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ABSTRACT

This report concerns the design and clinical use of a reflective brachial photoplethysmograph. A plethysmograph is an instrument to obtain tracings showing volume changes of a part of the body. Originally this related to volume variations due to blood circulation within the body part of interest. The instrument is said to have been invented by Mosso of Turin around 1870 [1], known in Italian as a "pletismografo", and first reported in Scientific American in July 1872. A photoplethysmograph is an optical detector that indicates the volume of blood in or passing through an area of tissue. By placing the photoplethysmograph at or near the site of a human artery the pulse waveform can be detected and measured. The photoplethysmograph can be transmissive or reflective. There are a variety of sites around the body that are commonly used for detecting the pulse waveform including the finger, the ear lobe, and the foot. The device developed in this work is a reflective detector that uses the brachial artery as a photoplethysmographic site. There appear to be no prior indications in academic or patent literature of this site being used with this type of detector and consequently the authors believe this device to be novel and worthy of reporting to the research community.

1 Introduction

During the development of an experimental procedure to correlate pulse transit time (PTT) with blood pressure [1] in humans, it became necessary to find a waveform detection site on the same artery but 'upstream' of the finger. Pulse transit time is the time an arterial pressure wave takes to travel between two points along the same artery. The brachial artery was the natural site as it rises to the surface of the upper arm before sinking deep into the tissue of the lower arm. This location is known as the *anticubital fossa*. Here the brachial artery splits into the ulnar and radial arteries. The traditional method of measuring PTT is to use the R-wave of the ECG as one timing reference and to use a consistent feature of the photoplethysmograph (PPG) as the other. Having conducted experiments using this technique, a consistent jitter in the PTT data suggested that there may have been variation in the timing of the R-wave. The timing variation is indicated as arising from isovolumetric contraction period variations [3]. It is assumed, in experiments using this timing reference that the R-wave of the ECG coincides with contraction of the left ventricle and systole. The source of jitter observed in the PTT data needed to be researched to establish if the ECG was the source and to examine whether a better timing reference could be found. The clearest approach was to cannulise a variety of subjects with a brachial cannula [3], giving a direct measurement of the pulse waveform. Fitting the subjects with an ECG and a finger photoplethysmograph would provide all the data needed to examine the problem. Cannulisation is both uncomfortable and risky in subjects so another means of looking at the brachial pulse waveform was required. The reflective brachial PPG was developed as a result of the unique requirements of this experimental protocol.

2 Technical development

The most common application of photoplethysmography is in pulse oximetry [4]. It was apparent from product literature that reflective oximetry probes existed for paediatric use. Leading commercial companies Nellcor© and Nonin© produced service manuals that indicated the type of circuit topology used in this sort of device.

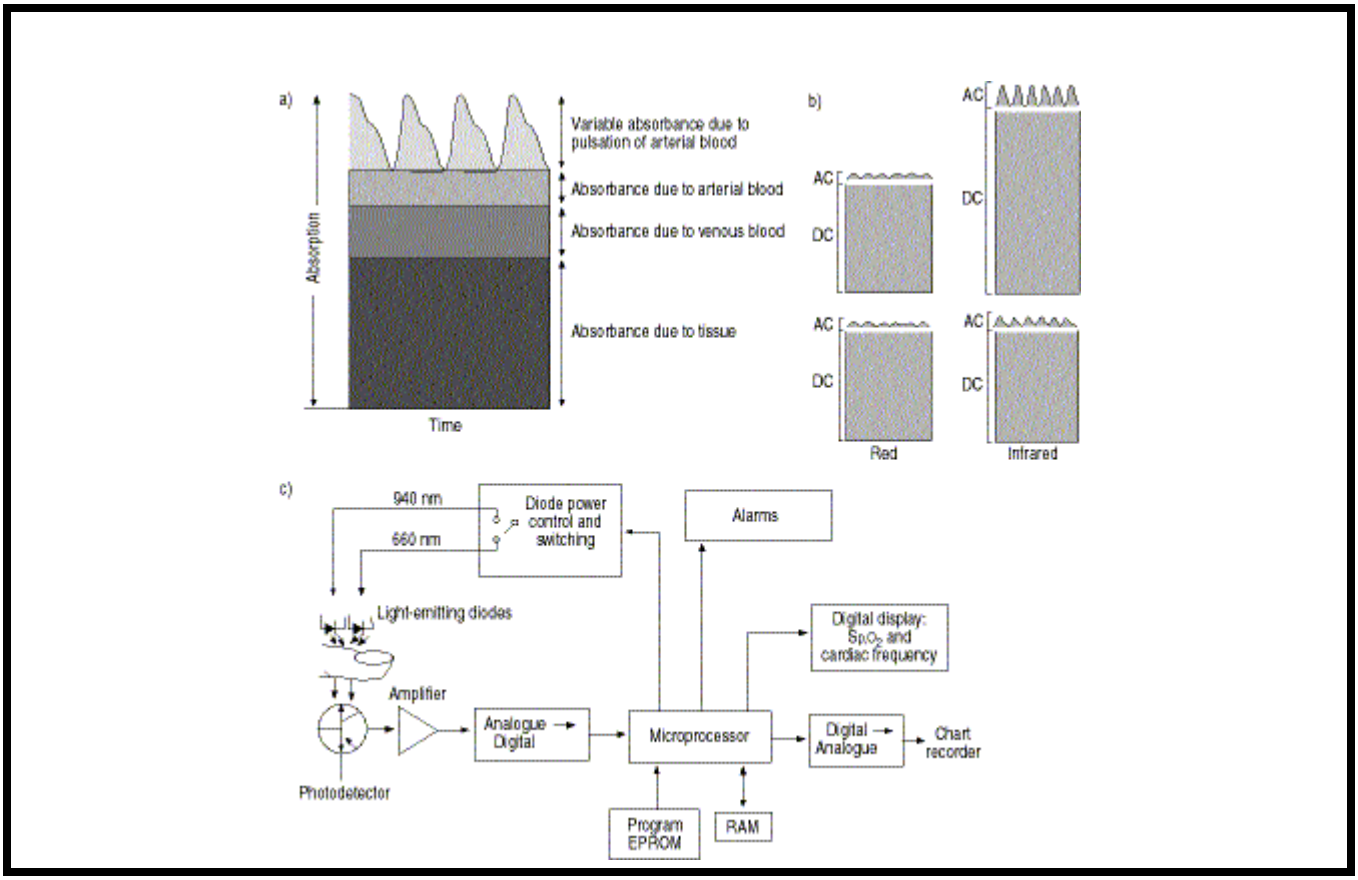


Figure 1 Principles of oximeter operation (reproduced by permission of Dr J Moyle)

Referring to *figure 1(a)* there are four absorption strata at a PPG site indicated. The majority of light produced by an oximetry probe is scattered by tissue, venous blood and arterial blood. A small proportion of light experiences variable scatter due to the pulsation of arterial blood. Some proportion of this light is detected by an optoelectronic detector and contains the signature of the pulsatile component of arterial blood flow. From *figure 1(b)* it is apparent that the proportion of the detected light that contains the pulsatile component is a fraction of the total received light. This large DC offset clearly has implications in the design of a pulse oximeter or PPG. Oximetry concerns the measurement of relative absorption of two different frequencies of light. This provides a metric of the ratio of oxyhaemoglobin to deoxyhaemoglobin. In the development of a PPG probe we can choose to work with either infrared or red light as both will show the variances in absorption required to produce a PPG characteristic. It was decided to use infrared light because of the availability of infrared detectors 'tuned' to different frequencies. Additionally, it was felt that an infrared detector would be less prone to interference from changes in ambient lighting conditions. *Figure 1(c)* shows a typical oximeter system

diagram. In common with this diagram most literature shows the oximeter or PPG in the transmissive configuration. The optical properties that give rise to transmissive changes also give rise to reflective changes. Consequently a reflective probe should look similar to a transmissive probe except in its physical configuration *(Figure 3(b)).*

2.1 Circuit design

A number of disposable oximetry probes were sourced with the intention of experimenting with them as the front end of the amplification circuit. The probes however contained parts that had no useable references or numbers and consequently it was not possible to examine data sheets. It was decided that both the circuit and the probe would need to be developed from first principles. As the probe was required to operate in conjunction with a transmissive finger probe and an electrocardiograph (ECG) amplifier, the other circuits were constructed onto a single printed circuit board (PCB) [4].

The PPG circuit comprises a current source driving an infrared LED. This was implemented with a linear regulator and rheostat arrangement, as a pulse width modulated (PWM) source would have potentially introduced noise into the system. The compromise here is that the source may need to be adjusted manually if the signal, detected by the photodiode, is not strong enough. The infrared detector, Texas Instruments TSL262, comprises a photodiode and an initial FET gain stage, detecting the transmitted light through the finger. The signal from the photodiode contains a large DC component or offset. Superimposed on this is an AC characteristic reflecting the pulsatile component of the circulation. This component can vary between 0.01% and 1% of the DC level. The signal from the IR detector is filtered and then amplified on the circuit board. A Butterworth, single pole, band pass filter is used, where the lower cut-off frequency is 0.05Hz. The upper cut-off frequency is 10Hz. The gain of the filter is G=522. To ensure the 'cleanest' possible signal, the PPG system is implemented with high quality active and passive components on a double sided PCB. A system diagram is shown in *figure 2.* There are two aspects to the circuit. An infrared source is controlled by a variable current source, a reflected component is detected, filtered and amplified.

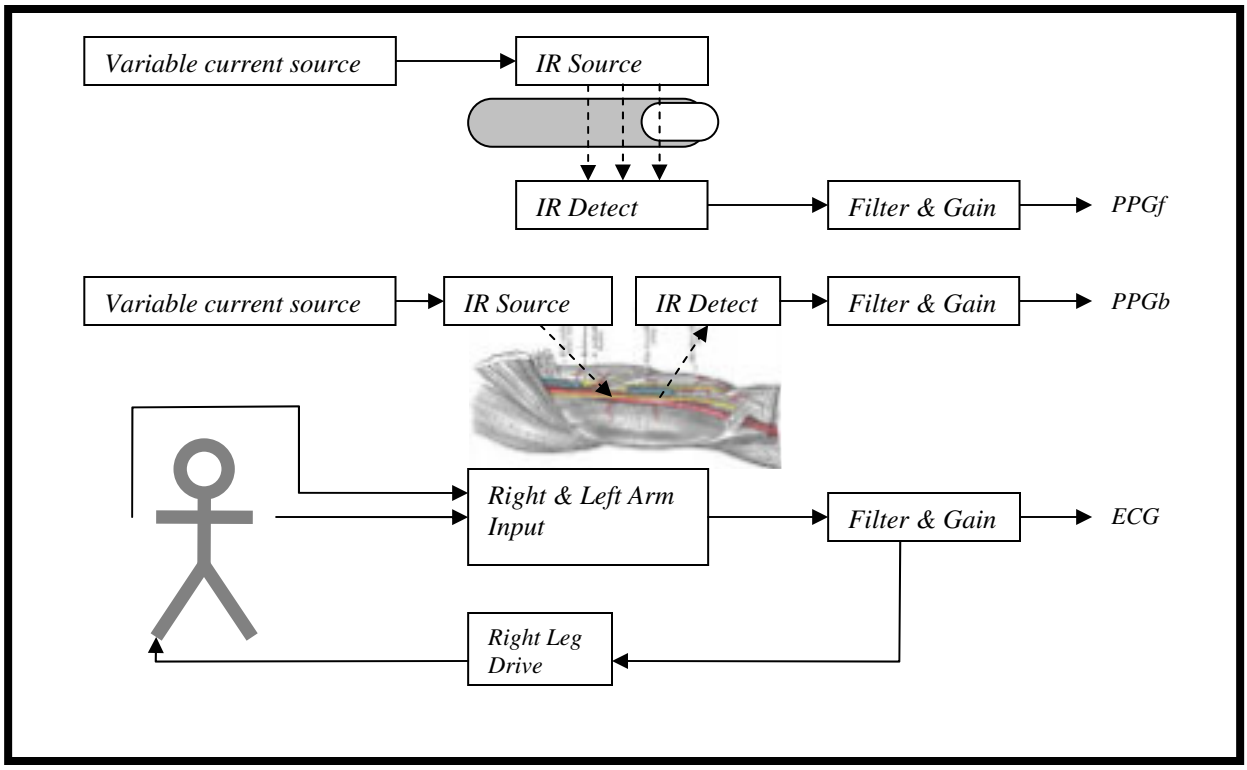


Figure 2 System diagram of ECG, PPGf and PPGb circuit

2.2 Probe design

Two probes were designed and tested. *Figure 2(a)* shows the first prototype. Three high intensity infrared light emitting diodes (LEDs) or emitters surround an IR detector. Both emitters and detectors are housed in an opaque black plastic disk and are flush with the surface. There is a raised ring around the detector such that when the front surface is pressed against the flesh there is no direct path for the IR light to travel between emitters and detector. The probe is positioned above the elbow, at the point where the brachial pulse is detected by hand. The emitters bathe an area around and including the brachial artery with IR light. The detector, if it is placed directly over the artery, will detect variations in reflected light and consequently the pulsatile waveform. This prototype probe worked well in tests conducted in the Dep't of Anaesthesiology in St Vincent's Hospital, under the direction of Dr Leo Kevin. The probe was however highly sensitive to motion, and being of rigid construction, had to be pushed into the flesh causing a certain degree of discomfort as well as unpredictable behaviour given any movement.

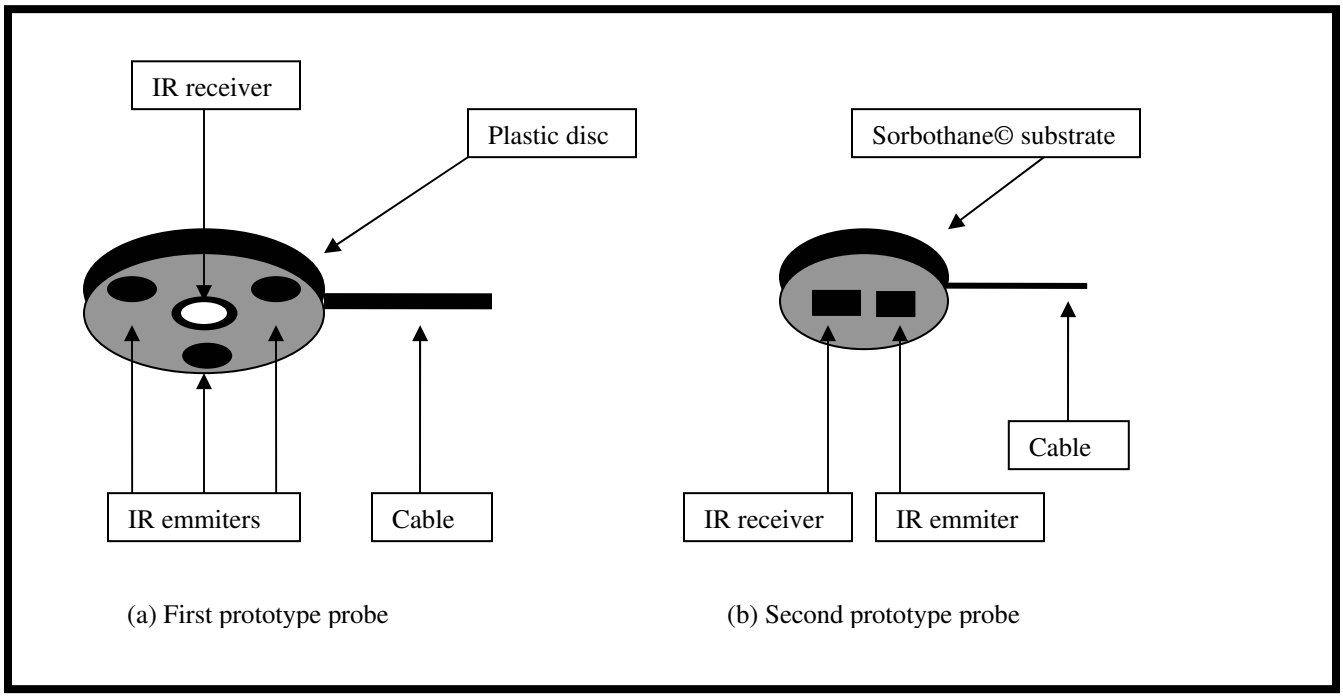


Figure 3 Probe designs

In *figure 2(b)* a number of changes to the design are shown. The principal change is reduction from three emitters to one emitter. This was due to the fact that in the first prototype the detector output level frequently saturated indicating too much IR light. The second change was the move to a flexible opaque medium called Sorbothane©. This is a viscoelastic material similar to a very low durometer rubber that conforms to the contour of the skin but can retain the emitter and detector components. Sorbothane, unlike rubber, is latex free and more suitable than rubber as a skin contact probe under biocompatibility guidelines. This probe proved substantially more stable in operation. Being slightly smaller than the first prototype, it lent itself to better attachment to the skin. The cable on this probe was lighter than the first prototype and this may also have contributed to enhanced stability. The performance enhancements achieved through the use of a smaller, more flexible probe with a lighter cable suggest that a further reduction in size and a telemetric solution to cabling could optimise this probe.

3 Clinical Testing

Having decided that the second prototype probe was superior to the first a trial was set up in which the probe would be used in a clinical setting. A series of 10 subjects were fitted with the probe and a data logging program was developed to collect the resulting data.

Figure 5 shows a sample of output from the brachial PPG. The sample has the morphology that might be expected but demonstrates both a high frequency component and some baseline wander. Predictably

enough, a spectrum produced from the data in Matlab© shows the high frequency component to be due to 50Hz mains 'breakthrough'. The baseline wander, occurring at about 0.8Hz, may have physiological significance but this is beyond the scope of this report. It is notable that there was no EMG type noise detected. This is due to the fact that the probe, being optical in nature, is galvanically isolated from the body. *Figure 6* shows the signal after band-pass filtering. The time and amplitude are represented in samples where the A/D has an input range -10V to 10V. The A/D card offers 12 bit resolution so 1 sample = 4.88 x 10^{-3} V. The data is sampled at 10kHz so 1 sample = 1 x 10^{-4} seconds. The offline filter was implemented in Matlab© and is a 4 pole Butterworth design.

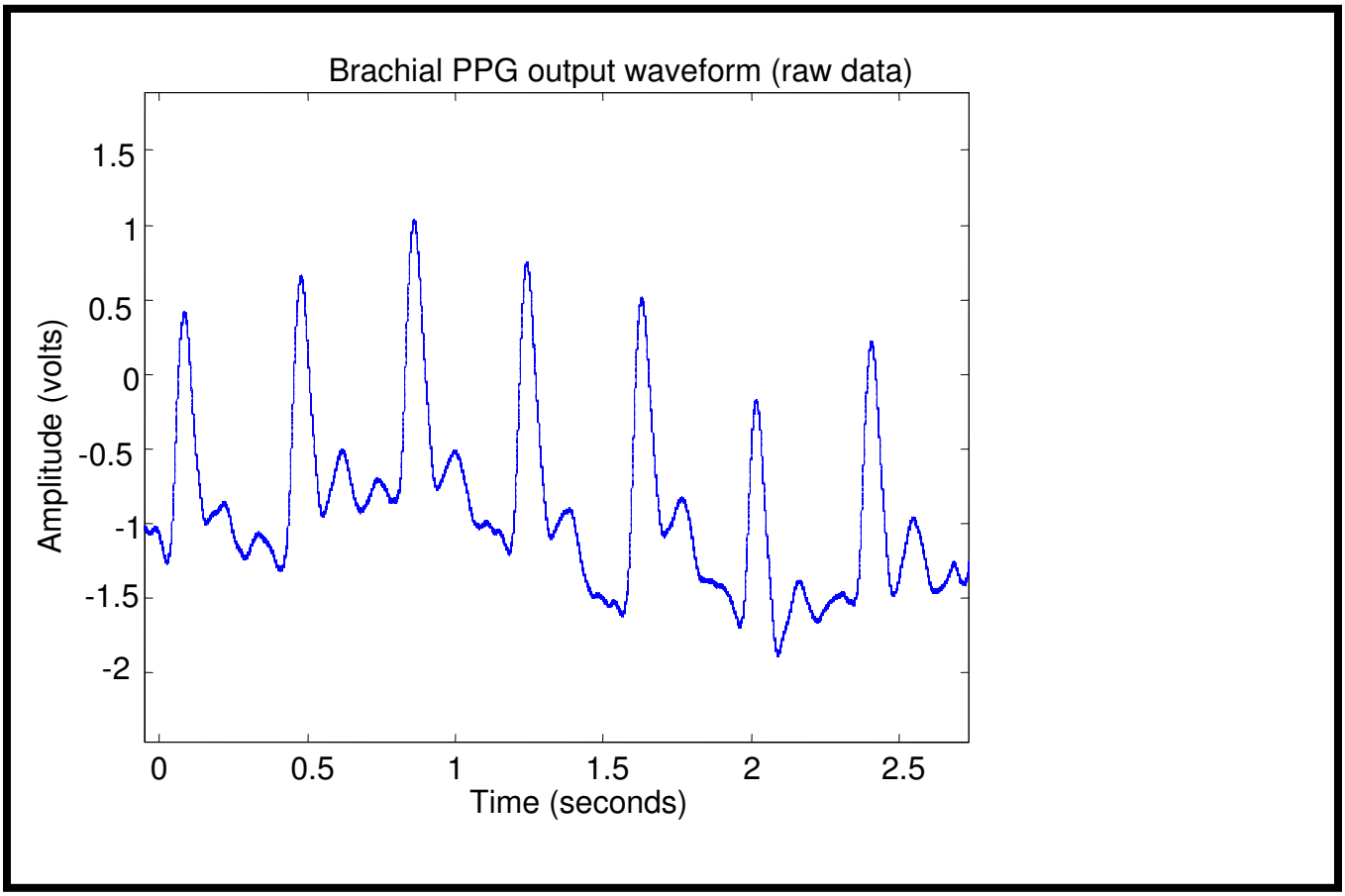


Figure 5 PPG probe output unfiltered

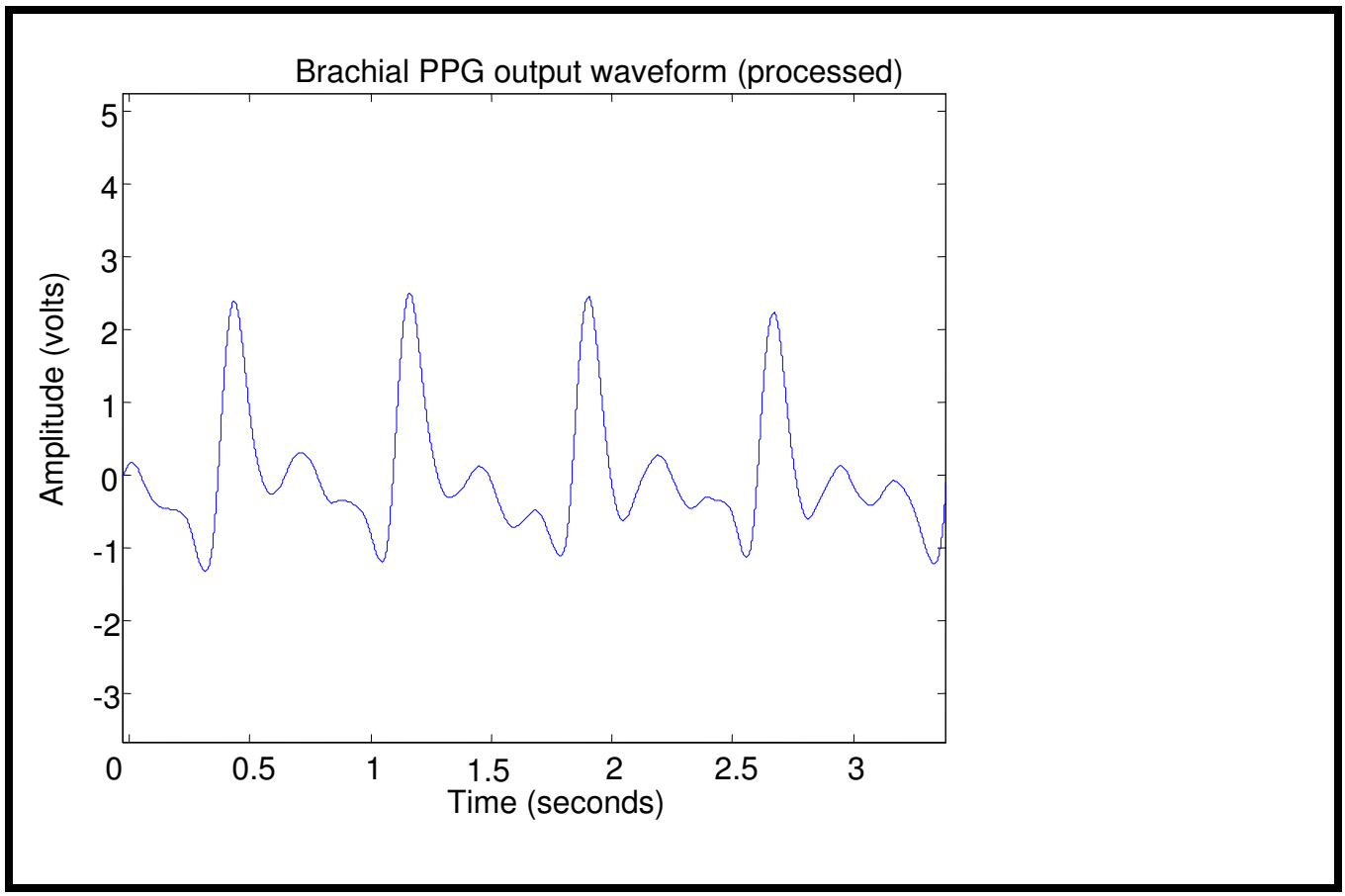
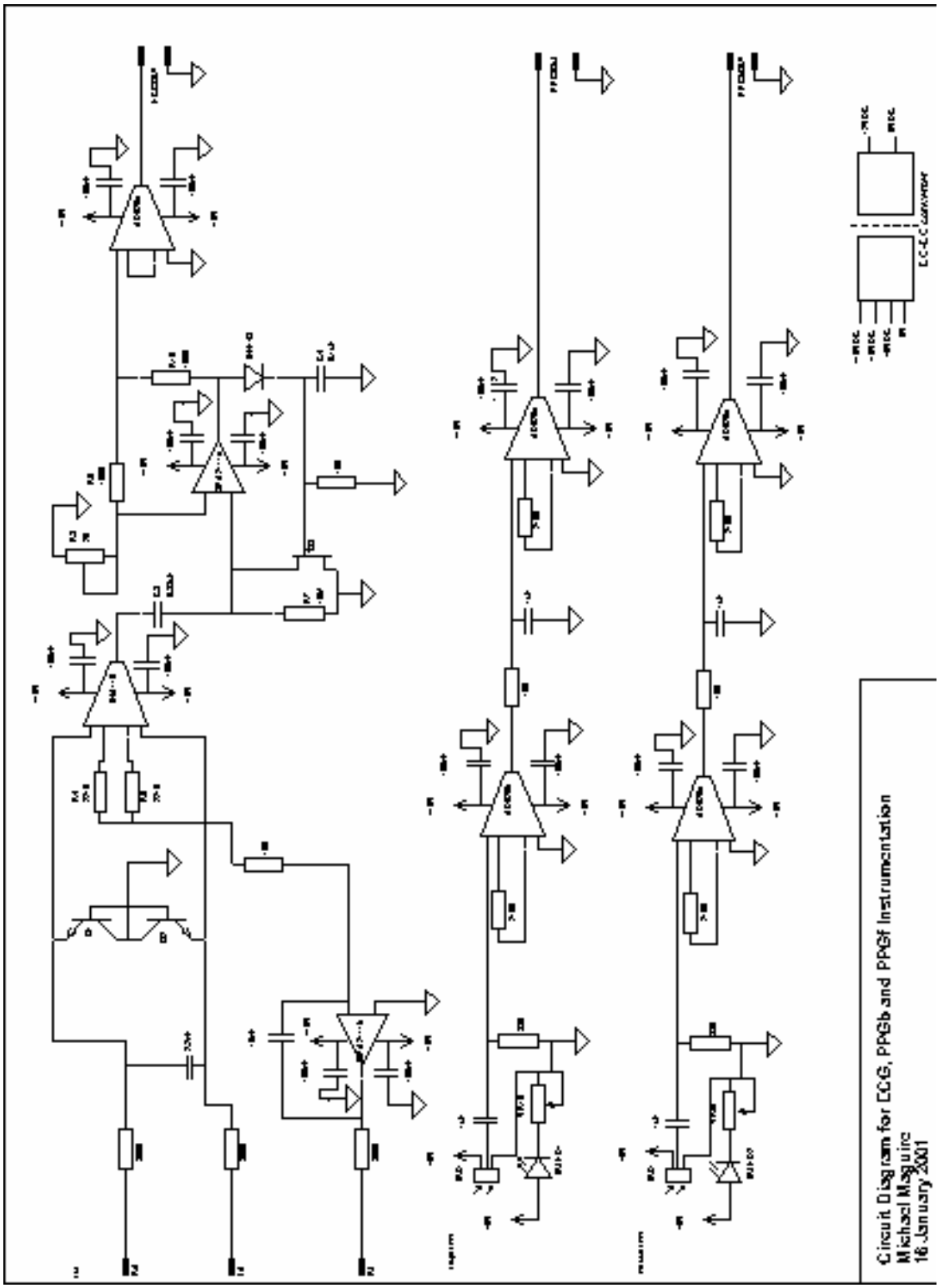


Figure 6 PPG probe output band pass filtered


Circuit Diagram for CCG, PPGb and PPGf Instrumentation
Michael Maguire
16 January 2001

4 Discussion

The objectives of this work were to design, and use a reflective brachial PPG in a clinical environment. The process of design took much from the oximetry application and as such was completed quickly and inexpensively. The use of instrumentation amps as gain stages might be seen as 'overkill' in this application. In a commercial embodiment of such a device this might be so however for the purposes of this work the flexibility that they provided enabled experimentation with many emitter/detector pairs and filter parameters.

The probe was sensitive to motion and on occasion saturated the amplifier when movement occurred. This situation was improved by the use of an arm splint. The splint comprised a short length of cardboard tube cut down its long axis. The splint was secured to the arm using loose fitting Velcro© straps.

A variety of solutions to the issue of resilience and stability of the signal may form the core of further work. Among the suggestions was the possibility of a matrix of detectors or emitter/detector pairs covering the general area in which the brachial artery is found. Some system of polling the matrix elements to read the strongest signal or an aggregated signal from several elements seems plausible. The second interesting suggestion was that of an automated gain circuit that would compensate for variations in the output signal. This would be achieved by controlling the output intensity of the emmiter/s and varying the gain of the detector.

The final objective was to use the device in a clinical setting. The device proved simple to use being attached to the arm with surgical tape. It was necessary to monitor the output on an oscilloscope to ensure there was a good signal. In the first probe the use of three high power emitters generated a degree of heat that was at times uncomfortable for the subject. The second probe, produced no perceivable heating effects.

The brachial photoplethysmographic probe is a device with much development potential. Its usefulness in examining pulse transit time has been demonstrated in subsequent work [6]. The authors believe that this non-invasive device may be useful in looking at the morphology of the blood pressure characteristic and has commercial potential in the area of ambulatory blood pressure measurement.

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