A MULTI-CHANNEL SYSTEM FOR USE IN CARDIAC ELECTROPHYSIOLOGIC STUDIES

by

Barry Neil Wyatt

A dissertation submitted to the Faculty of Medicine, University of Cape Town, in partial fulfilment of the requirements for the degree of Master of Science in Medicine in the field of Biomedical Engineering.

August 1991

DECLARATION

I, ...Barry Neil Wyatt..., hereby declare that the work on which this thesis in based is original (except where acknowledgements indicate otherwise) and that neither the whole or any part of it has been, is being, or is to be submitted for another degree in this or any other University.

I empower the University to reproduce for the purpose of research only, either the whole or any portion of the contents.

Signed: Signed by candidate

Signature removed

Date: 1991-08-15

ABSTRACT

The location of accessory pathways in Wolff-Parkinson-White syndrome patients is performed manually during open heart surgery at Groote Schuur Hospital, using a hand-held roving electrode. This manual procedure is slow and tedious, prolonging the operation and the time for which the patient remains on cardiac bypass.

A multichannel electrogram acquisition and display system with a storage facility would significantly reduce the time taken and improve the reliability of locating the accessory pathways. Having considered a number of currently available cardiac mapping systems it was decided that a new system be developed for specific application within Groote Schuur Hospital. The main design goals of this system are to improve accuracy, increase reliability and enhance the speed of the entire mapping procedure with direct benefit to staff and patients.

The system is based on an IBM compatible computer and allows for the acquisition of a maximum of thirty-two electrogram inputs. A typical configuration would acquire twenty epicardial, two reference (one each from atrium and ventricle), one roving electrode and two surface lead signals. The epicardial signals are obtained from a custom-built electrode belt which is placed around the heart over the atrioventricular groove.

The project includes the development of front-end hardware and software for processing, display and storage of electrogram signals. The relative activation times of the signals are displayed under software control in order to facilitate the location of any accessory pathway(s).

ACKNOWLEDGEMENTS

My grateful thanks to :-

the Foundation for Research Development, Pretoria; the Frank Forman Grant, Duncan Baxter Scholarship, University of Cape Town and the Cardiac Clinic, Groote Schuur Hospital for their financial assistance during this dissertation.

Mr Mladen Poluta and Professor Rob Scott Millar for their supervisory roles, advice and the many hours of proof reading.

Mr Andre Roux for his help and advice in the circuit board design and manufacture, and his wife for her wonderful hospitality during the hours of work in their home.

my parents for whose verbal and financial support I am deeply indebted.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

GLOSSARY

-3dB cutoff frequency The passband of a filter is the range of frequencies that are relatively unattenuated by the filter. The passband extends to the -3dB point where the gain is 3dB lower (30% reduction in amplitude) than that of the midband region. The frequency at this point is the -3dB cutoff frequency (see figure).

Pass band The range of frequencies between the -3dB points of a bandpass filter (see figure).

Roll-off The rate at which the attenuation of a signal increases at frequencies away from the passband. Typically expressed in dB/octave.

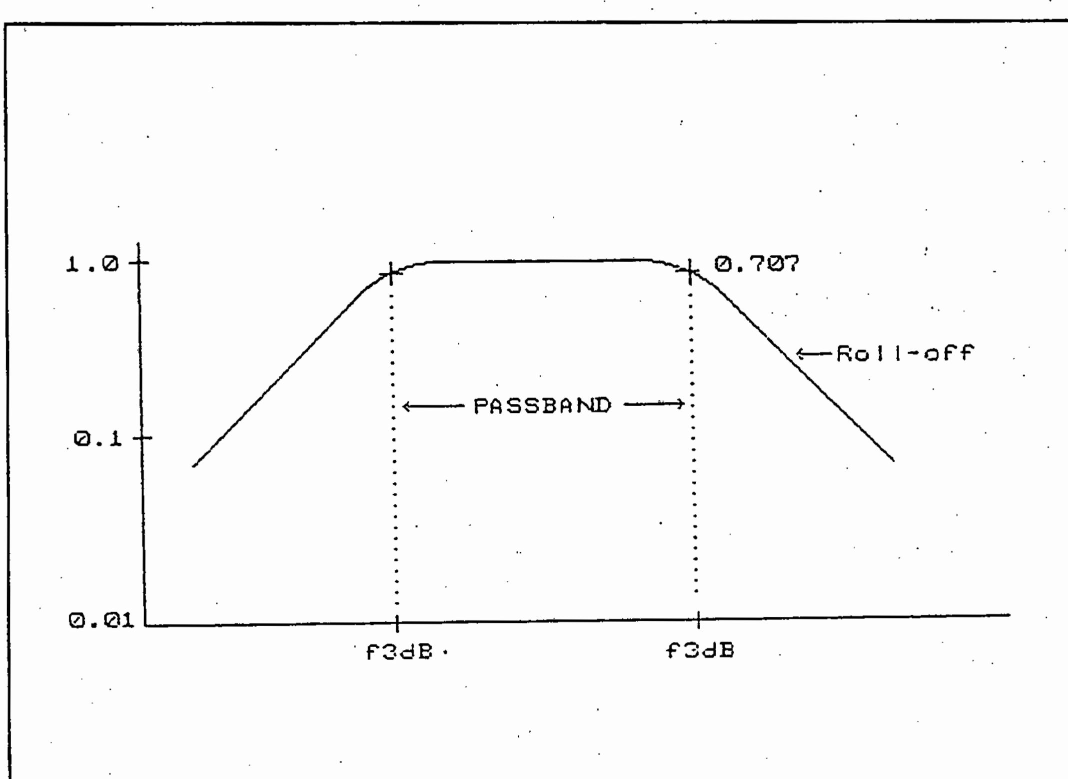


Figure i **The amplitude response of a bandpass filter showing the -3dB Cuttoff Frequency, Pass Band and Roll-off.**

Aliasing is due to undersampling (sampling at a lower frequency than permissible) of a signal. Undersampling causes signal frequency components to appear 'folded back' down near zero frequency. If the amplitude of these signals is high this phenomenon will cause distortion and noise in the resultant signal.

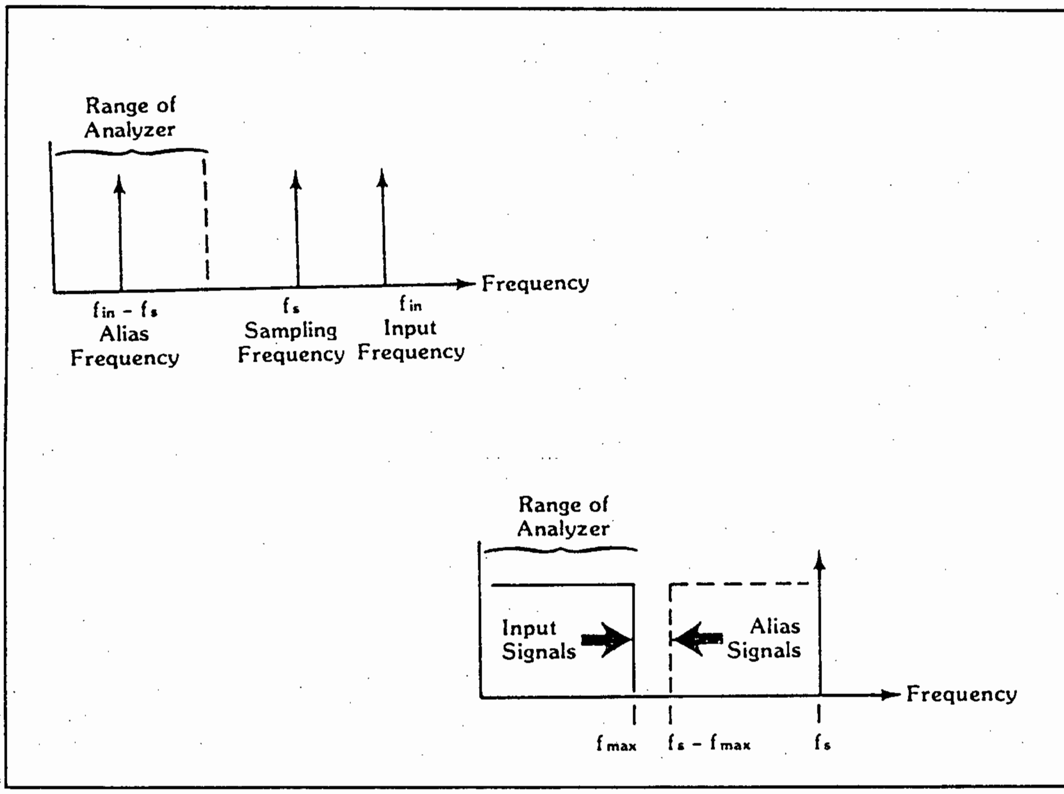


Figure ii **The Phenomenon of Aliasing due to undersampling**

Nyquist criterion This is closely linked to aliasing and states that in order to prevent aliasing the minimum sampling rate must be greater than twice the highest frequency component in the signal.

Random Access Memory (RAM) is the volatile memory of a computer. The memory contents are lost when the power is turned off. Conventional memory is located between 0K and 1024K and is the maximum amount of memory that can be addressed by the 8088 microprocessor on which the IBM PC is based. The designers of the original PC divided conventional memory space into a 640K block of memory to be used by MS-DOS programs (low DOS memory) and a 384K block of high memory for system hardware. The term, conventional memory, is sometimes used to refer just to memory from 0 to 640K.

Direct memory access (DMA) When data must be transferred at high rates DMA allows direct communication between a peripheral device and computer (RAM) memory. No programming is involved during the actual transfer of data; bytes are moved between peripheral and memory via the data bus without program intervention.

Common mode rejection ratio (CMRR) A characteristic feature of biopotential recording is that the body is not at ground potential, but at some higher potential common to both electrode inputs to the differential amplifier. This common-mode potential must be rejected. Consequently, the differential amplifier must exhibit very low gain for common-mode signals. This desirable characteristic given by a single figure of merit, the common-mode rejection ratio:

$$CMRR = \frac{Differential\text{-}Gain}{Common\text{-}mode\text{-}Gain}$$

For example a typical differential amplifier will have a differential voltage gain of 60 dB and a common-mode gain of 0.6 dB giving a CMRR of 100 dB.

Surge arrestors are used to protect measurements devices against large input voltages and often take the form of gas discharge (neon) tubes. These devices do not conduct current until some predefined voltage is reached. At this, or higher, voltages the device acts like a low resistance and diverts current from the measurement circuit while dissipating the applied energy.

CHAPTER ONE

INTRODUCTION

The electrocardiogram (ECG) represents the electrical activity of the heart. This activity is usually recorded by means of electrodes placed on the surface of the body and is used in the diagnosis and classification of pathological cardiac conditions. In the normal activation sequence beginning at the sinus node the wave of electrical activation is conducted through the atrial myocardium, internodal tracts and internodal fibres. The impulse arrives at the atrioventricular (AV) node before atrial muscle contraction is complete; it proceeds through the AV nodal tissue and then through the His-Purkinje system to reach the ventricular muscle fibres (Figure 1.1).

The orderly sequence of conduction through the entire myocardium is essential for correct functioning of the heart. Disturbance to this regularity and order may result in haemodynamic complications which frequently result in decreased cardiac output. Due to the complexity of the heart's activation sequence there are many different types of arrhythmia that can occur. An arrhythmia is a disturbance of the heart rhythm and can take the form of tachycardias (where the normal cardiac rhythm is accelerated), bradycardias (where the normal rhythm is slowed) or a variety of irregularities of the rhythm. Arrhythmia initiation and persistence is related to the properties of cardiac tissue and how it conducts the electrical impulses that give rise to cardiac contraction.

Tachycardias may originate in either of the ventricles or supra-ventricularly. Pre-excitation is one of the causes of a supraventricular tachycardia and occurs when cardiac muscle is excited by a stimulus other than the normal stimulus originating from the sino-atrial node. The WPW syndrome is the most common of the pre-excitation syndromes. The pre-excitation and tachycardia occur due to the presence of an electrical connection or connections (accessory pathway/s) between the atria and ventricles other than the atrioventricular node.

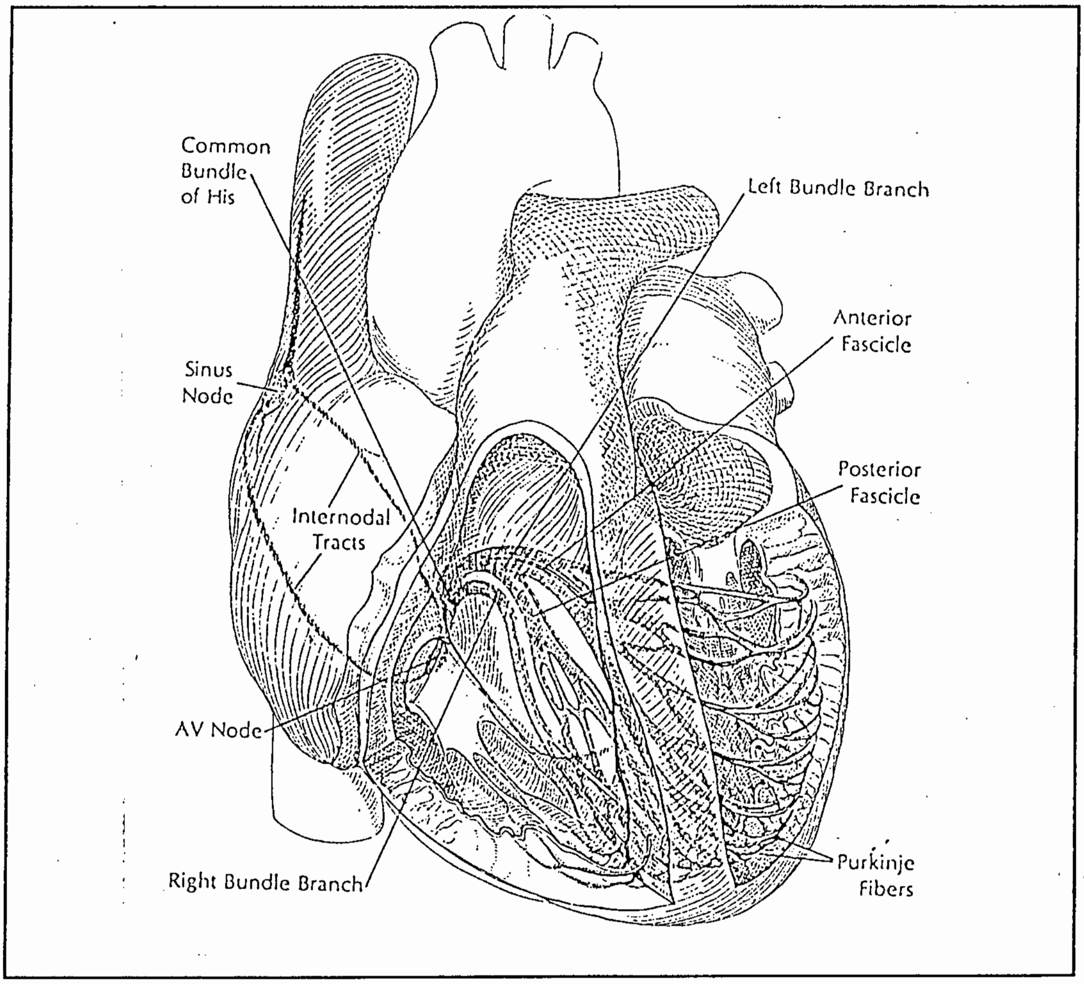


Figure 1.1 **The Conduction System of the Heart** (Katz and Selwyn 1983).

Cardiac rhythm disturbance is symptomatic of a range of cardiac disorders. As the complexity of diagnosis and treatment of abnormal heart activity has increased, so has the demand for more precise and specific ECG measurements. One way of meeting this need is to place the electrodes closer to the heart: on, or near, the internal heart surface (endocardially), the external heart surface (epicardially) or within the myocardium. Endocardial signals are usually obtained from catheter electrodes positioned in one or more cardiac chambers during procedures referred to as electrophysiologic (EP) studies.

EP studies are a great asset in aiding the understanding of cardiac conduction pathologies and their disruptive effect on the cardiac rhythm. One is able to diagnose syndromes such as WPW by means of the standard surface ECG but EP studies are used to confirm the diagnosis and to guide corrective therapy.

Measurement in these situations has been undertaken most successfully by means of multichannel data acquisition systems. In such systems signals from multiple electrodes placed inside or around the heart are acquired and displayed after basic processing such as amplification, filtering and isolation. Increasingly such systems are computer-based allowing for additional processing, more sophisticated user-interfaces and data storage/recall possibilities.

It was proposed by the Cardiac Clinic at Groote Schuur Hospital that a multichannel data acquisition system be developed. Such a system would form the basis for EP studies and have particular applicability to WPW syndrome and the localisation of accessory pathways. Initial investigation and specification led to the design and implementation of a computer-based system for acquisition, storage, processing and display of electrograms.

A review of literature in Chapter 2 attempts to provide background material pertinent to a study of cardiac mapping, data acquisition and signal processing. The review highlights EP studies, WPW syndrome mapping, data acquisition and signal processing techniques.

The specifications for the system are given in Chapter 3 while the hardware and software design considerations and implementation are presented in detail in Chapter 4 and 6 respectively. The system is designed to acquire up to 32 channels of electrogram data. The input signals are amplified, filtered, digitised and transferred to the computer where they are stored onto hard disk. The system software oversees this entire process as well as providing for the retrieval and display of the signals.

The creation of a patient waveform database is explained in Chapter 5 while Chapter 7 provides discussion and evaluation of the implemented design. Conclusions and recommendations for further investigation are presented in Chapter 8.

CHAPTER TWO

LITERATURE SURVEY

2.1 Introduction

A survey of the published material was conducted in order to provide the theoretical basis for this work. Firstly an overview of cardiac mapping is presented in which there is discussion on EP studies and WPW syndrome mapping. These were investigated in order to promote a clear understanding of the principles and conditions under which the diagnosis and study of cardiac arrhythmias is undertaken. Secondly the hardware and software techniques used in data acquisition and cardiac mapping are covered. Attention is also brought to bear on detecting the activation time of the acquired signals and on commercial cardiac mapping systems.

2.2 A Perspective on Cardiac Mapping

Beginning in the late 1970s, modern technological support for analysis of cardiac activation sequences during cardiac surgery has been introduced at several research centres (Ideker et al 1979; El-Sheriff et al 1981; Wit et al 1982; Witkowski and Corr 1984; Cardinal et al 1984; Parson and Downar 1984 and Bonneau et al 1987) both for laboratory research and as aids to surgical treatment of rhythm disturbances. These disturbances in the electrical activation patterns of the heart have come under increasingly diverse investigations in recent years (Jenkins et al 1987; Mehra et al 1983; Josephson et al 1980; Parson et al 1982 and Downar et al 1984).

The contractions of the heart are initiated by a propagating electrical wave of depolarisation which traverses the myocardium and sequentially activates the cardiac cells. This activation sequence can be mapped by analyzing the potentials recorded at multiple sites over or in the heart (Cochrane et al 1983). Cardiac activation mapping is increasingly being used to investigate arrhythmia mechanisms in experimental studies (Klein et al 1979; El-Sheriff et al 1981 and Wit et al 1982) and to localise the origin of abnormal rhythms (arrhythmogenic foci) that are to be ablated surgically or isolated in patients that are refractory to drug therapy (Parson and Downar 1987; Gallagher et al 1982; de Bakker et al 1984 and Downar et al 1984).

In situations where the arrhythmia is either highly unstable or polymorphic, a multichannel system capable of detailed activation sequence determination from one or only a few beats is desirable, and it is for these arrhythmias that most of the computer-based mapping-systems, with elaborate electrode arrays and powerful processors, have been developed (Buckles et al 1990).

2.2.1 Cardiac electrophysiologic studies

A cardiac EP study involves the insertion of several electrode catheters into the chambers of the heart (Gallagher et al 1978). These catheters are used to stimulate the heart and record local endocardial electrograms as shown in Figure 2.1. Stimulation is performed at predefined rates for fixed durations, or the stimulation frequency is gradually increased (incremental pacing). Premature extrastimuli may also be delivered during normal (sinus) rhythm or at the completion of a pacing sequence.

The goals of an EP study in general are: (1) the diagnosis of the electrical abnormality; (2) analysis of the arrhythmias that result from this electrophysiologic abnormality; (3) correlation with the symptoms that result from the arrhythmias; and (4) assessment of the ability to affect any or all of the above with pharmacologic, surgical, or other therapy (Untereker 1989).

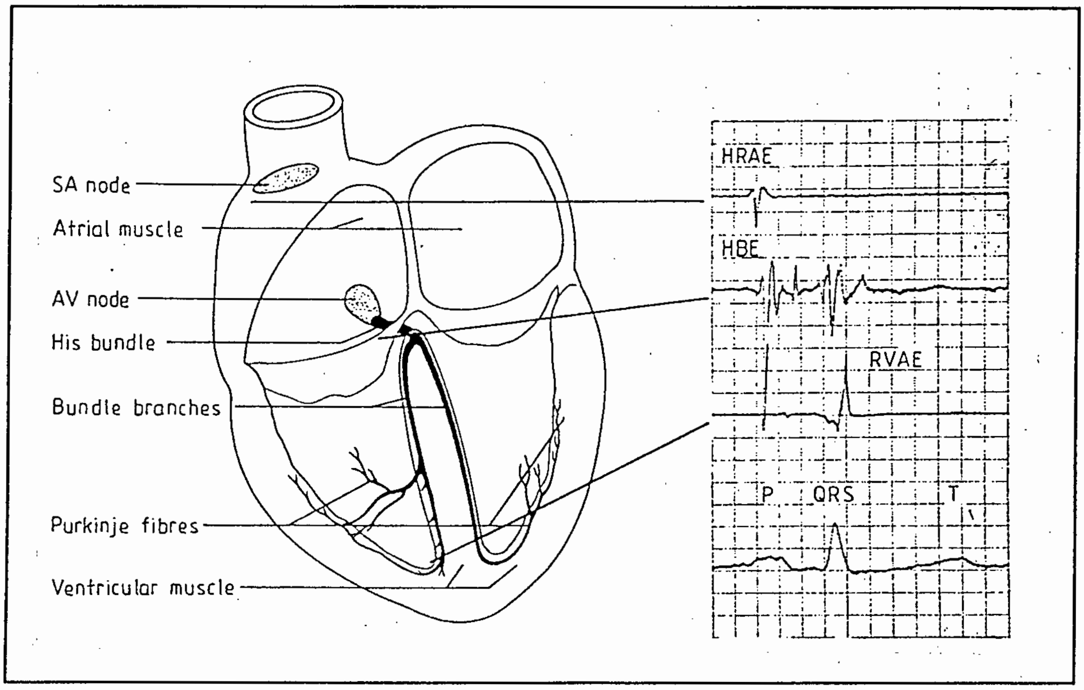


Figure 2.1 **Electrophysiology of the Heart Showing Typical Intracardiac Electrograms. HRAE, high right atrial electrogram; HBE, His bundle electrogram; RVAE, right ventricular apex electrogram. The lower trace shows a standard (lead I) surface electrogram.** (Cochrane et al 1983)

The study of the electrophysiology of the human heart has advanced rapidly since the development by Scherlag et al (1969) of a reproducible method of recording electrical activity from the region of the His bundle. Many procedures have been developed as useful indicators of cardiac electrical function or as aids to the localisation of abnormalities (Cochrane et al 1983). These tests are now performed routinely in many cardiac EP studies laboratories and over twenty years of clinical experience have amply demonstrated the value of the combined techniques of programmed stimulation and intracardiac recordings in the investigation and management of serious cardiac arrhythmias (Josephson and Seides 1979). These studies are however becoming increasingly complex and time consuming both during the procedure and during the subsequent analysis of the records (Ross et al 1980).

Table 2.1 **Routine Tests Performed during EP studies.**

(Cochrane et al 1983)

Basic conduction intervals
Basic cycle length
Intra-atrial conduction time
Atrioventricular (AV) nodal conduction time
His-Purkinje conduction time
Intraventricular conduction time
Sinus node recovery time
Sino-atrial conduction time
Incremental atrial pacing
Atrial extrastimulus
Incremental ventricular pacing and ventricular extrastimulus
Tachyarrhythmia initiation

By way of example, during an EP study performed by Antolini and Kirchner (1984) the activation parameters automatically measured were the cycle length and conduction times through the right atrium, AV node and His-Purkinje system. These times, shown in Figure 2.2, are obtained from four signals: one surface ECG lead, a high right atrial electrogram, a His bundle electrogram and, during the atrial pacing, the signal coming from the stimulator. Detection of the QRS is from the surface ECG, the high right atrial depolarisation is from the high right atrial and the low right atrial electrograms, His and ventricular depolarisations are from the His bundle electrogram.

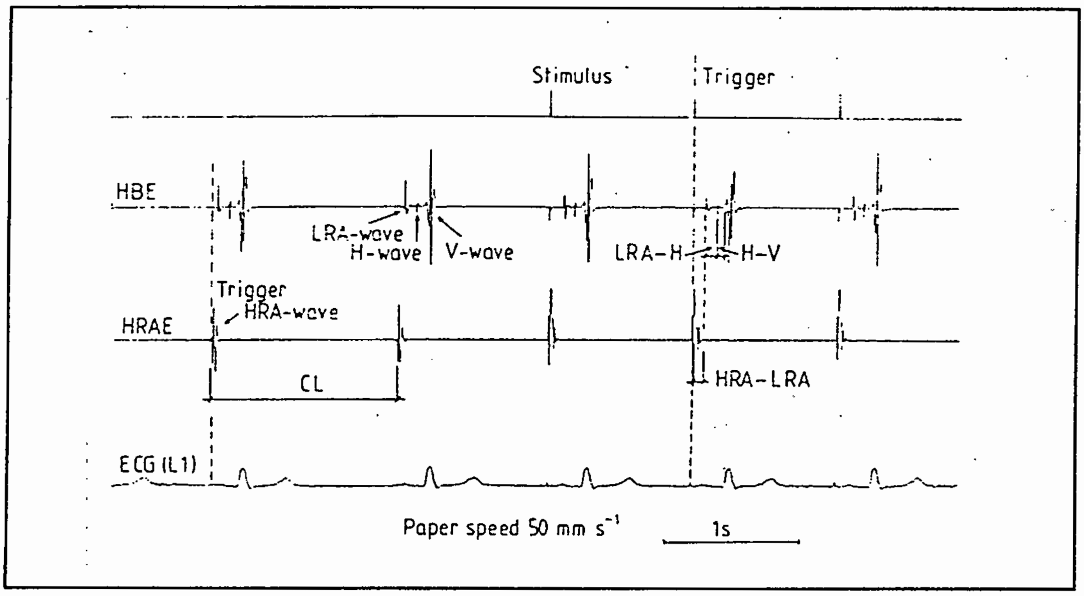
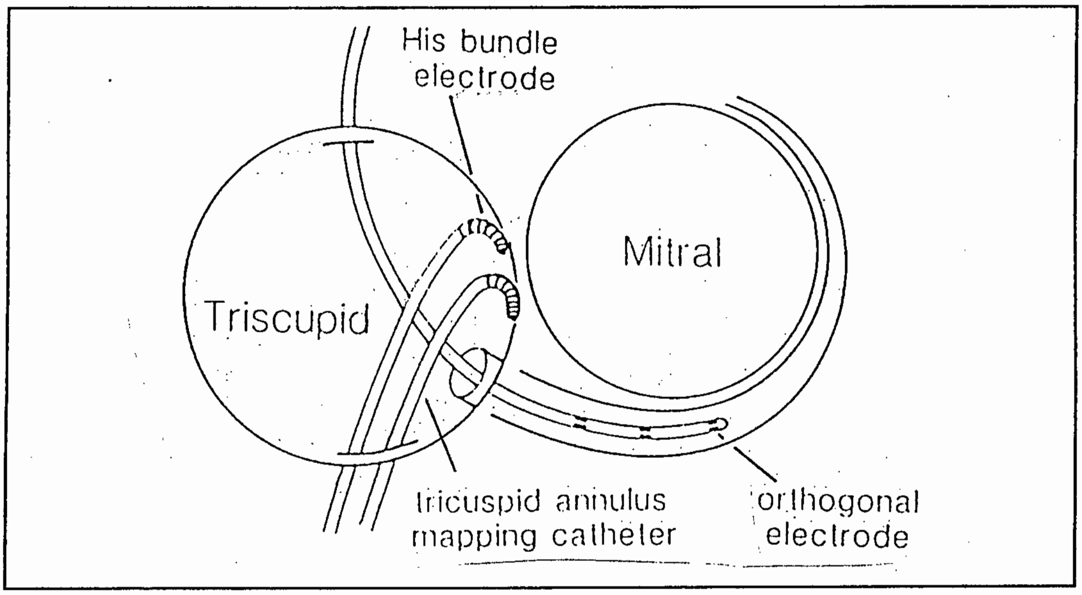


Figure 2.2 **Signals Analyzed During an EP Timing Study. From top to bottom: stimulator output, His bundle electrogram, high right atrial electrogram and standard ECG lead I. Both sinus and paced atrial cycles are shown with the trigger used to synchronise the measuring system.** (Antolini, Kirchner et al 1984)

Cardiac EP studies have evolved into widely employed clinical tools, often indispensable in evaluating patients with specific cardiac arrhythmias. Patients undergoing surgery for recurrent arrhythmias require some form of EP study determined by the type of arrhythmia and the method of surgical correction. The purpose of the EP study is to determine the probable mechanism of the arrhythmia, the components of the circuit, and the localisation of the substrate. Attempts at localisation should take place both during preoperative EP studies and at the time of surgery because intraoperative localisation procedures are important to define more precisely the arrhythmogenic substrate to be approached surgically and to localise additional pathways in patients with WPW syndrome or multiple tachycardias (Josephson and Harken 1983; Swerdlow et al 1986 and Miller et al 1985).



<u>**Figure 2.3**</u> **The AV Region as Viewed Fluoroscopically in the left Anterior Oblique Projection, illustrating the mapping technique. Large circles represent the tricuspid and mitral valve annuli.** (Jackman et al 1989)

Localisation by catheter techniques in the EP study laboratory is deemed necessary for directing surgical procedures (Zipes et al 1989) for the following reasons:

* the arrhythmia or arrhythmias may not be inducible in the operating room
* the only arrhythmia or arrhythmias induced intraoperatively may not be mappable
* not all arrhythmias observed spontaneously or induced by

preoperative EP studies are induced in the operating room

* rapid interpretation of intraoperative data may be difficult
* intraoperative catastrophes may preclude induction or mapping, or both, of arrhythmias.

Recent developments in EP studies include catheter ablation of specific areas of the myocardium (Seally et al 1976 from Zipes et al 1989). One example of this ablation technique is the ablation of the AV junction described by Gallagher et al (1982). This method of catheter ablation has eliminated the need for intraoperative surgical ablation of the AV junction in 85% or more of the cases (Cox 1985).

2.2.2 Wolff-Parkinson-White (WPW) syndrome mapping

The WPW syndrome is characterised by the presence of one or more abnormal electrical connections (accessory pathways) between the atria and ventricles. This results in two routes for the electrical impulse to proceed from the atria to the ventricles: along the His bundle and through the accessory pathway.

In the WPW syndrome, the presence of an accessory AV pathway leads to early activation of a part or the whole of the ventricle by an impulse originating in the sinus node or atrium (Figure 2.4). This second pathway between atrium and ventricle makes it possible for arrhythmias to occur. The two most frequently occurring arrhythmias are a circus movement tachycardia and atrial fibrillation.

During circus movement tachycardia, the impulse travels in a circuit consisting of atrium -> AV node -> His bundle -> bundle branch ->

ventricle -> accessory pathway -> atrium. A circus movement tachycardia with AV conduction over the AV node and ventriculoatrial conduction over the accessory AV pathway is due to uni-directional block in one of the two AV pathways (Figure 2.5). This can occur with a critically timed atrial or ventricular premature complex, by reaching a critical sinus rate, or after the administration of a drug.

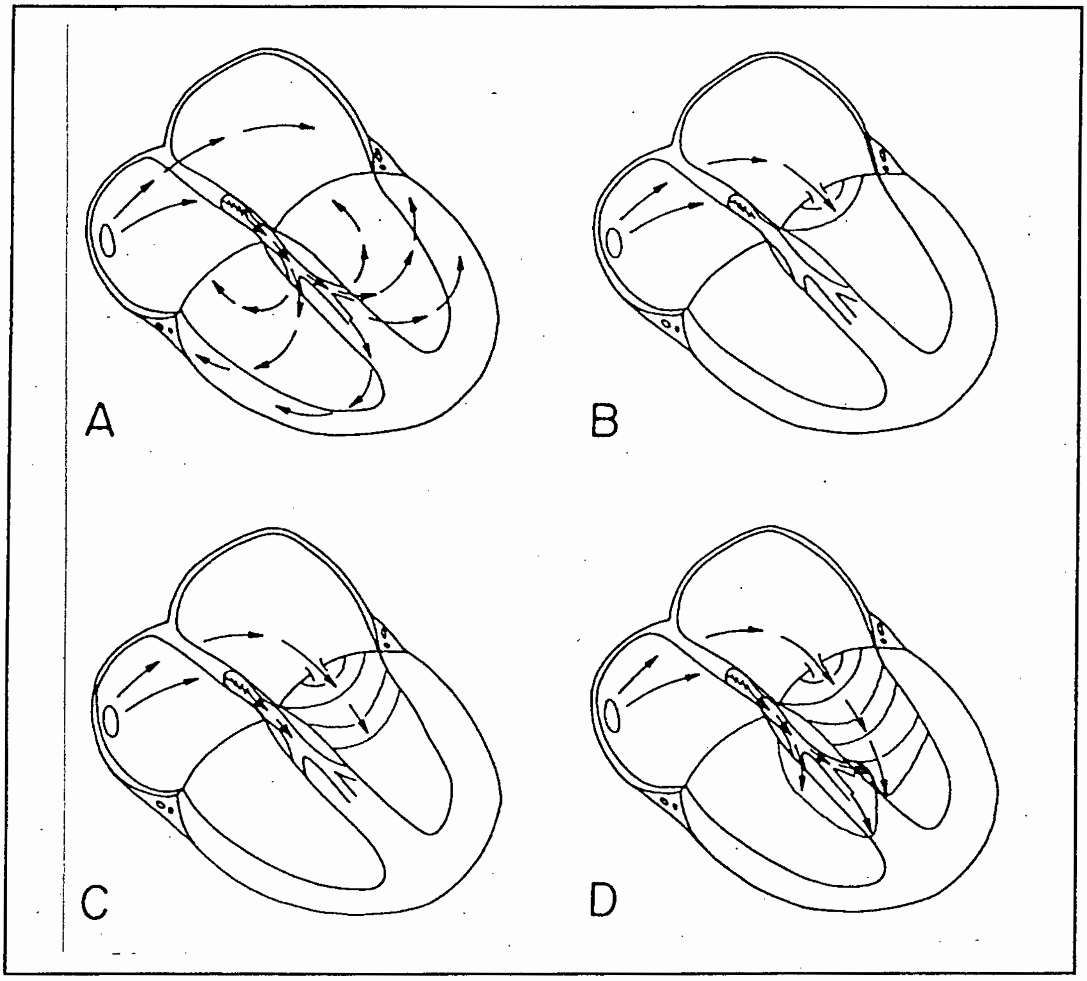


Figure 2.4 **(A) Normal Spread of Electrical Activation of the Heart During Sinus Rhythm. (B-D) Spread of Electrical Activation of the Heart During Sinus Rhythm in WPW Syndrome With an Accessory Pathway in Left Free-wall Position** (Cox and Ferguson 1989).

Atrial fibrillation in patients with the WPW syndrome can prove to be an extremely serious arrhythmia in the setting of an accessory pathway with a short anterograde refractory period. In this situation the

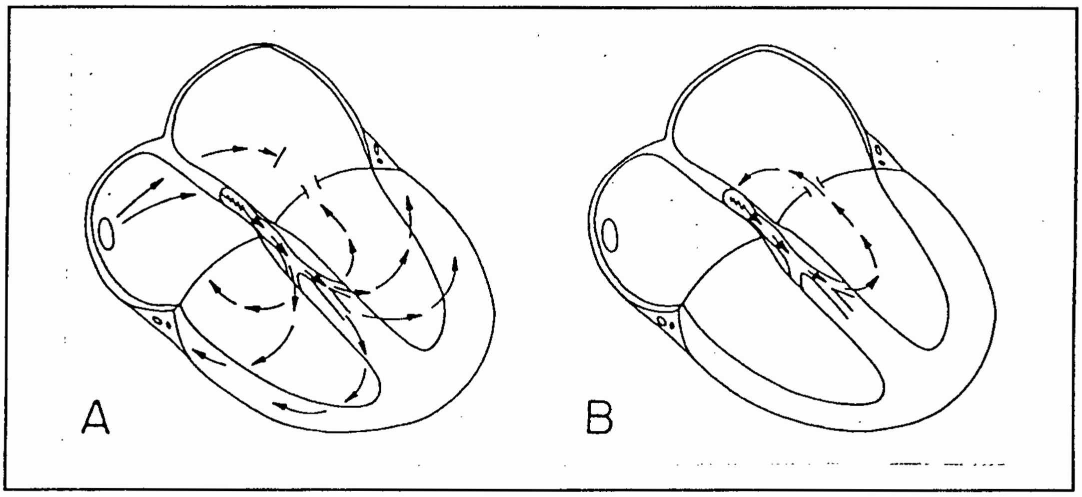


Figure 2.5 **A Macro-reentrant Circuit due to Unidirectional Block and Accessory Pathway in the Left Free Wall.**

occurrence of ventricular fibrillation is possible. The EP properties of the accessory pathway are an important determinant of the incidence and rate of a circus movement tachycardia and of ventricular rate during atrial fibrillation. EP studies using multipolar electrode catheters provide information about these properties of the accessory pathways and their possible locations (Zipes et al 1989).

Although EP studies frequently provide accurate and reliable data (Nzayinambaho et al 1989), the physical locations of the sites to be ablated must be verified in situ by electrical activity mapping. Successful ablation of accessory AV pathways requires precise knowledge of the locations of the atrial and ventricular insertions of the muscle fibres. In particular, septal and paraseptal pathways in the vicinity of the AV node and His-Purkinje system must be carefully isolated

(Buckles et al 1990).

Mapping in EP studies is the process whereby electrogram signals are acquired and displayed. It is performed during normal sinus rhythm, or by using atrial or ventricular pacing to increase the disparity between normal and abnormal conduction pathways, or after initiation of tachycardia. Ablation is usually performed at the site of earliest recorded activation during the tachycardia (Fisher 1988). Thus, a complete, accurately drawn, isochronal map is not absolutely required (Ideker et al 1989). Multichannel mapping is nevertheless very important and Buckles et al (1990) discusses the problems associated with the use of a single channel system in mapping WPW syndrome and the need, if only a single channel is available, to use pharmacological manipulations in order to locate accessory pathways accurately.

The measurement methods outlined in the paragraph below are practised by Dr. J. Cox and his associates of Barnes Hospital, St. Louis, Missouri. The success rate of these procedures in localising accessory pathways has been 100% (Cox 1989).

Before epicardial measurements are taken the chest is opened and the heart is exposed. Epicardial pacing and sensing electrodes are sutured onto the atrium and ventricle near the suspected site of the accessory pathway. A band electrode (Figure 2.6) is placed along the ventricular side of the AV groove and electrograms are recorded simultaneously from sixteen bipolar electrodes during normal sinus rhythm and also during atrial pacing. The electrogram signals are acquired, displayed and one beat is then selected for processing.

These simultaneously recorded electrograms are processed by a computer system and Figure 2.7 is an example of the computer screen showing the activation times recorded at each electrode. The electrogram recorded from the electrode located nearest to the accessory pathway shows the earliest activation. The time intervals in milliseconds are measured relative to activation at a reference electrode. Three surface lead representations of the selected beat are shown in the right of the figure.

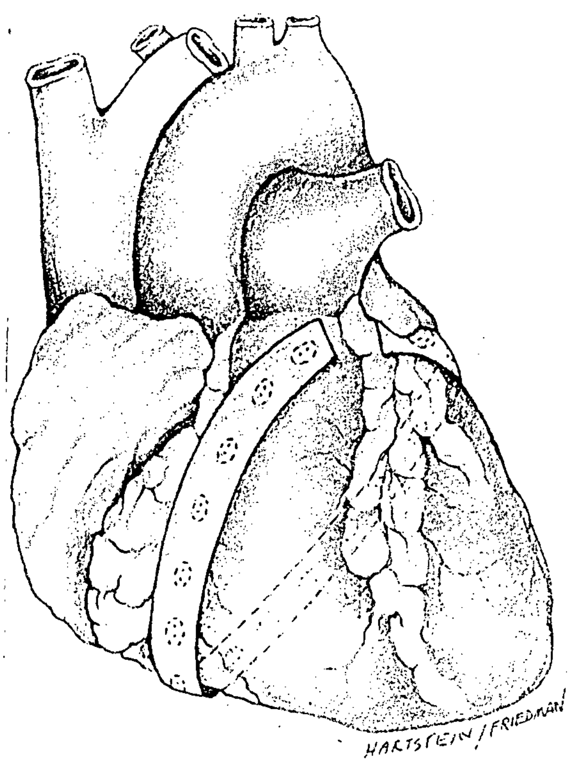


Figure 2.6 The Band Electrode in Position on the Ventricular Side of the AV Groove.

If multiple electrodes are used, they must remain fixed on the heart until after the map is drawn (or must be replaced in exactly the same locations). This is necessary to ensure correlation between information displayed in the map and physical sites on the heart surface. For example, if an accessory pathway (corresponding to a site of earliest ventricular activation) is indicated to be between electrodes 12 and 13, then the surgeon in the operating room or the investigator in the laboratory needs to locate electrodes 12 and 13 on the heart surface and attempt to identify this region by direct inspection.

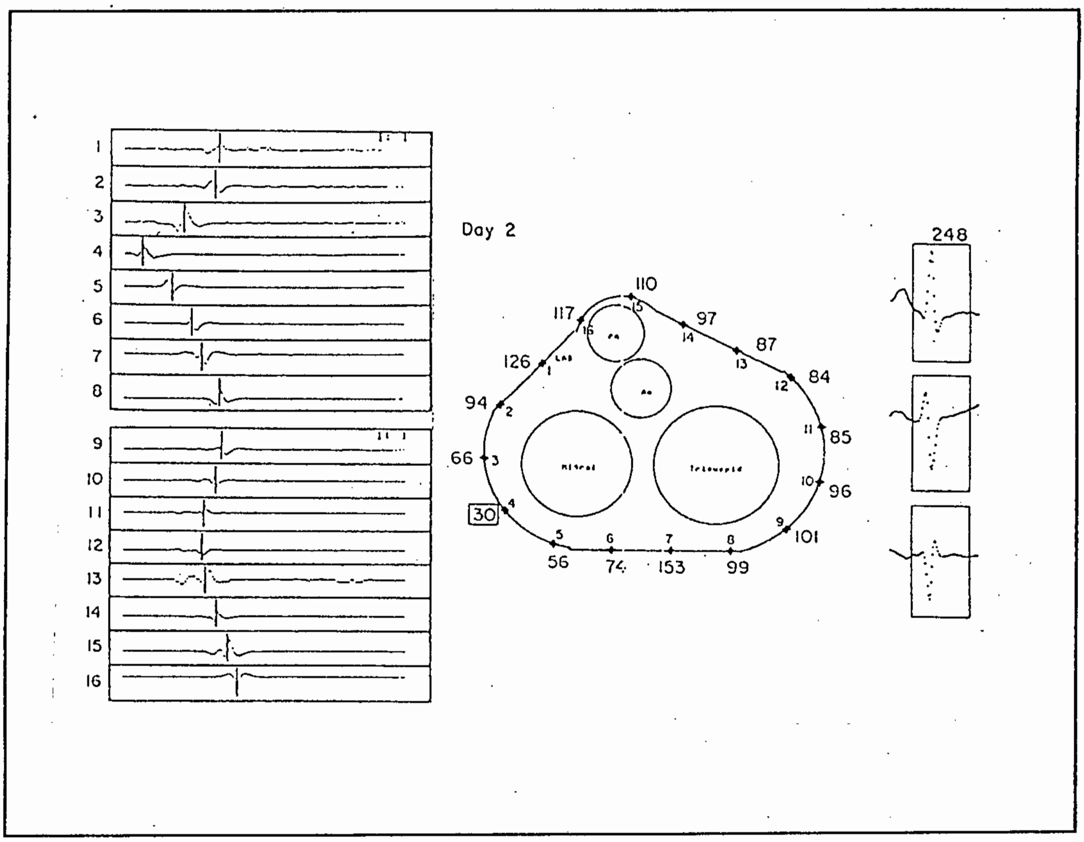


Figure 2.7 **The Graphics Terminal Display Showing the Activation Sequence of Sixteen Electrodes in the Band Configuration Around the Heart.** (Cox and Ferguson 1989)

Some tachycardias may be caused by micro-reentrant circuits. These circuits may be no more than 5mm in diameter (Wit et al 1982) and may be considerably smaller (Spach et al 1988). To map such small circuits, distances between electrodes should be known with an error of less than

1 mm. One method of ensuring this small interelectrode spacing while reducing the total number of electrodes is to use close electrode spacing only where it is needed; for example, in regions thought to be the sites of arrhythmia origin. In such cases a map can first be produced using wide electrode spacing. Areas of interest in this map can then be mapped again with closer interelectrode spacing while the arrhythmia is reinduced. This method does assume that the arrhythmia is unchanging and can be induced more than once (Ideker et al 1989).

Accessory pathways in WPW syndrome can be localised to one of four anatomic areas: the left free-wall, right free-wall, posterior septal and anterior septal regions (refer to figure in Appendix A for details of these regions). All the atrio-ventricular connections of these pathways must insert into the ventricle somewhere within these anatomic boundaries. If an accessory pathway is found in any particular region then correct dissection of the entire region will ensure ablation of the pathway.

2.3 Data Acquisition and Signal Processing Techniques

"A microcomputer-based electrophysiologic system analyzer combining computer-assisted protocol operation, automated event detection and analysis, data collection and archival, and patient report capability promises to provide an extremely useful and cost- effective tool for clinical and research electrophysiologic studies" (Gillette et al 1989).

Bonneau et al (1987) lists the basic requirements for an efficient cardiac mapping system as:

* simultaneous recording of cardiac potentials at multiple sites to

shorten the duration of investigations in antiarrhythmia surgery and to allow observation of transient phenomena that last only a few beats, which is not possible with multiple sequential recordings

* rapid analysis of the cardiac activation sequences to guide interventions such as changes in the location of the recording electrodes or ablation of the arrhythmogenic sites
* interactive software to allow rapid and easy control of all the functions of the system, such as setting up the amplification and sampling rate, recording data, detecting local activation times on the electrograms, generating activation maps and managing data.

These requirements have been met to varying degrees by existing systems which have a wide range of technical specifications and capabilities for data acquisition as well as for data analysis (Downar et al 1984; Dhein et al 1988 and Hernandez et al 1984). Most systems are based on a digital approach and since medical-grade amplifiers meeting the multiple-channel requirement and safety standards in a compact package are not commonly available this first part of the system has usually been locally constructed.

As for the A/D conversion, some groups have used a straightforward approach by interconnecting signal outputs from their own amplifiers to the standard A/D converter of available data acquisition systems while others have built their own A/D conversion hardware, interfacing it to a computer, and developed software for data acquisition and map generation (Antolini and Disertori et al 1984; Buckles et al 1990; de Bakker et al 1984; Witkowski and Corr 1984; van Heuningen et al 1984;

Jenkins et al 1987 and Hoeks et al 1988). In general this approach has resulted in systems better suited to the requirements of electrophysiological investigations (Bonneau et al 1987).

Antolini and Disertori (1984) for example has developed a system for on-line real-time measurement of the EP activation intervals. The system is based on Motorola Micromodules (Motorola Inc 1979) with the addition of two home-made cards: an analogue signal conditioning module and a real-time clock.

Buckles et al (1990), on the other hand, acquires the electrogram signals by means of a physiological recorder widely used for EP studies (PPG VR-16) and then feeds the signals to a Compaq 386/40 micro computer (Compaq Deskpro) via a Data Translation DT2801A A/D board.

Wikowski and Corr (1984) uses locally constructed amplification, gain and digitisation boards that feed the signals to a high density tape recorder for storage. Analysis is then performed off-line by transferring data to a digital computer system.

The approaches outlined above have most often been implemented under the supervision of individual research teams or institutions engaged in cardiac mapping. The construction of a specific mapping system is undertaken by them in order to satisfy the particular requirements of their application. The reader is referred to the literature for specific details of the various systems.

2.3.1 Hardware and software techniques

Data acquisition involves a number of different techniques to ensure accurate acquisition and display of signals. These electronic techniques address issues such as interference and noise reduction, amplification, filtering, isolation and circuit protection. There are many areas for consideration within this vast field and even the recommendations for standardisation of bandwidth and digital signal processing techniques in automated electrocardiology given by Bailey et al (1990) highlight only a subset of the applicable areas within the field of software and hardware techniques used in data acquisition. In no way attempting to cover the entire field the areas of particular importance for this application are covered briefly in the following paragraphs.

Interference and Noise Reduction. Bioelectric recordings are often disturbed by an excessive level of interference. The cause of such disturbance is not obvious and the use of equipment with very good specifications does not guarantee interference-free recordings (Metting van Rijn et al 1990). In most bioelectric measurements an interference level of 1-10 μV peak-to-peak (less than 1% of the peak-to-peak value of the desired signal) is acceptable (Spekhorst et al 1988).

Interference can be attributed to internal or external sources. <u>Internal</u> sources include activity of bioelectricity generators (such as respiratory muscles, see Figure 2.8) and movement artifacts of the electrode-patient interface, as well as electrical noise of the signal acquisition and processing circuitry.
<u>External</u> sources include mains power conductors and high frequency

generators such as x-ray and electrosurgical units. This interference is coupled to the measurement system by capacitive (electric) or inductive (magnetic) coupling between the sources and the patient and/or the leads connecting the patient to the system. The reader is referred to Appendix C for a detailed explanation of the sources of interference and the methods used for their reduction.

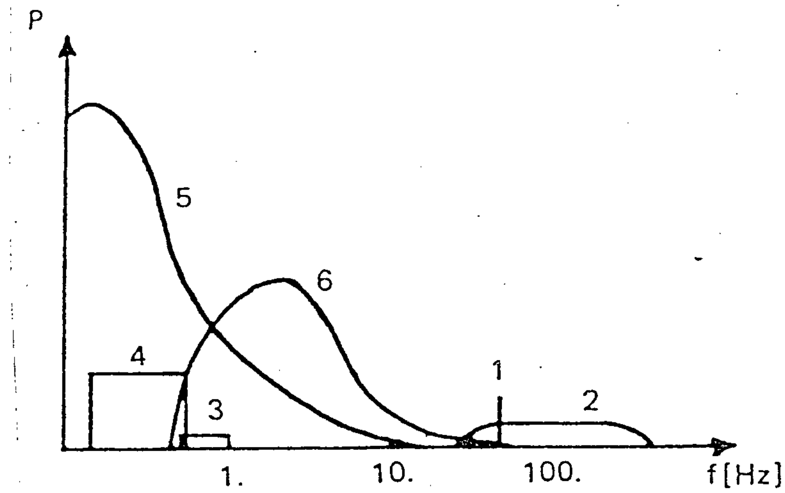


Figure 2.8 **Power Spectrum of ECG with noises and other signals.**

1 - Mains interference 2 - Muscular artifacts

3 - Movement artifacts 4 - Respiratory artifacts

5 - DC offset potential 6 - ECG signal

Amplification of the input signals is needed to increase the signal levels to well above the noise level of the system's electronic circuitry. Standard amplification for surface ECG signals is 1000X and less (typically 500X) for epicardial or intracardiac signals, depending on whether unipolar or bipolar measurements are made.

Filtering is used to exclude (or diminish) the effect of unwanted signals (such as noise) and can be achieved by both hardware and software techniques or a combination of the two. It is important to realise that the filtering in a mapping system will distort the waveform and timing of the signal (Thomas 1982), thus altering the waveform. The filtering criteria of an acquisition system differ depending on the type of investigation being performed.

Cutoff values (-3dB points) of filters for the surface ECG leads are typically 0.1-100 Hz while for internally-derived signals they are 30 - 250 Hz. A brief experiment to ascertain the bandwidth for intra-cardiac signals was perfromed and involved playback of recorded signals via a variable bandpass filter onto an oscilloscope. The filter cutoff values were decreased from maximum outer settings of 0.1 and 500 Hz and the cardiologist was asked to indicate values beyond which excessive distortion or loss of information occured. These values coincided closely with values routinely used (30 - 250 Hz) in the clinical setting.

Baseline detection. Cerutti et al (1980) states that one of the most important operations prior to any automatic processing of ECG tracings is that of baseline detection and correction and corresponding low-frequency noise reduction. Such steps may be accomplished by means of software using different algorithms and criteria. It is possible to use a threshold method which eliminates the complexes in which the baseline has a fluctuation larger than the threshold value and equalises it in the other complexes (Pipberger 1965). Other methods use frequency spectrum filtering (high pass filter), Fourier expansion, linear correction and linear regression (using the least squares methods) (Mac Farlane and Lawrie 1974), higher order polynomial regression (cubic regression in Meyer and Keiser (1977, from Cerutti et al 1980)) and finally sinusoidal functions regression.

Wellner and Brodda (1976, from Cerutti et al 1980) has demonstrated how much the measurements of the onset, maximum and off-set of the various ECG waves are affected by varying the methods of baseline detection: while the QRS does not seem to be too sensitive, the values for P and T waves vary considerably from one method to another.

Authors have claimed that a high-pass filter aggressive enough to remove the most common baseline drifts from standard ECG tracings should have its lower -3 dB point above 0.5 Hz (Meyer and Keiser 1977 from Cerutti et al 1980). This differs from the FDA and AHA Standards which require the -3 dB point not to be above 0.05 Hz, particularly to safeguard the low frequency components in the diagnostic ECG, such as the ST displacements. It should be noted that these components are not normally considered in EP mapping studies.

Isolation. The proliferation of monitoring equipment in hospitals, clinics and research establishments has increased the risk of electrical shock to the patient. This is especially so in the case of the critical care or operating environments where patients are connected to a number of electro-medical devices. Should an invasive procedure be performed the patient contact impedance is furthur lowered, for example, in the case of an EP study all the patient leakage current would flow through the heart.

The patient must be protected by providing a very high impedance path between the patient and the mains supply (and other sources or sinks of electrical current). The path impedance must be high enough to limit the worst-case current to below 10 μA. A current of 100 μA is sufficient to induce ventricular fibrillation (Hill and Dolan 1976). This high impedance path is obtained by using either isolation amplifiers, transformers or optical isolators within the electronic circuitry attached to the patient (see Smith 1986).

Digitisation of a signal involves both interval sampling and quantization of that signal. Sampling of the signal will introduce error that is a function of the sampling rate while the quantization error will depend on the number of bits used to represent any particular voltage. Electrograms are frequently recorded by immediately converting the input analogue signals to digital form. When such techniques are implemented it is important to ensure that all the required information is retained in the digitized signal. Rosenbaum et al (1989) states that despite the importance of accurate delineation of arrhythmia pathways, the precision of this technique and its dependence on computer sampling rate have not been systematically evaluated.

In his study to ascertain the required minimum sampling rates Barr and Spach (1977) states "sampling rates as high as 15 000 samples/second were required to record accurately extracellular waveforms of the ventricular conduction system. Decreasing sampling rates were required as the recording site shifted through the ventricle to the body surface where sampling rates as high as 1 500 samples/second were necessary".

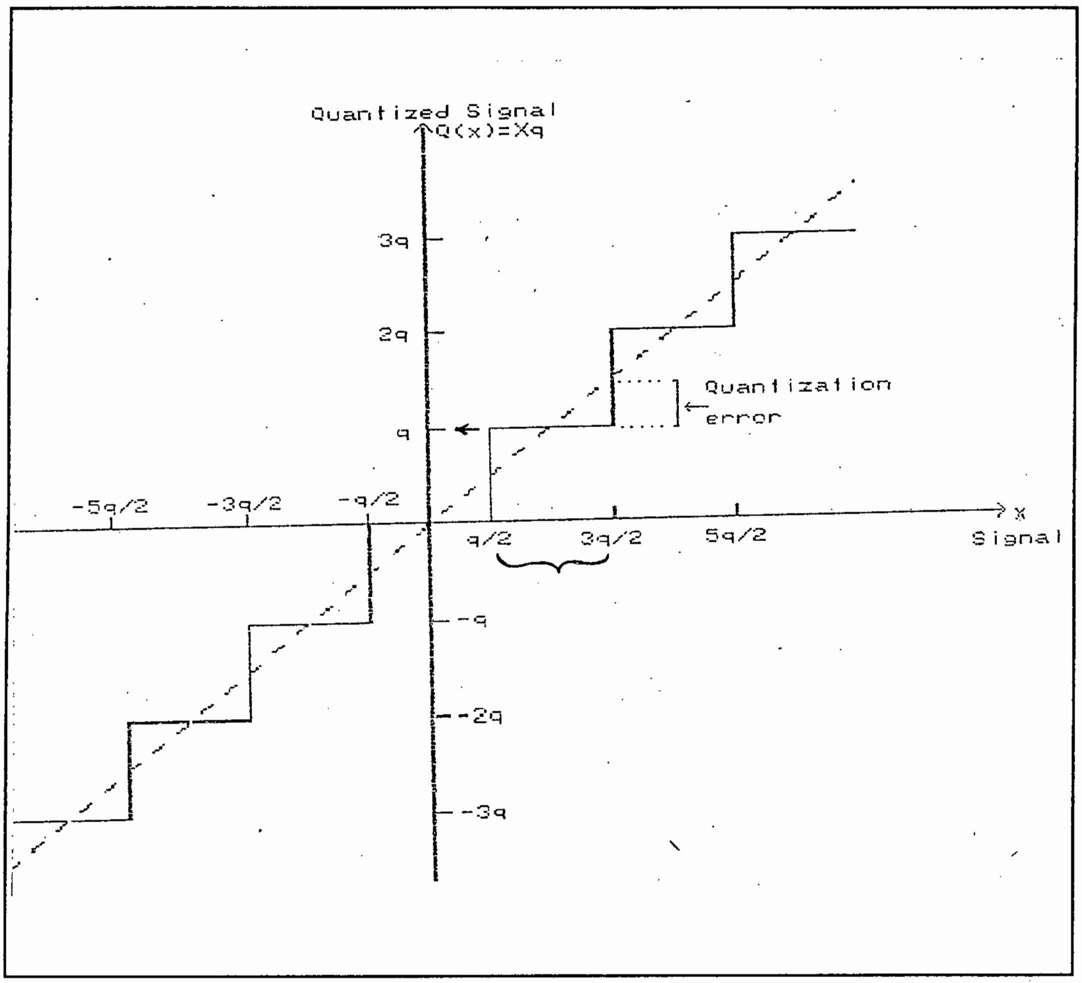


Figure 2.9 **Signal Quantization (eg: input voltage levels between q/2 and 3q/2 are assigned the digitized value of q).**

The quantization used in the representation of each sample is an important factor in determining the fidelity of the digitized signal (Figure 2.9). The more levels used to represent the signal the more closely the digital signal approximates the analogue signal. Typical digitisation integrated circuits use eight or twelve bits, but systems are available in which the number of bits is sixteen or even more.

Display. The user-interface is an important aspect of system design and the clear display of the electrogram data is a primary goal. It is therefore important that the highest resolution possible is made available for the graphical output. The ability to manually detect the activation times from the electrograms on the screen is the deciding factor as to whether or not the resolution is sufficient.

2.3.2 Activation time detection

The detection of activation time is an important requirement in the mapping of WPW patients because ablation is usually performed at the site of earliest recorded activation during the tachycardia. The electrogram activation time at an electrode site is a weighted mean of the activation times of the surrounding myofibres with each weighting factor inversely proportional to some power of the distance from the myofibre to the electrode (Ideker et al 1989).

When a small extracellular electrode is placed very close to a muscle fibre, the time of maximum downslope recorded from the electrode is approximately the time of maximum upslope of the action potential of the adjacent sarcolemma (Spach et al 1972; Spach and Kootsey 1985). It is this time of maximum downslope that is taken as the activation time recorded from a unipolar electrode (Ideker et al 1989). Unipolar recordings reflect influences from the whole heart with the intrinsic deflection corresponding to activation of the tissue in close proximity to the electrode (Witkowski and Corr 1984).

Simple model studies indicate that the activation time for a bipolar electrode should be taken at the time of peak absolute potential (Funada et al 1983, from Ideker et al 1989), a time which has been shown to correspond closely to the time of maximum downslope in a simultaneously recorded unipolar electrode (Blanchard et al 1985 and Blanchard et al 1988, from Ideker et al 1989). Even though the peak of the bipolar signal is thought to correspond to the time of activation (Itatsu 1954 from Ideker et al 1989) the presence or absence of activation in a bipolar recording is, according to Claydon et al (1985) better determined from the maximum absolute slope instead of the maximum amplitude.

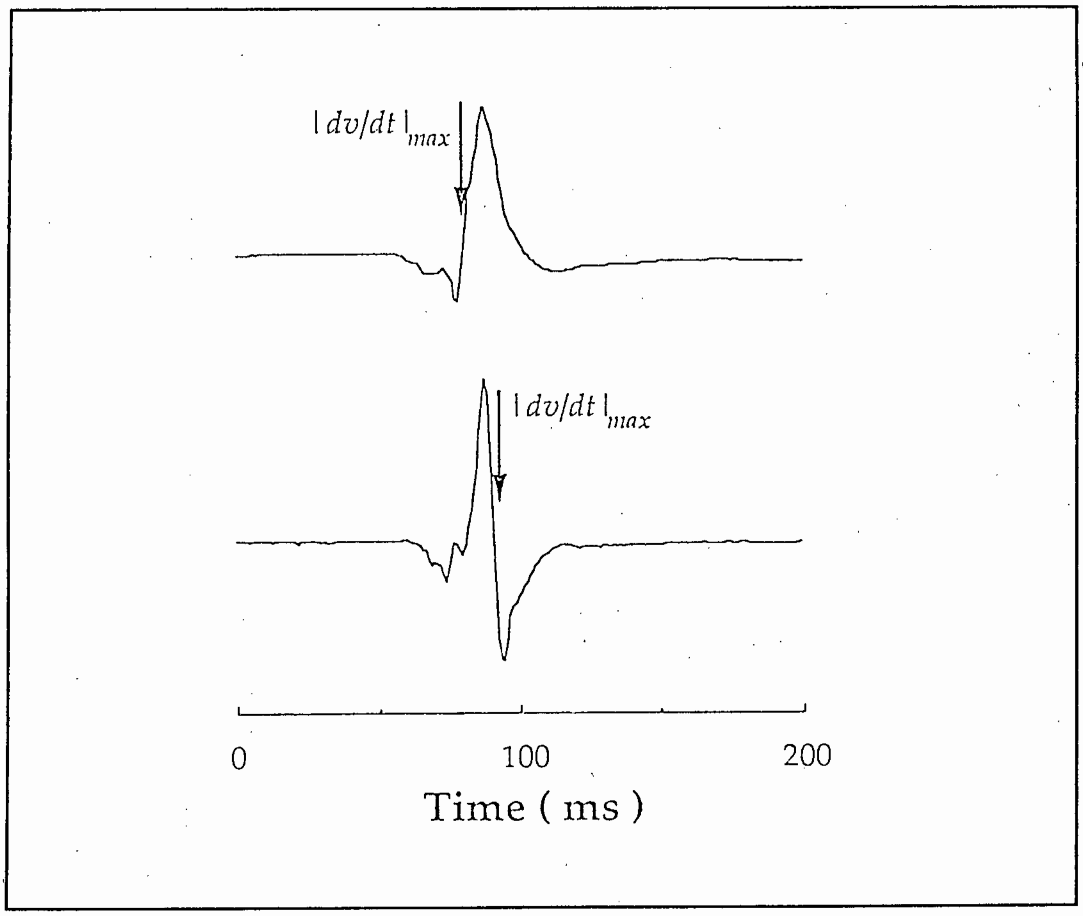


Figure 2.10 **Unipolar Deflection Detection using maximum first derivative criteria.** (Rosenbaum et al 1989)

In his study on the precision of electrophysiological mapping Rosenbaum et al (1989) used an automated estimate of local electrical activation based on the maximum first derivative ($|dV/dT|_{max}$). Local activation was thus defined as the time coinciding with the onset of the electrogram's earliest rapid deflection. Figure 2.10 shows the detection of two electrograms of similar morphology using the maximum first derivative criteria. Rigid adherence to maximum derivative criteria would result in a 9 millisecond delay of local activation in the lower signal despite the appearance of nearly simultaneous activation waveforms at both recording sites.

There is not yet agreement on whether unipolar or bipolar electrodes are better for detecting activation. For detecting local activation, the bipolar electrode has the advantage that its recording field falls off much faster with distance from the electrode so that it is less sensitive to distant activation fronts. This sensitivity to distant activation fronts can be further reduced by decreasing the distance between the two poles of the bipole (Ideker et al 1989).

The sampling rate is an important factor in determining whether an activation has occurred as well as determining the timing of the activation (Barr and Spach 1977). With few exceptions (Wit et al 1982; Witkowski and Corr 1984) most mapping systems sample at a rate of 1 000 samples/second or less. Recent preliminary evidence suggests that this rate may be too slow even for unipolar electrodes. Frequencies greater than 500 Hz, the highest that can be detected with a sampling rate of 1 000 samples/second according to the Nyquist theory, appear to contain information about the presence or absence of activation (Smith et al 1987 from Ideker et al 1989).

2.3.3 Commercial systems

Dedicated mapping systems are available commercially, for example the Bard Cardiac Mapping System (Bard, Billerica), the Life Science Data Acquisition and Analysis for the Macintosh (BIOPAC Systems, Goleta), the CardioMapp (Arrhythmia Research Technology, Inc, Austin) and the Cardimap (Ingg. Battaglia-Rangoni, Bologna).

The Bard cardiac mapping system is a system consisting of a 64 channel amplifier interfaced to a specially configured microcomputer with 4 megabytes (Mb) random access memory (RAM - see glossary), a 40 Mb internal hard disk drive, a high resolution colour monitor and a graphics printer. Data capture, analysis, storage, and isochrone map generation are controlled by menu-driven, dedicated software. Up to 56 intracardiac signals (unipolar or bipolar) and 7 surface ECG signals can be acquired simultaneously during an 8 second "window" for any rhythm of interest. Each signal is amplified, filtered and digitized. These signals can then be displayed for subsequent analysis on a high resolution graphics screen. Up to 16 channels can be displayed on the monitor in real time. A variety of electrode arrays including A-V groove bands, socks and plaques are available for use with the system for a variety of mapping requirements. This entire system retails at approximately R400 000.

The BIOPAC system consists of a number of input modules (ECG, EEG, temperature, pressure) that connect to the computer and acquire data under the control of the "MP100" data acquisition software. The Software is written specifically for the Macintosh. Each input module contains a maximum of eight channels and adaptors are necessary for multichannel expansion.

The CardioMapp line of cardiac mapping systems consists of three basic models. The most powerful in the range is the CardioMapp 255 which allows the simultaneous acquisition of 255 input channels at a sampling rate of 512 or 1024 samples per second. There is a maximum of nine ECG channels and six auxiliary cardiac channels. The input devices are the standard keyboard and a trackball. Data is saved onto an erasable optical disk with removable media.

The Cardimap consists of 36 differential input channels that are amplified, filtered and digitized before being stored to computer memory. The software allows for modification of the gain, filtering or sampling rate before acquisition. Storage of the data is on floppy disk. A 35 electrode isopotential map is drawn and manipulation of these maps is possible for sequential step mapping, smoothing and isointegral mapping.

The specific needs at GSH are for a single data acquisition and mapping system that can be used for surgically oriented WPW mapping as well as EP studies. The commercial systems mentioned above do not fulfil these specific requirements for the following reasons:

1. Systems dedicated to intraoperative mapping are not easily adaptable to meet existing or future requirements.
2. They are not available locally and are prohibitively priced.
3. There is no local backup for repair and maintenance of these systems.

CHAPTER THREE

SYSTEM SPECIFICATION

3.1 Introduction

In the method presently used for intra-operative WPW mapping at GSH the surgeon places a probe sequentially at sites around the AV groove, usually during AV re-entry or RV pacing. The activation times of the signals recorded from the probe are compared to a reference electrode. The process of recording takes 10-15 minutes, after which the traces are examined and activation times calculated. This takes a further 10-15 minutes. A significant reduction of this time would be both beneficial and advantageous. Simultaneous acquisition of signals from many sites would allow for mapping of single beats.

3.2 Discussion of requirements

The proposed method is to record simultaneously from multiple sites via an electrode array. The electrode array would be in the form of a belt and would be placed in position on the heart by the surgeon. This method would involve:-

* simultaneous display of multiple channels together with reference and surface ECG signals
* computer calculation of activation times at each site relative to the reference electrogram
* display of the activation times numerically and graphically.

The requirements of this automated mapping system are:-

* simultaneous recording of electrograms from twenty four electrodes plus surface ECG (at least two leads) and atrial and ventricular reference electrograms
* the ability to choose between unipolar or bipolar recording
* a frequency response of 0.1 - 500 Hz (filtering 30-500 Hz for

electrograms and 0.1 or 1.0 - 100 Hz for surface ECG)

* display of these signals in real-time (on a slave monitor)
* capture of a segment on computer screen for editing
* ability to exclude unsuitable complexes from analysis
* automatic selection of activation points on the signal (preferably choice of maximum slope or maximum amplitude)
* the ability to modify the activation points via a cursor
* the ability to update the segment repeatedly if necessary
* the ability to record signals from a manual probe when necessary and to integrate them with the above
* graphic display of relative activation times at each electrode site
* simple, menu driven user interface, and other ergonomic features
* compliance with accepted standards of patient safety (isolation) including protection from defibrillation discharges
* the system should also be able to accept inputs from electrode catheters so it can be used in EP studies.

The system required by the Cardiac Clinic at Groote Schuur Hospital is to have thirty two channels of analogue input. For the purposes of the thesis (constraints of time and finance) it was decided that only four input channels be implemented. A major design criterion however is to ensure ease of expansion from the four channels of the prototype system to the thirty two channels of the desired system. With this expansion in mind it is necessary that the software caters for the acquisition and storage of all 32 channels.

Ergonomically, several requirements are considered to be important in the design of a suitable system. These are:-

Ease of use. This is a prime consideration and much thought was given to designing a system that a cardiologist, with minimal knowledge of computing, could learn and use with ease in theatre and in the EP study laboratory.

Efficiency. The system must guarantee a rapid response to user input. For applications requiring data storage and retrieval, thought must be given to efficient methods of storing and retrieving data as well as the response of the computer to various keystrokes.

Data validation. It is accepted that some types of data input error will arise either because the information is not available at the time of input or for other reasons. Each item of operator input need not be individually validated however, but where the information being input is important for data retrieval purposes (such as the patient name or date of birth) or for other reasons obvious validation checks should be included at data entry.

Adaptability. In a research oriented department, adaptability is essential for the refinement or expansion of existing applications and for the development of new applications.

3.3 System specifications

The data acquisition unit includes 32 channels with instrumentation amplifiers and allows software programmable gain and bandwidth. Signals are sampled, digitized and transferred to the computer on an optically isolated data bus. Table 3.1 gives the hardware specifications of the acquisition system.

The software of the system is written in Turbo Pascal (v5, Borland International). It allows for the storage and graphical display of all

(32) input channels. A timing graph displays the activation times of all the channels. Only eight channels are displayed on the screen at any one time. The activation time is automatically detected for 24 of the electrogram inputs and these are displayed graphically. A choice of activation detection algorithms is available - either using the maximum absolute slope or the maximum amplitude within a specified time window. Manual editing of the activation identifiers is possible.

Table 3.1 **Functional Specifications of System Hardware**

Number of channels	32
Resolution	12 bit plus sign bit
Sampling rate	2 kHz per channel
Programmable gains	1 2 4 8 16 32 64 128
Programmable bandwidths (Hz)	
High Pass	0.1, 1, 30, 50
Low Pass	100, 250, 400, 500
Input impedance	> 10 MΩ
CMRR	90 dB
Dynamic range (input amplifier)	±0.5 V (no saturation)
Overvoltage protection	6 500 V
Isolation	Opto-isolators

CHAPTER FOUR

HARDWARE DEVELOPMENT

4.1 Introduction

The hardware aspects of the data acquisition system include the analogue interface to the patient, signal conditioning of the input waveforms and the digital interface to the computer. The hardware incorporates circuitry for the safety of the patient (from electrical shock) as well as protection of the electronics from large voltage inputs. Due to the low amplitude of the input electrogram signal, recording is particularly susceptible to interference from other physiological and external electrical signals. Sources of interference, and the measures used to reduce such interference, are covered in greater depth in Appendix C.

Bearing in mind the applications of the acquisition system the requirements for signal conditioning are:

* Increase the amplitude of the electrogram signal sufficiently to allow direct interface with the input of the analogue to digital converter.
* Apply conventional bandpass filtering to the electrogram and surface ECG signals for attenuation of low and high frequency interference.
* Apply anti-aliasing filtering to the signals prior to sampling.
* Isolate the patient from dangerous voltages that may occur in the event of equipment failure.
* Protect the input circuitry from high input voltages (such as defibrillation discharges).

These specific requirements have been achieved by partitioning the hardware into a number of different printed circuit boards, each with its own specific function. As mentioned previously only four channels are implemented, but expansion from this prototype is possible by interconnecting duplicates of the existing boards together in the correct configuration.

The hardware for the four channels consists of four printed circuit boards plus an IBM prototype board that fits into the computer. Due to the modularity in construction and function each board is explained in a separate section. Interconnections between the boards are made by means of electrical cables and connectors. The circuit diagrams for the boards are given in Appendix D.

4.2 Analogue Input

The analogue input board is represented by circuit diagrams WPWELEC1 and WPWELEC2 - BOARD ONE. A block diagram of the board showing its functions is shown in Figure 4.1. The board consists of four channels A, B, C and D. An electrogram amplifier for each channel was designed and constructed on this board with the following characteristics :-

* True differential-guarded amplification with high common mode rejection ratio (CMRR, see glossary) of 90 dB (manufacturer's specifications - Burr Brown) that eliminates the need for power-line-frequency notch filtering which can produce spurious ringing in the signal. The amplification choices available for this primary stage can be set to 1, 10, 100 or 1000 times by means of jumpers on the board. The small input signals from the heart (approximately 0.5mV to 10mV) are prone to interference and in an attempt to reduce this interference a ground plane has been used to shield the board and the instrumentation amplifiers are positioned in close proximity to this ground plane and the input

signals.

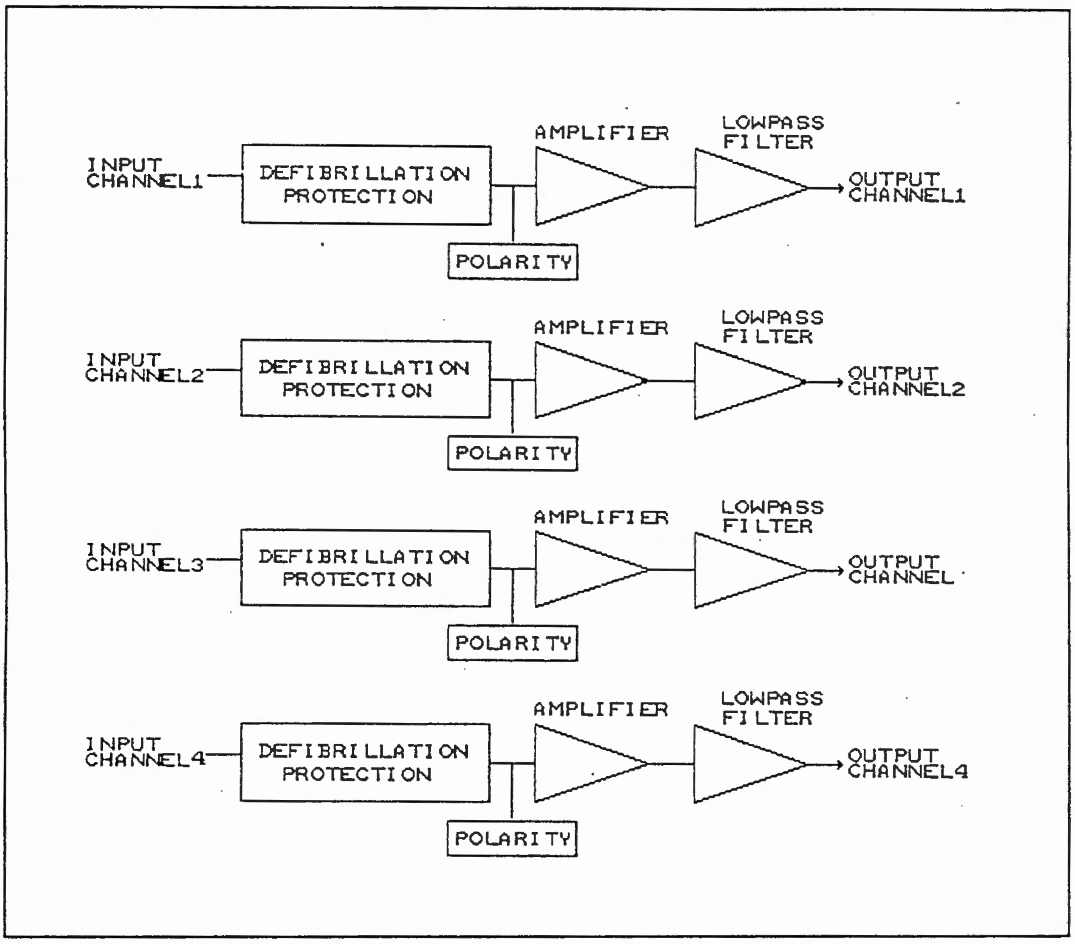
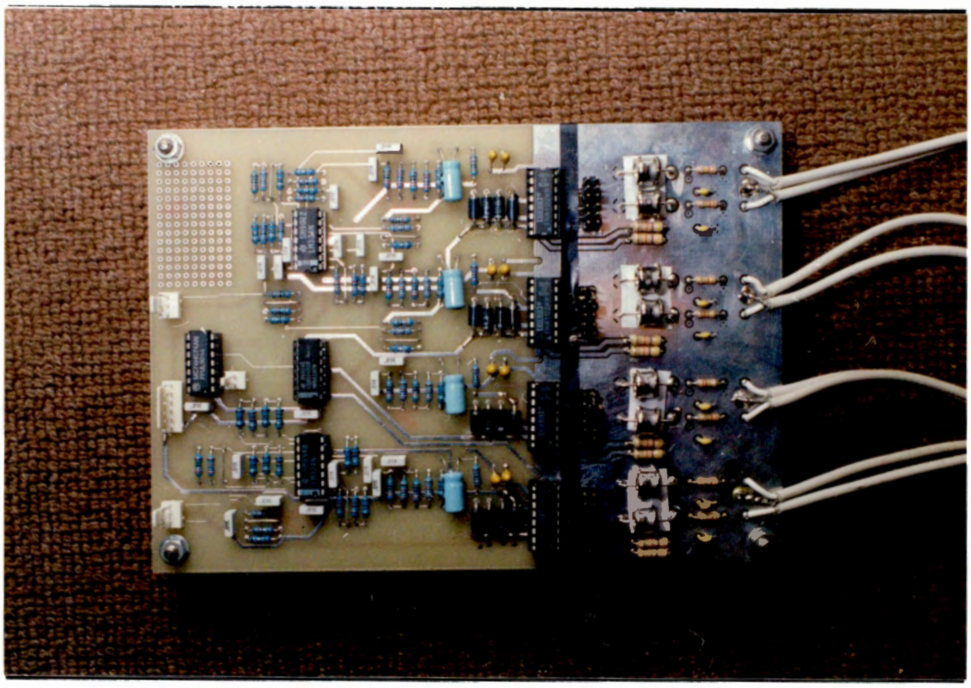


Figure 4.1 **Block Diagram of the Analogue Input Board.**

* Protection of the sensitive instrumentation amplifier against defibrillation pulses and other large input voltages is achieved by a combination of surge arrestors (see Glossary), with a breakdown voltage of 90V, and overvoltage protection diodes. The designed capability of withstanding defibrillation pulse potentials without damage (these are pulsed differential inputs of approximately 6 500 V maximum) has not been tested.
* Filtering consisting of a high-pass, single pole, passive resistor-capacitor filter with a -3dB cutoff of 0.1 Hz (used to remove DC offset from the signal) and a low-pass, four-pole,

Butterworth active filter to minimise aliasing (see Glossary) of components from higher frequency input signals. For a more detailed discussion on noise reduction see Appendix C.

* The software selection allowing a choice of either bipolar or unipolar acquisition modes is implemented on this board.



<u>Figure 4.2</u> **Photograph of the Analogue Input Board.**

A Butterworth filter is used as its reasonably linear phase response ensures that the resultant distortion of the signal is kept to a minimum. The highest frequency of interest during EP studies at GSH is 500 Hz (determined through conversation with Professor Scott Millar, senior cardiologist at GSH, and observations made during EP studies) and the filter therefore has a maximum -3dB cutoff frequency of 500 Hz.

4.3 Analogue Filtering

The analogue filtering board (circuit diagram WPWELEC3 - BOARD 2) comprises microprocessor-programmable universal active filtering for each of eight input channels. This board is therefore designed to accept the output from two analogue input boards. Each filter is configured as a high pass filter stage followed by a low pass filter

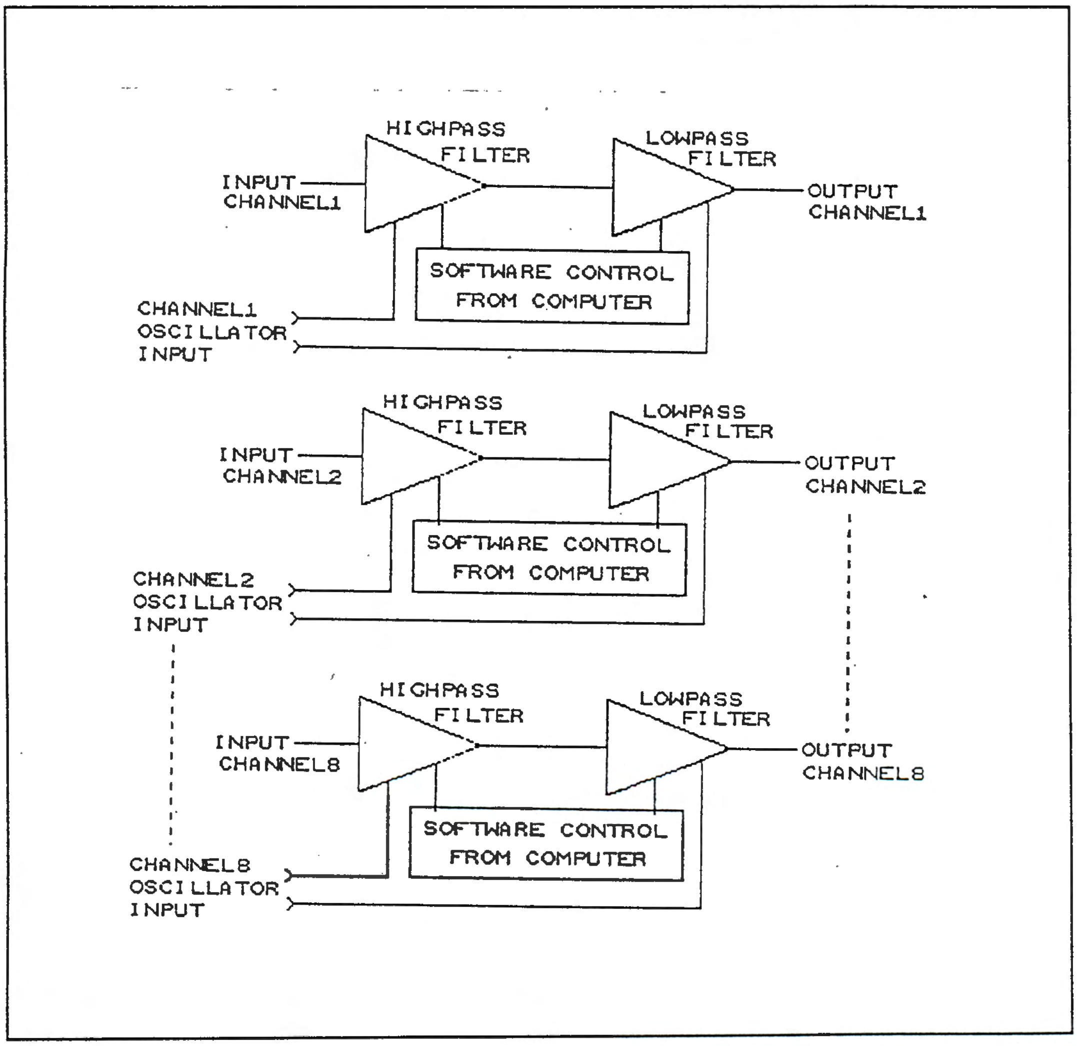


Figure 4.3 **Block Diagram of the Analogue Filtering Board**

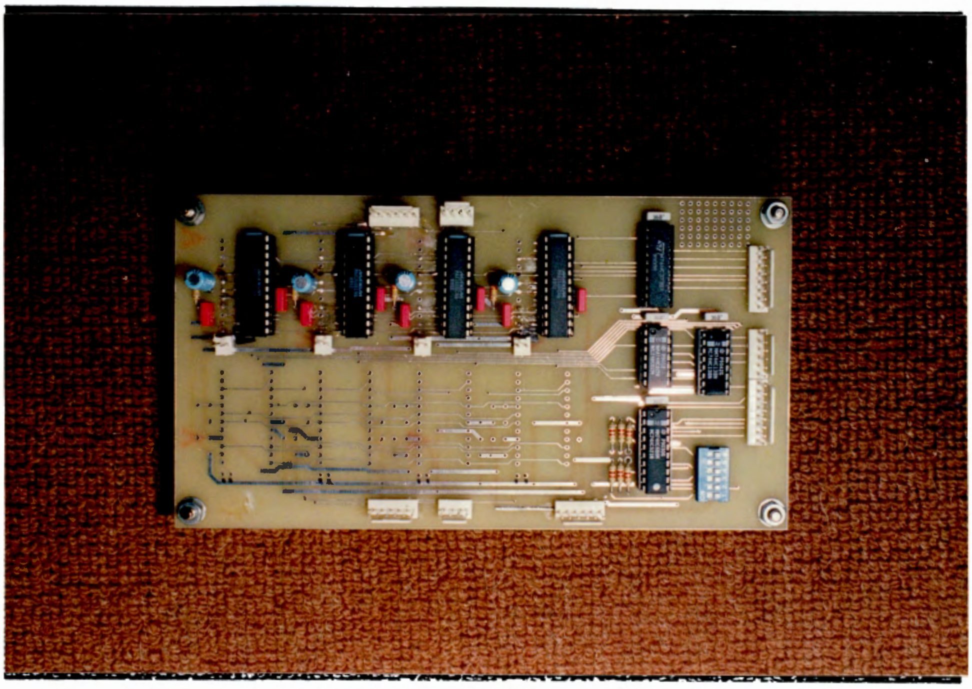


Figure 4.4 **Photograph of the Analogue Filtering Board**

Table 4.1 **- 3dB cutoff frequencies for software selection.**

Highpass Filter (Hz)	Lowpass Filter (Hz)
0.1	100
1	250
30	400
50	500

stage. This combination provides a band pass filter with maximum and minimum -3dB cutoff frequencies at 500 Hz and 0.1 Hz respectively. The specific cutoff frequencies of each channel are controlled by means of an input oscillator frequency fed from the oscillator control board, and software fed from the computer. Table 4.1 shows the cutoff frequency choices available. The necessary address decoding for software control of the filtering is also found on this board.

4.4 Oscillator control

This board (circuit diagram WPWELEC4 - BOARD3) comprises the address decoding and digital logic necessary to provide software control of the input oscillator frequencies that are fed to the filters of the analogue filtering board.

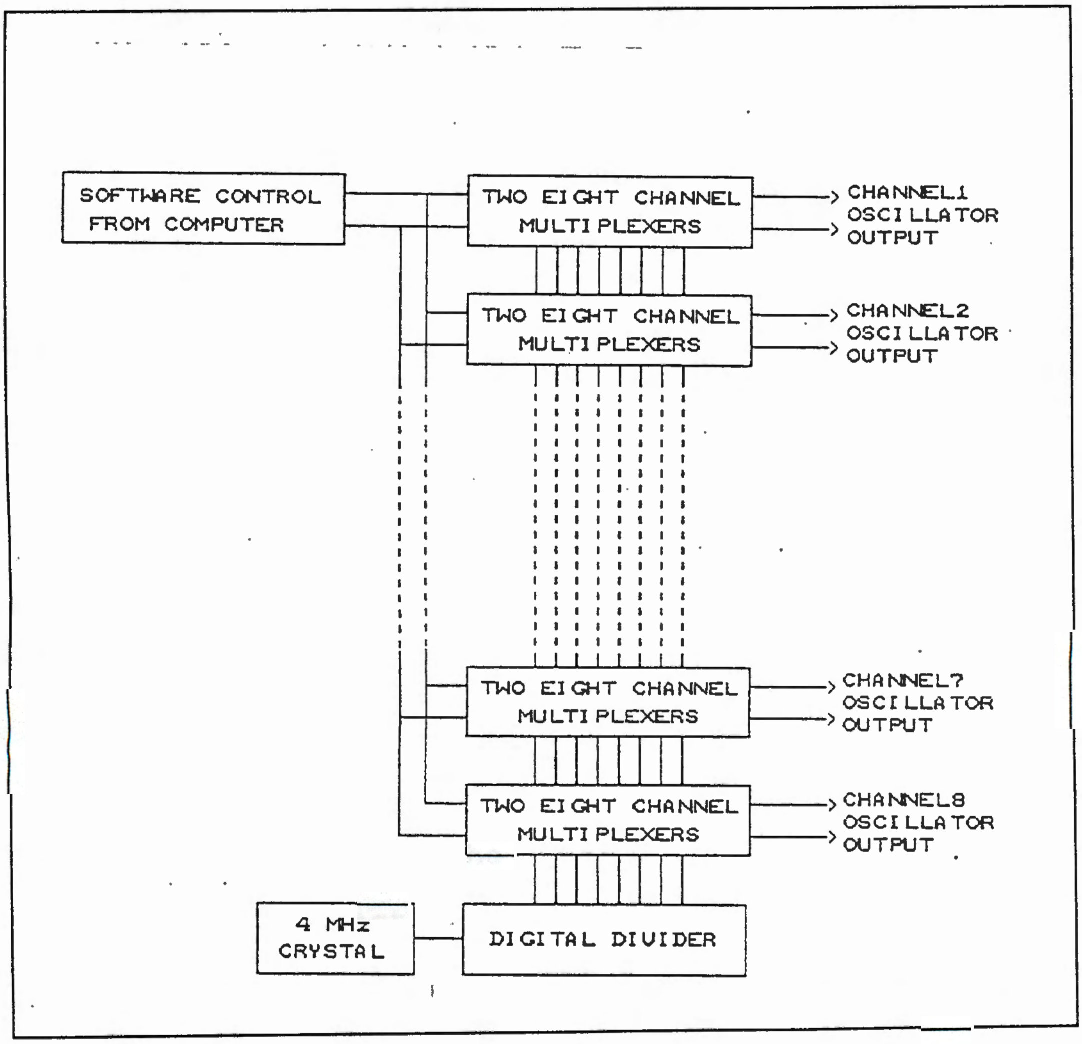


Figure 4.5 **Block Diagram of the Oscillator Control Board**

channels are implemented before the signals are passed to the analogue to digital converter (ADC). This is a twelve bit (plus one sign bit) converter which gives a resolution of 1 in 4096 in the quantization of the signal (with an input of 50 mV to BOARD ONE this gives a resolution of 122 μV).

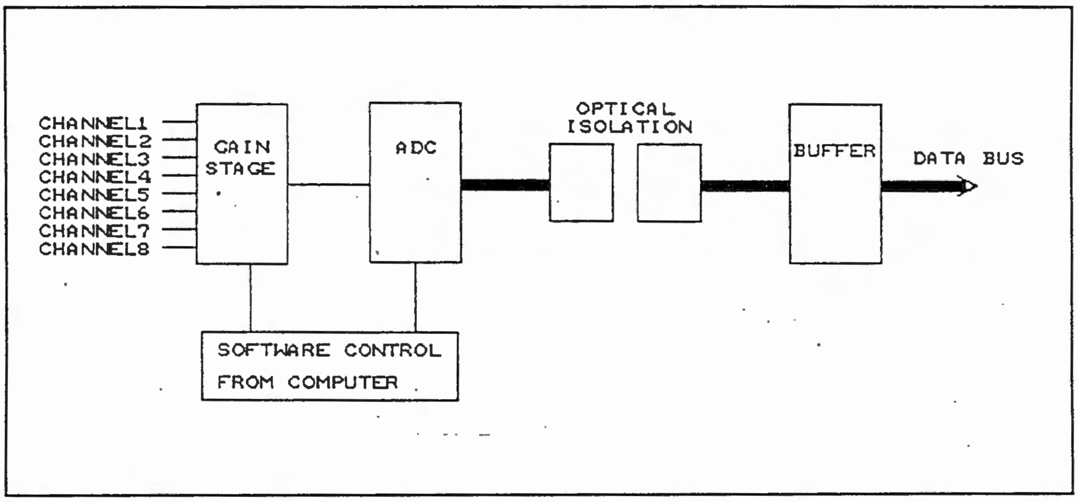


Figure 4.7 **Block Diagram of the Amplification, Digitisation and Isolation Board**

It is necessary that each channel be sampled at a frequency of at least 1 kHz to satisfy the Nyquist criterion (see Glossary), ie sampling at more than twice the frequency of the highest signal component. The ADC samples each channel at a frequency of 2 kHz. This rate was chosen after consideration of the following factors:-

* The greater the sampling rate the greater the accuracy in signal reproduction and activation time detection.
* The lower the sampling frequency the less demand there is on memory space and the speed of the system.
* A 2 kHz signal was readily obtainable from the digital oscillator board.

* Applying the Nyquist criterion an absolute minimum sampling frequency of 1 kHz is necessary for each channel. Due to the slow roll-off (see Glossary) of the Butterworth filter a frequency of greater than 1 kHz is desirable. The sampling frequency for the system was chosen to be 2 kHz and allows a margin of safety needed to ensure minimal aliasing and accuracy in the reproducibility of signals.

The internal circuitry of the ADC needs its own input clock to enable sampling and a frequency of 3 MHz is fed to the ADC from a separate crystal and oscillator circuit.

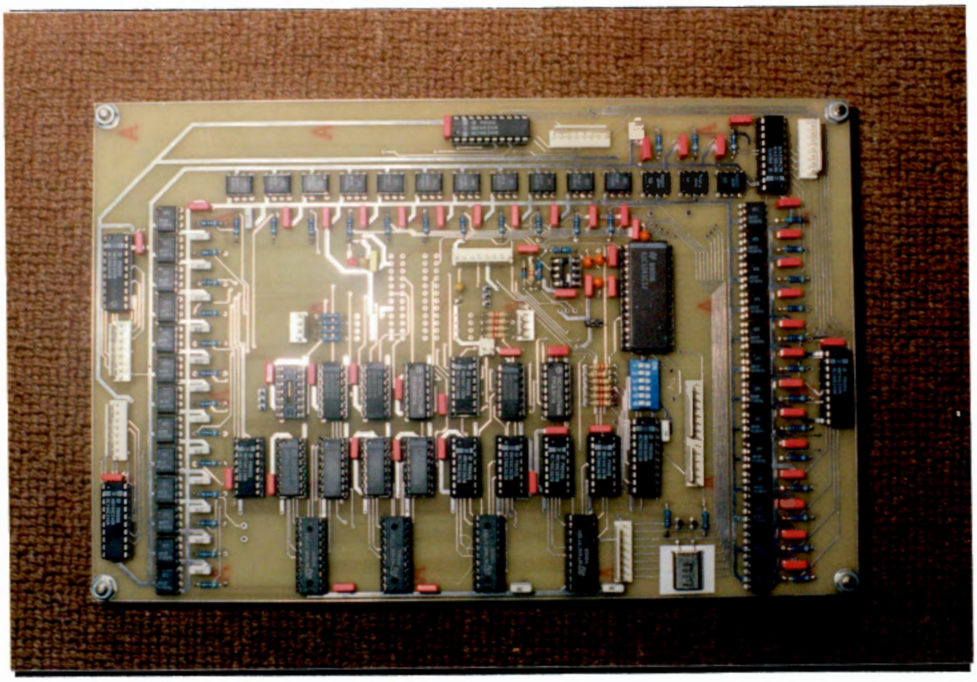


Figure 4.8 **Photograph of the Amplification, Digitisation and Isolation Board**

Protection of the patient from electric shock is critical. It was decided to achieve the high degree of isolation required by means of opto-isolators. Opto-isolators are only linear within a small range of operating voltages and linearity errors result if they are placed in the analogue signal path. Using the isolators in the signal path once it has been digitised circumvents linearity problems because the opto-isolator is switched either off or on. It is this method of digital isolation that has been implemented.

Once the data has been digitised and transmitted across the isolation barrier it is buffered and transferred via ribbon cable to the interface board which resides within the computer.

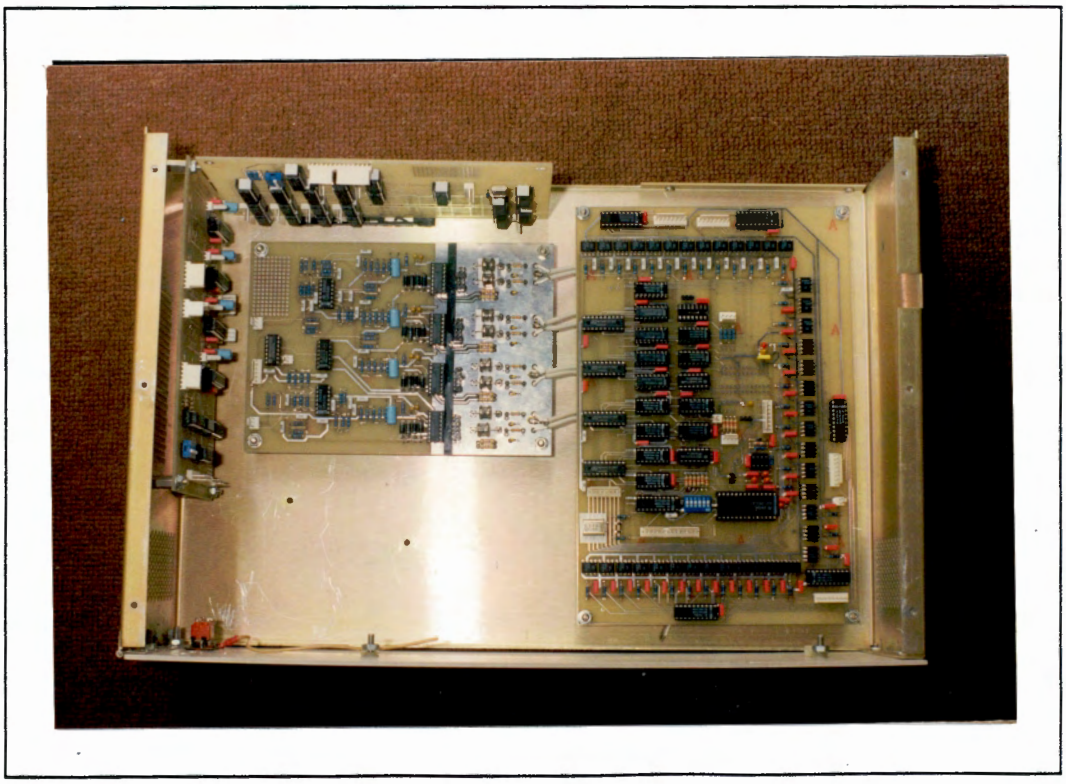


Figure 4.9 **Photograph of the Complete Four Channel Prototype System (excluding the interconnections between boards).**

4.6 Computer interface

The computer interface board (circuit diagram WPWELEC9) is implemented on an IBM prototyping board (Bicc-Vero Electronics, Gmund). The circuitry on this board consists of the necessary buffering, line drivers, logic and control circuitry to interface the system boards to the computer bus.

The direct memory access (DMA) circuitry is needed to realise a high speed of data transfer. Due to the limitation of MS-DOS (Microsoft, Redmond), whereby a maximum of only 64 Kbytes of data can be transferred in one DMA operation, a special technique of swopping DMA channels has been implemented which is referred to as dual DMA.

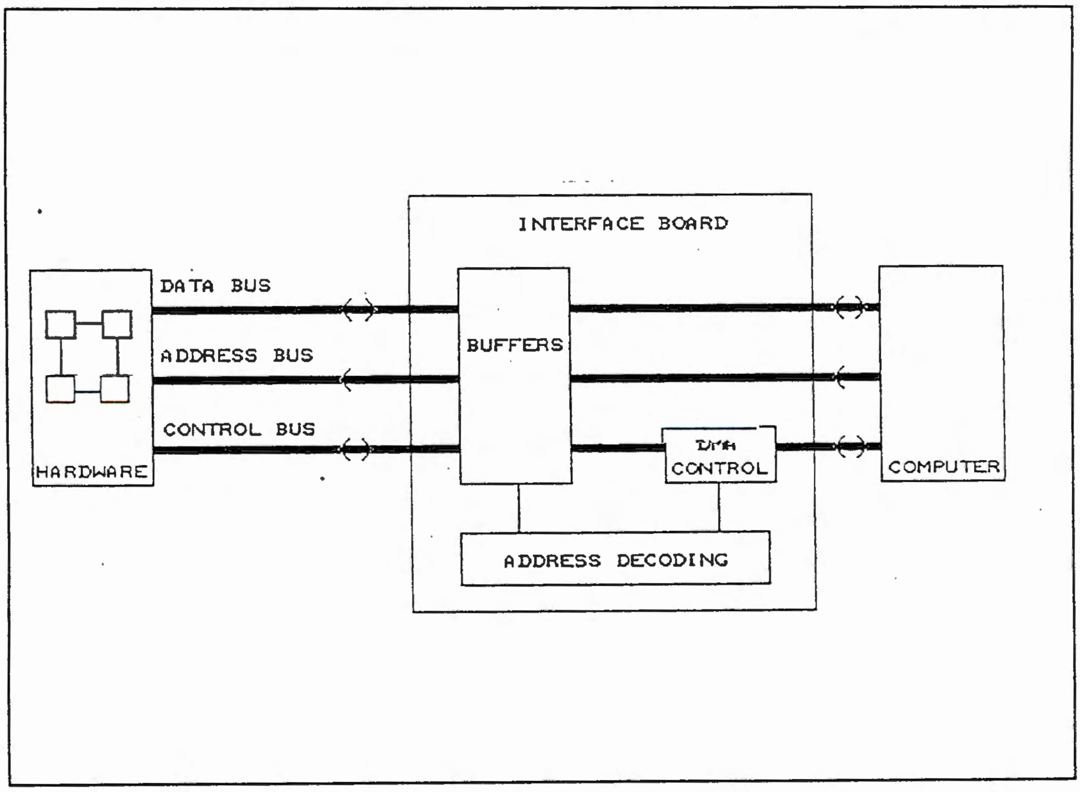


Figure 4.10 Block Diagram of the Computer Interface Board

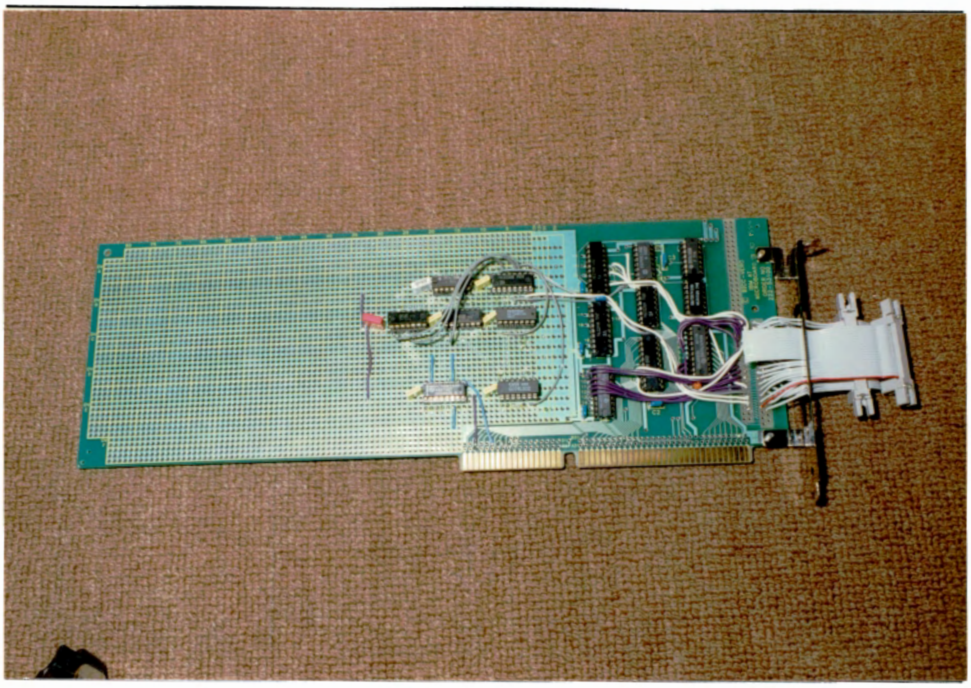


Figure 4.11 **Photograph of the Computer Interface Board**

4.7 Ergonomic Considerations

The hardware of the acquisition system has been designed in such a manner that its presence is as transparent as possible to the operator. Features such as software controlled gain and filtering have been implemented to achieve this goal.

The acquisition and processing of data is a major component of system function and a high speed (16 MHz) 386SX computer processor was considered necessary to ensure that frustration of the operator does not arise due to a slow-functioning system. With micro-technology changing rapidly it may well benefit the users to upgrade the system.

The display of the electrograms is of great importance and for this reason a high resolution, colour monitor is mandatory. The monitor used is a 14 inch multiscan monitor controlled by a 16 bit VGA colour card (giving a resolution of 640 x 480). This resolution is considered adequate but a larger monitor (20 inch) would effectively increase the resolution and facilitate the reading of data from the screen.

CHAPTER FIVE

PATIENT DATABASE

5.1 Introduction

The creation of a patient database was both necessary and useful for the development of the acquisition system. The experience gained while in the EP studies laboratory and the cardio-thoracic theatres and the practical problems encountered during this time helped a great deal with the final implementation of both hardware and software.

Electrogram data was recorded intraoperatively on magnetic tape during WPW mapping procedures as well as during routine EP studies in the Cardiac Clinic at Groote Schuur Hospital during 1990 and 1991. Representative segments from six of these recording sessions were then digitised and stored for use in the database.

5.2 Purpose of the Database.

The need to become familiar with the conditions under which the system will be used as well as the need to test the system software necessitated the creation of a patient waveform database. The system will be used in both EP studies as well as intraoperatively in WPW syndrome mapping and it is therefore necessary for the database to contain recorded signals from both of these mapping situations in order to ensure the system's ability to fulfil the stated requirements.

The software testing was necessary to establish a practical, pleasing and informative method of displaying the signals on the computer screen. Detection of the activation times and the overall ease of use of the software are important design criteria that further necessitated the formation of a database.

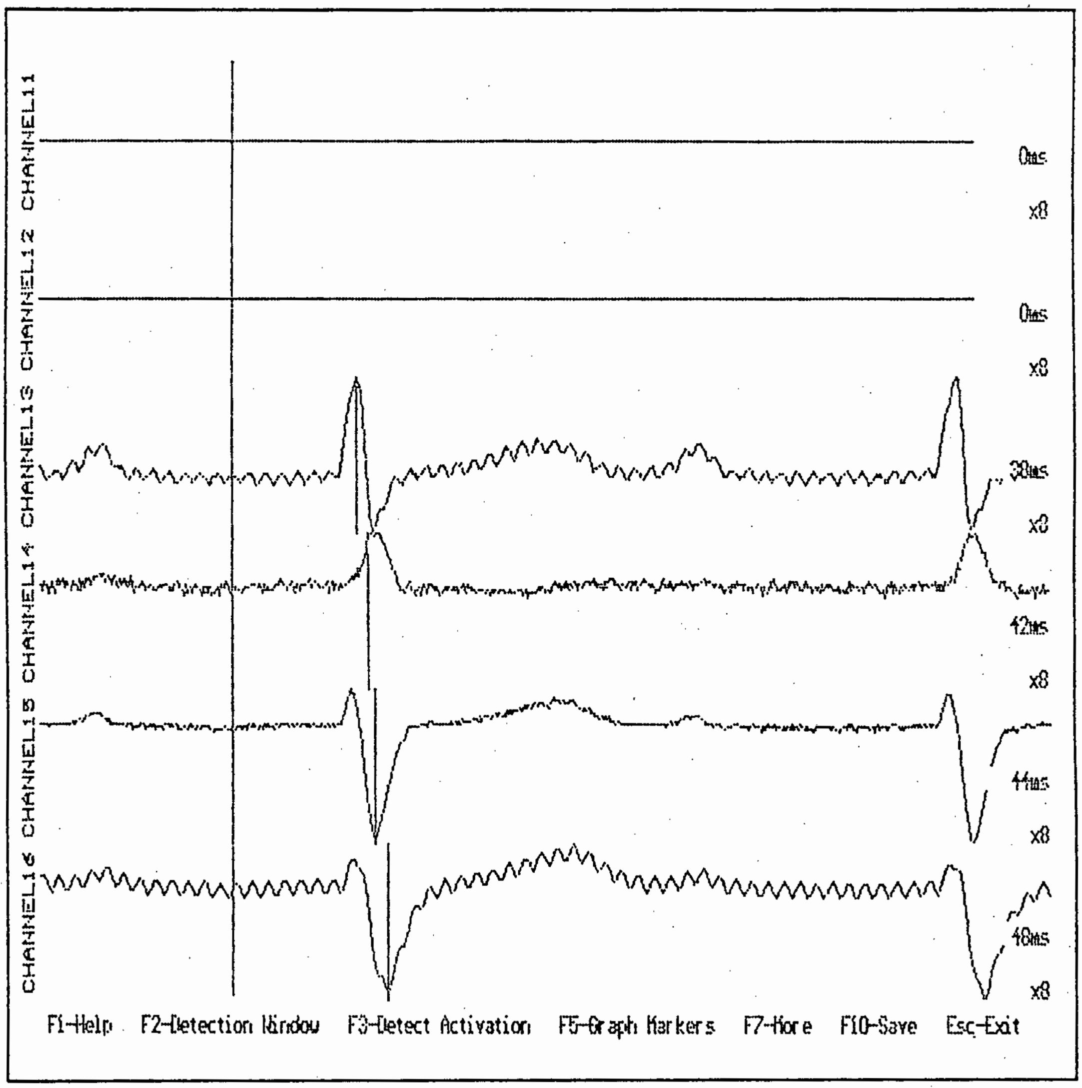
5.3 Recording method.

Mapping procedures in the Cardiac Clinic make use of the VR12 (Electronics for Medicine, Pleasantville) data acquisition system for the real-time display and processing of electrogram and surface ECG signals. A Mingograf (Siemens, Erlangen) is attached directly to the VR12 to provide the necessary hardcopy of the waveforms. The input channels of the VR12 are connected, after filtering and amplification, to a socket on the rear panel that is used for interfacing to a computer. Four of these channel outputs were connected to a four channel, magnetic tape, FM data recorder (R-61, TEAC, Tokyo) with a bandwidth of dc - 625 Hz.

The TEAC was used to acquire patient waveforms at various times during each procedure. The majority of electrogram signals obtained were from within the heart during catheterisation in EP studies and included His bundle, high right atrial, coronary sinus, atrial and ventricular electrograms. A listing of the patients and the data recorded is given in Appendix B.

5.4 Conversion and reviewing of data

The recorded signals were transferred to computer by means of the PC30D data acquisition board (Eagle Electric, Cape Town) and the program Status-30 (v2.02) supplied with the board. This data was then stored onto diskette to form the database. Due to the amount of recorded data and the large size of the digitized data files only three to six recordings of approximately four seconds each were digitized from each patient recording.



<u>Figure 5.1</u> Screen Copy of Waveforms from a WPW Syndrome EP Study.

The unit WPWDREAD was written in order to review the database information from diskette. Once working correctly this program was converted into a Turbo Pascal procedure and used in the main mapping program of the acquisition system. It must be remembered that the database comprises only four channels of data and for this reason only four (channels nine to twelve) of the display are active.

Figure 5.1 is a hardcopy printout of the computer screen showing the four recorded signals of a patient undergoing EP studies for WPW syndrome mapping. The reference marker was placed in position on channel 14 and the activation time markers were derived using the maximum amplitude detection algorithm. The reader is referred to Appendix B for further details regarding the database.

CHAPTER SIX

SOFTWARE DEVELOPMENT

6.1 Introduction

The software is the major component of the interface between the operating cardiologist and the acquisition system thus facilitating the necessary signal processing and control of the data recording.

The software is written in the high-level language Turbo Pascal (version 5, Borland). The main program is called WWEPSMAP.EXE and has direct control of the menu options presented to the user. As each menu option is selected by the operator, so the different units are called to perform the required functions.

The following units have been written and are responsible for controlling the tasks indicated:-

* WWEPSIO .TPU - controls the input and output functions associated with patient data manipulation and maintenance of the patient records on disk
* WWEPSBRD.TPU - configures the acquisition boards according to the user's selection
* WWEPSDMA.TPU - controls the DMA associated with acquiring the data to memory and saving it to disk
* WWEPSPLT.TPU - reads the acquired (or database) data from disk and displays it on the screen. Manipulation of the traces and activation detection are controlled by this unit.

6.2 The user interface

Figure 6.1 is a representation of the flow of the program WWEPSMAP.EXE and its associated units. When the program is run from DOS it initialises all variables and then proceeds to configure the acquisition boards to the default mode. The settings for this default mode are shown in Figure 6.2.

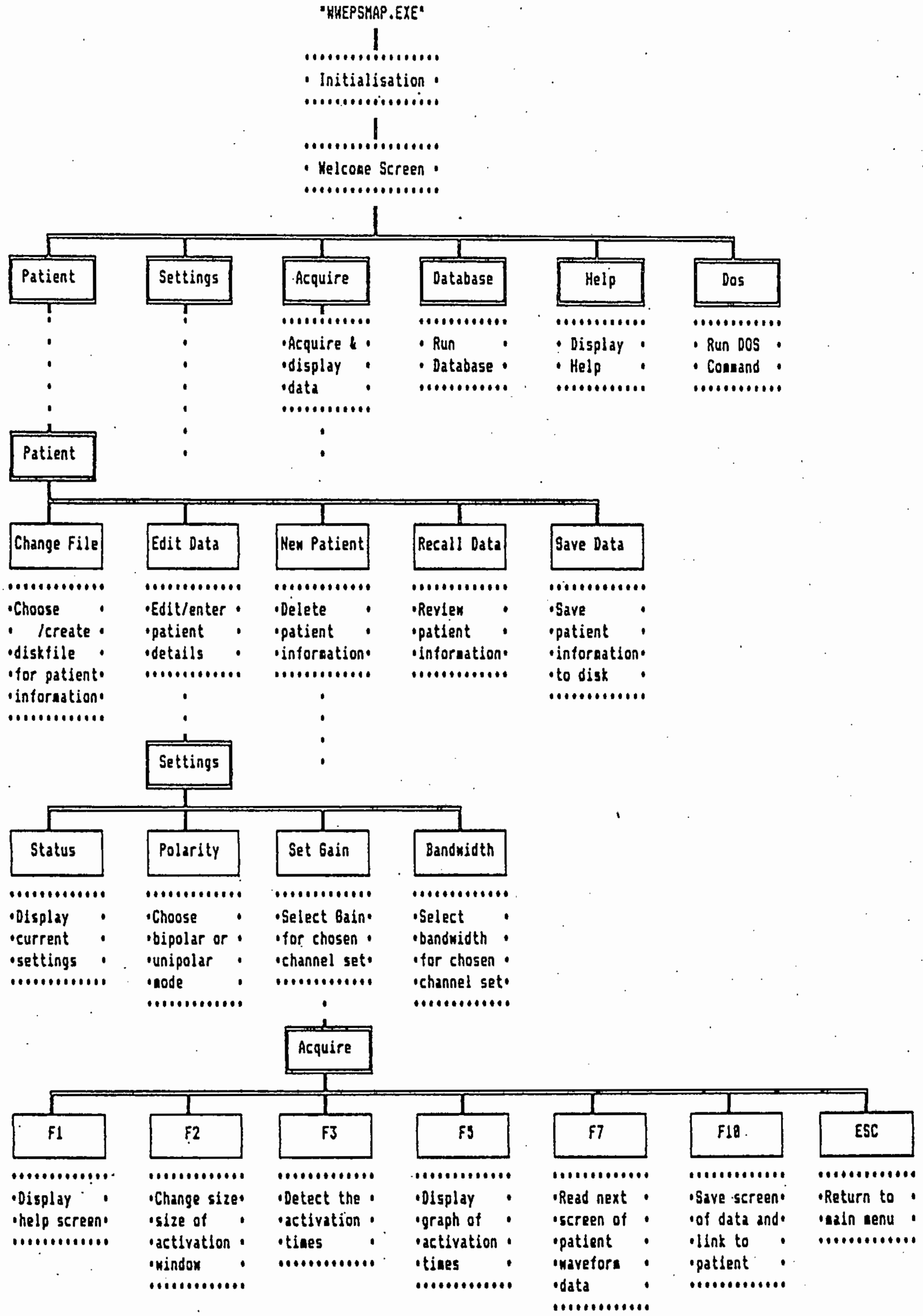


Figure 6.1 Flow of Main Software Program

Figure 6.2 The STATUS Option showing Default Board Configuration.

Figure 6.3 The Menu Choices of WWEPSMAP.EXE

The "Welcome" screen briefly introduces WWEPSMAP.EXE after which the main menu choices are displayed to the user (Figure 6.3). The purpose of each menu option is indicated in the flowchart representation of Figure 6.1.

The general sequence followed by the operator is:-

* Enter the patient information into the data entry windows and save to disk.
* Select the desired gain for each of the four channel sets. The 32 channels of the proposed system have been divided into four channel sets of eight channels each, viz 1-8, 9-16, 17-24 and 25-32. Whenever a setting is chosen it affects all the channels of that particular channel set.
* Select the desired highpass and lowpass filter settings for each of the channel sets.
* Select either bipolar or unipolar acquisition mode.
* Check that the acquisition boards have been configured correctly using the "Status" command.
* "Acquire" the data.
* Manipulate the displayed traces as required (for WPW syndrome mapping the automatic activation detection algorithms can be selected and time intervals measured).

In addition to the above there are a number of other options available to the operator:-

* Help is context sensitive and available whenever 'F1 - Help' is shown on the bottom line of the screen
* The user may select or create new diskfile names into which the

patient data records are saved

* If the patient has been mapped previously then the patient details may be reviewed and selected using the "Recall Data" command. The "Database" command will then automatically recall the mapping data previously saved under that patient name
* If the system memory allows, any DOS function is available without exiting the program by means of the "DOS" command.

6.3 Data acquisition

Once the "Acquire" command has been given the computer begins a series of instructions in order to acquire the input data. Firstly the hardware DMA channels five and six are programmed to accept thirteen bits of data (twelve data and one sign bit) from the acquisition boards and transfer this to computer memory. Secondly the computer is instructed to save this incoming data to disk and finally, the computer resets the channel counter to 'Channel one' and awaits a 'data available' request from the acquisition boards.

During the period of DMA acquisition there is very little time for extra computation. The data is received sequentially from all 32 channels at a rate of 64kHz and each DMA controller is programmed in the 'demand mode' (transfer of data only occurs when requested by the acquisition boards). This requires 16 bits of data (the width of the memory data bus) to be transferred to memory every 0.5 microseconds. Saving the data to disk immediately after acquisition is necessitated by the memory limitations (640 Kbytes RAM) of the present system because the amount of acquired data (4 seconds x 32 channels x 2 000 samples/second x 16 bits = 512 Kbytes) is too large to reside in memory along with the controlling software program (133 Kbytes for the program and 19 Kbytes for the heap).

It was decided to acquire four seconds of data in order to allow a long enough period for important data to be acquired yet keep the amount of stored data to a minimum. Once these four seconds of data have been sampled and stored to disk the first 600 bytes of data for each channel is read into memory (the resolution of the VGA screen is 640 pixels across - with 40 pixels reserved for text this leaves 600 pixels for the display of waveforms). The first eight channels are then displayed on the screen and the computer awaits input from the operator. The operator may display any eight sequential channels on the screen using the scrolling function of the 'up', 'down', 'Page Up' and 'Page Dn' arrow keys.

Each screen of data displays approximately one second of eight of the acquired waveforms. This data and the corresponding data of the other 24 channels can be saved on disk and automatically linked to the current patient record by the "F10" option. The database option in the main menu section is used to recall these previously saved screens.

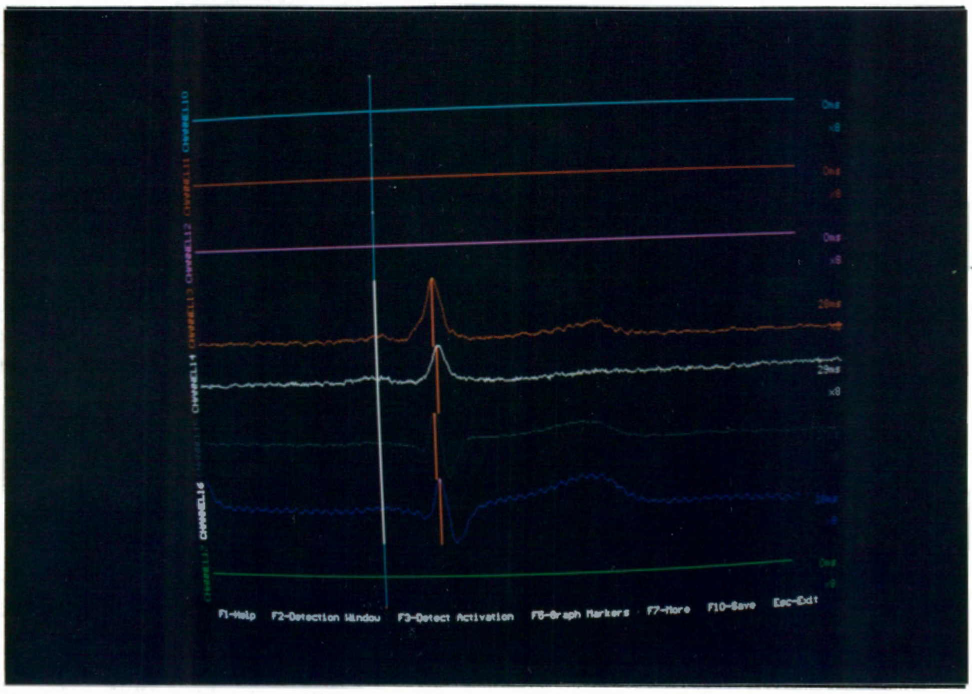
6.4 WPW syndrome specific software

The specifications require that the acquisition system include specific routines that allow automatic detection of activation times. The manner in which this has been implemented is now explained.

Once the acquisition of data is complete and the first eight channels of data have been displayed on the screen, the automatic activation time detection routine may be chosen. Activation time detection only occurs on channels 9 through to 32. It is assumed that these channels (or a subset of them) will be linked to the mapping electrode array. Channels 1 to 8 are for external leads and other signals.

An activation complex (heart beat) is selected by moving the reference marker to any point within this complex (on any one of the channels). The purpose of having a global reference marker is to ensure that the cardiologist is not limited whilst choosing a reference channel on which to base the measurements. Pressing the "F3" function key causes the computer to apply the activation detection algorithm to each channel. On completion red markers are placed at points of maximum amplitude or slope (see below). The activation detection is performed within a specified time window centred on the reference marker position. The default width of this window is 300 milliseconds but can be changed using the "F2" function.

The markers may be moved manually if the cardiologist disagrees with the indicated activation times. Movement of the markers is by means of the arrow keys (for left and right movement) and the space bar (to swop arrow key control between the reference marker and activation time detection markers). On the right of the screen the absolute time difference between the reference marker and the activation markers is shown for each of the displayed channels. The function key "F5" provides a graphical representation of the activation times of channels 9-32. The option to highlight a certain channel gives the cardiologist the ability to indicate the position of the crux of the heart in relation to the electrode band. Highlighting of a particular channel is achieved using the left and right arrow keys.



<u>Figure 6.4</u> Photograph showing Acquired Waveforms and Activation Time Detection Markers.

Two algorithms, maximum amplitude or maximum slope criteria, are available for the automatic detection of activation. Either one of these two algorithms can be chosen using the "F4" function key. The maximum amplitude algorithm simply establishes the position of the absolute maximum amplitude of the signal within the search window and places the detection marker at this point. The maximum slope algorithm calculates the first derivative at each point of the signal within the search window and places the activation marker at the position of the greatest absolute derivative. The reader is referred to section 2.3.2 for more details of these algoritms.

6.5 Ergonomic Considerations

The ergonomics of a system are important in the overall usability and efficiency of the system. The following factors highlight features of the software that implement such ergonomic considerations.

* In order to simplify the configuring and use of this system the number of commands available to the operator is minimised.
* Most choices that need to be made during the running of the software are selected by means of the available menus or function keys.
* Context sensitive help is available for operators unfamiliar with the system and the bottom line of the screen displays what is required of the operator.

Other than the actual time during which the acquisition of data is performed the computer responds almost instantaneously to the operator's commands by means of information windows or visual confirmation of each command.

To facilitate the entry of data during the creation of each patient record the standard word-processing commands (Tab, delete, insert, enter etc) are available. Obvious illegal entries (such as entering digits for a patient name or letters for a patient's birthdate) are ignored, but stringent checks have been omitted during data input in order to promote smooth data entry. The data entered may be modified and corrected at any time by the operator but must be saved to disk following any such modification.

The use of colour to enhance the different channel waveforms on the screen helps to clearly distinguish each waveform from the others and is useful while examining the waveforms.

CHAPTER SEVEN

SYSTEM EVALUATION AND DISCUSSION

7.1 Hardware Evaluation

Evaluation of the hardware was considered necessary in order to check for design errors and to ensure that the hardware performed according to specification.

The evaluation of the analogue circuitry (on Board One) was performed using a sine-wave input (Hewlett Packard Function Generator Model 3314A) and comparing input and output signals on an oscilloscope (Kikusui Model 7101A). The gain response of all four channels (Figure 7.1) shows the close matching of the gain between the channels. In this figure the gain and frequency axes are plotted on a logarithmic scale and the gain response has been normalised by dividing the gain by the mid-band gain value (19 dB for a gain setting of 10).

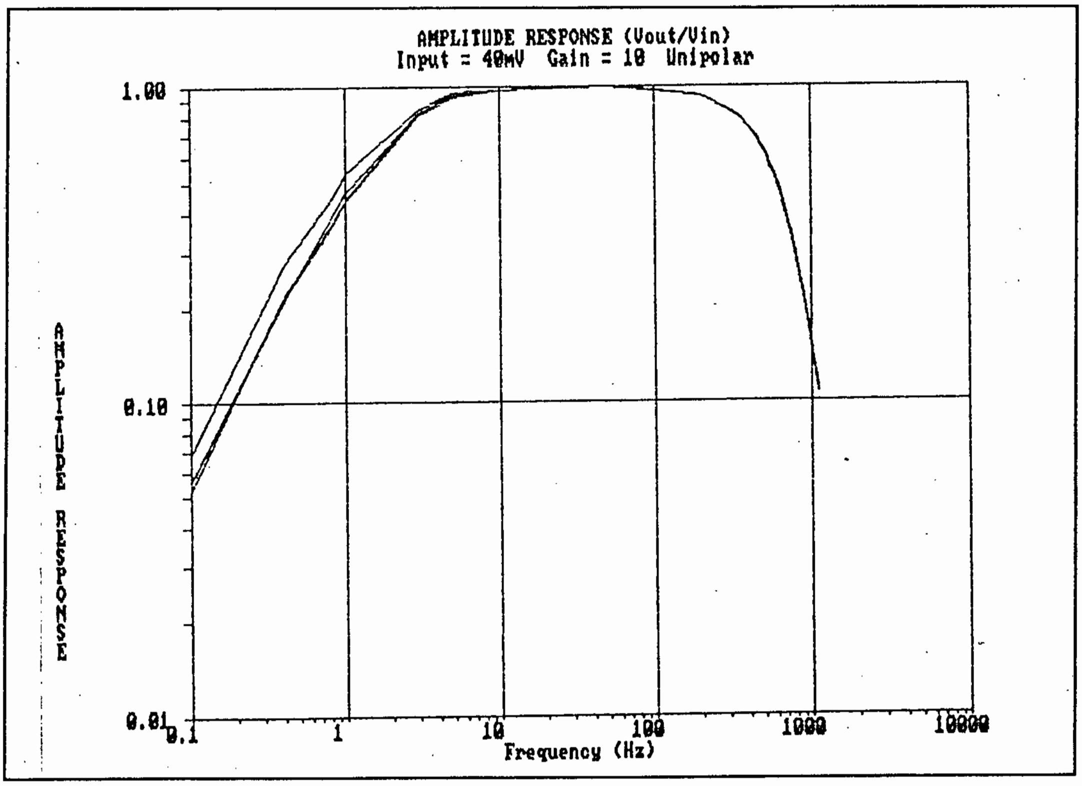


Figure 7.1 **Gain Response of Channels One to Four**

The effect of using a passive filter for the low frequency cutoff can be seen in the slow rolloff of the response at the lower frequencies, which results in the -3 dB cutoff being close to 2 Hz. This should not adversely affect the system's performance because low frequency signals (such as ST displacements) are not a concern in EP studies. The faster rolloff at high frequencies (corresponding to the 4 pole active filter used at these frequencies) results in the -3 dB highpass cutoff at the desired anti-aliasing value of 500 Hz.

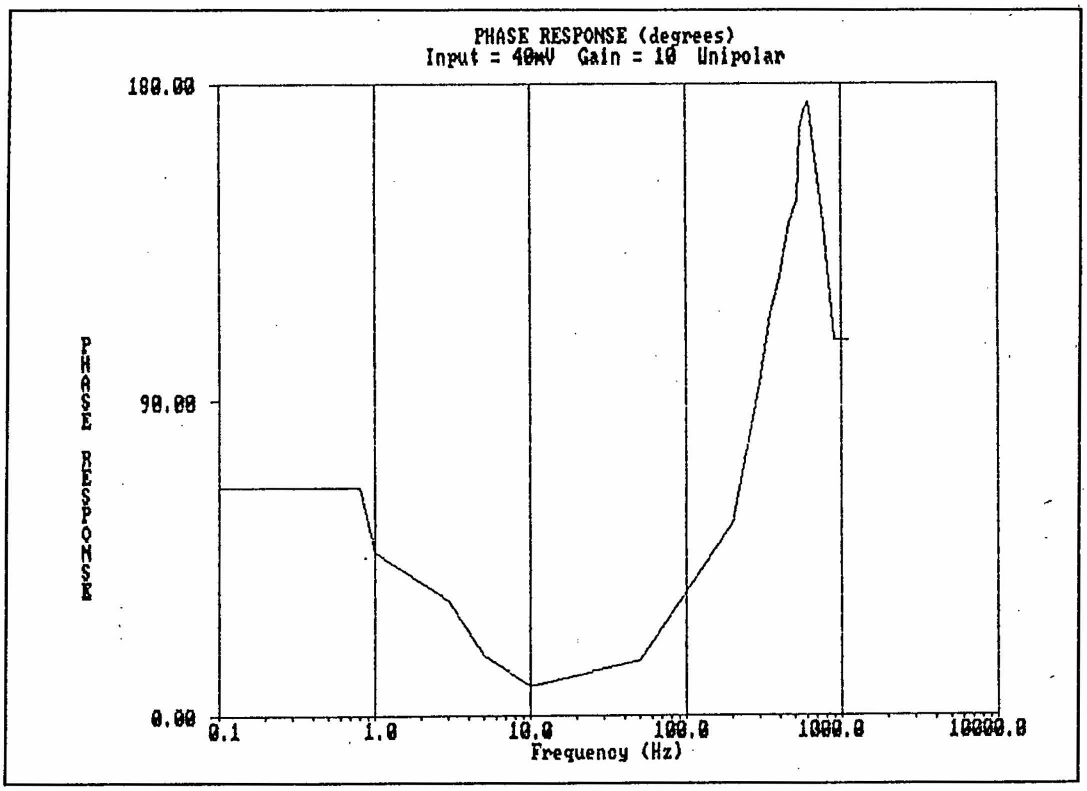


Figure 7.2 **Phase Response of Channel One** (circuitry on Board One)

Figure 7.2 shows the phase response of channel one. The phase response is displayed linearly on the vertical axis in degrees. Within the passband from 1 to 500 Hz (measured from figure 7.1) the phase changes from 47° to 144°. The phase response then increases above 500 Hz to almost 180° at 600 Hz. This increase falls outside the pass band and will therefore have only a limited effect on the signal. The maximum deviation from linearity (within the passband) is approximately 90 degrees. This variation in phase within the passband is larger than the design sought to achieve and further tests (in the form of comparisons between existing and newly acquired waveforms) will need to be performed to ensure that unacceptable distortion of the signal does not occur and that the activation time detection algorithms are not adversely affected.

The four digital boards form a functional unit and need to be tested in combination. The testing of these boards was performed by means of a logic analyzer (Hewlett Packard Model 1630D) and oscilloscope (Kikusui Model 7101A). The addresses on the various boards were verified for correct operation using a short test program. The functioning of the data transfer from the A/D converter was tested using a sinewave input to all four channels of 20 mV at 500 Hz. A test program very similar to the unit WWEPSDMA was used to initialise the computer's DMA channels and the memory contents were displayed on screen while DMA transfer occurred.

The original design did not function as expected and a number of changes had to be made :

i. Resistors of 330Ω were placed on the inputs to each of the optoisolators, thereby preventing excessive loading of the optoisolators.
ii. Address line A0 was left out of the addressing to the boards as it is used in sixteen bit computers to decode eight or sixteen bit bus

transfers. The boards are therefore addressed using lines A1-A9. The address range is 220-25F (Hex).

iii. The interface circuitry of the IBM interface board was used but modified slightly to allow for the change in (ii) above. The "enable" decoding was also changed to allow the board to write to the computer memory during the DMA process.

iv. The /IOW command from the computer was cascaded using two NAND gates to increase the access time that was required to overcome the delay due to capacitance effects in the connecting ribbon cable.

v. The data buffer for data lines D0-D7 was tristated (high-impedance inactive state) using the CDTIN line to prevent the bus contention that occurred.

vi. The voltage reference of the analogue to digital converter was connected to the 5 volt supply rail as the circuit initially implemented did not function correctly.

vii. A larger capacity power supply that could provide the 0.75 Amperes drawn by the circuitry was designed and implemented.

viii. The original logic used to read data from the A/D converter prevented synchronous timing with the computer and therefore the read line of the ADC1241 was permanently enabled.

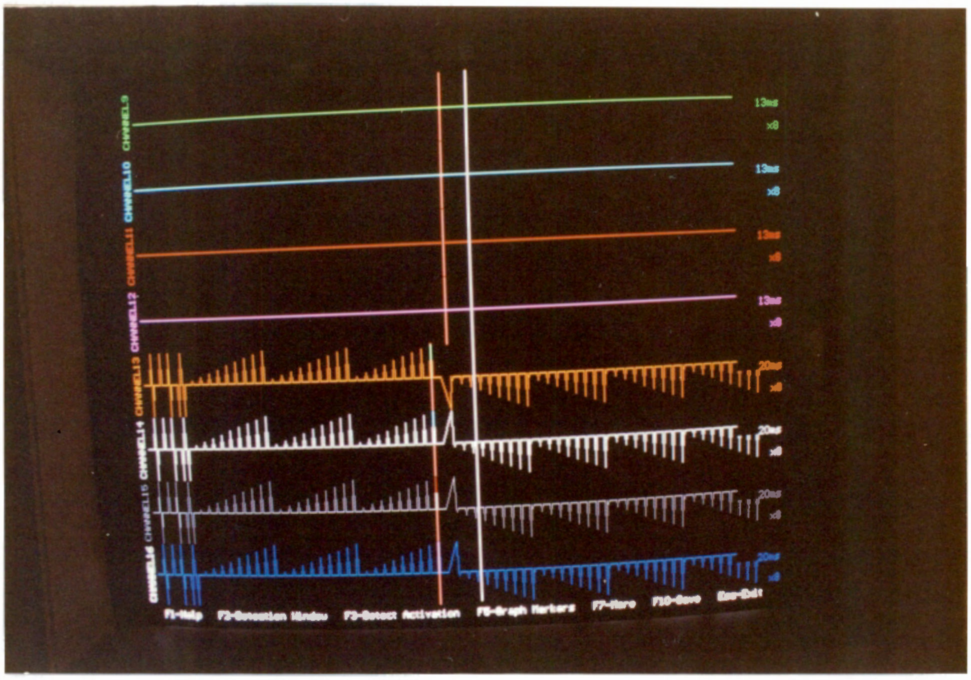
ix. The PGA100 programmable gain amplifier could not be obtained locally and it was decided to use a simple eight channel multiplexer as a replacement component for the purposes of the prototype system and implement the necessary amplification in software.

For the purpose of demonstration and testing the hardware is powered by a conventional split-rail power supply. This is clearly unsuitable for the final system where isolation is obligatory. The use of batteries to power the system hardware provides isolation of the patient from possible electrical shocks as well as providing a relatively constant voltage level. Battery operation on the other hand may not be practical in the clinical situation where operation times may be lengthy (battery operation times are limited) and trouble-free equipment is a necessity. Although a battery charger and charge level indicator could be added, operating time is still limited by the battery capacity. Using an isolated power supply may be more expensive but the added convenience may well make it more appropriate for implementation in this system.

7.2 Software Evaluation

In order to check the accuracy of the timing and activation detection algorithms used within WWEPSMAP.EXE an artificial record was created using a text editor and added to the database as a test pattern. Within this test record waveforms with closely matching slopes and amplitudes were placed alongside each other. The software was then run using this test record and the timing and detection algorithms were evaluated while the resolution and clarity of the display were noted. Figure 7.2 shows the test waveform used. The display and activation detection algorithms were found to perform correctly and no errors in detection were found. The timing is accurate to 0.5 milliseconds and this resolution is more than adequate for application in EP studies.

The ease of use and reliability of any software is best evaluated when used in the actual working situation. One indicator of the ergonomic performance or "user-friendliness" of the software is to ascertain how quickly an operator can learn the keystrokes and commands needed to quickly an operator can learn the keystrokes and commands needed to efficiently operate the program. In the case of activation time detection the number of times the cardiologist disagrees with the chosen activation times, and needs to correct them manually, is a good indicator of accuracy and reliability. Professor Rob Scott Millar, a senior cardiologist in the Cardiac Clinic at GSH, was asked to use the computer program and evaluate it in the context of an EP study. The software was evaluated without the hardware attached and for this reason the database files were used as the input.



<u>Figure 7.3</u> **Display of the Test Waveforms Used in the Evaluation of the Detection Algorithms**

Professor Scott Millar found the software to be user-friendly and to meet the specifications of the system. The program was simple to use and familiarisation with the commands and keystrokes occurred very quickly. The quality and clarity of the waveforms displayed is high and of adequate resolution to allow for manual placement of activation detection markers. The activation detection algorithms were consistent and accurate in the detection of the maximum amplitude and maximum slope of a waveform.

Future improvements suggested by Professor Scott Millar consisted of the following alterations:-

* Rather than manual positioning of a reference marker, it was suggested that the user be able to select a "reference channel". On selection of this channel the reference marker would then be placed on a computer-detected feature of that particular channel. The detection algorithms used for positioning of the reference marker would be the same as those used for the activation time detection on other channels.
* The detection algorithm should not detect voltages below a certain amplitude threshold. In the present case where there is no threshold, a low amplitude noise spike may interfere with the maximum slope detection algorithm resulting in incorrect activation time detection.
* A line graph of activation times versus electrode site number may well be easier to interpret than the bar graph display presently used.
* The ability to exclude channels from the activation time graph is necessary.
* Considering future uses of the system it would be necessary to detect pacemaker spikes (pulse widths down to 1ms).

* Increasing the size of the monitor to 17 or 20 inches will enhance the clarity of the display and allow for viewing from a greater distance (necessary in the clinical environment).

7.3 Financial Considerations

The circuit board layouts were professionally designed by a private engineering concern (Esanru, Durbanville). The layout design and manufacturing costs for the boards are shown in Table 7.1. The indicated costs of components are approximate. Figures in brackets represent setup costs which had to be paid for each board that was made.

Table 7.1 Cost of Printed Circuit Board Manufacture

Board	Layout R	Manufacture R	Components R
Board 1	300	(130) 136	200
Board 2	400	(130) 121	100
Board 3	400	(130) 123	100
Board 4	400	(130) 300	500
Interface Board	-	410	50

The above figures represent the hardware costs for the prototype system of four working channels. The costs per board will be reduced when more of the same type are made, for example when the other twenty eight channels of the system are added.

Approximately 14 man-months were spent on the total system development, with one-third of this time spent on on hardware development and two-thirds on software.

7.4 Discussion

After initial bench testing, the system hardware was modified as outlined in section 7.1. Likewise, the user-interface and software aspects were evaluated and suggestions were made for improvements in appearance and function (Section 7.2). The system was not implemented or tested in the clinical situation. Although the original aim had been to provide a four-channel system (upgradeable to 32 channels) for clinical use, this requirement was not met. During the course of the development, the system requirements had changed to include a greater number of channels (minimum of 40) which could be expanded to double or even triple this number for higher-resolution tachycardia mapping. This increased capability necessitated a new design approach which went beyond the requirements of this thesis. In addition, the time needed for system development had been underestimated. It was therefore decided to use the system development to demonstrate the feasibility of such a system concept, to identify design constraints and to provide a platform for further development.

For purposes of bench testing, a non-isolated power supply for the front-end circuitry (up to the opto-isolator stage) was constructed and used. This would have to be replaced by an isolated power supply complying with both circuit requirements (+12V, -12V, +5V and -5V) and safety standards relating to leakage current and isolation breakdown voltage.

As stated previously the system had been specified and designed to allow upgrading to 32 channels. Expansion to the full complement of channels is achieved by interconnecting multiples of the existing boards together. Figure 7.4 is a layout of the complete system displaying schematically how the boards are connected together. Obvious connections such as power lines are not shown.

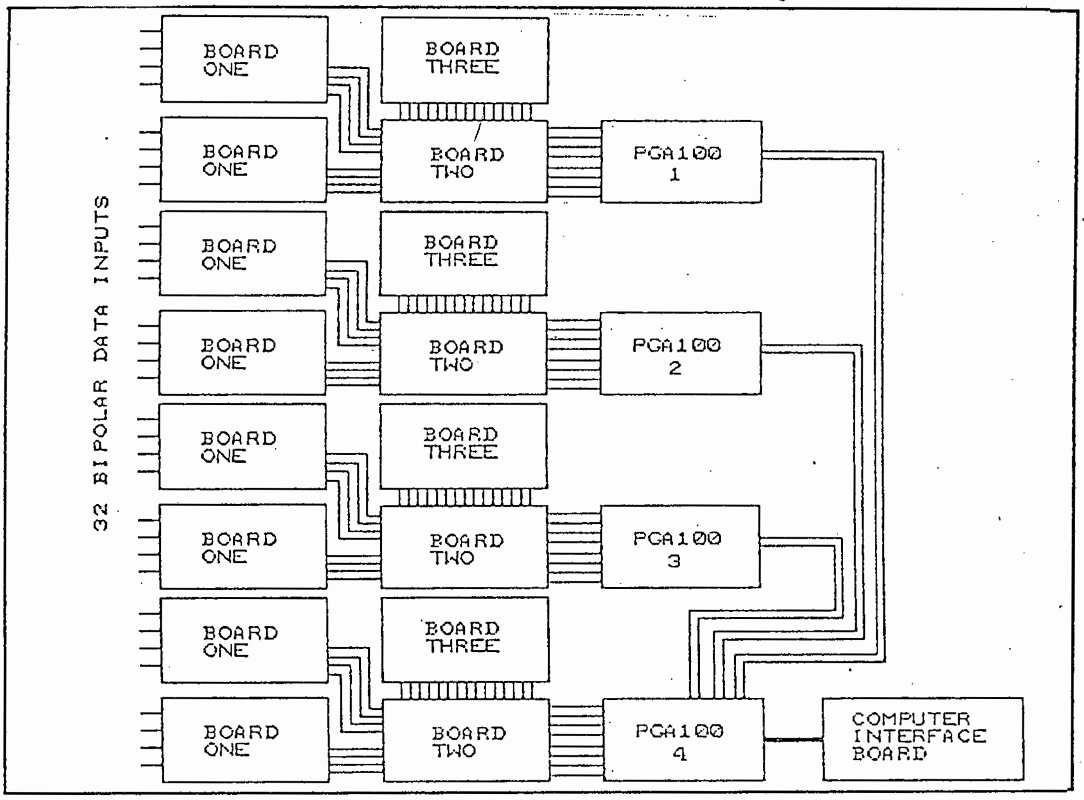


Figure 7.4 **Block Diagram of Interconnections Between Boards for the Complete 32 Channel System.**

In evaluating the software a number of suggestions were made - most could be easily implemented. Other, more complex expansions would involve the addition of routines and modification of the existing program. The system presently allows for 128 Kbytes of acquired data (DOS limits the total memory available for programs and data to 640 Kbyte; 1 Kbyte = 1024 bytes). At a sampling frequency of 2,0 kHz per channel, the data memory would be filled in just over a second for the full complement of 32 channels. Clearly this is inadequate for clinical applications. Expansion of the software for more realistic real-time acquisition, including real-time storage to disk and display in graphic mode, is needed. Limitations imposed by DOS in terms of memory size and allocation, and real-time data transfer and display modalities would have to be considered.

CHAPTER EIGHT

CONCLUSIONS AND RECOMMENDATIONS

8.1 Conclusions

A 32 channel data acquisition system for mapping, storage and display of electrograms during cardiac EP mapping studies was designed. Four channels of this design were implemented in a prototype system which consists of four interconnecting boards external to the computer and one interface board within the computer. The following is a list of the aims that were achieved for the thesis :-

Each channel to be acquired is sampled at 2 kHz. At the user command a four second window of data is acquired and stored to disk.

The frequency response of the system is software programmable within the limits of 0.1 and 500 Hz (choices for low frequency cutoff are 0.1, 1, 30 and 50 Hz and for high frequency cutoff are 100, 250, 400 and 500 Hz).

Amplification is achieved by means of software (choices are 1, 2, 4, 8, 16, 32, 64 and 128).

Software manipulation allows viewing of the sampled waveforms on a high resolution (640 x 480) VGA monitor. Timing calculations and the automatic detection of activation times for 24 of the input channels are available for specific application in WPW syndrome mapping.

The frequency response of the analogue circuitry (on Board One) conforms to the specifications. The phase response however varies to a large degree and may not be acceptable within the clinical situation.

Data may be read by means of dual DMA from the four input channels of the prototype hardware to computer memory.

The system could not be evaluated in the clinical situation for the following reasons :-

An isolated power supply was not implemented.

Only four input channels were implemented.

Problems associated with memory allocation and storage of data for real-time operation.

8.2 Recommendations for Future Work

Recommendations for future work are:-

* Revised specification and implementation of full complement of channels.
* 8 - 16 channel real-time display of waveforms with variable sweep speed.
* Improved selection (and range) of bandwidth and gain settings.
* Further evaluation of phase response and possible modification of circuitry.
* DOS memory assignment for data acquisition and graphical manipulation.
* Implementation of isolated front-end circuitry (power supply).
* Use of a 20 inch super VGA monitor.
* Use of a mouse or tracker-ball in both text and graphics mode.
* Real-time storage of all incoming data onto hard-disc.
* Backup of edited data onto magnetic or optical media.
* Circuitry for stimulus generation and control during EP studies.

REFERENCES

ANTOLINI R, KIRCHNER M, MONGERA A, DISERTORI M, FURLANELLO F.

1984

Real-time beat-to-beat measurement of conduction intervals during cardiac electrophysiological studies.

Clinical Physiology and Physiological Measurements; 5(3):171-183

ANTOLINI R, DISERTORI M, KIRCHNER M, MONGERA A, VERGARA G, IANAMA G, GUARNERIO M, FURLANELLO F.

1984

Automated on-line data analysis during electrophysiological studies in humans.

In: The Application of Computers in Cardiology. Querglas GM. ed., Elsevier Science Publishers B.V. North Holland; 129-132.

ANTOLINI R, KIRCHNER M, MONGERA A, DISERTORI M, FURLANELLO F.

1988

On-line interval measurement during invasive cardiac electrophysiologic testing.

Pace; 11:33-46

BAILEY JJ, BERSON AS, GARSON JR A, HORAN LG, MACFARLANE PW, MORTARA DW, ZYWIETZ C.

1990

Recommendations for standardization and specifications in automated electrocardiography: bandwidth and digital signal processing.

Circulation; 81(2):730-739

BARR RC, SPACH MS.

1977

Sampling rates required for digital recording of intra and extracellular cardiac potentials.

Circulation; 55(1):40-48

BLANCHARD SM, BUHRMAN WC, TEDDER M, SMITH WM, IDEKER RI, LOWE JE.

1988

Concurrent activation detection from unipolar and bipolar electrodes. (abstract)

Pace; 11:525

BLANCHARD SM, DAMIANO RJ, SMITH WM, et al.

1985

Estimating activation times between unipolar electrode sites.

Proceedings 38th Annual Conference on Engineering in Medicine and Biology; 5:123

BONNEAU G, TREMBLAY G, SAVARD P, GUARDO R, LEBLANC R, CARDINAL R, PAGE P, NADEAU R.

1987

An integrated system for intraoperative cardiac activation mapping.

IEEE Transactions on Biomedical Engineering; 34(6):415-423

BUCKLES DS, HAROLD ME, GILLETTE PC, CASE CL, CRAWFORD FA.
1990
Computer-enhanced mapping of activation sequences in the surgical treatment of supraventricular arrhythmias.
Pace; 13(11):1401-1407

CARDINAL R, SAVARD P, CARSON DL, PERRY J, PAGE P.
1984
Mapping of ventricular tachycardia induced by programmed stimulation in canine preparations of myocardial infarction.
Circulation; 70(1):136-148

CERUTTI S, GATTI E, MASCIADRI L.
1980
Digital filtering and baseline-drift correction algorithms in the computerised pre-processing of ECG tracings.
In: Medinfo 80, Lindberg/Kaihara(Editors), North Holland Publishing Company; 231-235.

CLAYDON FJ, PILKINGTON TC AND IDEKER RE
1985
Classification of heart tissue from bipolar and unipolar intramural potentials.
IEEE Transactions on Biomedical Engineering; BME-32(7):513-520.

COCHRANE T, DUNLOP AW, NATHAN AW, CAMM AJ.
1983
Cardiac electrophysiological studies: computer analysis using a digitiser and interactive visual display unit.
Clinical Physiology and Physiological Measurements; 4(3):321-331.

COX JL.
1989
Historical perspectives in the development of cardiac arrhythmia surgery.
In: Arrhythmia Surgery, Operative Clinics. Published by Washington University School of Medicine, Division of Cardiovascular Surgery, Office of Continuing Medical Education.

COX JL AND FERGUSON TB.
1989
Cardiac arrhythmia surgery.
In: Current Problems in Surgery, Year Book Medical Publishers, Inc, Editor-in-chief: Wells, Jr. SA; XXVI(4):204-228

COX JL.
1985
The status of surgery for cardiac arrhythmias.
Circulation; 71(3):413-417

DE BAKKER JMT, JANSE MJ, VAN CAPELLE FJL, DURRER D.
1984
An interactive computer system for guiding the surgical treatment of life-threatening ventricular tachycardias.
IEEE Transactions on Biomedical Engineering; 31(4):362-368

DHEIN S, RUTTEN P, KLAUS W.
1988
A new method for analysing the geometry and timecourse of epicardial potential spreading.
International Journal of Biomedical Computing; 23:201-207

DOWNAR E, PARSON ID, MICKLEBOROUGH DA, CAMERON L, YAO C, WAXMAN MB.
1984
On-line epicardial mapping of intraoperative ventricular arrhythmias: initial clinical experience.
Journal of the American College of Cardiology; 4(4):703-714

EL-SHERIF N, SMITH RA, EVANS K.
1981
Canine ventricular arrhythmias in the late myocardial infarction period 8. Epicardial mapping of reentrant circuits.
Circulation Research; 49:255-265

FISHER JD.
1988
Stimulation as a key to tachycardia localization and ablation.
Journal of the American College of Cardiology; 11(4):889-893.

FUNADA T, IWASE T, IWA T.

1983

Method for computerized display of epicardial maps.

Medical and Biological Engineering and Computing; 21:418-423

GALLAGHER JJ, SVENSON RH, KASELL JH, GERMAN LD, BARDY GH, BROUGHTON A, CRITELLI G.

1982

Catheter technique for closed-chest ablation of the atrioventricular conduction system.

The New England Journal of Medicine; 306:194-200

GALLAGHER JJ, KASELL JH, COX JL, SMITH WM, IDEKER RE, SMITH WM.

1982

Techniques of intraoperative electrophysiologic mapping.

American Journal of Cardiology; 49(1):221-240

GALLAGHER JJ, KASELL J, SEALY WC, PRITCHETT ELC, WALLACE AG.

1978

Epicardial mapping in the Wolff-Parkinson-White syndrome.

Circulation; 57:854-866

GILLETTE P, ZMIJEWSKI M, SHELTON MB.

1989

The evolution of computer application to assist during clinical electrophysiologic testing.

Journal of Electrocardiology; 22(supplement):218-222

HERNANDEZ C, MARQUEZ-MONTES J, LINACERO G, RUFILANCHAS J, GONZALEZ M, CABO C, CASTILLO-OLIVARES J.
1984
Endo-epicardial mapping of arrhythmias in surgery. A versatile microprocessor based system.
In: The Applications of Computers in Cardiology: State of the Art and New Perspectives, G. Martin Quetglas et al. (eds), Elsevier Science Publishers B.V. (North Holland); 55

HILL DW, DOLAN AM.
1976
Intensive care instrumentation. Grune and Stratton, New York, NY; 214-220

HOEKS AP, SCHMITZ GM, ALLESSIE MA, JAS H, HOLLEN SJ AND RENEMAN RS.
1988
Multichannel storage and display system to record the electrical activity of the heart.
Medical and Biological Engineering and Computing; 26:434-438

IDEKER RE, SMITH WM, BLANCHARD SM, REISER SL, SIMPSON EV, WOLF PD.
1989
The assumptions of isochronal cardiac mapping.
Pace; 12(3):456-478.

IDEKER RE, SMITH WM, WALLACE AG, KASELL J, HARRISON LA, KLEIN GJ, KINICKI RE, GALLAGHER JJ.
1979
A computerized method for the rapid display of ventricular activation during intraoperative study of arrhythmias.
Circulation; 59:449-458.

ITATSU H.
1954
Theoretical interpretation of contiguous bipolar ECG and its relationship to the time of arrival of activation. Part 1: some fundamental studies.
Japanese Circulation Journal; 18:1-10

JACKMAN WM, FRIDAY KJ, FITZGERALD DM, BOWMAN AJ, YEUNG-LAI-WAI JA AND LAZARA R.
1989
Localization of left free-wall and posteroseptal accessory atrioventricular pathways by direct recording of accessory pathway activation.
Pace; 12(1):204-214

JENKINS JR, CLEMO HF, BELARDINELLI L.
1987
A microcomputer system for on-line study of atrioventricular node accommodation.
Pace; 10:1301-1308

JOSEPHSON ME, HARKEN AH.

1983

Surgical therapy of arrhythmias.

In: Rosen MR, Hoffman BF (eds): Cardiac therapy. Boston Martinus Nijhoff Publishing; 337-385

JOSEPHSON ME, SEIDES SF.

1979

Clinical cardiac electrophysiology. Techniques and interpretations. Lea and Febiger, Philadelphia; 98

JOSEPHSON ME, HOROWITZ LN, SPIELMAN SR, GREENSPAN AM, VANDEPOL C, HARKEN AH.

1980

Comparison of endocardial catheter mapping with intraoperative mapping of ventricular tachycardia.

Circulation; 61(2):395

KATZ AM, SELWYN A (eds)

1983

The cardiac arrhythmias. An overview of the electrophysiology, diagnosis and management of clinical arrhythmias. Sinauer Associates, Sunderland; 6-27

KLEIN GJ, IDEKER RE, SMITH WM, HARRISON LA, KASELL J, WALLACE AG, GALLAGHER J.
1979
Epicardial mapping of the onset of ventricular tachycardia initiated by programmed stimulation in the canine heart with chronic infarction.
Circulation; 60(6):1375

MAC FARLANE PW AND LAWRIE TDV.
1974
An introduction to automated electrocardiogram interpretation.
Butterworths, London; 67

MEHRA R, ZEILER RH, GOUGH WB, EL-SHERIF N.
1983
Reentrant ventricular arrhythmias in the late myocardial period. Electrophysiologic-anatomic correlation of reentrant circuits.
Circulation; 67(1):11-16.

METTING VAN RIJN AC, PEPER A, GRIMBERGEN CA.
1990
High-quality recording of bioelectric events. Part 1: interference reduction, theory and practice.
Medical and Biological Engineering and Computing; 28:389-397

MEYER CR, KEISER HN.
1977
Electrocardiogram baseline noise estimation and removal using cubic splines and state-space computation techniques.
Computers and Biomedical Research; 10:81-83

MILLER JM, MARCHLINSKI FE, HARKEN AH, HARGROVE WC AND JOSEPHSON ME.
1985
Subendocardial resection for sustained ventricular tachycardia in the early period after acute myocardial infarction.
The American Journal of Cardiology; 55:980-984

NZAYINAMBAHO K, WALEFFE A, KULBERTUS H AND BROHET C.
1989
Localization of the Accessory Pathway in Ventricular Preexcitation (WPW) by means of Combined ECG and VCG recordings.
Journal of Electrocardiology; 22 Supplement:183-188

PARSON I, DOWNAR E.
1984
Clinical instrumentation for the intra-operative mapping of ventricular arrhythmias.
Pace; 7:683-692

PARSON I, MENDLER P, DOWNAR D.

1982

On-line cardiac mapping: an analog approach using video and multiplexing techniques.

American Journal of Physiology; 242:H526-H535

PARSON ID, DOWNAR E.

1987

Cardiac mapping instrumentation for the instantaneous display of endocardial and epicardial activation.

IEEE Transactions on Biomedical Engineering; 34(6):468-472

PIPBERGER H.

1965

Computer analysis of electrocardiograms.

In: Waxman B and Stacy R (eds) Computers in Biomedical Research Vol 1 Academic Press, New York; 125

ROSENBAUM DS, KAPLAN DT, WILBER DJ, SAUL JP, RUSKIN JN AND GARAN H

1989

The precision of electrophysiological mapping: localizing depolarization wave fronts from digitized extracellular electrograms and the role of data sampling rate.

Journal of Cardiovascular Electrophysiology; 1(1): 2-14

ROSS DL, FARRE J, BAR FWHM, VANAGT EJ, DASSEN WRM, WIENER I, WELLENS HJJ.

1980

Comprehensive clinical electrophysiologic studies in the investigation of documented or suspected tachycardias: time, staff, problems and costs.

Circulation; 61(5):1010-1016

SCHERLAG BJ, LAU SH, HELFANT RH, BERKOWITZ WO, STEIN E AND DAMATO AN.

1969

Catheter technique for recording His bundle activity in man.

Circulation; 39:13

SEALY WC, GALLAGHER JJ AND WALLACE AG.

1976

The surgical treatment of WPW syndrome: Evolution of improved methods for identification and interruption of the Kent bundle.

Annual of Thoracic Surgery; 22:443-457

SMITH C.

1986

Design of a general-purpose, optically isolated amplifier.

Journal of Clinical Engineering; 11(5):349-353

SMITH JM, KAPLAN DT, ROSENBAUM D, et al.

1987

Determinants of precision in epicardial activation time estimation.

Procedings of Computers in Cardiology, 101-104

SPACH MS, KOOTSEY JM.
1985
Relating the sodium current and conductance to the shape of transmembrane and extracellular potentials by simulation: effects of propagation boundaries.
IEEE Transactions on Biomedical Engineering; 32:743-755

SPACH MS, BARR RC, SERWER GA, KOOTSEY JM, JOHNSON EA.
1972
Extracellular potentials related to intracellular action potentials in the dog purkinje system.
Circulation Research; 30(5):505-519

SPACH MS, DOLBER PC, HEIDLAGE JF.
1988
Influence of the passive anisotropic properties on directional differences in propagation following modification of the sodium conductance in human atrial muscle: a model of reentry based on anisotropic discontinuous propagation.
Circulation Research; 62:811-832.

SPEKHORST H, SIPPENSGROENEWEGEN A, DAVID GK, METTING VAN RIJN AC, BROEKHUYSEN P.
1988
Radiotransparent carbon fibre electrode for ECG recordings in the catheterization laboratory.
IEEE Transactions on Biomedical Engineering; 35(5):402-406

SWERDLOW CD, MASON JW, STINSON EB, OYER PE, WINKLE RA, DERBY GC.
1986
Results of operations for ventricular tachycardia in 105 patients.
Journal of Thoracic Cardiovascular Surgery; 92:105-113

THOMAS CW.
1982
Electrocardiographic measurement system response.
In: Liebman J, Plonsey R and Gillette PC, (eds): Pediatric Electrocardiography. Williams and Wilkins, Baltimore, Maryland; 40-59

UNTREKER WJ.
Invasive electrophysiologic study: Its role in the pre-excitation syndromes
In: Lawrie TDV and Macfarlane PW (eds): Comprehensive Electrocardiology Theory and Practice in Health and Disease Vol 1. Pergammon Press, New York; 97-115

VAN HEUNINGEN R, GOOVAERTS HG AND DE VRIES FR.
1984
A low noise isolated amplifier system for electrophysiological measurements: basic considerations and design.
Medical and Biological Engineering and Computing; 22:77-85

WELLNER U AND BRODDA K.
1976
Sensitivity of VCG parameters to baseline adjustment procedures.
In: Advances in Cardiology, vol.16, Karger, Basel; 201

WIT AL, ALLESSIE MA, BONKE FIM, LAMMERS W, SMEETS J AND FENOGLIO JJ.
1982
Electrophysiologic mapping to determine the mechanism of experimental ventricular tachycardia initiated by premature impulses: experimental approach and initial results demonstrating reentrant excitation.
The American Journal of Cardiology; 49:166-185

WITKOWSKI FX, CORR PB.
1984
An automated simultaneous transmural cardiac mapping system.
American Journal of Physiology; 247:H661-H668

ZIPES P, AKHTAR M, DENES P, DESANCTIS RW, GARSON JR A, GETTES LS, JOSEPHSON ME, MASON JW, MYERBURG RJ, RUSHKIN JN, WELLENS HJJ.
1989
Guidelines for clinical intracardiac electrophysiologic studies.
A report of the American College of Cardiology / American Heart Association Task Force on Assessment of Diagnostic and Therapeutic Cardiovascular Procedures (Subcommittee to assess clinical intracardiac electrophysiologic studies).
Circulation; 80(6): 1925-1939

SELECTED BIBLIOGRAPHY

EGGEBRECHT LC

1990

Interfacing to the IBM Personal Computer

2nd ed. USA. Howard W. Sams and Company.

HOROWITZ P AND HILL W.

1989

The Art of Electronics

2nd ed. Cambridge. Cambridge University Press

WEBSTER JG (editor in chief)

1988

Encyclopedia of Medical Devices and Instrumentation.

Vol 2, USA, John Wiley and Sons.

APPENDIX A

WPW SURGICAL TECHNIQUES

A1. Surgical Treatment Of Accessory Pathways

Many different techniques and procedures are used for the surgical ablation of accessory pathways in WPW syndrome. The surgical procedure outlined below is performed by Dr. J Cox of Barnes Hospital in St Louis, USA. This procedure was chosen because it achieves a success rate of practically 100% and uses a multipoint measurement system that is similar to the one envisioned for this thesis (Cox and Ferguson 1989).

A1.1 Surgical procedure for ablation of accessory pathways

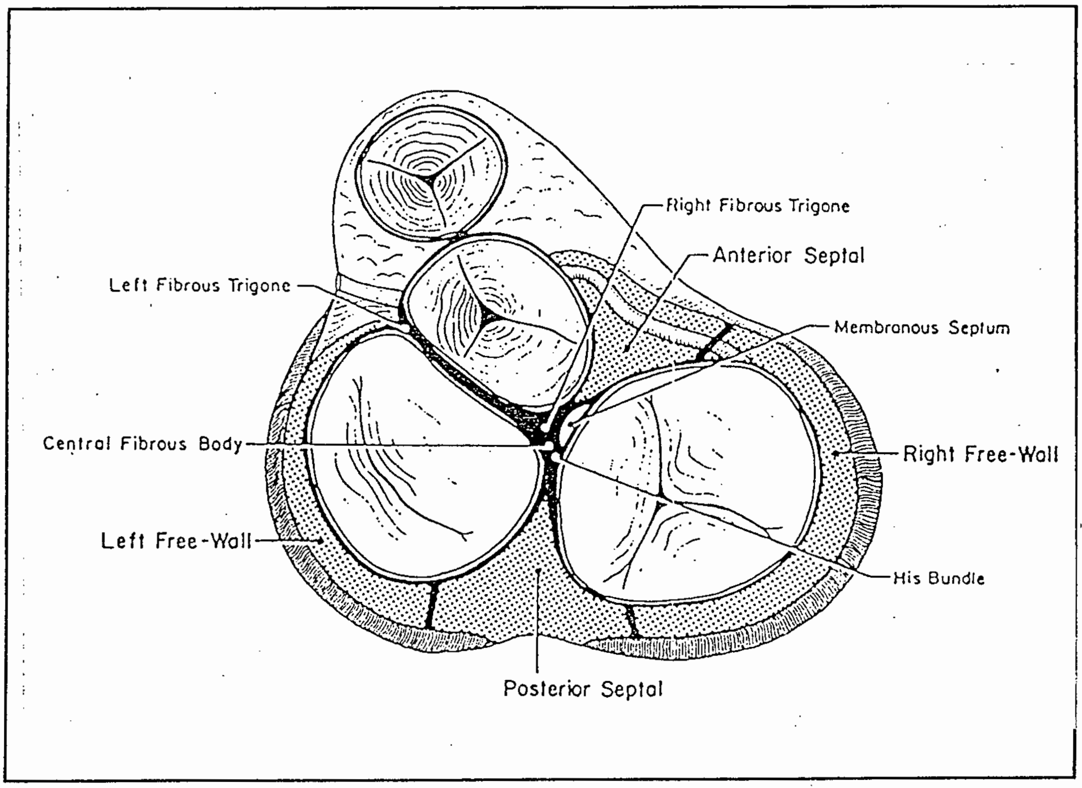
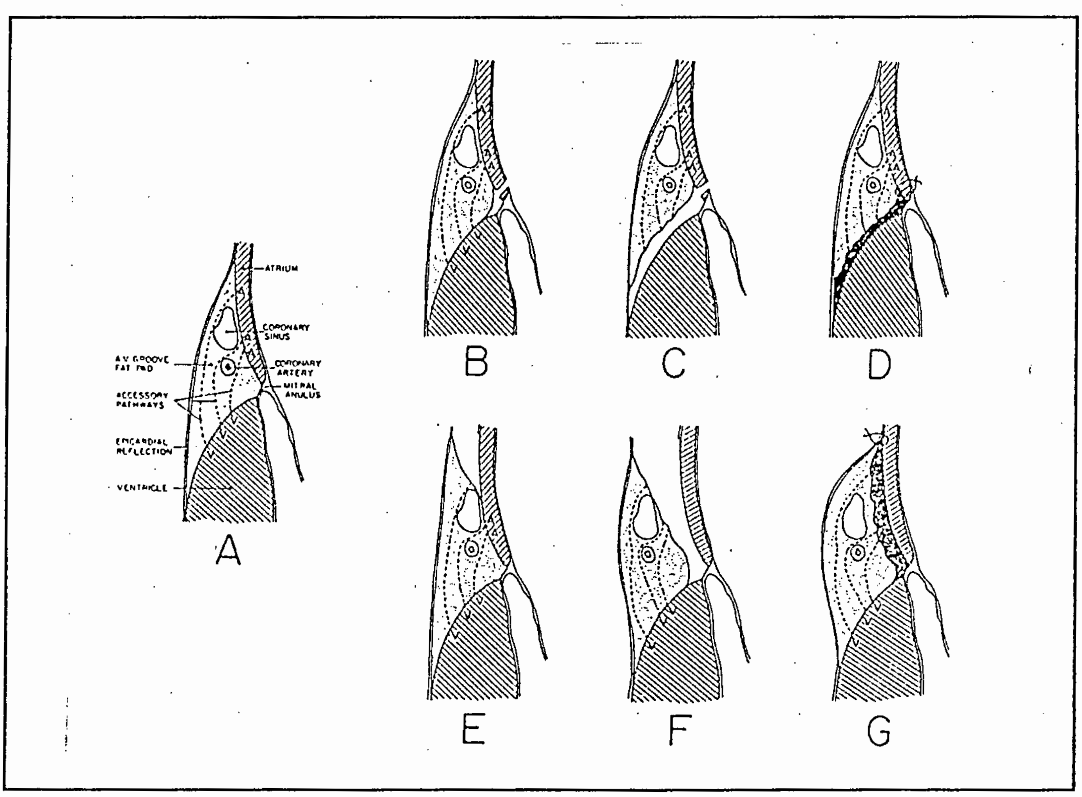


Figure A1 **Superior View of the Heart With Atria Removed Demonstrating the Boundaries of the Four Anatomic Areas Where Accessory Pathways Can Occur** (Cox and Ferguson 1989).

Accessory pathways in WPW syndrome can be localised to four anatomic areas of the heart : the left free-wall, right free-wall, posterior septal and anterior septal regions. Figure A1 shows a superior view of the heart with the atria removed. Boundaries of the four anatomic areas where accessory pathways can occur are shown and all the atrio-ventricular connections of these pathways must insert into ventricle somewhere within these anatomic boundaries. If an accessory pathway is found in any particular region then correct disection of the entire region will ensure ablation of the pathway.



<u>Figure A2</u> **Cross-sectional View of the Heart Showing (B-D) Endocardial and (E-G) Epicardial Procedures for Ablation of Accessory Pathways** (Cox and Ferguson 1989).

The surgical technique for ablation of accessory pathways is similar in all the four regions and for this reason only the technique for the ablation of left free-wall pathways will be discussed here.

There are two possible ablation approaches :- endocardial and epicardial. The epicardial technique was not very popular during the early 1970's because of the proximity of the incision to the coronary sinus and coronary artery as well as the inability at that time to cardioplegically arrest the heart. It was reintroduced in 1985 and using modern surgical methods both approaches have produced excellent results (Cox and Ferguson 1989).

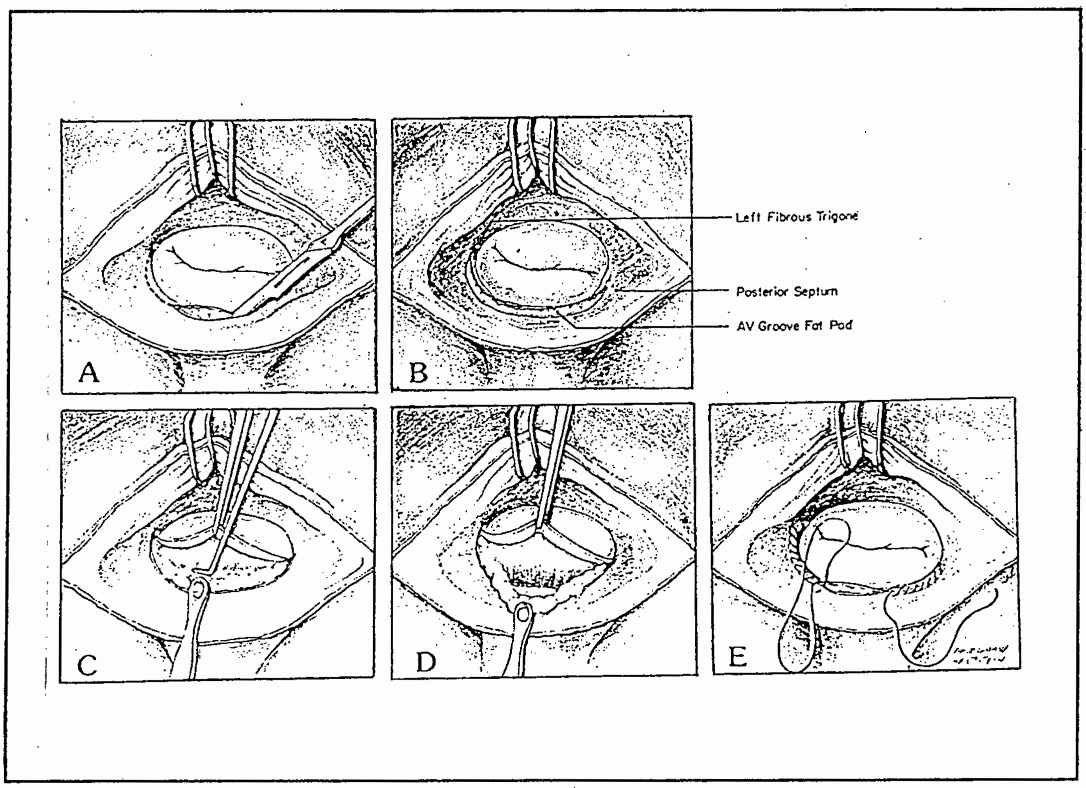


Figure A3 Diagrammatic Representation of Endocardial Surgical Ablation of the Left Free-wall Region (Cox and Ferguson 1989).

The epicardial approach to the left free-wall accessory pathway is by dissection from the atrial side of the AV groove. A plane of dissection is extended to the level of the posterior mitral valve annulus and is then carried further to the top of the posterior left ventricle. This divides the ventricle from all accessory pathways except those immediately adjacent to the mitral valve annulus. Although these persistent accessory pathways are few in number cryolesions at the level of the mitral valve annulus are performed to destroy any accessory pathway that may have survived the dissection. The cryo-probe uses liquid N2 and operates at a temperature of -60 C.

The endocardial technique approaches accessory pathways on the left free-wall through a left atriotomy after the heart has been arrested. A supra-annular incision is made 2mm above the mitral valve annulus, from the left fibrous trigone to the posterior septum. The plane of dissection is continued all the way to the epicardial reflection off the posterior left ventricle - this extensive incision is to ensure that any accessory pathway located in that region is cut. Even so a small accessory pathway immediately adjacent to the valve annulus may remain and to prevent this the annulus is cleaned meticulously and "squared off" with a nerve hook or knife. This dissection thus exposes the entire left freewall space and each of its boundaries leaving no other path for the accessory pathway to follow.

APPENDIX B

PATIENT DATABASE

The following pages contain the database used for this project. Each page shows the recording sheet for each patient and a printout of an example screen of the data for that patient (note the difference in morphology in the waveforms). Patient names and numbers have been altered.

In each recording sheet the following were used:-

Counter	The value of the tape counter recorded at various times.
Channel	The channel used to record from a particular electrode.
Position	The position of the electrode.
HP	The highpass frequency settings (low frequency cutoff).
LP	The lowpass frequency settings (high frequency cutoff).
Gain	The multiplication factor applied to the incomming signal. (1 = 1 mV/cm)

EPS Recording Sheet No. 1 900914

Patient #1
Born: 1930
Tape No: 1

Counter	Channel	Position	HP	LP	GAIN
0		CALIBRAT	.01	30	
0+	1	Lead I	30	250	1
	2	AVf	30	250	1
	3	Lead III	30	250	1
	4	RV apex	30	250	1
125	4	RV apex	30	250	.5
140					

Comments:
Operation involved placing of pacemaker leads.
Pacemaker input MOBITZ I Model 438

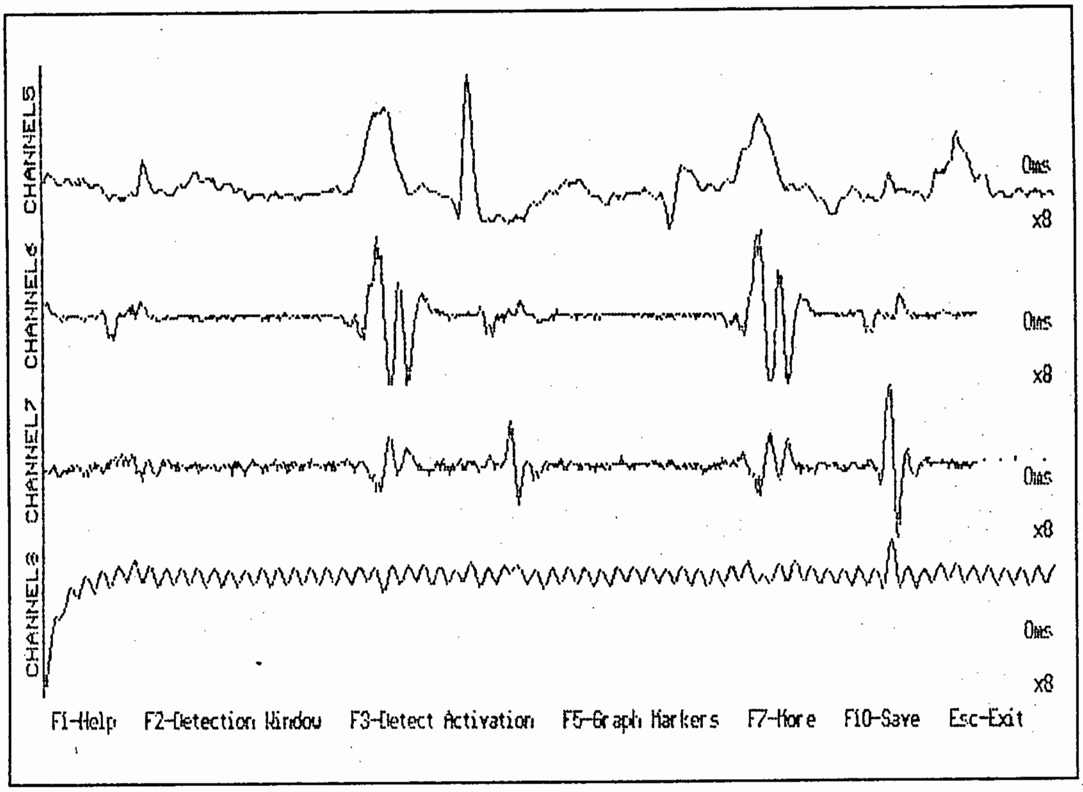


Figure B.1 **Screen Printout of Patient 1**

EPS Recording Sheet No. 2 900917

Patient #2
Born: 25/02/1938 R/S: 7 GSH
Tape No: 1

Counter	Channel	Position	HP	LP	GAIN
140		CALIBRAT			
145	1	Lead I	.01	30	1
	2	HIS	30	250	.2
	3	High RA	30	250	1
	4	--	30	250	
202					

.5mv p/p on surface In reciprocating tachycardia.
Pause, then pacing as out of RT.

Counter	Channel	Position	HP	LP	GAIN
202	1	HIS	30	250	.2
	2	High RA	30	250	1
	3	CS(dist)	30	250	.5
246					

Comments:
281 Pause. HIS G = .1 HRA G = .2 Continue
292 CS being moved - 320
HRA off, (now mapping catheter). Pacing to get tachycardia.
368 - Gains changing
398 Mapping - position changing.

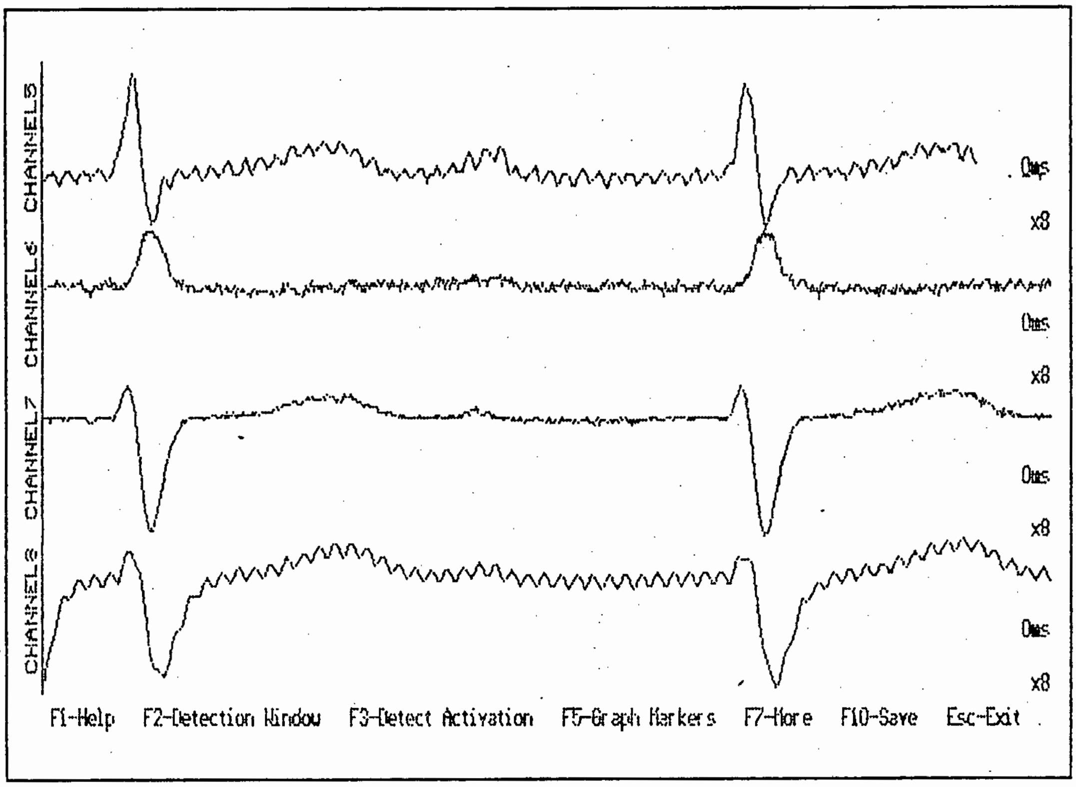


Figure B.2 Screen Printout of Patient 2

EPS Recording Sheet No. 3 900918

Patient #2 (repeat)
Born: 25/02/1938 R/S: 7 GSH
Tape No: 1

Counter	Channel	Position	HP	LP	GAIN
398		CALIBRAT			
501	1	STD I	.01	250	1
	2	HRA	30	250	.05
	3	HIS	30	250	.05
	4	RV	30	250	.2
638					

Shock 100J to HIS.
580 - Pacing at 50pps - 318

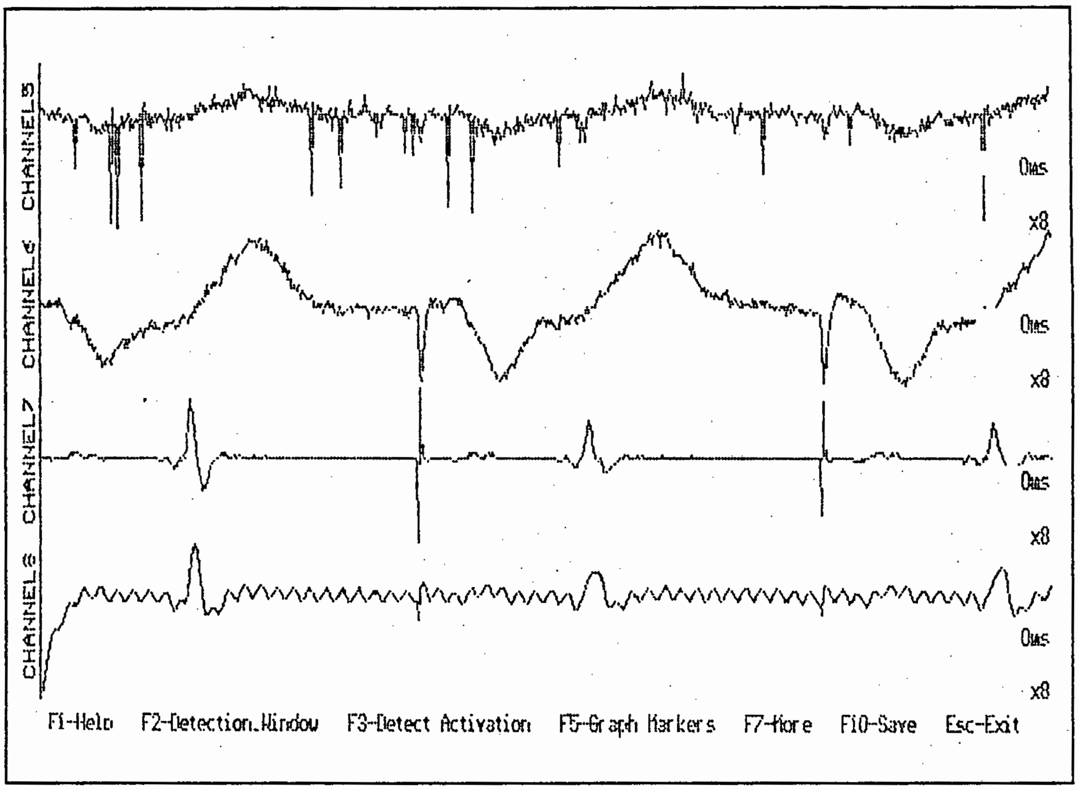


Figure B.3 Screen Printout of Patient 2.

EPS Recording Sheet No. 4 901011

Patient #3
Born: 05/05/1931 R/S 3 GSH
Tape No: 1

Counter	Channel	Position	HP	LP	GAIN
638		CALIBRAT			
640	1	LEAD I			1
	2	RA	30	250	.05
	3	HIS	30	250	1
	4	VENT	30	250	.2
690	Stimulate/Pace 600ms				
737	Atrial Pacing getting faster				
760	Ventricular Pacing				
834					

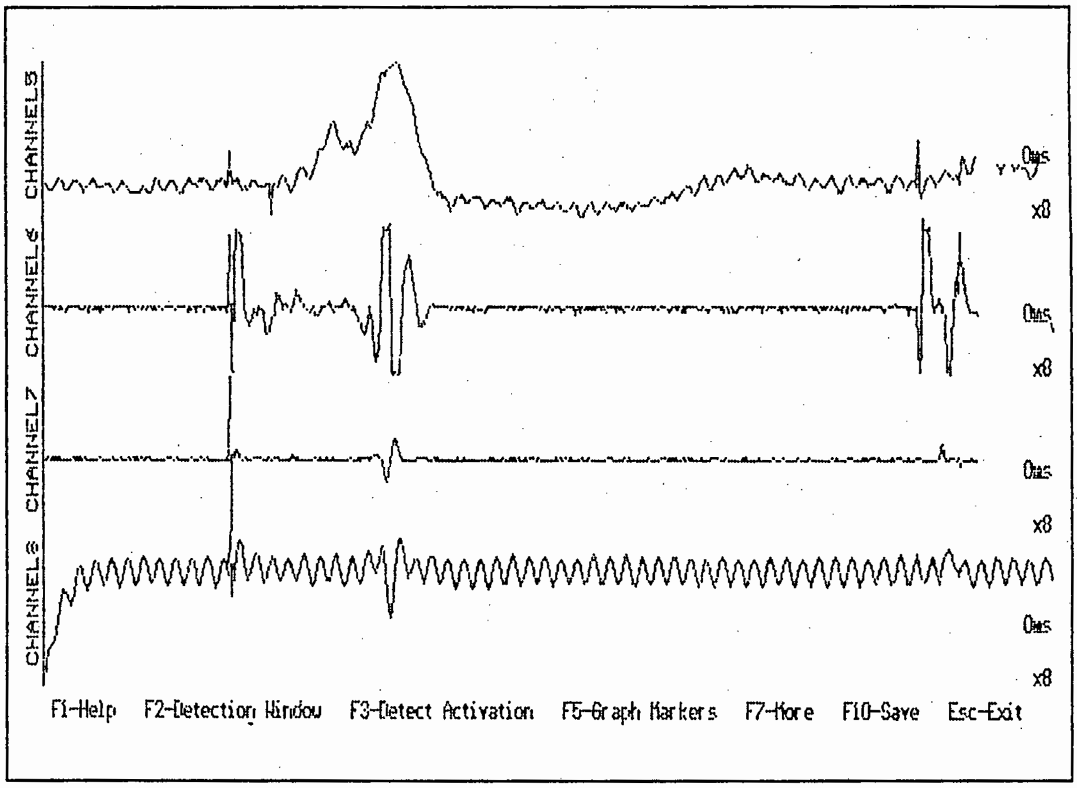


Figure B.5 Screen Printout of Patient 3

EPS Recording Sheet No. 5 901016

Patient #4
Born: 02/03/1927 R/S 2 GSH
Tape No: 1 Reset after 834:-

Counter	Channel	Position	HP	LP	GAIN
0	0	CALIBRAT			
004	1	LEAD I	.01	25	1
	2	V6	.01	25	.1
	3	RA	30	250	.05
	4	RV	30	250	.2
128	Trying to break AV node reentry.				
183	Pacing RA for AV node refractory period				

Tape No: 2

0		CALIBRAT			
006	1	LEAD I	.01	25	1
	2	V6	.01	25	1
	3	RA	30	250	.05
	4	RV	30	250	.2
107		Pacing RV			
108	1	V2	.01	25	1
	2	HIS	30	250	.05
150		Inc atrial pacing.			
180					

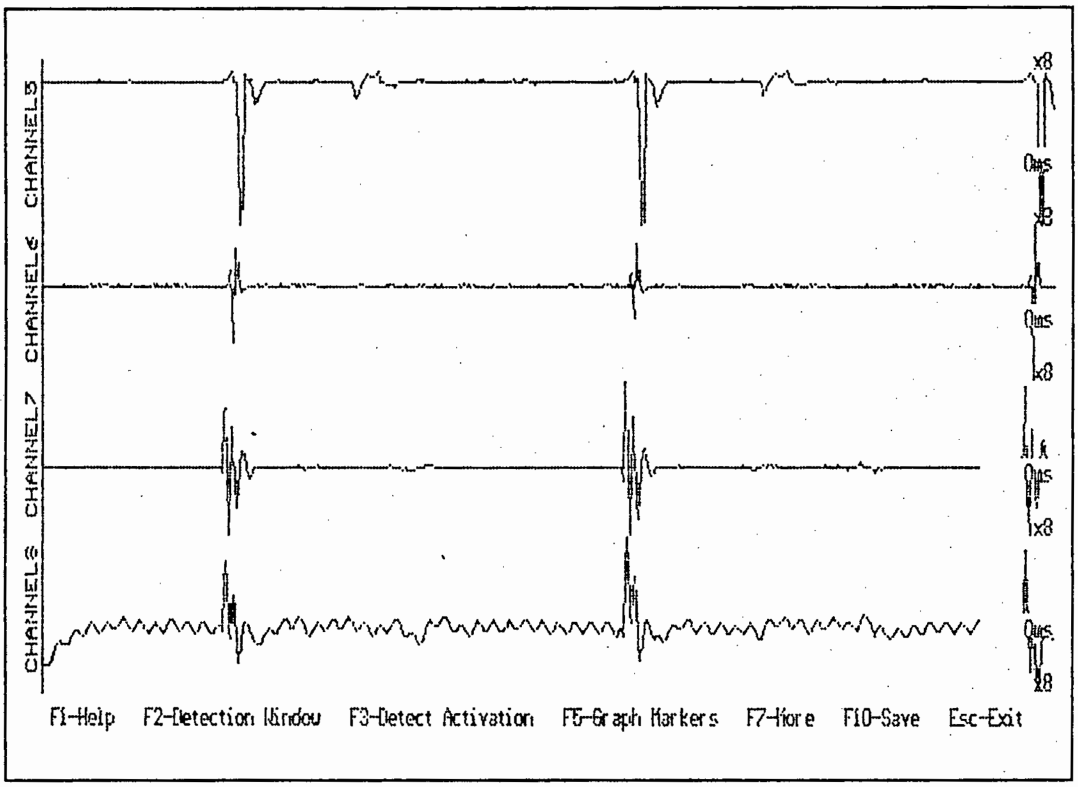


Figure B.6 **Screen Printout of Patient 4**

EPS Recording Sheet No. 6 901017

Patient #5
Born: 04/06/1985 R/S 2 GSH
Tape No: II

Counter	Channel	Position	HP	LP	GAIN
185		CALIBRAT			
190	1	aVF	.01	250	1
	2	RA	30	250	.05
	3	RAc	30	250	.05
	4	HIS	30	250	1
212	1	aVF	.01	250	1
	2	RA	.01	250	.05
	3	RAc	30	250	.05
	4	V6	30	250	1
362	Pacing 440 (HR = 136)				
408					

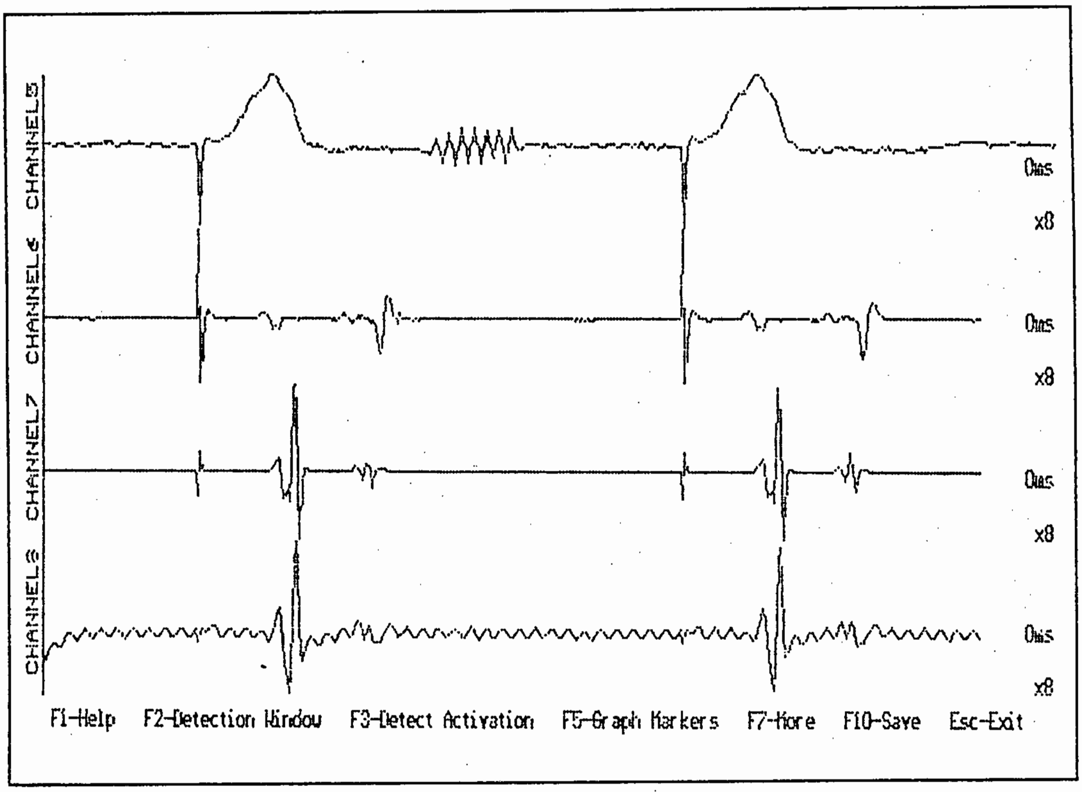


Figure B.7 Screen Printout of Patient 5

EPS Recording Sheet No. 7 901016

Patient #6
Born: 18/04/1940 R/S 3
Tape No: II

Counter	Channel	Position	HP	LP	GAIN
408		CALIBRAT			
410	1	LEAD I	.01	50	1
	2	aVF	.01	50	1
	3	V2	30	25	.5
	4				
446	1	LEAD I	.01	50	1
	2	RA	30	250	.05
	3	Cor	30	250	.1
	4	RV	30	250	.5
571					

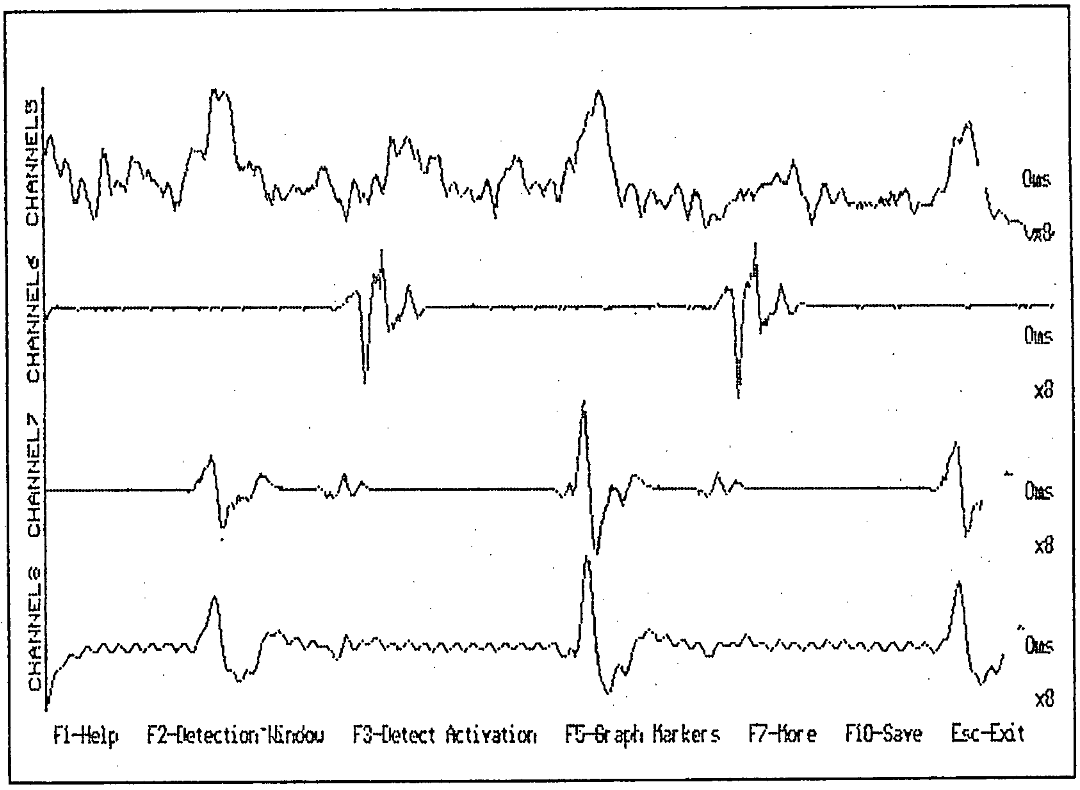


Figure B.8 Screen Printout of Patient 6

APPENDIX C

SIGNAL QUALITY CONSIDERATIONS

C.1 Interference in Biological Recordings

The goal of a bioelectric recording system is the faithful reproduction fo the desired signals from muscle or nerve. The undesired signals are called noise, interference or artifact. An upper level an upper level of interference of $10\mu V$ peak-to-peak can be accepted in most bioelectric measurements. Noise sources may be divided into internal or external interference sources.

Internal Interference is mainly due to signals generated by muscles or motion of the patient skin beneath electrodes. To reduce the noise caused by muslce the electrodes should be placed as close a spossible to the source of the signal and as far as possible from large muscle masses. Motion artifact occurs when the skin beneath the electrode stretches. This stretching causes the skin potential between the inside and outside barrier layer of skin to change from approximately 35 mV to 25 mV. Motion artifact can be reduced by abrasion or puncture of the skin before electrode placement which reduces the skin potential.

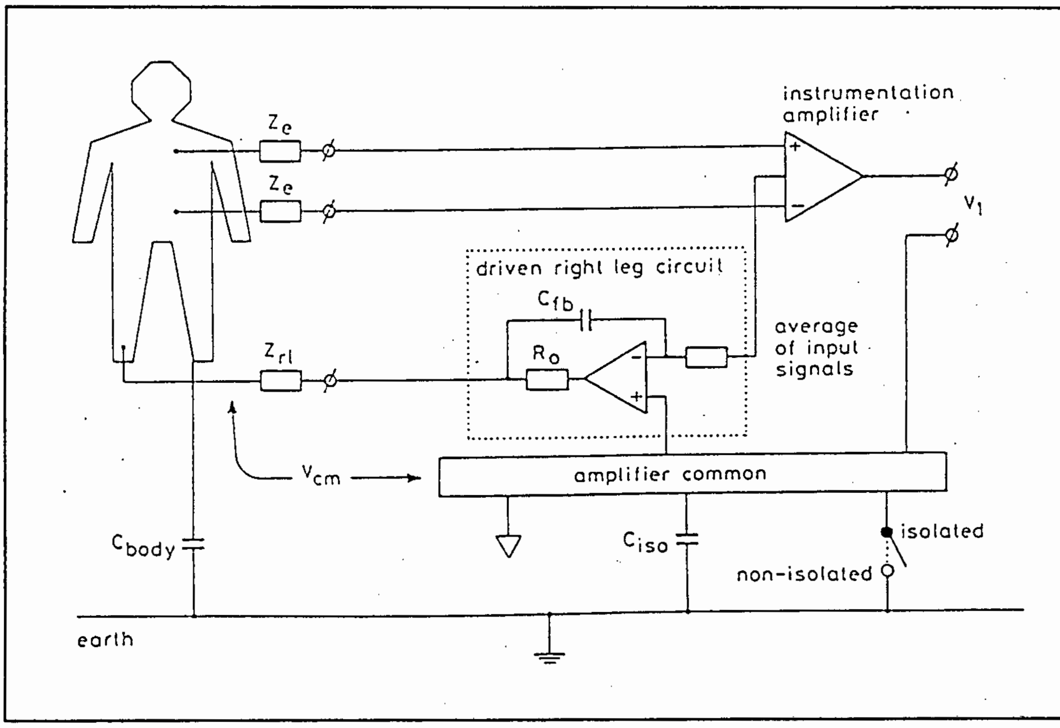
External Interference is mainly due to static electricity, radiofrequency interference and powerline interference. Static electricity is caused by nursing staff or patient movement in low humidity conditions. It usually appears when a high impedance occurs beneath one of the electrodes and simple skin abrasion reduces the interference.

Surgeons use electrosurgical units to cut and coagulate tissue during surgery. Sensitive electromedical equipment should be designed to be capable of working in such radiofrequency fields of up to 100 mV/m and modulated at 50 Hz (Webster 1984).

Interference from the mains power may originate from currents through the body, into the amplifier or into the measurement cables. These currents are due to capacitive coupling between patient, power lines and earth. If the measurement cables form a loop then magnetically induced interference may also occur (Metting van Rijn 1990). Suppression in this particular case of magnetic induction is simple in theory; by twisting of the measurement cable pairs (Webster JG 1988).

The portion of current flowing between patient and amplifier common causes what is termed the common mode voltage. Reduction of this type of interference is achieved with the use of modern instrumentation amplifiers with high common mode rejection ratios (CMRR). A CMRR of 80-90 dB is common, as in the INA102 (National Instruments, Austin). Even with high CMRR such as this it is important to reduce the common mode voltage as much as possible. This is because the common mode voltage may be converted into a differential input voltage when there are differences in electrode impedances. This mechanism is often called the "potential divider effect" (Metting van Rijn, 1990).

Connecting the amplifier common to earth is an effective method of reducing the common mode voltage. If this is done the only source of interference is the capacitance between the patient and the mains (Van Heuningen 1984). If this earth connection cannot be made, due to isolation considerations, then another method of largely reducing the common mode voltage is to use a driven right leg circuit. Here an extra amplifier drives the patient at the same voltage as the voltage of the amplifier common (this equals the common mode voltage).



<u>Figure C.1</u> Driven Right Leg Circuit. If the gain of the circuit is high, the common-mode voltage V_{cm} is much smaller than the voltage across Z_{rl}

Shielding is a very effective method of reducing interference currents in cables. Here the signals to be measured travel along cables that are surrounded by a conductor connected to the signal earth. Guarding is the technique used when the shield surrounding the measurement cables is driven by the right leg amplifier. Metting van Rijn (1990) cautions the use of this technique in combination with other interference reduction techniques due to the possibility of feedback, resulting in instability.

C.2 Filtering in sampled data systems

Signal processing is commonly performed by means of a computer which samples at discrete time intervals and digitises the results. For physiological signals, the spectral density of interest is normally below some frequency of interest f_i. Often the signal level above f_i is low while noise remains white with a power greater than the signal. The input signal thus needs to be passed through an analogue filter to attenuate frequencies higher than the frequency of interest. The Nyquist criterion gives a lower bound for the sampling frequency fs for valid estimation of the signal.

$$fs > 2fi$$

In practice it is necessary to sample at a higher frequency than fs due to aliasing where power at high frequencies is added to the frequencies of interest (van Heuningen, 1984). One cannot continue to increase the sampling frequency however as the higher the sampling frequency, the greater the amount of data, and thus the more memory necessary for data storage and manipulation. A trade-off is therefore necessary between high sampling frequency and the maximum amount of data, and subsequent processing, necessary.

C.3 Isolation

The electrical contact to a patient being monitored is usually of low resistance because of the attachment of electrodes and use of electrode gel. When transcutaneous electrodes or intravenous catheters are used a very low-impedance path is formed between patient and monitoring equipment. Isolation of the patient from sources of electrical shock is therefore of utmost importance.

The safety of the patient can be insured by using an isolation amplifier between the monitoring equipment and the patient (Smith 1986). The isolation amplifier provides a very high impedance path between patient and monitoring equipment. The path impedance must be high enough to limit the worst-case current to below 10 μA. A current of 100 μA is sufficient to induce ventricular fibrillation with endocardial and epicardial electrodes at mains frequency (Hill 1976).

A preferred maximum isolation current of 1 μA should be the aim. This can be achieved by the use of transformers or optical isolators. Transformers are normally larger and heavier than optical isolators and usually withstand lower isolation voltages.

The limiting factors with optical isolators is that they are inherently slower than ordinary bipolar transistors. Special combinations of low power schotky are however giving very high speeds of operation. A typical nonlinear forward transfer function of the optical isolator can cause problems as a linear transfer function between input and output is essential for faithfully collecting analogue biomedical signals foe analysis or monitoring. The temperature dependence of the isolator also introduces non-linearities because of the current transfer ratio. A reduction in amplitude of quarter a percent per degree Celcius is common with an increase in the temperature of the optoisolator (Smith 1986).

APPENDIX D

CIRCUIT BOARD DESIGN

ANALOGUE INPUT BOARD (BOARD ONE)

Circuit Schematics

Artwork - Component Side

Solder Side

Component Layout

ANALOGUE FILTERING BOARD (BOARD TWO)

Circuit Schematic

Artwork - Component Side

Solder Side

Component Layout

OSCILLATOR CONTROL BOARD (BOARD THREE)

Circuit Schematic

Artwork - Component Side

Solder Side

Component Layout

AMPLIFICATION AND ISOLATION BOARD (BOARD FOUR)

Circuit Schematics

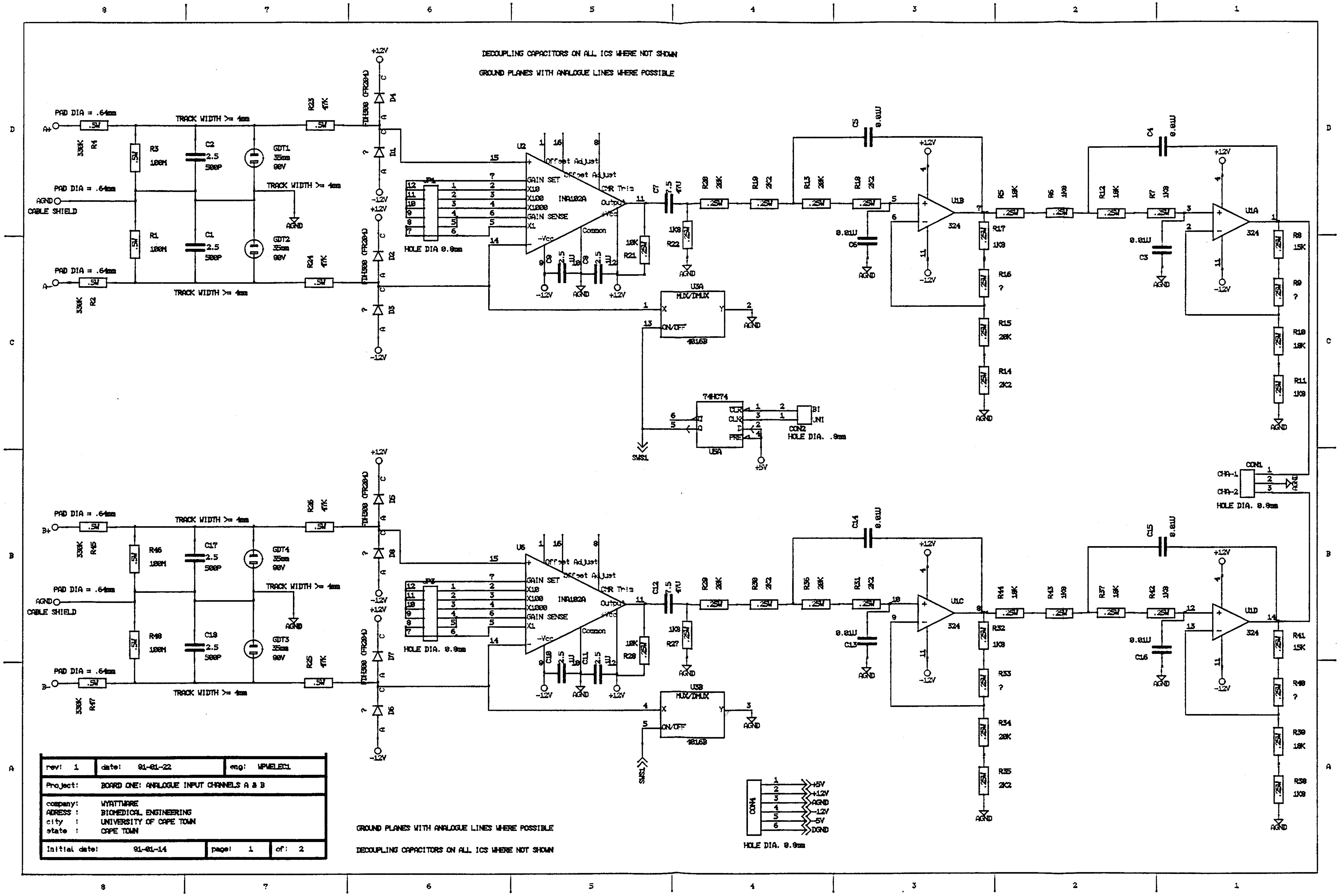
Artwork - Component Side

Solder Side

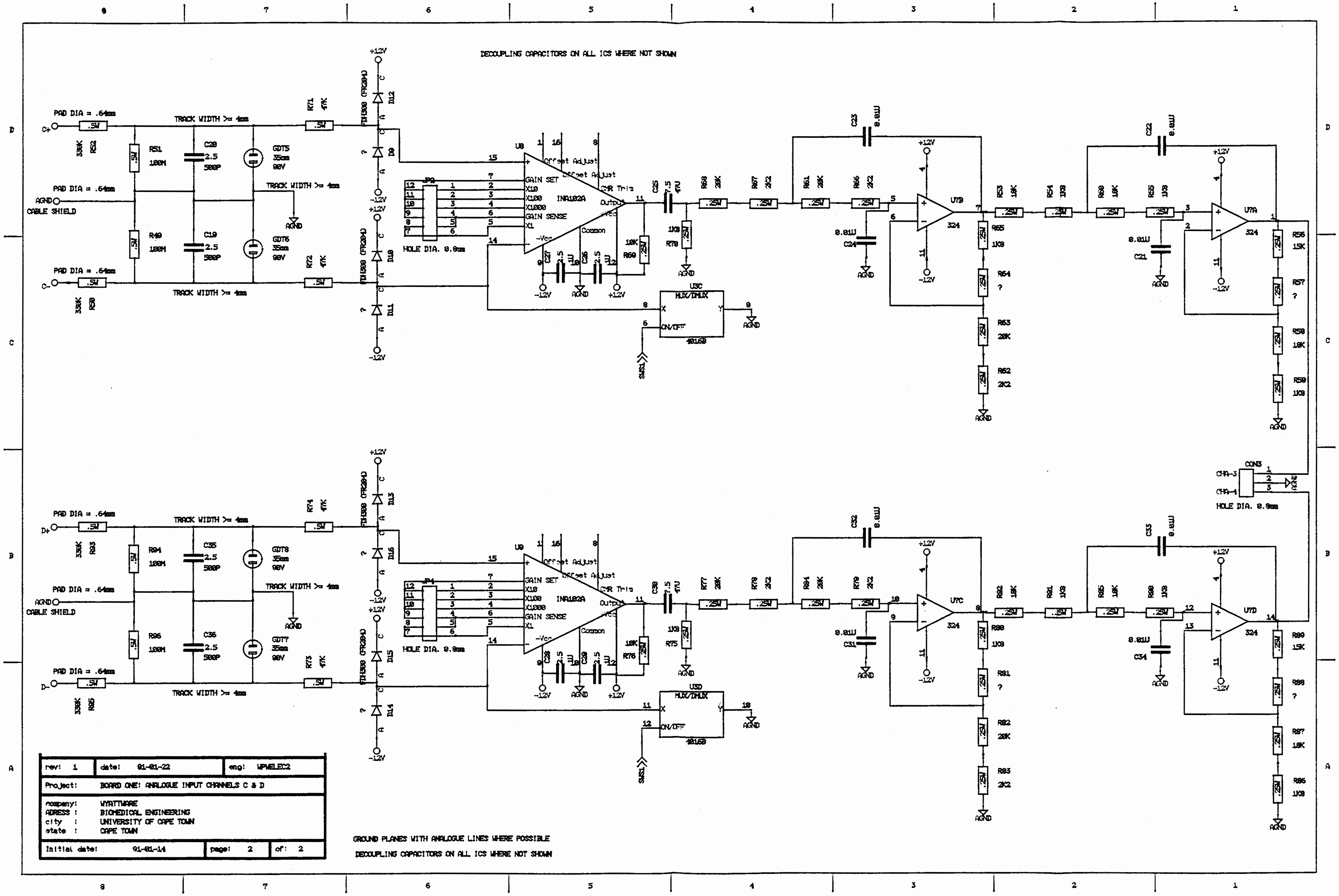
Component Layout

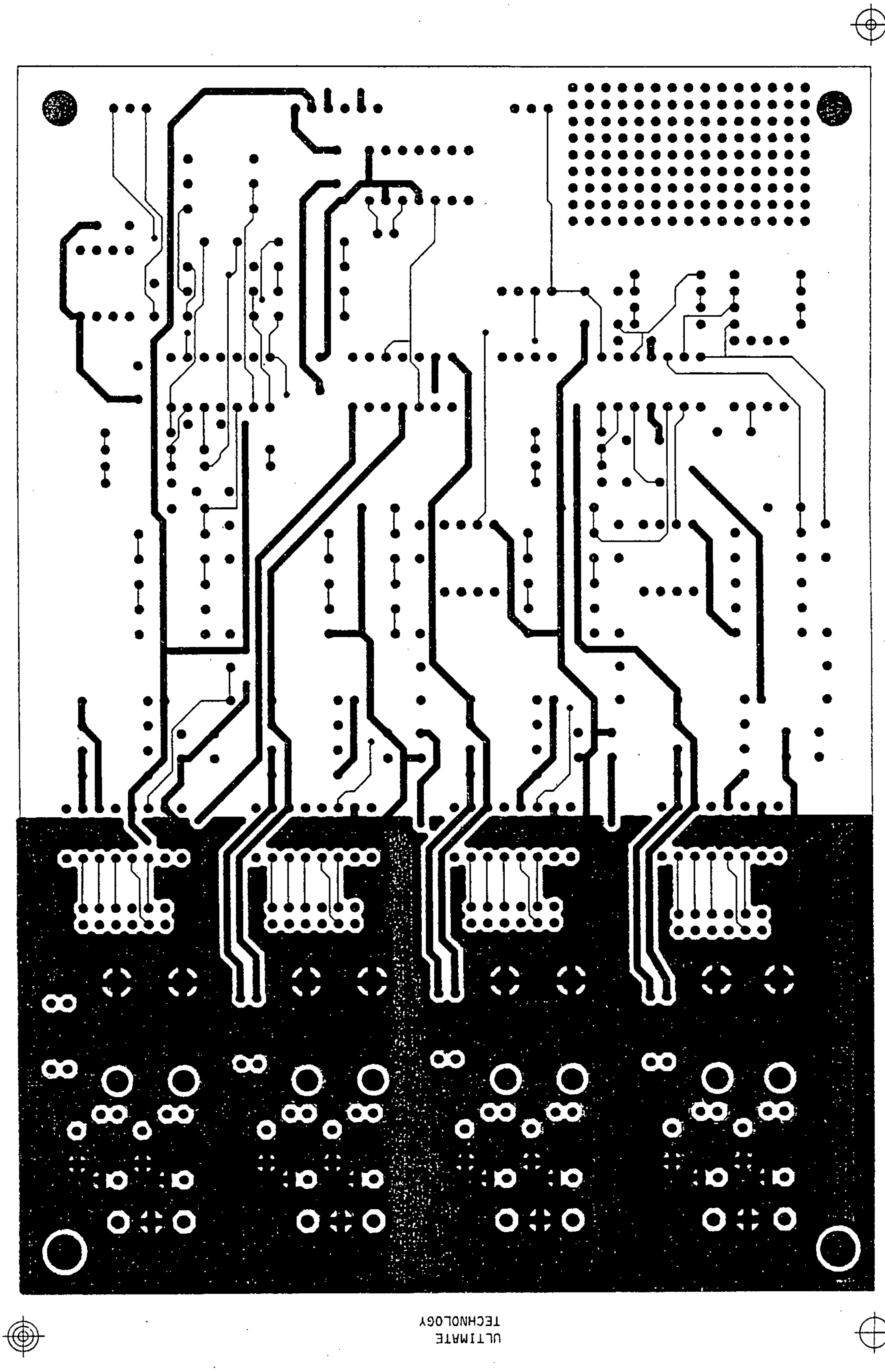
INTERFACE BOARD

Circuit Schematic



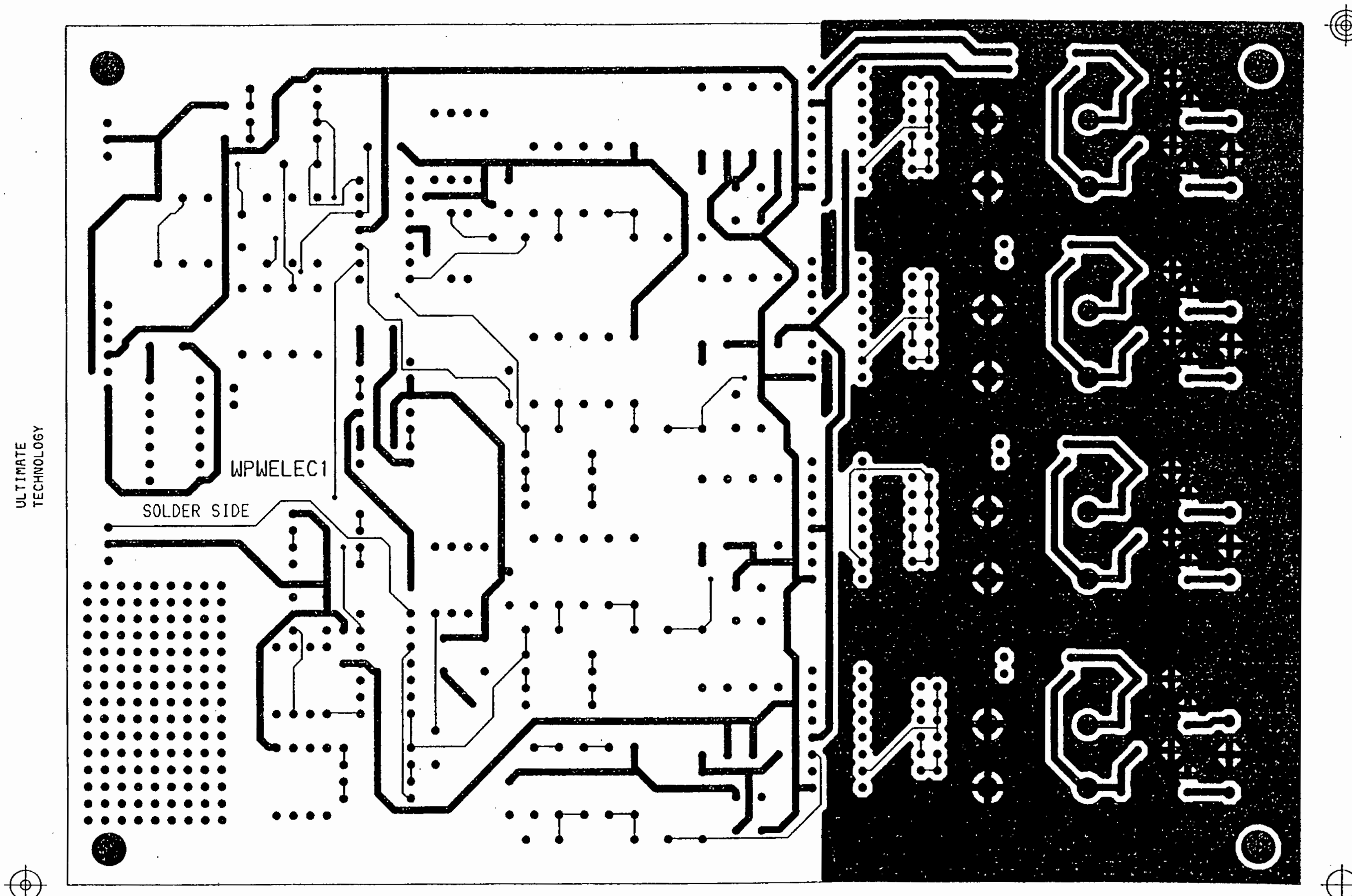
DECOUPLING CAPACITORS ON ALL ICS WHERE NOT SHOWN
GROUND PLANES WITH ANALOGUE LINES WHERE POSSIBLE
PAD DIA = .64mm
TRACK WIDTH >= 4mm
AGND
CABLE SHIELD
HOLE DIA 0.8mm
INA102A
MUX/DMUX
ON/OFF
4016B
74HC74
CON2
HOLE DIA. .8mm
CON1
CHA-1
CHA-2
HOLE DIA. 0.8mm
CON4
+5V
+12V
AGND
-12V
-5V
DGND
rev: 1
date: 91-01-22
eng: WPWELEC1
Project: BOARD ONE: ANALOGUE INPUT CHANNELS A & B
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 91-01-14
page: 1
of: 2
GROUND PLANES WITH ANALOGUE LINES WHERE POSSIBLE
DECOUPLING CAPACITORS ON ALL ICS WHERE NOT SHOWN


DECOUPLING CAPACITORS ON ALL ICS WHERE NOT SHOWN
PAD DIA = .64mm
TRACK WIDTH >= 4mm
AGND
CABLE SHIELD
HOLE DIA. 0.8mm
INA102A
MUX/DMUX
4016B
324
CON3
CHA-3
CHA-4
rev: 1
date: 91-01-22
eng: WPWELEC2
Project: BOARD ONE: ANALOGUE INPUT CHANNELS C & D
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 91-01-14
page: 2
of: 2
GROUND PLANES WITH ANALOGUE LINES WHERE POSSIBLE
DECOUPLING CAPACITORS ON ALL ICS WHERE NOT SHOWN

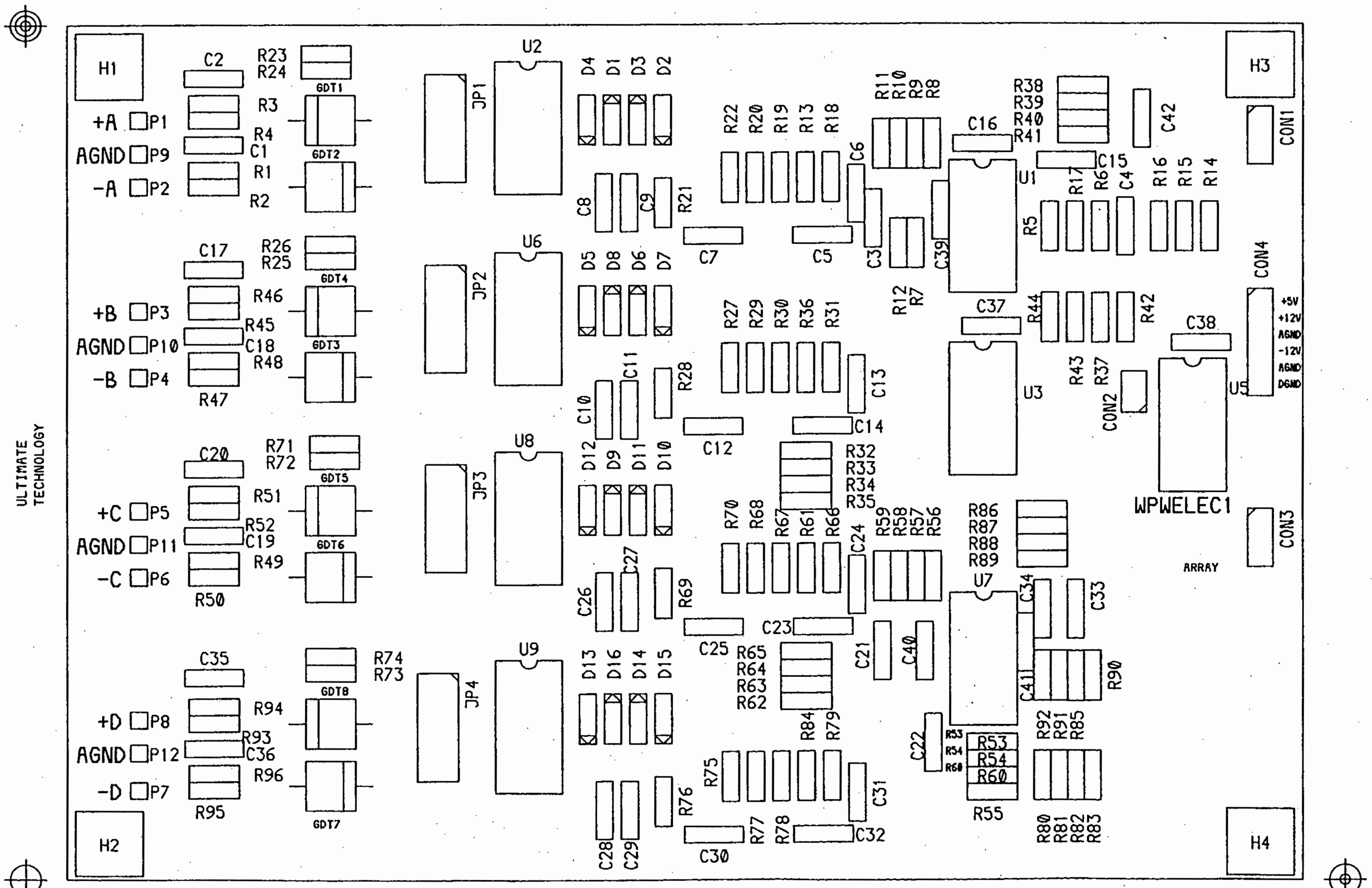


WPWELEC1.P0 (APR. 16. 1991) (10:42) (PCB) SCALE: 150%
DRILL REF PNT: 0.000, 0.000 (INCH)

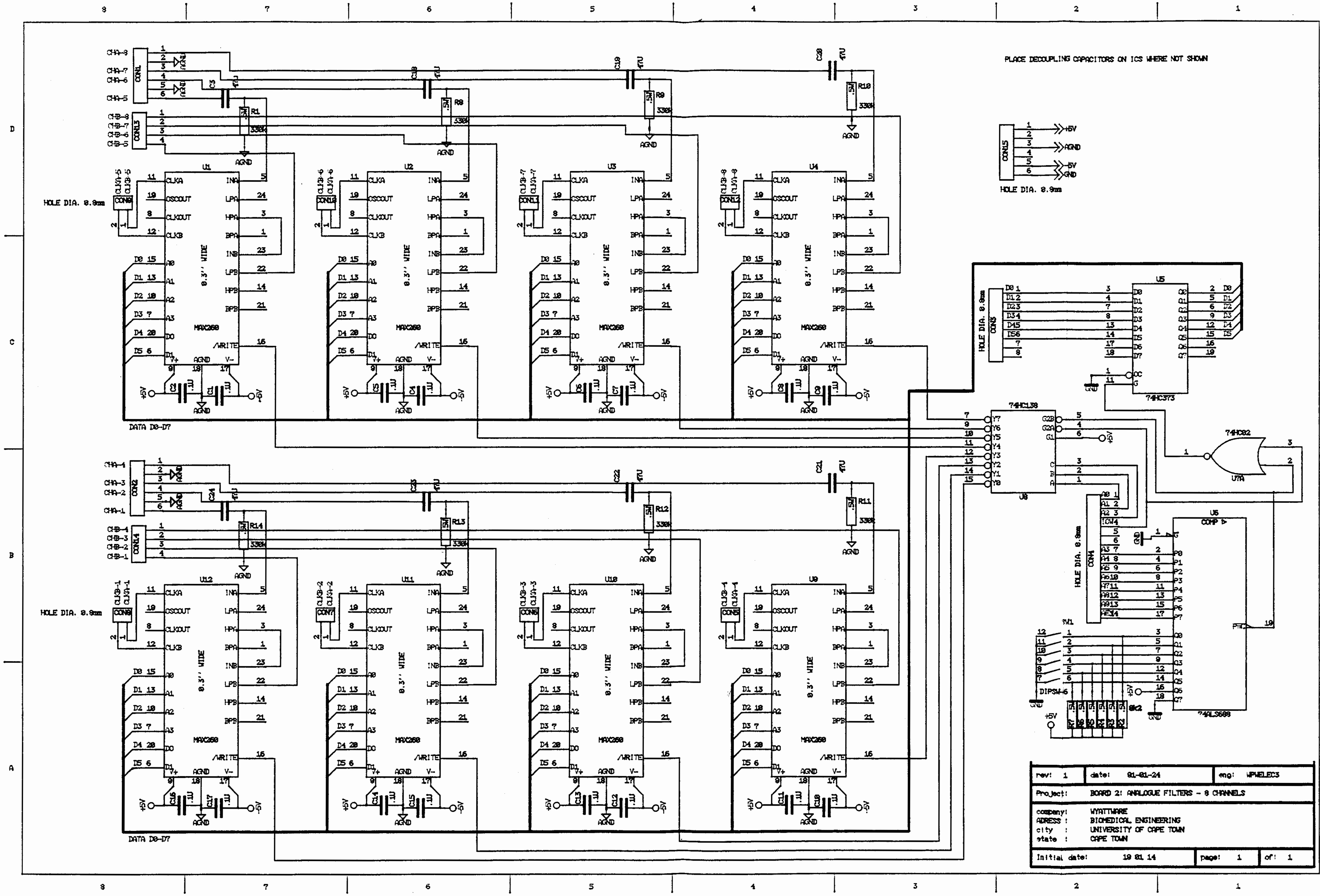
ULTIMATE
TECHNOLOGY

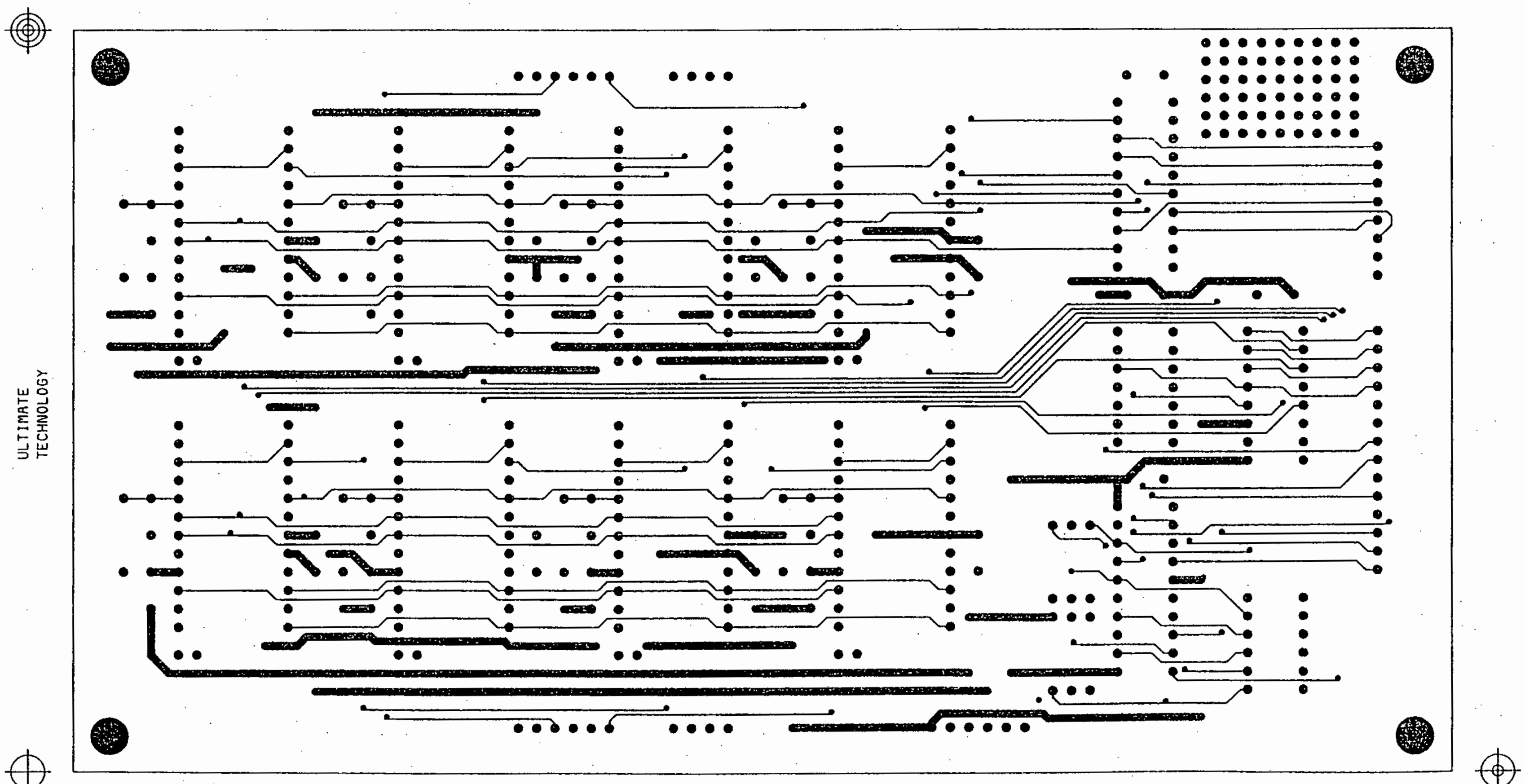


WPWELEC1.P1 (APR. 16. 1991) (10:45) (PCB) SCALE: 150% REFLECTED
DRILL REF PNT: 0.-625. -1.-425 (INCH)



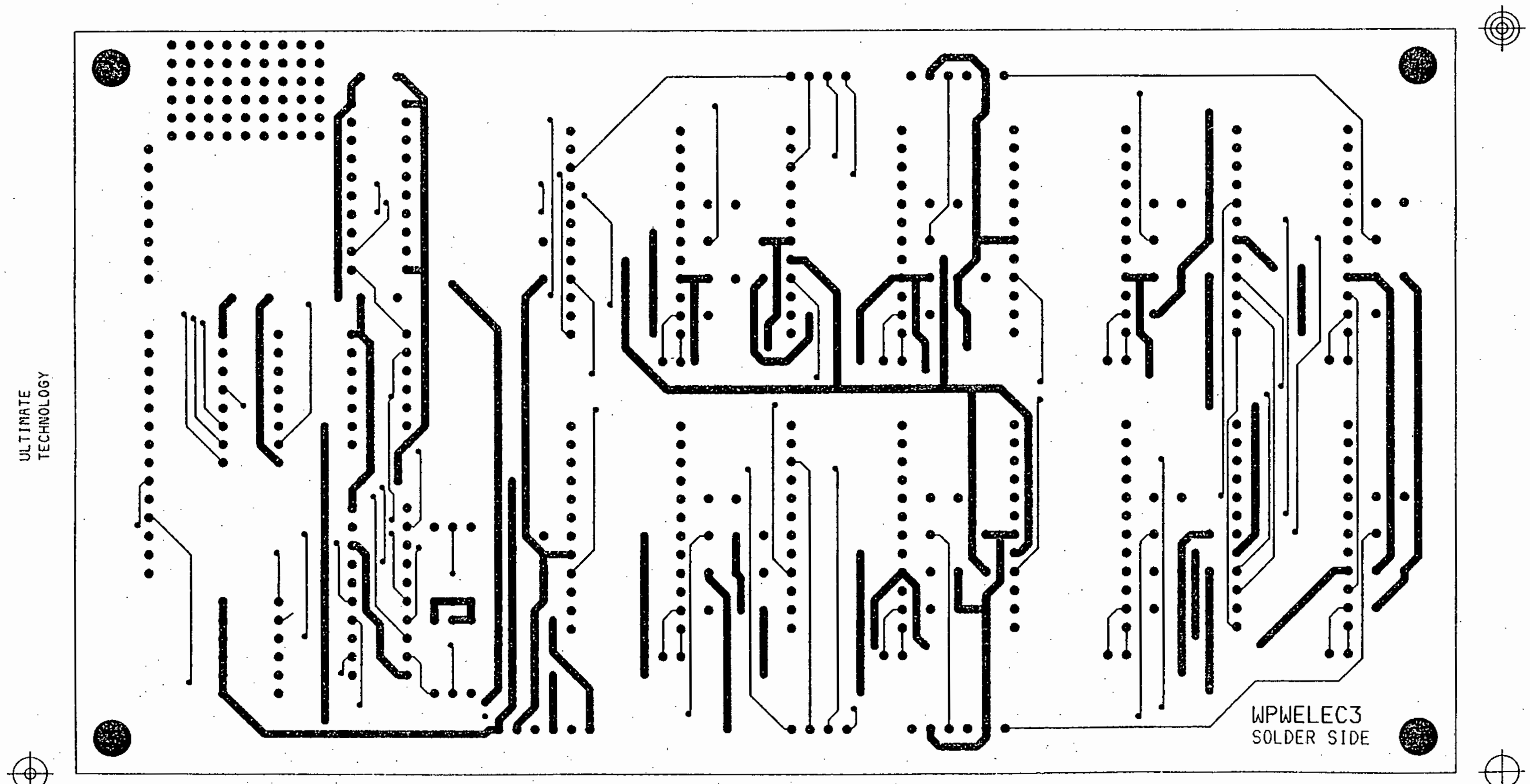
WPWELEC1.P3 (APR. 16, 1991) (10:48) (PCB) SCALE: 150%
DRILL REF PNT: 0.-625, -1.-425 (INCH)


PLACE DECOUPLING CAPACITORS ON ICS WHERE NOT SHOWN
HOLE DIA. 0.9mm
DATA D0-D7
rev: 1
date: 91-01-24
eng: WPWELEC3
Project: BOARD 2: ANALOGUE FILTERS - 8 CHANNELS
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 19 01 14
page: 1
of: 1

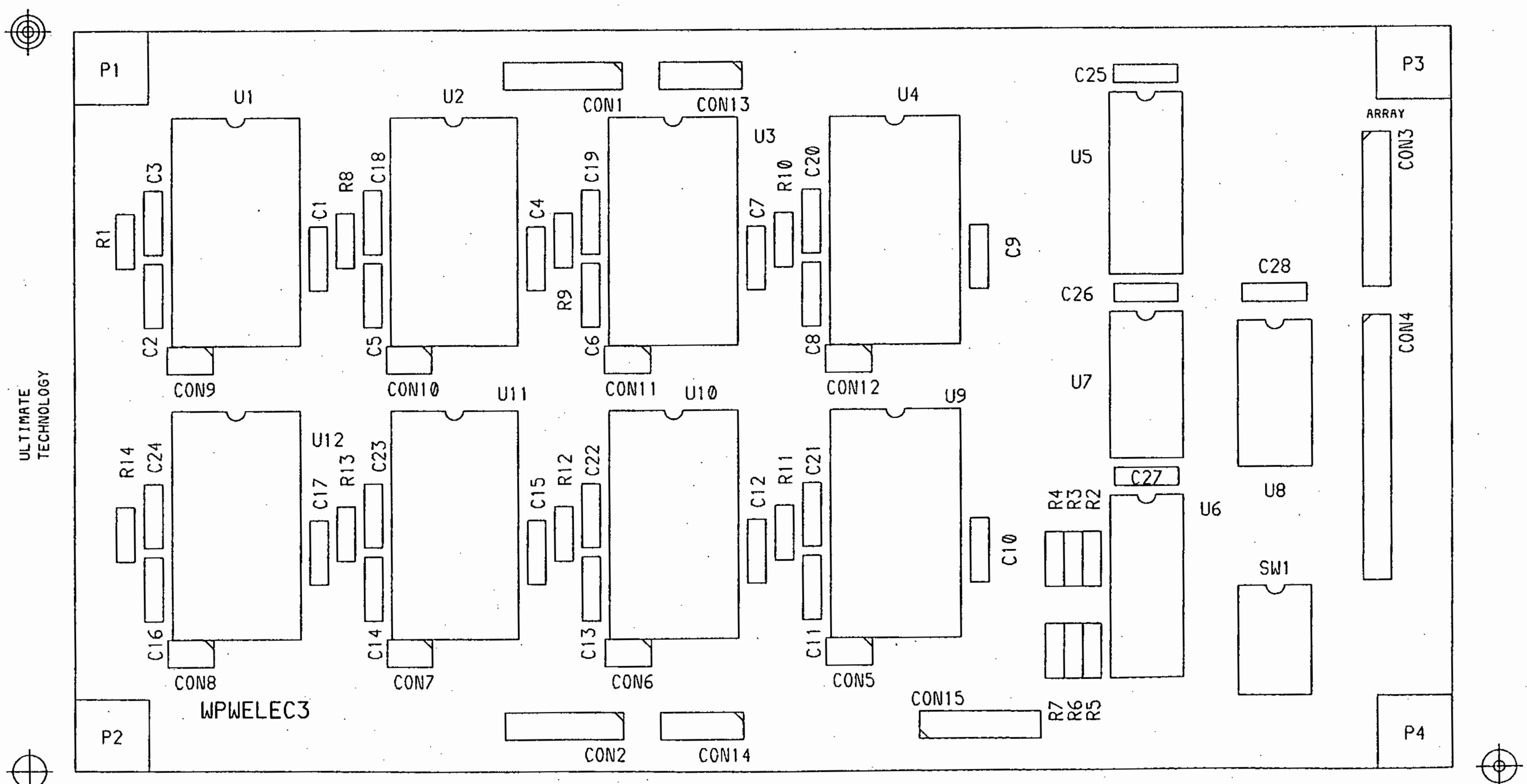


WPW3.P0 (APR. 23, 1991) (11:59) (WHYATT) SCALE: 150%
DRILL REF PNT: 0.000, 0.000 (INCH)

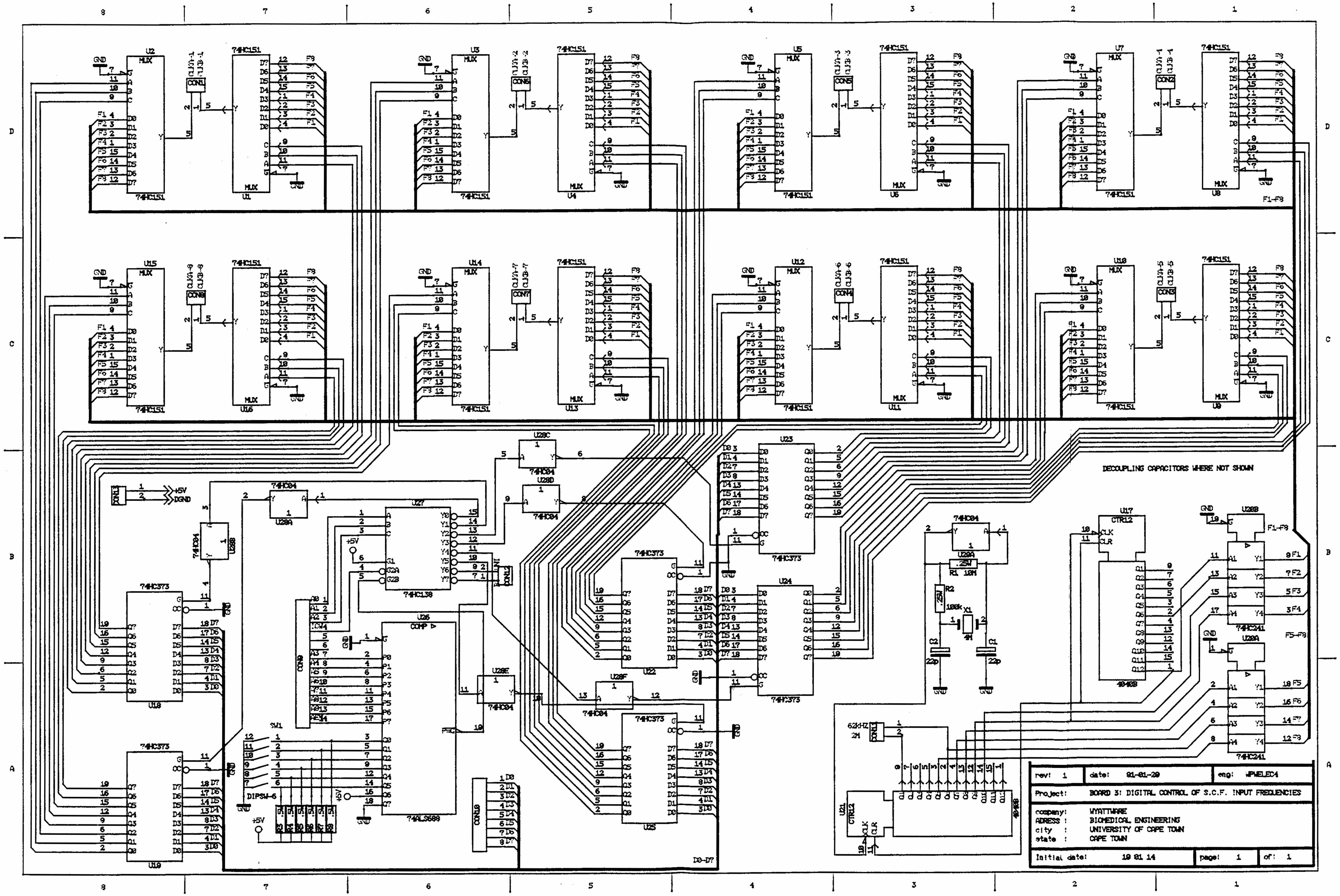
ULTIMATE
TECHNOLOGY

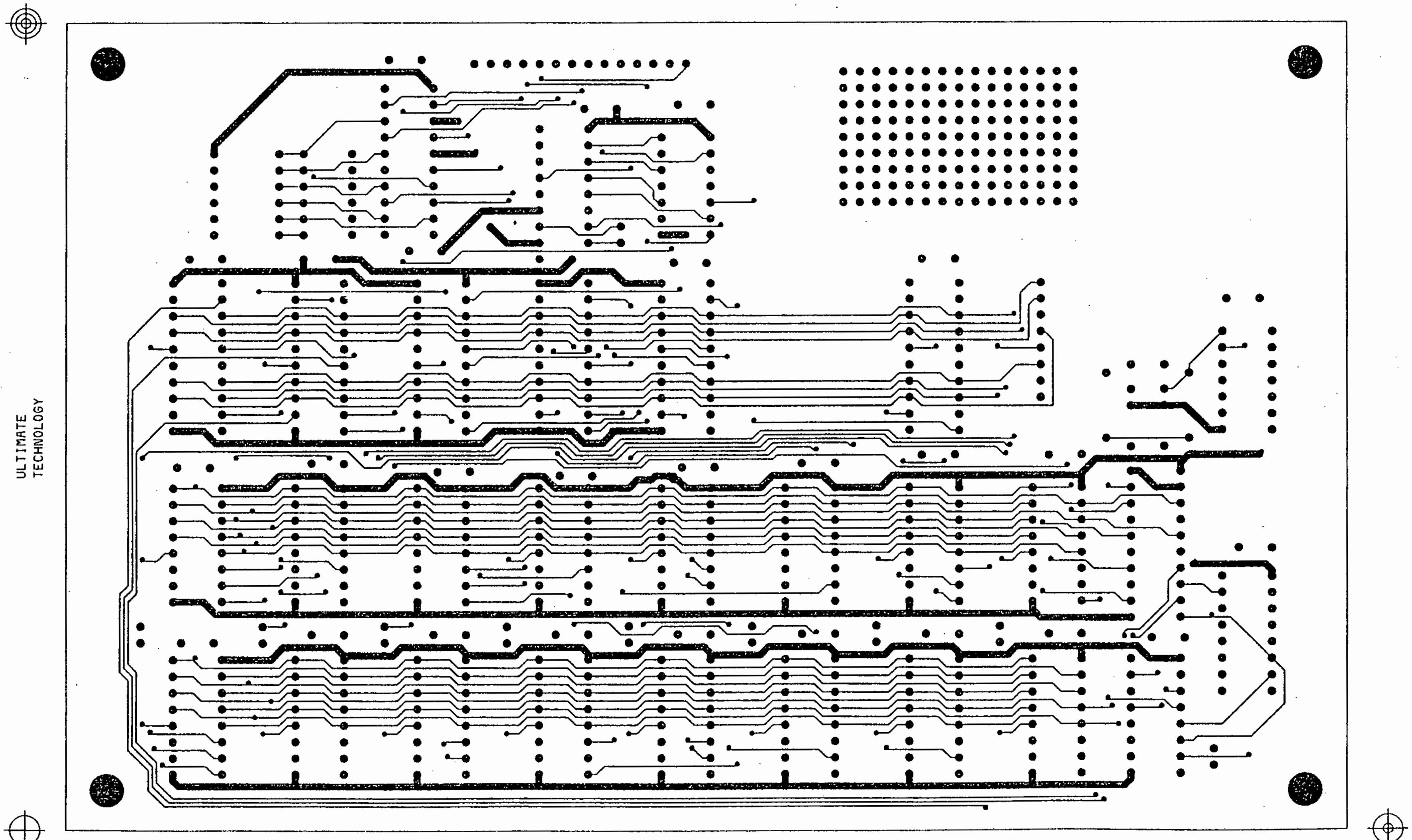


WPW3.P1 (APR. 23, 1991) (11:59) (WHYATT) SCALE: 150% REFLECTED
DRILL REF PNT: 0.-675, 0.-50 (INCH)

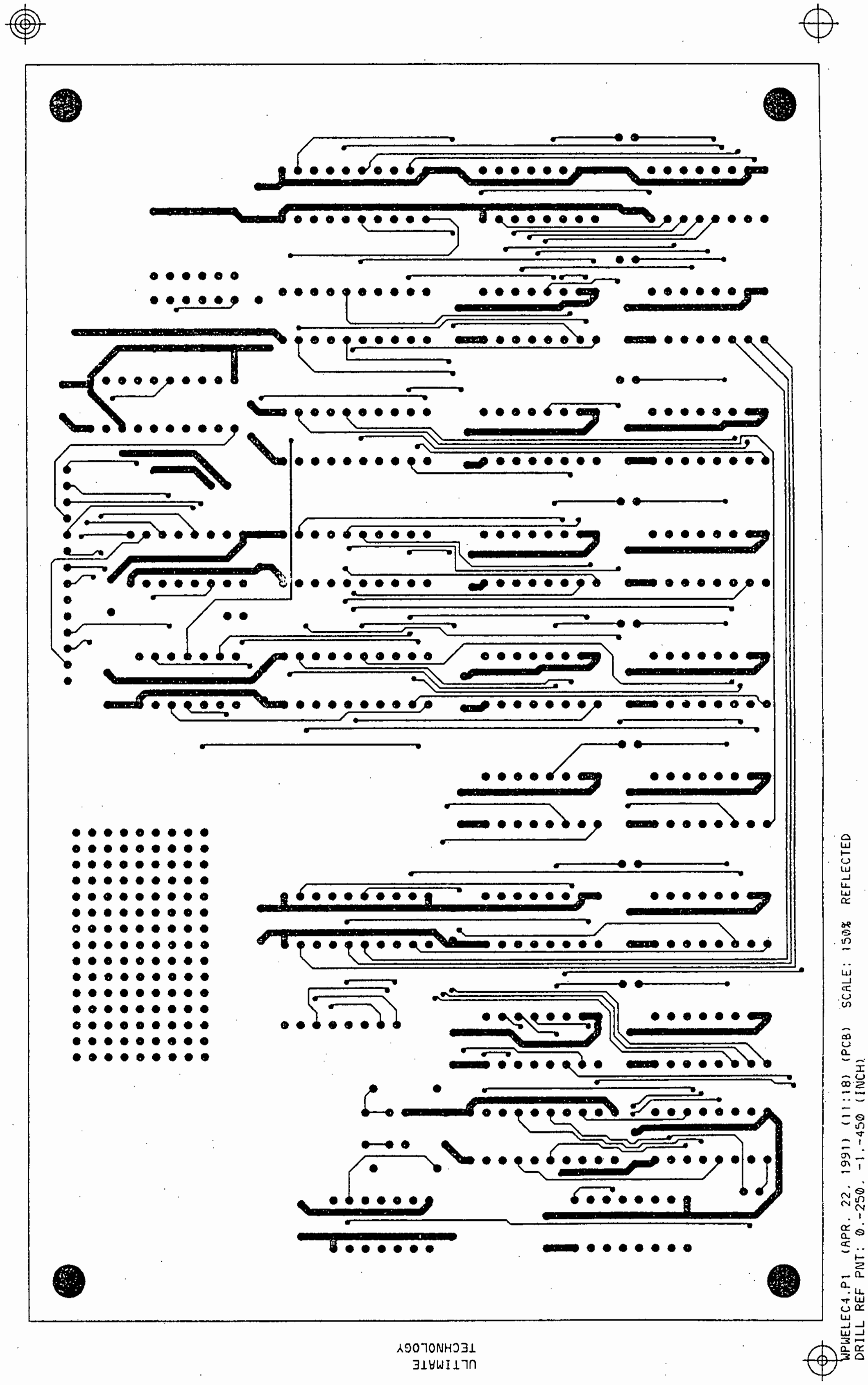


WPW3.P3 (APR. 23. 1991) (12:00) (WHYATT) SCALE: 150%
DRILL REF PNT: 0.-675. 0.-50 (INCH)


DECOUPLING CAPACITORS WHERE NOT SHOWN
rev: 1
date: 91-01-29
eng: WPWELEC4
Project: BOARD 3: DIGITAL CONTROL OF S.C.F. INPUT FREQUENCIES
company: WYATTWARE
adress : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 19 01 14
page: 1
of: 1

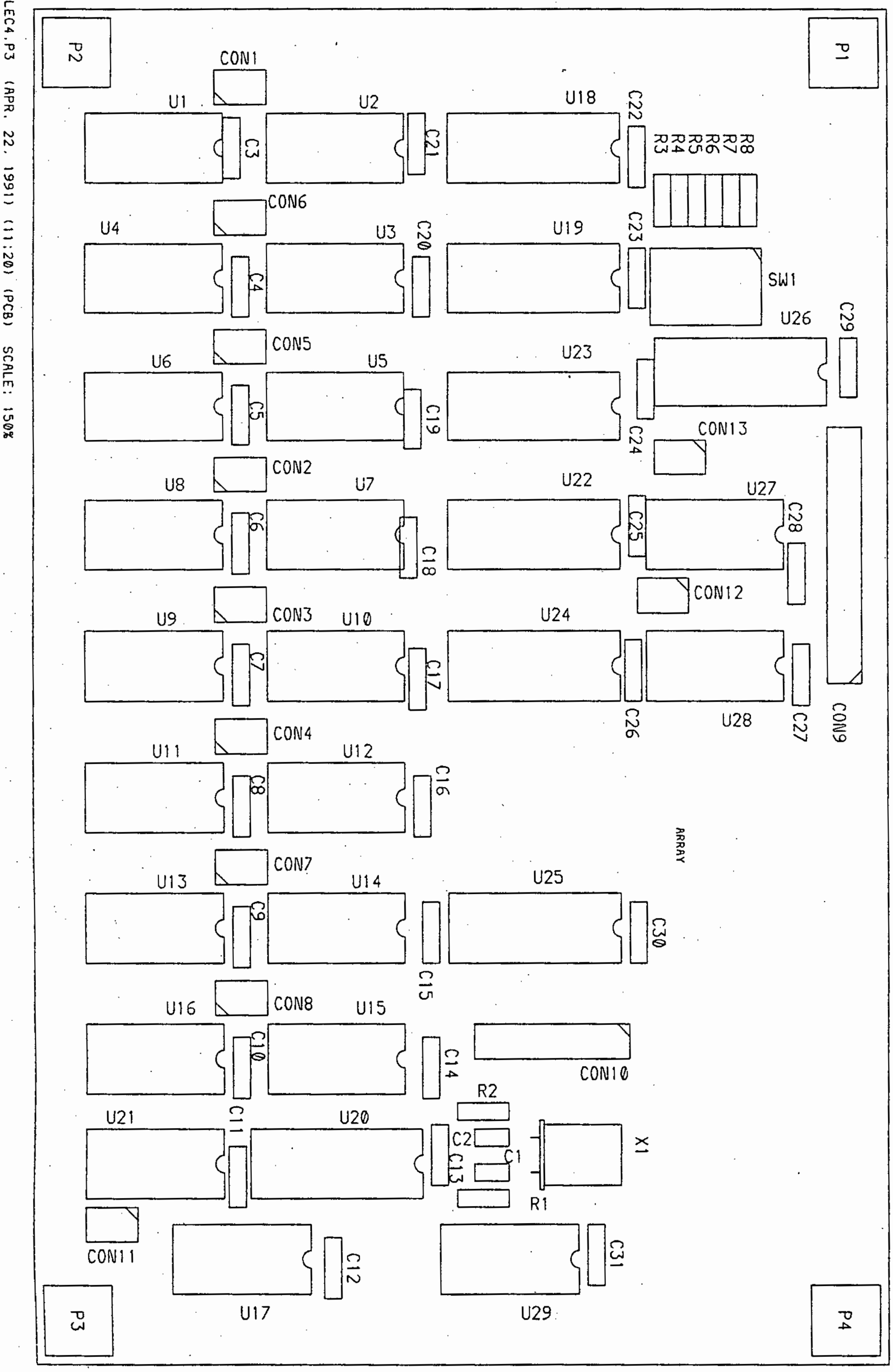


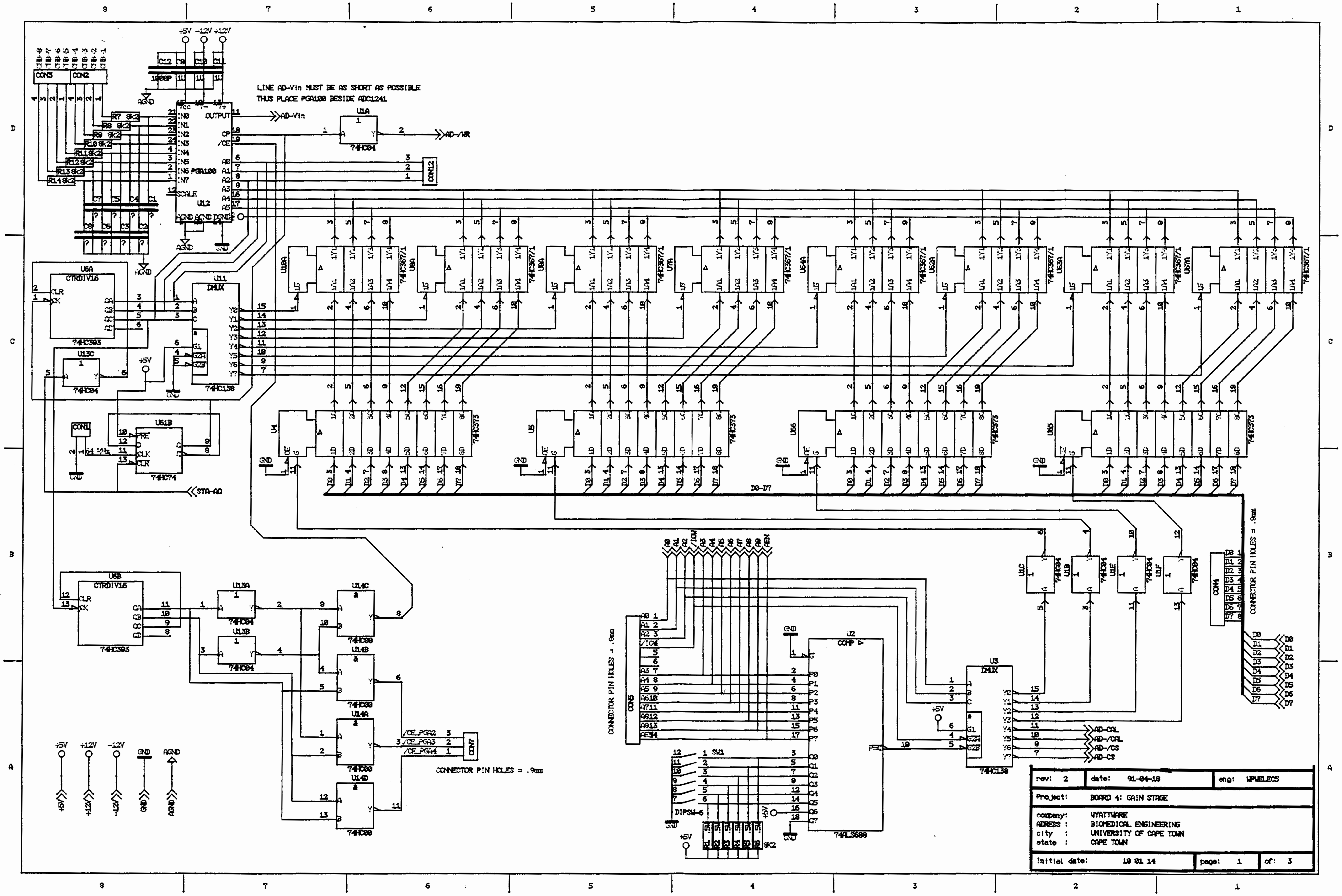
WPWELEC4.P0 (APR. 22, 1991) (11:17) (PCB) SCALE: 150%
DRILL REF PNT: 0.000, 0.000 (INCH)

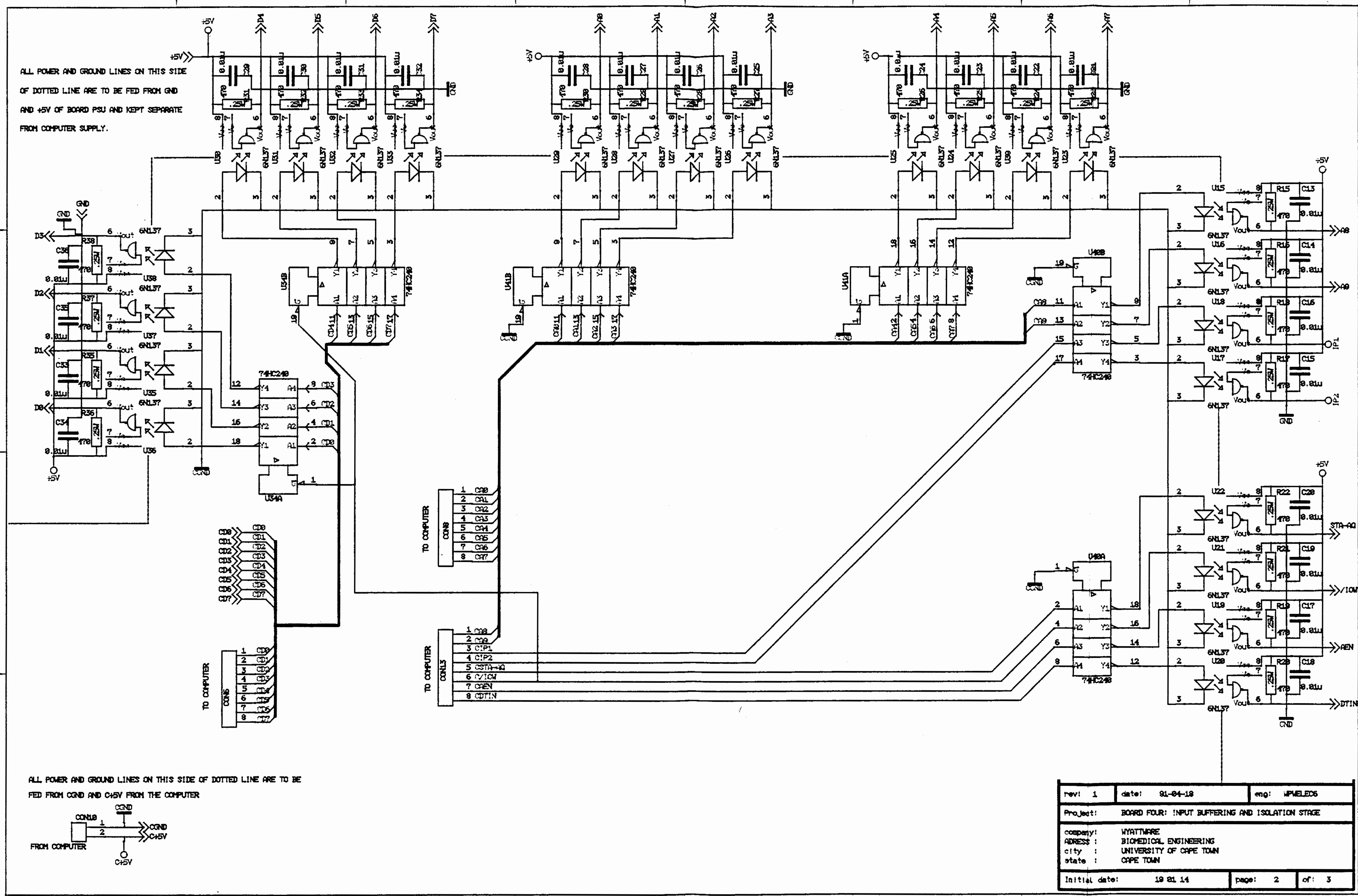

ULTIMATE
TECHNOLOGY
WPWELEC4.P1 (APR. 22, 1991) (11:18) (PCB) SCALE: 150% REFLECTED
DRILL REF PNT: 0.-250, -1.-450 (INCH)

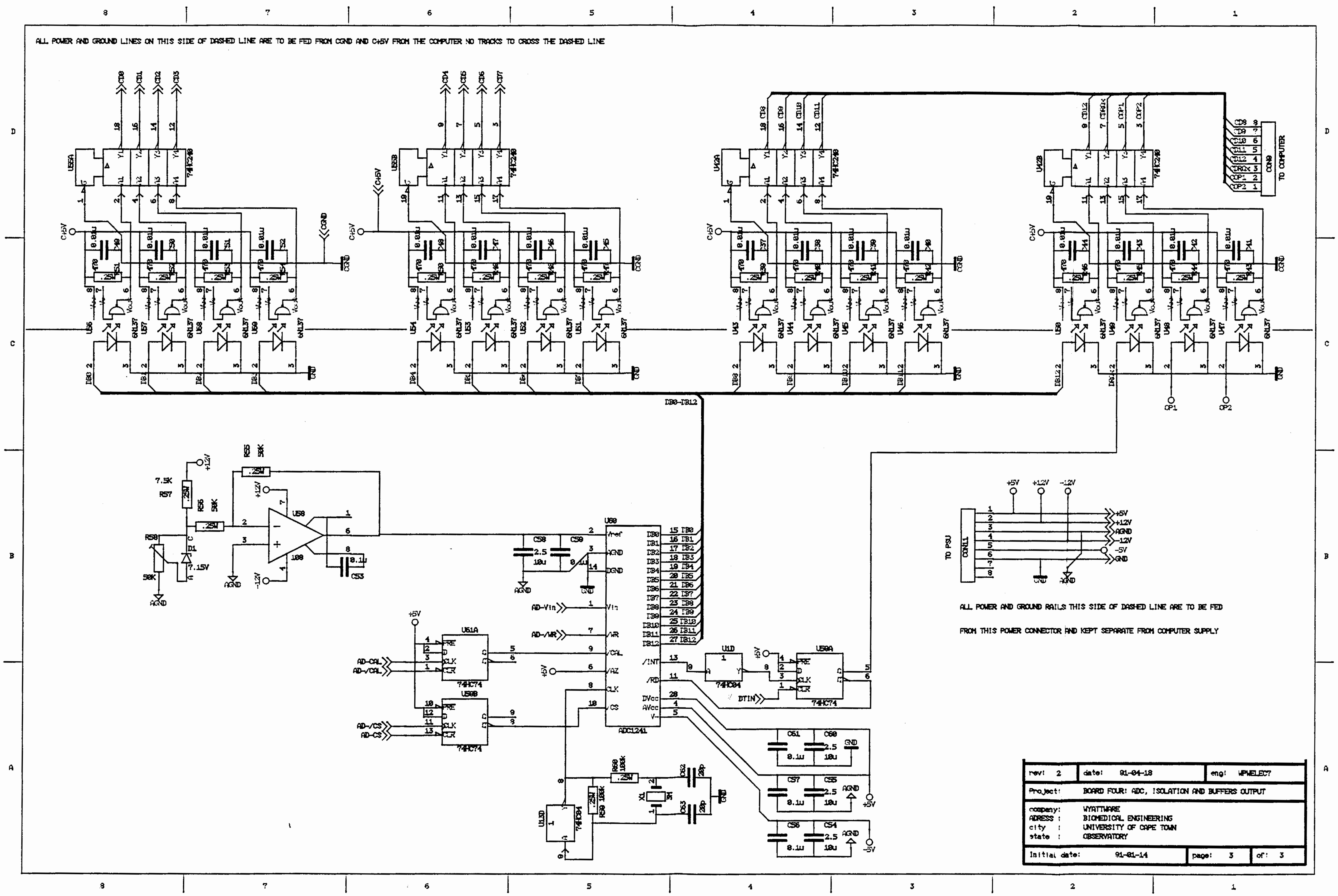
ULTIMATE
TECHNOLOGY

WPWELEC4.P3 (APR. 22, 1991) (11:20) (PCB) SCALE: 150%
DRILL REF PNT: 0.-250, -1.-450 (INCH)

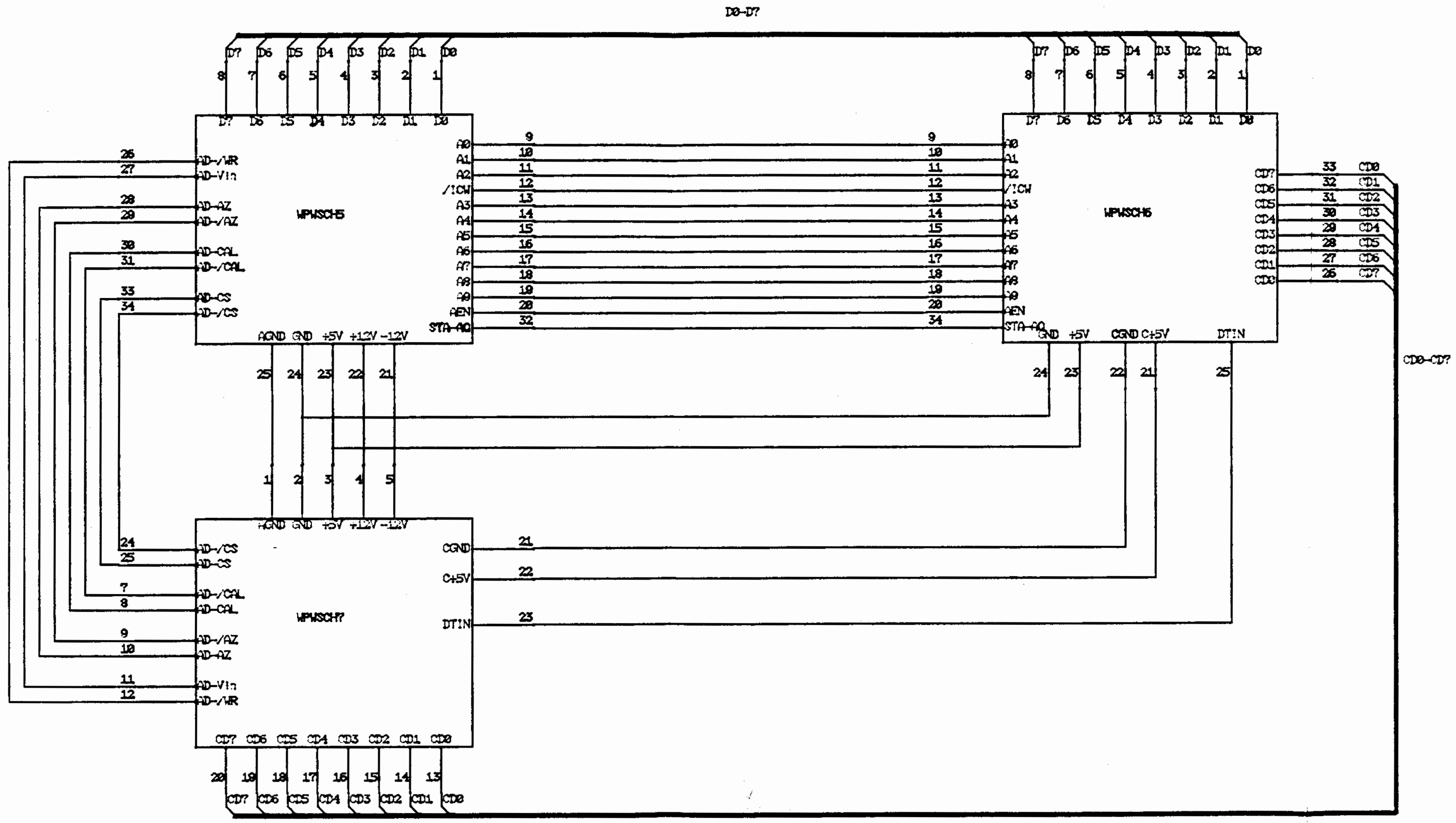



LINE AD-Vin MUST BE AS SHORT AS POSSIBLE
THUS PLACE PGA100 BESIDE ADC1241
CONNECTOR PIN HOLES = .9mm
rev: 2
date: 91-04-18
eng: WPWELECS
Project: BOARD 4: GAIN STAGE
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 19 01 14
page: 1
of: 3

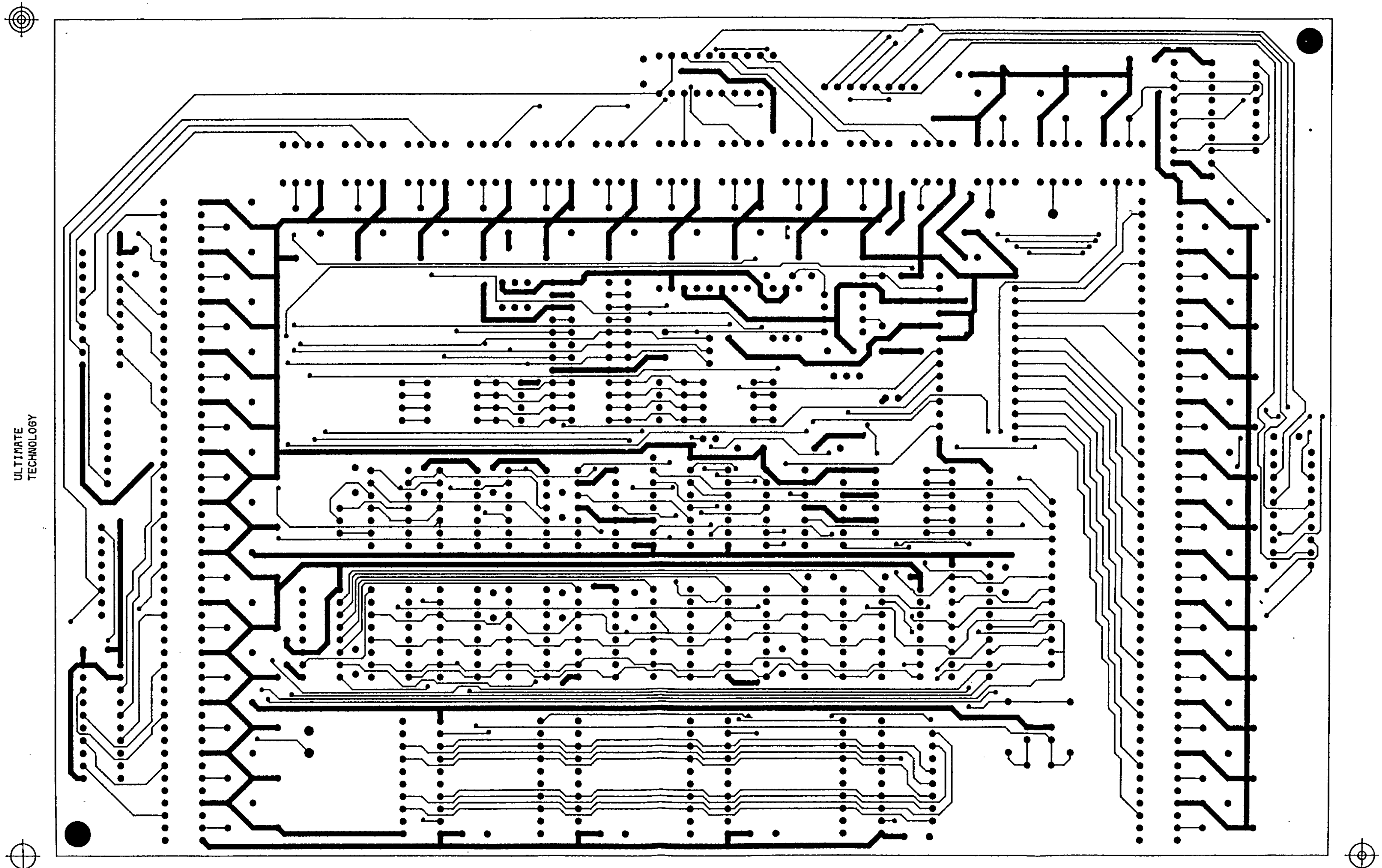

ALL POWER AND GROUND LINES ON THIS SIDE OF DOTTED LINE ARE TO BE FED FROM GND AND +5V OF BOARD PSU AND KEPT SEPARATE FROM COMPUTER SUPPLY.
ALL POWER AND GROUND LINES ON THIS SIDE OF DOTTED LINE ARE TO BE FED FROM CGND AND C+5V FROM THE COMPUTER
CON10
FROM COMPUTER
CGND
C+5V
TO COMPUTER
CON8
CON13
CON6
6N137
74HC240
rev: 1
date: 91-04-18
eng: WPWELEC6
Project: BOARD FOUR: INPUT BUFFERING AND ISOLATION STAGE
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 19 01 14
page: 2
of: 3



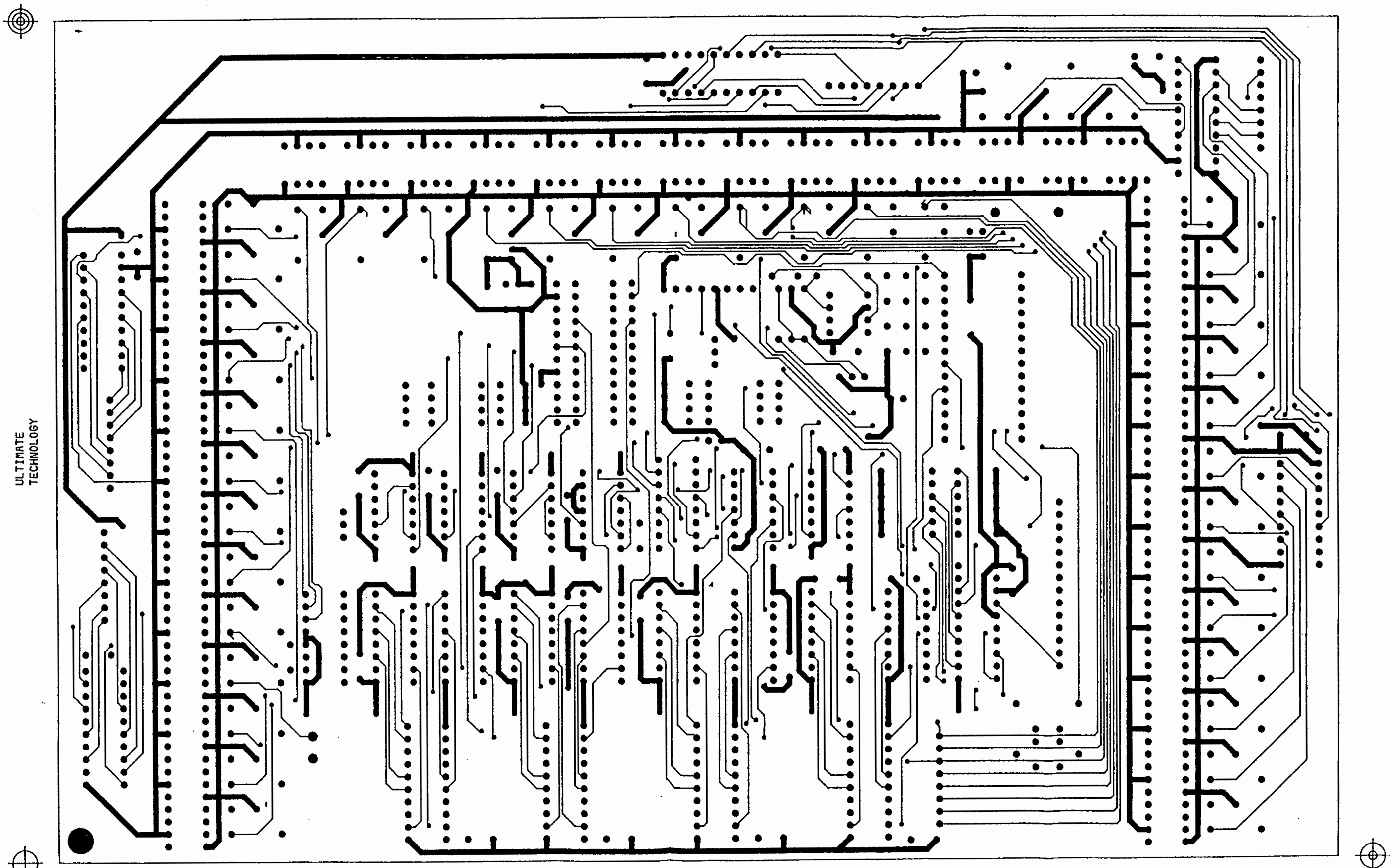
ALL POWER AND GROUND LINES ON THIS SIDE OF DASHED LINE ARE TO BE FED FROM CGND AND C+5V FROM THE COMPUTER NO TRACKS TO CROSS THE DASHED LINE
74HC240
6N137
DB0-DB12
TO COMPUTER
CON9
ADC1241
74HC74
74HC04
TO PSU
CON11
ALL POWER AND GROUND RAILS THIS SIDE OF DASHED LINE ARE TO BE FED
FROM THIS POWER CONNECTOR AND KEPT SEPARATE FROM COMPUTER SUPPLY
rev: 2
date: 91-04-18
eng: WPWELEC7
Project: BOARD FOUR: ADC, ISOLATION AND BUFFERS OUTPUT
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : OBSERVATORY
Initial date: 91-01-14
page: 3
of: 3



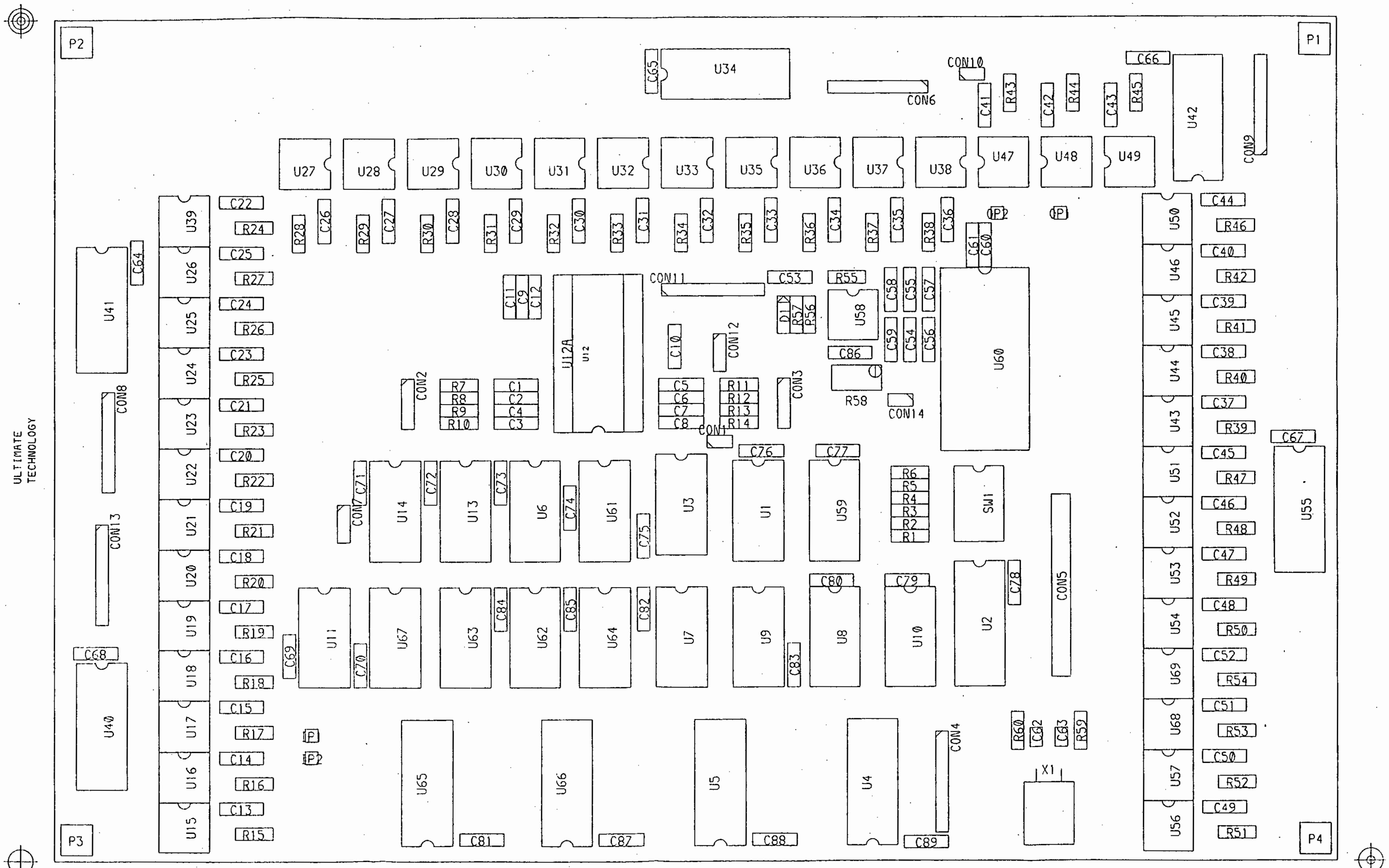
rev: 1	date: 91-03-19	eng: WPWELEC8
Project:	CONNECT BOARD FOR WPWELEC5, 6, 7.	
company:	WYATTWARE	
ADRESS :	BIOMEDICAL ENGINEERING	
city :	UNIVERSITY OF CAPE TOWN	
state :	OBSERVATORY	
Initial date: 91-01-14	page: 0	of: 3



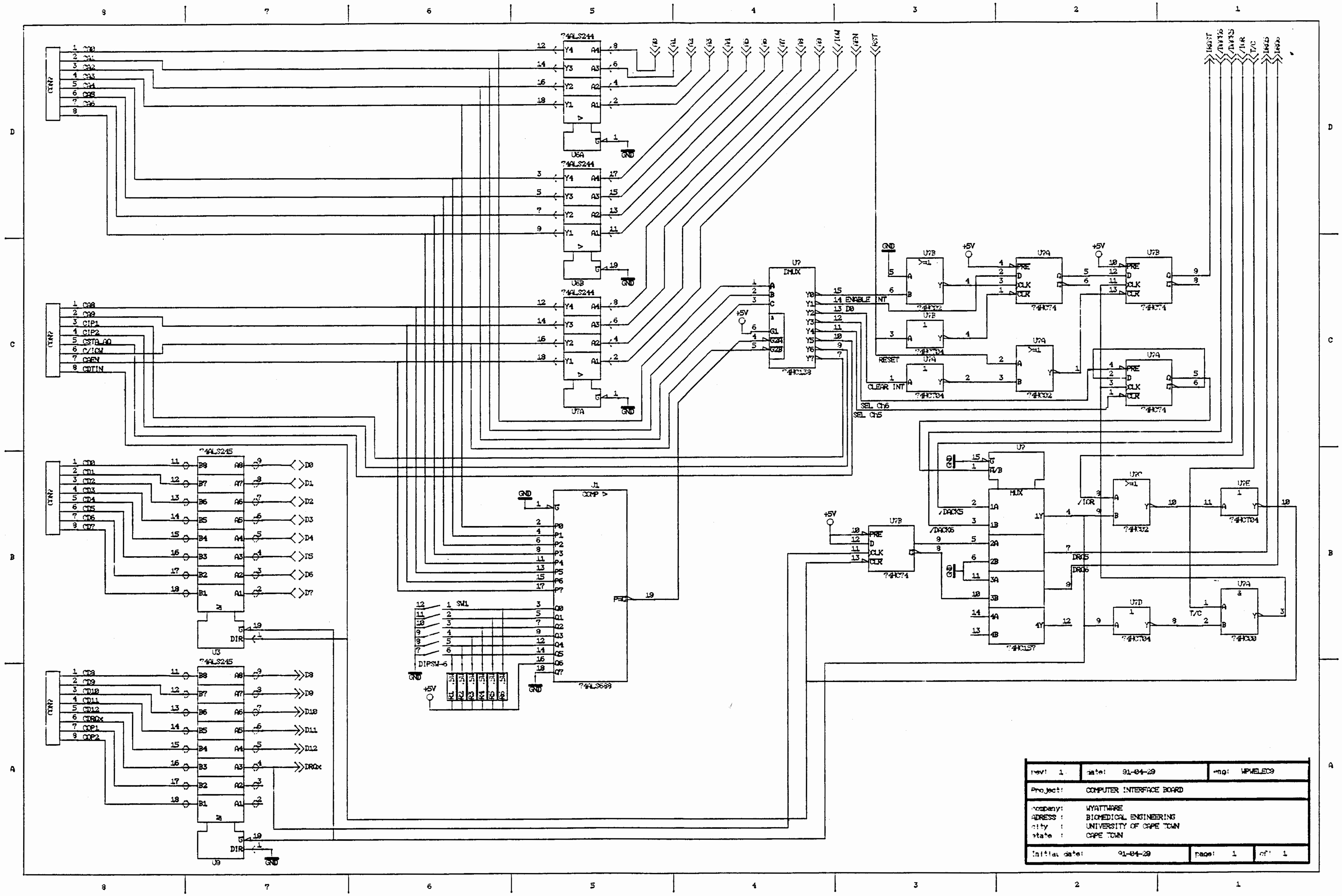
WPWSCH8.P0 (JUL. 26, 1991) (16:41) (PCB) SCALE: 140%
DRILL REF PNT: 0.000, 0.000 (INCH)



WPWSCH8.P1 (JUL. 26. 1991) (16:43) (PCB) SCALE: 140%
DRILL REF PNT: 0.000. 0.000 (INCH)



WPWSCH8.P3 (JUL. 26. 1991) (16:45) (PCB) SCALE: 140%
DRILL REF PNT: 0.000, 0.000 (INCH)


74ALS244
U6A
U6B
U7A
74ALS245
U3
U9
74HC138
74HC74
74HC32
74HC04
74HC157
74HC00
74ALS688
DIPSW-6
SW1
rev: 1
date: 91-04-29
eng: WPWELEC9
Project: COMPUTER INTERFACE BOARD
company: WYATTWARE
ADRESS : BIOMEDICAL ENGINEERING
city : UNIVERSITY OF CAPE TOWN
state : CAPE TOWN
Initial date: 91-04-29
page: 1
of: 1