelength at which the extinction coefficients of two substances, in this case oxyhaemoglobin and deoxyhaemoglobin, are equal (i.e. 805 nm). The extinction coefficient of carboxyhaemoglobin is about the same as that of oxyhaemoglobin at the wavelength of 660 nm while it is almost transparent in the infrared region. Methaemoglobin absorbs much light in the red region of the spectra and its extinction coefficient remains higher than that of oxyhaemoglobin in the infrared region (WUKITSCH et al., 1988; MANNHEIMER et al., 1997a).
Figure 5.1: Extinction coefficients (Lmmol\(^{-1}\)cm\(^{-1}\)) of the four most common haemoglobin species; oxyhaemoglobin (HbO\(_2\)), deoxyhaemoglobin (Hb), carboxyhaemoglobin (COHb), and methemoglobin (MetHb) at the wavelengths of interest in pulse oximetry (TREMPER and BARKER, 1989)

5.4 WAVELENGTHS USED IN PULSE OXIMETRY

Pulse oximeters determine arterial blood oxygen saturation by measuring the light absorbance of tissue at two different wavelengths and using the arterial blood pulsation to differentiate between absorbance of arterial blood and other absorbers (skin, bone, venous blood). A good choice of wavelength is where there are large differences in the extinction coefficients of deoxyhaemoglobin and oxygenated haemoglobin. Another criterion for the wavelength selection is the relative flatness of the absorption spectra (see more detail explanation in section 6.1) around the chosen wavelength (MOYLE, 1994; MANNHEIMER et al., 1997b). The two conventional wavelengths used in pulse oximetry are the 660 nm (red) and 940 nm (near infrared) (Figure 5.2).
Figure 5.2: Absorption coefficient versus wavelength of oxyhaemoglobin (solid) and deoxyhaemoglobin (dashed) (MANNHEIMER et al., 1997a)

The choice of the near infrared at 940 nm was easy, as the absorption spectra of HbO$_2$ and Hb are reasonably flat at this wavelength, so that slight variation in peak wavelength will make little difference to calibration. The advantage of emitting light at 660 nm is that there is a comparatively large difference in absorption between deoxyhaemoglobin and oxyhaemoglobin at this point on the spectrum, yielding detectable changes in absorption with small change in oxygenation. Although the curves are not horizontal at 660 nm, they are considerably flatter than at shorter wavelengths (MENDELSON, 1992).

5.5 BEER’S LAW IN PULSE OXIMETRY

Theoretical descriptions of pulse oximetry often begin with a discussion of the Beer-Lambert Law of light absorption. By itself however, this approach is incomplete, as it does not adequately account for the effects of light scattering. In conventional practice, the effects of light scattering are accounted for by empirically calibrating the sensor and oximeter, and this appears to work well, but only up to a certain point (FINE and WEINREB, 1993; WEBSTER, 1997). Assumptions inherently made during an empirical calibration are valid only for a limited range of saturations, and become invalid under extreme conditions. The Beer-Lambert Law approach will be followed here in order to derive the form of equations utilised in pulse
oximetry. The effects of scattering on practical measurements of arterial saturation will also be described.

According to the Beer-Lambert Law, as described in the first section of this chapter, light attenuation after passing through a non-scattering absorbing solution may be written as a linear combination of the absorbance from each of the contributors (Equation 5.4). In a scattering medium such as tissue, the path length is not simply the width of the sample cuvette (Figure 5.3).

![Figure 5.3: Schematic diagrams for use in considering the Beer-Lambert Law with and without scattering. In the figure on the left, the light is not scattered, thus the path length is given by the width of the sample cuvette, \( d \). On the right, each photon or ray, follows a random path characterised by the scattering and absorption properties of the material. The average distance travelled by the light is given by \( d_{\text{max}} \) and is necessarily greater than the width of the cuvette. In both figures, the shaded and unshaded dots are considered to represent different spectral light absorbers (MANNHEIMER et al., 1997b)

The actual path length of the detected light instead becomes a complicated function of both the scattering and absorption properties of the medium. In order to account for the effects of scattering in this discussion, let \( d_{\text{max}} \) refer to the effective mean path length of the detected light, thus combining all of the effects of scattering into one term.

If we apply Equation 5.3 at one point in time within the cardiac cycle (at \( t_1 = \text{diastole} \)) shown in Figure 5.4, we observe the attenuation effects of all the absorbing and scattering components (DC photoplethysmographic component) of the tissue, e.g. non-pulsating arterial blood, venous blood, and other tissue. By applying Equation 5.3 again at a different time within the
cardiac cycle (at \(t_2\)=systole), and then subtracting, we are left with only the change in absorbance due to the pulsatile change in arterial blood concentration (AC photoplethysmographic component), since the tissue concentrations of the other constituents do not change over this time period.

![Figure 5.4: Absorbed and transmitted light in living tissue. The amount of absorbed light correlates with the pulsation of arterial blood. A constant amount of light is absorbed by the skin pigmentation, bone, other tissue, venous blood and other non-pulsating part of the arterial blood. More blood is present in the arteries during systole (\(t_2\)) than during diastole (\(t_1\)) and therefore more light is absorbed (WEBSTER, 1997)](image)

The unknown input intensity value \(I_{in}\) from Equation 5.3 cancels since it is unchanging. We find;

\[
A(t_2) - A(t_1) = \ln(I_{out}(t_2)) - \ln(I_{out}(t_1)) = d_{max} \epsilon_i(\lambda) ([c_i(t_2)] - [c_i(t_1)])
\]

where \(\epsilon_i(\lambda)\) and \(c_i\) represent the extinction coefficient and concentration of the substance \(i\) and refer only to the arterial blood in the tissue. We can replace \(\epsilon_i(\lambda)\) with a linear combination of the oxyhaemoglobin (HbO\(_2\)) and deoxyhaemoglobin (Hb) contributions in the arterial blood as a function of the saturation SO\(_2\). The last equality of Equation 5.7 becomes:

\[
d_{max} \epsilon_i(\lambda) \Delta[c_i] = d_{max} [SO_2 \epsilon_{iHbO_2}(\lambda) + (1-SO_2) \epsilon_{iHb}(\lambda)] \Delta[c_i]
\]

\[(5.8)\]
where,
\[ SO_2 = \frac{c_{HbO_2}}{c_{HbO_2} + c_{Hb}} \] from Equation 5.5

and \( \Delta[c_i] \) refers to the change in tissue arterial blood concentration between the two points in time.

Pulse oximeters perform the measurement indicated in Equation 5.7 at two wavelengths and calculate their ratio, \( R \):

\[
R = \frac{[\ln(I_{out}(t_2)) - \ln(I_{out}(t_1))]_{\lambda_1}}{[\ln(I_{out}(t_2)) - \ln(I_{out}(t_1))]_{\lambda_2}} = \frac{[SO_2 \epsilon_{HbO_2}(\lambda_1) + (1 - SO_2) \epsilon_{Hb}(\lambda_1)] \Delta[c_i]_{\lambda_1} d^{ex}_{\lambda_1}}{[SO_2 \epsilon_{HbO_2}(\lambda_2) + (1 - SO_2) \epsilon_{Hb}(\lambda_2)] \Delta[c_i]_{\lambda_2} d^{ex}_{\lambda_2}} \tag{5.9}
\]

The superscripts and subscripts of \( \lambda_1 \) and \( \lambda_2 \) refer to the values of these parameters at the two wavelengths used for pulse oximetry (red (R) and infrared (IR)).

Equation 5.9 will be used as the basis for the discussions below, but first consider two simplifying (but incorrect) assumptions that are commonly made at this point. First, the path lengths at the two wavelengths are considered to be equal, even in the presence of scattering, and cancel as they appear both in the numerator and denominator of Equation 5.9. The second assumption is that the \( \Delta[c_i] \) terms are also equivalent at the two wavelengths and cancel. Recall that this term refers to the pulsatile changes of the tissue’s blood concentration that causes the detected light levels to modulate. Once these two simplifications are made, Equation 5.9 can be rewritten:

\[
R = \frac{[\ln(I_{out}(t_2)/I_{out}(t_1))]_{\lambda_R}}{[\ln(I_{out}(t_2)/I_{out}(t_1))]_{\lambda_{IR}}} = \frac{[SO_2 \epsilon_{HbO_2}(\lambda_R) + (1 - SO_2) \epsilon_{Hb}(\lambda_R)]}{[SO_2 \epsilon_{HbO_2}(\lambda_{IR}) + (1 - SO_2) \epsilon_{Hb}(\lambda_{IR})]} \tag{5.10}
\]

Equation 5.10 indicates a measurement of a ratio \( R \) that is dependent only on the arterial saturation (\( SO_2 \)) and the absorption properties of \( HbO_2 \) and \( Hb \).

We can solve this equation for \( SO_2 \) (now referred to as \( SpO_2 \); indicating arterial blood oxygen saturation by pulse oximetry) and thus estimate the saturation based on the measurement:
$$SpO_2 = \frac{\varepsilon_{Hb}(\lambda_R) - R \varepsilon_{Hb}(\lambda_{IR})}{R[\varepsilon_{HbO_2}(\lambda_{IR}) - \varepsilon_{Hb}(\lambda_{IR})] + \varepsilon_{Hb}(\lambda_R) - \varepsilon_{HbO_2}(\lambda_R)}$$ \hspace{1cm} (5.11)

Experience in calibrating pulse oximeters has shown that the simplifications mentioned above are not completely valid (MANNHEIMER et al., 1997b). The solid line graphed in Figure 5.5 shows the readings that would be seen empirically if the system were calibrated based solely on Equation 5.11, utilising extinction coefficients available in the literature (ZIJLSTRA et al., 1991) for a 660 nm and 890 nm wavelength pair.

**Figure 5.5:** Displayed SpO\(_2\) values for a pulse oximeter if calibrated using the simple Beer-Lambert Law (solid line). SaO\(_2\) is shown on the abscissa, the ordinate is the displayed SpO\(_2\). The Scattering Theory used here derived based on the works of Bonner et al. (MANNHEIMER et al., 1997b)
5.6 LIGHT SCATTERING

The resulting error of the calibration curve derived from the Beer-Lambert Law using Equation 5.11, shown in Figure 5.5, is generally attributed to the effects of scattering and multiple scattering in the tissue that cause the path lengths to be unequal at the two wavelengths (DELPY, 1988; SCHMITT, 1991).

Light traversing tissue suffers a loss in intensity due to the effects of both scattering and absorption (COPE and DELPY, 1988). Light scattering causes the deviation of a light beam from its initial direction. It occurs when light interacts with an object of a size similar to the magnitude of the wavelength of the light and a change in the index of refraction at the interface of this object (WEBSTER, 1997). The wavelengths of infrared and red light do have the same order of magnitude as the geometric dimensions of red blood cells (approximately 7μm in diameter). The discontinuity in the index of refraction at the interface between plasma and red blood cells and the high proportion of red blood cells in blood yield a highly light scattering medium. Light that is scattered once will likely be scattered again by cells and therefore multiple scattering occurs (STEINKE and SHEPHERD, 1992). Multiple scattering increases the optical path length and therefore increases the measured attenuation.

The intensity of the light scattered by the tissue depends on such factors as the red blood cell concentration in the blood; on the size, shape, orientation, and index of refraction of the scattering particles; on the tissue thickness; and on the aperture cone of the detector (FINE and WEINREB, 1995). The thickness of the tissue, the distance between the emitters and the detector, and the concentration of haemoglobin will vary from patient to patient and the shape and orientation of the red blood cells is irregular. Thus, it is difficult to develop a physical model which, can be used under different circumstances.

It would be very useful to find a relationship between SaO₂ and the ratio R that takes the scattering theory into consideration (SHIMADA et al., 1984). One attempt is described below.
According to BONNER et al. (1987), the mean path length travelled by detected photons for large emitter-detector separations is approximately proportional to \( r(\mu_a / \mu_s)^{1/2} \), where \( r \) is the emitter-detector separation, \( \mu_a \) is total absorption coefficient for the blood perfused tissue, and \( \mu_s \) is the reduced scattering coefficient. Both \( \mu_a \) and \( \mu_s \) have the units of inverse length. Since the absorption and scattering coefficients are wavelength-dependent, and \( \mu_a \) varies with oxygen saturation the path lengths can indeed be expected to be different. As long as the path length ratio is constant, however, or is a function of \( \text{SaO}_2 \), their unknown value can be absorbed into an empirical calibration, as in fact is done in practice. Calculating the path lengths \((d_{\text{max}})_R\) and \((d_{\text{max}})_\text{IR}\) according to published formula (BONNER et al., 1987), and incorporating them to Equation 5.9, results in a calibration estimate which better matches empirical observations (dashed line in Figure 5.5). Most of the remaining differences between empirical and theoretical calibration are likely to relate to specific features of the sensor and tissue bed that cannot be readily incorporated into numerical models.

### 5.7 CALIBRATION OF MODERN PULSE OXIMETERS

It could be considered that accurate values of \( \text{SpO}_2 \) can be obtained by application of simultaneous equations describing the Beer-Lambert Law, but this is not the case, and most modern pulse oximeters apply the “red:infrared ratio” (\( R \)) to a “look-up table” (MOYLE, 1994).

\[
R(\text{ratio}) = \frac{AC_{660} / DC_{660}}{AC_{940} / DC_{940}}
\]  

(5.12)

The manufacturers calibrate pulse oximeters empirically by correlating the measured ratio (\( R \)) of AC/DC signals from the red and infrared photoplethysmographs obtained from a large group of healthy volunteers with arterial \( \text{SaO}_2 \) values generally greater than 70%. Blood \( \text{SaO}_2 \) values are obtained directly from a standard \textit{in vitro} oximeter such as an Instrumentation Laboratory CO-Oximeter. This calibration procedure involves desaturating the subjects by asking them to breathe hypoxic gas mixtures and collecting optical measurements of blood samples at different steady-state oxygenation levels. Consequently, different brands of pulse oximeters may display slightly
different values, depending on the internal calibration of the oximeter. A typical relationship between the ratio (R) and SaO\textsubscript{2} is shown in Figure 5.6.

![Graph showing empirical relationship between arterial saturation and "red:infrared" ratio (R) (MOYLE, 1994)](image)

**Figure 5.6:** Empirical relationship between arterial saturation and "red:infrared" ratio (R) (MOYLE, 1994)

At approximately 85% SaO\textsubscript{2}, the amount of light absorbed by Hb and HbO\textsubscript{2} is about the same. Hence the amplitudes of the red and infrared signals are equal (i.e. R/IR=1.0)

One of the limitations of this traditional calibration method is the limited range of oxygen saturation that can be acquired. Ethical issues prevent intentional desaturation of healthy subjects below a certain point due to risk of hypoxic brain damage. A more detail discussion on the calibration limitations is presented in chapter 10.

### 5.8 TECHNICAL DEVELOPMENTS OF COMMERCIAL PULSE OXIMETERS

A typical commercial pulse oximeter consists of an opto-electronic probe that is applied to the patients periphery, usually the finger and a microprocessor-controlled electronic system. This system processes the AC and DC photoplethysmographic signals at the two wavelengths, red and infrared, and
displays the blood oxygen saturation measurement ($\text{SpO}_2$) in percent together with heart rate.

The technology of pulse oximetry is well reviewed in a highly readable account by POLOGE (1987). This section describes briefly the main electronic blocks comprising the pulse oximeter system, such as, the analogue and digital processing, the microprocessor, the power supply, and the front display.

5.8.1 Basic Block Diagram of a Pulse Oximeter

Figure 5.7 shows a simplified block diagram of a pulse oximeter.

**Figure 5.7:** Block diagram of a commercial pulse oximetry showing the major electronics components

**Emitter Driver**

The pulse oximeter probe, discussed in detail in the following chapter, has a single photodetector receiving signals from the infrared (IR) and red (R) emitters. The currents through the emitters are controlled by the emitter driver (Figure 5.7) which comprises a pair of current sources (R and IR) and a multiplexer. Emitter drive currents may vary between 60 mA to 120 mA. The multiplexer turns the red and infrared emitters on and off at a rate of approximately 325 Hz (but varies with different manufacturers) in sequence (WEBSTER, 1997). The timing signal from the microprocessor controls the multiplexer.
Preamplifier
The photodetector generates a current proportional to the light incident upon it. The multiplexed PPG signals from the photodetector are received by a preamplifier. The preamplifier consists of a differential current-to-voltage amplifier that converts the current output of the photodetector into a signal voltage.

Demodulator and filtering
The inputs to this circuit are the mixed PPG signal output (red and infrared) of the current-to-voltage amplifier and the timing signal from the microprocessor. The microprocessor along with the information stored in the EPROM, calculates the time period each signal component is present in the photodetector output. Switching at the right time results in the two components (red and infrared) getting separated. To eliminate the high frequency switching noise, low pass filters are provided (POLOGE, 1987).

Analogue-to-Digital Converter (ADC)
This is the point in the system where digital electronics take over. The PPG signals are converted by an analogue-to-digital converter to a form suitable for manipulation by the microprocessor. In most of the commercial pulse oximeters the analogue-to-digital conversion is accomplished by using a sample and hold circuit, which holds a voltage until it is sampled by the ADC controlled by a routine written in the memory of the microprocessor (POLOGE, 1987; WUKITSCH et al., 1988).

Microprocessor
The most important component of the pulse oximeter system is the microprocessor. The digitised information at the output of the ADC is processed by the microprocessor to compute the blood oxygen saturation from the ratio (R) derived from the signal at the red wavelength compared with the signal at the infrared wavelength, as described in the previous sections.
A synopsis of some of the most important parts and functions of the microprocessor are highlighted below. The memory on board the microprocessor consists of latches, buffers, decoders, RAMs, ROMs and EPROMs. Latches are needed to demultiplex the address and data bus information. EPROMs are erasable and reprogrammable memory devices that store information such as the calibration curves, compensation requirements, etc., which may need occasional change. The set of instructions to be executed by the pulse oximeter and all the algorithms responsible for the calculation of SpO\textsubscript{2} and heart rate is stored in the ROMs. Finally decoders are used to decode the address and data information to generate the required control signals which control the multiplexers and demultiplexers used in the system. The microprocessor is also used to adjust and control the gain of the PPG signals by providing programmable gain amplifiers (MENDELSON, 1992; WEBSTER, 1997).

**Displays and Status Messages**
Apart from displays of blood oxygen saturation, heart rate, and photoplethysmograph, most pulse oximeters also display messages such as:

“no probe connected"
“battery low”
“noisy signal”
“probe off patient”
“searching for pulse”
“insufficient signal”

Flashing logos on liquid crystal screens convey similar information but may be more difficult to interpret since a single logo may be used for multiple problems. The least desirable systems indicate “pulse search”, for just about any abnormal saturation, which is a clear disadvantage (ALEXANDER et al., 1989).
Alarms
All pulse oximeters must have alarms for low oxygen saturation and for low and high heart rate. These alarms have to be both in audio and visual form. Comparators, power amplifiers, drivers and speakers constitute the audio alarm section. Liquid Crystal Display (LCD) bar graphs and blinkers are used for the visual section. The alarm limit settings of pulse oximeters should be set to default values when the device is switched on. Common default values are $\text{SpO}_2$ less than 95% and heart rate between 60 to 100 beats/minute. These levels can be adjusted by the user (MOYLE, 1994).

Power Supply
The pulse oximeter obviously needs a power supply. Any consideration of the source of power must include whether or not it will be necessary to use the device away from mains electricity, and if so, for how long. The internal power requirements of the pulse oximeter may be considered in three sections: the light emitting diodes; the electronics; and the display.

The power supplies present in most pulse oximetry boards are switched mode power supplies (SMPS) with an optically isolated power transformer, isolating the mains input from the rest of the pulse oximetry system, therefore preventing any form of electric shock to the patient (WEBSTER, 1997). These power supplies are capable of generating low voltages at high currents. The more portable pulse oximeters contain either sealed lead acid secondary cell batteries or nickel-cadmium batteries, which will allow the pulse oximeter to be used for a minimum of twelve hours on battery operation (depending on the manufacturers). Battery chargers are provided and should be internal rather than a separate item.
CHAPTER 6

Pulse Oximeter Probes

The probe of the original pulse oximeter as described by YOSHIYA et al., (1980) was based on a bulky fibre optic cable (Figure 4.6). The fibre optic cable in this implementation was used only as a guide to conduct light from a quartz halogen lamp to the remote measurement side and to conduct the light transmitted through the tissue back to the photodetector. The light source and the photodetector were both housed inside the oximeter. Narrow-bandpass interference optical filters were used in combination with a mechanical chopper to select properly the red and infrared wavelengths (MENDELSON, 1992).

An improved design of a noninvasive pulse oximeter sensor was introduced in the United States in the early 1980s. This much simpler and inexpensive design, which dominates most of the commercial pulse oximetry probes nowadays, consists of a pair of small red and infrared emitters and a single highly sensitive silicon photodetector mounted inside either a reusable spring loaded clip, or a disposable adhesive wrap (MENDELSON, 1992).

A flexible cable connecting the probe and the pulse oximeter unit carries electric power to the emitters and the signal from the photodetector. The cable may also contain conductors for a temperature sensor, to detect the temperature of the probe and the underlying skin.

6.1 OPTICAL COMPONENTS USED IN PULSE OXIMETRY PROBES

6.1.1 Emitters
Light emitted by light emitting diodes in a pulse oximetry probe is partially reflected, transmitted, absorbed, and scattered by the skin, the blood and other tissues before it reaches the detector.
The choice of light emitting diodes for pulse oximetry is limited to the wavelengths and intensities available in the market. Their small size, excellent drive characteristics, and large light output over a very narrow bandwidth make them the ideal choice for the source of light at both the red and infrared wavelengths used for pulse oximetry. Early in the development of pulse oximetry probes only a small combination of wavelengths were available, but the range has increased enormously in the recent years. Emitters are available over an approximately 700 nm range of wavelengths, from blue in the visible spectrum into near infrared. The common emitter wavelengths of 660 nm and 940 nm are suitable for pulse oximetry, and are available on the market at relatively low cost. The wavelength of the red emitter(s) is typically chosen to be in regions of the spectra where the absorption coefficient of Hb and HbO\textsubscript{2} are markedly different (e.g., 660 nm)(Figure 6.1). The infrared wavelength is typically chosen to be in the spectral region where the difference in absorption coefficients of Hb and HbO\textsubscript{2} is relatively small (e.g., 940 nm). The spectral response of the photodetector must overlap the emission spectra of the red and infrared emitters (MENDELSON and OCHS, 1988).

![Figure 6.1: Absorption spectra of oxygenated and deoxygenated haemoglobin, showing the two most commonly used wavelengths for pulse oximetry (MOYLE, 1994)](image)
Some of the predominant factors relating to the performance of the emitters used for pulse oximetry are the radiated power (or light output) and the size of the emitters. The radiated power of an emitter is measured in milliwatts. The typical radiated power of both the red and infrared emitters used in pulse oximetry is 1 mW at 20 mA dc (WEBSTER, 1997). Modern manufacturing techniques have minimised emitters to sizes smaller than a millimetre in length or diameter.

As there is a great variability in the thickness and pigmentation of the tissue area to which the pulse oximetry probe is applied, such as the finger, the energy output of the emitters must be variable so that the semiconductor detector used is not overloaded. This is achieved by varying the current through the emitters when they are switched on. The diode emitters used for pulse oximetry do have some disadvantages. One of the problems in the design of pulse oximeters is the variation of the centre wavelength (or peak wavelength) of different emitters of the same type. In fact, almost no two examples are exactly the same (POLOGE, 1987), and there may be a variation of up to 15 nm in the centre wavelength (Figure 6.2).

![Figure 6.2: Centre wavelength variation of emitters of the same type from the same lot, 660 ± 15 nm (POLOGE, 1987)]
From the extinction curves shown in Figure 6.1, it is evident that any shift in wavelength of the light source leaves the oximeter working with a different set of extinction coefficients. This change in extinction appears in Equation 5.4 and results in an erroneous calibration curve and therefore an error in the measured oxygen saturation. The change in extinction is most pronounced at the 660 nm wavelength on the deoxyhaemoglobin curve due to the extremely steep slope of the extinction curve in this region of the spectra (see Figure 6.1). The lower the oxygen saturation, the more heavily the deoxyhaemoglobin is weighted and the greater the error in the saturation that is measured with a shifted wavelength red LED. The precise wavelength used in the infrared region is of less concern. The extinction curves in this area of the spectra are flatter, and any shift in LED centre wavelength causes a smaller error than would be generated by the same shift in the red LED.

In order to solve this problem, pulse oximeters can compensate for a number of different emitter peak wavelengths. The first step in solving the problem is by testing each individual emitter with a spectrophotometer to experimentally determine the wavelength of light at which the emitter has its highest power output. The emitters are then separated into a certain number of groups, with each group having a small distinct range of wavelengths. Knowing the peak wavelengths for a particular emitter pair allows the proper set of calibration curves, specific to that wavelength combination, to be chosen from the entire family of curves that exist. The pulse oximeter interrogates each new probe to find out which calibration curve must be used to accurately estimate arterial blood oxygen saturation. The most common technique is to include in the probe connector a coding resistor with a specific value. Each unique resistor value represents to the pulse oximeter those pairings of emitter wavelengths that correspond to one calibration curve. The microprocessor inside the pulse oximeter simply sends a current through the resistor and measures the voltage drop across it, in effect finding the value of the coding resistor. By finding this voltage value in a lookup table, the microprocessor can indirectly determine the proper calibration curve to be used for that probe (WEBSTER, 1997).
6.1.2 Photodetectors

The photodetector is a fundamental device found in the pulse oximetry probe. The photodetector detects the energy back scattered by tissue from both infrared and red emitters and gives an output current proportional to the detected radiation level. This current is then converted to a voltage which is passed on to the pulse oximetry processing unit. The selection of a photodetector in pulse oximetry depends on factors such as performance, size and cost. The photodetector usually used with pulse oximeter probes is a silicon photodiode, but devices such as photocells, phototransistors, and integrated circuit (IC) sensors can be used as well (MENDELSON and OCHS, 1988; WUKITSCH et al., 1988; WEBSTER, 1997).

Photodiode Characteristics

The photodiode is the photodetector of choice for use in pulse oximetry probes because of their relatively low cost and linear output current response to incident light. When selecting a photodiode for use in pulse oximetry, parameters such as sensitivity, spectral response and packaging should be considered.

Sensitivity: The sensitivity is normally expressed as the output current level for a known incident light level at a specified temperature.

Spectral Response: Photodiode manufacturers normally specify the spectral response by providing the wavelength of peak sensitivity. The designer of pulse oximetry probes should keep in mind the wavelengths of interest in pulse oximetry (660 nm to 940 nm) when deciding on an appropriate photodiode. The spectral response of the selected photodetector must overlap the emission spectra of the red and infrared emitters (MENDELSON and OCHS, 1988).

Packaging: Photodiodes are available in three types of packages, the can package, the ceramic stem package, and the resin mold package. In the can package the photodiode chip is mounted on the metallic stem and is sealed with a cap that has a window to allow incident light to reach the
semiconductor surface (Figure 6.3a). In the ceramic stem package, the photodiode chip is mounted on a ceramic stem and is coated with resin (Figure 6.3b). In the resin mould package the photodiode chip is mounted on a lead frame and moulded with resin (Figure 6.3c). This is the most common package type used in pulse oximetry probes today.

**Figure 6.3:** Typical photodiode packaging (WEBSTER, 1997)

6.2 TRANSMITTANCE PULSE OXIMETRY PROBES

6.2.1 Principle

The majority of the commercially available pulse oximeters use transmittance sensors in which the pulsating arterial bed is positioned between the emitters and the photodetector, and the light is transmitted through the arterial bed to
measure the arterial oxygen saturation of the blood (MOYLE, 1994). Figure 6.4 shows a general transmission probe.

![Figure 6.4: Transmission Pulse Oximetry Probe (MOYLE, 1994)](image)

The probe comprises two emitters, with emission wavelengths at 660 nm in the red range and at 940 nm in the near infrared range. The emitters are switched on alternately so that light of one particular wavelength will pass through the tissue, and the transmitted light will be detected by the photodiode. The intensity of the light emerging from the tissue is attenuated by the blood present in the tissue. This varies with the arterial pulse and is used as a measure to indicate the pulse rate. The absorption coefficient of oxyhaemoglobin is different from that of deoxyhaemoglobin at the two monitoring wavelengths. Thus, differences in the amount of light absorbed by the arterial blood at the two wavelengths can be used to estimate arterial oxygen saturation, as described in chapter 5.

### 6.2.2 Sensor Placement

In transmission pulse oximetry probes the emitters are mounted opposite the photodetector, as the detector has to detect the maximum amount of light transmitted through tissue. The photodiode should be placed as close as possible to the skin without exerting force on the tissue. If the force applied by the probe on the tissue is significant, the blood under the tissue, where the probe is placed, may clot due to the external force applied. Normally transmission probes are placed on the patient’s finger, ear lobe or toe.
6.3 REFLECTANCE PULSE OXIMETRY PROBES

Recently, several companies have introduced reflectance pulse oximeter sensors, where the emitters and the photodetector are mounted side by side facing the skin (MENDELSON, 1992). Figure 6.5 shows a general reflection probe. Pulse oximeters with reflectance probes are used to monitor oxygen saturation based on the intensity of reflected light. The idea of using light reflection instead of transmission was first described by Brinkman and Zijstra (1949), where they showed that blood oxygen saturation can be monitored by measuring the amount of light reflected (back scattered) from the tissue (WEBSTER, 1997).

![Figure 6.5: Reflection Pulse Oximetry Probe (MOYLE, 1994)](image)

6.3.1 Principle

Arterial blood oxygen saturation by reflectance pulse oximetry is measured by analysing the pulsatile photoplethysmographs, which make use of reflected light intensities. The light from the emitters in the reflectance sensor is diffused by the skin in all directions. This suggests that to detect most of the backscattered radiation from the skin, the photodetector must be able to detect light from an area concentric with the emitters. The output of the photodetector is processed by the pulse oximeter, and estimates the oxygen saturation of the pulsatile blood.

6.3.2 Sensor Placement

In a reflection pulse oximetry probe the emitters and the photodetector are placed adjacent to each other (Figure 6.5). Reflection pulse oximetry probes can be used to measure arterial oxygen saturation at virtually any place on
the human body. The most common surfaces for monitoring oxygen saturation with a reflectance pulse oximetry probe are the forehead and temple.

6.3.3 Distance Between Detector and Emitters

One of the considerations when designing a reflection pulse oximetry probe is the distance between the detector and the emitters. This distance should be such that photoplethysmographs with both maximum and minimum pulsatile components can be detected. There are two techniques that can enhance the quality of the photoplethysmographs. One way is to use large emitter driving current, which determines the effective penetration depth of the incident light, which increases light intensity. For a given emitter/detector separation, using higher levels of incident light, a larger pulsatile vascular bed can be illuminated. As a result the reflected photoplethysmographs will contain a larger AC component. In practice, the emitter driving current is limited by the manufacturer to a specified maximum power dissipation. The other way is to place the detector close to the emitters. If the detector is placed too close to the emitters, the detector will be saturated as a result of the large DC component obtained by the multiple scattering of the light due to the non-blood components (skin, tissue etc).

A study by MENDELSON and OCHS (1988) revealed that for a constant emitter intensity the intensity of the backscattered light decreases in direct proportion to the square of the distance between the photodetector and the emitters; thus the photodetector should be mounted close to the emitters. Thus the selection of a particular separation distance involves a trade-off. A larger photoplethysmographic signal can be achieved by placing the detector further apart from the emitters but a higher emitter driving current is needed to overcome absorption due to increased optical path length. A separation of 4-5 mm between the emitters and the photodetector suggested by MENDELSON and OCHS (1988) provides the best sensitivity in terms of detecting adequate large pulsatile components.
6.4 REFLECTANCE VERSUS TRANSMITTANCE PROBES

The major disadvantage of the transmittance probes is that the sensor application is limited to peripheral sites of the body such as the finger, toes, ear lobe and nose in the adults or on the foot or palms in infants where transmitted light can be readily detected. Reflectance probes can be placed on virtually any part of the body where light reflection due to tissue can be expected (MENDELSON and OCHS, 1988).

The basic advantage of transmittance probes over reflectance probes is the larger intensity of light usually detected by the photodetector. Practically the major limitation in reflection pulse oximetry is the comparatively low-level photoplethysmographs typically recorded from low-density vascular areas of the skin. The feasibility of reflection pulse oximetry is essentially dependent on the ability to design a sensor that can detect adequate photoplethysmographic signals from various locations of the body (MENDELSON and OCHS, 1988).

6.5 MRI PROBES

When a pulse oximeter with either transmission or reflection probes is used during magnetic resonance imaging (MRI), the very high magnetic field strengths involved with this imaging modality may give erroneous readings or no readings at all (WAHR et al., 1995). The problem is due to voltages being induced in any conductor being moved in the “permanent” MRI magnetic field and to the radio frequency magnetic pulses generated during MRI. To overcome the problem manufacturers of pulse oximeters have produced pulse oximeters especially designed for use with MRI scanners. All of the electronic components, including the emitters and the photodetector, are in the housing of the main unit. The light energy is directed to and from the patient by optical fibres. The complete pulse oximeter is kept beyond the influence of the magnetic field (approximately 3 m) from MRI and the light from the emitters is transmitted through optical fibres and the transmitted/reflected light is brought back through optical fibres to the
photodetector (WEBSTER, 1997). Figure 6.6 shows a typical clip type MRI probe using optical fibres.

![Image of an MRI pulse oximetry probe](image)

**Figure 6.6: MRI pulse oximetry probe (WEBSTER, 1997)**

### 6.6 FOETAL PROBES

Several designs of foetal pulse oximetry probes have been developed over the last few years. The design and the application of one of these probes is described below.

Figure 6.7 shows a reflectance pulse oximeter probe used for foetal monitoring (WEBSTER, 1997). Before the application of the probe in the cervix an abdominal examination is performed to determine the position of the foetus and the state of the cervix. A cable, which is stiffer near the probe, is used to guide the probe into the correct position. The probe must be placed beyond the presenting part and the transcervical region (just beyond the cervix). Figure 6.8 shows that the cable bends around the head of the foetus and conforms to the curve of the mother’s pelvis. The cable contains calibration grooves and markings to aid the physician in placement. The probe is placed on the temple of the foetus and therefore has less interference from hair. The system electronics have increased sensitivity to small signals to accommodate low amplitude foetal pulses (WEBSTER, 1997).
6.7 REUSABLE PROBES

Probes, which can be, used more than once in monitoring oxygen saturation are called reusable probes. Figure 6.9 shows the most common reusable probe, which is the clip or clamp type sensor. Generally all probes with nonadhesive or disposable adhesive sensors are reusable probes. Figure 6.10 shows a reusable sensor with disposable adhesive wrap and Figure
6.11 shows a reusable reflectance sensor applied over the forehead with a disposable adhesive pad. The main advantage of reusable pulse oximetry probes is the low use cost involved. However, reusable probes require cleaning between patients to minimise the risk of cross contamination (KELLEHER, 1989).

**Figure 6.9:** Clip type reusable pulse oximetry probe (WEBSTER, 1997)

**Figure 6.10:** Reusable pulse oximetry probe (WEBSTER, 1997)

**Figure 6.11:** Reusable reflectance pulse oximetry probe (WEBSTER, 1997)
6.8 DISPOSABLE PROBES

In the past few years many pulse oximeter manufacturers have produced disposable probes. As the name indicates disposable probes are discarded after they have been used. One of the advantages of disposable probes is the elimination of any form of cross contamination between patients since disposable probes are used on a single patient. Disposable probes are all adhesive probes and because of that they decrease the effect of movement artefact as they secure the sensor in the proper position and the relative motion between the patient and the probes is minimal. Disposable adhesive probes are used for both short term and long term monitoring. Figure 6.12 shows a typical disposable pulse oximetry probe.

![Disposable pulse oximetry probe](WEBSTER, 1997)

Figure 6.12: Disposable pulse oximetry probe (WEBSTER, 1997)
CHAPTER 7

Applications of Pulse Oximetry

Pulse oximeters are non-invasive, easy to use, and readily available. Pulse oximetry provides information about the blood oxygen saturation, heart rate, and pulse amplitude. Due to these characteristics, it has an abundance of clinical applications. This may seem as an overstatement, but as more and more pulse oximeters are coming in use, more and more hypoxaemic events are being seen as precursors of pathological events. Some of the main areas in which it is used are anaesthesia, patient transport, childbirth, neonatal and paediatric care, dentistry and oral surgery, sleep studies and many other applications.

The use of pulse oximetry has been described in nearly every setting in the hospital, outpatient and veterinary clinics, and even homes. KELLEHER reviewed 220 references in a review published in 1989. SEVERINGHAUS' follow-up review in 1992 found more than 500 new reports between 1989 and October 1991 (KELLEHER, 1989; SEVERINGHAUS and KELLEHER, 1992). Nearly 1346 further reports have been published since October 1991. This chapter gives an overview of the applications of pulse oximetry.

7.1 ANAESTHESIA

The pulse oximeter is most often and most importantly used when anaesthesia is given. The oximeter is considered a relatively safety monitor in the operating theatre and anaesthetic (induction) room as it ensures that the anaesthetist is aware of any decrease in oxygenation. Haemoxaemia has been shown to be much more common in the anaesthetic room, operating theatre and recovery areas than originally assumed (MOYLE, 1994).
Several studies have shown that desaturation is often a problem during induction of anaesthesia. This is the most critical time of general anaesthetic because of the side effects of the induction agents, onset of neuromuscular blockade, loss of protective reflexes, mechanical problems with the airway, respiratory depression, and problems with tracheal intubation. The mean oxygen saturation in patients studied by DRUMMOND and PARK (1984) fell from 96% before induction to 85% one minute after induction.

A study by MOLLER et al. (1992) observed 296 anaesthetised patients chosen randomly. Two anaesthetists were present at each procedure, one giving the anaesthetic but unable to observe the pulse oximeter and the other observing the pulse oximeter (MOLLER et al., 1992). One or more episodes of mild hypoxaemia (SpO$_2$ 86-90%) occurred in 53% of the patients. Episodes of severe hypoxaemia (SpO$_2$ < 81%) were recorded in 20% of patients. The mild hypoxaemic episodes lasted up to 34.6 minutes (mean 2.3 minutes) and 70% of these episodes were not detected by the anaesthetist giving the anaesthetic. In the remaining 30% the anaesthetist detected hypoxaemia with a mean delay of 70 seconds, and after a mean period of 57 seconds the SpO$_2$ was indicated as being > 90%. The authors conclude that hypoxaemic episodes are common and that pre-oxygenation and supplemental oxygen are required in all patients until full arousal postoperatively.

Pulse oximetry became an international standard for monitoring during anaesthesia in the early 1990’s following the publication of the Harvard minimum standards for monitoring of 1986. It is a vital monitoring technique in anaesthesia and is necessary for the continuous monitoring of blood oxygen saturation in patients under anaesthesia and the determination of any desaturation events.
7.2 EMERGENCY MEDICINE

Anaesthetists were quick to see the enormous benefit to the safety of the patient of using pulse oximeters, and doctors, nurses, and medical technicians involved in emergency care soon recognised the advantage of pulse oximetry. A retrospective study by ANDERSON et al. (1988) showed that failure to recognise hypoxaemia has been identified as one of the major avoidable causes of death in trauma patients. Pulse oximeters are now regularly used in accident and emergency departments and also in the pre-hospital care of the sick and injured.

7.2.1 Pre-hospital Care: Monitoring on the road and in the air.

Pulse oximeters can be very beneficial for monitoring in ambulances and aircraft since they are non-invasive, easy to use, and portable. Both ambulances and helicopters are used for patient transport, during which vital signs need to be monitored (WEBSTER, 1997). Pulse oximeters to be used in ambulances or helicopters should be lightweight and portable so the paramedics team can apply the monitor as soon as they reach the patient. During transport, in the case of an ambulance, the bouncing of the vehicle can cause the pulse oximetry probe to be displaced and temporarily lose the signal. It is important for probes used in transport to fit properly and snugly on the patient. In helicopters, since flights may take longer than ambulance rides, battery life is an important consideration. Also, in helicopters noise interferes a great deal with the ability to hear alarms. Therefore it is crucial that displays be readable. Visual indications of problems such as a lighted display, which flashes when a patient’s saturation falls below a particular level, are required.

The importance of pre-hospital pulse oximetry is confirmed by a recent study discussed in WEBSTER (1997) in which 14 of 50 patients were found to be hypoxic (oxygen saturation < 90% for longer than one minute) en route to the hospital; only four had been clinically recognised as hypoxic.
7.2.2 Pulse Oximetry in the Accident and Emergency Department

The role of pulse oximetry in the accident and emergency department is varied and includes routine monitoring of acutely ill patients, monitoring oxygenation of patients undergoing minor procedures under some form of sedation.

Pulse oximetry should also be used during procedures without sedation in which the airway is “shared” or may be compromised. Tracheal suction and gastric lavage both cause transient hypoxaemia (LAMBERT and CRINNION, 1989). Pulse oximetry in the accident and emergency department saves time and allows arterial oxygenation to be assessed in the majority rather than the minority of patients.

7.3 POSTOPERATIVE RECOVERY

The period between the end of surgery and when the patient is fully conscious is often when hypoxaemia is most likely to go unnoticed. The greatest risk during this period is respiratory failure. For these reason the pulse oximeter must be used regularly in the postoperative recovery phase (MOYLE, 1994).

Blood oxygen saturation is also compromised in patients during transport to the recovery room. Monitoring during transport has been examined in children and adults. PULLERITS et al. (1987) monitored 71 healthy paediatric patients during transport and found that 28.1% had SpO₂ values ≤ 90%, while only 45% of these desaturated patients had observable cyanosis. In a similar study of adult patients, TYLER et al. (1985) found that 35% had SpO₂ values ≤ 90%, and 12% had SpO₂ falls to 85% or less. Both of these studies conclude that due to the high incidence of desaturation and the inability to clinically recognise it, all patients should receive supplemental oxygen and should be monitored continuously with a pulse oximeter during transport from the operation theatre to the recovery room (KELLEHER, 1989).
7.4 CHILDBIRTH

The process of labour and delivery is a stressful time for both the mother and the foetus. If the foetus does not have sufficient metabolic reserve to withstand this ordeal it is at risk of hypoxia and of sustaining subsequent brain damage. Pulse oximetry is used to monitor arterial oxygen saturation of both the mother and the foetus during childbirth. Due to the inaccessibility of the foetus special apparatus is needed for monitoring. Factors such as hypovolaemia, hypertension, anaemia, maternal position and anaesthesia can cause desaturation and hypoxaemia in a woman during labour (MINNICH et al., 1990). Amniotic fluid embolism can occur when amniotic fluid escapes into the mother's circulatory system. The embolism can cause the mother to develop a pulmonary shunt and thus cause arterial desaturation. Pulse oximetry monitoring during labour can help detect problems early (QUANCE, 1988).

Foetal monitoring can indicate foetal distress and hypoxia. Pulse oximetry monitoring is crucial in difficult births such as breech presentation and caesarean section. Foetal monitoring can also detect acidemia, which results when a foetus experiences an increase in hydrogen ion concentration. Pulse oximeters can detect this problem because increasing pH causes the oxygen dissociation curve to shift to the right, resulting in low saturation levels (DILDY et al., 1994). Many difficulties have been encountered when attempting to monitor foetal pulse oximetry. The obvious problem is that the foetus is not accessible. A device is needed to advance the probe into the uterus and position it properly on the foetus head (the most frequent monitoring site). The position of the probe may be altered during cervical dilation and that will make the monitoring very difficult, inaccurate or not possible. Also the foetal head is often covered with hair, amniotic fluid and maternal blood all of which hinder the ability to obtain a stable and accurate signal. All these challenging difficulties in monitoring oxygen saturation in the foetus using pulse oximetry made it a very interesting and demanding research area that exploded over the last few years. The potential application of pulse oximetry to foetal monitoring during labour has already been
demonstrated by several groups (GARDOSI et al., 1991; MENDELSON, 1992). Several sensors have been developed for the monitoring of the foetus and were presented in section 6.5.

7.5 NEONATAL AND PAEDIATRIC CARE

Low and high arterial oxygen levels can both be damaging to newborn infants. Infants who are hypoxic may develop organ damage, and hypoxia may also cause pulmonary hypertension (MOYLE, 1996). The great concern of neonatal paediatricians is to prevent retinopathy of prematurity, formerly known as retrolental fibroplasia (MOYLE, 1994). Retinopathy is caused by high levels of retinal oxygenation, which may lead to blindness in premature babies. It is therefore essential to monitor accurately the oxygen levels of sick and premature newborn infants.

Pulse oximetry is becoming a common monitoring technique for assessing arterial oxygen saturation in neonates. A foetus generally has an oxygen saturation of about 50%. Within the first 15 minutes after birth, it normally rises to 90% (WEBSTER, 1997). Continuous on-line indication of the oxygenation of the sick neonate is needed.

Paediatric patients undergoing repair of cardiac anomalies can also benefit from pulse oximetry monitoring. For example, when applying pulmonary bands to the pulmonary artery, there is an optimal saturation range to be obtained while the patient is breathing 100% oxygen. Arrival at this optimal level is facilitated by the pulse oximeter (BOWES, et al., 1989).

Other applications in the paediatric population include intensive care unit monitoring, for which it has been shown that pulse oximetry out-performs transcutaneous pO2 monitoring (MOYLE, 1994). Pulse oximetry in neonates and infants is as safe a technique as in adults, with less morbidity than transcutaneous oxygen monitoring, as long as the oxyhaemoglobin
dissociation curve is taken into account, the probe is positioned properly, and care is taken with the fixing of the probe (MOYLE, 1994).

7.6 DENTISTRY AND ORAL SURGERY
Many papers have recommended the use of pulse oximetry during general anaesthesia for dental procedures, pointing out that pulse oximeters should be applied to all patients even for short procedures. HOVAGIM et al. (1989) monitored 46 patients undergoing dental procedures with a pulse oximeter. Of these, 36 received some form of conscious sedation during the procedure, while 10 received only local anaesthesia. In the control group (36 patients) there were only five episodes of mild hypoxaemia (3-5% below baseline saturation), whereas in the sedated group there were 316 episodes, 151 of which were mild, 132 moderate (6-10% below baseline saturation), and 33 severe (> 10% below baseline saturation). The results of this study confirm the need of continuous monitoring of oxygen saturation by pulse oximetry in dental surgery (HOVAGIM et al., 1989).

7.7 SLEEP STUDIES AND EXERCISE
Many people become desaturated during sleep or heavy exercise. The cause of desaturation during sleep is due to a disorder known as sleep apnoea (WEBSTER, 1997). Desaturation can occur during heavy exercise due to poor ventilation or chronic obstructive pulmonary disease (COPD). The use of pulse oximetry during sleep and exercise aids in the diagnosis of these respiratory problems (WEBSTER, 1997).

7.8 FUTURE APPLICATIONS OF PULSE OXIMETRY
Although pulse oximetry seems to be at the peak of its development, there is always room for further improvement and optimisation. Many of these improvements relate to specific applications. Improvements, which will increase the performance of pulse oximetry during transport, are to lengthen
the battery life in portable units. Reducing the occurrence of false alarms would be beneficial in all applications, but especially during long term monitoring when staff cannot always be in the room.

The application of foetal pulse oximetry is exciting on-going research by many groups. In this unique application, sensor attachment to the foetal scalp must be improved. Other potential applications of pulse oximetry have been suggested for monitoring oxygen saturation via the retinal fundus in the eye (DE KOCK et al., 1993). This approach offers the potential advantage of monitoring cerebral oxygenation directly, as opposed to monitoring oxygen saturation from peripheral locations. The direct application of pulse oximetry to an organ such as the liver the kidney or the gut will be a very useful application in determining organ blood flow and perfusion, regardless if the patients SpO₂ as measured from an extremity (finger) is normal.
CHAPTER: 8

Limitations of Pulse Oximeters

Although generally reliable, pulse oximeters have been reported to fail and many of their physiological and technical limitations have been described (MOYLE, 1994). This chapter reviews the limitations of this clinical tool.

8.1 LIMITATIONS OF PULSE OXIMETERS DUE TO CALIBRATION ASSUMPTIONS

Initially the conversion from absorbancy ratios to arterial oxygen saturation as described in previous chapters was based directly on a Beer-Lambert calculation, but the effects of reflection and scattering of light even within the pulsatile fraction of arterial blood led to overestimation of oxygen saturation (KELLEHER, 1989). Better results have come from using experimental derived calibration curves (Figure 8.1). These curves are based on measurements in healthy young volunteers after induction of hypoxaemia (MOYLE, 1994; SINEX, 1999).

Figure 8.1: Relationship of red/infrared ratio to oxygen saturation, as typical pulse oximeter calibration curve (SINEX, 1999)
An unavoidable limitation is that pulse oximeters can only be as accurate as their empirical calibration curves. Understandably, researchers were limited in the degree of haemoxaemia inducible in these volunteers, to an $\text{SaO}_2$ of approximately 75% to 80%. Therefore the shape of the curve below these levels must be extrapolated, with obvious implications for the accuracy of pulse oximetry at low saturation levels. Early accuracy studies showed such great inaccuracy and bias at low oxygen saturation that manufacturers revised early calibration curves and software (SEVERINGHAUS and NAIFEH, 1987). More recent studies, however, continued to show significant bias, increasing as oxygen saturation decreases, although it has been justifiably pointed out that few, if any, clinical treatment decisions will hinge on whether the oxygen saturation is actually 50% or 60% (KELLEHER, 1989).

8.2 LIMITATIONS OF PULSE OXIMETERS DUE TO INTERFERENCE

Many substances in the blood can interfere optically with pulse oximetry. This interference generally takes the form of false absorbers, or components besides deoxyhaemoglobin or oxyhaemoglobin that will absorb light within the red and near-infrared wavelengths used in pulse oximetry.

8.2.1 Dyshaemoglobinaemias

The most significant potential false absorbers in the circulation are carboxyhaemoglobin ($\text{COHb}$) and methaemoglobin ($\text{MetHb}$) (KELLEHER, 1989). Being two-wavelength devices, pulse oximeters can only deal with two haemoglobin species (deoxyhaemoglobin and oxyhaemoglobin). As discussed in chapter 5 (section 5.2.2) this would be adequate to measure functional $\text{SaO}_2$ if $\text{COHb}$ and $\text{MetHb}$ did not absorb red or infrared light at the wavelengths used by pulse oximeters. Unfortunately, this is not the case, and both $\text{COHb}$ and $\text{MetHb}$ will cause errors in the pulse oximeter readings. Of the dyshaemoglobins, $\text{COHb}$ is the most dangerous as it is fairly common and makes the pulse oximeter to overestimate oxygen saturation values. Carboxyhaemoglobin is caused mainly by the inhalation of carbon monoxide.
Common sources of carbon monoxide are the combustion engines fumes, barbecues, tobacco smoke and inadequately ventilated combustion of coal or gas in heating systems. The absorption spectrum of COHb is shown in Figure 8.2. In the infrared range (940 nm), COHb absorbs very little light, whereas in the red range (660 nm), it absorbs as much light as oxyhaemoglobin. This is clinically illustrated by the fact that patients with carboxyhaemoglobinemia appear red (TREMPER and BARKER, 1989). Therefore, to the pulse oximeter, COHb looks like oxyhaemoglobin at 660 nm, while, at 940 nm COHb is relatively transparent. The effect of COHb on pulse oximeter values has been evaluated experimentally in dogs (BARKER and TREMPER, 1987). In this study, the pulse oximeter saturation (SpO₂) was found to be given approximately by:

\[
\text{SpO}_2 = \frac{\text{HbO}_2 + 0.9 \times \text{COHb}}{\text{total Hb}}
\]

Moyle (1996) also indicates that for every 1% of COHb circulating, the pulse oximeter over-reads by approximately 1%. Therefore, is considered negligent to use a pulse oximeter on patients who have been at risk of carbon monoxide inhalation.

Methaemoglobin (MetHb) is formed as the iron in haemoglobin is oxidised from the ferrous to the ferric state. MetHb levels may be high congenitally or as a result of exposure to a number of agents, most notably anaesthetics, sulfa drugs, and nitrites (EISENKRAFT, 1988). The effects of MetHb on pulse oximetry can also be predicted by the extinction curves (Figure 8.2). MetHb has nearly the same absorbance as deoxyhaemoglobin at 660 nm, while it has a greater absorbance than the other haemoglobins at 940 nm. This is consistent with the clinical observation that methaemoglobinemia produces very dark, brownish blood (TREMPER and BARKER, 1989). Thus, it would be expected to produce a large pulsatile absorbance signal at both wavelengths. The effect of MetHb on pulse oximeter readings has also been measured in dogs (BARKER et al., 1989). In this study they found that as MetHb levels increased, the pulse oximeter saturation tended toward 85%.
and eventually became almost independent of the actual \( \text{SaO}_2 \) (BARKER et al, 1987). In other words, in the presence of high levels of \( \text{MetHb} \), \( \text{SpO}_2 \) is erroneously low when \( \text{SaO}_2 \) is above 85%, and erroneously high when \( \text{SaO}_2 \) is below 85%. This is explained by the fact that \( \text{MetHb} \) causes a large pulsatile absorbance at both wavelengths used by pulse oximeters, thereby adding to both the numerator and denominator of the absorbance ratio (R) (Equation 5.12) and forcing this ratio towards unity. As shown in Figure 8.1, an absorbance ratio of one corresponds to a saturation of 85% on the calibration curve. Pulse oximetry error during methemoglobinaemia has also been confirmed in a clinical report (EISENKRAFT, 1988).

![Figure 8.2: Haemoglobin extinction curve, demonstrating relative absorptions of the four major classes of haemoglobin (SINEX, 1999)](image)

In summary, both COHb and MetHb can have significant effects on pulse oximetry readings. When the presence of these Hb species is suspected, pulse oximetry should be supplemented by \textit{in vitro} multiwavelength CO-Oximetry.
8.2.2 Bilirubin and Intravenous Dyes

Taylor and Whitwam (RALSTON et al., 1991) state that a raised bilirubin concentration causes underestimation of oxygen saturation by pulse oximetry, but give no reference to support this. Conversely three groups (see references in Ralston) could not show any effect on the accuracy of SpO\textsubscript{2} readings with bilirubin levels up to 84.3 mg/dil (normal range; 0.1 to 1.0 mg/dil). The absorption spectra of bilirubin has a broad peak at 460 nm and two much smaller peaks at 560 nm and 600 nm. Thus, bilirubin is unlikely to have any detectable effect on the absorption of the 660 nm and 940 nm wavelengths used by commercial pulse oximeters (VEYCKEMANS et al., 1989). Bilirubin can cause errors in both COH\textsubscript{b} and MetH\textsubscript{b} levels as measured by in vitro CO-Oximeter because of the overlap of its absorption spectra with these Hb species. Thus the presence of bilirubin in the arterial blood will not introduce any significant errors in pulse oximetry measurements, but may cause a discrepancy between pulse oximetry and CO-Oximeter readings.

Intravenous dyes are known to have potentially profound effects on pulse oximetry readings, resulting in falsely low measured oxygen saturations. Methylene blue, with very high absorbance at 660 nm, causes an impressive spurious decrease in SpO\textsubscript{2} (KESSLER et al., 1986; SCHELLER et al., 1986; SIDI et al., 1987). As little as 5 ml of methylene blue administered to human subjects can decrease SpO\textsubscript{2} by 1%. Lesser decreases from baseline SpO\textsubscript{2} occur after intravenous administration of indocyanine green and indigo carmine. Onset of these changes occurs 30 to 45 seconds after dye administration, with recovery to baseline oxygen saturation within 3 minutes. Dye clearance is prolonged in debilitated, very young, or elderly subjects. In animal studies, administration of fluorescein has been observed to have an effect on pulse oximetry (SIDI et al., 1987). False desaturation with these intravenous dyes is also observed with the in vitro CO-Oximeter.
8.3 LIMITATIONS OF PULSE OXIMETERS DUE TO SIGNAL ARTEFACT

Most commonly, problems in pulse oximetry arise from signal artefact. The presence of a sharp pulsatile waveform displayed on those oximeter models featuring a photoplethysmograph is no guarantee against signal artefact (MOYLE, 1996). Signal artefact results from false sources of signal or from a low signal-to-noise ratio. False signal can arise from detection of non-transmitted light (ambient sources or optical shunt) or from non-arterial sources of alternating signal. A low signal-to-noise ratio results from inadequate signal complicated by an excess of physiological or technical noise.

8.3.1 False Signal

The oximetry system as outlined in the previous chapters assumes that the sum of the light absorbed and the light transmitted is equal to the incident light, with no other light loss or gain affecting the detector. Ambient light, however, is potentially a major source of interference. Recognising this, designers of pulse oximeters divided the emitter and detector activities into three sensing periods, cycling at hundreds of times per second. Two of these periods use light emitted by the emitters at each of the two incident wavelengths. In the third period neither emitter is activated, and the photodiode detector measures only ambient light, the influence of which is subsequently eliminated from the emitter-illuminated sensing periods.

However, cases of ambient light interference still occur (HANOWELL et al., 1987; KELLEHER, 1989). Implicated sources include fluorescent lighting, surgical lamps, fiberoptic instruments, and sunlight. Covering the probe with an opaque shield offers a simple solution.

Optical shunting occurs when light from the emitters reaches the photodetector without passing through an arterial bed (Figure 8.3). This occurs most commonly with inappropriate probe selection, as when a digit
probe is placed on the ear lobe, or with probe misplacement (WEBSTER, 1997).

![Diagram of pulse oximetry](image)

**Figure 8.3:** Ambient light interference and optical shunt in pulse oximetry. Optical shunt occurs when light from the emitters reaches the photodiode without passing through arterial bed (WEBSTER, 1997)

In both ambient light interference and optical shunt, the end result is the addition of false signal to both wavelengths emitted (KELLEHER and RUFF, 1989). As with methaemoglobinaemia, the result will be a ratio of relative absorbances approaching unity, with a corresponding displayed of SpO₂ of 85%. At normal saturations this will manifest as a falsely lowered saturation. More dangerously, errors caused by ambient light interference and optical shunt will lead to overestimation of saturation in patients with a true SaO₂ of less than 85%. This overestimation has been demonstrated with probe mulposition in hypoxic patients (BARKER et al., 1993).

Non-arterial sources of alternating signal most commonly result from motion artefact. Motion artefact, such as during shivering, seizure activity or exercise, are usually recognised by false or erratic heart rate displays or by distorted photoplethysmographic waveforms (Figure 8.4) (SEVERINGHAUS and KELLEHER, 1992; HANNING and ALEXANDER-WILLIAMS, 1995).
Figure 8.4: The photoplethysmographic signal of a subject at rest is periodic (a) and during exercise is not periodic (b) (WEBSTER, 1997)

Also, false readings caused by patient motion during cardiopulmonary resuscitation have been described. Repetitive cough and the cycling of mechanical ventilators offer other common potential sources of interference. When interference does occur from motion artefact it again tends to be additive into both the red and infrared wavelength channels (POLOGE, 1987), with a resultant absorbance ratio (R) becoming nearer to one and an $\text{SpO}_2$ approaching 85%.

Although different oximeter models employ different data processing, the saturation values acquired per second are typically averaged over a period of 2 to 16 seconds before a reading is given, serving in part to limit the impact of motion artefact (KIDD and VICKERS, 1989). However, this averaging period will also delay presentation of data, extending time to first reading as well as response time to saturation changes. The trade-off between limiting artefact and extending response time can be adjusted in many oximeters by variable averaging period settings. Oximeters also attempt to minimise motion artefact with internal algorithms that eliminate readings, which are likely due to motion (TREMPER and BARKER, 1989).

In addition to extracorporeal motion, there are other sources of false alternating signal. Arterial pulsations can be transmitted into the venous circulation in venous congestion causing artificially lowered oxygen saturation
readings (SEVERINGHAUS and SPELLMAN, 1990). This same phenomenon occurs in tricuspid regurgitation. Arterial pulsations with a large dicrotic notch, as in aortic regurgitation, combined aortic stenosis and regurgitation can be processed incorrectly at twice the true heart rate. This will have little impact on oxygen saturation but has obvious effects on displayed heart rate (SEVERINGHAUS and KELLEHER, 1992).

8.4 OTHER PHYSIOLOGICAL AND TECHNICAL LIMITATIONS OF PULSE OXIMETERS

8.4.1 Anaemia
Anaemia appears to adversely affect the accuracy of pulse oximetry, although the mechanism is unclear, and it may do so only in the presence of hypoxia. In theory, anaemia should not affect pulse oximetry at all, as the ratio of relative absorbances should be preserved and unchanged by changes in total haemoglobin concentration within the sample. However, a retrospective study found an underestimation of oxygen saturation by pulse oximetry in anaemic subjects, inversely proportional to Hb concentration and most pronounced at an SpO\textsubscript{2} less than roughly 80% (SEVERINGHAUS and KOH, 1990). A recent study of non-hypoxic human patients with acute anaemia from haemorrhage down to a Hb value as low as 2.3 g/dl (normal range:12 to 14 g/dl) showed good accuracy with pulse oximetry (JAY et al., 1994).

8.4.2 Skin pigmentation and nail polish
Skin pigmentation has shown variable effects on pulse oximetry (VOLGYESI and SPAHR-SCHOPFER, 1991). Primarily, dark pigmentation appears to make adequate light penetration more difficult, with significantly more signal detection failures. It has been suggested that the less pigmented nail beds may offer particularly good probe sided in these patients (SEVERINGHAUS and KELLEHER, 1992). More concerning, are some studies of pigmentation effects which have shown overestimation of oxygen saturation, both in models (VOLGYESI and SPAHR-SCHOPFER, 1991) and in human subjects (RIES et al., 1989). The mechanism of such an artefact if present, is unclear,
and could arise from low signal-to-noise effects, or as a reflection of the relative lack of darkly pigmented subjects in calibration trials.

Similarly, there is evidence for the effects of nail polish on pulse oximetry, but little consensus on occurrence or degree (KATARIA and LAMPKINS, 1986; COTE et al., 1988; RUBIN, 1988). When demonstrated, most interference appears to arise from blue or black polish. It is probably advisable to avoid possible interference by using other probe sites, removing the nail polish, or even simply placing the probe sideways on the digit to remove the nail from the transmission path.

8.4.3 Light emitting diode centre wavelength variability
Although LEDs are reliable sources of narrow wavelength light in some cases, there can be a slight variability between LEDs in actual wavelength emitted (TREMPER and BARKER, 1989). A very small error in wavelength can translate into an error in the estimation of SpO$_2$ by the pulse oximeter, particularly at the steep portions of the curve in the red range at 660 nm as discussed in section 6.1.1. In general, the manufacturer rejects LEDs that emit too far from the nominal wavelength. However, some manufacturers have dealt with this problem in the past by compensatory changes in software and data processing (WEBSTER, 1997) (see section 6.1.1).

8.4.4 Effects of temperature
An exposure of the body to low temperatures can cause changes in peripheral perfusion, which may cause inaccuracy. The temperature dependence of LEDs in pulse oximetry probes is unlikely to affect the pulse oximetry values. REYNOLDS et al. (1992) showed that there was a 5.5 nm increase in peak wavelength for a 660 nm LED, and a 7.8 nm increase in peak wavelength for a 950 nm LED as temperature increased from 0$^\circ$ to 50$^\circ$C. REYNOLDS et al. (1992) calculated the extinction coefficients of Hb and HbO$_2$ at different wavelengths and temperatures. Then, they substitute these values into the relationship between SpO$_2$ and ratio (R) given in the equation which was derived from Beer’s Law (Equation 5.11). They showed (see Figure 8.5) that the effect of shifts in wavelength of the LEDs on pulse
oximetry accuracy is negligible as the temperature increases from 0°C to 50°C (REYNOLDS et al., 1992).

Figure 8.5: Calibration curves derived from the Beer’s Law when substituting, the extinction coefficient of Hb and HbO₂ calculated at different wavelengths (660 nm and 950 nm) and temperatures [0°C (solid line) to 50°C (dashed line)], into the relationship between SpO₂ and ratio (R) (WEBSTER, 1997)

The AC PPG amplitudes are reduced during cold exposure, which might cause the pulse oximeter to be more sensitive to motion artefacts, for example those caused by shivering. These artefacts may cause the pulse oximeter to give erroneous values of SpO₂. The conclusion of Reynold’s temperature studies is that inaccuracies in pulse oximeter readings at extreme temperatures are far more likely to be caused by reductions in peripheral perfusion, rather than the result of temperature dependence of the LEDs in the pulse oximeter probe.

8.4.5 Electromagnetic interference
Electromagnetic interference (EMI) can affect the accuracy of pulse oximeters and other medical devices. It may be generated by many sources, mostly man made but it may also result from atmospheric events.

8.4.6 Interference due to Electrocautery
Electrocautery can interfere with pulse oximetry by artifactually decreasing SpO₂ readings, or by setting off false alarms (WAHR et al., 1995). The cause
is the wide spectrum of radio frequency emissions picked up directly by the photodetector in the pulse oximeter probe (BLOCK and DETKO, 1986). Careful placement of the electrocautery return plate and pulse oximetry probe may alleviate this problem (KELLEHER, 1989). Some recently released pulse oximeter models report detected electromagnetic interference. A new signal extraction technology (MASIMO) claims to be more accurate in the presence of electrocautery, but has not been fully tested (WAHR et al., 1995).

8.4.7 Interference due to Magnetic Resonance Imaging
Interference from magnetic resonance imaging (MRI) is also a source of pulse oximetry inaccuracy. MRI sometimes interferes with pulse oximetry, apparently because of radio frequency (RF) signal detection; an abrupt change in SpO2 as imaging begins may indicate such interference (WEBSTER, 1997). Grounding and filtering of the radio frequency have been shown to reduce or eliminate these problems, but this may require special probe extender/couplers, with grounding of the oximeter cable shield to the nuclear magnetic resonance magnet case (WEBSTER, 1997). Pulse oximeters adapted to the MRI environment have been recently introduced on the market offering compact non-metallic housing and are battery operated. Fibre optic lead probes are also used to keep the electronics outside the bore of the MRI magnet (SHELLOCK and SLIMP, 1989). Because there are no electric cables extending through the magnetic resonance imager bore, there is no possibility of RF burns to the patient or RF induced noise in the signal conveyed to the processor of the pulse oximeter (WAHR et al., 1995). However, the fibre optic lead probes are relatively delicate and easily broken. They are also very expensive to repair and purchase.

8.5 LIMITATIONS OF PULSE OXIMETERS DUE TO INADEQUATE PULSATILE PERFUSION
Apart from the physiological and technical limitations of pulse oximeters described in this chapter, they have been also reported to fail in patients with compromised peripheral perfusion (FREUND et al., 1991; MOLLER et al.,
Pulse oximetry is a pulse dependent technique, and any significant reduction in the amplitude of the pulsatile component of the photoplethysmographic signal can lead to dubious values for blood oxygen saturation ($\text{SpO}_2$) or complete failure. Hence, pulse oximeters require adequate peripheral perfusion to operate accurately. When peripheral perfusion is poor, as in states of hypovolaemia, hypothermia and vasoconstriction, oxygenation readings become unreliable or cease (PALVE and VUORI 1989, PALVE 1992a, PALVE 1992b). Such clinical situations occur, for example, after prolonged operations, especially hypothermic cardiopulmonary bypass surgery. The problem arises because conventional pulse oximetry sensors must be attached to the most peripheral parts of the body, such as finger, ear or toe, where pulsatile flow is most easily compromised. Measurements at sites other than the finger or ear, such as the forehead and nose, give no improvement in poorly perfused patients (ROSENBERG and PEDERSEN, 1990; CLAYTON et al., 1991). Thus, $\text{SpO}_2$ readings are often unobtainable at just the time when they would be most valuable.

A study of pulse oximetry failure at four University of Washington Hospitals reported that the incidence of failures greater than 30 minutes in duration ranged from 0.56% to 4.24% (mean 1.12%) of 11046 surgical procedures (FREUND et al., 1991). The risk of failure in this study was increased in association with ASA (American Standard of Anesthesiology; scale 1 to 5, with 5 the most critically ill patient) physical status 3, 4, and 5, advanced age (50 to 60 years old), and prolonged surgery. A large prospective study from Denmark on 20802 patients reported a mean pulse oximetry failure rate of 2.5% and as high as 7.2% in ASA physical status 4 patients (MOLLER et al., 1993).

A more recent study by REICH et al. (1996) reviewed case files of 9203 computerised anaesthesia records. Pulse oximetry failure was defined as the presence of at least one continuous gap in pulse oximetry data of at least 10 minutes duration. Ten minutes duration was chosen arbitrarily as the criterion for a clinically significant pulse oximetry data failure based on the likelihood
that ten minutes of continuous hypoxaemia would result in an adverse outcome. The proportion of cases with at least one continuous pulse oximetry data failure of at least ten minutes was 9.18%. The mean duration of all failures was 17 minutes, and the mean duration of anaesthesia was 213 minutes. Table 8.1 shows the distribution of pulse oximetry data failures by type of surgery.

**Table 8.1: Pulse Oximetry Data Failure ≥ 10 minutes by Surgery Classification (REICH et al., 1996)**

<table>
<thead>
<tr>
<th>Classification of Surgery</th>
<th>% of Cases with at Least one Failure (total)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Abdominal</td>
<td>6 (2544)</td>
</tr>
<tr>
<td>Cardiac with cardiopulmonary bypass</td>
<td>31 (775)</td>
</tr>
<tr>
<td>Cardiac without cardiopulmonary bypass</td>
<td>7 (268)</td>
</tr>
<tr>
<td>Genitourinary</td>
<td>8 (541)</td>
</tr>
<tr>
<td>Gynecologic</td>
<td>4 (537)</td>
</tr>
<tr>
<td>Head/neck</td>
<td>4 (829)</td>
</tr>
<tr>
<td>Intrathoracic</td>
<td>8 (324)</td>
</tr>
<tr>
<td>Orthopedic</td>
<td>9 (1329)</td>
</tr>
<tr>
<td>Other</td>
<td>7 (834)</td>
</tr>
<tr>
<td>Plastic/reconstructive</td>
<td>7 (173)</td>
</tr>
<tr>
<td>Rectal</td>
<td>4 (101)</td>
</tr>
<tr>
<td>Transplantation</td>
<td>6 (116)</td>
</tr>
<tr>
<td>Vascular</td>
<td>11 (832)</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td>9 (9203)</td>
</tr>
</tbody>
</table>

The predictors for pulse oximetry failure in Reich’s study were ASA physical status 3, 4, and 5 patients, the type and duration of the operation, especially cardiopulmonary bypass surgery, vascular, reconstructive or neuro-surgery, hypothermia, and hypotension.

So the question becomes, what to do in such cases, particularly those in which the oximeter is unable to find an adequate pulse at any of the available peripheral sites at just the time when SpO₂ readings would be most clinically valuable. There is, therefore, a need to find a means of solving this frustrating and serious clinical problem.

In an attempt to avoid the difficulties associated with conventional measurements of arterial blood oxygen saturation during conditions of poor
peripheral perfusion and pulsation, it has been proposed to use the upper oesophagus as a measurement site. ATLEE et al. (1995) presented results of blood oxygen saturation measurements obtained at the cricopharyngeus muscle in the upper oesophagus (14 ± 1 cm from incisors) using a “transoesophageal” probe where the optical components of the oximetry sensor were incorporated into a traditional anaesthesia oesophageal stethoscope. They compared their “transoesophageal” probe SpO\textsubscript{2} measurements with simultaneous SpO\textsubscript{2} measurements from conventional pulse oximetry probes (Nellcor N-200: N-200F) and arterial oxygen saturation (SaO\textsubscript{2}) measurements using an in vitro CO-Oximeter (O:482 CO-Oximeter) in 16 anaesthetised adult patients (ASA 1-3). The results showed that the “transoesophageal” probe underestimated or overestimated SpO\textsubscript{2} values depending on the geometry of the sensor (PRIELIPP et al., 1996; BORUM 1997). Another limitation of this design was the difficulty in placing the probe accurately at the cricopharyngeus muscle, as the procedure required considerable expertise. It was also found that electrocautery interference resulted in more frequent signal dropout and delayed signal reacquisition than for a peripheral pulse oximetry probe. The study by ATLEE et al. (1995) did not present any investigations into the morphology or the quality of PPG signals at the cricopharyngeus muscle or at any other depths within the oesophagus.

To overcome the drawbacks of Atlee’s design, which can operate only in the upper oesophagus, and the difficulties which are associated with attempts to measure arterial blood oxygen saturation in the poorly perfused peripheral circulation, the investigation of PPG signals within the whole depth of the oesophagus is suggested. Pulsatile signals necessary for SpO\textsubscript{2} estimation may be larger and more consistent at this site. The use of this central probe location should increase the likelihood of obtaining readings of oxygen saturation from the group of patients described above. In the work discussed here, such a probe location is investigated. This novel location – the oesophagus – could eventually allow pulse oximeters to operate reliably in patients undergoing cardiopulmonary bypass surgery under hypothermia in addition to providing benefits for other patient groups where their peripheral
circulation is compromised. Also, such central monitoring location might be very useful in patients with severe burns where there may be no suitable peripheral site to attach a conventional pulse oximeter probe.

To validate the hypothesis that perfusion in the oesophagus might be preferentially preserved, purposely made miniaturised oesophageal probes have been constructed and are described in detail in the next chapter.
CHAPTER 9

Design and Evaluation of Reflectance Oesophageal and Finger Photoplethysmographic (PPG) Probes

9.1 INTRODUCTION
To overcome the limitations of commercial pulse oximeters especially in cases of poor peripheral perfusion, as described in the previous chapter, two oesophageal photoplethysmographic (PPG) probes were developed. These new probes were used to investigate PPG signals within the whole depth of the oesophagus, where pulsatile signals necessary for $\text{SpO}_2$ estimation may be larger and more consistent.

A prototype reflectance oesophageal PPG probe (Oesophageal PPG Probe I) operating at one wavelength (infrared) was first constructed. This probe has been initially used to investigate infrared AC PPG signals in the deep oesophagus of a healthy awake volunteer. Later on, the probe was used for the first clinical trials on ASA physical status 1 anaesthetised patients. A second oesophageal PPG probe (Oesophageal PPG Probe II) which operates at two wavelengths, red and infrared (conventional wavelengths for pulse oximetry) was also designed and constructed. This probe was used for the detailed investigation of PPG signals at two wavelengths, red and infrared, from the whole depth of the human oesophagus of ASA physical status 1 anaesthetised patients. The Oesophageal PPG Probe II was also used for extensive clinical trials on cardiopulmonary surgery patients (ASA 2, 3,4) to investigate PPG signals and estimate blood oxygen saturation in the oesophagus. A finger reflectance PPG probe optically identical to the Oesophageal PPG Probe II was also constructed to facilitate comparisons between the two sites (oesophagus and finger).
The details of the fabrication of the oesophageal probes and the finger probe as well as the performance evaluation tests at all stages of the probes design is the subject of this chapter.

9.2 OESOPHAGEAL PPG PROBE I

The prototype reflectance *Oesophageal PPG Probe I* was constructed utilising two surface mount infrared emitting diodes (IREDs) and a surface mount silicon diode photodetector (Figure 9.1). The photodetector detected radiation back scattered by the tissue from both infrared emitters and gave an output current proportional to the detected radiation level. A 4-core cable carried the power to the IREDs in the probe from the main PPG processing unit and also the detected PPG signals from the photodetector (Figure 9.1).

![Figure 9.1: Block diagram of the Oesophageal PPG Probe I connected to the PPG Processing System](image)

### 9.2.1 Optical Components

The infrared emitters (IREDs) used for this probe were ceramic chip surface mount types (dimensions: 3.2 mm x 1.27 mm) with peak emission wavelength at 880 nm (CR 50 IRH, ELCOS GmbH). The photodetector was a surface mount silicon PhotoPinDiode (dimensions: 4.57 mm x 3.81 mm) which had the positive side on the front and the negative side on a ceramic contact base (CFD 10, ELCOS GmbH). The optical and electrical characteristics and package details of the IRED emitters and the diode photodetector are shown in Table 9.1 and Table 9.2 respectively.
### Table 9.1: Optical, Electrical and Package Specifications of the Infrared Emitter (CR 50 IRH)

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak emission wavelength (λ)</td>
<td>880 nm</td>
</tr>
<tr>
<td>Radiated power (100 mA) / (20 mA)</td>
<td>9.2 mW / 2.0 mW</td>
</tr>
<tr>
<td>Power dissipation</td>
<td>130 mW</td>
</tr>
<tr>
<td>Forward voltage (100 mA) / (20 mA)</td>
<td>1.65 V / 1.35 V</td>
</tr>
<tr>
<td>Min. reverse voltage, V_r</td>
<td>8.0 V</td>
</tr>
<tr>
<td>Reverse leakage current</td>
<td>100 μA</td>
</tr>
<tr>
<td>Peak forward current at 10μs (I_{FSM})</td>
<td>800 mA</td>
</tr>
<tr>
<td>Continuous forward current (I_{C})</td>
<td>75 mA</td>
</tr>
<tr>
<td>Light emission angle</td>
<td>160° Degree</td>
</tr>
<tr>
<td>Operating temperature (T_{OP})</td>
<td>-25 to 80°C</td>
</tr>
<tr>
<td>Soldering temperature (T_{SOLD}), 10 s</td>
<td>250°C</td>
</tr>
</tbody>
</table>

### Package Details

![Package Diagram](image)

### Table 9.2: Optical, Electrical and Package Specifications of the silicon diode photodetector (CFD 10)

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak wavelength sensitivity (λ_{max})</td>
<td>880 nm</td>
</tr>
<tr>
<td>Range of spectral band width (50%)</td>
<td>500 ... 1000 nm</td>
</tr>
<tr>
<td>Active area</td>
<td>7.34 mm²</td>
</tr>
<tr>
<td>Sensitivity (S) @ V_R=5V, E_V=1kLx</td>
<td>85 nA/Lx</td>
</tr>
<tr>
<td>Junction capacitance (C_j) @ V_R=3V, f=1 MHz</td>
<td>50 pF</td>
</tr>
<tr>
<td>Open circuit voltage (V_o) @ E_V=1kLx</td>
<td>365 mV</td>
</tr>
<tr>
<td>Light reverse current (I_{L}) @ V_R=5V, E_V=1kLx</td>
<td>85 mA (typ)</td>
</tr>
<tr>
<td>Reverse dark current (I_{R}) @ V_R=10 V</td>
<td>5 μA (typ)</td>
</tr>
<tr>
<td>Reverse voltage (V_R) Max. @ 25°C</td>
<td>32 V</td>
</tr>
<tr>
<td>Power dissipation (P_V)</td>
<td>200 mW</td>
</tr>
<tr>
<td>Operating temperature (T_{OP})</td>
<td>-25 to +80°C</td>
</tr>
<tr>
<td>Soldering temperature (T_{SOLD}), 10 s</td>
<td>260°C</td>
</tr>
</tbody>
</table>

### Package Details

![Package Diagram](image)
9.2.2 Mechanical Construction of Oesophageal PPG Probe I

The photodetector was mounted between the IREDs to detect radiation back scattered by the tissue from both infrared emitters and gave an output current proportional to the detected radiation level. The distance between each emitter and the photodetector was 5 mm (Figure 9.2b). Experimental studies by MENDELSON and OCHS (1988) showed that a separation of 4 to 5 mm between the emitters and the photodetector provides the best sensitivity in terms of detecting adequately large pulsatile photoplethysmographic signals.

The emitter and photodiode chips were mounted on the copper side (Figure 9.2b) of an epoxy glass copper clad single sided eurocard (dimensions: 25 mm x 4 mm x 1.0 mm) (RS 435-484). The copper tracks (see Figure 9.2a) were hand-etched using a stanley knife (RS 546-742) and a needle file set (RS 545-121). Low temperature (200°C to 250°C) soldering was used for the mounting of the surface mount components on the copper tracks of the board.

![Diagram](image)

**Figure 9.2:** Top view of the Oesophageal PPG Probe I. (a) shows the hand-etched copper tracks (b) surface mount components (IREDs and photodetector) mounted on the board.
A four-core screened multicore cable (RS 367-347) (see specifications in Table 9.3) carried the power to the IREDs in the probe from the main PPG processing unit and also the detected PPG signal from the photodetector (Figure 9.1). This cable was electrically screened to minimise electromagnetic interference. Figure 9.3 shows a close-up photograph of the complete design of the Oesophageal PPG Probe I.

**Table 9.3: Specifications of Screened Multicore Cable (RS 367-347)**

<table>
<thead>
<tr>
<th>General Description</th>
<th>7/0.01 mm (0.055 mmsq) tinned copper stranded conductors covered by 0.2 mm PVC, overall tinned copper braided screen, outer sheath grey PVC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max. working voltage</td>
<td>250 V r.m.s.</td>
</tr>
<tr>
<td>Max. current per core</td>
<td>0.25 A r.m.s.</td>
</tr>
<tr>
<td>Max. diameter: 4 core / 6 core</td>
<td>3.4 mm / 4.0 mm</td>
</tr>
<tr>
<td>Core/Screen capacitance (nominal)</td>
<td>85 pF/m</td>
</tr>
<tr>
<td>Resistance per core</td>
<td>384 Ω/km</td>
</tr>
</tbody>
</table>

**Figure 9.3: Photograph of the infrared Oesophageal PPG Probe I**
The *Oesophageal PPG Probe I* with the cable attached was designed to fit into a plastic transparent disposable stomach/oesophageal tube (Pennine Healthcare, Derby, UK). The oesophageal tube used was a size French 20 (external diameter: 6.66 mm, internal diameter: 4.66 mm, length 780 mm, without X-ray detectable line) mainly used for gastric lavage (washout) or other gastric surgical procedures. The tube has a closed end with two side drainage holes (Figure 9.4).

![Oesophageal tube, French 20](image)

**Figure 9.4: Oesophageal tube, French 20**

The oesophageal tube was sealed at the bottom to prevent gastric juices or oesophageal mucous entering the drainage holes of the stomach tube and reaching the electronic and optical components of the probe during clinical measurements. The sealing of the tube (above the drainage holes) was
achieved using a custom manufactured (Medical Electronics Workshop) perspex cylindrical rod (length: 20 mm, diameter 5 mm) (Figure 9.5a). The perspex rod was first lubricated using K-Y Lubricating Jelly (Johnson & Johnson Ltd, Maidenhead, U.K.). The bottom end of the plastic oesophageal tube was warmed using a hot air blower to soften the plastic and allow it to expand easily. The perspex rod was pushed from the upper drainage hole towards the top of the tube (see Figure 9.5b) until the bottom end of the rod was situated above the upper drainage hole of the tube (Figure 9.5c). The diameter of the rod was slightly greater than the diameter of the tube (approximately 0.3 mm) therefore when pushed into the tube a tight contact with the plastic was accomplished, which kept it firmly in position. In its final position the end of the Oesophageal PPG Probe 1 was approximately 60 mm from the end of the stomach tube, see Figure 9.5c.

Figure 9.5: Positioning of perspex rod in oesophageal tube (a) shows the perspex rod and the bottom end of the oesophageal tube with the two drainage holes; (b) the perspex rod is pushed from the upper drainage hole towards the top of the tube; (c) final position of the rod in the tube with the Oesophageal PPG Probe 1 in situ
9.2.3 Performance Evaluation of Oesophageal PPG Probe I

The functionality of the Oesophageal PPG Probe I was tested in the laboratory. The Probe was inserted into a stomach tube (French 20) until the end of the probe itself was approximately 6 cm from the bottom end of the tube. Prior to insertion of the probe into the stomach tube the plastic tube was warmed using a conventional hot air blower. The warming of the plastic made the sliding of the probe into the tube more easy since the clearance between the tube and the probe was minimal (approximately 0.33 mm) (Figure 9.6).

![Diagram of Oesophageal PPG Probe I](image)

Figure 9.6: Side and cross sectional view of the Oesophageal PPG Probe I. In the cross sectional view the probe is shown inserted in the oesophageal tube.

The cable was connected to a prototype PPG processing system (PPG Processing System 1) which will be described in detail in the next chapter. This processing system comprised a constant current source used to drive the infrared emitters continuously at 40 mA, a transresistance amplifier to convert the output current of the photodetector into a signal voltage and a PPG pre-amplification and filtering stage. The output of the PPG Processing System 1 was connected to a digital storage oscilloscope (Tektronix 2211). Infrared AC PPG signals were obtained and displayed on the oscilloscope screen by placing an index finger flat on the plastic stomach tube on top of
the probe facing the optical components. The laboratory lights were switched
off to eliminate any interference (as described in section 8.3.1) from the room
fluorescent lights being detected by the photodetector. No significant
pressure was applied on the plastic tube by the finger. The PPG signals were
of good quality, large amplitudes (400 mV to 600 mV) and high signal-to-
noise ratio. Infrared (IR) AC PPG signals were recorded using a 10-bit
analogue to digital card (ADC) and Picolog, a data acquisition and logging
system, (Pico Technology Ltd, Cambridge, UK) and a personal computer.
Typical IR AC PPG traces from the finger using the Oesophageal PPG Probe
I are shown in Figure 9.7.

![Figure 9.7: Finger Infrared AC PPG traces obtained using the Oesophageal PPG Probe I](image)

PPG traces from the finger using the Oesophageal PPG Probe I were
recorded continuously for approximately 30 minutes. During the 30 minutes
of monitoring there was no subjective indication of any noticeable rise in
temperature in the finger. This suggested that the temperature rises directly
on top of the 1 mm thick insulating plastic of the oesophageal tube due to the
infrared emitters were insignificant. \textit{In vitro} and \textit{in vivo} temperature tests of the emitters are described later on in the chapter.

9.2.4 Limitations of Oesophageal PPG Probe I

Despite the successful operation of the probe there were two limitations that made it not so practical. The first limitation was the size of the probe itself. The probe would only slide down to the bottom of the stomach tube, as described in the previous sections, if the plastic of the tube was softened (by warming) prior to insertion of the probe. To eliminate this problem it was suggested to change the type of stomach tube size from French 20 (internal diameter: 4.33 mm) to French 22 (internal diameter: 5.33 mm). A French 22 stomach tube would easily allow the probe to slide down to the bottom of the tube without any warming of the plastic. The change of stomach tube from French 20 to 22 size was undesirable and was disapproved by clinical collaborators since a French 22 is not a routinely used tube in anaesthetics (unlike French 20 tubes) and its size is rather large for application into the oesophagus.

The second limitation was the way the tube was sealed at the bottom (see section 9.2.3). The size and the rigidity of the perspex rod, which was custom manufactured for sealing the drainage holes of the stomach tube, compromised the flexibility of the oesophageal tube, with the possibility of causing problems during insertion of the tube in the human oesophagus.

Also, the \textit{Oesophageal PPG Probe I} was designed to operate only at one wavelength (infrared) which made it a not suitable probe for estimating blood oxygen saturation by pulse oximetry since pulse oximeters require two wavelengths (red and infrared).

9.3 OESOPHAGEAL PPG PROBE II

To overcome the limitations of the \textit{Oesophageal PPG Probe I}, as discussed above, a new PPG probe (\textit{Oesophageal PPG Probe II}) which allowed the investigation of oesophageal PPGs at both red and infrared wavelengths.
(conventional wavelengths for pulse oximetry) within the whole length of the oesophagus was constructed. This new and optimised probe comprised two infrared and two red surface mount emitters and a surface mount photodetector (Figure 9.8). The photodetector detected radiation back scattered by the tissue from both infrared and red emitters and gave an output current proportional to the detected radiation level. A 6-core cable carried the power to the IRED and RED emitters in the probe from the main PPG processing unit and also the detected PPG signals from the photodetector (Figure 9.8).

![Figure 9.8: Block diagram of the Oesophageal PPG Probe II connected to the PPG processing System](image)

### 9.3.1 Optical Components
The infrared emitters (IREDs) and the photodetector used for this probe were the same as the ones used for Oesophageal PPG Probe I. The red emitters used were ceramic chip surface mount (dimensions: 3.2 mm x 1.27 mm) with peak emission wavelength at 655 nm (CR 10HRT, ELCOS GmbH). The optical and electrical characteristics and package details of the IRED and RED emitters and the diode photodetector are shown in Tables 9.1, 9.4 and 9.2 respectively.
Table 9.4: Optical, Electrical and Package Specifications of the Red Emitter (CR 10HRT)

<table>
<thead>
<tr>
<th>Optical and Electrical Characteristics at 25°C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak emission wavelength ($\lambda$)</td>
</tr>
<tr>
<td>Luminous intensity (at $I_F$=20 mA)</td>
</tr>
<tr>
<td>Power dissipation $P_{tot}$</td>
</tr>
<tr>
<td>Forward voltage ( at 20 mA)</td>
</tr>
<tr>
<td>Min. reverse voltage, $V_R$</td>
</tr>
<tr>
<td>Reverse leakage current</td>
</tr>
<tr>
<td>Peak forward current at 10µs ($I_{FSM}$)</td>
</tr>
<tr>
<td>Continuous forward current ($I_F$)</td>
</tr>
<tr>
<td>Light emission angle</td>
</tr>
<tr>
<td>Operating temperature ($T_{OP}$)</td>
</tr>
<tr>
<td>Soldering temperature ($T_{SOLD}$), 10 s</td>
</tr>
</tbody>
</table>

Package Details

9.3.2 Mechanical Construction of Oesophageal PPG Probe II
The photodetector was mounted between the red and infrared emitters to detect radiation back scattered by the tissue from both infrared and red emitters and gave an output current proportional to the detected radiation level. The distance between the emitters and the photodetector was again 5 mm (Figure 9.9a) for the reasons explained in section 9.2.2. The emitter and photodiode chips were mounted on the copper side (Figure 9.2a) of an epoxy glass copper clad single sided eurocard (dimensions: 20 mm x 3.5 mm x 1.0 mm). The copper tracks (see Figure 9.2b) were hand-etched using a stanley knife (RS 546-742) and a needle file set (RS 545-121). Low temperature (200°C to 250°C) soldering was used for the connecting of the surface mount components to the copper tracks of the board.
Figure 9.9: Top view of the Oesophageal PPG Probe II. (a) layout of the surface mount components (IRED and RED emitters and photodetector) mounted on the board; (b) shows the hand-etched copper tracks.

Insulating wrapping wire (30 AWG) was used for the interconnection (using low temperature soldering) of the optical components (emitters and photodiode). A six-core screened cable (RS 367-369) (see specifications in Table 9.3) carried the power to the IRED and RED emitters in the probe from the main PPG processing unit and also the detected PPG signals from the photodetector (Figure 9.8). Figure 9.10 shows a close-up photograph of the complete Oesophageal PPG Probe II.
The Oesophageal PPG Probe II with the cable attached was designed to fit into a French 20 gauge plastic transparent disposable stomach/oesophageal tube (same as the Oesophageal PPG Probe I). The oesophageal tube was again sealed at the bottom therefore preventing gastric juices or oesophageal mucous entering the drainage holes of the stomach tube and reaching the electronic and optical components of the probe during clinical measurements. The sealing of the tube for this probe design was achieved using a solid rubber stopper (dimensions: upper diameter: 6 mm; lower diameter: 4 mm; length: 14 mm) (Fisher Scientific, Loughborough, U.K.) part number (SYH-450-043V) (Figure 9.11a). The rubber stopper was first lubricated, using K-Y Lubricating Jelly (Johnson & Johnson Ltd, Maidenhead, U.K.). Then was pushed from the upper drainage hole of the tube towards the top of the tube, (Figure 9.11b) until the bottom end of the rubber stopper was situated approximately 1 cm above the upper drainage hole of the tube (Figure 9.11c). No heating of the plastic tube was required prior to placing the rubber stopper in the tube. The upper diameter of the rubber stopper is greater than the diameter of the tube (diameter difference: 1.34 mm) therefore when
pushed into the tube a tight contact with the plastic was accomplished, which kept it firmly in position. In its final position the end of the *Oesophageal PPG Probe II* was approximately 60 mm from the end of the stomach tube, see Figure 9.11c.

![Diagram of rubber stopper in oesophageal tube](image)

**Figure 9.11:** Positioning of rubber stopper in oesophageal tube (a) shows the rubber stopper and the bottom end of the oesophageal tube with the two drainage holes; (b) the rubber stopper is pushed from the upper drainage hole towards the top of the tube; (c) final position of the rubber stopper in the tube with the *Oesophageal PPG Probe II* in situ

### 9.3.3 Performance Evaluation of Oesophageal PPG Probe II

The functionality of the *Oesophageal PPG Probe II* was tested in the laboratory. The *Oesophageal PPG Probe II* connected to the 6-core screened cable (external diameter 4.0 mm) was slide along a plane containing the axis and diameter of the conventional disposable transparent stomach tube (20 French gauge) of internal diameter of 4.66 mm. There was
little clearance between the wall of the stomach tube and the probe (approximately 0.43 mm) thereby minimising relative movement (Figure 9.12). Despite the small clearance between the probe and the internal walls of the plastic tube, the probe was able to slide all the way down to the bottom of the tube very easily and without any warming of the tube.

**Figure 9.12:** Side and cross sectional view of the *Oesophageal PPG Probe II*. In the cross sectional view the probe is shown inserted in the oesophageal tube;

The reduction of the probe width from 4.0 mm (*Oesophageal PPG Probe I*) to 3.5 mm (*Oesophageal PPG Probe II*) solved one of the limitations of the *Oesophageal PPG Probe II*, which was the need to warm the tube prior to sliding the probe into the tube. Also, the replacement of the 20 mm perspex rod which was used to seal the drainage holes of the oesophageal tube of the *Oesophageal PPG Probe II* with the rubber stopper, was found to be an adequate solution since the rubber stopper was smaller, softer and much more flexible than the perspex rod. Hence, with this new technique in sealing the drainage holes of the oesophageal tube, the bending properties of the plastic tube were not compromised as it would have been with the perspex rod, especially at the time of insertion of the tube in the human oesophagus, where bending of the bottom end of the tube was necessary.
The cable was connected to a mini 6-pin DIN socket located on the front panel of the PPG processing system (*PPG Processing System 2*) which will be described in detail in the following chapters. Red and infrared PPG measurements were made consecutively by means of mechanical switches, which allowed the changeover from one wavelength to the other. Infrared and red AC PPG signals were obtained by placing an index finger flat on the plastic stomach tube over the probe facing the optical components. Again, the laboratory lights were switched off to eliminate any interference (as described in section 8.3.1) from the room fluorescent lights. *Picolog* was used to acquire the signals and display them on a PC computer screen (see section 9.2.3). The PPG signals were again of good quality and large amplitudes for both red and infrared wavelengths. Typical red and infrared AC PPG traces from the finger using the *Oesophageal PPG Probe II* are shown in Figure 9.13.

![Figure 9.13: Infrared and red AC PPGs obtained from the finger using the *Oesophageal PPG Probe II*](image)

**9.3.4 Temperature tests on Oesophageal PPG Probe II**

The operating current of the red and infrared emitters was relatively low (45 mA) and they were thermally insulated from the tissue by the 1 mm thick plastic wall of the oesophageal tube. Despite the fact that the operating
current of the emitters was well below the maximum rated value of 75 mA, it was still necessary to confirm that the red and infrared sources would not cause any direct thermal damage to the oesophagus. Temperature tests both in vitro and in vivo were conducted to investigate the possibility of excessive temperature rises in the oesophagus during PPG measurements.

**In vitro Measurements**

Measurements were made of the changes in temperature at the outside and inside of the oesophageal tube. In order to simulate conditions in the oesophagus, the oesophageal tube with the *Oesophageal PPG Probe II* inside was immersed in a static water bath at 37°C, as water has thermal properties similar to those of soft tissue. Figure 9.14 shows the experimental set-up for the in vitro temperature measurements at the outside of the oesophageal tube.

![Figure 9.14: Block diagram of the in vitro temperature test (outside oesophageal tube)](image)

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124
Two type K thermocouples were used as temperature sensors. The first thermocouple was attached to the outside of the oesophageal tube adjacent to one of the infrared emitters (Figure 9.14). The thermocouple was secured in place by taping (using electrical tape) the leads of the thermocouple on the plastic oesophageal tube. The second thermocouple was used to monitor the temperature of the water bath (Figure 9.14). Two thermocouple amplifiers (AD595CQ, Analog Devices) were used to produce linear voltage outputs with sensitivity 10 mV/°C. The AD595CQ is a complete instrumentation amplifier and thermocouple cold junction compensator on a monolithic chip. It combines an ice-point reference with a pre-calibrated amplifier to produce a high level, low impedance, 10 mV/°C output directly from the thermocouple signal. Table 9.5 gives a brief outline of the electrical and package specifications of the AD595CQ thermocouple amplifier.

### Table 9.5: Specifications of the AD59CQ Thermocouple Amplifier

| Absolute Maximum Ratings |  |
|--------------------------|--|---|
| +Vs to -Vs               | 36 Volts (max) |
| Common mode input voltage | -Vs-0.15 (min), +Vs (max) |
| Differential input voltages | -Vs (min), +Vs (max) |
| Operating temperature range | -55°C to +125°C |

### Temperature Measurement

| Calibration error at +25°C | ± 1°C (max) |
| Stability vs. Temperature  | ± 0.025°C (max) |

The two temperature signals from the thermocouple amplifiers were subtracted \((T_{\text{tube}} - T_{\text{water}})\) using operational amplifier circuits and the temperature difference displayed on a chart recorder (Servogor, Model 464) (Figures 9.14 & 9.15).
After a constant baseline reading was achieved, the infrared light emitters were switched on and the temperature difference was monitored on the chart recorder (Figure 9.16). The measurement was repeated for a red emitter (Figure 9.17).

Figure 9.16: *In vitro* temperature rise test with the thermocouple located at the outside surface of the plastic oesophageal tube adjacent to an infrared emitter; Display: (Ttube – Twater); Time: right to left

**Figure 9.15:** Temperature processing circuit
The rise in temperature directly on top of the emitters inside the oesophageal tube was also measured. The experimental set-up for this temperature test was exactly the same as the one described previously (Figure 9.14), apart from the fact that the thermocouple from the surface of the oesophageal tube was now placed directly on top of one of the infrared emitters (one nearest to the multicore cable of the probe) inside the tube. A second thermocouple was again used to measure the water bath temperature, which was set at 37°C. The temperature processing circuit (Figure 9.15) was again used to subtract the two temperatures (\( T_{\text{emitter}} - T_{\text{water}} \)) and display the difference on the chart recorder. After a constant baseline reading was achieved, the infrared light emitters were switched on and the temperature difference was monitored (Figure 9.18). The temperature experiment was repeated for a red emitter as well (Figure 9.19).
Table 9.6 summarises the results from \textit{in vitro} temperature tests.

\textbf{Table 9.6:} Results from the \textit{in vitro} temperature tests for the infrared and red emitters

\begin{tabular}{|c|c|c|c|}
\hline
 & RED (°C) & INFRARED (°C) & RED (°C) & INFRARED (°C) \\
\hline
Temperature rises with the thermocouple placed at the inside of the plastic oesophageal tube adjacent to either the red or infrared emitter & 0.2 & 0.15 & 3.0 & 2.0 \\
\hline
Temperature rises with the thermocouple placed at the outside surface of the plastic oesophageal tube adjacent to either the red or infrared emitter & & & & \\
\hline
\end{tabular}

\textbf{In vivo Measurements}

\textit{In vivo} simulations were also made in three normal healthy volunteers. The oesophageal tube with the PPG probe inside was placed in contact with the mucosa of the cheek inside the mouth (Figure 9.20). Two type K thermocouples were attached to the outside of the oesophageal tube. The first thermocouple was attached on the outside surface of the plastic oesophageal tube directly on top of one of the infrared emitters. The second thermocouple was attached at the front of the tip outside the oesophageal tube (approximately 7 cm from the bottom end of the probe) to monitor the
temperature of the mucosa of the cheek inside the mouth (Figure 9.20). The two temperature signals from the thermocouple amplifiers were subtracted ($T_{\text{tube}} - T_{\text{mouth}}$) using the same operational amplifier circuits (Figure 9.15) used for the *in vitro* temperature testing and the temperature difference displayed on a chart recorder (Servogor, Model 464).

![Block diagram of the *in vivo* temperature test](image)

**Figure 9.20:** Block diagram of the *in vivo* temperature test

The difference of the two temperatures in the mouth was recorded until a constant baseline was attained. The infrared light emitters were switched on and the temperature was monitored. Steady state conditions were achieved after approximately 10 minutes and monitoring was continued for a further 8 minutes. A typical section of the temperature trace where the infrared
emitters were switched on is shown in Figure 9.21. The measurement was repeated for a red emitter (Figure 9.22).

**Figure 9.21:** *In vivo* temperature rise test (mucosa of the mouth) with the thermocouple located at the outside of the plastic oesophageal tube adjacent to an infrared emitter; Display: \((T_{\text{tube}} - T_{\text{mouth}})\); Time: right to left

**Figure 9.22:** *In vivo* temperature rise test (mucosa of the mouth) with the thermocouple located at the outside of the plastic oesophageal tube adjacent to a red emitter; Display: \((T_{\text{tube}} - T_{\text{mouth}})\); Time: right to left

In these *in vivo* tests the rise in temperature at the outside surface of the oesophageal tube, when in contact with the cheek mucosa, was less than 0.7°C (mean 0.64 °C, n=3) for the red emitter and 0.6°C (mean 0.52 °C, n=3) for the infrared emitter in all cases. The higher temperature rise in the *in vivo* measurements compared with the *in vitro* measurements may be due to the cheek mucosa making poorer thermal contact with the wall of the stomach tube than the water.
None of these temperature rises would be expected to result in tissue damage and therefore, it was concluded that there would be negligible risk of thermal injury to the oesophagus using the *Oesophageal PPG Probe II*.

### 9.4 FINGER REFLECTANCE PPG PROBE

A finger reflectance PPG probe, optically identical to the *Oesophageal PPG Probe II* has also been designed and constructed comprising two infrared and two red surface mount emitters and a surface mount photodetector. The design of this finger probe was intended to facilitate comparisons between the two sites (oesophagus and finger).

#### 9.4.1 Optical Components

The infrared and red emitters and the photodetector used for this finger PPG probe were the same as the ones used for the *Oesophageal PPG Probe II*. The optical and electrical characteristics and package details of all optical components used for this probe are shown in Tables 9.1, 9.4 and 9.2 respectively.

#### 9.4.2 Mechanical Construction of the Finger PPG Probe

The photodetector was mounted between the red and infrared emitters to detect radiation back scattered by the tissue from both infrared and red emitters and gave an output current proportional to the detected radiation level. The distance between the emitters and the photodetector was again 5 mm (Figure 9.23a) for the reasons explained in section 9.2.2. The emitter and photodiode chips were mounted on the copper side (Figure 9.23a) of an epoxy glass copper clad single sided eurocard (dimensions: 20 mm x 10 mm x 1.6 mm). The copper tracks (see Figure 9.23b) were hand-etched using the same techniques as the previous oesophageal PPG probes. Low temperature soldering was again used for the mounting of the surface mount components on the copper tracks of the board. A six-core screened cable (RS 367-369) (see specification in Table 9.3) carried the power to the IRED and RED emitters in the probe from the main PPG processing unit and also the detected PPG signal from the photodetector (Figure 9.8). A damaged commercial finger clip type pulse oximeter probe was used to accommodate
the copper board of the new reflectance finger PPG probe. The optical components and cable of the commercial finger probe were removed and only the finger clip mechanism was used in the finger probe design.

* Figure 9.23: Top view of the Finger PPG probe. (a) layout of the surface mount components (IRED and RED emitters and photodetector) mounted on the board; (b) shows the hand-etched copper tracks

* All measurements are in mm

The bottom side of the copper board of the newly constructed finger probe was glued on to the bottom rubber side of the finger clip using a multi-purpose clear adhesive (RS 315-8584). A clear insulating plastic material
(thickness 0.05 mm) was wrapped around the surface of the probe to prevent direct contact of the optical components with the finger when inserted into the clip. Figure 9.24 shows a close-up photograph of the complete design of the reflectance finger probe.

![Photograph of the Reflectance Finger Probe](image)

**Figure 9.24**: Photograph of the Reflectance Finger Probe

### 9.4.3 Performance Evaluation of the Finger PPG Probe

The functionality of the finger probe was tested in the laboratory. The 6-core cable of the finger PPG probe was connected to the PPG processing system (*PPG Processing System 2*) which will be described in detail in the following chapters. Red and infrared PPG measurements were made consecutively by means of mechanical switches, which allow the changeover from one wavelength to the other. Infrared and red AC PPG signals were obtained by placing an index finger flat in the finger clip on top of the probe facing the optical components. Again, the laboratory lights were switched off to eliminate any interference from the room fluorescent lights. *Picolog* was used to acquire the signals and display them on the computer screen. The signals were of large amplitudes and high signal-to-noise ratio. Typical red and
infrared AC PPG traces from the finger using the reflectance finger PPG probe are shown in Figure 9.25.

Figure 9.25: Infrared and red AC PPGs obtained from the finger using the reflectance finger PPG probe
CHAPTER 10

PPG Processing System 1

10.1 INTRODUCTION

A prototype one wavelength (infrared) PPG data acquisition system has been developed to detect, pre-process, sample, record and display infrared AC PPG signals on a laptop personal computer. It consisted of two identical PPG processing channels (Figure 10.1), which allowed the monitoring of oesophageal and finger infrared AC PPG signals simultaneously using the Oesophageal PPG Probe I and the Finger Probe discussed in the previous chapter.

![Diagram of PPG Processing System 1]

Figure 10.1: Basic block diagram of the infrared PPG Processing System 1

The PPG Processing System 1 was used for the first PPG measurements in the deep oesophagus of a healthy awake volunteer. This system was also
used for a preliminary pilot study to investigate infrared AC PPG signals in the deep oesophagus in anaesthetised patients. No previous PPG measurements appear to have been reported for the human oesophagus.

The technical details of the construction of this processing system as well as the performance evaluation tests at all stages of the system’s design will be the subject of this chapter. The clinical methods and the results from all the infrared PPG measurements using the PPG Processing System 1 will also be presented and discussed.

10.2 INSTRUMENTATION

A detail block diagram of one of the two identical PPG channels (oesophageal) is shown in Figure 10.2.

![Figure 10.2: Detail block diagram of one of the two identical channels of the infrared PPG Processing System 1](image)

The Oesophageal PPG Probe I (see Figure 9.3) was connected to the PPG Processing System 1. The infrared emitters (IREDs) were driven by an IRED driver which is a constant current source comprising an operational amplifier and a series transistor. The photodetector detected the energy back scattered by tissue from both infrared emitters and gave an output current proportional to the detected radiation level. The photodetector (I-V) amplifier
was an operational amplifier connected in a transresistance configuration that converted the photodetector current into a signal voltage. The signal was then passed to a signal conditioning circuit comprising a PPG filter and Pre-amplifier circuit. The final stage consisted of a variable amplification stage and a DC level shifting circuit, which gave an output signal compatible with the unipolar input range of the ADC-11 analogue-to-digital converter. The whole PPG Processing system 1 was battery operated (± 9 V).

A detailed description of each of the main blocks of the PPG Processing System 1 is presented below.

**10.2.1 IRED driver**
The infrared emitters (IREDS) in each channel (oesophageal and finger) were driven by an IRED driver circuit which is a constant current source comprising a low power operational amplifier (TL084, Texas Instruments, Dallas, Texas, USA) and a series NPN transistor (BC184L) (Figure 10.3). The TL084 operational amplifier features high slew rates and low input bias and offset currents and requires 1.4 mA per amplifier (see Table 10.1 for electrical specifications). The output intensity depends on the forward current flowing through the IREDs, which was maintained at 40 mA by the constant current source. The IREDs were operated well below their maximum rated current of 75 mA.

![Figure 10.3: IRED driver circuit (constant current source)](image-url)
The feedback in the circuit forces a voltage $V_{in}$ across $R3$, giving an emitter current (and therefore an output current):

$$I_E = \frac{V_{in}}{R3} \quad (10.1)$$

**Calculations:**

$$V_{in} = \left( \frac{R2}{R1 + R2} \right) V_{cc} = \left( \frac{0.96K}{1.2K + 0.96K} \right) 9V = 4V \quad (10.2)$$

Therefore, $I_E \cong I_C = \frac{V_{in}}{R3} = \frac{4V}{100R} = 40mA$

**Note:** The variable resistor (R2) was set at 0.96K

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**Table 10.1:** Main Electrical and Package Specifications of the TL084 Op-Amp

<table>
<thead>
<tr>
<th>Electrical Characteristics at 25°C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating supply voltage range</td>
</tr>
<tr>
<td>Input offset voltage</td>
</tr>
<tr>
<td>Input offset current</td>
</tr>
<tr>
<td>Common mode input voltage range</td>
</tr>
<tr>
<td>Common mode rejection ratio</td>
</tr>
<tr>
<td>Supply current per amplifier (no load)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Symbol and Package Details</th>
</tr>
</thead>
</table>

**10.2.2 Photodetector I-V Amplifier**

Transresistance amplifiers or current-to-voltage converters, are amplifiers that convert an input current to an output voltage. The current generated by the photodetector (CFD 10), in *Oesophageal PPG Probe I*, was converted to a voltage using an operational amplifier (TL084) connected in a transresistance configuration (Figure 10.4).
Because of the virtual ground, the operational amplifier maintains zero voltage across the photodiode. Current flows through the feedback resistor \( R_f \) and creates a voltage at the output that is proportional to the light intensity as given by:

\[
V_o = I_d R_f \tag{10.3}
\]

The transresistance gain is then equal to the value of the feedback resistor. The feedback resistor should be made as large as possible to minimise noise. This is because the feedback resistor is the dominant source of noise in the circuit. This thermal (Johnson) noise increases as a function of the square root of the value of the feedback resistance.

\[
\text{thermal noise} = \sqrt{4kTBR} \tag{10.4}
\]

where \( k \) is Boltzmann’s constant, \( T \) is absolute temperature, \( B \) is the noise bandwidth (Hz), \( R \) is the feedback resistance (\( \Omega \)), while the signal voltage increases as a function of \( R \). Therefore the signal-to-noise ratio improves by the square root of the feedback resistance as the feedback resistance is increased (WEBSTER, 1997). The capacitor in the feedback loop \( C_f \) minimises gain peaking and improves stability.
10.2.3 PPG filter & Pre-amplifier

The infrared DC and AC coupled PPG signal at the output of the transresistance amplifier passed through to a band-pass filter and pre-amplifier circuit (Figure 10.5). The PPG pre-amplifier and filter circuit consisted of a first order high-pass active filter, to remove the DC PPG component, and a Butterworth 2-pole low pass active filter which attenuated high frequencies. The frequency response was 0.48 Hz to 20 Hz (at -3dB) with a gain of 1.58 in the pass band. The low-pass cut-off frequency was chosen to be 20 Hz because the PPG signal frequency components do not extend much beyond 12 Hz (see section 3.1).

![Diagram of PPG filter and pre-amplifier](image)

Notes
1. Resistor type: 0.5W High Stability Carbon Film Resistors (tolerance +/- 5%)
2. Capacitors: polyester mini case capacitors (capacitance tolerance +/- 5%)

**Figure 10.5:** Band-pass and pre-amplification circuit: The frequency response is 0.48 to 20 Hz (at -3 dB) with a gain of 1.58 in the pass-band. (a): The first op-amp configuration is a high pass active filter with a gain of 1 (b): The second op-amp configuration is a Butterworth 2-pole low pass active filter with a gain of 1.58.

**High-Pass Active Filter**

Figure 10.5a shows a one op-amp high pass filter. Such a circuit is useful for passing a small AC signal that rides on top of a large DC voltage, because capacitor $C_1$ blocks the DC. The frequency response equation is:

$$\frac{V_o(j\omega)}{V_i(j\omega)} = \frac{Z_o}{Z_i} = -\frac{R_2}{j\omega C_1 + R_1} = \frac{j\omega R_2 C_1}{1 + j\omega C_1 R_1} = -\frac{R_2}{R_1} \frac{j\omega}{1 + j\omega R_1}$$

(10.5)
where $\tau = R_1 C_1$. For $\omega << 1/\tau$, the circuit behaves as a differentiator because $C_1$ is the dominant input impedance. For $\omega >> 1/\tau$, the circuit behaves as an inverting amplifier, because the impedance $R_1$ is large compared with that of $C_1$. The gain of this high pass filter design was:

$$\text{Gain} = -\frac{R_2}{R_1} = -\frac{150K}{150K} = -1$$

The cut-off frequency $f_c$ is given by the relation $\omega \tau = 2\pi f_c \tau = 1$

Therefore the cut-off frequency for the high pass filter in this design was equal to:

$$f_c = \frac{1}{2\pi \tau} = \frac{1}{2\pi R_1 C_1} = \frac{1}{2\pi \left( 50 K \Omega \right) \left( 0.2 \mu F \right)} = 0.48 \text{ Hz}$$

(10.6)

The cut-off frequency of the high-pass filter was chosen to be around 0.48 Hz to primarily eliminate the DC component of the PPG signal and also to exclude some of the low frequency respiratory component (0.2 Hz to 0.3 Hz) from the AC PPG signal.

**Low-pass Active Filter**

Frequencies above 20 Hz were attenuated using a 2-pole voltage controlled voltage source (VCVS) Butterworth low-pass active filter shown in Figure 10.5b. The VCVS filter is a variation of the Sallen-and-Key type filter with the exception that it replaces the unity-gain follower with a non-inverting amplifier with a gain ($K$) greater than one (HOROWITZ and HILL, 1989). The VCVS filter was designed using Table 10.2 shown below.

**Table 10.2: VCVS LOW-PASS FILTERS (HOROWITZ and HILL, 1989)**

<table>
<thead>
<tr>
<th>Poles</th>
<th>Butterworth $K$</th>
<th>Bessel $f_n$</th>
<th>Butterworth $K$</th>
<th>Bessel $f_n$</th>
<th>Chebyshev (0.5 dB) $f_n$</th>
<th>Chebyshev (2.0 dB) $f_n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>1.586</td>
<td>1.272</td>
<td>1.268</td>
<td>1.231</td>
<td>1.842</td>
<td>0.907</td>
</tr>
<tr>
<td>4</td>
<td>1.152</td>
<td>1.432</td>
<td>1.084</td>
<td>0.597</td>
<td>1.582</td>
<td>0.471</td>
</tr>
<tr>
<td></td>
<td>2.235</td>
<td>1.606</td>
<td>1.759</td>
<td>1.031</td>
<td>2.660</td>
<td>0.964</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
A filter with a flat pass-band was required so that the pulsatile component of the PPG signal was not distorted. The Butterworth filter was selected because it produces the flattest pass-band at the expense of steepness in the transition region (from pass-band to stop-band) (HOROWITZ and HILL, 1989). In the 2-pole low-pass Butterworth filter (Figure 10.5b) all sections have the same values of R and C, given simply by \( RC = \frac{1}{2\pi f_c} \), where \( f_c \) is the desired –3dB frequency of the entire low-pass filter;

\[
\text{Therefore, } f_c = \frac{1}{2\pi RC} = \frac{1}{2\pi (2 \text{K}\Omega \times 1\mu\text{F})} = 19.4 \text{ Hz}
\]

The gain of this low-pass filter was:

\[
\text{Gain (A_v)}_{LP} = \frac{R5 + R6}{R6} = \frac{47\text{K} + 82\text{K}}{82\text{K}} = 1.58
\]

The overall gain of the band-pass filter was equal to:

\[
\text{Gain (A_v)}_{BP} = \text{Gain (A_v)}_{LP} \times \text{Gain (A_v)}_{HP} = 1 \times \frac{58}{158} = -1.58
\]

A linear analogue circuit analysis simulation program, Analyser (Number One Systems Ltd, Cambridge, UK) was used to plot out the theoretical frequency response of the band-pass filter. Analyser accepts a circuit description in the form of a Netlist, which is generated internally using a built-in Netlist Editor window. The results from the band-pass filter simulation are shown in Figure 10.6. The maximum gain in the pass band was approximately 1.58, as calculated. The frequency response was 0.48 Hz to 19.4 Hz for -3 dB and was flat over approximately 1.0 Hz to 10.0 Hz.
The frequency response of the band-pass filter was also measured experimentally in the laboratory, and is shown in Figure 10.7. The lower and upper cut-off frequencies of the band-pass filter from the experimental analysis were 0.43 Hz and 20 Hz respectively for -3 dB and the response was flat over approximately 2.0 Hz to 10.0 Hz.

**Figure 10.7**: Experimental frequency response of the band-pass filter
10.2.4 Variable amplification and DC level shifting stage

After the PPG pre-amplifier filter circuit the PPG signal was amplified using a variable amplification circuit (inverting amplifier) with a gain in the range of 0.68 to 100.68 (Figure 10.8a). The final stage comprised a DC level shifting circuit which gave an output signal compatible with the unipolar (0 to 2.5 volts) input range of the analogue-to-digital converter (ADC) (Figure 10.8b).

![Variable amplification and DC level shifting circuit](image)

**Notes**
1. Resistor type: 0.5W High Stability Carbon Film Resistors (tolerance +/- 5%)
2. Potentiometers: Single Turn Carbon Track Potentiometer 20mm, 400mW

**Figure 10.8:** (a) PPG variable amplification circuit; (b) DC level shifting circuit

The DC offset circuit was configured as a summing amplifier (adder) which is a variation of the inverting op-amp configuration. The input AC PPG signal could be offseted by “injecting” a DC voltage at the inverting input of the operational amplifier as shown in the circuit in Figure 10.8b. The overall gain of the *PPG Processing System* 1 could be varied from approximately 1 to 160.

10.2.5 Analogue-to-Digital Converter (ADC)

The PPG signals at the output of the conditioning circuit were sampled and recorded using a 10-bit data acquisition system (type ADC11, Pico Technology Ltd., Cambridge, UK) and a laptop personal computer. The ADC - 11 is an analogue-to-digital converter, which, connects to the printer...
port of an IBM PC compatible computer (Figure 10.9). It plugs directly into the parallel printer port requiring no external power, and taking up no expansion slots. The ADC-11 has 11 analogue input channels and a digital output. Detail specifications for the ADC-11 are shown in Table 10.2.

![Photograph of the analogue-to-digital converter ADC-11](image)

**Figure 10.9:** Photograph of the analogue-to-digital converter ADC-11

**Table 10.3:** Specifications of the analogue-to-digital converter (ADC-11)

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolution</td>
<td>10 bits</td>
</tr>
<tr>
<td>Number of analogue inputs</td>
<td>11</td>
</tr>
<tr>
<td>Analogue input range</td>
<td>0-2.5 V</td>
</tr>
<tr>
<td>Maximum sampling rate</td>
<td>18ksps (33MHz 386/486 PC)</td>
</tr>
<tr>
<td></td>
<td>12ksps (8MHz 286 PC)</td>
</tr>
<tr>
<td>Linearity</td>
<td>±1 LSB</td>
</tr>
<tr>
<td>Accuracy</td>
<td>± 1 %</td>
</tr>
<tr>
<td>Overload protection</td>
<td>± 30 V</td>
</tr>
<tr>
<td>Analogue input impedance</td>
<td>&gt; 200 KΩ</td>
</tr>
<tr>
<td>Number of digital outputs</td>
<td>1 (TTL)</td>
</tr>
<tr>
<td>Digital output impedance</td>
<td>Approx. 1 - 3 KΩ</td>
</tr>
<tr>
<td>Input connector</td>
<td>25 way female D-type</td>
</tr>
<tr>
<td>Connection to computer</td>
<td>25 way male D-type</td>
</tr>
</tbody>
</table>

The infrared AC PPG signals from the oesophagus and the finger were recorded using the *PicoLog* data logger (Pico Technology Ltd., Cambridge, UK). The *PicoLog* data logger is a powerful and flexible program, which enables the user to carry out data logging and some basic analysis. It collects sets of measurements from the ADC-11 analogue-to-digital converter and displays the signals in real time. It can also store the data on disk in a spreadsheet format.
10.2.6 Mechanical Construction

The two identical PPG channels (oesophageal and finger) were developed on a single Eurocard prototyping stripboard (dimensions: 160 mm x 100 mm x 1.6 mm) (RS 435-181) using soldering techniques (Figure 10.10).

![Photograph of the PPG Processing System 1](image)

The Processing system was enclosed in a versatile aluminium case with a steel base and a slide-on top cover offering easy access. A size 5 aluminium case (RS 501-468) (dimensions: 250 mm x 202.8 mm x 64 mm) was used which could easily accommodate the eurocard stripboard horizontally. The front panel (Figure 10.11) of the processing system comprised:

- a power on/off switch (RS 316-989) with a power on illuminating red LED
- two continuously variable amplification control knobs (one for each channel)
- two continuously variable offset control knobs (one for each channel)
- two six-way mini DIN sockets (RS 183-1798) (one for each channel) for connecting the oesophageal and finger PPG probes
- two nickel plated insulated BNC connectors (RS 304-5444) (one for each channel) for the output of analogue infrared AC PPG signals
The back panel of the processing system comprised the power supply of the system (± 9 V). Two non-rechargeable alkaline PP3 batteries (RS 325-8476) were used in this system. Apart from the 2xPP3 battery holder (RS 501-244) the back panel also incorporates 4 mm round insulating sockets allowing the PPG Processing System 1 to be operated from a bench type dual (± 9 V) power supply.

10.3 CLINICAL MEASUREMENTS WITH PPG PROCESSING SYSTEM 1

10.3.1 Healthy Volunteer

The following procedure was adopted in measuring the healthy oesophagus of a male awake volunteer. The airway was prepared with lignocaine spray (10 mg per spray) and benzocaine lozenges each containing 10 mg of benzocaine. Ten minutes was allowed to ensure maximum effect. The nasopharynx was sprayed with lignocaine spray and also left for ten minutes. A French 20 gauge oesophageal transparent stomach tube containing the Oesophageal PPG Probe 1 (see section 9.2) was lubricated on its outside surface with aqueous and lignocaine gel. The tip of the tube was softened with warm water and introduced into the oesophagus via the nasopharynx.
The insertion of the probe into the oesophagus of the awake volunteer was almost intolerable which indicated that the probe could only be applied in anaesthetised patients. After a lot of attempts the oesophageal probe was inserted into the deep oesophagus at a depth of 36 cm from the nose. During the oesophageal PPG measurement a PPG signal was also obtained from a finger using the identical reflection finger probe. Nothing was taken by mouth for three hours following the procedure and there were no complications.

10.3.2 Anaesthetised Patients
Four adult elective surgery patients (urological, gynaecological and general surgery) were recruited to this preliminary pilot study. Following induction of general anaesthesia the oesophageal PPG probe was inserted into a sealed oesophageal stomach tube which was lubricated with aqueous gel. The tube was then inserted through the mouth into the oesophagus under direct vision (Figure 10.12). The stomach tube was advanced into the oesophagus until the end of the probe itself was approximately 30 cm to 35 cm from the upper incisors. The identical reflectance finger probe was placed on the finger of the patient. Simultaneous infrared AC PPG traces from the oesophagus and the finger were recorded for approximately 5 minutes in the anaesthetic room.

Figure 10.12: The oesophageal PPG probe contained within the stomach tube is seen placed in the oesophagus via the mouth (courtesy of M. Savva)
10.4 RESULTS

10.4.1 Healthy Volunteer
A typical section, of the Picolog output as shown on the computer screen, of the infrared AC PPG traces obtained from the deep oesophagus and finger of a healthy volunteer, is presented in Figure 10.13. The electrical characteristics and gain of the PPG signal conditioning circuit for the oesophageal channel and the finger channel were identical for all the measurements. This facilitated realistic comparisons between the two signals. As can be seen from Figure 10.13, the AC PPG amplitudes from the oesophagus are at least a factor of two greater than those obtained from the finger. The oesophageal signal appears to be artefact free with a high signal-to-noise ratio.

![Figure 10.13: Oesophageal and Finger infrared AC PPG signals from a healthy oesophagus](image)

Figure 10.13: Oesophageal and Finger infrared AC PPG signals from a healthy oesophagus
10.4.2 Anaesthetised Patients

Figure 10.14 shows typical infrared AC PPG traces obtained from the deep oesophagus and finger of an anaesthetised patient with the mechanical ventilator temporarily switched off for 10 seconds. The oesophageal PPG signals appear to be of good quality with high signal-to-noise ratio. The AC PPG amplitudes from the oesophagus are at least a factor of two greater than those obtained simultaneously from the finger.

![Photoplethysmographic Signals](image)

**Figure 10.14:** Oesophageal (top) and finger (bottom) infrared AC PPG signals from an anaesthetised patient

When the mechanical ventilator was switched on, the oesophageal PPG traces were modulated by an artefact synchronous with the approximately 5 second period of the ventilator, as shown in Figure 10.15. The magnitude of the artefact is of the order of 400% of the PPG peak-to-peak amplitude but varies considerably from patient to patient.
10.5 DISCUSSION AND CONCLUSIONS

This chapter described the investigation of infrared AC PPG signals from the human deep oesophagus. These measurements appear to be the first PPG measurements in the human oesophagus.

The developed circuitry has been evaluated and it has been shown that good quality PPG signals with high signal-to-noise ratio and large amplitudes can be measured in the human deep oesophagus. Simultaneous infrared AC PPG signals from the identical reflectance probes in the oesophagus and on a finger were recorded and compared. On the basis of the data obtained, it is concluded that the oesophageal PPG signals are significantly larger than the finger PPG signals for both the healthy volunteer and the anaesthetised patients. In this study no detailed data analysis was performed since the investigation was primarily aiming to test the operability of the various electronic circuits in detecting, processing and estimating the amplitude of
PPG signals available in the oesophagus and displaying them on the screen of the laptop computer.

Despite the successful operation of the prototype *PPG Processing System 1*, there were limitations that needed to be addressed. The first limitation was concerning safety issues. Even though the PPG system and the laptop computer were battery operated, therefore any change of electrical hazard was minimal, there was a possibility of operating the laptop computer on its mains adapter (unintentionally or intentionally by a different user), which could cause real hazard to the patient. This was possible, since there were no electrical isolation between the input side (patient) and the output side (laptop) of the two channels (oesophageal and finger) in the prototype *PPG Processing System 1*. A second limitation was that the PPG system 1 did not provide precise gain control of the two channels (oesophageal and finger), therefore accurate quantitative studies between the oesophageal and finger PPGs were not possible. A third limitation was that the system operated only with one wavelength (infrared). To be able to estimate arterial oxygen saturation in the oesophagus by pulse oximetry, adequate PPG signals at both red and infrared wavelengths are required, (see section 5.4) which the *PPG Processing System 1* did not offer.

To eliminate the limitations of the *PPG Processing System 1*, as discussed above, a new processing system (*PPG Processing System 2*) was designed and constructed. This new system not only overcame the technical limitations of the *PPG Processing System 1* but also allowed more detailed quantifying PPG studies at both red and infrared wavelengths in the oesophagus in a larger group of anaesthetised patients. This new *PPG Processing System 2* is the subject of the next chapter.
CHAPTER 11

PPG Processing System 2

11.1 INTRODUCTION

To eliminate the limitations of the PPG Processing System 1, which operates only at one wavelength (infrared) and does not provide any electrical isolation between the patient side and the monitoring side, a new PPG processing system (PPG Processing System 2) was developed. A basic block diagram of the PPG Processing System 2 is shown in Figure 11.1.

![Figure 11.1: Basic block diagram of the red and infrared PPG Processing System 2](Image)

The new PPG Processing System 2 operates at two wavelengths, red and infrared, and was developed to detect, pre-process, sample, record and display the red and infrared AC PPG output signals on a laptop personal
computer. It consisted of two identical electrically isolated PPG processing channels (see Figure 11.1), which allowed the monitoring of oesophageal and finger PPG signals. The *PPG Processing System 2* was used to perform detail investigations of AC PPG signals, at red and infrared wavelengths, in the mid-third of the oesophagus of anaesthetised ASA 1 patients. The technical details of the construction of this processing system as well as the performance evaluation tests at all stages of the system's design will be the subject of this chapter. The clinical trial methods and the results from all the red and infrared PPG measurements using this *PPG Processing System 2* will also be presented and discussed.

### 11.2 INSTRUMENTATION

A detail block diagram of one of the two identical PPG channels (oesophageal) is shown in Figure 11.2.

![Figure 11.2: Detail block diagram of one of the two identical channels of the red and infrared *PPG Processing System 2*](image)

The new *Processing System 2* allowed the monitoring of oesophageal and finger AC PPG signals by using the *Oesophageal PPG Probe II* (see Figure 9.10) and the identical reflectance finger probe (see Figure 9.24). Red and infrared AC PPG measurements were made consecutively. At a given time,
the same wavelength was used in the finger and oesophageal channels; mechanical switches effected the changeover between red and infrared wavelengths in each channel (Figure 11.3).

11.2.1 Input Side Circuitry

The red (R) and infrared (IR) emitters in each channel (oesophageal and finger) were driven by a constant current source, identical to the one described in the previous chapter, section 10.2.1 (Figure 10.3). A mechanical switch, single pole double throw (SPDT), was used at the output of the current source to turn the red and infrared emitters on/off consecutively (Figure 11.3). The red and infrared emitter output intensities depended on the forward current, which was again maintained at 40 mA in each case. This operating current was well below the maximum rated value of 75 mA for the emitters.

The photodetector detected the energy backscattered by the tissue and gave an output current proportional to the detected light intensity. The output current from the photodetector was converted into a signal voltage using a transresistance amplifier identical to the one used for the PPG Processing System 1 (Figure 10.4). The output of the transresistance amplifier was connected to a band-pass filter identical to the one described in the previous chapter (Figure 10.5).
11.2.2 Isolation and Output Side Circuitry

An analogue isolation amplifier was used to electrically isolate the input side (patient side) from the output side in each channel (Figure 11.2). The input sides of the two channels were physically isolated from each other (see Figure 11.1). The isolation amplifier used in each channel was a Hewlett Packard type HCPL7820. This was an inexpensive 8-pin opto-electronic device. The HCPL7820 high common rejection ratio (CMR) isolation amplifier consisted of an analogue-to-digital converter (ADC) optically coupled to an integrated output digital-to-analogue converter (DAC). Table 11.1 presents more detailed specifications (electrical and package) of the HCPL7820 isolation amplifier. The output of the band-pass filter (see Figure 10.5) was transmitted across the isolation barrier via the isolation amplifier. The differential output of the isolation amplifier was converted to a single ended signal by a differential amplifier with a gain of one (Figure 11.4).

Table 11.1: Technical specifications of the Analogue Isolation Amplifier HCPL7820

<table>
<thead>
<tr>
<th>Description</th>
<th>Symbol</th>
<th>Characteristic &amp; units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supply Voltages</td>
<td>$V_{DD1}, V_{DD2}$</td>
<td>5.0 V (typ)</td>
</tr>
<tr>
<td>Input Voltage</td>
<td>$V_{IN+}, V_{IN-}$</td>
<td>-200 mV to +200 mV</td>
</tr>
<tr>
<td>Operating Temperature</td>
<td>$TA$</td>
<td>-40 to 100 °C</td>
</tr>
<tr>
<td>Insulation Resistance at VIO=500 V</td>
<td>$R_s$</td>
<td>$\geq 1 \times 10^{12}$ Ω</td>
</tr>
<tr>
<td>Input Offset Voltage</td>
<td>$V_{os}$</td>
<td>0.45 mV</td>
</tr>
<tr>
<td>Gain: HCPL7820</td>
<td>$G$</td>
<td>8.0</td>
</tr>
<tr>
<td>Input DC Common-Mode Rejection Ratio</td>
<td>CMRR$_{IN}$</td>
<td>52 dB</td>
</tr>
<tr>
<td>Average Input Bias Current</td>
<td>$I_{IN}$</td>
<td>-1 μA</td>
</tr>
<tr>
<td>Input Supply Current</td>
<td>$I_{DD1}$</td>
<td>11.1 mA</td>
</tr>
<tr>
<td>Small-Signal Bandwidth (-3dB)</td>
<td>$f_{3dB}$</td>
<td>200 kHz</td>
</tr>
<tr>
<td>Output Supply Current</td>
<td>$I_{DD2}$</td>
<td>10.0</td>
</tr>
</tbody>
</table>

Package Specifications
The signal from the output of the differential output was then passed to a precision variable amplification stage with a gain in the range of 0.08 to 4.0 (Figure 11.4). The precision switching variable amplification stage was constructed using a rotary switch kit (make before break, 1 pole, 11 way; RS 352-244). Metal film precision resistors (0.125 W, 0.1%) were used for the eleven amplification values. Table 11.2 summarises all the amplification stages for the PPG Processing System 2 including the 11 positions of the variable amplification stage.

<table>
<thead>
<tr>
<th>Amplif. Switch Position</th>
<th>Band-pass Filter Gain</th>
<th>Opto-Isolator Gain (fixed)</th>
<th>Differential Amplifier Gain</th>
<th>Variable Amplification Gain</th>
<th>Overall Gain</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>0.08</td>
<td>1.0</td>
</tr>
<tr>
<td>2</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>0.2</td>
<td>2.5</td>
</tr>
<tr>
<td>3</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>0.4</td>
<td>5.0</td>
</tr>
<tr>
<td>4</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>0.8</td>
<td>10.0</td>
</tr>
<tr>
<td>5</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>1.0</td>
<td>12.5</td>
</tr>
<tr>
<td>6</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>1.5</td>
<td>18.8</td>
</tr>
<tr>
<td>7</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>2.0</td>
<td>25.0</td>
</tr>
<tr>
<td>8</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>2.5</td>
<td>31.3</td>
</tr>
<tr>
<td>9</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>3.0</td>
<td>37.5</td>
</tr>
<tr>
<td>10</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>3.5</td>
<td>43.8</td>
</tr>
<tr>
<td>11</td>
<td>1.58</td>
<td>8</td>
<td>1</td>
<td>4.0</td>
<td>50.0</td>
</tr>
</tbody>
</table>

The final stage of the processing system comprised a DC level shifting circuit which gave an output signal compatible with the unipolar input range (0 to 2.5 V) of the analogue-to-digital converter (ADC-11) (Figure 11.4).

The AC PPG signals (red or infrared) at the output of the level shifting circuit were sampled using the 10-bit analogue-to-digital converter (ADC-11) connected to the parallel port of a laptop personal computer. The signals recorded and displayed on the laptop computer corresponded to infrared and red AC PPGs from the mid-third of the oesophagus and the finger of anaesthetised patients.
Notes
1. Resistor type: 0.5W High Stability Carbon Film Resistors (tolerance +/- 5%)
2. Capacitors: polyester mini case capacitors (capacitance tolerance +/- 5%)
11.3 MECHANICAL CONSTRUCTION

11.3.1 Electronic Circuits
The two isolated processing channels (oesophageal and finger) were constructed on a Eurocard prototyping stripboard (dimensions: 160 mm x 100 mm x 1.6 mm) (RS 435-181) using soldering techniques. Figure 11.5 shows a photograph of the PPG Processing System 2. The isolation barrier between the input side and the output side is indicated (see Figure 11.5).

![Figure 11.5: Photograph of the two isolated processing channels](image)

11.3.2 Power Supply
Three pairs of PP3 (±9 V) batteries, identical to the ones used in the PPG Processing System 1, were used to power the PPG Processing System 2. One pair was used to power the common output sides of the two processing channels (oesophageal and finger). The other two pairs were used to power the input sides of each channel (oesophageal and finger) independently.

11.3.3 Enclosure
The PPG Processing System 2 was enclosed in a Eurocard black case consisting of moulded ABS top and bottom covers screwed to a chassis assembly with brushed aluminium front and back panels (RS 506-996) (dimensions W, H, D: 247 mm x 102 mm x 220 mm) (Figure 11.6).

![Figure 11.6: Photograph of the complete PPG Processing System 2](image)

The front panel (Figure 11.7) of the processing system comprised:

- two power on/off switches (one for input oesophageal channel side, one for the input finger channel side) with a power on indicating illuminating red LED
- two wavelength selection toggle switches (SPDT; RS 352-761)
- two rotary variable amplification selector switches (one for each channel)
- two continuously variable offset control knobs (one for each channel)
- two six-way mini DIN sockets (RS 183-1798) (one for each channel) for connecting the oesophageal and finger PPG probes

![Figure 11.7: Photograph of the front panel of the PPG Processing System 2](image)

The back panel of the Eurocard black case, housed the three (2XPP3) battery holders, two nickel plated insulated BNC connectors (RS 304-5444) (one for each channel) for the output of the infrared and red AC PPG signals, and a power on/off switch for the common output side.

### 11.4 PERFORMANCE EVALUATION

To minimise the risk to the patient from any form of electrical hazard, in the case of accidental or unintended mains power up of the laptop computer, the input (patient side) and output (monitoring side) circuits of the system were isolated using the two HCPL7820 analogue isolation amplifiers (Figure 11.1). Furthermore, the *Oesophageal PPG Probe II* was isolated from the tissue by the plastic wall (1 mm) of the stomach tube and the identical reflectance finger probe was insulated from the finger by a transparent plastic sheath (thickness of 0.05 mm). Also, the input sides of the two channels were physically isolated from each other (Figure 11.1).

#### 11.4.1 PPG Processing System 2 Electrical Safety Test
The *PPG Processing System 2* was electrically safety tested using a Rigel Safety Tester Model 233. The Model 233 Electrical Safety Tester is a portable unit housed in an attached case. Inside the case, the tester’s electronics circuits are contained within a separate metal enclosure.

**Insulation Resistance Test**

The insulation resistance between the patient applied parts (probes) and the output circuitry was measured for the oesophageal and finger channels. A diagram of the test principle is shown in Figure 11.8. This test is performed for all class 1 instruments with an applied part. i.e. Class1, Type BF and CF.

![Diagram for the Insulation Resistance Test](image)

**Figure 11.8:** Diagram for the Insulation Resistance Test

In this test, the input side at the mini DIN input socket of one of the two channels (oesophageal or finger) of the *PPG Processing System 2* was connected to the 500 VDC source from the Tester (Figure 11.9). The 500 VDC was applied consecutively during the test to all individual pins of DIN socket. The output side (BNC connector) of the *PPG Processing System 2* was connected to the earth terminal on the Tester (Figure 11.9). The insulation resistance test was repeated for all combinations of inputs and outputs for both, oesophageal and finger channels.
Figure 11.9: PPG Processing System 2 connected on the Safety Tester for the Insulation Resistance Test

Mains on Applied Part Test

The current flowing across the isolation barrier with mains voltage on the applied parts, and with mains voltage at the output terminals was measured for both channels. A diagram of the test principle is shown in Figure 11.10.

Figure 11.10: Diagram for the Mains on Applied Parts Safety Test

In this test, the input side at the mini DIN input socket of one of the two channels (oesophageal or finger) of the PPG Processing System 2 was connected to the mains via the Safety Tester (Figure 11.11). The output side
(BNC connector) of the *PPG Processing System 2* was connected to the earth terminal of the Safety Tester (Figure 11.11). Mains was applied to the five input pins of the DIN socket, one pin at a time with the Safety Tester Programme button set at BF, CLASS 1. Mains on applied parts test was performed for all combinations of inputs and outputs for both channels. The same test was then repeated with the Safety Programme button set at CF, CLASS 1.

![Figure 11.11: PPG Processing System 2 connected on the Safety Tester for the Mains on Applied Parts Test](image)

### 11.4.2 Results of *PPG Processing System 2* Electrical Safety Test

In the safety tests described above the insulation resistance between the input and output circuits of the *PPG Processing System 2* was too large to register on the tester meter, but was well in excess of 100 MΩ for each channel. The test with mains on the patient applied parts and on the output gave leakage currents across the isolation barrier of less than 9 μA in all cases. This is well below the 50 μA current limit for Class 1 and 2 type CF instruments.

### 11.5 CLINICAL MEASUREMENTS WITH THE *PPG PROCESSING SYSTEM 2*

Twenty adult patients (two male, eighteen female, average age (± SD): 43 ± 16.6) who were to undergo tracheal intubation as a routine part of
general anaesthesia for elective urological, gynaecological and general surgery were studied. Table 11.3 summarises the details of all twenty patients studied.

Table 11.3: Patient Details

<table>
<thead>
<tr>
<th>Patient</th>
<th>Sex</th>
<th>Age</th>
<th>Type of operation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Female</td>
<td>36</td>
<td>Laparoscopy / Hysterectomy</td>
</tr>
<tr>
<td>2</td>
<td>Female</td>
<td>43</td>
<td>Stapedectomy</td>
</tr>
<tr>
<td>3</td>
<td>Female</td>
<td>32</td>
<td>Tonsillectomy</td>
</tr>
<tr>
<td>4</td>
<td>Female</td>
<td>25</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>5</td>
<td>Female</td>
<td>22</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>6</td>
<td>Female</td>
<td>72</td>
<td>Total abdominal Hysterectomy / Bilateral Salphingo-ophorectomy (T.A.H./B.S.O.)</td>
</tr>
<tr>
<td>7</td>
<td>Female</td>
<td>30</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>8</td>
<td>Female</td>
<td>28</td>
<td>Tonsillectomy</td>
</tr>
<tr>
<td>9</td>
<td>Female</td>
<td>45</td>
<td>T.A.H./B.S.O.</td>
</tr>
<tr>
<td>10</td>
<td>Female</td>
<td>27</td>
<td>Laparoscopy / Hysterectomy</td>
</tr>
<tr>
<td>11</td>
<td>Female</td>
<td>25</td>
<td>Laparoscopic Ovarian Cystectomy</td>
</tr>
<tr>
<td>12</td>
<td>Male</td>
<td>60</td>
<td>Nasal Polypectomy</td>
</tr>
<tr>
<td>13</td>
<td>Female</td>
<td>75</td>
<td>T.A.H./B.S.O.</td>
</tr>
<tr>
<td>14</td>
<td>Female</td>
<td>33</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>15</td>
<td>Female</td>
<td>31</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>16</td>
<td>Female</td>
<td>47</td>
<td>Vaginal Hysterectomy</td>
</tr>
<tr>
<td>17</td>
<td>Female</td>
<td>51</td>
<td>Umbilical Hernia Repair</td>
</tr>
<tr>
<td>18</td>
<td>Male</td>
<td>62</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>19</td>
<td>Female</td>
<td>60</td>
<td>Laparoscopic Cholecystectomy</td>
</tr>
<tr>
<td>20</td>
<td>Female</td>
<td>56</td>
<td>Varicose Veins</td>
</tr>
</tbody>
</table>

The *Oesophageal PPG Probe II* (see Figure 9.10) was inserted into a sealed disposable transparent, French 20, stomach tube. A new sealed stomach tube was used for each patient. The exterior surface of the tube was lubricated with aqueous gel prior to insertion through the mouth into the oesophagus. General anaesthesia was induced with intravenous propofol; a muscle relaxant (atracurium or vecuronium) was given, and the trachea was intubated with an endotracheal tube. The stomach tube was advanced into the oesophagus under direct vision until the end of the *Oesophageal Probe II*, distal to the multicore cable, was between 25 cm and 30 cm from the upper incisors. The lungs were mechanically ventilated and anaesthesia was maintained using nitrous oxide (70%, in oxygen) and isoflurane (approximately 1.5% inspired concentration). The identical reflectance finger probe (see Figure 9.24) was also placed on the index finger of the patient.
Simultaneous AC PPG traces from the oesophagus and the finger were recorded for approximately 15 minutes.

### 11.6 DATA ANALYSIS AND STATISTICS

Data files recorded by the data logging software (*Picolog*) were analysed offline. The peak-to-peak amplitudes of the oesophageal and finger AC PPG signals, at red or infrared wavelength, for each patient were measured on printouts from *Picolog* using a ruler, and the means, standard deviations (SD), and standard errors (SE) were calculated. The statistical significance of the differences between the oesophageal and finger PPG amplitudes was assessed by performing a One Way Analysis of Variance (ANOVA) for both infrared and red wavelengths using *SigmaStat* (SPSS Ltd, Birmingham, UK).

### 11.7 RESULTS

Figure 11.12 shows typical red and infrared AC PPG traces obtained from the middle third of the oesophagus and the finger of an anaesthetised patient with the mechanical ventilator temporarily switched off for approximately twenty seconds. The oesophageal PPGs appear to be of good quality with large amplitudes at both red and infrared wavelengths. The electrical characteristics and gain of the oesophageal and finger channels were identical for all measurements. The large deviations at the centre of each trace are due to the switching artefact when changing between red and infrared wavelengths. When the mechanical ventilator was switched on, the oesophageal PPG traces were modulated by an artefact synchronous with the approximately 5 second period of the ventilator, as shown in Figure 11.13. The magnitude of the artefact was of the order of 30% of the peak-to-peak PPG amplitude but varied considerably from patient to patient. The finger PPG traces were also modulated by a very small artefact, especially at the infrared wavelength, again synchronous with the period of the ventilator (see Figure 11.13).
Figure 11.12 Typical AC PPG traces for the red and infrared wavelengths from the middle third of the oesophagus and the finger of an anaesthetised patient with the mechanical ventilator temporarily switched off.

Figure 11.13: AC PPG traces for the red and infrared wavelengths from the middle third of the oesophagus and the finger of an anaesthetised patient with the mechanical ventilator switched on.
Table 11.4 shows the means, standard deviations (SD), and standard errors (SE) of the peak-to-peak amplitudes of the red and infrared AC PPGs for the oesophagus and the finger calculated at the output of the differential amplifier, before the variable amplification and level shifting stage (see Figure 11.4). These results are also illustrated in Figure 11.14. In the first three patients studied no PPG measurements were possible at the red wavelength due to technical problems (the red wavelength switching mechanism was wired wrong for both channels). Finger PPG measurements at either wavelength were not possible on the eighth patient due to a complete failure of the finger probe (the DIN plug of the finger probe was damaged accidentally in the anaesthetic room immediately prior to the study).

Table 11.4. Means, standard deviations (SD) and standard errors (SE) of the peak-to-peak amplitudes of the infrared and red AC PPGs for the oesophagus and the finger

<table>
<thead>
<tr>
<th>Patient No</th>
<th>Oesophageal IR (mV)</th>
<th>Oesophageal RED (mV)</th>
<th>Finger IR (mV)</th>
<th>Finger RED (mV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>15.8</td>
<td>---</td>
<td>7.2</td>
<td>---</td>
</tr>
<tr>
<td>2</td>
<td>31.5</td>
<td>---</td>
<td>12.8</td>
<td>---</td>
</tr>
<tr>
<td>3</td>
<td>30.4</td>
<td>---</td>
<td>17.9</td>
<td>---</td>
</tr>
<tr>
<td>4</td>
<td>70.5</td>
<td>48.3</td>
<td>15.4</td>
<td>9.7</td>
</tr>
<tr>
<td>5</td>
<td>48.2</td>
<td>20.2</td>
<td>15.5</td>
<td>7.9</td>
</tr>
<tr>
<td>6</td>
<td>32.2</td>
<td>10.7</td>
<td>23.6</td>
<td>10.0</td>
</tr>
<tr>
<td>7</td>
<td>29.6</td>
<td>11.6</td>
<td>14.2</td>
<td>4.5</td>
</tr>
<tr>
<td>8</td>
<td>37.9</td>
<td>9.0</td>
<td>---</td>
<td>---</td>
</tr>
<tr>
<td>9</td>
<td>19.2</td>
<td>6.4</td>
<td>14.8</td>
<td>4.0</td>
</tr>
<tr>
<td>10</td>
<td>20.8</td>
<td>8.9</td>
<td>4.7</td>
<td>3.2</td>
</tr>
<tr>
<td>11</td>
<td>25.5</td>
<td>16.1</td>
<td>5.0</td>
<td>2.5</td>
</tr>
<tr>
<td>12</td>
<td>19.7</td>
<td>8.9</td>
<td>2.1</td>
<td>0.9</td>
</tr>
<tr>
<td>13</td>
<td>29.1</td>
<td>9.7</td>
<td>37.8</td>
<td>14.5</td>
</tr>
<tr>
<td>14</td>
<td>27.0</td>
<td>12.0</td>
<td>12.9</td>
<td>3.3</td>
</tr>
<tr>
<td>15</td>
<td>26.9</td>
<td>10.2</td>
<td>9.3</td>
<td>6.7</td>
</tr>
<tr>
<td>16</td>
<td>29.6</td>
<td>15.4</td>
<td>47.2</td>
<td>17.9</td>
</tr>
<tr>
<td>17</td>
<td>23.4</td>
<td>5.3</td>
<td>23.7</td>
<td>5.6</td>
</tr>
<tr>
<td>18</td>
<td>17.9</td>
<td>9.5</td>
<td>12.8</td>
<td>3.3</td>
</tr>
<tr>
<td>19</td>
<td>9.4</td>
<td>7.4</td>
<td>1.8</td>
<td>1.5</td>
</tr>
<tr>
<td>20</td>
<td>40.1</td>
<td>13.9</td>
<td>35.6</td>
<td>9.1</td>
</tr>
</tbody>
</table>

Size (n) 20 17 19 16

Mean 29.2 13.2 16.5 6.5

Std. Dev 13.2 9.8 12.4 4.8

Std. Err 2.9 2.4 2.8 1.2
Figure 11.14: Mean peak-to-peak AC PPG signals (± SD) at red and infrared wavelengths from the mid-third of the oesophagus and the finger.

The ratio of the oesophageal to finger AC PPG amplitudes was calculated for the infrared and red wavelengths for each patient (Table 11.5).

Table 11.5. Means, standard deviations and standard error of the ratio of the oesophageal to finger AC PPG amplitudes at infrared and red wavelengths

<table>
<thead>
<tr>
<th>Patient No</th>
<th>Oesophageal IR / Finger IR</th>
<th>Oesophageal RED / Finger RED</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.2</td>
<td>---</td>
</tr>
<tr>
<td>2</td>
<td>2.6</td>
<td>---</td>
</tr>
<tr>
<td>3</td>
<td>1.7</td>
<td>---</td>
</tr>
<tr>
<td>4</td>
<td>4.6</td>
<td>5.0</td>
</tr>
<tr>
<td>5</td>
<td>3.1</td>
<td>2.6</td>
</tr>
<tr>
<td>6</td>
<td>1.4</td>
<td>1.1</td>
</tr>
<tr>
<td>7</td>
<td>2.1</td>
<td>2.6</td>
</tr>
<tr>
<td>8</td>
<td>---</td>
<td>---</td>
</tr>
<tr>
<td>9</td>
<td>1.3</td>
<td>1.6</td>
</tr>
<tr>
<td>10</td>
<td>4.4</td>
<td>2.8</td>
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<tr>
<td>11</td>
<td>5.1</td>
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<tr>
<td>12</td>
<td>9.2</td>
<td>9.4</td>
</tr>
<tr>
<td>13</td>
<td>0.8</td>
<td>0.7</td>
</tr>
<tr>
<td>14</td>
<td>2.1</td>
<td>3.6</td>
</tr>
<tr>
<td>15</td>
<td>2.9</td>
<td>1.5</td>
</tr>
<tr>
<td>16</td>
<td>0.6</td>
<td>0.9</td>
</tr>
<tr>
<td>17</td>
<td>1.0</td>
<td>0.9</td>
</tr>
<tr>
<td>18</td>
<td>1.4</td>
<td>2.9</td>
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<td>5.1</td>
</tr>
<tr>
<td>20</td>
<td>1.1</td>
<td>1.5</td>
</tr>
</tbody>
</table>

Size (n) 19 16
Mean 2.86 3.13
Std. Dev 2.13 2.43
Std. Err 0.50 0.63
Before commencing into any statistical analysis of the data in Table 11.4, a normality test on the data was performed. Normality refers to the assumption that a population follows a standard, “bell” shaped Gaussian distribution, also known as a “normal” distribution. Normality was tested using the Kolmogorov-Smirnov test provided by the statistical software package *Sigmastat*. *Sigmastat* automatically performs a normality test when running a statistical procedure. The results of the normality test give the Kolmogorov-Smirnov (K-S) distances and p values (defined below) computed for each column of Table 11.4, and whether or not the data in each column selected passed or failed the normality test. Table 11.6 shows the results of the normality test performed on the data of Table 11.4.

<table>
<thead>
<tr>
<th>Normality Test (Kolmogorov-Smirnov)</th>
<th>Tuesday, October 31, 2000, 18:04:27</th>
</tr>
</thead>
</table>

**Data source:** Data 1 in Notebook

<table>
<thead>
<tr>
<th>Oes IR: K-S Dist. = 0.211 P = 0.020 Failed</th>
</tr>
</thead>
<tbody>
<tr>
<td>Oes R: K-S Dist. = 0.264 P = 0.003 Failed</td>
</tr>
<tr>
<td>Fing IR: K-S Dist. = 0.218 P = 0.018 Failed</td>
</tr>
<tr>
<td>Fing R: K-S Dist. = 0.166 P &gt; 0.200 Passed</td>
</tr>
</tbody>
</table>

*A test that fails indicates that the data varies significantly from the pattern expected if the data were drawn from a population with a normal distribution.*

*A test that passes indicates that the data matches the pattern expected if the data were drawn from a population with a normal distribution.*

**K-S Distance:** The Kolmogorov-Smirnov distance is the maximum cumulative distance between the histogram of the data and the gaussian distribution curve of the data.

**P values:** The p values represent the observations for normality using the Kolmogorov-Smirnov test. If the p value computed by the test is greater than the p value set in the test (0.05), the data can be considered normal.

The results from the normality test can be represented graphically by either a histogram of the residuals, or a normal probability plot of the residuals. For the data in Table 11.4 normal probability plots were produced. A typical graphical example for normally distributed data and non-normally distributed data using the normal probability plot of residuals is shown in Figure 11.15a and Figure 11.15b respectively. The normality probability plot graphs the
frequency of the raw residuals. Plots with residuals that fall along the gaussian curve indicate that the data were taken from a normally distributed population (Figure 11.15a). The X-axis is a linear scale representing the residual values. The Y-axis is a probability scale representing the cumulative frequency of the residuals.

Figure 11.15: Normal probability plot of the residuals; (a) Normally distributed data (Finger Red) (b) Not-normally distributed data (Oesophageal Red)

To see if there was any significant difference between the mean oesophageal and mean finger PPG amplitudes at the two wavelengths, a
Kruskal-Wallis One Way Analysis of Variance on Ranks was performed. A non-parametric test was used, as most of the data were not normally distributed. A significant difference was found between the groups p<0.001. To isolate the groups that differed an all pairwise multiple comparison procedure (Dunn’s method) was applied. Dunn’s test is used for ANOVA on Ranks when the sample sizes in the different treatment groups are different (Sigmasat, Statistical Software User’s Manual). There were statistically significant differences between the PPG amplitudes in the mid-oesophagus and the PPG amplitudes at the finger at the infrared wavelength. There were no significant differences between the PPG amplitudes in the mid-oesophagus and the PPG amplitudes at the finger at the red wavelength. Table 11.7 summarises all the results from the ANOVA test.

**Table 11.7:** Results of One Way Analysis of Variance Test between the infrared and red oesophageal at the mid-third of the oesophagus and finger AC PPG amplitudes (n = 20)

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Missing</th>
<th>Median</th>
</tr>
</thead>
<tbody>
<tr>
<td>Oes IR</td>
<td>20</td>
<td>0</td>
<td>28.050</td>
</tr>
<tr>
<td>Fing IR</td>
<td>20</td>
<td>1</td>
<td>14.200</td>
</tr>
<tr>
<td>Oes RED</td>
<td>20</td>
<td>3</td>
<td>10.200</td>
</tr>
<tr>
<td>Fing RED</td>
<td>20</td>
<td>4</td>
<td>5.050</td>
</tr>
</tbody>
</table>

**All Pairwise Multiple Comparison Procedures (Dunn’s Method):**

<table>
<thead>
<tr>
<th>Comparison</th>
<th>P&lt;0.05</th>
</tr>
</thead>
<tbody>
<tr>
<td>Oes IR vs Fing RED</td>
<td>Yes</td>
</tr>
<tr>
<td>Oes IR vs Oes RED</td>
<td>Yes</td>
</tr>
<tr>
<td>Oes IR vs Fing IR</td>
<td>Yes</td>
</tr>
<tr>
<td>Fing IR vs Fing RED</td>
<td>Yes</td>
</tr>
<tr>
<td>Fing IR vs Oes RED</td>
<td>No (NS)</td>
</tr>
<tr>
<td>Oes RED vs Fing RED</td>
<td>No (NS)</td>
</tr>
</tbody>
</table>

**11.8. DISCUSSION AND CONCLUSIONS**

The objective of this stage was to develop a new processing system which will allow PPG measurements in the middle third of the oesophagus at red
and infrared wavelengths. In this system the red and infrared AC PPG measurements were made consecutively using mechanical switches to effect the changeover between red and infrared wavelengths. Using the new isolated *PPG Processing System 2* and the reflectance *Oesophageal PPG Probe II* and finger PPG probe it has been demonstrated that measurable PPG signals with high signal-to-noise ratios at both infrared and red wavelengths can be obtained from the mid-third of the oesophagus. The amplitudes of the oesophageal PPGs were on average approximately three times larger than those obtained simultaneously from a finger for both wavelengths (2.86 ± 2.13 for the infrared and 3.13 ± 2.43 for the red) although there was considerable variability. The PPG signals obtained from the mid-third of the oesophagus seemed to be less affected by the ventilator artefact (compare Figures 10.17 with 11.13). The magnitude of the artefact from the deep oesophagus (see Figure 10.17) and from the mid-third of the oesophagus (see Figure 11.13) was of the order of 400% and 30% of the PPG peak-to-peak amplitude respectively. This may suggest that as the monitoring depth is further away from the diaphragm (deep oesophagus) the modulation on the PPG traces caused by the ventilator becomes smaller.

The red (655 nm) and infrared (880 nm) wavelengths used in this processing system were suitable for pulse oximetry and in principle it would be feasible to estimate arterial blood oxygen saturation (SpO₂) values in the mid-oesophagus. However, it was not possible to perform preliminary manual calculations to estimate arterial blood oxygen saturation with this system because: (a) the red and infrared AC PPG signals were not recorded simultaneously and (b) the *PPG Processing System 2* was not design to record red and infrared DC PPGs simultaneously. Measurements of SpO₂ can only be performed if the AC and DC PPG signals at both wavelengths are recorded simultaneously. This can be done when the emitters, red and infrared, are illuminated alternately using time multiplexing techniques. The photodiode signal representing light from both red and infrared emitters in sequence is then separated into red and infrared PPGs using a demultiplexer or demodulator. The value of blood oxygen saturation can then
be derived by calculating the ratio (R) of the quotient of the simultaneously recorded AC and DC PPG amplitudes at the red (655 nm) and infrared (880 nm) wavelengths (see section 5.7). The ratio (R) could then be used to compute the arterial oxygen saturation using an empirically derived calibration curve.

To overcome the limitations of the PPG Processing System 2, as discussed above, a new processing system (PPG Processing System 3) was designed and constructed. This new system not only overcame the technical limitations of the PPG Processing System 2 but also allowed more detailed quantifying PPG studies at both red and infrared wavelengths, recorded simultaneously, at various depths within the oesophagus of anaesthetised patients. This new PPG Processing System 3 is the subject of the next chapter.
12.1 INTRODUCTION
To eliminate the limitations of the PPG Processing System 2, in which red and infrared AC PPG measurements (no DC PPGs) were made consecutively (not simultaneously) using mechanical switches to effect the changeover between red and infrared wavelengths, a new PPG Processing System 3 was developed. This new electrically isolated oesophageal PPG time-multiplexed system 3 was developed to detect, pre-process, sample, record and display simultaneously the red and infrared AC and DC PPG output signals on a laptop personal computer. Furthermore, this new system incorporated a 3-lead ECG channel, which was used as a timing reference for the oesophageal PPG signals. ECG traces were displayed simultaneously alongside the red and infrared AC and DC PPGs on the laptop computer. A detailed block diagram of the system is shown in Figure 12.1.

The PPG Processing System 3 was used to perform detail investigations of AC and DC PPG signals at five oesophageal depths (15 cm, 20 cm, 25 cm, 30 cm, and 35 cm) at red and infrared wavelengths, on anaesthetised ASA 1 patients, to determine an optimal position for reliable SpO₂ monitoring. The technical details of the construction of this processing system, as well as the performance evaluation tests at all stages of the system’s design, will be the subject of this chapter. The clinical methods and the results from all PPG measurements will also be presented and discussed.
12.2 INPUT SIDE CIRCUITRY OF THE OESOPHAGEAL PPG CHANNEL

12.2.1 Master Clock and Timing Generator Circuit

This section of the processing circuit (see Figure 12.1) was used to generate the timing signals for controlling the switching (ON/OFF) of the red and infrared emitters (discussed in the next section). These timing control signals were also used for synchronising the demultiplexer, for separating the mixed (red and infrared) PPG signals at the output of the photodetector current-to-voltage (I-V) amplifier into red and infrared PPG signals (as discussed later in the chapter). A block diagram illustrating the major circuit sections of the master clock and timing generator circuit is shown in Figure 12.2.
Figure 12.2: Block diagram of the Master Clock and Timing Generator circuit

Master Clock

A HA7210 crystal oscillator was used as the master clock to obtain the first running frequency of 2.4576 MHz (Figure 12.2). The HA7210 is a Pierce oscillator optimised for very low power consumption (5 μA @ 32 kHz), requiring no external components except a bypass capacitor and a series mode crystal (Figure 12.3).

Figure 12.3: Schematic Diagram of the Master Clock and Timing Generator Circuit
The 2.4576 MHz clock from the crystal oscillator was fed into a frequency divider. The divider used was the MC14536B 24-stage frequency divider (Figure 12.3) set-up to send out a pulse every 3.33 ms (300 Hz). This frequency was used by the 4-bit shift ring counter on the next stage to produce the timing control signals Q0 (Red) and Q2 (IR) (Figure 12.3). The frequency divider divided the frequency of the crystal oscillator (2.4576 MHz) by $2^{13}=8192$, therefore resulting into a running frequency of 300 Hz. The 24-stage frequency divider consists of a chain of 24 flip-flops with each flip-flop dividing the frequency of the previous one by two. The inputs A, B, C, D (pins 9, 10, 11, 12) were used to select the flip-flop stage to be connected to the decode out pin (pin 13) (Table 12.1). The output of the frequency divider (pin 13) was a 300 Hz, 50% duty cycle square wave during free run.

Table 12.1: Truth table of the select inputs (A, B, C, D) of the MC14536B Frequency Divider

<table>
<thead>
<tr>
<th>Input</th>
<th>Stage selected For decode out</th>
</tr>
</thead>
<tbody>
<tr>
<td>D C B A</td>
<td></td>
</tr>
<tr>
<td>0 0 0 0</td>
<td>9</td>
</tr>
<tr>
<td>0 0 0 1</td>
<td>10</td>
</tr>
<tr>
<td>0 0 1 0</td>
<td>11</td>
</tr>
<tr>
<td>0 0 1 1</td>
<td>12</td>
</tr>
<tr>
<td>0 1 0 0</td>
<td>13</td>
</tr>
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<td>14</td>
</tr>
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<td>15</td>
</tr>
<tr>
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<tr>
<td>1 0 0 1</td>
<td>18</td>
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</tr>
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</tr>
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</tr>
<tr>
<td>1 1 1 0</td>
<td>23</td>
</tr>
<tr>
<td>1 1 1 1</td>
<td>24</td>
</tr>
</tbody>
</table>

Ring Shift Counter

A 4-bit ring shift counter was designed (see Figure 12.5) to generate the sequential control waveforms (Q0 and Q2) described in Figure 12.3. The ring shift counter used was similar to a synchronous counter because the clock input to each J-K flip-flop (LS7476) was driven by the same clock input
(Figure 12.3). Their outputs (Q0 to Q3) did not count in true binary, but instead provided a repetitive sequence of digital output levels.

In the case of a 4-bit ring shift counter, used in this circuit (see Figure 12.3), the output of each flip-flop was HIGH (+5 V) for one clock period, then LOW (0 V) for the next three, then repeated, as shown in Figure 12.4. To form the ring shift counter shown in Figure 12.3, the Q-1 output of each J-K flip-flop stage was fed to the J-K input of the next stage and the Q-1 output of the last stage was fed back to the J-K input of the first stage. Before applying clock pulses, the shift counter was preset with 1-0-0-0 (on reset (Rd) input pins). A brief description of the operation of the 4-bit ring shift counter is presented below.

The 1 kΩ and 0.001 μF RC circuit connected to the power supply (Figure 12.3) provided a LOW-then-HIGH a soon as the power was turned on, forcing a HIGH-LOW-LOW-LOW at Q0-Q1-Q2-Q3, which was the necessary preset condition for a ring shift counter. At the first negative clock input edge, Q0 went LOW because just before the clock edge J0 was LOW (from Q3) and K0 was HIGH (from 13). At the same clock edge, Q1 went HIGH because its J-K inputs were connected to Q0-10, which were 1-0. The Q2 and Q3 flip-flops remained Reset (LOW) because their J-K inputs saw a 0-1 from the previous flip-flops. Now, the ring shift counter was outputting a 0-1-0-0 (input clock period 2). At the negative edge of period 2, the flip-flop outputs responded to whatever levels were present at their J-K inputs, the same as explained at the beginning of the paragraph. That was, since J2-K2 were looking back (connected to) at Q1-11 (1-0) then Q2 went HIGH. All other flip-flops were looking back at a 0-1, so they would Reset (LOW). This cycle repeated continuously. Figure 12.4 shows the timing diagram of the 4-bit ring-shift counter. The Q0 and Q2 outputs of the ring shift counter were used to turn the red and infrared emitters on and off at the desired sequence as described in the following section.
12.2.2 Emitter Driver

The emitters (R and IR) were driven by a pair of identical constant current sources, one for each wavelength, which were time multiplexed using analogue switches (MC14053) that turned the red and infrared emitters on and off at 75 Hz. Figure 12.5 shows the emitter driver circuit. This circuit provided 40 mA of pulse current to each pair of emitters (20 mA per emitter) (red and infrared). The constant current sources used in this circuit were identical to the one described in section 10.2.1. The control timing signals Q0 (RED) and Q2 (IR), from the master clock and timing generator circuit described previously, controlled the switching (on/off) of the analogue switches inside the MC14053 analogue multiplexer (see Figure 12.5). The MC14503 analogue multiplexer comprises three digitally controlled analogue switches. This device allows independent control for each switch. Table 12.2 presents more detail specifications on the MC14053 analogue multiplexer. Channel select input A (pin 11) of the multiplexer controlled the X-switch (IR wavelength), and channel select input C (pin 9) controlled the Z-switch (RED wavelength) (see Figure 12.5). The current that drove each pair of emitters depended on the reference voltage (Vref) which was switched to the constant current sources for the red (VR) and infrared (VIR) emitters by the analogue switches (X and Z) of the multiplexer (see Figure 12.5). In this design the reference voltage was maintained constant at +5 V for both VIR and VR inputs, therefore the same current (40 mA) drove both red and infrared emitters (Figure 12.5).
Figure 12.5: Emitter Driver Circuit

The red and infrared emitters were never on at the same time, although during part of the emitter switching cycle they were both off to allow the photodetector to detect ambient light. Figure 12.6 shows the required timing diagram of the Emitter Driver Circuit of the PPG Processing System 3. The duty cycle in the PPG Processing System 3 was 1/4. The red emitters were on for the first quarter cycle, then all emitters were off for the second quarter cycle. The infrared emitters were on for the third quarter cycle, and all were again off for the final quarter cycle (Figure 12.6).

Table 12.2: Specifications of the MC14503 Analogue Multiplexer

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Parameter</th>
<th>Value</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>VDD</td>
<td>Positive DC supply Voltage</td>
<td>-0.5 to +18</td>
<td>V</td>
</tr>
<tr>
<td>Vin</td>
<td>Input voltage, all inputs</td>
<td>-0.5 to VDD</td>
<td>V</td>
</tr>
</tbody>
</table>
I

DC current drain per pin

25

mA

T_A

Operating Temp.

-55 to +125

°C

IDD

Quiescent current at +5Vdc

0.005

µA

### Functional / Package Specifications

#### LOGIC DIAGRAM
MC5474HC4053
Triple Single-Pole, Double-Position Plus
Common Off

#### FUNCTION TABLE - MC5474HC4053

<table>
<thead>
<tr>
<th>Control Inputs</th>
<th>Select</th>
<th>ON Channels</th>
</tr>
</thead>
<tbody>
<tr>
<td>L</td>
<td>L</td>
<td>Y0 X0</td>
</tr>
<tr>
<td>L</td>
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<td>Z0 Y0 X1</td>
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<tr>
<td>L</td>
<td>L</td>
<td>Z0 Y1 X0</td>
</tr>
<tr>
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<td>L</td>
<td>Z0 Y1 X1</td>
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<td>L</td>
<td>L</td>
<td>Z1 Y0 X0</td>
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<td>L</td>
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</tr>
<tr>
<td>L</td>
<td>L</td>
<td>Z1 Y1 X0</td>
</tr>
<tr>
<td>L</td>
<td>L</td>
<td>Z1 Y1 X1</td>
</tr>
<tr>
<td>L</td>
<td>L</td>
<td>Z1 X X X</td>
</tr>
<tr>
<td>H</td>
<td>H</td>
<td>NONE</td>
</tr>
</tbody>
</table>

X = Don't Care

#### CHOPPING FREQUENCY = 75 Hz

Clock input

-5V 0V

Q0
(Red)

-5V 0V +5V

Q2
(IR)

RED ON BOTH OFF IR ON BOTH OFF RED ON BOTH OFF IR ON BOTH OFF

Figure 12.6: Timing Diagram of PPG Processing System 3

The effect of ambient light could have been estimated during the off periods and used to correct the results, although it was deemed unnecessary as the Oesophageal PPG Probe II was in “total darkness”, in the collapsed tube of the oesophagus.

### 12.2.3 Preamplifier

The photodetector was receiving the signals reflected from a pair of infrared emitters and from a pair of red emitters and generated a current proportional to the light incident upon it. Each pair of emitters was turned on at a rate of
75 Hz in the sequence described in section 12.2.2 (Figure 12.6). The preamplifier circuit consisted of a transresistance (current-to-voltage) amplifier and an inverting amplifier (Figure 12.7). The transresistance amplifier converted the photodiode current into a mixed signal voltage containing red and infrared PPGs (a more detail description of the circuitry and operation of the transresistance amplifier can be found in section 10.2.2). The signal was then passed into an inverting amplifier with a gain of one to invert the negative going output signal of the transresistance amplifier.

**Figure 12.7: Preamplifier Circuit**

12.2.4 Demultiplexer
The mixed PPG signal from the output of the inverting amplifier (Figure 12.7) was then fed into a demultiplexer (Figure 12.8) to separate the red and infrared PPG signals into two independent channels (red and infrared). The inputs to the demultiplexer (MC14052) were the inverting amplifier output and the control timing signals (Q0 and Q2) from the master clock and timing generator circuit (Figure 12.8).
Figure 12.8: Demultiplexing of the mixed red and infrared PPG signals

The MC14052 analogue demultiplexer comprises two digitally controlled analogue switches. Table 12.3 presents more detail specifications on the MC14052 analogue demultiplexer. Channel select input A (pin 10) of the demultiplexer controlled the X-switch (IR wavelength), and channel select input B (pin 9) controlled the Y-switch (RED wavelength) (see Figure 12.8).

The output multiplexed PPG signal from the inverting amplifier (Figure 12.7) was fed into the analogue switches (X and Y) of the demultiplexer (pins 13 and 3). During the Q0 active time (timing signal Q0 is HIGH) the red emitters were on and the Y-switch was closed (X-switch was open), which allowed the red PPG signal to pass (pin 2). The opposite occurred during the Q2 active time, where the infrared emitters were on and the X-switch was closed (Y-switch was open) and therefore, the infrared signal passed through (pin 14). Switching at the right time, Q0 and Q2 resulted in the two components (red and infrared) getting separated.

Table 12.3: Specifications of the MC14502 Analogue Demultiplexer

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Parameter</th>
<th>Value</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>VDD</td>
<td>Positive DC supply Voltage</td>
<td>-0.5 to +18</td>
<td>V</td>
</tr>
<tr>
<td>Vin</td>
<td>Input voltage, all inputs</td>
<td>-0.5 to VDD</td>
<td>V</td>
</tr>
</tbody>
</table>
12.2.5 Filtering and Attenuation

In order to eliminate the high frequency (75 Hz) switching noise from the demultiplexer, caused by the timing control signals Q0 and Q2, the red and infrared PPG signals were low-pass filtered. Two identical first order active low pass filters, one for the red channel and one for the infrared channel were constructed. The input circuit of the filter was a single low-pass RC network connected to the non-inverting input of a unity gain op-amp. Simply stated, this was a voltage follower with an RC filter between the input signal and the non-inverting input. The cut-off frequency was set at 40.2 Hz. One of the two identical low-pass filters (infrared) is shown in Figure 12.9. The red and infrared PPG signals were then attenuated, using a simple inverting amplifier (see Figure 12.9) with a gain of 0.037, before passing the PPG signals (red and infrared) through the isolation barrier of the analogue isolation amplifier to the output side of the processing system. Attenuation was necessary since the maximum input positive voltage (VIN+) of the isolation amplifier was 200 mV (see Table 11.1) and the average PPG amplitude at the output of the low pass filter was approximately 4.0 V.
12.3 ISOLATION AND OUTPUT SIDE CIRCUITRY OF THE OESOPHAGEAL PPG PROCESSING SYSTEM 3

Two analogue isolation amplifiers (HCPL7820) were used to pass the two PPG signals (red and infrared) from the input side to the output side of the processing system. The position and number of analogue isolation amplifiers used in the processing system was considered. The introduction of the isolation amplifiers after the low-pass and attenuation circuit (Figure 12.9) was found to be the most appropriate one since only two isolation amplifiers were needed to isolate the red and infrared PPG signals from the input side to the output side of the processing system. If the isolation stage were to take place before the demultiplexer at the output of the preamplifier circuit (Figure 12.7), then three isolation amplifiers were needed. One to pass the mixed red and infrared PPG signal across the isolation barrier and two more to pass the timing control signals (Q0 and Q2) that drive the demultiplexer (Figure 12.8). This option was dismissed since the higher number of analogue isolation amplifiers used in the circuit would increase the circuit complexity and the possibility of electrical faults and also the overall cost of the processing system.

The isolation amplifiers used were identical to those described in section 11.2.2 of the previous chapter. Since, the output side circuitry was identical.
for both red and infrared wavelengths only one (infrared) will be discussed here.

The differential output of the isolation amplifier was converted into a single ended signal by a differential amplifier with a gain of one (Figure 12.10). The infrared PPG signal which consisted of a large amplitude DC PPG component and a small amplitude AC PPG component (approximately 1% to 2% of the DC component) was split into two independent channels (AC and DC) using filters.

The AC PPG component was extracted using a band-pass filter (Figure 12.10). The band-pass filter used consisted of a first order high-pass active filter, to block the DC PPG component, and a Butterworth 2-pole low pass active filter which attenuated high frequencies. The frequency response was 0.48 Hz to 20 Hz (at -3dB) with a gain of 1.58 in the pass band (Figure 12.10). The low-pass cut-off frequency was chosen to be 20 Hz because the PPG signal does not extend much beyond 12 Hz (see section 3.1). A more detailed analysis of the band-pass filter was presented in section 10.2.3. The AC PPG signal from the output of the band-pass filter was fed into a fixed amplification stage comprising an inverting amplifier with a gain of 13.6 (Figure 12.10). The signal was then passed through a precision variable amplification stage, identical to the one described for the PPG Processing System 2 (section 11.2.2), with a gain in the range of 0.08 to 4.0 (Figure 12.10). Table 12.4 summarises all the amplification stages in the PPG Processing System 3 including the 11 positions of the variable amplification stage.
Figure 12.10: Isolation and Output Side circuitry of the Oesophageal channel of the PPG Processing System 3
Table 12.4: Analysis of all gain stages of the PPG Processing System 3 at different switch positions

<table>
<thead>
<tr>
<th>Position</th>
<th>Low-pass filter</th>
<th>attenuation</th>
<th>Opto-isolator gain</th>
<th>Diff. Amp.</th>
<th>Band-pass filter</th>
<th>Invert. ampl.</th>
<th>Variable stage</th>
<th>Overall gain</th>
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<td>1.58</td>
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<td>0.2</td>
<td>1.3</td>
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<td>3.0</td>
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<tr>
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<td>0.037</td>
<td>8</td>
<td>1</td>
<td>1.58</td>
<td>13.6</td>
<td>3.5</td>
<td>22.4</td>
</tr>
<tr>
<td>11</td>
<td>1</td>
<td>0.037</td>
<td>8</td>
<td>1</td>
<td>1.58</td>
<td>13.6</td>
<td>4.0</td>
<td>25.6</td>
</tr>
</tbody>
</table>

Finally, the AC PPG component passed through a DC level shifting circuit which enabled the user to DC level shift the PPG signal up and down the display screen of the laptop computer (Figure 12.10).

The DC PPG component was extracted from the combined AC and DC PPG signal at the output of the differential amplifier using a unity gain 2-pole, active low pass Butterworth filter with a cut-off frequency at 0.15 Hz (Figure 12.10).

### 12.4 INPUT SIDE CIRCUITRY OF THE ECG CHANNEL

An isolated 3-Lead ECG channel was also constructed for monitoring the R waves of the QRS complex of the ECG signal, which were used as a timing reference for the PPG signals. The ECG signals were sensed by electrodes placed on the body of the patient, and transferred to an instrumentation amplifier, which pre-amplified the signal. The signals were then filtered by a band-pass filter before being passed across the isolation barrier of the analogue isolation amplifier (HCPL7820) for further signal processing. Figure 12.11 shows a detailed schematic diagram of the ECG channel.

#### 12.4.1 Instrumentation amplifier

The instrumentation amplifier in Figure 12.11 was constructed using three operational amplifiers (TL084). A high common-mode rejection ratio (CMRR) was achieved by adjusting the 22 k\(\Omega\) potentiometer to about 12 k\(\Omega\). The gain of the instrumentation amplifier was 11.9 and was derived from:
The gain of the instrumentation amplifier was relatively low when compared with conventional ECG circuits that use instrumentation amplifiers (gains of approximately 100). Low gain was necessary since the maximum input positive voltage (VIN+) of the isolation amplifier, further on in the circuit, was 200 mV (see Table 11.1) and the average ECG amplitude at the output of the instrumentation amplifier with a fixed gain of 11.9 was 120 mV to 150 mV.

12.4.2 Band-pass Filter
The output of the instrumentation amplifier was fed into a band-pass filter (see Figure 12.11). The band-pass filter used consisted of a first order high-pass active filter, to remove the DC component in the ECG signal, and a Butterworth 2-pole low pass active filter which attenuated high frequencies. The frequency response was 0.1 Hz to 41 Hz (at –3 dB) with a gain of 1.25 in the pass-band.

12.5 ISOLATION AND OUTPUT SIDE CIRCUITRY OF THE ECG CHANNEL
A third analogue isolation amplifier (HCPL7820) was used to isolate the ECG signal at the output of the band-pass filter at the input side, from the output side of the processing system. The differential output of the isolation amplifier was converted to a single ended signal by a differential amplifier with a gain of one (Figure 12.11). The ECG signal was then passed to a variable amplification stage, with a gain in the range of 1 to 21 (Figure 12.11). Finally the ECG signal passed through a DC level shifting circuit for the same reason as explained in the penultimate paragraph of section 12.3.
Figure 12.11: Schematic diagram of the 3-lead ECG channel of the PPG Processing System 3
12.6 SIGNAL DIGITISATION

12.6.1 Introduction

All four output oesophageal PPG (infrared and red AC and DC) signals and the ECG signals were digitised by an analogue-to-digital (ADC) card. The card used was a PCMCIA National Instruments Data Acquisition Card 1200 (DAQCard-1200) (National Instruments Corporation, Austin, Texas). The DAQCard-1200 was a low-power analogue input, analogue output, digital I/O, and timing I/O card. It could provide up to 100 kS/s (single channel), 12-bit performance on eight single-ended analogue inputs. It had two connectors, a 68-pin PCMCIA bus connector on one end and a 50-pin I/O connector on the other end. The voltage input range was software programmable for 0-10 V (unipolar) or ±5 V (bipolar). A software programmable gain amplifier had gain selections of 1, 2, 5, 10, 20, 50, or 100. The card also had two, 12-bit successive approximation, ADCs with analogue signal resolution of 2.44 mV at a gain of 1. The standard 50-pin I/O connector of the DAQCard-1200 enabled the connection of all analogue signals (PPGs and ECGs) to the card via a 50-way ribbon cable. The small size and weight of the DAQCard-1200 coupled with its low-power consumption made this card ideal for applications using battery powered laptop personal computers such as the one described in this thesis.

12.6.2 Laptop Personal Computer

The computer used for all the PPG and ECG signal data acquisition, processing displaying and data storage was an IBM compatible laptop (DELL LATITUDE CPi D266XT) personal computer (DELL Corporation, USA). The microprocessor in the computer was an Intel Pentium II with a clock frequency of 266 MHz and 512 Kb cache memory. The laptop also had a 64 Mb of RAM, 4 Gb hard disk space, a modular 3½ inch floppy disk and CDROM drive, and two lithium ion battery packs (up to six hours of continuous operation with both batteries). It also had a high performance 128-bit graphics controller that gave a resolution of 1024 x 768 pixels in true colour (32-bit) to a 13.3 inch TFT colour display. The operating system was WINDOWS 95 version 4.0 (Microsoft corporation, Seattle, USA).
12.6.3 Installation and Configuration of the DAQCard-1200
The DAQCard-1200 was a plug and play device and was completely software configurable. It supported switchless and jumperless configuration. All resources on this device (DAQCard-1200) including base address, DMA channels, and IRQ levels were fully software configurable. The laptop computer was able to detect and install the device automatically on power up. The appropriate DAQ configuration utility (provided by National Instruments) was then installed to obtain a mapping for the newly installed device to a NI-DAQ device number. When the DAQ configuration utility detected the new plug and play device in the computer, it assigned the first available device number (1) to the new device. Then the utility assigned the default resources to the new device (e.g., I/O address DMA channel and IRQ level).

The analogue input range for the DAQCard-1200 was selected to be bipolar (± 5 V). The analogue input mode was selected to be referenced single-ended (RSE). RSE configuration meant that all input signals were referenced to a common ground point that was also tied to the DAQ-1200 analogue input ground.

12.7 SIGNAL ACQUISITION USING LABVIEW
The digitised PPG and ECG signals were further analysed by a virtual instrument implemented in LabVIEW version 4.0 (National Instruments, Newbury, U.K) on the laptop computer. Before the detailed description of the software written to process and display the PPG and ECG signals on the laptop computer, a brief introduction to LabVIEW definitions is essential for understanding the rest of the section.

12.7.1 Introduction to LabVIEW
LabVIEW is a program development environment, much like modern C or BASIC development environments. However, LabVIEW is different from
those applications in one important respect. Other programming systems use text-based languages to create lines of code, while LabVIEW uses a graphical programming language, G, to create programs in block diagram form.

LabVIEW is a general-purpose programming system with extensive libraries of functions for any programming task. It includes libraries for data acquisition, serial instrument control, data analysis, data presentation and data storage. It also includes conventional program development tools, so the user can set breakpoints, animate the execution to see how data passes through the program, and single-step through the program to make debugging and program development easier. LabVIEW programs are called virtual instruments (VIs) because their appearance and operation can imitate actual instruments.

A VI consists of an interactive user interface, a data flow diagram that serves as the source code, and icon connections that allow the VI to be called from higher level VIs. More specifically, VIs are structured as follows:

- The interactive user interface of a VI is called the front panel, because it simulates the panel of a physical instrument. The front panel can contain knobs, push buttons, graphs, and other controls and indicators. Data can be entered using a mouse or keyboard, and then the results can be viewed on the computer screen.

- The VI receives instructions from a block diagram, which is constructed in G. The block diagram consists of the executable code that is created using nodes, terminals, and wires.

- VIs are hierarchical and modular. They can be used as top level programs, or as subprograms within other programs. A VI within another program is called a subVI.
Some of the basic concepts of the G language are described in the following list.

- **Loops and charts:** G has two structures to repeat execution of a sub-diagram, the *While Loop* and the *For Loop*. Both structures have resizable boxes where a sub-diagram to be repeated is placed inside the loop structure. The *While Loop* executes as long as the value at the conditional terminal in TRUE. The *For Loop* executes a set number of times. Charts are used to display real-time trend information to the operator.

- **Case and Sequence Structures:** The *Case structure* is a conditional branching control structure, which executes a subdiagram based on a certain input. A *Sequence structure* is a program control structure that executes its subdiagrams in numeric order.

- **Attribute Nodes:** They are special block diagram nodes that are used to control the appearance and functional characteristics of controls and indicators.

- **Arrays, Clusters and Graphs:** An array is a resizable collection of data elements of the same type. A cluster is a collection of data elements of the same or different types. Graphs commonly are used to display data.

### 12.8 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT 1 (VI-1)

A PPG signal Acquisition Virtual Instrument 1 (VI-1) was developed in *LabVIEW*, to sample, record, store and display simultaneously the red and infrared AC and DC PPG signals and the ECG signals on the screen of the laptop computer. The block diagram of the virtual instrument is shown in Figure 12.12. This VI-1 read the PPG and ECG data, converted them into a spreadsheet format and saved them into a file specified by the user at the start-up of the VI-1 and finally displayed the signals in real time on a chart on
the front panel of VI-1. The VI-1 is described in more detail in the following paragraphs.

12.8.1 Analogue Input Acquisition

The acquisition of the time-sampled data (PPGs and ECGs) was performed using four Intermediate analogue input VIs, *AI CONFIG*, *AI START*, *AI READ*, and *AI CLEAR* (Figure 12.12). Intermediate VIs are used when speed and efficiency is required. A brief description of the four intermediate VIs used, follows in the next paragraph.

At start-up (iteration zero) the *VI-1* called the *AI CONFIG VI* to configure the channel group and hardware and allocate a data buffer (Figure 12.12). On each iteration, this VI called the *AI START* and *AI READ VIs* (see Figure 12.12). *AI START VI* set the scan rate and started the acquisition. The *VI-1* stored the measurements in a buffer as they were acquired and the *AI READ* retrieved them from the buffer and returned all the data as an array of voltages. This array of voltages comprised the acquired PPG and ECG signals which, were displayed in real time on a chart at the front panel of *VI-1* (see Figure 12.13). In the programming of *VI-1* a circular buffer technique of data acquisition was used whereby data was continuously acquired into a circular acquisition buffer at the same time that the *VI* read the acquired data and processed it. If an error occurred, the *VI-1* called the *GENERAL ERROR HANDLER VI*, which determined whether an error had occurred. If an error had occurred, the *VI-1* created a description of the error and displayed a dialog box. It then called the *AI CLEAR VI* to clear the acquisition in progress. In the design of *VI-1* the program was set up to call the *AI READ VI* from a continuous loop (*while loop*) since the total number of data points to be acquired was not known. A stop button was placed on the front panel of the *VI-1* so that the user could stop the acquisition of data at any time (see Figure 12.13). A description of the main input and output configurations of the four intermediate analogue input VIs shown in the block diagram in Figure 12.12 follows.
**AI CONFIG VI**

**Device** is the device number assigned to the DAQCard-1200 during configuration. It equals 1 (default value). **Channels** is a list of analogue input channels to use. *VI*-1 used 5 channels (0 to 4): AC PPG Red, DC PPG Red, AC PPG Infrared, DC PPG Infrared, and ECG. **Buffer size** is the number of scans that the buffer holds. *VI*-1 was configured to a buffer size of 100000 scans. **Input Limits** is the highest and lowest expected voltage level of the signal of measurement. In *VI*-1 this was configured to +5 V for the high limit and 0 V for the low limit.

The *device number* the *channels* the *buffer size* and the high and low *input limits* can be controlled from the numeric controls located at the front panel of *VI*-1 (see Figure 12.13). Once they are set they stay the same every time the *VI*-1 starts a new acquisition.

**AI START VI**

**Scan rate** is the number of scans per second the rate at which *VI*-1 acquires the data. This was equivalent to the sampling rate per channel, configured at 60 Hz. The 60 Hz sampling rate satisfied the Nyquist criterion of the sampling theory which states that a continuous-time signal can be reconstructed from discrete, equally spaced samples if the sampling frequency is at least twice that of the highest frequency in the time signal. For the oesophageal AC PPG signals (red and infrared) the sampling rate of 60 Hz was more than adequate since their highest frequency component was at approximately 20 Hz (see section 12.3). The ECG channel was slightly under-sampled since its highest frequency component was at 41 Hz (see section 12.4.2). A sampling rate of approximately 100 Hz was needed for the ECG channel in order to satisfy the Nyquist criterion. Sampling all channels (five) at 100 Hz using the DAQCard-1200 was not possible since the card could not handle that speed for all channels and was causing the *Virtual Instrument 1* to crash every time a sampling rate of 100 Hz was selected. The maximum sampling rate that the card could handle without crashing the *VI*-1 was 60 Hz. **Number of scans to acquire** is the total number of scans *LabVIEW* acquires before the acquisition completes. A
scan is one point per channel. For VI-1, number of scans to acquire was configured to zero (0), which implies that LabVIEW will acquire data indefinitely into the buffer until the acquisition is cleared with the AI CLEAR VI or the stop button at the front panel of the VI is pressed.

The scan rate and number of scans to acquire can also be set using the numeric controls located at the front panel of VI-1 (see Figure 12.13).

AI READ VI

number of scans to read is the number of scans the VI-1 is to retrieve from the acquisition buffer (configured at 2 scans). A numeric control of the number of scan to read is located at the front panel of the VI-1 which allows the user to change the number of scans to read at any time during the acquisition. Scan backlog indicates on the front panel how much data remains in the buffer after each retrieval, and is an indication of how well the application is keeping up with the acquisition rate. If the backlog increases with time it is an indication that the scan rate is too high and will eventually overwrite the previous data in the circular buffer. Voltage data is a 2D array that contains the analogue input data (PPG and ECG signals) (see Figure 12.12). The array is scan-ordered which means that each row contains the data of a single scan. Each column holds the data for a single channel. The voltage data is connected to a chart for displaying the PPG and ECG signals in real time at the front panel of VI-1. Also the voltage data is connected to an Array to Spreadsheet String function which is used for storing the data into a file (see next section).
Figure 12.12: Block Diagram of the PPG Signal Acquisition Virtual Instrument 1 (VI-1)
12.8.2 Data Storage

The acquired data from the PPG and ECG signals was saved into a file in a spreadsheet format using *LabVIEW* File I/O functions. At start-up of VI-1 the program created the file for storing the data using the *OPEN/CREATE/REPLACE File I/O function VI* (see Figure 12.12). This function was programmed to open a new file or replace a file (configure function input set to 2), if it existed and the user gave permission, interactively using dialogue. A dialogue prompt was specified with a default file name *Data.TXT* (Figure 12.12). The voltage data (PPGs and ECGs) at the output of the *AI READ VI* was passed into an *Array to Spreadsheet String* function (see Figure 12.12) which converted the voltage data 2D array into a spreadsheet string. The behaviour of the *Array to Spreadsheet String* function was controlled by a *format string*. A format string was composed of one or more format specifiers (codes), which determined what action to take to process a given parameter. The format of the string was configured to be “%.3f” (see Figure 12.12) which means:

- “%” begins the formatting specification
- “.” separates width and precision
  - *width* (optional), specifies the minimum character field width of the output. In this configuration it was missing which means the output is as long as necessary to contain the converted input parameter.
  - *precision* (optional), specifies the number of digits to the right of the decimal point. In this configuration it was set to 3.

The output spreadsheet string from the *Array to Spreadsheet String* was passed into a *WRITE FILE VI*, which wrote the data to the file specified at the initialisation of VI-1. When the stop button was pressed at the front panel of the VI-1 the program stopped the acquisition and closed the file using the *CLOSE FILE VI* (see Figure 12.12).
12.9 DEVELOPMENT OF A SIGNAL DISPLAY VIRTUAL INSTRUMENT 2 (VI-2)

A new Virtual Instrument (VI-2) was developed in LabVIEW to display the stored data (PPGs and ECGs) written to the spreadsheet file in string format by VI-1 (see previous section). This new VI-2 was used for the offline display and evaluation (post-processing and analysis) of the acquired PPG and ECG signals. The block diagram of the VI-2 is shown in Figure 12.14. This VI-2 used one of the FILE functions called Read from Spreadsheet File VI (Figure 12.14).
On start-up the VI-2 prompted the user to select the file to open. When the file was selected, the Read from Spreadsheet File VI opened the file and called the Spreadsheet String to Array function (embedded in the block diagram of the Read from Spreadsheet File VI). This function converted the stored spreadsheet string to a numerical array of the same dimensions and representation as it had when it was stored by VI-1. The Read from Spreadsheet File VI was enclosed in a while loop which ran continuously until the Read from Spreadsheet File VI reached the end of file (eof) or the stop button at the front panel of VI-2 was pressed (Figure 12.15).
12.10 MECHANICAL CONSTRUCTION

12.10.1 Electronic Circuits
The two isolated processing channels (oesophageal PPG and ECG) were constructed on two separate Eurocard prototyping stripboards, dimensions: 160 mm x 100 mm x 1.6 mm and 90 mm x 55 mm x 1.6 mm, respectively using soldering techniques. Figure 12.16a and b shows a photograph of the two channels with the isolation barrier indicated.

12.10.2 Power Supply
Three pairs of PP3 (± 9 V) batteries were used to power the PPG Processing System 3. One pair was used to power the common output side of the two processing channels (oesophageal and ECG). The other two pairs were used to power the input sides of each channel (oesophageal and ECG) independently.
Figure 12.16: Photograph of the two isolated processing channels: (a) Oesophageal PPG channel; (b) ECG channel
12.10.3 Enclosure

The PPG Processing System 3 was enclosed in a Eurocard black case, identical to the one used in Processing System 2 (see section 11.3.3). The case accommodated the two eurocard stripboards horizontally, stacked on top of each other separated by nylon insulating spacers.

The front panel of the Processing System 3 (Figure 12.17) comprised:

- two power on/off switches (one for each input side, oesophageal and ECG)
- two rotary variable amplification selector switches (one for the Oesophageal Infrared AC PPG and Oesophageal Red AC PPG)
- three continuously variable offset control knobs (Oesophageal Infrared AC PPG, Oesophageal Red AC PPG and ECG)
- two mini DIN sockets (RS 183-1798) (one for each channel) for connecting the Oesophageal PPG Probe II and the 3-lead ECG

Figure 12.17: Photograph of the front panel of the PPG Processing System 3
The back panel of the Eurocard black case housed the three (2XPP3) battery holders and a 50-pin connector (Universal IDC Connector Plug, RS 471-092). The connector connected the five analogue output channels (Infrared AC and DC PPG, Red AC and DC PPG and ECG) of the *Processing System 3* to the analogue-to-digital card (DAQCard-1200) in the laptop computer.

### 12.11 EVALUATION OF THE HARDWARE AND SOFTWARE DEVELOPMENTS OF THE **PPG PROCESSING SYSTEM 3**

#### 12.11.1 Evaluation of the Hardware and Software Stages

The functionality of all the electronic circuits and the software developments, as described in this chapter were successfully bench tested in the laboratory. Figure 12.18 shows the timing control signals (Q0 and Q2) from the Master Clock circuit (see Figure 12.3) and the output of the transresistance amplifier (see Figure 12.7). The traces in Figure 12.18 were recorded using a TEK 2211 storage oscilloscope and Hewlett Packard ColorPro plotter (Model 7440A). The top timing signal (see Figure 12.18) is the Q0 (red) output followed by the Q2 (infrared) output of the master clock and timing generator circuit. The bottom trace is the negative going output of the transresistance amplifier indicating the on/off switching of the red and infrared emitters. During this testing a finger PPG probe was connected to the *Processing System 3* with the index finger placed in the clip of the probe.
Figure 12.18: Testing parts of the electronic circuits of PPG Processing System 3

The overall hardware and software testing was performed with the PPG Processing System 3 connected to the laptop computer using the ribbon cable that connected to the 68-pin PCMCIA bus connector (National Instruments) of the DAQCard-1200 in the laptop and was terminated with a 50-pin female connector that attached on the 50-pin connector on the PPG Processing System 3 enclosure. The finger PPG probe (6-pin DIN connector) and the 3-lead ECG cable (6-pin DIN connector) were connected to the processing system. The PPG finger probe was clipped on the left index finger and the ECG leads were connected to the right and left arms and right leg using ECG skin surface electrodes. The Signal Acquisition Virtual Instrument 1 (VI-1) implemented in LabVIEW was loaded and the PPG Processing System 3 was switched on. PPG and ECG traces were recorded for approximately 30 minutes and displayed on the laptop screen (Figure 12.19). During the monitoring time all the electronic circuits of the Processing System 3 and the VI-1 operated successfully.
12.11.2 **PPG Processing System 3 Electrical Safety Test**

The **PPG Processing System 3** was electrically safety tested using the Rigel Safety Tester Model 233 (see section 11.4.1). The insulation resistance between the patient applied parts (probes) and the output circuitry was measured for the oesophageal PPG channel and ECG channel. Also, the current flowing across the isolation barrier with mains voltage on the applied parts, and with mains voltage at the output terminals was measured for both channels. The procedures followed for both electrical safety tests (insulation resistance and mains on applied parts) were identical to the ones used for the **PPG Processing System 2** (see section 11.4.1).

The insulation resistance between the input and output circuits of the **PPG Processing System 3** was too large to register on the tester meter, but was well in excess of 100 MΩ for each channel. The test with mains on the
patient applied parts and on the output gave leakage currents across the isolation barrier of less than 9 μA in all cases. This was well below the 50 μA current limit for Class 1 and Class 2 type CF instruments.

12.12 PATIENTS AND MEASUREMENTS

The PPG Processing System 3 and Virtual Instrument 1 were used to perform detail investigations of oesophageal PPG signals on anaesthetised ASA 1 (American Standard of Anesthesiology; scale 1 to 5, with 5 the most critically ill patient) patients. AC and DC PPG signals at five oesophageal depths (15 cm, 20 cm, 25 cm, 30 cm, and 35 cm) at red and infrared wavelengths were investigated to determine an optimal position for reliable SpO₂ monitoring.

Local ethics committee approval was obtained prior to commencing the study of anaesthetised patients. Sixteen adult elective surgery patients (urological, gynaecological and general surgery) were recruited to the study (six male, ten female, average age (± SD): 49.1 ± 14.1). Table 12.5 shows the patients’ details. Any patients undergoing head, neck or throat procedures were excluded.

Table 12.5: Patient Details

<table>
<thead>
<tr>
<th>Patient</th>
<th>Sex</th>
<th>Age</th>
<th>Type of operation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Male</td>
<td>65</td>
<td>Transurethral Prostatectomy (TURP)</td>
</tr>
<tr>
<td>2</td>
<td>Male</td>
<td>35</td>
<td>Percutaneous Nefrolithotomy</td>
</tr>
<tr>
<td>3</td>
<td>Female</td>
<td>41</td>
<td>Laparoscopic cholecystectomy</td>
</tr>
<tr>
<td>4</td>
<td>Male</td>
<td>62</td>
<td>Left knee amputation</td>
</tr>
<tr>
<td>5</td>
<td>Male</td>
<td>26</td>
<td>Septorhinoplasty</td>
</tr>
<tr>
<td>6</td>
<td>Female</td>
<td>56</td>
<td>Left Mastectomy</td>
</tr>
<tr>
<td>7</td>
<td>Female</td>
<td>48</td>
<td>Mastectomy</td>
</tr>
<tr>
<td>8</td>
<td>Male</td>
<td>49</td>
<td>Bone Marrow Harvest</td>
</tr>
<tr>
<td>9</td>
<td>Female</td>
<td>41</td>
<td>Laparotomy</td>
</tr>
<tr>
<td>10</td>
<td>Female</td>
<td>20</td>
<td>Myringoplasty</td>
</tr>
<tr>
<td>11</td>
<td>Female</td>
<td>64</td>
<td>Repair of incisional Hernia</td>
</tr>
<tr>
<td>12</td>
<td>Female</td>
<td>41</td>
<td>Laparoscopy</td>
</tr>
<tr>
<td>13</td>
<td>Male</td>
<td>67</td>
<td>Radical nephrectomy</td>
</tr>
<tr>
<td>14</td>
<td>Female</td>
<td>54</td>
<td>Segmental Mastectomy</td>
</tr>
<tr>
<td>15</td>
<td>Female</td>
<td>57</td>
<td>Segmental Mastectomy</td>
</tr>
<tr>
<td>16</td>
<td>Female</td>
<td>59</td>
<td>Parathyroidectomy</td>
</tr>
</tbody>
</table>
Following induction of general anaesthesia with intravenous propofol, a muscle relaxant (atracurium or vecuronium) was given, and the trachea intubated. The lungs were mechanically ventilated and anaesthesia was maintained using nitrous oxide (70%) in oxygen and isoflurane (approximately 1.5% inspired concentration). During the oesophageal measurements, values of blood oxygen saturation from a commercial finger pulse oximeter (Markette, Tram 200A; Markette Electronics, Milwaukee, Wisconsin) were recorded. Heart rate, systolic and diastolic blood pressures were also noted from the theatre monitors at two minute intervals.

The oesophageal PPG probe was inserted into a sealed oesophageal stomach tube, which was lubricated with aqueous gel. The tube was then inserted through the mouth into the oesophagus under direct vision. The stomach tube was advanced into the oesophagus until the end of the probe itself was 35 cm from the upper incisors. PPG traces from the oesophagus at both wavelengths, along with ECG traces, were recorded simultaneously for approximately 5 minutes at this depth. Measurements were repeated at 30, 25, 20 and 15 cm from the upper incisors.

12.12.1 Data Analysis and Statistics

Data files recorded by the LabVIEW virtual instrument 1(VI-1) were analysed offline. Patients were only accepted into the final analysis if measurable oesophageal PPGs were present at both wavelengths at all five depths together with an ECG record. Patients 3, 13 and 16 were excluded from the final analysis. During the measurements on patient 3 there was a complete power failure (PP3 batteries were discharged) in the Processing System 3. The study on patient 13 was purely observational and no data were collected. Patient 16 was excluded from the analysis due to failure of the infrared emitters of the PPG probe.

The amplitudes of the oesophageal AC PPG signals for each patient were measured on printouts from LabVIEW using a ruler, and the means and standard deviations (SD) calculated. Preliminary Pulse transit times (PTTs) to the oesophagus were determined similarly by measuring the time interval...
from the R wave of the QRS complex on the ECG and the upstroke of the PPG signal. The statistical significance of the differences between the oesophageal PPG amplitudes at different oesophageal depths was assessed by performing a non-parametric One Way Analysis of Variance (ANOVA) for both infrared and red wavelengths using SigmaStat (SPSS Ltd, Birmingham, UK). A non-parametric analysis was performed since the PPG data were not normally distributed. The PTT data were normally distributed and a parametric ANOVA test was used to analyse the differences between the PTT values measured at the five depths. A value of $p<0.05$ was considered statistically significant.

Manual measurements on the AC and DC PPG signals at both wavelengths allowed the estimation of blood oxygen saturation for each patient at each of the five oesophageal depths. The value of blood oxygen saturation was derived by calculating the ratio (R) of the quotients of the AC and DC amplitudes at the red (655 nm) and infrared (880 nm) wavelengths. The ratio was then used to compute the arterial oxygen saturation using an empirically derived calibration curve (see Figure 5.6)(MOYLE, 1994).

$$R(ratio) = \frac{AC_{655} / DC_{655}}{AC_{880} / DC_{880}}$$

These preliminary estimations of oesophageal blood oxygen saturation ($S_o^pO_2$) for each patient at each depth were compared with blood oxygen saturation values obtained simultaneously from the commercial finger pulse oximeter ($S_f^fO_2$).

12.13 RESULTS
Measurable AC and DC PPG traces at both wavelengths were obtained in the oesophagus at all five depths in thirteen patients. Figure 12.20 depicts typical traces from one patient for the five depths each recorded during temporary cessation of mechanical ventilation.
Figure 12.20. AC and DC PPG signals corresponding to red (R) and infrared (IR) wavelengths from five oesophageal depths and ECG traces with the mechanical ventilator temporarily switched off. The amplitudes of both red and infrared signals increase as the depth increases from 15 cm reaching a maximum at 25 cm.

When the mechanical ventilator was on an artefact synchronous with the period of the ventilator (Figure 12.21) modulated the PPG traces.
Figure 12.21. AC and DC PPG signals corresponding to red (R) and infrared (IR) wavelengths from five oesophageal depths and ECG traces with the mechanical ventilator switched on. The amplitudes of both red and infrared signals increase as the depth increases from 15 cm reaching a maximum at 25 cm.

The magnitude of the artefact was of the order of 11% at the depth of 15 cm, 20% at 20 cm, 113% at 25 cm, 109% at 30 cm and 91% at 35 cm of the PPG peak-to-peak amplitude respectively. From the 13 patients included in the analysis, it was generally observed that the oesophageal PPGs were less affected by the ventilator artefact in the upper oesophagus (15 cm to 20 cm) than in the middle and deep oesophagus (20 cm to 35 cm) although there was considerable variability from patient to patient (see Figure 12.21).
Table 12.6 gives the mean \( \pm \) SE of the AC PPG amplitudes at both wavelengths at the five oesophageal depths for the 13 patients. The AC PPGs in the mid to lower oesophagus (depths of 20 cm or greater) had larger mean amplitudes at both wavelengths than those in the upper oesophagus (15 cm). The maximum mean oesophageal amplitude for each wavelength occurred at the depth of 25 cm. The mean value of the AC PPG amplitude at 25 cm was a factor 4.8 higher than that at 15 cm at the infrared wavelength and a factor of 6.7 higher at the red wavelength. The mean amplitudes of the red PPG signals at all depths were smaller than the corresponding infrared PPG signals by a factor of approximately 0.45, typical ratio (R) (red/infrared) value of normal high oxygen saturation (see section 5.6). The error of these manual amplitude measurements were of the order of \( \pm 3\% \) which was insignificant compare to the physiological variability of the PPG signals at the five oesophageal depths. These results are also illustrated in Figure 12.22.

Table 12.6: Mean \( \pm \) SE of AC peak-to-peak PPG amplitudes (mv) at two wavelengths and five oesophageal depths (n = 13)

<table>
<thead>
<tr>
<th>Oes. depth</th>
<th>15 cm</th>
<th>20 cm</th>
<th>25 cm</th>
<th>30 cm</th>
<th>35 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infrared (880 nm)</td>
<td>134 ± 13</td>
<td>565 ± 77</td>
<td>649 ± 106</td>
<td>467 ± 56</td>
<td>367 ± 53</td>
</tr>
<tr>
<td>Red (655 nm)</td>
<td>51 ± 5</td>
<td>322 ± 72</td>
<td>342 ± 72</td>
<td>177 ± 33</td>
<td>125 ± 17</td>
</tr>
</tbody>
</table>

Figure 12.22: Mean (\( \pm \) SE) AC PPG Peak-to-Peak Amplitudes at two wavelengths and five Oesophageal Depths
To see if there was any significant difference between the mean PPG amplitudes at the five oesophageal depths, a Kruskal-Wallis One Way Analysis of Variance on Ranks was performed (see section 11.7). A non-parametric test was used, as some of the data were not normally distributed (see section 11.7). A significant difference was found between the groups p<0.001. To isolate the groups that differed an all pairwise multiple comparison procedure (Dunn’s method) was applied (see section 11.7). The results are shown in Table 12.7 together with the corresponding mean of the PPG amplitude differences at the two depths being compared. The PPG amplitudes in the upper oesophagus at 15 cm were statistically significant smaller than the amplitudes at all other depths at the infrared wavelength. This was also true for the red wavelength except that there was no significant difference between the amplitudes at the depths of 15 cm and 35 cm (see first four rows of Table 12.7).

Table 12.7: Results of comparisons between the infrared and red oesophageal AC PPG amplitudes at five oesophageal depths (n = 13) showing mean PPG amplitude differences

<table>
<thead>
<tr>
<th>Oes. Depth Pairs (cm)</th>
<th>Mean Infrared PPG Amplitude Difference (mv)</th>
<th>Infrared (880 nm) p value</th>
<th>Mean Red PPG Amplitude Difference (mv)</th>
<th>Red (655 nm) p value</th>
</tr>
</thead>
<tbody>
<tr>
<td>15cm / 20cm</td>
<td>431.3</td>
<td>P&lt;0.05</td>
<td>270.5</td>
<td>P&lt;0.05</td>
</tr>
<tr>
<td>15cm / 25cm</td>
<td>514.8</td>
<td>P&lt;0.05</td>
<td>291.3</td>
<td>P&lt;0.05</td>
</tr>
<tr>
<td>15cm / 30cm</td>
<td>332.5</td>
<td>P&lt;0.05</td>
<td>125.6</td>
<td>P&lt;0.05</td>
</tr>
<tr>
<td>15cm / 35cm</td>
<td>238.3</td>
<td>P&lt;0.05</td>
<td>73.6</td>
<td>NS</td>
</tr>
<tr>
<td>20cm / 25cm</td>
<td>83.5</td>
<td>NS</td>
<td>20.8</td>
<td>NS</td>
</tr>
<tr>
<td>20cm / 30cm</td>
<td>-98.8</td>
<td>NS</td>
<td>-144.9</td>
<td>NS</td>
</tr>
<tr>
<td>20cm / 35cm</td>
<td>-207.3</td>
<td>NS</td>
<td>-211.8</td>
<td>NS</td>
</tr>
<tr>
<td>25cm / 30cm</td>
<td>-182.3</td>
<td>NS</td>
<td>-165.7</td>
<td>NS</td>
</tr>
<tr>
<td>25cm / 35cm</td>
<td>-303.7</td>
<td>NS</td>
<td>-231.1</td>
<td>NS</td>
</tr>
<tr>
<td>30cm / 35cm</td>
<td>-118.1</td>
<td>NS</td>
<td>-61.4</td>
<td>NS</td>
</tr>
</tbody>
</table>

Table 12.8 gives the mean ± SD of the pulse transit times (PTTs) to the oesophagus measured at five oesophageal depths using the ECG “R” wave as reference. The error from these manual calculations was of the order of ± 15% to ± 20%. For these preliminary PTT measurements this order of error
was acceptable since variability between PTT values were up to 50%. These results are illustrated in Figure 12.23.

**Table 12.8:** Mean ± SD of the pulse transit times (PTTs) to the oesophagus

<table>
<thead>
<tr>
<th>Oesophageal depth</th>
<th>15 cm</th>
<th>20 cm</th>
<th>25 cm</th>
<th>30 cm</th>
<th>35 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulse Transit Time (ms)</td>
<td>111 ± 24ms</td>
<td>106 ± 41ms</td>
<td>95 ± 41ms</td>
<td>67 ± 30ms</td>
<td>91 ± 30ms</td>
</tr>
<tr>
<td>Number of patients (n)</td>
<td>10</td>
<td>10</td>
<td>10</td>
<td>9</td>
<td>9</td>
</tr>
</tbody>
</table>

**Figure 12.23:** Oesophageal pulse transit times ± SD using the ECG “R” wave as reference

To see if there were any significant differences between the PTTs at the five oesophageal depths, a One Way Analysis of Variance was performed (data were normally distributed, see section 11.7). It was found that there were no statistically significant differences between the mean transit times at the five depths. However, there appeared to be a minimum mean PTT value at the depth of 30 cm (see Figure 12.23), which corresponded to the oesophageal region vascularised by direct branches from the aorta (ROMANES, 1972). By
contrast the upper and lower oesophagus, where the mean PTT values are higher, were supplied by longer routes, via the inferior thyroid artery and left gastric arteries, respectively (ROMANES, 1972). The oesophageal pulse transit times (PTTs) were consistent with previously reported measurements. SUGO et al. (1999) measured PTT values (PTTs were measured as intervals from the R wave of the ECG to the onset of each pulse wave) ranging from 169 ms at the nose to 311 ms at the toe. The values decreased as the measurement site got closer to the heart. The oesophageal PTT values were all less than those quoted by SUGO et al. (1999). Despite that there was no significance between the oesophageal PTT values there was a trend (see figure 12.23) which suggests that the PTT values become smaller probably due to the close proximity of the oesophagus to the heart. Further studies are needed to confirm this hypothesis.

Oesophageal blood oxygen saturation ($S_{\text{po}}^2O_2$) calculations were performed for eleven patients at all five oesophageal depths, as explained in section 12.12.1, and compared with finger blood oxygen saturation ($S_{\text{pf}}^1O_2$) values obtained simultaneously from the commercial pulse oximeter. Table 12.9 shows for each depth the mean ± SD of the difference between the oesophageal and finger blood oxygen saturation values. The calculated oesophageal blood oxygen saturation values were on average 6.5% lower than those obtained simultaneously from the finger. The consistent underestimation by the hand calculations was possibly due to the use of a published empirical calibration curve for the wavelengths 660 nm and 940 nm (MOYLE, 1994) instead of a curve specifically for the wavelengths used in the Oesophageal PPG Probe II (655 nm and 880 nm).

<table>
<thead>
<tr>
<th>Oesophageal depth</th>
<th>15 cm</th>
<th>20 cm</th>
<th>25 cm</th>
<th>30 cm</th>
<th>35 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>($S_{\text{po}}^2O_2 - S_{\text{pf}}^1O_2$)%</td>
<td>-6.6 ± 9.5%</td>
<td>-11.5 ± 10.9%</td>
<td>-8.8 ± 5.6%</td>
<td>-3.6 ± 5.1%</td>
<td>-2.7 ± 4.4%</td>
</tr>
</tbody>
</table>

12.14 DISCUSSION AND CONCLUSIONS
To overcome the limitations of the *PPG Processing System 2*, as discussed in the previous chapter in which red and infrared AC PPG measurements (no DC PPGs) were made consecutively (not simultaneously) using mechanical switches to effect the changeover between red and infrared wavelengths, a new *PPG Processing System 3* was constructed and described in this chapter. This new system not only overcame the technical limitations of the *PPG Processing System 2* but also allowed more detailed quantitative PPG studies at both red and infrared wavelengths, recorded simultaneously, using time multiplexing techniques, at various depths within the oesophagus of anaesthetised patients. A new 3-lead ECG channel was also incorporated in this *PPG Processing System 3* for monitoring the R waves of the QRS complex, which were used as a timing reference for the oesophageal PPG signals. A new data acquisition and recording system, *Virtual Instrument 1*, was also developed to sample, record, display and store the acquired PPG and ECG signals using *LabVIEW*.

The preliminary objective of this stage was to characterise PPGs in healthy (ASA 1) anaesthetised patients undergoing low risk surgery and to determine whether there would be sufficient PPG amplitudes at red and infrared wavelengths throughout the oesophagus to make pulse oximetry feasible.

Using the reflectance *Oesophageal PPG Probe II* and *Processing System 3*, it has been demonstrated that measurable PPG signals (AC and DC) with high signal-to-noise ratios at both infrared and red wavelengths can be obtained from all five oesophageal depths investigated (see Figure 12.20). These depths cover the whole range from the upper to the deep oesophagus. The maximum PPG amplitude appeared to be in the mid-oesophagus at 25 cm from the upper incisors (see Figure 12.22). The PPG signals obtained from the upper oesophagus (20 cm and 15 cm) seemed to be less affected by the ventilator artefact when compared with the PPGs obtained in the deeper oesophagus (25 cm, 30 cm and 35 cm) (see Figure 12.21). This supported the hypothesis, discussed in the previous chapter that as the monitoring depth was further away from the diaphragm (upper
oesophagus) the modulation on the PPG traces caused by the ventilator became smaller.

The oesophageal measured pulse transit times (PTTs) were consistent with previous measurements at peripheral sites (SUGO et al., 1999). Despite that there was no significance between them, they showed a trend (Figure 12.23) that the minimum value occurred when the oesophageal measurement site was supplied by the shortest and most direct arterial route. The high error in these preliminary measurements was due to the difficulty in determining the point on the PPG signal representing the upstroke of the pulse signal. Also, due to the slight undersampling of the ECG signal there was an error introduced in selecting the peak of the R wave of the QRS complex of the ECG signal. A more accurate method (using software perhaps) for measuring the peak of the R wave and the upstroke of the PPG signal are needed. Such measurements may give useful clinical information on the vasculature of the oesophagus. Further studies are needed to confirm this hypothesis.

The broad agreement between the calculated values of oesophageal blood oxygen saturation and the commercial oximeter values (Table 12.9) supported the hypothesis that the oesophagus might be an effective alternative site for monitoring blood oxygen saturation.

Despite the successful operation of the Processing System 3 and the Virtual Instrument 1 there were limitations that made the overall system not so practical to use.

The first main limitation of the PPG Processing System 3 was the power supply. It comprised three pairs of PP3 batteries, which made the wiring of the system more complex and therefore open to electrical faults. Also, if only one of the six batteries used was discharged, that was enough to compromise the operability of one channel or the whole system. The batteries were not rechargeable and their capacity was not sufficient for long
term monitoring (batteries lasted for approximately 2 hours of continuous monitoring).

A second limitation of the PPG Processing System 3 was the absence of a finger PPG channel. The introduction of another PPG finger channel (identical to the oesophageal) would allow more detail comparison studies between oesophagus and finger.

The third limitation of the system was the undersampling of the ECG analogue input signal (i.e. ECG). Using the 12-bit DAQCard-1200 was not possible to sample the five analogue channels at the desired sampling rate (100 samples/s). The maximum sampling rate that the Virtual Instrument 1 could display the analogue signals without crashing was at 60 samples/s. Also, the DAQCard-1200 offered only eight analogue input channels which were not enough if a system with oesophageal, finger and ECG channels was to be used.

Also, there was room for further optimisation of the Virtual Instrument 1. There was a need for designing a new virtual instrument, which would be able to estimate online continuous blood oxygen saturation and heart rate. The new virtual instrument could incorporate more signal software controls such as, signal gain control, signal offset control etc. The incorporation of such controls could replace some of the hardware stages of the processing system (variable amplification stage, offset control stage) which would again made the processing system less open to electrical faults.

To overcome the limitations of the PPG Processing System 3 and the Virtual Instrument 1, as discussed above, a new processing system (PPG Processing System 4) and virtual instrument (Virtual Instrument 3) were developed. This new system (hardware and software) not only overcame the technical and software limitations of the PPG Processing System 3 and Virtual instrument 1 but also allowed more detailed oesophageal and finger PPG and SpO₂ studies in patients with poor peripheral circulation in whom conventional pulse oximetry fails. Also, a new 16-bit data acquisition card,
which offers higher signal resolution, more input analogue signals and operates at higher speeds than the 12-bit DAQCard-1200 was introduced. This new system is the subject of the next chapter.
CHAPTER 13

PPG Processing System 4

13.1 INTRODUCTION

To eliminate the hardware and software limitations of the PPG Processing System 3 and Virtual Instrument 1, as discussed at the end of chapter 12 a new PPG Processing System 4 and a new Virtual Instrument 3 (VI-3) were developed. A simplified block diagram of the Processing System 4 is shown in Figure 13.1. This new electrically isolated Processing System 4 comprised three electrically isolated channels, two identical PPG channels (oesophageal and finger) and a 3-lead ECG channel. The input sides (patient side) of all three channels were electrically isolated from the common output side using new Burr-Brown ISO122 analogue isolation amplifiers (Figure 13.1). Also, the input sides of the three channels were physically isolated from each other (see Figure 13.1). Oesophageal and Finger time multiplexed AC and DC PPG traces (obtained at red and infrared wavelengths) together with ECG traces, were processed by the new Processing System 4 and digitised by a new 16-bit data acquisition card. The Processing System 4 was further optimised (Processing System 4a) by redesigning and constructing all three channels using Printed Circuit Board (PCB) technology.

The digitised PPG and ECG signals at the output of the Processing System 4 were further analysed by a new Virtual Instrument 3 implemented in LabVIEW. Oesophageal and Finger PPG signals and ECG traces were displayed simultaneously on the front panel of the new Virtual Instrument 3 (VI-3) on the laptop computer. Virtual Instrument 3 incorporated algorithms for providing signal gain and offset control. Algorithms were also developed for the estimation of oesophageal and finger SpO₂. This new VI-3 also incorporated a battery condition algorithm. The Virtual Instrument 3 was furthered optimised, Virtual Instrument 3 version a (VI-3a) by introducing an
algorithm for calculating heart rate. A final new version (version b) of *Virtual Instrument 3 (VI-3b)* was developed. This new VI-3b introduced online digital filters for the AC PPG and ECG signals.

The *PPG Processing System 4a* and the *Virtual Instrument VI-3b* were used in the study of patients undergoing high risk surgery (cardiopulmonary bypass surgery), who had an arterial cannula inserted routinely, therefore allowing rigorous validation of oesophageal SpO₂ measurements with co-oximetry and/or blood gases analysis.

The technical details of these new Processing Systems (versions 4 and 4a) and *Virtual Instruments* (versions 3, 3a, and 3b) as well as the performance evaluation tests at all stages of the system’s design will be the subject of this chapter.

**Figure 13.1:** Simplified block diagram of the *PPG Processing System 4*
13.2 PPG PROCESSING SYSTEM 4

A detailed block diagram of one of the two identical PPG channels (oesophageal) and the ECG channel is shown in Figure 13.2. The new Processing System 4 allowed the monitoring of oesophageal and finger AC and DC PPG signals by using the Oesophageal PPG Probe II (see Figure 9.10) and the identical reflectance finger probe (see Figure 9.24). Red and infrared AC and DC PPG measurements were made simultaneously.

Figure 13.2: Detailed block diagram of the Processing System 4 showing one of the two identical PPG channels (oesophageal) and the ECG channel

13.2.1 Input Side Circuitry of the Oesophageal PPG Processing Channel

The input side circuitry of the Oesophageal PPG channel of the Processing System 4 was almost identical to the input side circuitry of the Processing System 3 (discussed in section 12.2), and will not be repeated in this section, apart from the attenuation stage. Figure 13.3a and 13.3b shows the complete
input side circuit of the Oesophageal PPG channel of the Processing System 4. In the input side circuitry of the Processing System 4 the attenuation stage was eliminated since there was no need to attenuate the red and infrared PPG signals before passing them to the isolation amplifier (compare Figure 12.9 and Figure 13.3). This was due to the introduction of a new Burr Brown ISO122 analogue isolation amplifier (discussed in the next section) in the circuit where it’s maximum input positive voltage (VIN+) was much greater than the average PPG amplitude at the output of the filtering stage (Figure 13.3b).

13.2.2 Isolation and Output Side Circuitry of the Oesophageal PPG Processing Channel

Two new analogue isolation amplifiers were used to pass the two oesophageal PPG signals (red and infrared) from the input side to the output side of the oesophageal channel (see Figure 13.4). The isolation amplifiers used were Burr-Brown ISO122. The ISO122 was an easy to use (requiring no external components) precision isolation amplifier. The input to the isolator was transmitted digitally across the isolation barrier. The supply range of this isolation amplifier was between ±4.5 V to ±18 V (set at ±12 V). This wide supply range was one of the advantages of the ISO122 when compared with the HCPL7820 isolation amplifier, used in Processing Systems 2 and 3. The power supply of the HCPL7820 had to be set at 5 V using two voltage regulators (one for the input supply and one for the output supply) (see Figure 12.10) The introduction of the ISO122 eliminated the two voltage regulators from the circuit. The input signal to the isolation amplifier was converted into a current by an internal 200 kΩ integrator input resistor. That made it possible to have an input voltage of up to ±9.5 V with the input and output power supplies at ±12 V. This feature was another advantage over the HCPL7820 isolation amplifier (see section 13.2.1) where it’s maximum input positive voltage (VIN+) was ±200 mV and attenuation of the input signals was required.
Master Clock

Emitter Driver

(a)
Figure 13.3: Detail Schematic Diagram of the Input Side Circuitry of the Processing System 4; (a) Emitter Driver and Timing Generator Stages (b) Preamplifier, Demultiplexer and Filtering Stages
Also, the ISO122 had a single ended output and not a differential output as the HCPL7820, therefore a differential amplifier to convert it into a single ended signal was not necessary (Figure 13.4). The key specifications of this isolation amplifier are summarised in Table 13.1.

**Table 13.1: Technical specifications of the Analogue Isolation Amplifier ISO122**

<table>
<thead>
<tr>
<th>Description</th>
<th>Symbol</th>
<th>Characteristic &amp; units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supply Voltages</td>
<td>$V_{S1}, V_{S2}$</td>
<td>±15 V (typ)</td>
</tr>
<tr>
<td>Input Voltage @ ±15 V</td>
<td>$V_{IN+}, V_{IN-}$</td>
<td>±12.5 V (typ)</td>
</tr>
<tr>
<td>Voltage Rated Continuous AC 60 Hz</td>
<td></td>
<td>1500 VAC</td>
</tr>
<tr>
<td>Isolation Mode Rejection @ 60 Hz</td>
<td></td>
<td>140 dB</td>
</tr>
<tr>
<td>Barrier Impedance</td>
<td></td>
<td>$10^{14}$ Ω</td>
</tr>
<tr>
<td>Operating Temperature</td>
<td>$T_A$</td>
<td>-25 to 85°C</td>
</tr>
<tr>
<td>Input Offset Voltage</td>
<td>$V_{os}$</td>
<td>±20 mV (typ)</td>
</tr>
<tr>
<td>Nominal Gain</td>
<td>$G$</td>
<td>1.0</td>
</tr>
<tr>
<td>Small-Signal Bandwidth (-3dB)</td>
<td>$f_{3db}$</td>
<td>50 kHz</td>
</tr>
<tr>
<td>Quiescent Current $V_{S1}$ $V_{S2}$</td>
<td></td>
<td>±5.0 mA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>±5.5 mA</td>
</tr>
</tbody>
</table>

The infrared and red PPG signals which consisted of a large amplitude DC PPG component and a small amplitude AC PPG component (approximately 1% to 2% of the DC component) were split into independent channels (Infrared AC and DC and Red AC and DC) using filters (see Figure 13.4). The AC PPG components (for both wavelengths) were extracted using two identical band-pass filters, one for the infrared and one for the red wavelength built around the op-amps U1A and U2A respectively (Figure 13.4. The band-pass filters used consisted of a first order high-pass active filter, to block the DC PPG component, and a Butterworth 2-pole low pass active filter which attenuated high frequencies. The frequency response was 0.48 Hz to 20 Hz (at -3dB) with a gain of 1.58 in the pass band. The DC PPG components of both wavelengths were extracted from the combined AC and
DC PPG signals at the output of the isolation amplifiers using two identical unity gain 2-pole, active low pass Butterworth filters, built around U1B and U2B, with a cut-off frequency at 0.15 Hz (see Figure 13.4). Both filters used to split the red and infrared PPG signals to their AC and DC PPG components were identical to the ones used in the Processing System 3 (see section 12.3) and will not be discussed here in any more detail.

Figure 13.4: Isolation and Output Side circuitry of the Oesophageal channel of the PPG Processing System 4
The output side circuitry of the Oesophageal and Finger channels of the Processing System 4 did not include a variable amplification stage or a DC level shifting stage as in Processing Systems 2 and 3. This minimised the hardware of the Processing System 4, which resulted in a more compact and reliable (less electrical and wiring faults) system. The variable amplification stages and the DC level shifting stages for both Oesophageal and Finger channels were incorporated as software controls into the new Virtual Instrument 3 which will be discussed in the following sections.

13.2.3 Input Side Circuitry of the ECG Channel

As discussed in section 12.4 an isolated 3-Lead ECG channel was again constructed for monitoring the R waves, which were used as a timing reference for the oesophageal and finger PPG signals. Figure 13.5 shows a detail schematic diagram of the ECG channel. The input side circuitry of the ECG channel used for Processing System 4 was again identical with the one used in Processing System 3 with only one alteration. The custom-made instrumentation amplifier of the ECG channel of Processing System 3 (see Figure 12.11) was replaced with an 8-pin Burr-Brown INA114, general purpose instrumentation amplifier (see Figure 13.5). A single external resistor \( R_G \) could set the gain \( G \) from 1 to 10000. The formula used for calculating the gain of the instrumentation amplifier was given by the manufacturer in the data sheets of the chip.

\[
\text{Gain} \left( G \right) = 1 + \frac{50k\Omega}{R_G}
\]

The gain of the instrumentation amplifier in Processing System 4 was set at 100 using an external 1 k\( \Omega \) trimmer resistor set at \( R_G = 505.0 \Omega \) (Figure 13.5). The INA114 was laser trimmed by the manufacturers for very low offset voltage (50 \( \mu V \)), drift (0.25 \( \mu V/\degree C \)) and high common rejection (115 dB at \( G = 1000 \)). It operated from a wide range of power supplies (±2.25 V to ±18.0 V, and quiescent current was 2.2 mA (typical).
The output ECG signal from the instrumentation amplifier was passed through a band-pass filter (Figure 13.5), identical to the one used in the ECG channel of Processing System 3 (see section 12.4.2).

### 13.2.4 Isolation and Output Side Circuitry of the ECG Channel

An ISO122 analogue isolation amplifier was used to isolate the ECG signal after the output band-pass filtering at the input side, from the output side of the Processing System 4. The output of the isolation amplifier was then passed to a fixed amplification stage, with a gain of 10 using an inverting amplifier constructed using op-amp U2C (Figure 13.5).

**Figure 13.5:** Schematic diagram of the 3-lead ECG channel of the PPG Processing System 4
13.3 POWER MANAGEMENT OF PROCESSING SYSTEM 4

To eliminate the high number (six) of PP3 batteries used in Processing System 3 (see section 12.10.2) this new Processing System used two heavy duty (+12 V) sealed lead-acid rechargeable (Yuasa NP) batteries (dimensions (each): L=178 mm x W=34 mm x H=67 mm; weight (each): 0.90 kg). Table 13.2 summarises their specifications.

Table 13.2: Technical specifications of the Yuasa NP Sealed Lead-Acid Batteries

<table>
<thead>
<tr>
<th>Battery Ratings and Storage</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Battery Type</strong></td>
</tr>
<tr>
<td>---------------------</td>
</tr>
<tr>
<td>Fully Charged</td>
</tr>
<tr>
<td>NP2.1-12</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Battery charge information</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Battery Type</strong></td>
</tr>
<tr>
<td>---------------------</td>
</tr>
<tr>
<td>NP2.1-12</td>
</tr>
</tbody>
</table>

The batteries were interconnected to produce the positive (+12 V) and negative (-12 V) rails required by Processing System 4 (Figure 13.6). The ±12 V output from the batteries was connected to the common output side of the three channels (Oesophageal, Finger, and ECG) (Figure 13.6).

The power was distributed to the three isolated input sides of the channels using three dual isolated DC-to-DC converters (Burr-Brown 722BG) (Figure 13.6). The 722BG converted a single +12 V DC input to a pair of bipolar output voltages of the same magnitude (±12 V) as the input voltage. The positive power potential (Vin=+12V) was applied to the “P+” input terminal (see Figure 13.6). The voltage for the input side “V+” terminal was obtained from “P+”. One of the requirements of the “P+” and “V+” inputs was that they had to remain positive with respect to “V-” at all times. For that reason a diode clamp was used across these inputs (Figure 13.6). Diodes were also
used to prevent the DC-to-DC converter and the output side circuitry from power supply reversal. The “E” pin enabled the converter when it was connected to the “V+”. An external capacitor (0.47 μF) was connected across “P+” and “V−” to reduce input ripple.

**Processing System 4**

Each of the outputs of the 722BG was filtered with an internal 0.22 μF capacitor. To reduce output ripple voltage even more, external capacitors (0.1 μF) were connected between each output and its common (Figure 13.6). The converter was capable of providing a total maximum output current of 100 mA (per output) at rated voltage. The two output sides of the converter were connected in parallel for higher output current (2 x 100 mA) (Figure 13.6). Table 13.3 shows the important electrical specifications of the 722BG dual isolated DC-to-DC converter.

**Figure 13.6:** Schematic diagram illustrating the power management of the Processing System 4
Table 13.3: Specifications of the 722BG Dual Isolated DC/DC Converter

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Conditions</th>
<th>722BG</th>
</tr>
</thead>
<tbody>
<tr>
<td>Input Voltage Range</td>
<td>Input-to-output, 5 seconds, min</td>
<td>5 to 15 VDC</td>
</tr>
<tr>
<td></td>
<td>Input-to-output, 1 minute, min</td>
<td></td>
</tr>
<tr>
<td>Isolation (test voltages)</td>
<td>Input-to-output</td>
<td>8000 Vpk, 2500 Vrms</td>
</tr>
<tr>
<td></td>
<td>Input-to-output, 5 seconds, min</td>
<td>8000 Vpk</td>
</tr>
<tr>
<td></td>
<td>Input-to-output, 1 minute, min</td>
<td>2500 Vrms</td>
</tr>
<tr>
<td>Isolation Impedance</td>
<td>Input-to-output</td>
<td>10 GΩ</td>
</tr>
<tr>
<td>Leakage Current</td>
<td>Input-to-output, 240 V, 60 Hz</td>
<td>1 μA</td>
</tr>
<tr>
<td>Rated Output Voltages</td>
<td>$I_{\text{LOAD}}=40 \text{ mA per output, } V_{\text{IN}}=15 \text{ VDC}$</td>
<td>14.2 VDC</td>
</tr>
<tr>
<td>Output Current</td>
<td>Any one output (max)</td>
<td>100 mA</td>
</tr>
</tbody>
</table>

13.4 MECHANICAL CONSTRUCTION OF THE PROCESSING SYSTEM 4

13.4.1 Electronic Circuits

The three isolated processing channels (oesophageal PPG, finger PPG and ECG) were constructed on three separate Eurocard prototyping stripboards, (dimensions: 160 mm x 100 mm x 1.6 mm) using soldering techniques. Figure 13.7 shows a photograph of one of the two identical PPG channels (oesophageal) and the ECG channel.

![Photograph of the isolated processing channels of Processing System 4](image)

*Figure 13.7: Photograph of the isolated processing channels of *Processing System 4*: (a) Oesophageal PPG channel; (b) ECG channel*
13.4.2 Enclosures

The PPG Processing System 4 was enclosed in a Eurocard black case, identical to the one used in Processing System 3 (see section 12.10.3). The case accommodated the three eurocard stripboards horizontally, stacked on top of each other. The batteries were enclosed in a separated enclosure.

The front panel connectors of the Processing System 4 comprised three mini DIN sockets (RS 183-1798) (one for each channel) for connecting the Oesophageal PPG Probe II, the Finger PPG Probe and the 3-lead ECG.

The back panel of the Eurocard black case housed a 68-pin male vertically mounting connector (National Instruments). The connector connected the nine analogue output channels (oesophageal and finger red and Infrared AC and DC PPGs and ECG) of Processing System 4 to the DAQCard-AI-16XE-50 (discussed later on in this chapter) via a 68-way ribbon cable. A tenth channel representing the battery condition signal was also connected to the 68-pin connector. The battery condition signal (approximately +12 V) was
attenuated by 50% before it was connected to the analogue input channel of the unipolar configured (0 to 10 volts) DAQCard-AI-16XE-50. Apart from the 68-pin male connector the back panel also incorporated three (+VCC, -VCC, GND) 4 mm round insulating sockets allowing the PPG Processing System 4 to be powered externally by the two heavy duty (+12 V) sealed lead-acid rechargeable batteries (discussed in section 13.3) or a conventional dual power supply for testing.

13.4.3 Battery Enclosure
The two sealed lead-acid rechargeable batteries were enclosed in a separate plastic box (dimensions: W=110 mm, L= 190 mm, H= 85 mm). The front panel of the battery box comprised a power on/off switch (RS 316-989) with a power on illuminating red LED. Three 4 mm round insulating sockets were also incorporated into the front panel to transfer the power from the power supply box to the enclosure of Processing System 4 (see Figure 13.6). Figure 13.8 shows a photograph of the Processing System 4 connected to the external power supply box.

Figure 13.8: Photograph of the completed Processing System 4.
13.5 DEVELOPMENT OF A PRINTED CIRCUIT BOARD (PCB) VERSION OF PROCESSING SYSTEM 4 (SYSTEM 4a)

The Processing System 4 was further optimised by redeveloping the three isolated channels using printed circuit board (PCB) technology. This new PCB system was called Processing System 4a. The introduction of PCB technology made the system more compact, robust and reliable. Almost all the wiring connections, which were a possible source of electrical faults, were eliminated. The circuitry for all three channels (oesophageal, finger and ECG) remained identical to the one described in detail in section 13.2. The layout of the new printed circuit boards was designed using Easy-PC (Number One Systems Ltd, Cambridge, UK). Since, the oesophageal and finger PPG channels were identical only one layout was designed. Both PCB layouts (PPG channel and ECG channel) are shown in Appendix A.

The artwork of both PCB layouts was printed on ultra-fine matt finish film (RS 180-4623). Three Photo-resist printed circuit boards (size: 180 mm x 100 mm x 1.2 mm) were used. An ultraviolet exposure unit (RS 196-5251) was used for the preparation of track layouts on the PCBs. The artwork was placed on the glass plate of the exposure unit, checking for correct side up. The plastic film of the photo-resist board was peeled off and the board was positioned onto the artwork, resist side down. The board was exposed for 3 minutes. A modular PCB preparation system (RS 512-9233) and a range of chemicals were then used for the preparation of the PCB boards. A brief description of the sequence is given below:

Develop: The board was placed in a processing tank of the modular PCB preparation system containing Universal Developer (RS 690-849). The image was fully developed in 2 to 3 minutes at 21°C. The board was then washed in a wash tank.

Etch: The board was then placed in a Bubble Etch Tank containing etchant prepared with Ferric Chloride Hexahydrate Crystals (RS 551-277).
bubble agitation the board was etched in approximately 6 minutes at 45°C. The board was then washed in a wash tank.

**Resist Strip:** The unwanted resist covering the etched circuit was removed by immersing the board in another processing tank containing *Photoresist Stripper* (RS 690-855). The resist was stripped in approximately 2 minutes at 40°C to 50°C. The board was then washed in a wash tank.

**Tin Plating:** The copper surface of the board was cleaned and then was immersed in a Process Tank containing *Tin Plating Solution* (RS 567-812). To achieve a minimum thickness of plating the immersion time was 7 minutes. The board was washed in hot water and then dried with a cloth.

All the PCB pads on the boards were drilled using a PCB bench drill (drill bit size 0.5 mm). After this stage all components were mounted on the boards using soldering techniques. Figure 13.9 shows a photograph of one of the two identical PPG channels (oesophageal) and the ECG channel constructed on PCBs.

![Figure 13.9: Photograph of the new PCB developed channels; (a) Oesophageal PPG (b) ECG](image)
The three PCB boards of *PPG Processing System 4a* were enclosed in a Eurocard black case, identical to the one used in *Processing System 4*. The case accommodated the three PCBs horizontally, stacked on top of each other. Due to the smaller size of the PCBs (compared with the three eurocard stripboards used for *Processing System 4*) there was enough room in the Eurocard case for incorporating the battery box (as describe in section 13.4.3). That eliminated the external connection of the power supply to the *Processing System 4a*, which made the system more compact, reliable and ergonomical.

The front panel connections of *Processing System 4a* comprised three mini DIN sockets (RS 183-1798) (one for each channel) for connecting the *Oesophageal PPG Probe II*, the *Finger PPG Probe* and the 3-lead ECG and a power on/off switch (RS 316-989). The back panel housed a 68-pin male vertical mounting connector identical to the one described for the *Processing System 4* (see section 13.4.2).
13.6 SIGNAL DIGITISATION

13.6.1 Introduction

All eight PPG (oesophageal and finger infrared and red AC and DC) signals, the ECG signal and the battery signal were digitised by a new analogue-to-digital (ADC) card. The card used was a PCMCIA National Instruments Data Acquisition Card AI-16XE-50 E series (DAQCard-AI-16XE-50) (National Instruments Corporation, Austin, Texas). The DAQCard-AI-16XE-50 was a low-power analogue input, analogue output, digital I/O, and timing I/O card. It could provide up to 20 kS/s (per channel), 16-bit performance on sixteen single-ended analogue inputs or 200 kS/s on a single channel. The DAQCard-AI-16XE-50 also featured digital triggering capability, as well as two 12-bit analogue outputs; two 24-bit, 20 MHz counter/timers; and eight digital I/O channels. It had two connectors, a 68-pin PCMCIA bus connector on one end and a 68-pin I/O connector on the other end. The voltage input range was software programmable for 0-10 V (unipolar) or ±10 V (bipolar). A software programmable gain amplifier had gain selections of 1, 2, 10, or 100. The card had a 16-bit successive approximation ADC with analogue signal resolution of 153 μV (gain of 1 and unipolar range) or 305 μV (gain of 1 and bipolar range). The standard 68-pin I/O connector of the DAQCard-AI-16XE-50 enabled the connection of all analogue signals (PPGs, ECGs and battery signal) to the card via the 68-way ribbon cable. The small size and weight of the DAQCard coupled with its low-power consumption made this card ideal for applications using battery powered laptop personal computers such as the one described in this thesis.

The computer used for all the PPG and ECG signal data acquisition, processing, displaying and data storage was the DELL Latitude used with the Processing System 3 (see section 12.6.2).

13.6.2 Installation and Configuration of the DAQCard-AI-16XE-50

The DAQCard-AI-16XE-50 was a plug and play device and was completely software configurable. It supported switchless and jumperless configuration. All resources on this device (DAQCard-AI-16X-50) including base address,
DMA channels, and IRQ levels were fully software configurable. The laptop computer was able to detect and install the device automatically on power up. The appropriate DAQ configuration utility software (provided by National Instruments) was then used to obtain a mapping for the newly installed device to a NI-DAQ device number. When the DAQ configuration utility detected the new plug and play device in the computer, it assigned the first available device number (1) to the new device. Then the utility assigned the default resources to the new device (e.g., I/O address, DMA channel and IRQ level).

The analogue input range for the DAQCard-AI-16XE-50 was selected to be unipolar (0-10 V). The analogue input mode was selected to be referenced single-ended (RSE). RSE configuration means that all input signals were referenced to a common ground point that was also tied to the DAQCard-AI-16XE-50 analogue input ground.

13.7 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT 3 (VI-3)

A new PPG signal Acquisition Virtual Instrument 3 (VI-3) was developed in LabVIEW (version 5.1). The block diagram of the Virtual Instrument 3 is shown in Figure 13.10. This VI-3 read the PPG and ECG data, converted them into a spreadsheet format and saved them into a file specified by the user at the start-up of the VI-3 and finally displayed the signals in real time on a chart on the front panel of VI-3 (Figure 13.11). Virtual Instrument 3 incorporated algorithms for providing signal gain and offset control. Algorithms were also developed for the online estimation of oesophageal and finger SpO₂. This new VI-3 also incorporated a battery condition algorithm. The description of VI-3 which follows in the next subsections is divided into five parts; the analogue input acquisition, the storage of data, signal gain and offset software control, the estimation of oesophageal and finger SpO₂ and the development of the battery condition algorithm.
13.7.1 Analogue Input Acquisition

The acquisition of the time-sampled data (PPGs, ECGs and battery signal) was performed using the four Intermediate analogue input VIs, \textit{AI CONFIG, AI START, AI READ, and AI CLEAR} described in detail in section 12.8.1 (Figure 12.12). The details of the functionality of these VIs will not be repeated in this section. Only a brief description of the main input and output configurations of the four Intermediate analogue input VIs shown in the block diagram in Figure 13.10 or more clearly in Figure 12.12 is given.

\textit{AI CONFIG VI}

\textbf{Device} is the device number assigned to the DAQCard-Al-16XE-50 during configuration. It equals to 1 (default value). \textbf{Channels} are a list of analogue input channels to use. \textit{VI-3} had 10 channels (0 to 9): Oesophageal AC PPG Red (6), Oesophageal DC PPG Red (1), Oesophageal AC PPG Infrared (7), Oesophageal DC PPG Infrared (2), Finger AC PPG Red (8), Finger DC PPG Red (3), Finger AC PPG Infrared (9), Finger DC PPG Infrared (4), ECG (5) and battery signal (0). \textbf{Buffer size} is the number of scans that the buffer holds. \textit{VI-3} was configured with a buffer size of 2500 scans. \textbf{Input Limits} are the highest and lowest expected voltage levels of the signal of measurement. In \textit{VI-3} Input Limits were configured to +10 V for the high limit and 0 V for the low limit.

The \textit{device number} the \textit{channels} the \textit{buffer size} and the high and low \textit{input limits} were controlled from the numeric controls located at the front panel of \textit{VI-3} (see Figure 13.11).

\textit{AI START VI}

\textbf{Scan rate} is the number of scans per second the \textit{VI-3} acquires the data configured at 100 scans/s. \textbf{Number of scans to acquire} is the total number of scans \textit{LabVIEW} acquires before the acquisition completes. A scan is one point per channel. For \textit{VI-3}, \textit{number of scans to acquire} was configured to zero (0), which implies that \textit{LabVIEW} will acquire data indefinitely into the buffer until the acquisition is cleared with the \textit{AI CLEAR VI} or the stop button on the front panel of \textit{VI-3} was pressed.
The **scan rate** and *number of scans to acquire* were set by numeric controls located at the front panel of *VI-3* (see Figure 13.11).

**AI READ VI**

*number of scans to read* is the number of scans the *VI-3* is to retrieve from the acquisition buffer (configured at 10). A numeric control of *the number of scans to read* was located at the front panel of the *VI-3*, which allowed the user to change the *number of scans to read* at any time during the acquisition. *Scan backlog* indicated on the front panel how much data remained in the buffer after each retrieval, and was an indication of how well the application was keeping up with the acquisition rate. If the backlog increased with time, it was an indication that the scan rate was too high and would eventually overwrite the circular acquisition buffer. *Voltage data* is a 2D array that contained the analogue input data (PPG and ECG signals) (see Figure 13.10). The array was scan-ordered which means that each row contained the data of a single scan. Each column holds the data for a single channel (for *VI-3* ten columns were required to hold the output signals from the *Processing System 4*). The *Voltage data* array was connected to an *Array to Spreadsheet String* function, which was used for storing the data into a file (see next section). In *VI-3* the output voltage data was separated into the ten different channels using an *Index Array* (see Figure 13.10). An *Index Array* can be used to extract any of the columns (channels) at the *Voltage Data* output of the *AI READ VI*. Therefore, all ten channels were separated using ten different *Index Arrays* with each one indicating the channel (column) to be extracted (Figure 13.10).

**13.7.2 Data Storage**

The acquired data from the PPG and ECG signals was saved into a file in a spreadsheet format using *LabVIEW* File I/O functions. The data storage procedure was exactly identical with the one described for *Virtual Instrument 1* (see section 12.8.2) and will not be repeated in this section.
Figure 13.10: Block Diagram of the Virtual Instrument 3 (VI-3)
13.7.3 Signal Gain and Offset Software Control

The gain and offset (DC level shifting) of the oesophageal and finger AC PPGs, at red and infrared wavelength, channels and the ECG channel were controlled using software implemented in VI-3. Figure 13.12 shows the part of the main block diagram of VI-3 illustrating the gain and offset control for the five channels.

The finger red and infrared AC PPG signals (F/R/AC and F/IR/AC) were controlled by the same gain numeric control (see Figure 13.12). The same was done with the oesophageal red and infrared AC PPG signals (O/R/AC and O/IR/AC). The ECG channel had its own gain control (Figure 13.12). The control of the gain was accomplished by connecting the output of each channel (output of index array) to a multiply numeric function (returns the product of the two inputs) (Figure 13.12). The other input of the multiply numeric function was the gain numeric control (knob type) which was located on the front panel of the VI-3 (see Figure 13.11). This gain numeric control allowed the user to change the gain of the signal by using the keyboard of the laptop computer to insert the exact number to be multiplied with the
signal to produce the desired gain. The gain could also be controlled by using the mouse of the laptop to turn the gain knob to the desired gain (Figure 13.11).

**Figure 13.12:** Gain and Offset Control of the Oesophageal and Finger AC PPG Channels and ECG Channel

The amplified signals from each channel were then passed into an *add* numeric function (computes the sum of the two inputs) (see Figure 13.12). The other input of the *add* numeric functions was connected to an offset numeric control (one for each signal) located again at the front panel of the VI-3. These offset numeric controls allowed the user to level shift (offset) the signal on the chart display using either the mouse or the keyboard of the laptop computer to input the desired amount of offset.
13.7.4 Estimation of Oesophageal and Finger SpO\textsubscript{2}

The algorithms used to estimate oesophageal and finger SpO\textsubscript{2} were identical, therefore only the oesophageal algorithm will be discussed here. The value of blood oxygen saturation was derived by calculating the ratio (R) of the quotients of the AC and DC amplitudes at the red (655 nm) and infrared (880 nm) wavelengths.

\[ R(\text{ratio}) = \frac{\text{AC}_{655}}{\text{DC}_{655}} / \frac{\text{AC}_{880}}{\text{DC}_{880}} \]  \hspace{1cm} 13.1

The ratio (R) was then used to compute the arterial oxygen saturation using an empirically derived calibration equation (WEBSTER, 1997).

\[ \text{SpO}_2 = 110 - (R) \times 25 \]  \hspace{1cm} 13.2

The equation was a linear approximation of an empirical calibration curve derived from a large group of healthy volunteers with arterial SaO\textsubscript{2} values generally greater than 70% (WEBSTER, 1997).

The description of the algorithms to estimate SpO\textsubscript{2} will be divided into two parts; the calculation of the ratio (R) and the estimation of SpO\textsubscript{2} by using the ratio (equation 13.1) and the equation 13.2.

**Calculation of the ratio (R)**

To be able to calculate the ratio (R) the VI-3 was set up to call the AI READ VI from a For Loop inside a continuous loop (While Loop) (Figure 13.13). The For Loop, executed the code inside its borders, called a subdiagram, a total of count times, where the count equalled the value connected in the count terminal (Figure 13.13). This enabled the continuous collection of enough data (AC and DC PPG) to calculate the ratio. The count terminal of the For Loop was set to 20 iterations. Figure 13.13 is the part of the block diagram of VI-3, which illustrates the calculation of the ratio (for the oesophageal channel). At every iteration of the For Loop the AI READ VI read ten points (samples or scans) from the circular buffer by setting the number of scans to read at the front panel of the VI-3 equal to ten. At the end of the 20th iteration of the For Loop 200 points of data were collected in the index arrays of each channel. Two hundred samples (sampling at 100 samples/s) were
enough (approximately two PPG cycles) to allow the estimation of the AC and DC PPG component for the red and infrared wavelength. At the end of the For Loop execution the data (200 samples) in the index arrays (Oesophageal Red AC (O/R/AC), Oesophageal Red DC (O/R/DC), Oesophageal Infrared AC (O/IR/AC), and Oesophageal Infrared DC (O/IR/DC)) were passed into four Array Max & Min array functions located outside the For Loop (Figure 13.13). These array functions searched for the maximum and minimum values in the numeric array input. For the AC PPG data (200 samples), at red and infrared wavelengths, the max (peak) and min (valley) values were found and then passed into a subtract numeric function (Figure 13.13).

![Diagram](image)

**Figure 13.13:** Illustration of the algorithm used for calculating the ratio (R) for the oesophageal channel
The pulsatile AC PPG component was calculated by subtracting the PPG AC_{min} from the PPG AC_{max} for both wavelengths (see Figure 13.13). The max DC PPG components for both wavelengths were also found.

The ratios of (AC/DC)_R and (AC/DC)_IR were calculated using two division numeric functions (see Figure 13.13). The final ratio (R) value as described in formula 13.1 was calculated by dividing the quotients of the AC and DC amplitudes at the red (655 nm) and infrared (880 nm) wavelengths using another division numeric function (Figure 13.13). The value of the ratio (R) (for both oesophageal and finger) was displayed on the front panel of the VI-3 using a numeric indicator (see Figures 13.11 and 13.13).

The For Loop operated continuously since it was called from inside a While Loop (Figure 13.13). Therefore, a new ratio value was calculated every two seconds. A stop button was placed on the front panel of the VI-3 (top right hand corner) so that the user could stop the acquisition of data at any time (see Figure 13.11).

Estimation of Oesophageal SpO_2

The ratio values that were calculated every two seconds were passed into a shift register. A shift register could transfer values from one iteration of a loop to the next. The shift register comprised a pair of terminals (see Figure 13.14) directly opposite to each other on the vertical sides of a While or a For loop. The right terminal stored the data upon the completion of an iteration. The data was then shifted at the end of the iteration and appeared on the left terminal at the beginning of the next iteration (Figure 13.14). Shift registers can be configured to remember values from several iterations, as shown in Figure 13.14. This feature can be very useful when averaging of data points in different iterations is needed. Shift registers should always be initialised with a specific value. Initialisation of the shift register is done by wiring a specific value to the left terminal from outside the loop.
After the *For Loop* shown in Figure 13.13 was executed (20 iterations) a value of $R$ was calculated which was passed into the *shift register* from the right to the left terminal located on the vertical wall of the *While Loop* of VI-3 (Figure 13.15). The *shift register* used in VI-3 was configured to remember (hold) five previously calculated ratio values. Figure 13.15 illustrates clearly the whole *VI* algorithm that let to the estimation of Oesophageal SpO$_2$ (also see the complete block diagram of VI-3 in Figure 10.10). The five values of the calculated ratios located at the left terminals of the shift register (see Figure 13.15) were then passed into a *Boolean Case Structure*. This *case structure* consisted of two *subcases* (TRUE or FALSE) one of which executed depending on the value of the Boolean condition wired to the *selector terminal* (TRUE subcase is shown in Figure 13.15). If the *selector terminal* was wired to a true condition the case structure was TRUE, and the algorithm inside the TRUE subcase was executed. Similarly, the same happened when the selector terminal was wired to a false condition, then the case structure was FALSE, and the algorithm inside the FALSE subcase was executed. The input to the *selector terminal* of the *case structure* was coming from a logic diagram configured in such a way that produced a TRUE condition if all five ratio values at the shift register inputs at the left terminal were within a set limit. This limit of ratio values reflected saturation values within the range of 100% (ratio value of 0.4) to 80% (ratio value of 1.1) taken
from an empirical calibration curve (WEBSTER, 1997). Each one of the ratio values was connected to an in range comparison function (Figure 13.15).

The in range comparison function returned a TRUE if $R$ (ratio value input) was greater than or equal to $l_0$ (0.4) and less than or equal to $h_i$ (1.1). Otherwise the function returned a FALSE. The outputs of the in range comparison functions were connected into a group of six AND Boolean functions (Figure 13.15). An AND Boolean function computes the logical AND of the two inputs (produces a TRUE output only when both inputs are TRUE). The sixth AND Boolean function was connected to the selector terminal of the Boolean case structure (Figure 13.15).  

![Figure 13.15](image.png)

**Figure 13.15**: Part of the Virtual Instrument 3 illustrating the algorithm for the estimation of Oesophageal SpO$_2$ when the Boolean case structure is true

The logical operation of the combination of AND Boolean functions was to produce a TRUE at the selector terminal if all five ratio values were within the preset range (0.4 to 1.1). If any of the five ratio values was not, then the final output of the sixth AND Boolean function would produce a FALSE at the input of the selector terminal.

In the case of a TRUE condition the algorithm inside the TRUE subcase (the one shown in Figure 13.15) was executed. Firstly the ratio values connected
to the case structure were passed into a compound arithmetic Boolean function which added the five ratio values together (Figure 13.15). The sum of the ratio values was then passed into a division numeric function, which divided the sum of the ratio values, by five producing the mean ratio value (Figure 13.15). The mean ratio value was then passed into a formula node. The formula node is a resizable box similar to the boolean case structure (previously described). Instead of containing a subdiagram, however, the formula node contains one or more formula statements. The formula shown in the formula node in Figure 13.15 was the empirically derived calibration equation (13.2) for estimating SpO\(_2\) when the ratio value is known. The formula node consisted of one input variable (ratio (R) value) located at the left vertical wall of the formula node and one output variable (SpO2) located at the right vertical wall of the formula node (Figure 13.15). Therefore, on execution of the formula node the blood oxygen saturation value was calculated. The calculated SpO\(_2\) value was then transferred to the output variable (SpO2) of the formula node (located on the right vertical wall of the formula node). The calculation of SpO\(_2\) was continuous (every two seconds) as long as the criteria for the ratio value, discussed above were satisfied. The output variable (SpO2) of the formula node was then connected to a two parameter Format into String function (Figure 13.15). A Format into String function converts input arguments into a string, whose format is determined by format string (Figure 13.15). The format of the string was configured to be “%.2f %s” (see Figure 13.15) which means:

“%” begins the formatting specification
“.” separates width and precision
  width (optional), specifies the minimum character field width of the output. In this configuration it was missing which means the output is as long as necessary to contain the converted input parameter.
  precision (optional), specifies the number of digits to the right of the decimal point. In this configuration it was set to 2.
“s” space

The calculated SpO\(_2\) value was connected to the first parameter of the Format into String function corresponding to a double precision floating number (DBL) (see Figure 13.15). The other parameter of the Format into String function was connected to a string variable (abc) corresponding to the
symbol “%”. The output of the *Format into String function* was connected to a string indicator located outside the *boolean case structure* (Figure 13.15). The final output of the *boolean case structure* when the condition at the *selector terminal* was true was the calculated SpO₂ number presented in the format described above with a % symbol at the end of the number (i.e. 98.75 %). The value of SpO₂ was displayed on the front panel of *VI-3* under the label *Oes. SpO₂* (see Figure 13.11).

If any of the five successive ratio values R in the shift register was not within the preset range (see Figure 13.15) then the output of the sixth *AND Boolean function* produced a FALSE at the input of the *selector terminal* (Figure 13.16). In this case the *Boolean Case Structure* executed the algorithm inside the FALSE subcase (Figure 13.16). The algorithm inside the FALSE subcase in *VI-3* was a string constant connected to the string indicator at the output of the *Boolean case structure* (Figure 13.16). The string constant was preset with the word “processing”. This meant that every time one of the ratio values was out of range the estimation of SpO₂ was not possible and the user was warned with the message “processing” on the front panel of *VI-3*.

*Figure 13.16*: Part of the *Virtual Instrument 3* illustrating the algorithm for the estimation of Oesophageal SpO₂ when the Boolean case structure is False
13.7.5 Development of a Battery Condition Algorithm

This algorithm was developed with the aim to alert the user of a low battery condition on the front panel of the Virtual Instrument 3. The part of the block diagram of VI-3, which illustrates the battery condition algorithm is shown in Figure 13.17.

![Image of battery condition algorithm](image)

**Figure 13.17:** Battery condition algorithm

The output of the *index array* representing the attenuated (by 50%) sum of the voltages from the two batteries (channel (0)) was connected to a *Max & Min Array*, which returned the maximum value from the 10 samples in the index array per iteration of the *For Loop* (Figure 13.17). This maximum value was then multiplied by two using a *multiply control function* (Figure 13.7). A numeric indicator on the front panel of VI-3 displayed continuously the true voltage of the power supply of the Processing System 4 or 4a. The output of the *multiply* function was passed into two *comparison* functions (*Greater and Less or Equal*) (Figure 13.17). Both comparison functions were also connected to a numeric constant set at 9.8. The 9.8 numeric constant represented the low voltage limit of the power supply of the Processing System 4 or 4a. At 9.8 V the functionality of the Processing System 4 was not compromised. The level of voltage that the Processing System 4 or 4a would fail to operate (at least one of the channels would fail) was at 9.0 V. Therefore the 9.8 V limit provided an adequate warning for the user indicating...
that the batteries needed charging. Bench testing showed that for the battery voltage level to drop from 9.8 V (warning status) to 9.0 V (failing) it would take two hours of continuous operation of the Processing System 4 or 4a.

The Greater comparison function returned a TRUE if the battery signal was greater than 9.8. The output of the Greater comparison function was connected to a square flat LED boolean indicator (Figure 13.17). A text message was written inside the area of the flat square LED on the front panel saying “Battery OK”. Therefore, as long the battery signal was larger than the reference voltage limit (9.8) the front panel was displaying continuously the message “Battery OK” (see Figure 13.11). When the output of the Greater comparison function was FALSE then the flat square LED was off.

The Less or Equal comparison function operated the opposite way of the Greater comparison function described above, and returned a TRUE when the battery signal was less or equal to the reference voltage limit of 9.8. The output of the Less or Equal comparison function was connected to a Boolean Case Structure (Figure 13.17). This case structure consisted of two subcases (TRUE or FALSE) one of which executed depending on the value of the Boolean condition wired to the selector terminal (TRUE subcase is shown in Figure 13.17). If the selector terminal was wired to a true condition the case structure was TRUE, and the algorithm inside the TRUE subcase was executed. Similarly, the same happened when the selector terminal was wired to a false condition, then the case structure was FALSE, and the algorithm inside the FALSE subcase was executed.

When the Less or Equal comparison function returned a TRUE, the algorithm inside the TRUE subcase was executed. The algorithm inside comprised a square wave generator (generates an array containing a square wave) where the input frequency (1 Hz) and amplitude (5.0) of the square wave are determined by the numeric controls connected to the inputs of the Square wave (Figure 13.17). The output of the Square wave was connected to an Index Array, which extracted the data of the square wave into a row of numbers (Figure 13.17). Those numbers representing the square wave were
passed into an *Equal* comparison function. The other input of the *Equal* comparison function was connected to a numeric constant set at 5. The *Equal* comparison function returned a TRUE if the output of the *Index Array* was equal to 5 otherwise it returned a FALSE. Since the running frequency of the Square wave was set at 1 Hz, the output of the index array was alternating between HIGH (5.0) and LOW (0) levels continuously at a rate of 1 Hz. The output of the *Equal* comparison function was connected to another square flat LED boolean indicator. A new text message was written inside the area of the flat square LED on the front panel saying “Low Battery”. Therefore, as long the battery signal was less than or equal to the reference voltage limit (9.8 V) the front panel was displaying continuously the flashing message (at a rate of 1 Hz) “Low Battery”. When the output of the *Less or Equal* comparison function was FALSE then the case structure was FALSE, and the algorithm inside the FALSE *subcase* was executed. There was no algorithm inside the FALSE *subcase*, therefore the flat square LED was off permanently and nothing was displayed on the front panel of the *VI-3*. Therefore, the battery condition algorithm kept the user continuously informed of the condition of the power supply of the *Processing System 4 or 4a*.

**13.8 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT VERSION 3a (VI-3a)**

**13.8.1 Introduction**

The *Virtual Instrument 3* was further optimised by introducing an algorithm for calculating heart rate. This new version of *Virtual Instrument (VI-3a)* was identical to *VI-3* apart from the heart rate algorithm, which will be the subject of this section.

The heart rate was derived using the finger infrared AC PPG signal. The algorithm used calculated the power spectrum of the PPG signal. The period of the signal was estimated and then was multiplied by 60 to give the number of pulses or beats per minute which reflected heart rate since the PPG signal...
follows the R (component of the QRS complex) wave of the electrical heart activity. The details of this algorithm are described below.

13.8.2 Heart Rate Algorithm

Figure 13.18 shows the part of the VI-3a block diagram (identical to the block diagram of VI-3 (see Figure 13.10)) that was used to calculate the heart rate. The finger IR AC PPG signal (inside the FOR Loop) was passed into the signal (time domain signal in volts) input of an auto power spectrum VI (Figure 13.18). The auto power spectrum VI computed the single-sided, scaled power spectrum of the AC PPG time-domain signal. The specifications of the power spectrum VI indicates that at least three cycles of the signal must be contained in the time-domain signal (finger AC PPG) for a valid estimate of the power spectrum. To satisfy the requirements of the power spectrum VI the number of iterations of the For Loop were changed from 20 (used by VI-3) (200 samples; two cycles) to 30 (used by VI-3a) (300 samples; three cycles).

The dt input of the auto power spectrum VI (see Figure 13.18) was connected to the actual scan rate output of the AI START VI (see Figure 13.10). dt represented the sample period of the time domain signal (finger IR AC PPG) in seconds. dt was also 1/fs where fs was the sampled frequency of the time domain signal (100 samples/s). The auto power spectrum VI computed the single-sided, scaled auto power spectrum in volts rms squared. This VI computed the power spectrum as:

$$\frac{\text{FFT}^*(\text{signal}) \times \text{FFT(\text{signal})}}{N^2}$$

where FFT is the Fast Fourier Transform of the time domain signal, N is the number of points in the signal array (300) and * denotes complex conjugate. The output of the Auto Power Spectrum VI was connected to the input of a Power and Frequency Estimate VI. This VI computed the peak frequency of the time-domain signal (Figure 13.18).
The *For Loop* operated continuously, since it was called from inside a *While Loop* (Figure 13.18), until the stop button on the front panel of the *VI-3a* was pressed to stop the acquisition. Therefore, a new peak frequency was calculated every three seconds (300 samples at 100 samples/s).

![Figure 13.18: Block diagram illustrating the algorithm for calculating Heart Rate (HR)](image)

After the *For Loop*, shown in Figure 13.18, was executed (30 iterations) a peak frequency was calculated which was passed into the *shift register* (see detail description in section 13.7.4) from the right to the left terminal located on the vertical wall of the *While Loop* of *VI-3a* (Figure 13.18). The *shift register* used in *VI-3a* was configured to remember (hold) two calculated frequency values. The two values of the calculated peak frequencies located at the left terminals of the *shift register* (see Figure 13.18) were then passed into a *Boolean Case Structure*. This *case structure* consisted of two *subcases* (TRUE or FALSE) one of which executed depending on the value of the Boolean condition wired to the *selector terminal* (TRUE subcase is shown in Figure 13.18). If the *selector terminal* was wired to a true condition...
the case structure was TRUE, and the algorithm inside the TRUE subcase was executed. Similarly, the same happened when the selector terminal was wired to a false condition, then the case structure was FALSE, and the algorithm inside the FALSE subcase was executed. The input to the selector terminal of the case structure was coming from a logic diagram configured in such a way that produced a TRUE condition if both frequency values at the shift register inputs at the left terminal were within a set limit. This limit of frequency values were 0.5 Hz to 1.5 Hz which reflected heart rate values 120 beats per minute (bpm) (tachycardia) and 40 bpm (bradycardia) respectively (Figure 13.18). Each one of the frequency values was connected to an in range comparison function (Figure 13.18).

The in range comparison function returned a TRUE if $x$ (peak frequency input) was greater than or equal to $lo$ (0.5 Hz) and less than or equal to $hi$ (1.5 Hz). Otherwise the function returned a FALSE. The outputs of the two in range comparison functions were connected into an AND Boolean function (Figure 13.18). The AND Boolean function was connected to the selector terminal of the Boolean case structure (Figure 13.18). The logical operation of the AND Boolean function was to produce a TRUE at the selector terminal if both peak frequency values were within the preset range (0.5 Hz to 1.5 Hz). If either of the two peak frequency values was not within the range, then the final output of the AND Boolean function would produce a FALSE at the input of the selector terminal.

In the case of a TRUE condition the algorithm inside the TRUE subcase (one shown in Figure 13.18) was executed. Firstly the peak frequency values connected to the case structure were passed into a compound arithmetic Boolean function which added the two peak frequencies together (Figure 13.18). The sum of the peak frequency values was then passed into a division numeric function, which divided the sum of the peak frequency values by two producing the mean peak frequency value (Figure 13.18). The mean peak frequency value was then passed into reciprocal control function, which divided one by the input frequency, therefore calculating the period of the ECG signal. The period of the signal was then multiplied by 60 using a
multiply numeric function, which resulted the number of heart beats per minute (bpm). The calculated heart rate (HR) value was then connected to a two parameter Format into String function (Figure 13.18). The format of the string was configured to be “%.0f %s” (see Figure 13.18) which means:

“%” begins the formatting specification
“.” separates width and precision
width (optional), specifies the minimum character field width of the output. In this configuration it was missing which means the output is as long as necessary to contain the converted input parameter.
precision (optional), specifies the number of digits to the right of the decimal point. In this configuration it was set to zero.
“s” space

The calculated HR value was connected to the first parameter of the Format into String function corresponding to a double precision floating number (DBL) (see Figure 13.18). The other parameter of the Format into String function was connected to a string constant corresponding to the symbol “bpm”. The output of the Format into String function was connected to a string indicator located outside the boolean case structure (Figure 13.18). The final output of the boolean case structure when the condition at the selector terminal was true was the calculated HR number presented in the format described above with a bpm symbol at the end of the number (i.e. 60 bpm). The value of HR was displayed on the front panel of VI-3a under the label HR bpm (see Figure 13.19)

If any of the two frequency values was not within the preset range then the output of the AND Boolean function produced a FALSE at the input of the selector terminal (Figure 13.20). In this case the Boolean Case Structure executed the algorithm inside the FALSE subcase (Figure 13.20). The algorithm inside the FALSE subcase in VI-3a was a string constant connected to the string indicator at the output of the Boolean case structure (Figure 13.20). The string constant was preset with the word “processing”. This meant that every time one of the peak frequency values was out of range the estimation of HR was not possible and the user was warned with the message “processing” on the front panel of VI-3a.
Figure 13.19: Front panel of the PPG Signal Acquisition Virtual Instrument 3a

Figure 13.20: Part of the Virtual Instrument 3a illustrating the algorithm for the calculation of Heart Rate when the Boolean case structure is False
13.9 DEVELOPMENT OF THE PPG SIGNAL ACQUISITION VIRTUAL INSTRUMENT VERSION 3b (VI-3b)

13.9.1 Introduction
The Virtual Instrument 3a was further optimised to make a new version called Virtual Instrument 3b (VI-3b). This VI-3b was the final version that was used for the detailed clinical measurements on patients undergoing cardiopulmonary bypass surgery. This new VI-3b introduced a new algorithm that allowed the online digital filtering of the Oesophageal and Finger red and infrared AC PPGs and ECG signals. The heart rate algorithm developed in the previous version of Virtual Instrument 3 (version 3a) was removed since it was decided that monitoring heart rate was not of primary interest in this study. The elimination of the heart rate algorithm allowed the calculation of oesophageal and finger SpO₂ every two seconds as was originally designed in VI-3 instead of every three seconds as in VI-3a. Despite the removal of the HR algorithm VI-3b still displayed ECG traces.

Therefore, the Virtual Instrument 3b incorporated algorithms for providing signal acquisition, data storage, online estimation of oesophageal and finger SpO₂, battery condition monitoring, and online digital filtering. All algorithms apart from the online digital filtering remained identical to those described in VI-3 and VI-3a and will not be discussed in this section.

13.9.2 Theory of Digital Filters
A brief discussion in digital filters is given in this section before the description of the digital filtering algorithm.

Modern sampling and digital signal processing tools have made it possible to replace analogue filters with digital filters in applications that require flexibility and programmability. Digital filters have the following advantages over their analogue counterparts.

- They are software programmable
- They are stable and predictable
They do not drift with temperature, humidity or require precision components

Digital filters can be used in LabVIEW when it is possible to control parameters such as filter order, cut-off frequencies, amount of ripple and stopband attenuation. When designing a digital filter the sampling theory must be taken into consideration. The sampling theory states that a continuous-time signal can be reconstructed from discrete, equally spaced samples if the sampling frequency is at more than twice that of the highest frequency in the time signal. Antialiasing analogue filters (upper cut-off of 20 Hz) were introduced in the Processing System 4 described earlier in section 13.2.2.

The digital filters used in VI-3b were Infinite Impulse Response Filters (IIR). Infinite Impulse Response Filters are digital filters whose impulse response can theoretically be infinite in duration. Some of the properties of IIR filters in LabVIEW are as follows:

- They check for the validity of the Nyquist criterion.

- Because the initial filter state is assumed to be zero, a transient proportional to the filter order occurs before the filter reaches steady state. The duration of the transient response, or delay, for lowpass and highpass filters is equal to the filter order. The duration of the transient response for bandpass and bandstop filters is twice the filter order.

13.9.3 Digital Filter Algorithm

The digital filter algorithm developed in VI-3b allowed the user to select from the front panel of VI-3b (see Figure 13.21) the type of the IIR filter (Bessel, Butterworth or Chebyshev), the type of the passband (lowpass, highpass, bandpass, or bandstop), as well as several other parameters. The block diagram illustrating the digital filter algorithm applied to the oesophageal AC red PPG signal is shown in Figure 13.22.
Figure 13.21: Front panel of the Virtual Instrument 3b (VI-3b)
The same algorithm was also applied to the Oesophageal infrared AC PPG, Finger AC red and infrared PPG signals and the ECG signal.

The output of the *index array* containing the oesophageal red AC PPG signal was passed into a *case structure* (Figure 13.22). This *case structure* consisted of three *subcases* (Butterworth, Bessel, Chebyshev) one of which executed depending on the condition at the *selector terminal* of the case structure (the Butterworth sublease is shown in Figure 13.22). The *selector terminal* of the *case structure* was connected to an *enumerated list control* (Figure 13.22).

![Figure 13.22: Block diagram of the digital filter algorithm](image)

This *list control* contained three different text options (Butterworth, Chebyshev and Bessel) corresponding to the three digital filters used in the algorithm. The format of the list control was like a drop menu where the user could scroll with the mouse and select the desired filter. For example if the Butterworth option in the *list control* was selected by the user at the front panel of VI-3b, then the *case structure* executed the algorithm in the *subcase*
corresponding to a Butterworth filter. The filter VIs corresponding to the three digital filters were embedded in each subcase, with each one corresponding to the type of filter indicated by the enumerator list control on the front panel of VI-3b. This is also illustrated graphically in Figure 13.23.

**Figure 13.23:** Illustration of the three subcases containing the filter VIs
The filter parameters (filter type, lower and high cut-off frequencies, filter order, passband ripple, and stopband attenuation ripple) were all controlled by the user from the front panel of VI-3b (Figure 13.21). All parameters were connected to the Filter VIs from outside the subcases (Figure 13.23). That allowed each of the three Filter VIs inside the subcases to connect only to the necessary parameters corresponding to the chosen Filter VI. Not all Filter VIs used the same parameters (Figure 13.23). The parameters that were not used stayed floating (unwired to the Filter VI) without affecting the functionality of the digital filter algorithm.

The Butterworth filter was used for all the clinical measurements performed with the Processing System 4. The configuration of the parameters for this filter is discussed briefly below.

**Filter type:** specified the passband of the filter according to the values; Lowpass (0), Highpass (1), Bandpass (2), and Bandstop (3). In VI-3b the filter type was set as a Bandpass (2).

**Input Signal (x):** Corresponded to the Oesophageal AC red PPG signal

**Sampling Frequency (f_s):** set at 200 samples/s

**High cut-off frequency:** Set at 15.0 Hz

**Low cut-off frequency:** Set at 0.45 Hz

**Filter Order:** Set at 2

**Filtered X:** Corresponds to the output Oesophageal AC infrared PPG array of filtered samples

The settings of the parameters were the same for the Oesophageal Infrared AC PPG channel and Finger AC red and infrared PPG channels. For the ECG channel the only parameters that were different were the low cut-off frequency (0.2 Hz) and the high cut-off frequency (20 Hz).
13.10 EVALUATION OF THE HARDWARE AND SOFTWARE DEVELOPMENTS

13.10.1 Evaluation of the Hardware and Software Stages

The functionality of all the electronic circuits and the software developments, as described in this chapter were successfully bench tested in the laboratory.

*Processing System 4 and Virtual Instrument 3 (VI-3)*

The *PPG Processing System 4* was connected to the laptop computer using a ribbon cable. This cable connected the 68-pin PCMCIA bus connector of the DAQCard--AI-16X-50 in the laptop and was terminated with a 50-pin female connector that attached on the 68-pin connector on the *PPG Processing System 4*.

The *Oesophageal PPG Probe II* the Finger PPG probe and the 3-lead ECG cable were connected to the three DIN connectors at the front panel of the *Processing System 4*. The oesophageal probe was inserted into a transparent oesophageal tube. An index finger (right) was placed flat on the plastic stomach tube on top of the probe facing the optical components of the probe. The PPG finger probe was clipped on the left index finger and the ECG leads were connected to the right and left arms and right leg using ECG skin surface electrodes. The *Signal Acquisition Virtual Instrument 3 (VI-1)* implemented in *LabVIEW* was loaded and the *PPG Processing System 4* was switched on. PPG and ECG traces were recorded for approximately 30 minutes and displayed on the laptop screen (Figure 13.24). During the monitoring time the oesophageal and finger channel SpO₂ values were calculated every two minutes and displayed on the front panel (Figure 13.24) of VI-3. Also, the battery condition algorithm executed successfully by displaying the battery voltage status on the front panel of VI-3 (Figure 13.24).
Figure 13.24: Front panel of VI-3 after execution using the Processing System 4

Processing System 4 and Virtual Instrument 3a (VI-3a)
The new Virtual Instrument 3a was tested using the same setup as the one described above for the Processing System 4 and VI-3. Both the processing system 4 and VI-3a operated successfully. The heart rate algorithm executed as described in section 13.8.2 resulting of a heart rate value being calculated every three seconds.

Processing System 4a and Virtual Instrument 3b (VI-3b)
The new PCB Processing System 4a was tested in the laboratory together with the final version of Virtual Instrument 3, Virtual Instrument 3b. The setup was identical to the one described above with the Oesophageal and Finger Probes and the ECG connected on the front panel of the Processing System 4a. The filtered (using the digital filter algorithm) PPG and ECG signals were recorded successfully for approximately 30 minutes and displayed on the front panel of VI-3b (Figure 13.25). Oesophageal and Finger SpO₂ values (updated every two seconds) and battery status signals were also displayed on the front panel of VI-3b (Figure 13.25).
13.10.2 Electrical Safety Test for PPG Processing Systems 4 and 4a

The PPG Processing System 4 and 4a were electrically safety tested using the Rigel Safety Tester Model 233 (see section 11.4.1). The insulation resistance between the patient applied parts (probes) and the output circuitry was measured for the oesophageal and finger PPG channels and ECG channel. Also, the current flowing across the isolation barrier with mains voltage on the applied parts, and with mains voltage at the output terminals was measured for all three channels. The procedures followed for both electrical safety tests (insulation resistance and mains on applied parts) were identical to the ones applied for the PPG Processing Systems 2 and 3 (see sections 11.4.1 and 12.11.2).

The insulation resistance between the input and output circuits of the Processing System 4 and 4a was too large to register on the tester meter, but was well in excess of 100 MΩ for each channel. The test with mains on the patient applied parts and on the output gave leakage currents across the
isolation barrier of less than 9 \( \mu A \) in all cases. This was well below the 50 \( \mu A \) current limit for Class 1 and 2 type CF instruments.

The *PPG Processing System 4a* and the *Virtual Instrument VI-3b* were used in the study of patients undergoing high risk surgery (cardiopulmonary bypass surgery), who had an arterial cannula inserted routinely, therefore allowing rigorous validation of oesophageal SpO\(_2\) measurements with co-oximetry and blood gases. This system also facilitated comparison studies between the oesophageal PPG channel with commercial finger transmission pulse oximeters. The clinical methods and the results from all PPG and SpO\(_2\) measurements will be presented in the next chapter.
CHAPTER 14

Investigation of Oesophageal Pulse Oximetry Using *Processing System 4a* and *Virtual Instrument 3b* in Adult Thoracic Surgery Patients

14.1 INTRODUCTION

The PPG *Processing System 4a* and the *Virtual Instrument 3b* were used to investigate patients undergoing high-risk operations, such as hypothermic cardiothoracic bypass surgery, in whom conventional pulse oximetry might fail postoperatively due to poor peripheral circulation. These patients had an arterial cannula inserted routinely, which allowed the rigorous validation of the oesophageal SpO₂ measurements with CO-Oximetry. The oesophageal SpO₂ values which were continuously being estimated (every two seconds) as described in chapter 13, were also compared with blood oxygen saturation values obtained from blood gas analysis (BGA) and from a commercial finger pulse oximeter.

Moreover, the study on cardiac patients provided data with which to address one of the fundamental questions put forward at the beginning of the investigation of oesophageal PPG signals. The question was: Are the Oesophageal PPG signals due entirely to pulsatile blood flow in arteries or are they influenced (partly or completely) by a cardiac or an aortic movement artefact?

The clinical methods and the results from all PPG investigations and SpO₂ measurements are presented and discussed in the following sections.
14.2 PATIENTS AND MEASUREMENTS

Local research ethics committee approval was obtained prior to commencing the study of ASA 2, 3 and 4 anaesthetised patients. Fifty adult patients (41 male and 9 female) aged 26 to 81 years undergoing elective cardiothoracic surgery were recruited to the study. Table 14.1 summarises the number of patients per type of operation. Patients with oesophageal pathology or previous surgery to the oesophagus or stomach were excluded.

Table 14.1: Patient Details

<table>
<thead>
<tr>
<th>Number of Patients</th>
<th>Type of operation</th>
</tr>
</thead>
<tbody>
<tr>
<td>36</td>
<td>Coronary Artery Bypass Graft (CABG) (continuous flow)</td>
</tr>
<tr>
<td>2</td>
<td>CABG (pulsatile flow)</td>
</tr>
<tr>
<td>7</td>
<td>Heart Valve Reconstruction</td>
</tr>
<tr>
<td>2</td>
<td>Heart Valve Reconstruction +CABG</td>
</tr>
<tr>
<td>1</td>
<td>Excision of Cervical Rib</td>
</tr>
<tr>
<td>1</td>
<td>Thoracotomy</td>
</tr>
<tr>
<td>1</td>
<td>Thoracoscopy</td>
</tr>
</tbody>
</table>

Intravenous access was gained with a 14-gauge cannula in a forearm and a 20-gauge arterial cannula was placed in a radial artery to allow continuous blood pressure monitoring during induction of anaesthesia.

Anaesthesia was induced with midazolam and etomidate followed by a dose of rocuronium or pancuronium. Fentanyl was used for analgesia. The trachea was intubated and the lungs were mechanically ventilated. Anaesthesia was maintained using isoflurane (approximately 1.2% inspired concentration) in a 1:2 mixture of oxygen and nitrous oxide. Heart rate, systolic and diastolic blood pressures were monitored continuously. Central temperature was measured from the nasopharynx and peripheral temperature from the left shoulder tip. The oesophageal PPG probe was inserted into a sealed (see section 9.3.2) 20 French gauge stomach tube, which was lubricated with aqueous gel. The tube was then inserted through the mouth into the oesophagus under direct vision. The stomach tube was advanced into the oesophagus until the end of the probe itself was 30 cm from the lips. A 30 cm
marker was positioned on the plastic oesophageal tube prior to insertion in the oesophagus (Figure 14.1).

![Image of a patient with an oesophageal probe inserted](image)

**Figure 14.1:** Photograph of an anaesthetised cardiopulmonary bypass patient in the induction room with the *Oesophageal PPG Probe II* placed in the oesophagus (prior to cannulation) at 30 cm from the lip (reproduced with consent of patient)

The right internal jugular vein was cannulated using a triple lumen cannula to allow central venous pressure monitoring. PPG signals were observed at various depths of the oesophagus as the probe was withdrawn, until the site of best signal (good signal-to-noise PPGs and acceptable ventilator artefact) was determined. The monitoring depth was measured using a ruler by measuring the distance from the 30 cm marker to the lip and then subtracting the measured value from 30 to result in the actual oesophageal depth of the probe (see Figure 14.2). The probe was then left at this depth for the duration of surgery unless sternotomy adversely affected the signal. PPG traces and SpO₂ values from the oesophagus and the finger, along with ECG traces, were recorded simultaneously. During the oesophageal and finger measurements, values of blood oxygen saturation from a commercial transmission type finger pulse oximeter (Markette, Tram 200A; Markette Electronics, Milwaukee, Wisconsin) were also recorded.
Figure 14.2: Measuring of the oesophageal monitoring depth using a ruler (reproduced with consent of patient)

Figure 14.3: Photograph showing the positioning of the Processing System 4a and the laptop in the operating theatre (reproduced with consent of patient)
Monitoring with the *Processing System 4a* was temporarily stopped during transportation of the patient from the induction room into the operating theatre. In the operating theatre the laptop computer and *Processing System 4a* were positioned on a theatre trolley approximately 1.5 m away from the patient, next to the main theatre monitoring computer system (see Figure 14.3). Monitoring in theatre was intermittent. Oesophageal, finger and ECG data were collected at the following times:

- At least 10 to 15 minutes prior to skin incision. During this time the depth of the oesophageal probe was reconfirmed (measured with a ruler), since there was a possibility that it might have moved during transport from the induction room into the theatre. PPG signals were recorded at this depth and any adjustments of the probe depth were made if needed.

- After sternotomy, PPG and ECG signals were monitored for approximately 15 minutes. During this time the chest was opened and any changes in blood pressure due to alterations in intrathoracic pressures and their effect on the PPG signal were noted.

- Ten to fifteen minutes prior to bypass. PPG signals were recorded until cessation of cardiac activity on bypass on the heart lung machine, and when no pulsatile signals were seen on the screen of the laptop. The heart lung machine is a peristaltic pump (rotating at approximately 300 rpm) that pumps the blood around the body since the heart is completely stopped. It also oxygenates the blood and removes carbon dioxide. Basically, the heart lung machine takes over completely the cardiac and pulmonary function (see Figure 14.4).

- The monitoring started again approximately ten minutes before the patient was coming off bypass and continued for another thirty minutes after the patient was completely off bypass and stable cardiac activity was established.

- The recording of signals in theatre continued during the closure of the chest for approximately 15 to 20 minutes.
The last monitoring took place in the intensive care unit where the patient was monitored continuously for approximately 30 to 60 minutes. During this period the patients were peripherally cold and therefore potentially at risk of peripheral pulse oximeter failure.

During the above recording periods samples of arterial blood (approximately 1 ml/sample) were drawn into 2ml heparinised syringes and analysed immediately by an Instrumentation Laboratories IL 482 CO-Oximeter or an Instrumentation Laboratories IL BG-1400 Blood Gas Analyser (BGA) (Instrumentation Laboratories, Lexington, Massachusetts, USA) depending on availability. During the CO-Oximetry blood analysis simultaneous measurements of total haemoglobin (Hb) and oxyhaemoglobin (HbO₂) were obtained from each blood sample. The arterial cannula was flushed with heparinised 0.9% normal saline solution (1000 units/500 ml) after sampling. Care was taken to ensure that the arterial line and the blood-sampling syringes were free of air bubbles.

**Figure 14.4:** Diagram illustrating the set-up of the heart lung machine during cardiopulmonary bypass surgery
14.2.1 Data Analysis and Statistics
The collected arterial blood oxygen saturation values and the PPG data files recorded by the *LabVIEW Virtual Instrument 3a(VI-3a)* were analysed offline. Patients were only accepted into the PPG amplitude analysis if measurable PPGs were present at both wavelengths in the oesophagus. Also, patients were only accepted into the final arterial blood oxygen saturation statistical analysis if at least one set of data (collected at the same time) of CO-Oximetry or Blood Gas Analysis with oesophageal, finger and commercial finger SpO₂ values existed. The PPG amplitudes of the oesophageal AC PPG signals for all fifty patients were measured on printouts from *LabVIEW*, and the means and standard deviations (SE) calculated. From the fifty patients studied, patient 1 was excluded from the final blood oxygen saturation data analysis (he was included in the PPG amplitude analysis). The measurements on patient 1 were purely observational and blood oxygen saturation data were not noted.

Linear Regression analysis was used to compare the blood oxygen saturation results from all pulse oximeters used (oesophageal, finger (using custom made finger probe) and finger (using a commercial pulse oximeter)) with CO-Oximetry and blood gases, the two "gold standard" methods. Linear Regression analysis is the most suitable when calibrating an approximate or simple method (like the oesophageal pulse oximeter) with a very precise method or one that produces a “true” value (like the CO-Oximeter or blood gas analyser) (BLAND and ALTMAN, 1986).

The “gold standard” for the calibration of pulse oximeters is the CO-Oximeter. The CO-Oximeter requires small samples of heparinised arterial blood from which, it provides oxygen saturation readings only for the instant at which the blood was withdrawn (MOYLE, 1994). The CO-Oximeter measures the oxygen saturation by a spectrophotometric technique as in pulse oximetry, but none of the commercial CO-Oximeters uses the same wavelengths as current commercial pulse oximeters (660 nm and 940 nm). Most CO-Oximeters operate in the visible range at four wavelengths (535 nm, 585.2 nm, 594.5 nm and 626.6 nm) (WEBSTER, 1997). CO-Oximeters analyse the
concentrations of different types of haemoglobin including deoxyhaemoglobin (Hb), oxyhaemoglobin (HbO2), carboxyhaemoglobin (COHb), and methaemoglobin (MetHb) (WEBSTER, 1997).

The blood gas analyser (BGA) is consider in the clinical world as a second “gold standard” in the calibration of pulse oximeters. The blood gas analyser has electrodes, which measure the partial pressures of oxygen and carbon dioxide and also the concentration of hydrogen ions in plasma (MOYLE, 1994). The oxygen is usually measured with a Clark polarographic electrode, and the carbon dioxide by a Severinghaus electrode. The value of oxygen saturation is derived from the measured oxygen pressure (PO2), taking into account the effects of carbon dioxide pressure (PCO2), pH, and temperature, and also assuming healthy adult human haemoglobin plasma (MOYLE, 1994).

The levels of agreement between the oesophageal and commercial finger probes as well as between the finger and the commercial finger probes were calculated using the between-method differences analysis outlined by ALTMAN and BLAND (1983). This type of comparison is different from calibration, where quantities are measured by a new method (i.e. oesophageal pulse oximeter) and the result compared with measurements made by a highly accurate method (i.e. CO-Oximetry). The Bland and Altman method suggests that the best way to look for an association between the two methods is to plot the difference between the methods against their mean. If there is no obvious relation between the difference and the mean then the lack of agreement can be summarised by calculating the bias, estimated by the mean difference (d) and the standard deviation of the differences (s). Provided differences within $d \pm 2s$ would not be clinically important we could use the two measurement methods or instruments interchangeably (BLAND and ALTMAN, 1986).
14.3 RESULTS FROM THE INVESTIGATION OF PPG SIGNALS IN CARDIAC PATIENTS

Measurable PPG traces at both wavelengths were obtained in the oesophagus of all fifty patients. Figure 14.5 depicts typical traces from one patient undergoing cardiopulmonary bypass surgery during the various monitoring periods as described above (monitoring depth at 17 cm). Figure 14.5a shows oesophageal and finger AC PPGs, obtained at both wavelengths, and ECG signals recorded after arterial cannulation. The signals in Figure 14.5b were recorded just before sternotomy. In Figure 14.5c the signals were recorded after the chest was open. Figure 14.5d shows the transition from non bypass to bypass time. When the heart lung machine was switched on (indicated in the Figure “on bypass”) the pulsatile PPG signals disappeared within the next 15 to 20 seconds. Figures 14.5e and 14.5f are showing PPG and ECG signals after bypass, during closing of the chest and postoperatively in the intensive care unit, respectively.
Figure 14.5: Oesophageal, Finger and ECG traces obtained from an anaesthetised patient undergoing cardiopulmonary bypass surgery; (a) during induction (after cannulation); (b) in operating theatre before sternotomy; (c) after sternotomy; (d) during bypass transition (before bypass, on bypass; (e) closing the chest; (f) in intensive care unit
The oesophageal PPG signals recorded from all patients (before and after bypass) were of good quality with large amplitudes and high signal-to-noise ratio. The monitoring oesophageal depth ranged from 14 cm to 28 cm, measured from the upper lip (mean ± SD: 17.8 cm ± 3.3 cm). Table 14.2 summarises the number of patients per monitoring depth.

Table 14.2: Summary of the oesophageal monitoring depth of the fifty patients

<table>
<thead>
<tr>
<th>Number of patients</th>
<th>Oesophageal Monitoring Depth</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>14 cm</td>
</tr>
<tr>
<td>6</td>
<td>15 cm</td>
</tr>
<tr>
<td>8</td>
<td>16 cm</td>
</tr>
<tr>
<td>9</td>
<td>17 cm</td>
</tr>
<tr>
<td>11</td>
<td>18 cm</td>
</tr>
<tr>
<td>7</td>
<td>19 cm</td>
</tr>
<tr>
<td>1</td>
<td>22 cm</td>
</tr>
<tr>
<td>3</td>
<td>27 cm</td>
</tr>
<tr>
<td>1</td>
<td>28 cm</td>
</tr>
</tbody>
</table>

Optimal oesophageal monitoring depth for each patient was considered the depth that oesophageal PPGs with good signal-to-noise ratio and acceptable
ventilator artefact could be obtained. The magnitude of the ventilator artefact at the depth range of 14 cm to 28 cm was of the order of 10% to 40% of the oesophageal PPG peak-to-peak amplitude. It was originally thought that with a ventilator artefact of that order it would be impossible to estimate continuously and accurately oesophageal blood oxygen saturation values unless the ventilator was switched off temporarily. But with practice it was found that ventilator artefact within the range of 10% to 40% of the oesophageal PPG peak-to-peak amplitude did not affect the continuous estimation of oesophageal SpO2.

Table 14.3 gives the mean ± SE of the AC PPG amplitudes at both wavelengths at the different oesophageal monitoring depths for the 50 patients. The amplitudes at the monitoring depths as described in Table 14.2 were separated into three groups; the upper oesophageal depths (14 cm to 17 cm), the mid oesophageal depths (18 cm to 22 cm), and the lower oesophageal depths (27 cm and 28 cm). The AC PPGs in the mid to lower oesophagus (depths of 18 cm or greater) had larger mean amplitudes at both wavelengths than those in the upper oesophagus (14 cm to 17 cm). These quantitative results were in close agreement with the more detail amplitude PPG study at five oesophageal depths of healthy anaesthetised patients described in section 12.13 (see Table 12.6). The error in these manual amplitude measurements was approximately ± 3%, which was insignificant compare with the physiological variability of the PPG signals at the various oesophageal depths.

Table 14.3: Mean ± SE of AC peak-to-peak PPG amplitudes (mv) at two wavelengths measured in the upper, mid and lower oesophagus

<table>
<thead>
<tr>
<th>Oesophageal depths</th>
<th>upper oesophagus (14 cm to 17 cm)</th>
<th>mid oesophagus (18 cm to 22 cm)</th>
<th>lower oesophagus (27 cm to 28 cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infrared (880 nm)</td>
<td>177 ± 18 mv</td>
<td>532 ± 72 mv</td>
<td>358 ± 42 mv</td>
</tr>
<tr>
<td>Red (655 nm)</td>
<td>69 ± 7 mv</td>
<td>222 ± 29 mv</td>
<td>183 ± 37 mv</td>
</tr>
<tr>
<td>Number of Patients (n)</td>
<td>27</td>
<td>19</td>
<td>4</td>
</tr>
</tbody>
</table>
14.4 RESULTS FROM AN INVESTIGATION TO DETERMINE THE CONTRIBUTION OF AORTIC OR CARDIAC MOVEMENT ARTEFACT TO THE OESOPHAGEAL PPG SIGNALS

The study of the thoracic patients, especially the patients undergoing cardiac surgery (n=47), where the heart was completely stopped and the circulation of the blood was taken over by the heart lung machine, confirmed that the oesophageal PPG signals were due to arterial pulsatile blood and not a product of cardiac or aortic movement artefact, as suggested by the hypothesis introduced in section 14.1 in this chapter. Three different investigations confirmed that the oesophageal PPGs were not movement artefacts and are described below.

14.4.1 Investigation of Oesophageal PPG signals during aortic cross-clamp

After the patient was on bypass the surgeon clamped the aorta. Therefore, no blood from the heart flowed into the aorta but instead the blood flowed through a plastic pipe, which was previously cannulated in the aorta, to the heart lung machine. This is illustrated schematically in Figure 14.4. At that moment (aortic cross-clamp) there was some heart electrical activity (visually confirmed) (see Figure 14.6) but there was no pulsatile blood circulating in the arterial system, therefore no pulsatile oesophageal PPG signals were recorded on the screen of the laptop. That proved that the oesophageal PPG signals were not cardiac movement artefacts since there were no oesophageal PPG signals recorded on the laptop despite the electrical activity (beating) of the heart. This observation was common to all patients undergoing cardiopulmonary bypass surgery.
Figure 14.6: Oesophageal and finger PPG signals during aortic cross-clamping; (a) illustrates the on bypass and aortic cross-clamp monitoring times; (b) after aortic cross-clamping, showing clearly the heart activity with no pulsatile oesophageal or finger PPG signals.
14.4.2 Investigation of Oesophageal PPG signals during pulsatile flow bypass

The second investigation was performed on two patients undergoing cardiopulmonary bypass surgery. In these two cases a different heart lung machine was used. This new heart lung machine was designed to operated with two different blood flow modes. One was the usual continuous flow mode (used routinely during most bypass surgery) and the other was pulsatile flow mode. In the pulsatile flow mode the machine basically imitates the heart by pulsing the blood in the arterial system the same way as the heart. The operator of the heart lung machine can set the pulsing rate to 60 pulses per second. Both of the patients went on bypass using the continuous blood flow mode. Figure 14.7a illustrates the transition between continuous flow and pulsatile flow of the heart lung machine (the blood flow modes can be changed during the operation). During the continuous flow time (after aortic cross-clamping) there was no heart activity and no pulsatile PPG signals (see Figure 14.7a).

![Figure 14.7: PPG signals during cardiopulmonary bypass surgery with the heart lung machine switched from continuous to pulsatile flow (measurements performed after aortic cross-clamping); (a) shows transition from continuous flow bypass to pulsatile flow bypass; (b) illustrates pulsatile PPG signals (oesophageal and finger) when there was no heart activity (during pulsatile flow bypass)](image-url)
When the machine was switched in the pulsatile flow mode, pulsatile PPGs (oesophageal and finger) appeared on the screen of the laptop (see Figure 14.7a and 14.7b). That again confirmed that the morphology of the oesophageal PPG signals was due to the pulsatile property of the arterial blood coming from the pulsatile flow heart lung machine and not due to a cardiac (since the heart was completely stopped, note the ECG on both Figures) mechanical movement.

14.4.3 Evidence from the quantitative estimations of Oesophageal SpO\(_2\)

The continuous estimation of oesophageal SpO\(_2\) in all 50 patients was another indication that the oesophageal PPGs were not mechanical movement artefacts but were signals due to arterial blood pulsations. The SpO\(_2\) values obtained from the oesophagus, as will be discussed in detail in the following section, were in agreement with SpO\(_2\) values obtained simultaneously from a commercial finger pulse oximeter. Also, the oesophageal SpO\(_2\) values agreed well with oxygen saturation values obtained from blood gas analysis using a CO-Oximeter (gold standard). If the oesophageal PPGs were just mechanical artefact signals caused by the
heart or the aorta then the likelihood of obtaining saturation values that were in agreement with the commercial oximeter or the CO-Oximeter would have been very small.

14.5 RESULTS FROM THE BLOOD OXYGEN SATURATION MEASUREMENTS

The blood oxygen saturation values obtained from the oesophageal and finger channels of Processing System 4a, and from the commercial finger pulse oximeter were collected and tabulated. Also the corresponding blood oxygen saturation values obtained from blood gas analysis using the blood gas analyser or the CO-Oximeter were also tabulated. A total of 36 values of oxygen saturation, from 17 patients, were obtained using the CO-Oximeter, and 155 values of blood oxygen saturation, from 49 patients, were obtained using the blood gas analyser. Measurements obtained from the CO-Oximeter were less than those obtained from the blood gas analyser due to technical and calibration problems with the CO-Oximeter.

14.5.1 Comparisons of blood oxygen saturation measurements between CO-Oximetry and oesophageal, finger, and commercial finger pulse oximeters

A total of 36 sets of data (oxygen saturation values) from 17 patients were used for the regression analysis, which gave the estimated slope and intercept of the regression line. An average (± SD) of 2.1 (± 1.2) blood samples were collected from each patient. CO-Oximetry measurements were not possible for all patients for the reasons mentioned above.

A comparison of SpO₂ readings obtained from the reflectance oesophageal pulse oximeter (y-axis) and the CO-Oximeter (x-axis) is shown in Figure 14.8. The equation of the best fit linear regression line was: \( y = -10.1 + 1.1 x \); \( r = 0.91 \); Standard Error of Estimate (SEE) = 0.71; \( P<0.001 \). The mean and standard deviation of the differences between the oesophageal pulse oximeter and the CO-Oximeter readings were +0.73 ± 0.72 %.
Figure 14.8: Comparison of SpO₂ measurements obtained from the Oesophageal Probe (y-axis) and the CO-Oximeter (x-axis) in 17 patients. y=-10.1+1.1x; r=0.91; SEE=0.71; n=36; P<0.001. The solid line represents the best fit linear regression line. The dashed line represents identity. The error bars represent oesophageal SpO₂ error of ± 0.8%.

The estimated measurement error in the oesophageal SpO₂ values was investigated. The main cause of error in these measurements was thought to be the ventilator artefact. A large modulation of the oesophageal PPG signals due to ventilator artefact could cause an error in the max and min amplitude of the PPG signal recorded by the algorithm for estimating the ratio (R) value and therefore an error in the final calculation of SpO₂ (see section 13.7.4). This algorithm detects the maximum (peak) and minimum (valley) amplitude of two PPG pulses (at both wavelengths) and then calculates the AC PPG component (of both wavelengths) by subtracting the maximum from the minimum value. Therefore, the modulation of the PPG signals due to ventilator artefact may result in AC peak-to-peak amplitude PPG values that do not represent the actual PPG signal. The measurement error was calculated for the range of ventilator artefacts, 15% to 30% of the PPG signal, which represents the range of ventilator artefact for the oesophageal depths 14 cm to 19 cm (monitoring depths for 45 patients). The order of SpO₂ error for the 15% and 30% ventilator artefact was 0.3% and 1.3%.
respectively. For ventilator artefacts greater than 40% (usually occurred at oesophageal depths of 20 cm or greater) the Processing System 4a was not able to estimate accurately and continuously an SpO₂ value. At these large ventilator artefacts (40% or greater) the calculated value of the ratio (R) was out of the specified range as described in section 13.7.4 and therefore the algorithm for estimating SpO₂ would continuously display the message "processing" at the front panel of VI-3b instead of an SpO₂ value. In all the fifty patients investigated the Oesophageal PPG Probe II was placed at oesophageal depths that the ventilator artefact was less then 40%. In five patients ventilator artefact of less than 40% was measured at oesophageal depths of greater than 22 cm. That shows that there is variability in ventilator artefact from patient to patient and it is not the rule that ventilator artefact is always greater than 40% at oesophageal depths of 20 cm or greater.

The error bars in Figure 14.8 represent the average error of 0.8%. For clinical purposes an SpO₂ error in the range of 0.3% to 1.3% is of no clinical significance and it can be ignored (an error greater than ± 3% is considered clinical significant). Also, the possibility of an error introduced by the resolution of the 16-bit data acquisition card was also investigated. The error was found to be extremely small (0.0003%) and therefore it was dismissed.

The regression analysis of the SpO₂ values obtained from the Finger (identical with oesophageal) pulse oximeter vs the SaO₂ values obtained from the CO-Oximeter is shown in Figure 14.9. The equation of the best fit linear regression line was: $y = 10.0 + 0.91 x$; $r = 0.74$; $\text{SEE} = 1.18$; $P<0.001$. The mean and standard deviation of the differences between the finger pulse oximeter and the CO-Oximeter readings were $+1.11 \pm 1.17\%$.

Figure 14.10 shows the linear regression analysis of the commercial finger SpO₂ vs the CO-Oximeter. The equation of the best fit linear regression line was: $y = 4.9 + 0.9x$; $r = 0.74$; $\text{SEE} = 1.26$; $P<0.001$. The mean and standard deviation of the differences between the commercial finger pulse oximeter and the CO-Oximeter readings were $+0.61 \pm 1.23\%$. 
Figure 14.9: Comparison of SpO₂ measurements obtained from the Finger Probe (y-axis) and the CO-Oximeter (x-axis) in 17 patients. The solid line represents the best fit linear regression line. $y=10.0+0.91x; r=0.74; \text{SEE}=1.18; n=36; P<0.001$. The dashed line represents identity.

Figure 14.10: Comparison of SpO₂ measurements obtained from the commercial Finger Probe (y-axis) and the CO-Oximeter (x-axis) in 17 patients. The solid line represents the best fit linear regression line. $y=4.9+0.9x; r=0.74; \text{SEE}=1.26; n=36; P<0.001$. The dashed line represents identity.
Although, there was no ventilator artefact on the finger PPG traces the results of the regression analysis of both finger pulse oximeters vs the CO-Oximeter (see Figures 14.9 and 14.10) were more scattered than the results from the oesophageal vs CO-Oximetry comparison. This may be due to variability in the amplitude of the PPG signals at the finger, since cardiopulmonary bypass patients are hypothermic during the operation and therefore that might compromise the amplitude of the PPG signals at the periphery and therefore compromise the accuracy of the estimation of SpO₂.

Also, a regression analysis of the red/infrared ratio (R values) as measured by the oesophageal pulse oximeter vs the CO-Oximetry SaO₂ values is shown in Figure 14.11. This analysis aimed to see the agreement between the empirical calibration equation used for the estimation of oesophageal SpO₂ with the equation derived from the regression analysis of the red/infrared ratio (R values) as measured by the oesophageal pulse oximeter vs the CO-Oximetry. The empirical calibration equation (SpO₂=110 – (R) x 25), (dashed line in Figure 14.11) used in the oesophageal SpO₂ algorithm (see section 13.7.4) is the result of a linear approximation of an empirical calibration curve derived from a large group of healthy volunteers with arterial SaO₂ values generally greater than 70% (WEBSTER, 1997).

The equation derived from the regression analysis of the red/infrared ratio (R values) as measured by the oesophageal pulse oximeter vs the CO-Oximetry, was SpO₂=106.4 – (R) x 18.8  (solid line in Figure 14.11). That shows that there is some agreement between the two equations. This new oesophageal calibration equation can replace the empirical calibration equation (SpO₂=110 – (R) x 25) in the Virtual Instrument 3b algorithm for estimating oesophageal SpO₂. The new calibration equation would result in oesophageal SpO₂ values closer to the CO-Oximetry values and therefore in a more accurate oesophageal pulse oximeter.
**Figure 14.11:** Comparison of the CO-Oximeter (y-axis) and the red/infrared Ratios measured by the oesophageal pulse oximeter (x-axis) in 17 patients. The solid line represents the best fit linear regression line. $y=106.4-18.8x; r=0.91; \text{SEE}=0.58; n=36; P<0.001$. The dashed line represents the empirical calibration equation ($y=110-25x$) used for the estimation of oesophageal SpO$_2$.

Table 14.4 shows a summary of the results of the regression analysis between saturation values (SpO$_2$) obtained from the three pulse oximeters (oesophageal, finger, and commercial finger) and SaO$_2$ values measured by a CO-Oximeter.

**Table 14.4:** Calculated values of the relationships between SpO$_2$ (oesophageal, finger and commercial finger pulse oximeters) and SaO$_2$ measured by CO-Oximeter

<table>
<thead>
<tr>
<th></th>
<th>Oesophageal vs CO-Oximeter</th>
<th>Finger vs CO-Oximeter</th>
<th>Commercial Finger vs CO-Oximeter</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean Difference (± SD)</td>
<td>0.73 ± 0.72 %</td>
<td>1.11 ± 1.17 %</td>
<td>0.61 ± 1.23 %</td>
</tr>
<tr>
<td>Standard Error of Estimate (SEE)</td>
<td>0.71 %</td>
<td>1.18 %</td>
<td>1.26 %</td>
</tr>
<tr>
<td>Correlation Coefficient (r)</td>
<td>0.91</td>
<td>0.74</td>
<td>0.74</td>
</tr>
</tbody>
</table>
14.5.2 Comparisons of blood oxygen saturation measurements from oesophageal, finger, and commercial finger pulse oximeters and values from blood gas analysis

Before comparing the estimated saturation values obtained from the three pulse oximeters (oesophageal, finger and commercial finger) with blood gas analysis, the second gold standard, a ALTMAN and BLAND (1983) test between the two gold standard methods (CO-Oximetry and blood gases) was performed. The result of this test showed whether the two methods can be used interchangeably.

A set of 36 measurements of blood oxygen saturation values obtained simultaneously using CO-Oximetry and blood gases from 17 patients were used for the comparison of the two gold standards. The between-method differences analysis as suggested by ALTMAN and BLAND (1983) was used to compare the two SaO₂ measuring methods.

Figure 14.12 is a plot of the difference between blood oxygen saturation values obtained by Blood Gas Analysis (BGA) and CO-Oximetry (CO-OX) against their mean. From the Figure it can be concluded that there is no obvious relation between the difference and the mean. Under these circumstances a calculation of the bias, estimated by the mean difference (d) and the standard deviation of the differences (s) could clarify more the degree of agreement between the two methods. For the SaO₂ data the mean difference (d) (CO-OX minus BGA) was −0.8% and the standard deviation (s) was 0.7%. Since, the differences were normally distributed (checked by running a Kolmogorov-Smirnov normality test using the statistical software backage Sigmastat, see section 11.7), 95% of them were expected to lie between d-2s and d+2s (or, more precisely between d-1.96s and d+1.96s) (Figure 14.12).
Figure 14.12: Difference against mean for SaO₂ data obtained from the CO-Oximeter (CO-OX) and blood gas analyser (BGA)

Provided the differences within d ± 2s would not be clinically important the two measurement methods could be used interchangeably. These are referred as the “limits of agreement” (BLAND and ALTMAN, 1986). For the SaO₂ data (CO-OX and BGA) the limits of agreement were:

\[
\begin{align*}
    d - 2s &= -0.8 - (2 \times 0.7) = -2.2 \% \\
    d + 2s &= -0.8 + (2 \times 0.7) = 0.6 \%
\end{align*}
\]

Thus, the SaO₂ results obtained from a blood gas analyser may be 2.2 % below or 0.6 % above the SaO₂ results obtained from a CO-Oximeter. The limits of agreement were within limits (±3%) that are considered to be clinically irrelevant, thereby making either machine (CO-Oximeter and Blood gas analyser) interchangeable.

Since, it was proven that the two gold standards could be used interchangeably the comparison of the estimated saturation values obtained from the three pulse oximeters (oesophageal, finger and commercial finger) with blood gas analysis was performed and is described below.
A total of 155 sets of data points from 49 patients were used for the regression analysis, which gave the estimated slope and intercept of the regression line. An average (± SD) of 3.5 (± 1.5) blood samples were collected from each patient.

A plot of SpO\textsubscript{2} readings obtained from the reflectance oesophageal pulse oximeter (y-axis) and the blood gas analyser (x-axis) is shown in Figure 14.13. The equation of the best fit linear regression line was: \( y = 12.3 + 0.88 x \); \( r = 0.86 \); Standard Error of Estimate (SEE) = 0.86; \( P<0.001 \). The mean and standard deviation for the differences between the oesophageal pulse oximeter and blood gases were 0.02 ± 0.88 %. The average estimated measurement error of the oesophageal pulse oximeter (0.8%), as discussed in the previous section, is also shown in Figure 14.13.

**Figure 14.13:** Comparison of SpO\textsubscript{2} measurements obtained from the oesophageal Probe (y-axis) and the Blood Gas Analyser (x-axis) in 49 patients; The solid line represents the best fit linear regression line. \( y=12.3+0.88x; \) \( r=0.86; \) SEE=0.86; \( n=155; \) \( P<0.001 \). The dashed line represents identity. The error bars represent oesophageal SpO\textsubscript{2} error of ± 0.8%.
The regression analysis of the SpO$_2$ values obtained from the Finger pulse oximeter vs the oxygen saturation values obtained from the Blood Gas Analyser is shown in Figure 14.14. The equation of the best fit linear regression line was: $y = 35.5 + 0.64x$; $r = 0.69$; $\text{SEE} = 1.09$; $P<0.001$. The mean and standard deviation of the differences between the finger pulse oximeter and the Blood Gas Analysis were $0.19 \pm 1.24\%$.

![Figure 14.14: Comparison of SpO$_2$ measurements obtained from the finger Probe (y-axis) and the Blood Gas Analyser (x-axis) in 49 patients; The solid line represents the best fitted linear regression line. $y=35.5+0.64x$; $r=0.69$; $\text{SEE}=1.09$; $n=155$; $P<0.001$. The dashed line represents identity.](image)

Figure 14.15 shows the linear regression analysis of the commercial finger SpO$_2$ vs the Blood Gas Analyser. The equation of the best fit linear regression line was: $y = 26.8 + 0.73x$; $r = 0.63$; $\text{SEE} = 1.48$; $P<0.001$. The mean and standard deviation for the differences between the Blood Gas Analyser and the commercial finger pulse oximeter were $0.33 \pm 1.54\%$. 
Figure 14.15: Comparison of SpO₂ measurements obtained from the commercial finger Probe (y-axis) and the Blood Gas Analyser (x-axis) in 49 patients; The solid line represents the best fit linear regression line. y=26.8+0.73x; r=0.63; SEE=1.48; n=155; P<0.001. The dashed line represents identity.

Again the scatter in Figures 14.14 and 14.15 (finger pulse oximeter and commercial finger pulse oximeter) is larger than the scatter in Figure 14.13 (oesophageal pulse oximetry) perhaps for the same reasons as described previously (compromised pulsatile signals at the periphery due hypothermia) in see section 14.5.1.

Also, a regression analysis of the red/infrared ratio as measured by the oesophageal pulse oximeter vs the Blood Gas Analysis oxygen saturation values are shown in Figure 14.16. The equation of the best fit linear regression line was: \( \text{SpO}_2 = 108.2 - 21.1 \times (R) \); \( r = 0.86 \); \( \text{SEE} = 0.84 \); \( P<0.001 \). The dashed line represents the empirical calibration equation \( \text{SpO}_2 = 110-25 \times (R) \) used for the estimation of oesophageal SpO₂ (see more details about the empirical calibration curve in the previous section).

The equation derived from the regression analysis of the red/infrared ratio (R values) as measured by the oesophageal pulse oximeter vs the blood gas
analyser ($\text{SpO}_2=108.2 - 21.1 \times (R)$) (solid line in Figure 14.16) shows that there is close agreement between the two equations. This new oesophageal calibration equation can replace the empirical calibration equation ($\text{SpO}_2=110 - (R) \times 25$) in the *Virtual Instrument 3* algorithm for estimating oesophageal $\text{SpO}_2$. The new calibration equation would result in oesophageal $\text{SpO}_2$ values closer to the blood gas analysis.

The comparison of the two regression analysis of the red/infrared ratios as measured by the oesophageal pulse oximeter vs the Blood Gas Analyser (Figure 14.16) and the CO-Oximeter (Figure 14.11) shows that the calibration equation derived from the blood gas analyser ($\text{SpO}_2 = 108.2 - 21.1 \times (R)$) was closer to the empirical calibration equation used in the algorithm for estimating oesophageal $\text{SpO}_2$ in the *Virtual Instrument 3b* equation ($\text{SpO}_2 = 110 - 25 \times (R)$) than the equation derived form the CO-Oximeter ($\text{SpO}_2=106.4 -18.8 \times (R)$). That was perhaps due to the larger number of
SaO$_2$ measurements performed with the blood gas analyser (n=155) compare with those obtained with a CO-Oximeter (n=36).

Although, the calibration equation derived from the blood gas analyser was closer to the empirical calibration equation used (Virtual Instrument 3b) than the calibration equation derived from the CO-Oximeter, the CO-Oximeter equation is the one recommended in replacing the empirical one in the algorithm. This is because the CO-Oximeter is considered a more accurate gold standard than the blood gas analyser.

Table 14.5 shows a summary of the results of the regression analysis between saturation values (SpO$_2$) obtained from the three pulse oximeters (oesophageal, finger, and commercial finger) and SaO$_2$ values obtained from blood gas analysis (BGA).

<table>
<thead>
<tr>
<th></th>
<th>Oesophageal vs BGA</th>
<th>Finger vs BGA</th>
<th>Commercial Finger vs BGA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean Difference (± SD)</td>
<td>0.002 ± 0.88 %</td>
<td>0.19 ± 1.24 %</td>
<td>0.33 ± 1.54 %</td>
</tr>
<tr>
<td>Standard Error of Estimate (SEE)</td>
<td>0.86 %</td>
<td>1.09 %</td>
<td>1.48 %</td>
</tr>
<tr>
<td>Correlation Coefficient (r)</td>
<td>0.86</td>
<td>0.69</td>
<td>0.63</td>
</tr>
</tbody>
</table>

14.5.3 Comparisons of blood oxygen saturation measurements between oesophageal, finger, and commercial finger pulse oximeters

Also, the 155 sets of blood oxygen saturation data points from 49 patients were used for the comparison of the three different pulse oximeters used in the study (the oesophageal and finger pulse oximetry channel of the Processing System 4a and the commercial finger pulse oximeter). The between-method differences analysis as suggested by BLAND and ALTMAN (1986) was used to compare the three pulse oximeters.
Figure 14.17 is a plot of the difference between the oesophageal (OES.) and commercial finger (CF) SpO₂ values against their mean. From the Figure it can be concluded that there is no obvious relation between the difference and the mean. Under these circumstances a calculation of the bias, estimated by the mean difference (d) and the standard deviation of the differences (s) could clarify more the degree of agreement between the two methods.

Figure 14.17: Difference against mean for SpO₂ data obtained from the oesophageal (OES.) and commercial finger (CF) pulse oximeters

For the SpO₂ data the mean difference (d) (commercial pulse oximeter minus oesophageal pulse oximeter) was −0.3% and the standard deviation (s) was 1.5%. Since the differences were normally distributed (Gaussian), 95% of them were expected to lie between d-2s and d+2s (or, more precisely between d-1.96s and d+1.96s) (Figure 14.17). Provided the differences within d ± 2s would not be clinically important the two measurement methods could be used interchangeably. These are referred as the “limits of agreement” (ALTMAN and BLAND, 1983). For the SpO₂ data (commercial finger and oesophageal) the limits of agreement were:

\[
\begin{align*}
    d - 2s &= -0.3 - (2\times1.5) = -3.3 \% \\
    d + 2s &= -0.3 + (2\times1.5) = 2.7 \%
\end{align*}
\]

Thus, the oesophageal pulse oximeter may be 3.3% below or 2.7% above the commercial pulse oximeter. The limits of agreement were approximately
within the ±3% accuracy of commercial pulse oximeters as given by the manufacturers. It can reasonably be argued that these levels of difference are clinically irrelevant, thereby making either device a satisfactory method for monitoring the patient’s blood oxygenation.

Figure 14.18 shows a comparison of blood oxygen saturation measured by the commercial finger pulse oximeter (CF) and the finger pulse oximeter (FING) of the Processing System 4a. The mean difference (d) was −0.5% and the standard deviation (s) was 1.3%. The CF SpO\textsubscript{2} readings were lower on average the FING SpO\textsubscript{2} readings. The limits of agreement were:

\[
d - 2s = -0.5 - (2\times1.3) = -3.1 \%
\]
\[
d + 2s = -0.5 + (2\times1.3) = 2.1 \%
\]

These limits of agreement were small enough (clinically insignificant) to be confident that the finger pulse oximeter could be used in place of the commercial finger pulse oximeter.

\[\text{Figure 14.18: Difference against mean for SpO}_2\text{ data obtained from the finger (FING) and commercial finger (CF) pulse oximeters}\]
A comparison of the two pulse oximeters of the *Processing System 4a* (oesophageal and finger) was also performed using the BLAND and ALTMAN (1986) method (see Figure 14.19).

The mean difference (d) was –0.2% and the standard deviation (s) was 1.0%. The limits of agreement were:

\[
\begin{align*}
d - 2s &= -0.2 - (2 \times 1.0) = -2.2 \% \\
d + 2s &= -0.2 + (2 \times 1.0) = 1.8 \%
\end{align*}
\]

Again, these limits of agreement were small enough (clinically insignificant) to be confident and that the finger pulse oximeter and the oesophageal pulse oximetry can be used interchangeable.

![Figure 14.19: Difference against mean for SpO₂ data obtained from the finger (FING) and oesophageal (OES) pulse oximeters of the *Processing System 4a*](image-url)
14.6 PATIENTS IN WHOM PERIPHERAL PULSE OXIMETRY FAILED: RESULTS OF THE OESOPHAGEAL BLOOD OXYGEN SATURATION MEASUREMENTS

From the fifty patients included in the study, five of the patients (10%) had periods of at least 10 minutes (continuous) in duration where both finger pulse oximeters (commercial and Processing System 4a) failed to record pulsatile PPG signals and estimate SpO₂ values. The oesophageal pulse oximeter operated successfully during the periods of finger failure. Figure 14.20 shows typical PPG traces obtained from the oesophageal and finger channel during peripheral pulse oximetry failure. The Oesophageal PPG traces obtained at both wavelengths were of good quality with large amplitudes and high signal-to-noise ratio. An oesophageal SpO₂ was estimated every two seconds without any problems. The finger channel was not able to record any PPG signals (see Figure 14.20) therefore the estimation of finger SpO₂ was not possible (finger SpO₂ channel displays “processing”, see Figure 14.20).

Figure 14.20: Typical PPG traces obtained from a cardiopulmonary bypass patient during times of finger failure
The five patients in whom the peripheral pulse oximetry failure occurred were all cardiac patients undergoing cardiothoracic bypass surgical procedures. In four of the patients the finger pulse oximeters failed postoperatively (within the first half hour after completion of the surgery) in the intensive care unit and in the fifth patient the failure occurred in the operating theatre before the patient went on bypass.

The oesophageal PPG signals recorded from the five patients that a finger failure occurred were of good quality with large amplitudes and high signal-to-noise ratio. The monitoring oesophageal depths of the five patients ranged from 15 cm to 22 cm, measured from the upper lip (15 cm (two patients), 16 cm (one patient), 17 cm (one patient), 22 cm (one patient)).

Table 14.6 gives the mean of the AC PPG amplitudes at both wavelengths at the different oesophageal monitoring depths for the five patients. The PPG amplitudes measured at the monitoring depths were separated into two groups; the upper oesophageal depths (14 cm to 17 cm), the mid oesophageal depths (18 cm to 22 cm). These PPG amplitudes, from the five patients in whom peripheral pulse oximetry failed, were in close agreement with the more detail amplitude PPG study in healthy anaesthetised patients at five oesophageal depths described in section 12.13 (see Table 12.6) and with the quantitative amplitude study of sick patients presented in this chapter (see Table 14.3).

Table 14.6: Mean of AC peak-to-peak PPG amplitudes (mv) at two wavelengths measured in the upper and mid oesophagus of five patients in whom peripheral pulse oximetry failed

<table>
<thead>
<tr>
<th>Oesophageal depths</th>
<th>upper oesophagus (14 cm to 17cm)</th>
<th>mid oesophagus (18 cm to 22 cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infrared (880 nm)</td>
<td>182 mv</td>
<td>477 mv</td>
</tr>
<tr>
<td>Red (655 nm)</td>
<td>65 mv</td>
<td>273 mv</td>
</tr>
<tr>
<td>Number of Patients (n)</td>
<td>4</td>
<td>1</td>
</tr>
</tbody>
</table>

These encouraging results indicate that the arterial blood circulation to the oesophagus is less subject to peripheral vasoconstriction and decreased
PPG amplitudes at times that the peripheral sites used for estimation of blood oxygen saturation fail.

During the time of the finger failure blood samples (a total of seven measurements for all five patients) were collected and blood gas analysis was performed. Oesophageal pulse oximetry values were also recorded at the same time with the blood sampling. A comparison of SpO₂ readings obtained from the reflectance oesophageal pulse oximeter (y-axis) and the blood gas analyser (x-axis) is shown in Figure 14.21. The equation of the best fit linear regression line was: \( y = 3.7 + 0.96x \); \( r = 0.83 \); Standard Error of Estimate (SEE) = 0.52; \( P < 0.001 \). The mean and standard deviation for the differences between blood gases and the oesophageal pulse oximeter were \( 2.0 \times 10^{-15} \), \( \pm 0.48 \% \). Figure 14.21 shows very good agreement between the oxygen saturation values obtained from the oesophageal pulse oximeter and the blood gas analyser.

![Figure 14.21](image_url)

**Figure 14.21**: Comparison of SpO₂ measurements obtained from the Oesophageal Probe (y-axis) and the Blood Gas Analyser (x-axis) in 5 patients in whom peripheral pulse oximetry failed; \( y=3.7+0.96x \); \( r=0.83 \); SEE=0.52; \( n=7 \); \( P<0.001 \). The solid line represents the best fit linear regression line. The dashed line represents identity.
14.7 DISCUSSION AND CONCLUSIONS

The study on the fifty thoracic patients brought the main objective of this project to a successful conclusion. Firstly, it proved that the human oesophagus could be used as an alternative site for monitoring arterial blood oxygen saturation by pulse oximetry. Secondly, it proved that the oesophagus can be used as a reliable site for measuring blood oxygen saturation in patients with poor peripheral circulation, such as the ones presented here, where a pulse oximeter attached to the periphery (usually the finger) fails to operate. This study also showed that the oesophageal pulse oximeter was in good agreement with both gold standards (CO-Oximetry and blood gas analysis).

Oesophageal PPG signals with large amplitudes and high signal-to-noise ratios were measured from various depths within the oesophagus (from the upper to the deep oesophagus) from all fifty thoracic (ASA 2, 3, and 4) patients. The oesophageal amplitudes from the fifty thoracic patients were in good agreement with previous oesophageal PPG amplitude investigations from various depths within the oesophagus in ASA1 patients. These results also indicate that the perfusion at this central site, the oesophagus, is less subject to peripheral vasoconstriction than are the peripheral sites used for pulse oximetry, such as the finger. From all the monitoring depths investigated, the upper oesophagus (17 cm to 18cm) was proven to be the most appropriate for measuring blood oxygen saturation. The main reason for that was that in the upper oesophagus the magnitude of the ventilator artefact was less than 30% of the oesophageal PPG peak-to-peak amplitude, which did not effect the continuous estimation of oesophageal SpO₂. The error (0.8 %) of the estimated SpO₂ values due to ventilator artefact was clinically insignificant and can be ignored.

Comparisons between the three pulse oximeters (oesophageal, finger and commercial finger) with the CO-Oximeter were performed. It was found that the oesophageal pulse oximeter was in very good agreement with the CO-Oximeter and with the finger and commercial finger pulse oximeters. The
mean difference between the commercial pulse oximeter and the CO-Oximeter was slightly smaller (0.61 %) in comparison with that of the oesophageal and the CO-Oximeter (0.73 %) which might suggest that the commercial finger pulse oximeter is more accurate than the oesophageal. This can be argued since the standard deviation of the differences between the commercial finger pulse oximeter and the CO-Oximeter was 1.23% and that of the oesophageal was 0.72 %.

Similarly, comparisons between the three pulse oximeters (oesophageal, finger and commercial finger) with blood oxygen saturation values calculated by a blood gas analyser were performed. Again, it was found that the oesophageal pulse oximeter was in good agreement with the blood gas analyser and with the finger and commercial finger SpO₂.

The results of the oesophageal pulse oximeter can become even more accurate and therefore more close to the CO-oximetry measurements (the most accurate gold standard). This can be done by replacing the empirical calibration equation, used in the algorithm of Virtual Instrument 3b for estimating oesophageal SpO₂, with the calibration equation found by the relationship between the red/infrared ratios measured by the reflectance oesophageal pulse oximeter and SaO₂ measured by the CO-Oximeter.

Also, the three pulse oximeters were compared using the statistical analysis suggested by Bland and Altman. The conclusion from these comparisons was that the three methods or the three instruments used for measuring blood oxygen saturation agreed sufficiently to allow them to be used interchangeably. Both oesophageal and finger pulse oximeters were within the “limits of agreement” (d±2s) of ±3%. This order of error was in agreement with the acceptable error (±3%) specified by the manufacturers of commercial pulse oximeters. Differences of the order of ±3% in saturation are not clinically important and the three different pulse oximeters can be used interchangeably.
Finally the oesophageal pulse oximeter was found to be reliable and accurate (compared with blood gas analysis) in cases of poor peripheral perfusion (five patients) where both the finger pulse oximeters failed to estimate oxygen saturation values for at least ten minutes in duration. From the fifty thoracic patients used in this study 10% had a finger pulse oximetry failure. This percentage of finger failure is in agreement with previously reported measurements (REICH et al., 1996). The oesophageal PPG amplitudes obtained from the five patients at times that the finger pulse oximetry failed were in close agreement with the more detail amplitude PPG study in healthy anaesthetised patients (see Table 12.6) and with the quantitative amplitude study of sick patients (see Table 14.3). These results suggest that the arterial blood circulation to the oesophagus is less subject to peripheral vasoconstriction and decreased PPG amplitudes than are the peripheral sites used for pulse oximetry such as the finger. Therefore, the human oesophagus not only can be used as an alternative SpO\textsubscript{2} monitoring site but also it can be used as a possible SpO\textsubscript{2} monitoring site in cases of poor peripheral circulation where peripheral pulse oximeters fail.
CHAPTER 15

INVESTIGATION OF PPG SIGNALS IN HUMAN VISCERAL ORGANS USING THE OESOPHAGEAL PPG PROBE II AND PROCESSING SYSTEM 4

15.1 INTRODUCTION

Measurement of blood oxygen saturation ($\text{SpO}_2$) from an extremity such as the finger may not accurately reflect splanchnic oxygen saturation values. In many critically ill patients, poor tissue oxygenation is due to disordered regional distribution of blood flow, despite high global blood flow and oxygen delivery. Splanchnic ischaemia may ultimately lead to cellular hypoxia and necrosis and may well contribute to the development of multiple organ failure and increased mortality (JURY OF THE CONSENSUS, 1996). Rapid detection of a significant change in tissue oxygenation could enable earlier and more successful intervention and restoration of splanchnic blood flow and should improve survival in critically ill patients (JURY OF THE CONSENSUS, 1996).

Techniques used to measure tissue oxygenation such as polarographic oxygen electrodes, magnetic resonance spectroscopy and positron emission tomography remain research tools (JURY OF THE CONSENSUS, 1996). Manual fluid tonometry for estimating intestinal hypoxia, is expensive, intermittent, operator dependent and time consuming; the recently introduced automatic device is more convenient but is even more expensive (JURY OF THE CONSENSUS, 1996). Methods such as laser Doppler, Doppler ultrasound, and intravenous fluorescein have been previously explored to assess intestinal ischaemia in animals (PEARCE et al., 1987; FERRARA et
al., 1988; DENOBILE et al., 1990; MACDONALD et al., 1993). Many of these techniques are complex and expensive and none of them directly measures oxygenation. Therefore there is a need for a simple, reliable, and continuous method for estimating visceral organ SpO$_2$. Animal studies have also shown that pulse oximetry could be used to monitor intestinal oxygen saturation (MACDONALD et al., 1993). The feasibility of estimating blood oxygen saturation in humans has been demonstrated by a study using a commercial transmission pulse oximeter on the colon (OURIEL et al., 1988). However, there are difficulties in applying commercial pulse oximeters to measurements in abdominal human organs because the probes are unsuitable and are not easily sterilizable. Moreover, none of the currently available probes could be left in the abdomen for prolonged postoperative monitoring.

As a preliminary to constructing a suitable pulse oximeter for monitoring abdominal organ SpO$_2$, the Oesophageal PPG Probe II and Processing System 4 were used for the investigation of PPG signals, obtained at red and infrared wavelengths, from the surface of the bowel, liver and kidney. The aim was to develop techniques to facilitate measurements on patients with compromised splanchnic circulation, which would be useful both intraoperatively and in intensive care.

15.2 METHODS
After obtaining approval from the Local Research Ethics Committee and informed, written consent, twelve adult patients undergoing elective laparotomy under general anaesthesia were studied. The seven women and five men ranged in age between 29 and 78 years, with an average age of 52 ± 4 years. Table 15.1 summarises patients’ details.
Table 15.1: Details of the Patients used for the Visceral PPG Study

<table>
<thead>
<tr>
<th>Patients</th>
<th>Sex</th>
<th>Age</th>
<th>Surgical procedures</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>F</td>
<td>49</td>
<td>Anterior sphincteroplasty</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>29</td>
<td>Bowel resection</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>38</td>
<td>Colectomy with ileo-rectal anastomosis</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>57</td>
<td>Repair of incisional hernia</td>
</tr>
<tr>
<td>5</td>
<td>M</td>
<td>54</td>
<td>Pancreatrico-Duodenectomy and right adrenalectomy</td>
</tr>
<tr>
<td>6</td>
<td>F</td>
<td>47</td>
<td>Laparotomy</td>
</tr>
<tr>
<td>7</td>
<td>M</td>
<td>57</td>
<td>Whipple’s procedure</td>
</tr>
<tr>
<td>8</td>
<td>F</td>
<td>47</td>
<td>Biliary reconstruction, re-do choledochojejunostomy</td>
</tr>
<tr>
<td>9</td>
<td>M</td>
<td>31</td>
<td>Cystoscopy, uretheric stenting, abdomino perineal excision of ileoanal pouch</td>
</tr>
<tr>
<td>10</td>
<td>M</td>
<td>78</td>
<td>Biliary bypass</td>
</tr>
<tr>
<td>11</td>
<td>M</td>
<td>62</td>
<td>Laparotomy, defunctioning colostomy</td>
</tr>
<tr>
<td>12</td>
<td>F</td>
<td>70</td>
<td>Laparotomy, defunctioning colostomy</td>
</tr>
</tbody>
</table>

The study was observational and patients’ surgical and anaesthetic management were as per routine. Induction with propofol 2-3 mg/kg and fentanyl 1-2 μg/kg was followed by atracurium 0.5 mg/kg, tracheal intubation and intermittent positive pressure ventilation. Anaesthesia was maintained with 1-2% isoflurane and 50-70% nitrous oxide in oxygen. Patients received boluses of fentanyl, morphine and atracurium as clinically indicated. All patients were haemodynamically stable and well oxygenated.

The reflectance *Oesophageal PPG Probe II* and the identical reflectance finger probe, described in detail in sections 9.3 and 9.4 respectively, were used for the simultaneous recording of PPG signals from the abdominal viscera and from the finger. Both probes were connected to *Processing System 4* (see section 13.2). *Virtual Instrument 3b* (see section 13.7) was used for the acquisition of the PPG signals. To facilitate comparison studies between red and infrared AC and DC PPG signals from the two channels (oesophageal and finger), the hardware and software gains of *Processing System 4* and *Virtual Instrument 3b* were identical for both channels. A commercially available transmission type pulse oximeter probe was also used on an adjacent finger to record finger SpO₂.

The *Oesophageal PPG II* was inserted into a sealed, disposable size 20 French gauge gastric tube. Sealing of the sterile gastric tube, for prevention of contamination, was achieved aseptically by cutting off the blind tip and
inserting it retrogradely into the distal lumen. The PPG probe, mounted on the end of a semi-rigid cable, was then passed down the gastric tube to its final position with the probe approximately 0.5 cm from the sealed end. The gastric tube containing the probe was then applied gently to the surface of each abdominal organ so that the emitted light was reflected from its surfaces (Figure 15.1). General theatre and operating lights were switched off. The probe was kept in place until a stable PPG signal was achieved. Simultaneous AC and DC PPG traces from each abdominal organ and the finger were recorded for approximately two minutes.

![Figure 15.1](image.png)

**Figure 15.1:** Reflectance abdominal organ PPG probe placed on the surface of the human bowel (courtesy of M. Savva)

### 15.2.1 Data Analysis and Statistics

Data files recorded by the *LabVIEW Virtual Instrument 3b* software were analysed offline. Patients were only accepted into the final analysis if measurable visceral PPGs were present for at least one abdominal organ. The amplitudes of the visceral and finger AC PPG signals for each patient were measured on printouts from *LabVIEW* using a ruler and the means and standard deviations (SD) were calculated. The statistical significance of the mean difference between the PPG amplitudes at the investigated abdominal
organs and the finger was assessed by performing a One Way Analysis of Variance (ANOVA) for both infrared and red wavelengths using SigmaStat (SPSS Ltd, Birmingham, UK). The data were normally distributed (checked by performing the Kolmogorov-Smirnov normality test (see more details in section 11.7). A value of \( p < 0.05 \) was considered statistically significant.

15.3 RESULTS

Measurable PPG signals were obtained from the surface of the bowel in all twelve patients, and depending on intra-operative accessibility, also from the liver (eight patients) and the kidney (six patients). Typical PPG traces obtained from the large bowel, kidney and liver, together with finger PPG traces at both wavelengths are shown in Figure 15.2. The low frequency artefact present on the liver and kidney traces was due to movement of the handheld probe especially when monitoring at areas were access was difficult (kidney or liver).

![Figure 15.2: (a) AC PPG traces from simultaneous measurements at the bowel and finger.](image)
Figure 15.2: (b) AC PPG traces from simultaneous measurements at abdominal organs (kidney, and liver) and the finger.
Table 15.2 gives the mean ± SD of the AC PPG amplitudes at both wavelengths at the three investigated abdominal organs and the finger. The PPG signals obtained from the abdominal organs and the finger had similar amplitudes and reasonably high signal-to-noise ratios. These results are also illustrated in Figure 15.3.

Table 15.2: Mean (± SD) of AC peak-to-peak PPG amplitudes (v) at two wavelengths at the three abdominal organs and the finger

<table>
<thead>
<tr>
<th>Investigated organ</th>
<th>Mean AC PPG Amplitudes (v)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Bowel (V)</td>
</tr>
<tr>
<td>infrared IR(880nm)</td>
<td>1.02 (±0.38)</td>
</tr>
<tr>
<td>red R(655 nm)</td>
<td>0.43 (±0.18)</td>
</tr>
<tr>
<td>Number of patients (n)</td>
<td>12</td>
</tr>
</tbody>
</table>

Figure 15.3: AC Peak-to-Peak Amplitudes, Mean (± SD), at two wavelengths from the three abdominal organs and the finger
The ANOVA test showed that there were no statistically significant differences between the PPG amplitudes recorded from the abdominal organs and those from the finger.

15.4 DISCUSSION AND CONCLUSIONS

Abdominal organ PPG signals at red and infrared wavelengths have been obtained with adequate signal-to-noise ratio using the *Oesophageal PPG Probe II*. This appears to be the first quantitative report of PPG signals from human abdominal organs, although the feasibility of recording PPG signals from the colon have been reported (did not show any PPG signals or any quantifying results on PPG signals from the colon) by OURIEL et al. (1988). The PPG amplitudes from both hollow and solid abdominal organs were on average, approximately the same as those obtained simultaneously from a finger for both wavelengths, although there was considerable variability. The primary objective of this work was to measure the amplitude of PPG signals from various abdominal organs as a preliminary to constructing a suitable pulse oximeter for monitoring abdominal organ SpO$_2$. However, some random comparison of online estimation of SpO$_2$ from the abdominal organs showed good agreement with SpO$_2$ values obtained simultaneously from the commercial finger pulse oximeter. These observations support the hypothesis that pulse oximetry (using a custom made probe such as the one described here) may be used as a monitoring technique, for abdominal organs such as the bowel, liver and kidney for intraoperative and prolonged postoperative monitoring. Further work needs to be carried out to validate this hypothesis.
CHAPTER 16

CONCLUSIONS

Pulse oximeter probes placed peripherally may fail to give accurate values of arterial blood oxygen saturation (SpO₂) when peripheral perfusion is poor (REICH et al., 1996; MOLLER et al., 1993; FREUND et al., 1991). The aim of this project was to develop and evaluate an oesophageal pulse oximeter, which would allow accurate and reliable measurements of oxygen saturation to be made in patients with compromised peripheral circulation, based on the hypothesis that blood flow to this central site may be preferentially preserved.

A new prototype miniature opto-electronic infrared reflectance Oesophageal PPG Probe I, small enough to fit into a conventional stomach tube (French 20) was developed (see section 9.2). A two channel (oesophageal and finger) battery operated PPG Processing System 1 was also constructed (chapter 10). Oesophageal PPG Probe I and PPG Processing System 1 were used to record for the first time infrared AC photoplethysmographic signals from the deep oesophagus of a healthy awake volunteer. The recorded infrared oesophageal AC PPG signals were compared with finger infrared AC PPG signals obtained from an identical custom made reflectance finger PPG probe (see section 9.4). The infrared AC PPG signals obtained from the human deep oesophagus were significantly larger in amplitude that those obtained from the finger. The infrared wavelength used for these measurements was suitable for pulse oximetry and therefore these encouraging results were the first positive indication of the feasibility of the oesophagus as a monitoring site for blood oxygen saturation.

The successful outcome from the healthy volunteer study convinced the ethics committee to allow the continuation of the study on healthy anaesthetised (ASA 1) patients. The infrared Oesophageal PPG Probe I was used for the investigation of AC infrared PPGs from the deep oesophagus of
a small group (four) of anaesthetised patients. Large infrared AC PPGs with good signal-to-noise ratio were obtained from all four patients. Such measurements had not been reported previously and were very hopeful as far as the feasibility of estimating SpO\textsubscript{2} in the oesophagus in anaesthetised patients. The next step was to try to measure AC PPGs in the oesophagus at a red wavelength (the second wavelength used by pulse oximeters).

The above encouraging results at infrared wavelength lead to a new more quantitative study of PPG signals in the middle third of the oesophagus in 20 ASA 1 anaesthetised patients, at two wavelengths red (665 nm) and infrared (880 nm). For this study a new Oesophageal PPG Probe II (section 9.3) and a new battery operated PPG Processing System 2 (chapter 11) were used. Again, the oesophageal AC PPG signals were of good signal-to-noise ratio for both wavelengths. The amplitudes of the oesophageal AC PPGs were on average three times larger than those obtained simultaneously from a finger for both wavelengths. However, the oesophageal PPG traces were modulated by an artefact synchronous with the approximately five second period of the mechanical ventilator. The magnitude of the artefact from the mid-third of the oesophagus was of the order of 40% of the PPG peak-to-peak amplitude. The results were exciting since both red and infrared AC PPGs were recorded for the first time from the middle third of the oesophagus in all 20 patients. Since both wavelengths used were suitable for pulse oximetry, in principle, it would have been feasible to estimate arterial SpO\textsubscript{2} values in the mid-oesophagus. However, it was not possible to estimate arterial blood oxygen saturation with this system because: (a) the red and infrared AC PPG signals were not recorded simultaneously and (b) PPG Processing System 2 was not design to record red and infrared DC PPGs. Measurements of SpO\textsubscript{2} can only be performed if the AC and DC PPG signals at both wavelengths are recorded simultaneously. Also, there was a concern over the ventilator artefact in relation with the estimation of SpO\textsubscript{2}. It was thought that the ventilator artefact would make the estimation of SpO\textsubscript{2} impossible or erroneous. Techniques to eliminate or minimise the ventilator artefact such as filtering were considered but they were not required
The new time multiplexed PPG Processing System 3 (chapter 12) was used to perform detail quantitative investigations of AC and DC PPG signals (obtained simultaneously) at five oesophageal depths (15 cm, 20 cm, 30 cm, and 35 cm) at red and infrared wavelengths on 16 anaesthetised ASA 1 patients. The objective was to determine an optimal position for reliable SpO₂ monitoring. A 3-lead ECG channel was also incorporated in this PPG Processing System. The PPG and ECG signals were analysed by means of a Virtual Instrument implemented in LabVIEW on a laptop personal computer.

It was found that the new Oesophageal PPG Probe II was able to record good quality red and infrared AC and DC PPG signals from all five oesophageal depths. The Oesophageal PPG Probe II could record PPGs from any depth within the oesophagus and was much more versatile than the “transesophageal” probe described by ATLEE et al. (1995), which could only be positioned in the upper oesophagus at the cricopharyngeous muscle. The maximum PPG amplitudes appear to be in the mid-oesophagus at 25 cm from the upper incisors. The PPG signals obtained from the upper oesophagus (20 cm and 15 cm) seemed to be less affected by the ventilator artefact when compared with the PPGs obtained in the deeper oesophagus (25 cm, 30 cm and 35 cm). This supported the hypothesis that as the monitoring depth was further away from the diaphragm (upper oesophagus) the modulation on the PPG traces caused by the ventilator became smaller. Manual offline oesophageal SpO₂ calculations were performed at the five oesophageal depths and the SpO₂ values were on average 6.5% lower than those obtained simultaneously from the commercial finger pulse oximeter. Despite the fact that the estimation of oesophageal SpO₂ was relied upon a published empirical calibration curve for the wavelengths 660 nm and 940 nm (MOYLE, 1994) instead of a curve specifically for the wavelengths used in the Oesophageal PPG Probe II (655 nm and 880 nm), the consistent underestimation was not far from the accuracy limits of ± 3% of commercial pulse oximeters. The SpO₂ results were very encouraging and proved for the first time that adequately large AC and DC PPG signals, that will allow the
estimation of SpO₂, can be recorded within the whole depth of the oesophagus. These results also indicated that the ventilator artefact might not affect the estimation of SpO₂ as much as it was thought initially.

Also, offline calculations of the pulse transit times (PTTs) to the oesophagus were performed. Although there do not seem to be any reported PTT measurements to the oesophagus in the literature, the measured PTTs were consistent with published PTT measurements at peripheral sites (SUGO et al., 1999). Although there was no significant difference between the oesophageal PTT values at the five oesophageal depths, they showed a trend that the minimum value occurred when the oesophageal measurement site was supplied by the shortest and most direct arterial route. Such measurements may give useful clinical information on the vasculature of the oesophagus. Further studies are needed to confirm this hypothesis.

Since the oesophagus was successfully shown to be an adequate site for monitoring PPGs and estimating SpO₂ in ASA 1 healthy anaesthetised patients, the next important step was to investigate sick ASA 2,3, and 4 patients in whom conventional pulse oximetry might fail due to poor peripheral circulation. A new isolated Processing System 4 was developed (chapter 13) on three PCB boards, one for the oesophageal PPG channel, one for the finger PPG channel and one for the 3-lead ECG channel. A new LabVIEW Virtual Instrument (section 13.7) allowed the continuous display of oesophageal and finger PPGs and the online estimation of oesophageal and finger SpO₂ values. This appears to be the first time that a Virtual Pulse Oximeter implemented in LabVIEW was developed.

In the extensive clinical study (chapter 14) on fifty ASA 2,3, and 4 anaesthetised patients undergoing high-risk thoracic operations, the majority of whom were cardiopulmonary bypass patients, SpO₂ values were estimated online. These patients had an arterial cannula inserted routinely, which allowed the rigorous validation of the oesophageal SpO₂ measurements with CO-Oximetry. The oesophageal SpO₂ values, which
were continually being estimated (every two seconds), were also compared with blood oxygen saturation values obtained from blood gas analysis (BGA) and from a commercial finger pulse oximeter.

The study on cardiac patients resolved the question over the contribution (partial or complete) of any aortic or cardiac movement artefact to the oesophageal PPG signals. The amount of this artefact had been in doubt since the beginning of the study. The investigation of oesophageal PPG signals during aortic cross clamping showed that pulsatile oesophageal PPGs ceased although there was still cardiac activity (see section 14.4.1). Also, the investigation of oesophageal PPGs during pulsatile flow bypass showed that pulsatile oesophageal PPG signals of arterial origin existed when the heart was completely stopped (see section 14.4.2). The results of these two investigations, along with the quite accurate estimation (see section 14.5) of oesophageal SpO$_2$ by the new system showed clearly that the oesophageal PPG signals are likely to be due entirely to pulsatile blood flow in arteries.

Oesophageal PPG signals with large amplitudes were measured from the upper to the deep oesophagus from all fifty thoracic (ASA 2, 3, and 4) patients. The oesophageal amplitudes were in good agreement with previous measurements in ASA 1 patients. The upper oesophagus (17 cm to 18cm) was proven to be the most appropriate for measuring SpO$_2$. The main reason was that in the upper oesophagus the magnitude of the ventilator artefact was less than 30% of the PPG peak-to-peak amplitude, which did not prevent the continuous estimation of oesophageal SpO$_2$. The estimated error in the SpO$_2$ values due to ventilator artefact was less than 1.3%, which would be clinically insignificant.

The SpO$_2$ results showed that the oesophageal pulse oximeter was in very good agreement with the CO-Oximeter and the blood gas analyser (section 14.5.1). Also, a comparison of the three pulse oximeters was made using a statistical analysis suggested by Bland and Altman (1986) (section 14.5.3).
The oesophageal and finger pulse oximeters were all within the “limits of agreement” (d ± 2s) of ± 3% which equal to the actual error (± 3%) usually specified by the manufacturers of commercial pulse oximeters. Differences of the order of ± 3% in saturation are not clinically important and it was concluded that the three different pulse oximeters could be used interchangeably.

Finally, the oesophageal pulse oximeter was found to be reliable and accurate (compared with blood gas analysis) in cases of poor peripheral perfusion (five patients) where both the finger pulse oximeters failed to estimate oxygen saturation values for at least ten minutes. From the fifty thoracic patients studied 10% had a finger pulse oximetry failure, which is in agreement with previously reported measurements (REICH et al., 1996). The mean oesophageal PPG amplitudes obtained from the five patients when the finger pulse oximetry failed were in close agreement with values of PPG amplitudes in healthy anaesthetised patients (see Table 12.6) and with those for sick patients (see Table14.3). These results show that in general the arterial blood circulation to the oesophagus is less subject to peripheral vasoconstriction and decreased PPG amplitudes than are the peripheral sites used for pulse oximetry such as the finger. Therefore, the human oesophagus not only can be used as an alternative SpO₂ monitoring site but also it can be used as a possible SpO₂ monitoring site in cases of poor peripheral circulation where peripheral pulse oximeters fail. These results were based on only five patients and, therefore, more work is needed to confirm these findings. This novel monitoring site, the oesophagus, can also find applications in patients who have burns or other serious injury where the oesophagus may be the only available site for pulse oximetry monitoring.

The Oesophageal PPG Probe II and the Processing System 4 were also used for the investigation of PPG signals, obtained at red and infrared wavelengths, from the surface of the bowel, liver and kidney. The aim of this pioneer investigation was to develop techniques to facilitate measurements on patients with compromised splanchnic circulation, which would be useful both intraoperatively and in intensive care. This preliminary study on 12
patients undergoing laparotomy showed for the first time that adequate PPG signals at both wavelengths could be obtained from the human bowel, liver and kidney. The PPG amplitudes from both hollow and solid abdominal organs were on average, approximately the same as those obtained simultaneously from a finger for both wavelengths. Some random comparisons of online estimations of SpO₂ from the abdominal organs showed good agreement with SpO₂ values obtained simultaneously from the commercial finger pulse oximeter. These observations support the hypothesis that pulse oximetry may be used as a blood oxygen saturation monitoring technique for abdominal organs for intraoperative and prolonged postoperative monitoring. Further work needs to be carried out to validate this hypothesis.

The goal of further efforts should be to develop a smaller oesophageal probe, which can be tolerated by conscious patients post operatively. A smaller oesophageal probe that can be introduced via the nasopharynx and remain in situ throughout the operation and postoperatively will be very advantageous. Such a small and more versatile probe, which can fit into a nasogastric tube can be designed and constructed either using smaller opto-electronic components than in the existing probe or using optical fibres. Long term monitoring of oesophageal SpO₂ should then be possible until cardiovascular stability and adequate peripheral perfusion are restored. This new probe could also be use for SpO₂ monitoring in paediatrics. The fibre optic oesophageal probe could be further developed to become a multifunction sensor registering not only oesophageal PPG and SpO₂ but also blood velocity using laser Doppler techniques, oesophageal pH and temperature. Telemetric techniques might also be applied to develop oesophageal pulse oximetry where the SpO₂ monitoring would be feasible in the absence of a cable connecting the oesophageal probe with the main processing unit. Such multifunction probes would be useful in anaesthesia and have many other clinical applications e.g. the measurement of organ perfusion in the stomach and jejunum.
The Virtual Instrument implemented in LabVIEW can be furthered optimised to replace most of the hardware circuitry of the PPG Processing System such as the timing generation, multiplexing and demultiplexing. A versatile pulse oximeter would in effect then be made from a laptop computer.

In summary, a new and clinically useful minimally invasive oesophageal pulse oximetry system with the potential of commercial exploitation has been successfully developed and evaluated for use with patients undergoing prolonged surgical procedures. The use of this novel pulse oximeter has proven for the first time that the oesophagus is a reliable and accurate monitoring site for blood oxygen saturation in healthy patients and in sick patients in whom conventional pulse oximetry fails due to poor peripheral circulation.
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