بسم الله الرحمن الرحيم

Palestine polytechnic university

College of Engineering & Technology

Electrical and Computer Engineering Department

Graduation Project

Design and implementation synchronizing Defibrillator as training kit

Project Team

Maymoun O.Al-Mohtaseb **Mousa A.Al-Qawasmi**

Yassar H.Zeed

Project Supervisor
Eng. ABDALLAH ARMAN

Hebron - Palestine
june,2009

جامعة بوليتكنك فلسطين
الخليل – فلسطين
كلية الهندسة والتكنولوجيا
دائرة الهندسة الكهربائية والحاسوب

اسم المشروع

Design and implementation synchronizing Defibrillator as training kit

أسماء الطلبة

ميمون عبيدة محفوظ المحتسب   موسى أنور موسى القواسمي
يسار حسين علي زيد

بناء على نظام كلية الهندسة والتكنولوجيا وإشراف ومتابعة المشرف المباشر على المشروع وموافقة أعضاء اللجنة الممتحنة تم تقديم هذا المشروع إلى دائرة الهندسة الكهربائية والحاسوب, وذلك للوفاء بمتطلبات درجة البكالوريوس في الهندسة تخصص هندسة الأجهزة الطبية.

توقيع المشرف

....

توقيع اللجنة الممتحنة

....

توقيع رئيس الدائرة

....

Abstract

Heart diseases has become one of most common diseases in our time and as the time pass by, a major increase in people that suffer from heart diseases, like ventricular fibrillation, ventricular tachycardia, atrial fibrillation, and atrial flutter .

So we decided in this project to design a device which treat heart diseases ,this device is called "defibrillator" its used to apply a strong electrical shock to the heart muscle undergoing a fatal arrhythmia the shocks can restore normal heart rhythms before the malfunctioning heart suffers sudden cardiac arrest, a seizure than can lead to death within minutes.

In certain types of arrhythmia the patient's ventricles maintain their ability to pump blood, as evidenced by the existence of an *R* wave feature in the ECG waveform. These arrhythmias are also correctable by electrical shock to the heart, but it is necessary to avoid delivering the shock during the ventricles' refractory period (the *T* wave of the ECG waveform),. The shock is usually time to occur approximately 30 microsecond after the *R* wave peak. , Human operator cannot be trusted to time the ECG waveform properly to void this problem. so the automatic electronic circuit is used A machine equipped with synchronized circuit is called a cardioverter.

This device will design it into a training kit, in order to consummate advantage for student's , by simplifying this device into main parts and presentment the job of each part equally.

الملخص

تعد الأمراض التي تصيب عضلة القلب من أكثر الأمراض شيوعا في وقتنا الحالي, مع الوقت أصبحت هذه الأمراض تشيع شيئا فشيئا بين الناس فملايين الناس تموت سنويا جراء هذه الأمراض ,و من هذه الأمراض حدوث ارتعاش في البطين أو الاذين او حدوث تسارع أو تباطأ في نبضات القلب نتيجة لحدوث مشكلة في كهربائية القلب الذاتية, هذه الأمراض في حال عدم معالجتها قد تودي الى موت الإنسان خلال دقائق قليلة.

لذلك قررنا في المشروع تصميم جهاز يدعى (جهاز الصدمات الكهربائية) يقوم بتزويد عضلة القلب بصدمة كهربائية إذا عانت عضلة القلب من أي من هذه الأمراض حتى يتم استعادة عملها بصورة صحيحة, الجهاز سيعتمد على اشاره تخطيط القلب (ECG) في تحدد الفترة الزمنية التي ستعطى فيها الصدمة الكهربائية لعضلة القلب , حيث سيقوم الجهاز بعملية استشعار لموجة(R) في تخطيط القلب واعطاء الصدمة الكهربائية بعد ظهور هذه الموجة بفترة قصيرة جدا حتى يتم استعادة عمل القلب بالصورة الصحيحة ويسمى هذا النوع بجهاز الصدمات الكهربائية المتزامن.(Synchronizing defibrillator).

حيث سنقوم من خلال المشروع بتصميم هذا الجهاز على شكل نموذج تعليمي مبسط يتم فيه تفصيل كل جزء من هذا الجهاز على حدة , ليتم تحقيق الفائدة لطلبة تخصص هندسة الأجهزة الطبية في إمكانية التعرف على أهم أجزاء هذا الجهاز وعلى مبدأ عمل كل جزء بصورة مبسطة.

Table of contents

List of Figures

Chapter one

Introduction

1.1 Introduction :

A Defibrillator is Device that used to apply a strong electrical shock to the heart muscle undergoing a fatal arrhythmia, the shocks can restore normal heart rhythms before the malfunctioning heart suffers sudden cardiac arrest, a seizure than can lead to death within minutes.

These arrhythmias are also correctable by electrical shock to the heart, but it is necessary to avoid delivering the shock during the ventricles' refractory period (the *T* wave of the ECG waveform), or the shock intended to correct the problem will create a much more serious arrhythmia such as ventricular fibrillation. The shock is usually time to occur approximately 30 microsecond after the *R* wave peak .

Human operator cannot be trusted to time the ECG waveform properly to void this problem.so the automatic electronic circuit is used A machine equipped with synchronized circuit is called a synchronized defibrillator.

This project has the purpose to design all circuits for synchronized defibrillator and ECG device .

1.2 Project objectives:

The main objectives of this project are :

- To study the physiology of the Heart and the disease .
- Design and implementation synchronizing defibrillator as training kit .
- Design and implementation ECG device.

1.3 What is the importance of the project ?

It is undeniable that nowadays people are more aware of the health conditions. One of the most widely used methods to support life .

Our project is very important, because it add a new technique in biomedical lab in Palestine polytechnic university .

The importance of this device comes from the following :

- Simple to use .
- Non invasive technique .

1.4 List of Abbreviations:

ECG	ElectroCardioGram
AC	Alternating Curret
DC	Direct Curret
SA nodes	Sinoatrial node
AV node	Atrioventricular node
R	Resister
C	Capacitor
W	Energy
VF	Ventricular Fibrillation
VT	Ventricular Tachycardia
SVT	Super Ventricular Tachycardia
AHA	American Heart Association
BTE	Biphasic Turncated Exponential
ICD	Implantable Cardioverter Defibrillator
AED	Automated External Defibrillator
ACLS	Advanced Cardiac Life Support
EMT	Emergency Medical Technician
LCD	Liquid Crystal Display
SLA	Sealed Lead Acid

1.5 Time Plan:

Time (Weeks)	Activates
5	Study Defibrillator physiology
11	Study defibrillator measurements and componants
13	Literature Review
14	Design Theory
15	Theoretical report Ready
16	Discussion for Project Theory
19	Design the Schematic blovk diagram
22	Peurchase Electronic Component for Project
30	Design Hardware System
31	Testing System Performance

1.6 Economical study :

Component	Quantity	Cost (sheqalim)
Solder board	1	60
Lm311	1	10
NE555N	2	20
IRF9532	1	15
IRF540	1	15
Capacitors	15	30
Resistors	35	40
Zener diode	2	7
Tramsistors	7	30
Personal		350
Instrumentation Amplifier AD620		40
Lm358	1	5
Power supply		70
Total		700

1.7 Project content :

Our report divided into ten section ; these sections are described as follows:

Section One : Introduction .
Section Tow : Definition and History of Defibrillator.
Section Three : Heart and Arrhythmias .
Section Four : Defibrillator Modes
Section Five : Defibrillator waveform and energy .
Section Six : Uses of Defibrillator
Section Seven : Types of Defibrillator .
Section Eight: Testing Defibrillator.
Section Nine : Block Diagram and Circuit analysis.
Section Ten : Testing and implementation.
Section Eleven : Conclusion and Future works .

Chapter two

Definition and History of Defibrillator

2.1 Definition:

Device that used to apply a strong electrical shock to the heart muscle undergoing a fatal arrhythmia. The shock changes ventricular fibrillation to an organized ventricular rhythm or changes a very rapid and ineffective cardiac rhythm to slower, more effective rhythm., the shocks can restore normal heart rhythms before the malfunctioning heart suffers sudden cardiac arrest, a seizure than can lead to death within minutes.

2.2 History:

Defibrillation was first demonstrated in 1899 by Prevost and Batelli, two physiologists from University of Geneva, Switzerland. They discovered that small electric shocks could induce ventricular fibrillation in dogs, and that larger charges would reverse the condition

The first use on a human was in 1947 by Claude Beck professor of surgery at Case Western Reserve University. Beck's theory was that ventricular fibrillation often occurred in hearts which were fundamentally healthy, in his terms "Heart too good to die", and that there must be a way of saving them. Beck first used the technique successfully on a 14 year old boy who was being operated on for a congenital chest defect. The boy's chest was surgically opened, and manual cardiac massage was undertaken for 45 minutes until the arrival of the defibrillator. Beck used internal paddles on either side of the heart, along with procaine amide, a heart drug, and achieved return of normal sinus rhythm.

These early defibrillators used the alternating current from a power socket, transformed from the 110-240 volts available in the line, up to between 300 and 1000 volts, to the exposed heart by way of 'paddle' type electrodes. The technique was often

ineffective in reverting VF while morphological studies showed damage to the cells of the heart muscle post mortem. The nature of the AC machine with a large transformer also made these units very hard to transport, and they tended to be large units on wheels
Before 1960the defibrillator was ac models applied 5 to 6 A of 60Hz ac across the patient's chest for 250 to 1000 ms. The success rate for ac defibrillators was useless for correction a trial fibrillation.

Until the early 1950s, defibrillation of the heart was possible only when the chest cavity was open during surgery. The technique used an alternating current from a 300 or greater volt source delivered to the sides of the exposed heart by 'paddle' electrodes here each electrode was a flat or slightly concave metal plate of about 40 mm diameter.

The closed-chest defibrillator device which applied an alternating current of greater than 1000 volts, conducted by means of externally applied electrodes through the chest cage to the heart, was pioneered by Dr V. Eskin with assistance by A. Klimov in Frunze, USSR in mid 1950s.

In 1959 Bernard Lawn commenced research into an alternative technique which involved charging of a bank of capacitors to approximately 1000 volts with an energy content of 100-200 joules then delivering the charge through an inductance such as to produce a heavily damped sinusoidal wave of finite duration (~5 milliseconds) to the heart by way of 'paddle' electrodes. The work of Lown was taken to clinical application by engineer Barouh Berkovits with his "cardioverter".

Since 1960, several different DC defibrillators have been devised. These defibrillators store a DC charge that can be delivered to the patient.

1962,the basic circuit as I will illustrate posterior was described first by Gurvich and Yuniech and was later popularized by Dr.Bernard Lown of Harvard University, who use the device clinically.

The damped sine wave defibrillator replaced the simpler alternating current because it produced fewer postshock arrhythmias and less tissue damage than the other types of ac defibrillator. Produce many defibrillator with different wave shape. and this waveform called Lown waveform and I will illustrate it posterior, and after that the developed at defibrillator in waveform by using the basic circuit and remove or add component, the most common forms of developed wave forms are (Lown, monopulse, tapered dc delay, and trapezoidal waveforms).

Chapter Three

Heart and Arrhythmias

3.1 Introduction to Heart:

Heart is a cardiac muscle that contract to pump the blood through all parts of body, by the circulatory system. So we can see in fig(3.1). that the human heart is divided internally into four chambers, Two upper chambers, (the right and left atria) received blood as it returns to the heart. These atria transfer blood to two lower chambers,(the right and left ventricles); which pump blood into major arteries of the circulatory system. The heart takes its untrue from coronary artery.

3.2 Heart Potential

Your muscles only work when they're sparked by an electric signal. For most muscles, the command comes from the brain via nerves. Your heart, though, supplies its own spark. Special areas in the heart called pacemakers generate electric charges every second that trigger your heart muscle to beat.

The heart pumps blood to the lungs and to all the body's tissues by a sequence of highly organized contractions of its four chambers. For the heart to function properly, the four chambers must beat in an organized manner. This is governed by the electrical impulse. A chamber of the heart contracts when an electrical impulse or signal moves across it. Such a signal starts in a small bundle of highly specialized cells located in Figure3.2

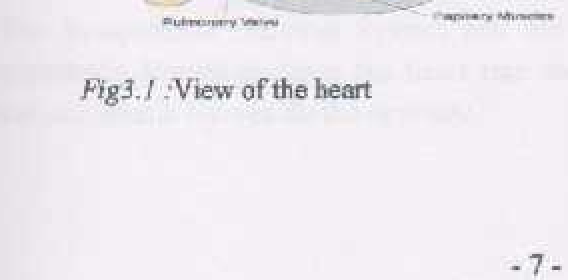


Fig3.1 :View of the heart

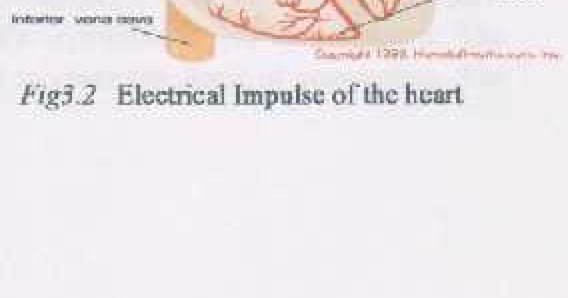


Fig3.2 Electrical Impulse of the heart

Electrical Impulse of the heart the right atrium — the sinoatrial node (SA node), also called the sinus node. A discharge from this natural "pacemaker" causes the heart to beat. This pacemaker generates electrical impulses at a given rate, but emotional reactions and hormonal factors can affect its rate of discharge. This allows the heart rate to respond to varying demands.

A special group of cells that have the ability to generate electrical activity on their own. These cells separate charged particles. Then they spontaneously leak certain charged particles into the cells. This produces electrical impulses in the pacemaker cells which spread over the heart, causing it to contract. These cells do this more than once per second to produce a normal heart beat of 72 beats per minute. The natural pacemaker of the heart is called the Sinoatrial node (SA node). It is located in the Right Atrium. The heart also contains specialized fibers that conduct the electrical impulse from the pacemaker (SA node) to the rest of the heart (see Figure3.2) .

The electrical impulse leaves the SA node (1) and travels to the right and left Atria, causing them to contract together. This takes .04 seconds. There is now a natural delay to allow the Atria to contract and the Ventricles to fill up with blood. The electrical impulse has now traveled to the Atrioventricular Node (AV node) (2). The electrical impulse now goes to the Bundle of His (3), then it divides into the Right and Left Bundle Branches (4) where it rapidly spreads using Purkinje Fibers (5) to the muscles of the Right and Left Ventricle, causing them to contract at the same time.

Any of the electrical tissue in the heart has the ability to be a pacemaker. However, the SA node generates an electric impulse faster than the other tissue so it is normally in control. If the SA node should fail, the other parts of the electrical system can take over, although usually at a slower rate. Although the pacemaker cells create the electrical impulse that causes the heart to beat, other nerves can change the rate at which the pacemaker cells fire and the how strongly the heart contracts. These nerves are part of the Autonomic Nervous System. The Autonomic Nervous System has 2 parts - The Sympathetic Nervous System and the Parasympathetic Nervous System. The Sympathetic Nerves increase the heart rate and increase the force of contraction. The Parasympathetic Nerves do the opposite.

3.3 Normal heart rate or pulse:

During each heartbeat, the muscles of the heart contract causing a wave of pressure which forces blood through the arteries. This wave of pressure is known as a pulse. There is one pulsation for each heartbeat. The pulse can be felt at various points on the body where the arteries are just under the skin, such as the temples, neck, crook of the elbow, wrist, groin, back of the knee, and the inside back of the ankle. The normal pulse rate varies with age. Below is a chart listing the range of heart rates and average heart rate for various ages With exercise or physical activity, the heart rate increases to supply the muscles with more oxygen to produce extra energy. The heart can beat up to 200 times per minute with extreme exercise. The brain sends nerve signals to the heart to control the rate. The body also produces chemical hormones, such as adrenaline, which can change the heart rate. When we are excited, scared, or anxious our heart gets a signal to beat faster. During a fever, the heart beats faster to bring more blood to the surface of the body to release heat and cool the body. The heart rate increases during and after a meal to send more blood to the digestive system.

The heart contracts (beats) as the electrical impulse moves through it. This normally occurs 60 to 80 times a minute when a person is at rest. The atria contract a split-second before the ventricles. This lets the atria empty their blood into the ventricles before the ventricles contract.

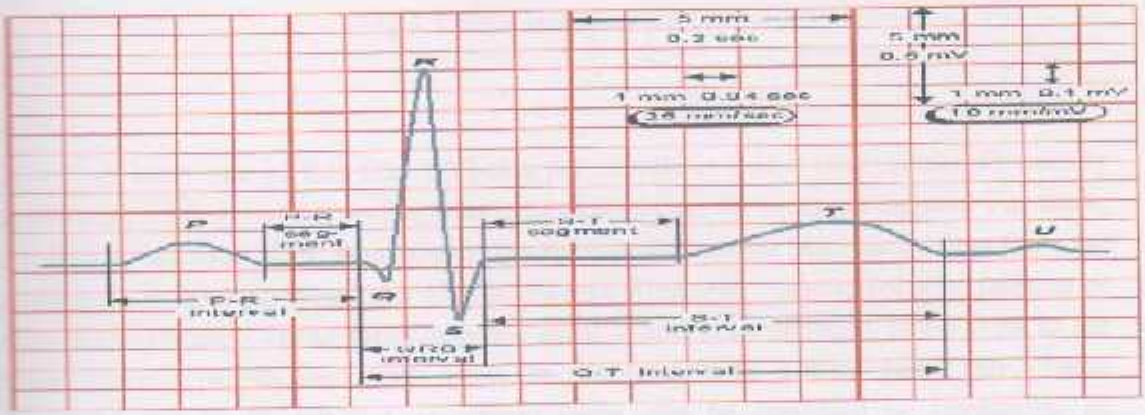


Fig3.3 :Normal ECG wave form .

3.4 Heart block and Arrhythmias

Arrhythmias:

Arrhythmias (or dysrhythmias) are problems that affect the electrical system of the heart muscle, producing abnormal heart rhythms. They can cause the heart to pump less effectively.

Many arrhythmias have no known cause. However, a number of factors can contribute to arrhythmias. They include coronary artery disease, high blood pressure, diabetes, smoking, excessive use of alcohol or caffeine, drug abuse and stress. Certain substances, including some over-the-counter and prescription medications, dietary supplements and herbal remedies are known to cause arrhythmias in some people.

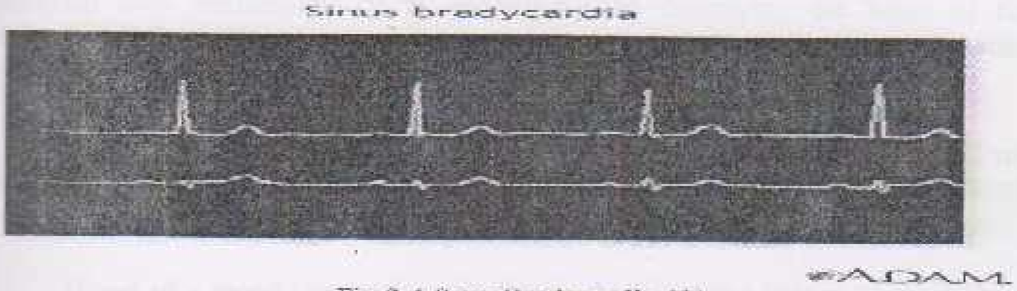


Fig 3.4:Snus Bradycardia {1}.

A heart rate of less than 60 beats per minute is called bradycardia. Physically active people often have a resting heart rate slower than 60 beats per minute. For them, this isn't abnormally slow. Bradycardia doesn't usually require treatment unless there are prolonged or repeated symptoms such as fatigue, dizziness, lightheadedness, fainting or near-fainting spells. These symptoms are usually due to insufficient blood flow to the brain. Elderly people are more prone to problems with a slow heart rate. These symptoms can be treated and, in certain cases, can be corrected by implanting an electronic pacemaker under the skin to speed up the heart rhythm, when the heart rate becomes too slow.

Tachycardia:

Rapid heart beating, called tachycardia or tachyarrhythmia, can produce palpitations, rapid heart action, chest pain, dizziness, lightheadedness, fainting or near fainting if the heart beats too fast to circulate blood effectively. Heartbeats may be either regular or irregular in rhythm.

When rapid heart beating starts in the ventricles — called ventricular tachycardia — it can interfere with the heart's ability to pump enough blood to the brain and other vital organs. This dangerous arrhythmia can change without warning into the most serious heart rhythm disturbance — ventricular fibrillation. In this, the lower chambers quiver and the heart can't pump any blood. Collapse and sudden cardiac death follow unless medical help is provided immediately.

If treated in time, ventricular tachycardia and ventricular fibrillation can be converted into normal rhythm with electrical shock. Rapid heart beating can be controlled with medications or by identifying and destroying the focus of rhythm disturbances. One effective way to correct these life-threatening rhythms is by using an electronic device called an implantable cardioverter / defibrillator.

Blood clots can form during atrial fibrillation, a disorder found in 2.2 million Americans. In atrial fibrillation the atria quiver instead of beating effectively. Blood isn't pumped completely out of them when the heart beats, so the blood pools and clots.

If part of a blood clot in the atria leaves the heart and lodges in an artery in the brain (or leading to it), a stroke results. About 15 percent of strokes occur in people with atrial fibrillation.

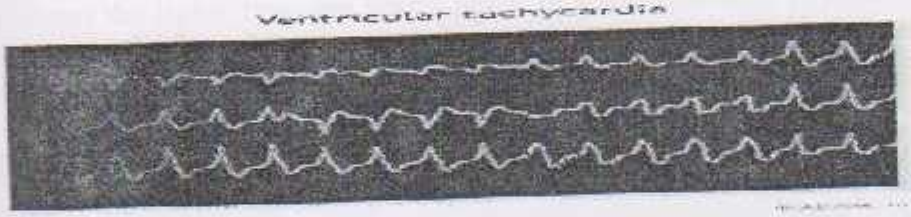
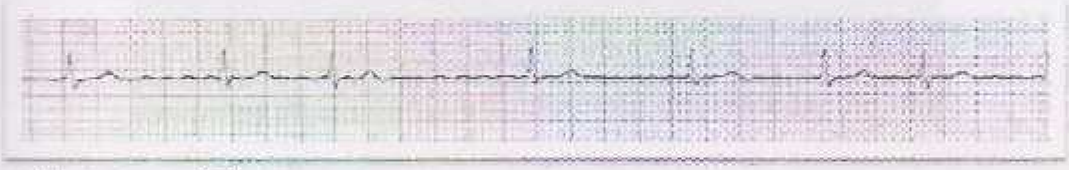


Fig 3.5 Ventricular tachycardia .{2}

Fibrillation:

Is the rapid, irregular, and unsynchronized contraction of the muscle fibers of the heart. There are two major classes of fibrillation, atrial fibrillation and ventricular fibrillation.

Atrial fibrillation can be a chronic condition, usually treated with anticoagulation and sometimes with conversion to normal sinus rhythm. In atrial fibrillation, the heart's upper chambers quiver rapidly - or "fibrillate" - at rates of approximately 400/minute. This chaotic electrical rhythm, although confined to the heart's upper chambers, the atria, usually has an adverse impact on the lower chambers, the ventricles, resulting in a fast and irregular pulse. Though some patients have no symptoms whatsoever from their atrial fibrillation, others may complain of palpitations,



breathlessness or fatigue

Fig 3.6 :Atrial fibrillation{3}

Ventricular fibrillation is rapidly fatal if not reversed by defibrillation. Sometimes the signal from the heart's upper to lower chambers is impaired or doesn't transmit. This is "heart block" or "AV block." This does not mean that the blood flow or blood vessels are blocked.

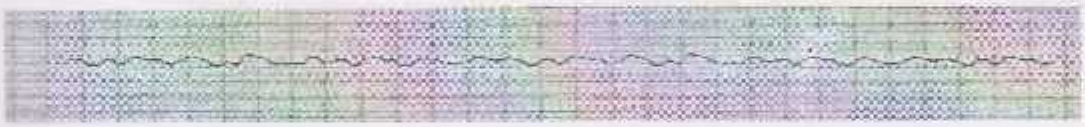


Fig 3.7: Ventricular fibrillation{4}

Heart block is classified according to the level of impairment — first-degree heart block, second-degree heart block or third-degree (complete) heart block.

First-degree heart block :

Also called first-degree AV block, is when the electrical impulse moves through the AV node more slowly than normal. The time it takes for the impulse to get from the atria to the ventricles (the PR interval) should be less than about 0.2 seconds. If it takes longer than this, it's called first-degree heart block. Heart rate and rhythm are normal, and there may be nothing wrong with the heart.

Certain heart medicines such as digitalis can slow conduction of the impulse from the atria to the ventricles and cause first-degree AV block. Also, well-trained athletes may have it. Generally, no treatment is necessary for first-degree heart block.

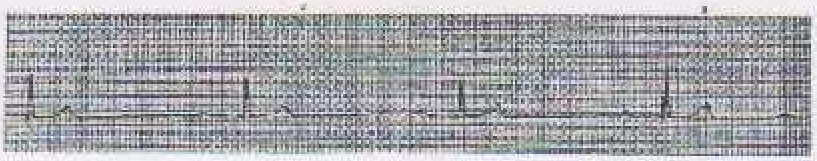


Fig 3.8 : First degree (incomplete) heart block.

second-degree heart block :

In this condition, some signals from the atria don't reach the ventricles. This causes "dropped beats." On an ECG, the P wave isn't followed by the QRS wave, because the ventricles weren't activated. There are two types:

- Type I second-degree heart block, or Mobitz Type I, or Wenckebach's AV block. Electrical impulses are delayed more and more with each heartbeat until a beat is skipped. This condition is not too serious but sometimes causes dizziness and/or other symptoms.

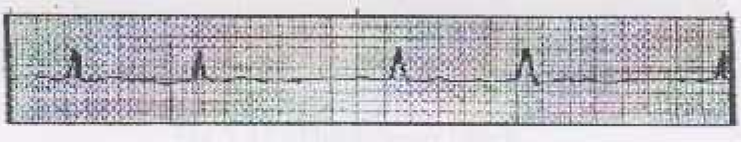


Fig 3.9 : Wenckebach

- Type II second-degree heart block, or Mobitz Type II. This is less common than Type I but generally more serious. Because electrical impulses can't reach the ventricles, an abnormally slow heartbeat may result. In some cases a pacemaker is needed.

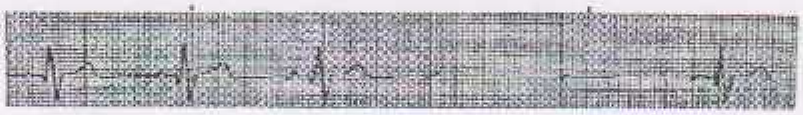


Fig 3.10: Classical second-degree heart block .

Third-degree or complete heart block :

Complete heart block (complete AV block) means that the heart's electrical signal doesn't pass from the upper to the lower chambers. When this occurs, an independent pacemaker in the lower chambers takes over. The ventricles can contract and pump blood, but at a slower rate than that of the atrial pacemaker.

These impulses are called functional or ventricular scope beats. They're usually very slow and can't generate the signals needed to maintain full functioning of the heart muscle. On the ECG, there's no normal relationship between the P and the QRS waves.

Complete heart block is most often caused in adults by heart disease or as a side effect of drug toxicity. Heart block also can be present at — or even before — birth. (This is called congenital heart block.) It also may result from an injury to the electrical conduction system during heart surgery. Complete heart block may be a medical emergency with potentially severe symptoms and a serious risk of cardiac arrest (sudden cardiac death). If a pacemaker can't be implanted immediately, a temporary pacemaker might be used to keep the heart pumping until .

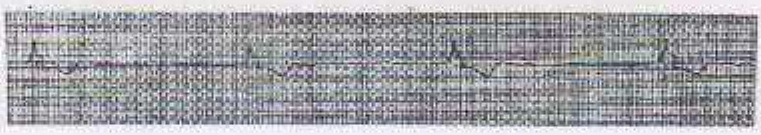


Fig 3.11 :Third heart block.

These rhythm disorders are the consequence of one of two mechanisms:

a) **The circus motion:**

The concept of the circus motion can be illustrated in the figure.3.12 .

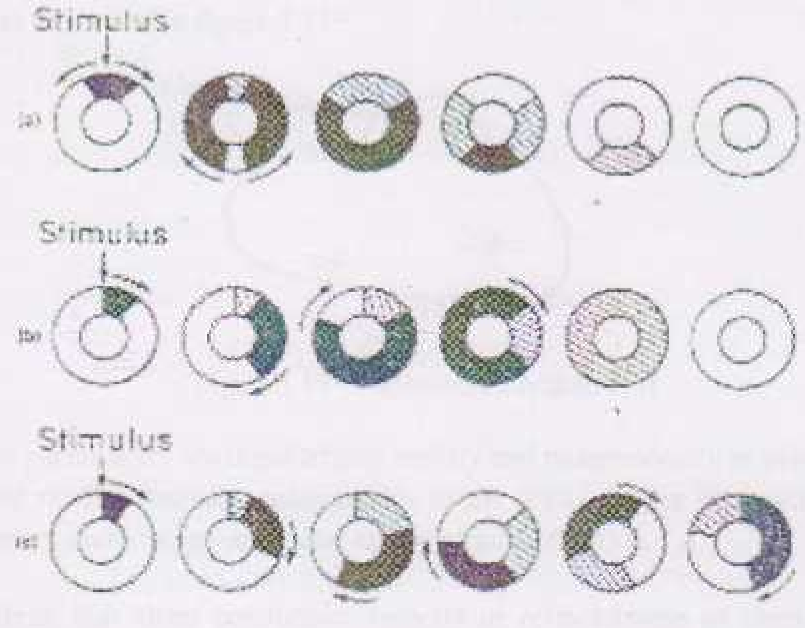


Fig3.12 :Circus motion.

As in figure.3.12 (a) ring of tissue is stimulated at the point indicated by the arrow, and excitation travels around the ring in both directions until the two excitatory wave fronts encounter each other at bottom of the ring. Since depolarized tissue cannot be excited, the stimulus wave front of activation is extinguished. This is the usual and normal sequence for heart tissues. In figure.3.12 (b and c), the effect of a wave front being propagated in only one direction, due to altered metabolism, and of temporary refractoriness of the excitable tissue in the other direction is shown.

In case (c), the stimulus may travel around the ring in perpetuity, sending off "daughter" pacemaker signals to the rest of the heart each time the circle is traversed. This mechanism is often proposed to explain atrial flutter, which is a very rapid rate of the atria of the heart, and ventricular tachycardia, which is the rapid, but still organized, arrhythmia of the ventricles.

b) Presence of multiple pacemaker sites:

There is more than one pacemaker located in the myocardial cells of the heart chamber, as illustrated in figure.3.13 :

Figure3.13 : Pacemaker Located :(5)

These pacemakers are depolarizing rapidly and independently to produce a chaotic and/or rapid rhythm disorder, recognizable in the (ECG ,this is frequently proposed as the mechanism underlying ventricular and/or atrial fibrillation.

Anything that alters conduction velocity or refractoriness of cardiac tissue, that alters the excitability or automaticity of pacemaker cells, that creates changes in membrane potentials of adjacent cardiac cells, or that converts latent pacemaker tissue to activate pacemaker tissue can lead to these rhythm disorders. In practice, most common cause of tachyarrhythmias is ischemia of the heart tissues. Less frequent causes are other forms of heart disease.

3.5 Detection of fibrillation:

We can detect the cardiac tachyarrhythmias by visual inspection of electrocardiographic tracings. This may be on either a monitor screen or on an ECG paper-strip recording. Occasionally, presence of a tachyarrhythmia is assumed under conditions in which a pulse less, apneic patient is encountered in field and no electrocardiographic recording capabilities are available. If a defibrillation shock is applied under these conditions, it is referred to as "blind" defibrillation.

Automated electrocardiographic detection devices, which use signal processing and decision algorithms for detection of tachyarrhythmias, include fibrillation, tachycardia, and flutter are also available. The defibrillators which are designed for automatic or semiautomatic use by emergency personal is provide with automated electrocardiographic detection.

The automated detection schemes are usually based on counting the rate of the QRS complexes per minute and/or looking for changes in a "normal" envelope for ECG waveforms. Also, some of the detection algorithms include analysis for impedance changes due to respiratory or cardiac activity, and in the event that electrographic criteria are met for tachyarrhythmia, the device will not diagnose fibrillation, if the patient has mechanical evidence for cardiac or respiratory activity. [5]

Chapter Four

Defibrillator modes

4.1 : Synchronized cardioversion (automatic) mode:

Synchronized Cardioversion refers to an electrical energy discharge that is synchronized with the large R or S wave of the QRS complex Synchronization in the early part of the QRS complex avoids energy delivery in the early phase of repolarization when ventricular fibrillation (VF) can be easily induced. Defibrillation refers to an unsynchronized discharge of energy and is only recommended for ventricular fibrillation (VF).

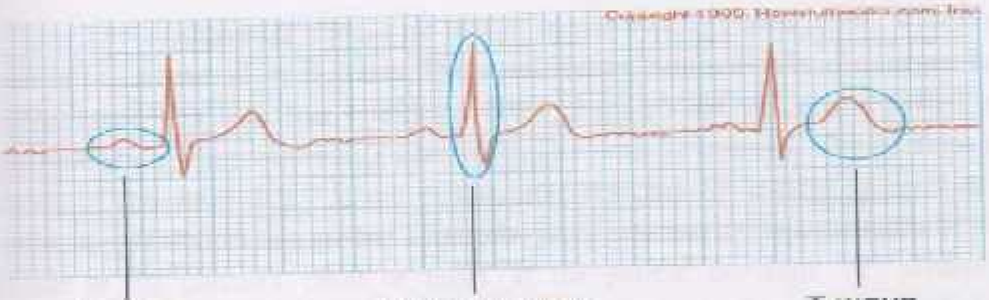


Fig 4.1 : ECG waveform. {6}

In certain types of arrhythmia the patient's ventricles maintain their ability to pump blood, as evidenced by the existence of an *R* wave feature in the ECG waveform. These arrhythmias are also correctable by electrical shock to the heart, but it is necessary to avoid delivering the shock during the ventricles' refractory period (the *T* wave of the ECG waveform), or the shock intended to correct the problem will create a much more serious arrhythmia such as ventricular fibrillation. The shock is usually time to occur approximately 30 microsecond after the *R* wave peak.

Human operator cannot be trusted to time the ECG waveform properly to void this problem. so the automatic electronic circuit is used A machine equipped with synchronized circuit is called a cardioverter.

Transient delivery of electrical current causes a momentary depolarization of most cardiac cells. This allows the sinus node to resume normal pacemaker activity.

In the presence of reentrant-induced dysrhythmia, electrical cardioversion interrupts the self-perpetuating circuit and restores a sinus rhythm. Electrical cardioversion is much less effective in treating arrhythmia caused by increased automaticity .

In an urgent or emergent setting, any patient with reentrant tachycardia having an either narrow or wide QRS complex (ventricular rate >150) who is unstable should be immediately treated with synchronized electrical cardioversion. Synchronized electrical cardioversion may be used to treat stable ventricular tachycardia (VT) that does not respond to a trial of intravenous medications. In hemodynamically stable patients, synchronized electrical cardioversion is also used to electively restore sinus rhythm in atrial fibrillation, atrial flutter, or other supraventricular tachycardia (SVT).

The cardioverter should be placed in the synchronized mode, which permits a search for a large R or S wave. The cardioverter automatically discharges an electric current that lasts less than 4 milliseconds and avoids the vulnerable period of cardiac repolarization when ventricular fibrillation (VF) can be induced. The operator should be aware of this brief delay as the cardioverter searches for a large positive or negative deflection. If deflections are too small, the physician can change the leads or place them closer to the patient's chest or heart. If the patient develops VF, always turn off synchronization to avoid delay in energy delivery to perform synchronized electrical cardioversion two electrode pads are used, each comprising a metallic plate which is faced with a saline based conductive gel. The pads are placed on the chest of the patient, or one is placed on the chest and one on the back. These are connected by cables to a machine which has the combined functions of an ECG display screen and the electrical function of a defibrillator. A synchronizing function (either manually operated or automatic) allows the cardioverter to deliver a reversion shock, by way of the pads, of a selected amount of electric current over a predefined number of milliseconds at the optimal moment in the cardiac cycle which corresponds to the R wave of the QRS complex on the ECG. Timing the shock to the R wave prevents the delivery of the shock during the vulnerable period (or relative refractory period) of the cardiac cycle, which could induce ventricular fibrillation. If the patient is conscious, various drugs are often used to help sedate the patient and make the procedure more tolerable. However, if the patient is haemodynamically unstable or unconscious, the shock is given immediately upon confirmation of the arrhythmia. When synchronized electrical cardioversion is performed as an elective procedure, the shocks can be performed in conjunction with drug therapy until sinus rhythm is attained. Multiple electrical shocks may cause burns of the epidermis at the pad sites. After the procedure, the patient is monitored to ensure stability of the sinus rhythm.

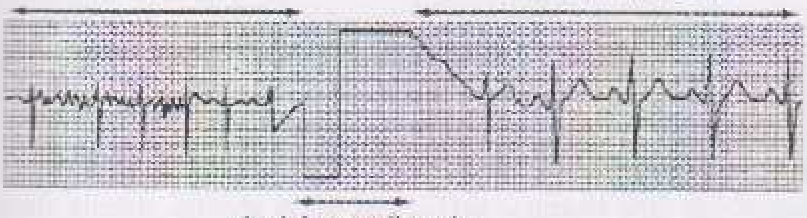


Fig 4.2 : Shock from cardioversion

The energy (shock dose) used for a synchronized shock is lower than that used for unsynchronized shocks (defibrillation). These low-energy shocks should always be delivered as synchronized shocks because if they are delivered as unsynchronized shocks they are likely to induce VF.

- The goal is to avoid shock delivery in the relative refractory period of the cardiac cycle, where a shock might cause VF.
- Synchronized electrical cardioversion is effective when a reentry circuit is the cause of the arrhythmia, as is interrupts the reentry pattern. Synchronized cardioversion is not effective in sinus tachycardia , junctional tachycardia, ectopic or multifocal atrial tachycardia because these rhythms are related to increased automaticity.
- Synchronized electrical cardioversion is not indicated for treatment of VF, pulseless VT, or unstable polymorphic VT. These arrhythmias require high-energy unsynchronized shocks (defibrillation).
- Cardioversion energies for biphasic machines have not yet been established, but we extrapolate from experience with monophasic machines. For reentry SVT, the recommendation is to start at 100J; for atrial fibrillation, start at 100J; for atrial flutter, start at 50J.
- When choosing between synchronized cardioversion and antiarrhythmic drugs, physicians often reach for drugs first because we are more comfortable with using them. All these drugs have undesirable effects such as hypotension and are likely to make an unstable patient even worse. A good rule of thumb is that electricity is faster, more effective and safer than drugs in an unstable patient.

This applies to sedation as well. While we don't want to be cruel, it is unlikely you are going to be able to achieve optimum sedation in an unstable patient because their blood pressure won't tolerate it. It is better to err on the side of light sedation and analgesia; you can apologize for the discomfort after you save your patient's life.

4.2 Unsynchronized Cardioversion(manual) mode :

There is no concern about the timing of the charge as there is no coordinated intrinsic electric activity in the heart. This contrasts with defibrillation, which can occur at any point in the cardiac cycle and is usually performed under emergency circumstances.

'The energy (shock dose) used for a un synchronized shock is higher than that used for synchronized shocks (defibrillation.

Chapter Five

Defibrillator wave form and the Energy

When sudden cardiac arrest strikes, CPR alone doesn't save lives - it is merely a temporary measure that maintains minimal oxygen flow to the brain. Early defibrillation is required to re-establish a regular heartbeat. A defibrillator can deliver a controlled electrical shock to a heart that has a life-threatening rhythm, such as ventricular fibrillation (VF). In VF, the heart's chaotic activity prevents blood from pumping to the body and brain. Voltage stored by the defibrillator conducts electrical current (a shock) through the chest by way of electrodes or paddles placed on the chest. This brief pulse of current halts the chaotic activity of heart, giving the heart a chance to re-start with a normal rhythm.

The defibrillator uses energy to deliver a shock. The amount of energy used depends on:

- How much voltage is used .
- How much current is delivered.
- The duration (length) of the shock.

Energy is measured in joules (J). External defibrillators may offer a range of energy selections. So-called "low energy" defibrillators are those that limit their energy selections to 200J or less. Escalating energy defibrillators offer a range of energies, starting with low energy levels with the option to increase the energy levels for subsequent shocks.

Many people confuse current and energy. This distinction is important in defibrillation, since defibrillators are often described in terms of energy (e.g., 200J) but it is their current - not the energy - that defibrillates. Successful defibrillation requires that enough current be delivered to the heart muscle during the shock.

Defibrillation requires a true middle-of-the-road approach. You must have enough current reach the heart to defibrillate the heart (stop the lethal rhythm), but not so much peak current that you risk damaging the heart. In fact, low-energy shocks from some defibrillators deliver higher peak current than higher-energy shocks from other types of defibrillators.

Impedance is the body's resistance to the flow of current. Some people naturally have higher impedance than others.Certain factors can also increase impedance, such as:

- A large and/or hairy chest.
- Very dry skin.
- Excess air in the lungs .
- Improper application of the defibrillation electrodes.

You can't tell if someone has high impedance simply by looking at him or her. If impedance is high, the heart may not receive enough current for defibrillation to be successful. More current may be delivered by increasing the voltage and by increasing the energy selected (more joules) on the defibrillator.

Biphasic waveforms adjust for impedance by varying the characteristics of their waveforms. How each waveform adjusts for impedance has important consequences - it may determine whether or not someone's life is saved. It is important to know how each biphasic waveform adjusts for impedance to ensure that high-impedance persons will have the same chance for survival as those who are easily defibrillated.

Many clinical studies demonstrating the success of low-energy biphasic waveforms were conducted in electro-physiology labs under ideal conditions. In real life, cardiac emergencies are much less predictable. Many factors affect the chance of defibrillation success: time elapsed before the first shock is given, placement of the electrode pads, the person's impedance level and certain health conditions. Therefore, it may take more current, a longer shock duration, and/or increased voltage to ensure success. Current flow changes with time during a defibrillation shock. When drawn on a graph, this is known as a waveform. Hearts respond differently to different waveforms, which is why the introduction of biphasic waveforms to external defibrillators can have a positive impact.

5.1 Monophasic versus Biphasic waveforms :

Modern defibrillators are classified according to 2 types of waveforms: monophasic and biphasic, defibrillators have used monophasic waveforms. Monophasic" means that the current delivered by the machine travels in only one direction between the paddles. This has been the standard way of doing things for many years, but is now seen as out of date, and is being replaced with a newer method, called "biphasic" .

Biphasic means that the current initially moves towards the positive paddle, then reverses direction and heads the other way. The difference for us at the bedside is that biphasic shocks seem to be just as effective as the monophasic ones, but at lower power levels. This is a good thing for a couple of reasons: first. less power applied means less trauma to the patient. Second, less power required means longer battery life, and apparently all implanted defibrillators now use biphasic shocks for this reason – they can also be made smaller. I remember seeing patients come in with what looked like a small brick implanted under the skin of their chests.

Available research shows that biphasic waveforms are more effective and pose less risk of injury to the heart than monophasic waveforms, even when the shock energy level is the same. This is why manufacturers of external defibrillators are now using biphasic waveforms in their devices , Although the latest research shows biphasic defibrillation to be more effective than monophasic, the International Guidelines 2000 published by the American Heart Association (AHA) state: "These new recommendations do not imply that care using past guidelines (for monophasic devices) is either unsafe or ineffective." However, biphasic waveforms are becoming the new standard of care in external defibrillators. That's why most organizations choose the biphasic waveform when purchasing a new external defibrillator today.

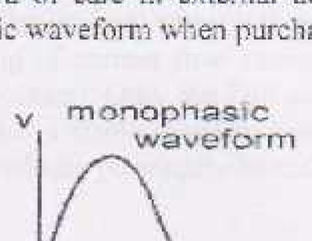


Fig 5.1: Monophasic Waveforms

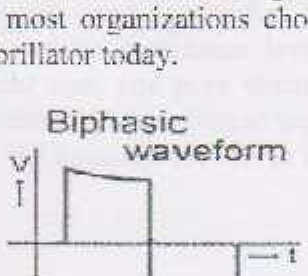


Fig 5.2 : Biphasic Waveforms

In the past there was only one type of transthoracic defibrillation, that of a standard dampened sine wave monophasic shock. Over the many years of study, the theory of impedance and timing of shocks has accumulated in today's standard practice of 25lbs of pressure (if using paddles) with three "stacked shocks". The key has been the sequential raising of the energy from 200j, to 300j, to a maximum of 360j, then having the subsequent shocks at 360j. In relation to the energy there have been many studies to evaluate the effect of multiple high-energy shocks on the heart muscle itself.

The truncated exponential biphasic waveform, in which the polarity is reversed part way through the pulse, has been used in internal pacemakers for over 10 years. There have been many studies done to prove some of the following points: With the Biphasic system there is higher success rate of initial shock conversion from VT (ventricular tachycardia) or VF (ventricular fibrillation) than monophasic (85.2% monophasic vs. 97.6% biphasic). The joules are significantly less (200j monophasic, 130 + 20j biphasic) which will affect reserve energy needs, Biphasic is more effective in reversing sustained VF. Biphasic defibrillation offers equal or better efficacy at lower energies than traditional monophasic waveform defibrillators-with less risk of post-shock complications such as myocardial dysfunction and skin burns.

The biphasic truncated exponential (BTE) waveform was originally developed for low-impedance internal cardiac defibrillation applications. It's been adapted for external defibrillation by two vendors. Heart stream pioneered the low-energy approach. The second BTE defibrillator, developed by Medtronic Physio-Control, uses a high-energy (over 200 joules) protocol. This approach is promoted as easier to adopt but exposes patients to potentially higher peak currents.

The rectilinear biphasic waveform was developed specifically for external defibrillation and takes into account high and varied patient impedance levels (the blocking of current flow caused by chest hair, large chest size, and poor electrode-to-chest contact). Only the Zoll defibrillator uses this waveform. The rectilinear waveform maintains a stable shape in response to impedance, and the constant current in the first phase reduces potentially harmful peak currents.

The biphasic truncated exponential (BTE) waveform was developed for internal use, where impedance is low. When it's used in a transthoracic device such as a defibrillator, impedance affects the waveform's shape. Research has shown that as the biphasic waveform's shape changes, its efficacy varies. The rectilinear waveform remains stable in shape, however, and current delivery dynamics are similar for patients over a wide range of impedances. This reduces the potentially adverse effect of patient impedance on successful defibrillation.

When impedance is low (50 ohms), a 360-joule BTE defibrillator delivers more current than required, exposing the patient to potentially harmful high peak currents. At an average patient impedance of 75 ohms, the 360-joule BTE and 200-joule rectilinear defibrillators are equally effective. With high-impedance (greater than 100 ohms), the 200-joule rectilinear shock delivers a higher average current than a 360-joule BTE shock, therefore making it more effective at lower energy levels.

A direct clinical comparison between the two types of biphasic waveforms has yet to be done in a prospective, randomised trial with appropriate controls. But the growing body of published, peer-reviewed human data points to some waveform-specific performance characteristics Higher energy doesn't necessarily mean you'll be raising the average current delivered. In a recently published study, researchers found that a high-energy BTE defibrillator needs nearly 50% more energy to deliver the same average current as a low-energy rectilinear defibrillator.

Although the very first commercial defibrillator used a biphasic waveform for the treatment of ventricular defibrillation, commercial external defibrillators in the western world adopted monophasic waveforms at least 30 years ago, and these have been used almost exclusively until recently. Thus, much of our clinical experience comes from the use of monophasic waveforms.

Conventional defibrillators produce monophasic shocks where the current flows in one direction. Biphasic waveform technology has developed from electrophysiological work on the design of implantable defibrillators. With biphasic shocks the direction of current flow is reversed at some point (usually near half way) during the discharge from the machine. External defibrillators that utilise biphasic waveforms are now available and licensed for clinical use. These devices have a number of advantages. Low energy biphasic shocks are as effective as higher energy monophasic shocks. This may result in less damage to the myocardium and a reduced frequency of post-shock contractility and arrhythmic problems. It allows smaller, lighter batteries to be used with a lengthening of the defibrillator battery life.

At present, different manufacturers of defibrillators use different energy levels. The precise waveforms used in biphasic shocks vary considerably with different models. The energy levels used with successive shocks may stay constant or escalate depending on the machine. Some of these parameters are programmable, and may be pre-selected by the user. At present, there is inadequate comparative data to be able to decide which is the most effective energy level, shock sequence, or biphasic waveform. It is therefore impossible to make definite recommendations. The Council considers that all currently available biphasic defibrillators have energy levels that are acceptable.

At present, different manufacturers of defibrillators use different energy levels. The precise waveforms used in biphasic shocks vary considerably with different models. The energy levels used with successive shocks may stay constant or escalate depending on the machine. Some of these parameters are programmable, and may be pre-selected by the user. At present, there is inadequate comparative data to be able to decide which is the most effective energy level, shock sequence, or biphasic waveform. It is therefore impossible to make definite recommendations. The Council considers that all currently available biphasic defibrillators have energy levels that are acceptable."

A wave of electrical current has a shape that can be drawn as a "waveform". The waveform shows how the flow of current changes over time during the defibrillation shock. The highest part of the current waveform is called "peak current". Too much peak current during the shock can injure the heart. It's the peak current (not energy) that can injure the heart.

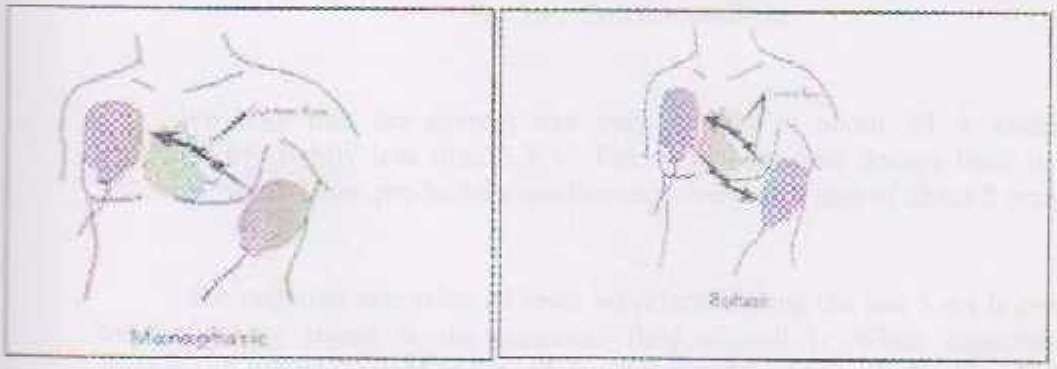


Fig 5.3:Energy direction in Monophasic and Biphasic Waveforms (7)

5.2 Types of waveform using in modern defibrillator:

1. Lown Waveform :-

The Lown waveform , was the standard for defibrillation until the late 1980s when numerous studies showed that a biphasic truncated waveform (BTE) was equally efficacious while requiring the delivery of lower levels of energy to produce defibrillation. A side effect was a significant reduction in weight of the machine. The BTE waveform, combined with automatic measurement of transthoracic impedance is the basis for modern defibrillators.

We can see the Lown waveform in the figure.5.4, shows the voltage and the current applied to the patient's chest plotted against time:

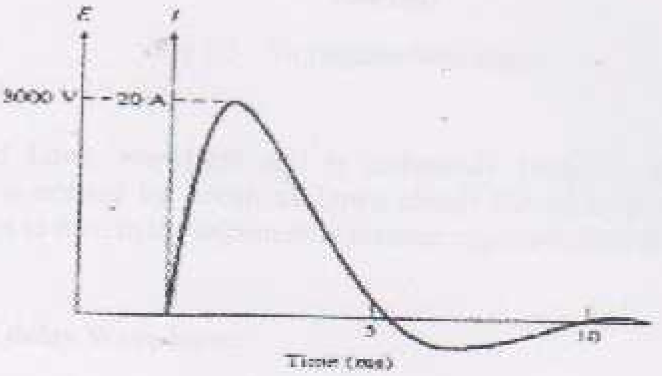


Fig 5.4 : Lown waveform

We note that the current rise very rapidly to about 20 A under the influence of slightly less than 3 KV. The waveform then decays back to zero within 5 ms, and then produces a smaller negative pulse, also of about 5 ms.

The negative excursion of lown waveform during the last 5 ms is produce by the energy stored in the magnetic field of coil L .When capacitor has discharged, the coil's field collapses, dumping energy back into the circuit.

2. **Monopulse Waveform:**

We can see the Lown waveform in the figure.5.5. shows the voltage and the current applied to the patient's chest plotted against time:

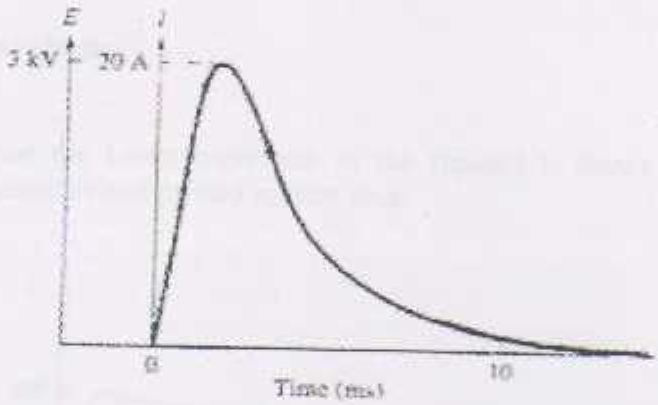


Fig 5.5 : Monopulse Waveform

Modified Lown waveform and is commonly found in certain portable defibrillators. It is created by circuit as Lown circuit but without inductor L_1.The wave form decays to zero in the exponential manner expected of an R-C network.

3. **Tapered DC delay Wave form:**

We can see the Lown waveform in the figure.5.6, shows the voltage applied to the patient's chest plotted against time:

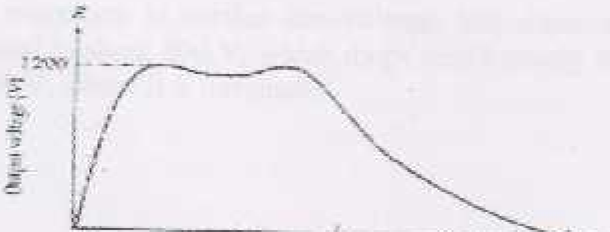


Fig 5.6: Tapered DC delay waveform

This waveform differs from two previous pulses in that it uses a lower amplitude and longer duration to achieve the energy level. The energy transferred is proportional to the area under the square of the curve. So we may have the same energy as in other form. This waveform is achieved by placing two L-C sections, in cascade with other.

4. Trapezoidal Waveform:

We can see the Lown waveform in the figure.5.7, shows the voltage applied to the patient's chest plotted against time:

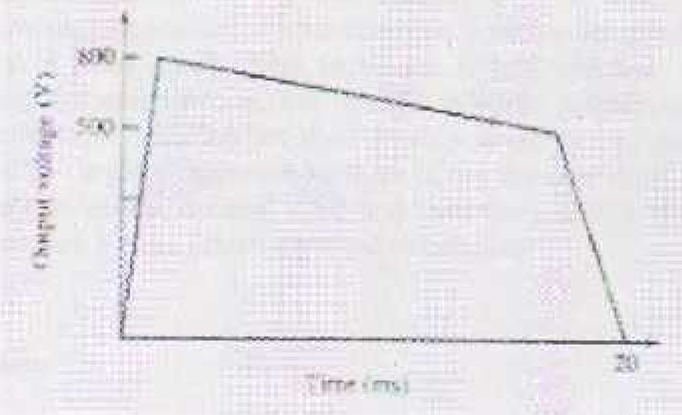


Fig 5.7 : Trapezoidal Waveform

This waveform is another low-voltage, long-duration shape. The initial output potential is about 800 V, which drops continuously for about 20 ms until it reaches 500 V, where it is terminated.

Section Six

Uses of defibrillator

6.1 Basically three uses for defibrillator

1. During cardiac surgery:

Frequently during cardiac surgery, the ventricles will spontaneously fibrillate or will be intentionally fibrillation, it is necessary to defibrillate in order to bring the patient out surgery successfully. In this case of fibrillation applied shocks directly to the heart using 5-30 J. Occasionally, direct defibrillation may be attempted with emergency ventricular fibrillation, such as occurs with a heart attack .This technique is less common than transchest defibrillation for emergencies, but is used in some institutions. The success rates for direct defibrillation are considerably lower during emergencies than during cardiac surgery, inasmuch as there is not the opportunity to control the environment, select the optimal time, and have the patient's heart perfused by cardiopulmonary bypass (extracorporeal circulation).

2. Cardioversion:

Application of synchronized, electrical shocks for treatment of arrhythmias other than ventricular fibrillation. This is virtually always a transchest shock. As stated previously, this includes atrial fibrillation, atrial flutter, and ventricular tachycardia. It is necessary to synchronize shocks to be certain that they do not fall during the T wave of the ventricular electrocardiographic complex, since this is associated with production of ventricular fibrillation which is a potentially fatal arrhythmia. Cardioversion is usually attempted at energy settings 5-20 J for ventricular tachycardia and atrial flutter and energies of 50-200 J for atrial fibrillation. In general, energies are low for the first shock, and the energy is increased gradually and only if previous shocks are unsuccessful in converting the arrhythmia to functional rhythm.

3. **Emergency fibrillation:**

In this case, the defibrillation shock is applied transchest, more commonly using precordial electrodes, but sometimes using anterior-to-posterior electrodes. Higher energy levels are required for this situation, with an initial shock of 200 j being followed by shocks 300 or 360 J if the lower energy is unsuccessful.

6.2 How Use the Defibrillator?

1. Turns the set energy control to the desired level and presses the charge button.
2. Capacitor C beings charging and will continue to charge until the voltage across the capacitor is equal to the desired set energy, and then release a tooting meaning ready to defibrillate.
3. Position the paddles as illustrated in figure 6.1, one electrode is placed on the chest directly over the heart, while the second electrode is placed on the left side of the patient chest, and press discharge button.

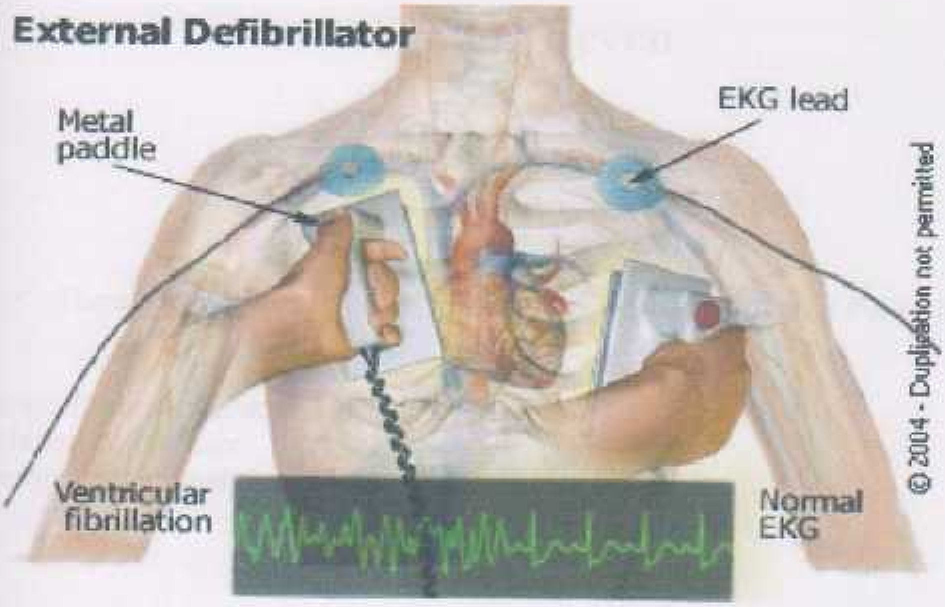


Fig 6.1 :External Defibrillator

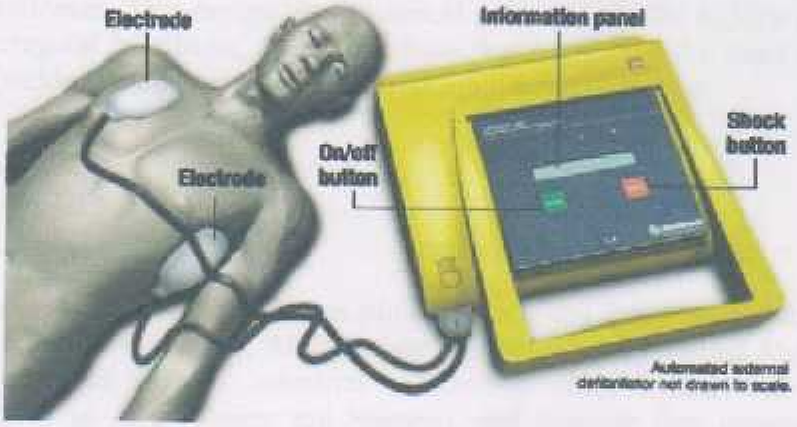


Fig 6.2 : External Defibrillator

We note that we should connect the ECG to the patient during applied the DC shock to see the cardiac waveform during fibrillation.

Section Seven

Types of Defibrillator:

✓ There are three main types of devices used for defibrillation of the heart :

1. Implantable cardioverter defibrillator (**ICD**).
2. Manual defibrillator.
3. automated external defibrillator (**AED**).

7.1 Implantable Cardioverter Defibrillator (ICD).

What is mint by (ICD)?

An implantable cardioverter defibrillator (ICD) is a small device that's placed in the chest or abdomen. This device uses electrical pulses or shocks to help control life-threatening, irregular heartbeats, especially those that could lead the heart to suddenly stop beating (sudden cardiac arrest). This device similar to pacemaker .

History of the (ICD) :

The development of the ICD was pioneered at Sinai Hospital in Baltimore by a team including Stephen Heilman, Alois Langer, Morton Mower, Michel Mirowski, and Mir Imran, with the help of industrial collaborator Intec Systems of Pittsburgh. Mirowski teamed up with Mower and Staewen, and together they commenced their research in 1969 but it was 11 years before they treated their first patient. Imran was the young engineer who was able to turn the bench top idea into a working device. Using analogue tape recordings of patients heartbeats, Imran developed algorithms and hardware that are the foundation of today's .

ICD's and Cardiac Rhythm Management industry. Similar developmental work was carried out by Shudder and colleagues at the University of Missouri.

More than a decade of research went into the development of an implantable defibrillator that would automatically sense the onset of ventricular fibrillation and deliver an electric counter shock within 15-20 seconds, converting the rhythm to sinus rhythm. Improved versions were programmed to be able to detect ventricular tachycardia, often a forerunner of ventricular fibrillation. These were then called implantable cardioverters.

The work was commenced against much skepticism even by leading experts in the field of arrhythmias and sudden death. There was doubt that their ideas would ever become a clinical reality. In 1972 Bernard Lown, the inventor of the external defibrillator, stated in the journal Circulation - "The very rare patient who has frequent bouts of ventricular fibrillation is best treated in a coronary care unit and is better served by an effective ant arrhythmic program or surgical correction of inadequate coronary blood flow or ventricular malfunction. In fact, the implanted defibrillator system represents an imperfect solution in search of a plausible and practical application".

The problems to be overcome were the design of a system which would allow detection of ventricular fibrillation or ventricular tachycardia. Despite the lack of financial backing and grants, they persisted and the first device was implanted in February 1980 at Johns Hopkins Hospital by Dr. Levi Watkins, Jr. Modern ICDs do not require a thoracotomy and possess pacing, cardio version, and defibrillation capabilities.Internal cardioverters defibrillators have also been used twice in dogs to prevent sudden death from arrhythmia. The first defibrillator was implanted at Washington State University by a team of cardiologists led by Dr Lynne Johnson in 2003. The patient was a Boxer dog with life threatening arrhythmias from arrhythmogenic right ventricular cardiomyopathy, an inherited disease.

On July 21rst 2008 , a second ICD was implanted in a 6-month-old German Shepherd dog with inherited ventricular arrhythmias. The 5-hour long surgery took place at Louisiana State University and was led by Dr Romain Pariaut. So far, these pets are the only two client-owned dogs that have received such a high-tech treatment.

An over view on (ICD) :

An implantable cardioverter defibrillator (ICD) has wires with electrodes on the ends that connect to one or more of heart's chambers. These wires monitor the heart rhythm. They also deliver high- or low-energy electrical pulses to the heart when it beats abnormally.

Single-chamber ICDs have wires that connect to one or both of the ventricles. These ICDs correct faulty electrical signaling within the ventricles. Dual-chamber ICDs have wires that connect to both an upper heart chamber (atrium) and a ventricle. These ICDs correct faulty electrical signaling between the two chambers .

Many modern ICD's use a combination of various methods to determine if a fast rhythm is normal, ventricular tachycardia, or ventricular fibrillation Rate discrimination evaluates the rate of the lower chambers of the heart (the ventricles) and compares it to the rate in the upper chambers of the heart (the atria). If the rate in the atria is faster than or equal to the rate in the ventricles, then the rhythm is most likely not ventricular in origin, and is usually more benign. If this is the case, the ICD does not provide any therapy.

Rhythm discrimination will see how regular a ventricular tachycardia is. Generally, ventricular tachycardia is regular. If the rhythm is irregular, it is usually due to conduction of an irregular rhythm that originates in the atria, such as a trial fibrillation.

Morphology discrimination checks the morphology of every ventricular beat and compares it to what the ICD believes is a normally conducted ventricular impulse for the patient. This normal ventricular impulse is often an average of a multiple of beats of the patient taken in the recent past.

7.2 What is the difference between ICD and Pacemaker ?

An ICD is similar to a pacemaker, but there are some differences. Pacemakers can only give off low energy electrical pulses. They are often used to treat less dangerous heart rhythms, such as those that occur in the upper chambers of the heart. Most new ICD's can act as both pacemakers and ICD's. the illustration compares an implantable cardioverter defibrillator and a pacemaker.

The wires with electrodes on the ends are inserted into the heart through a vein in the upper chest.. The wires with electrodes on the ends are inserted into the heart through a vein in the upper chest.

7.3 Manual defibrillators :

A manual defibrillator, or defibrillator paddles, are used by physicians to deliver high intensity electrical charges to patients in ventricular fibrillation or cardiac arrest, restoring normal heart rhythm and restarting the flow of blood.

Many popular television programs demonstrate emergency personnel loudly stating, "Clear!" before using the defibrillator paddles. This is done because anyone touching the person about to receive the shock may receive a shock himself or herself.

Manual defibrillators are also used in non-emergency situations in a procedure called elective cardio version. In this procedure, lower levels of electricity are used to restore normal heart patterns in patients with non-emergency arrhythmias (e.g., a trial fibrillation).

Cardio version is the term doctors use to describe the delivery of a shock that is carefully timed to the waveform of the heartbeat, usually under controlled circumstances.This contrasts with defibrillation, which can occur at any point in the cardiac cycle and is usually performed under emergency circumstances.

7.4 The automated external defibrillator(AED)

AEDs deliver a high-amplitude current impulse to the heart in order to restore normal rhythm and contractile function in patients who are experiencing ventricular fibrillation (VF) or ventricular tachycardia (VT) that is not accompanied by a palpable pulse. AEDs differ from conventional defibrillators in that AEDs can analyze the ECG rhythm to determine whether defibrillation is necessary; this eliminates the need for the user to interpret the cardiac rhythm before delivering a shock. AEDs are designed to be used primarily by first responders to cardiac emergencies, who may not be fully trained in advanced cardiac life support (ACLS). In the pre hospital setting, these emergency personnel can include emergency medical technicians (EMTs), firefighters law enforcement officers, and paramedics. More recently, flight attendants, security guards, and others (sometimes called traditional targeted responders) may be expected to use PAD units. AEDs can also be used in areas of the hospital where advanced life support personnel are not readily available.

Principles of operation :

AEDs can be classified as either fully automatic or semiautomatic. Fully automatic models require only that the user apply defibrillator electrodes to the patient and activate the unit, which then analyzes the ECG rhythm obtained through the disposable defibrillation electrodes and determines whether a defibrillation counter shock is needed; if it is, the device automatically charges and discharges.

Most AEDs are semiautomatic. These units analyze the patient's ECG and notify the operator when defibrillation is indicated and charge automatically. The operator then discharges the defibrillator. AED's can use visual messages and/or voice-synthesized instructions to notify the operator of the proper course of action.

AED's typically include a code documentation device, such as a cassette recorder, memory module, or personal computer data card; disposable adhesive defibrillation electrodes through which the cardiac rhythm can be monitored and the electric shock delivered; an LCD (liquid crystal display) or other display to give the user status messages (patient and/or defibrillator), to display the ECG waveform, or to prompt the user to initiate a shock; and audible voice prompts.

Defibrillation procedure of (AED's) :

The operator attaches two adhesive defibrillator electrodes to the cables or directly to the AED and applies the electrodes to the patient. One electrode is usually placed on the patient's chest near the upper-right eternal border, and the other is placed on the lower-left ribs over the apex of the heart. The adhesive electrodes must be carefully applied to ensure good contact with the skin because unlike paddles , These electrodes do not allow the user to apply pressure and thereby lower Tran thoracic impedance by reducing the resistance between the patient's skin and the electrodes.

All biphasic AEDs measure chest impedance during the first 1/1,000 of a second of the shock. The shock can then be altered to deliver the appropriate output for the particular patient.

After the electrodes have been attached to the patient, CPR is discontinued, and either the user activates the analysis function or the AED will automatically analyze the rhythm to determine whether defibrillation is necessary.

After analysis is initiated, all physical contact with the patient must cease for the remainder of the defibrillation process. Depending on the AED, analysis takes from 5 to 15 seconds; in fully automatic models, a shock is then automatically delivered when the rhythm analysis determines it is necessary.

Before the shock, the AED indicates that the capacitors are charging and warns that a shock is imminent. In semiautomatic units, LCDs, visual alarms, or voice-synthesized instructions prompt the user to press a button to deliver the shock. After the first shock, analysis is activated again either manually or automatically, and the defibrillator reanalyzes the cardiac rhythms to determine whether the heart has resumed its normal beat. If fibrillation is still occurring, the device alerts the user and advises another shock.

This procedure repeats until three shocks have been delivered to the patient. CPR should be resumed for one minute if normal rhythm has not been restored after the third shock; if the patient is confirmed to be pulse less and not breathing after CPR, the defibrillation procedure should be resumed.

Accuracy of (AED'S) :

The accuracy of an AED's rhythm-recognition algorithm is usually described in terms of sensitivity and specificity. Sensitivity represents the AED's ability to identify rhythms that should shocked (e.g., VF).

It is defined as the percent of correct decisions to shock out of the total number of truly shock able rhythms. Specificity represents the ability of an AED to identify rhythms that should not be shocked. It is defined as the percent of truly non shock able rhythms that the device made the correct decision not to shock. Ideally, AEDs would have 100% sensitivity as well as 100% specificity.

However, this is not realistic because there are trade-offs between the two. In general, AEDs favor specificity: they avoid shocking non shock able rhythms. In addition, the recommended practice of attaching AEDs only to pulse less, unconscious patients helps to ensure that they will not be used to analyze a large number of non shock able rhythms.

According to this standard, the sensitivity for recognizing VF at an amplitude of (200 μV) or greater should exceed 90% in the absence of artifact. For devices that detect VT, the sensitivity should exceed 75%. The specificity of the device in correctly differentiating non shock able rhythms should exceed 95% in the absence of artifact.

7.5 Battery care and maintenance

AEDs use sealed lead-acid (SLA), nickel-cadmium, or lithium batteries. When possible, SLA batteries not being used should be charged continuously at room temperature until they are fully charged, which can take from 4 to 24 hours. Units that offer disposable lithium batteries do not require recharging, thus simplifying AED use.

Most AED manufacturers recommend that batteries be charged after each use and replaced every one to two years.

Although battery life depends heavily on usage and maintenance practices, routine replacement better ensures that the battery will provide the desired performance when needed. Rechargeable batteries should not be used in PAD applications. For a more detailed discussion, see the Health Devices article on battery maintenance listed below.

One unit covered in this Product Comparison is designed for in-hospital monitoring and automatic defibrillation of patients who are deemed to be at high risk of sudden cardiac arrest.

The unit continuously monitors the patient's cardiac rhythms through disposable electrodes and responds by automatically delivering either cardio version pacing or shock within seconds of detecting VT or VF, depending on the rhythm detected. After setting up the device and attaching the electrodes to the patient, no clinical personnel are needed to deliver the therapy.

Section Eight

Testing Defibrillator

8.1 Defibrillator tester :

There are several defibrillator testers on the market. Most are basically integrating voltmeters that are calibrated in watt-seconds.

As we can see one of these type testers in figure 8.1.

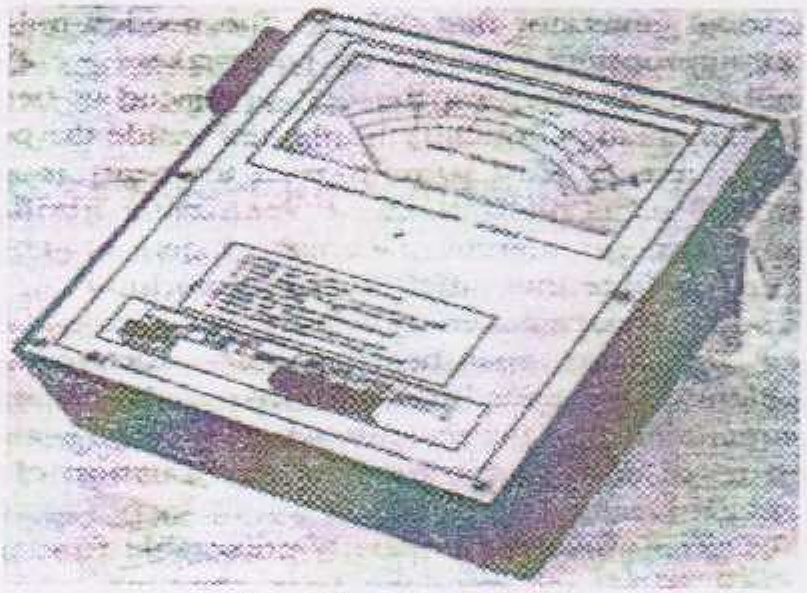


Fig 8.1: Defibrillator tester .

A 50 Ω dummy load is built into tester and connected between a pair of electrodes. The paddles are placed against the electrodes, and the capacitor is discharged into the load. The meter registers the delivered energy in watt-seconds.

Section Eight

Testing Defibrillator

8.1 Defibrillator tester :

There are several defibrillator testers on the market. Most are basically integrating voltmeters that are calibrated in watt-seconds.

As we can see one of these type testers in figure 8.1.

Fig 8.1: Defibrillator tester .

A 50 Ω dummy load is built into tester and connected between a pair of electrodes. The paddles are placed against the electrodes, and the capacitor is discharged into the load. The meter registers the delivered energy in watt-seconds.

It specifies a tester that has an oscilloscope output jack, so that the output waveform may be viewed on the oscilloscope. An oscilloscope camera is usually preferred for making a permanent record of the waveform, although at least one tester uses special digital circuitry that allows recording of the defibrillator output waveform on a 25-mm/s strip chart recorder.

8.2 Defibrillator Safety:

Defibrillators are potentially dangerous devices because of their high electrical output characteristics.The danger to the patient of unsynchronized shocks has already been presented, as has the synchronization design to prevent inadvertent precipitation of fibrillation by a cardioversion shock applied during the *T* wave.

There are other safety issues. Improper technique may result in accidental shocking of the operator or other personnel in the vicinity, if someone is in contact with the electric discharge pathway. This may occur if the operator is careless in holding the discharge electrodes or if someone is in contact with the patient or with a metal bed occupied by the subject when the shock is applied. Proper training and technique is necessary to avoid this risk.

Another safety issue is that of producing damage to the patient by application of excessively strong or excessively numerous shocks. Although cardiac damage has been reported after high-intensity and repetitive shocks to experimental animals and human patients, it is generally held that significant cardiac damage is unlikely if proper clinical procedures and guidelines are followed.

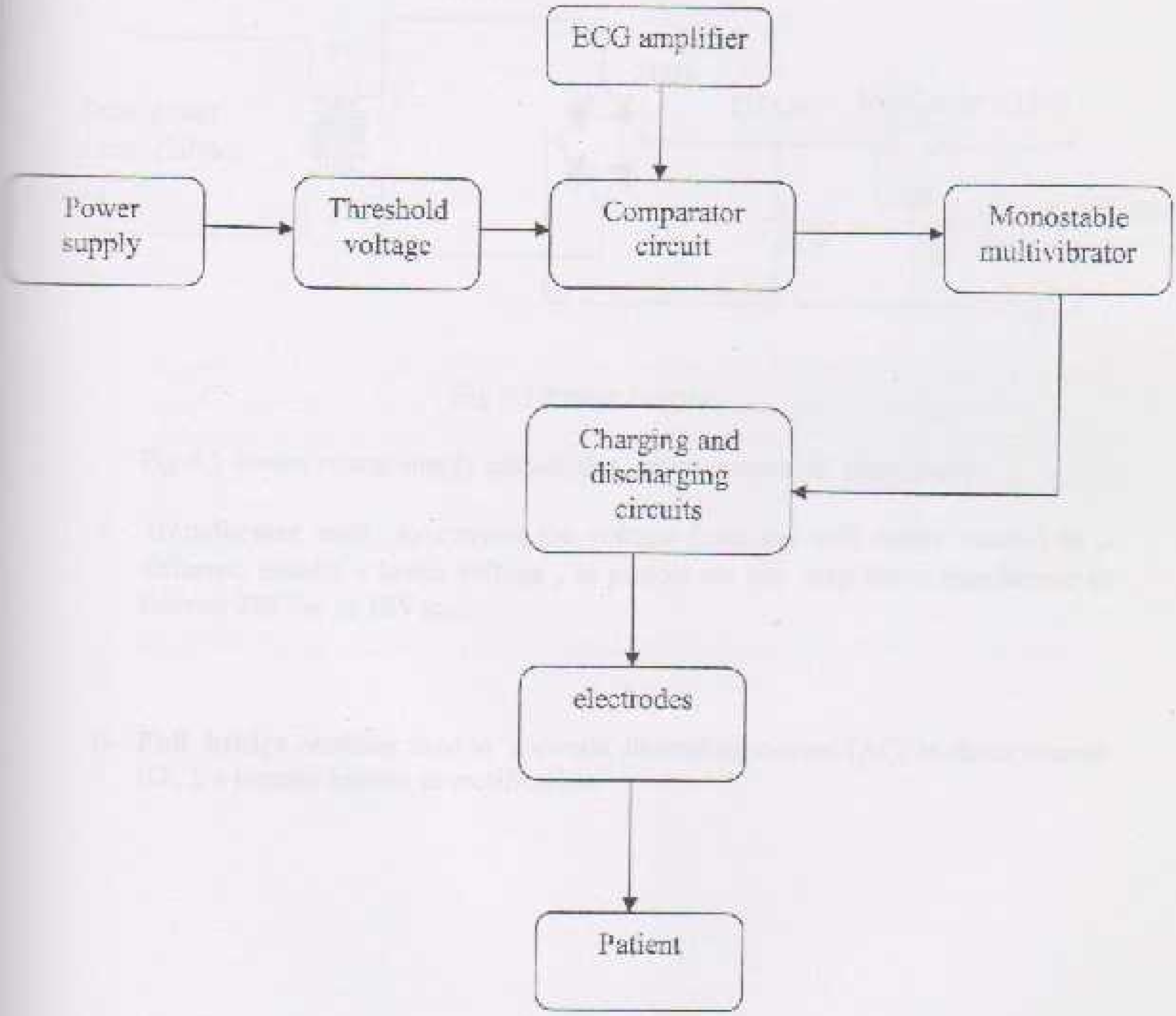
Failure of a defibrillator to operate correctly may also be considered a safety issue, since inability of a defibrillator to deliver a shock in the absence of a replacement unit means loss of the opportunity to resuscitate the patient. A recent review of defibrillator failures found that operator errors, inadequate defibrillator care and maintenance, and, to a lesser extent, component failure accounted for the majority of defibrillator failures.

Complications may affect patients or health care workers. Injury incidence is 1 case per 1700 shocks for paramedics in the field. The patient may become hypoxic or hypoventilate from sedation. Most burns from shocks are superficial partial-thickness burns, but a few are deep. Cardiac complications include dysrhythmia, hypotension, and pulmonary edema.

Inducible arrhythmias include bradycardia, atrioventricular (AV) block, systole, VT, and VF. In patients with acute coronary syndromes or acute myocardial infarction, bradycardia or AV blocks can be induced, and they may need an external or internal pacemaker. VT and VF commonly occur in patients with prior similar history.

Chapter Nine

Block Diagram and circuit analysis



9.1 Power Supply :

system that Conversion of one form of electrical power to another desired form and voltage. This typically involves converting 120 or 220 volt AC supplied by a utility company to a well-regulated lower voltage DC for electronic devices.

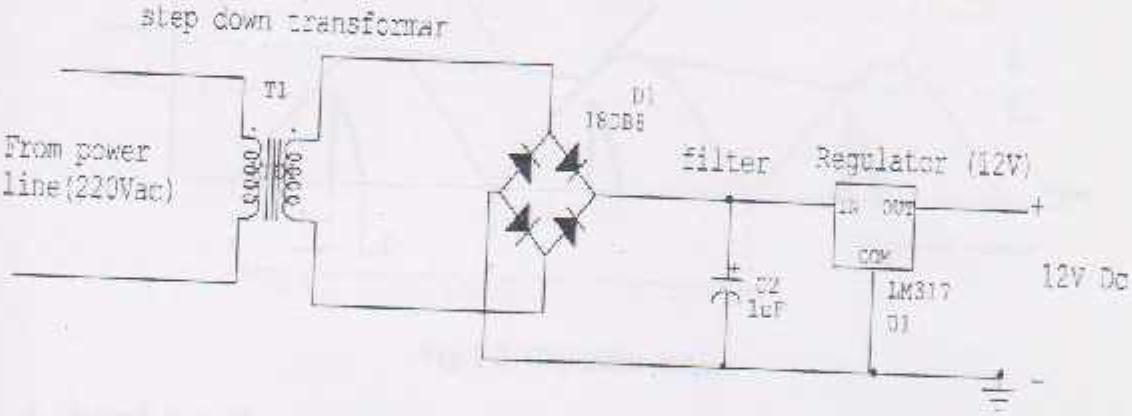


Fig 9.1:Power Supply

Fig 4.1 shown power supply circuit, this circuit consist of many parts:

a- **transformer** used to convert the voltage from the wall outlet (mains) to a different, usually a lower voltage , in project we use step down transformer to convert 220Vac to 12Vac.

b- **Full bridge rectifier** used to converts alternating current (AC) to direct current (DC), a process known as rectification.

c- **A capacitor** is used to smooth the pulsating current from the rectifier(filtering). Some small periodic deviations from smooth direct current will remain, which is known as ripple. These pulsations occur at a frequency related to the AC power frequency .

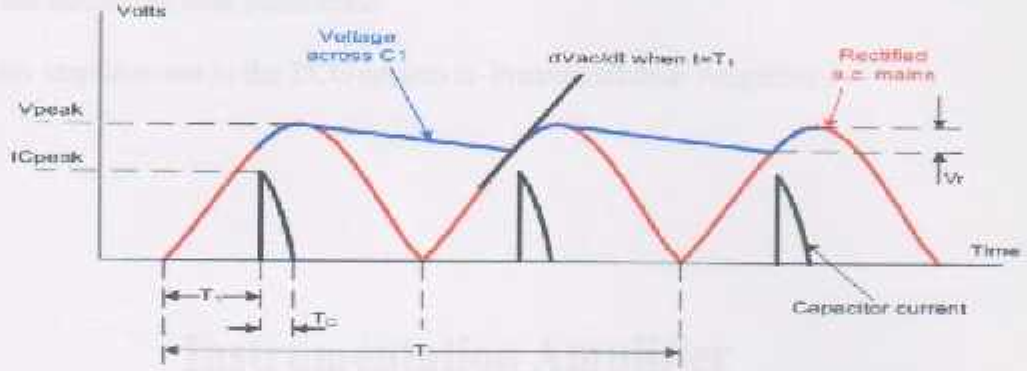


Fig 9.2: Capacitor output

d- **Regulator** The voltage produced by an unregulated power supply will vary depending on the load and on variations in the AC supply voltage. For critical electronics applications a linear regulator will be used to stabilize and adjust the voltage. This regulator will also greatly reduce the ripple and noise in the output DC current. Linear regulators often provide current limiting, protecting the power supply and attached circuit from over current.

In this project we use a positive regulator (7812), to get 12VDC output using to supplied voltage for all electronic circuit in project ,and charging the capacitor.

9.2 ECG amplifier :

The defibrillator gives the possibility to be synchronized to the wave **R** coming from ECG amplifier or to operate manually discharging the energy of the **LC** circuit across the electrodes with pushbutton .

The main amplifier use in the ECG system is Instrumentation Amplifier .

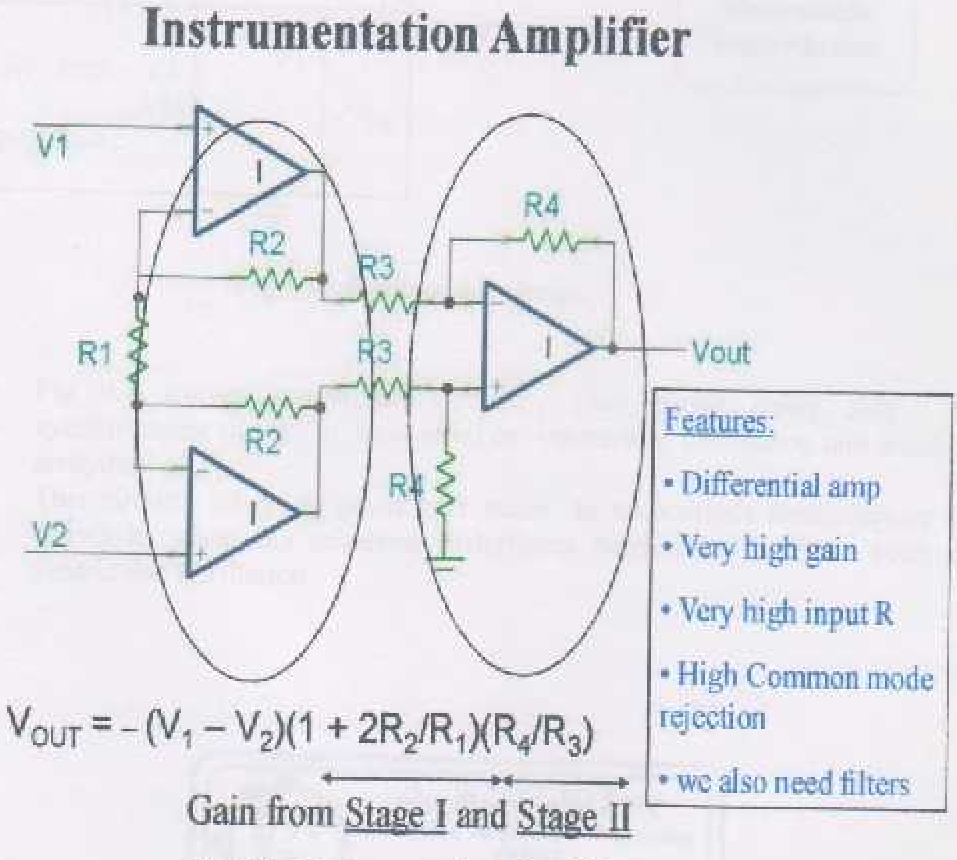


Fig 9.3 : Instrumentation Amplifier

9.3 Comparator circuit :

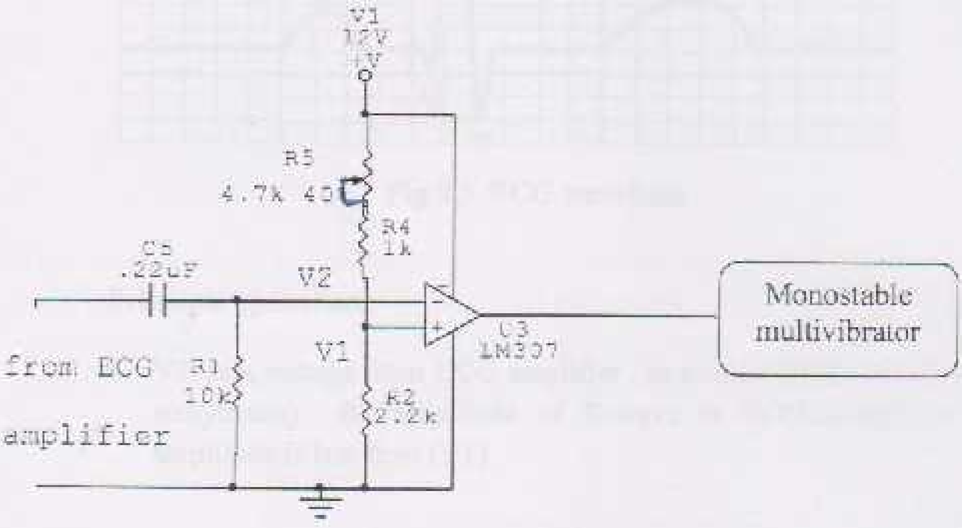


Fig 9.4 : Comparator circuit .

- Fig 9.4 shown comparator circuit , This circuit using only in synchronizing mode , to treat atrial or ventricular fibrillation and another arrhythmia .
- This circuit using to get trigger pulse to monostable multivibrator to switch it when any occurring arrhythmia happen to the heart such as ventricular fibrillation.

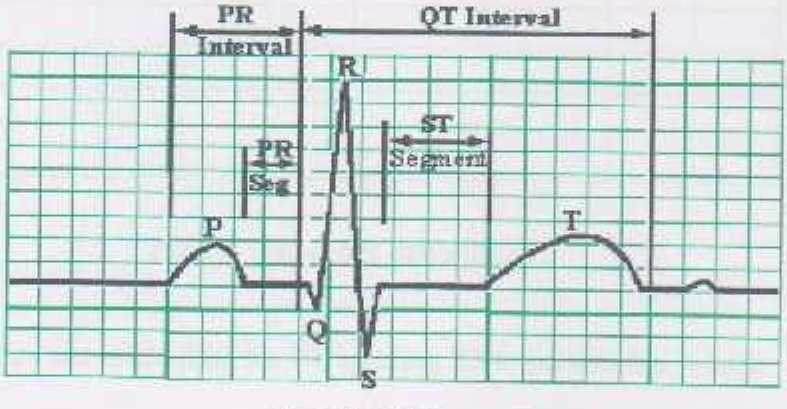


Fig 9.5 :ECG waveform

Principle operation:

1- V2 is a voltage from ECG amplifier , in normal ECG waveform (no arrhythmia) the amplitude of R-wave in ECG almost 3V this amplitude is less than (V1).

$$V1=\frac{R2}{R4+R5+R2}*12V = \frac{2.2}{1+4.7+2.2}*12 = 3.4V$$

The output of comparator will not give trigger pulse to switched on monostable multivibrator , and the defibrillator device will not give any shock .this case shown in fig 9.6 .

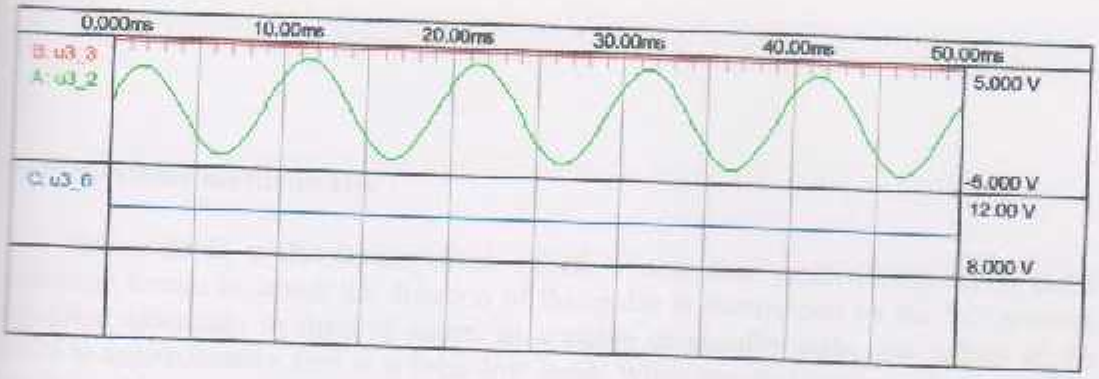


Fig 9.6 : No shock defibrillator .

- You can controlling for V1 by using R5 to get suitable voltage.

2- when any occurring arrhythmia happen to the heart such as ventricular fibrillation. The amplitude of R wave will increasing ⟶ V2 will increasing and become greater than V1 , in these case the comparator will give trigger pulse and switched on monostable multivibrator shown in fig 9.7.

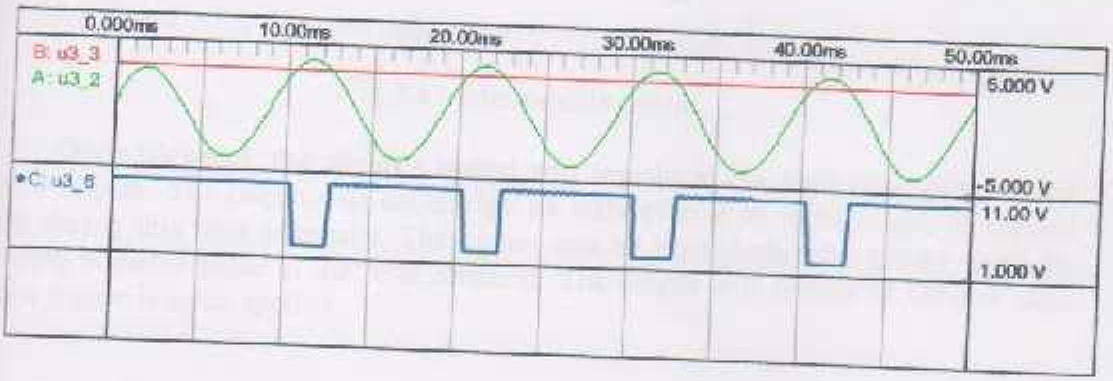


Fig 9.7 : Switched on monostable multivibrator .

where :

- Red color (B wave) represent for (V1) .
- Green color(A wave) represent for (V2) .
- Blue color (C wave)represent for (comparator output voltage)

9.4 Monostable multivibrator :

Monostable multivibrator often called a one shot multivibrator is a pulse generating circuit in which the duration of this pulse is determined by the RC network connected externally to the 555 timer. In a stable or standby state, the output of the circuit is approximately zero or a logic-low level. When external trigger pulse is applied output is forced to go high (+ V_{CC}). The time for which output remains high is determined by the external RC network connected to the timer. At the end of the timing interval, the output automatically reverts back to its logic-low stable state. The output stays low until trigger pulse is again applied. Then the cycle repeats. The monostable circuit has only one stable state (output low) hence the name monostable.

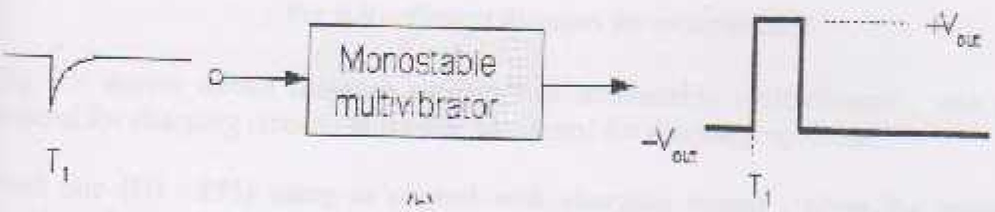


Fig 9.8 : Monostable circuit.

Once triggered, the circuit's output will remain in the high state until the set time, t elapses. The output will not change its state even if an input trigger is applied again during this time interval t. The circuit can be reset during the timing cycle by applying negative pulse to the reset terminal. The output will remain in the low state until a trigger is again applied.

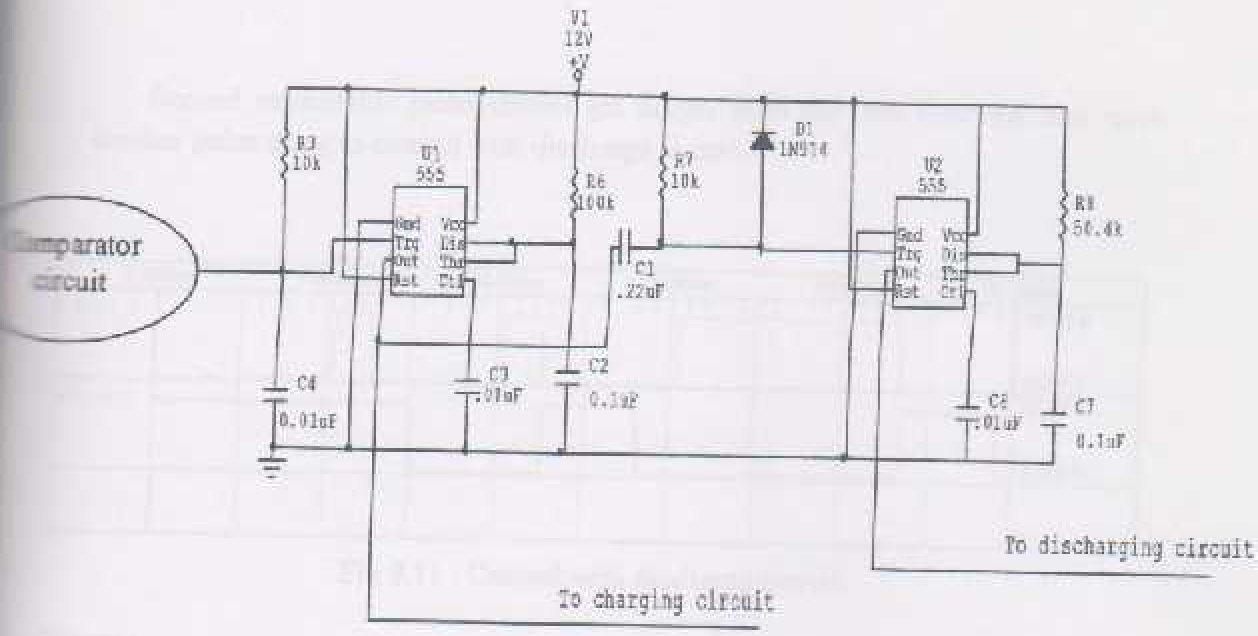


Fig 9.9 : Circuit diagram for monostable .

Fig 9.9 shown circuit diagram contain two monostable multivibrator , one using to control for charging circuit , and other to control for discharging circuit .

First one (U1 555) using to control with charging circuit , when the trigger pulse coming from comparator circuit , this circuit will generate pulse shown in fig 9.10 with known duration of the output pulse in seconds is approximately equal to:

$$t = 1.1 * R6 * C2$$

- you can control for these time by vary R6 to get suitable charging time .

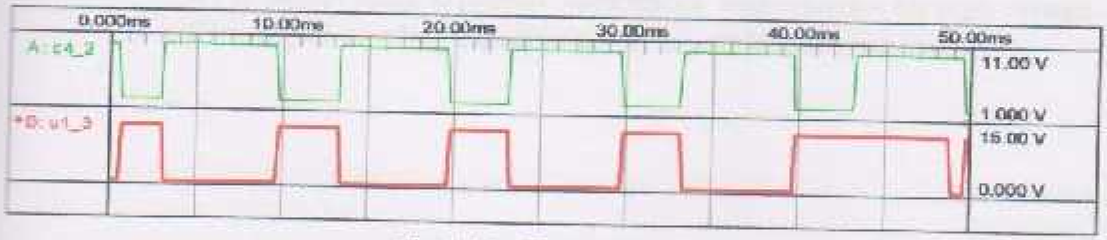


Fig 9.10 : Generate pulse .

Where:

- Green line (first wave) is a trigger pulse from comparator circuit.
- Red line is (second wave)output pulse from monostable multivibrator.

Second monostable multivibrator get trigger from the the first one, and give another pulse using to control with discharge circuit.

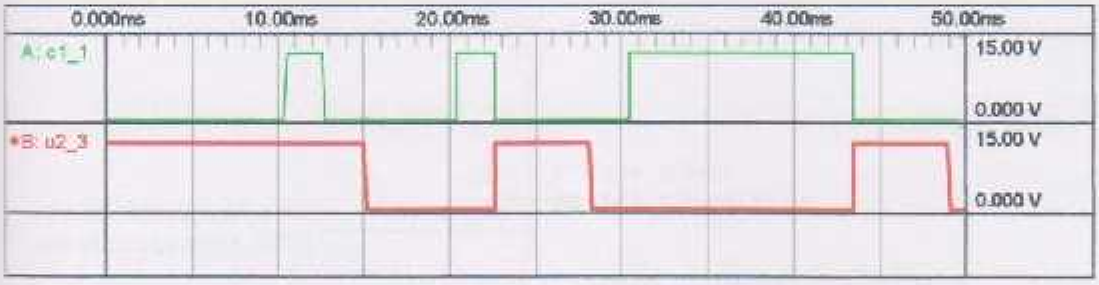


Fig 9.11 : Control with discharge circuit.

Where:

- Green line (first wave) is a trigger pulse from the first monostable multivibrator
- Red line(second wave) is output pulse from the second monostable multivibrator.

- duration of the output pulse in seconds is approximately equal to:

$$t = 1.1 * R8 * C7$$

- you can control for these time by vary R8 to get suitable dicharging time The monostable duration is programmable with the trimmer R8 in arrange between 0. 5 to 5 ms.
- D1 in circuit using to protect monostable multivibrator for high voltage .
- C1 using as coupling capacitor using to couple these two stage .

9.5 Charging and discharging circuit:

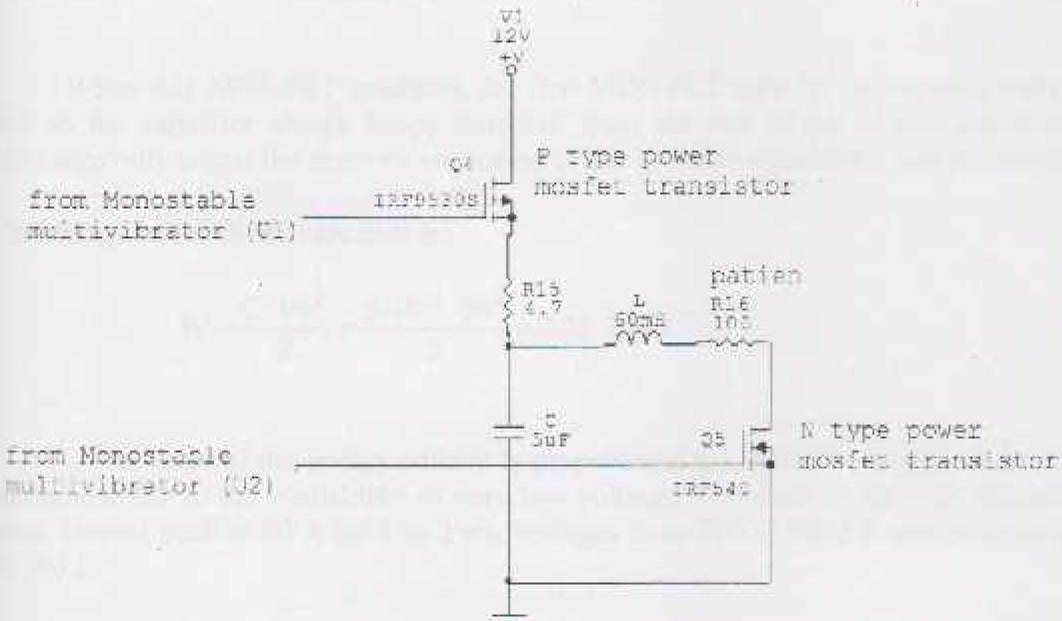


Fig 9.12 : Charging and discharging circuit .

monostable multivibrator (U1) generate output pulse duration fixed to about 10 ms. This output directly drives the MOS-FET type "p" ,this will active on fall edge of this pulse, which enables to charge the 5uF- capacitor from the power supply voltage (30V). while the MOS-FET type "n" is cut-off so that the MOS-FET behaves as open switch ,these case shown in fig 9.12.

Once the capacitor charge monostable pulse is finished, the second monostable (U2)is activated on its falling edge of the second pulse , and this determines the duration of the discharge on the 100 Ohm-resistance. which represents the section of the human body to which the electrodes are applied.

The 100 Ohm (R6) given to the simulation resistance of the human body Represents the average value between the possible values taken by the different situations of direct application to the heart, in which the effective value is lower and in the applications of the electrodes to the chest in which the resistance is higher and depends on the distance between the same electrodes.

When this MOS-FET conducts, the first MOS-FET type "p" is naturally cut off and so the capacitor charge keeps insulated from the rest of the circuit and it can discharge only across the network consisting of the 100 ohm-inductance and resistance.

The energy stored in the capacitor is:

$$W = \frac{C \cdot Vo^2}{2} = \frac{5 \cdot 10^{-6} \cdot 30^2}{2} = 2.25 \cdot 10^{-3} J$$

The value of the energy content is proportionally reduced in respect to the real equipment due to the availability of very low voltages in respect to the ones normally used: current peak of 60 A for 1 to 2 ms, voltages from 700 to 1000 V and energies up to 360 J.

The 5uF-charge and discharge capacities and the 60mH-inductance have been chosen according to the relation between these charge and discharge values of the capacitor.

To obtain a damped oscillating behavior of the discharge, the values of the inductive and capacitive component together with the resistance one simulating the concerned section of the human body, must satisfy the following relation:

$$R < 2\left(\frac{L}{C}\right)^{\frac{1}{2}}$$

With the components of the circuit the expression on the right (-200) satisfies the relation and the discharge effectively occurs in damped oscillatory mode.

The conduction phase duration of the MOS-FET type "n" determines the value of the energy transferred from the capacitor to the 100 Ohm-resistor which represents the patient and so enables the simulation of progressively increasing dosing of energy.

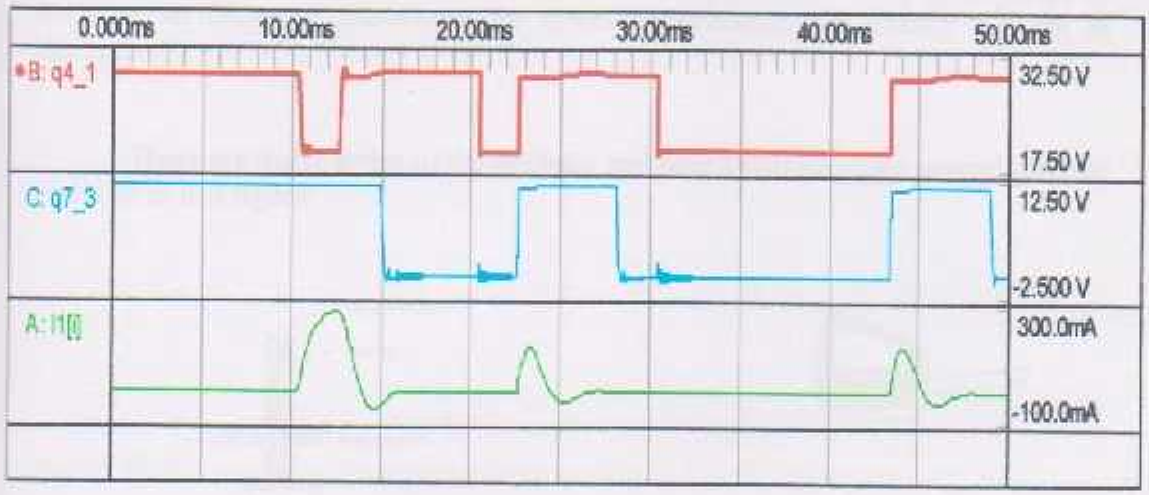


Fig 9.13 : Output waveform for defibrillator .

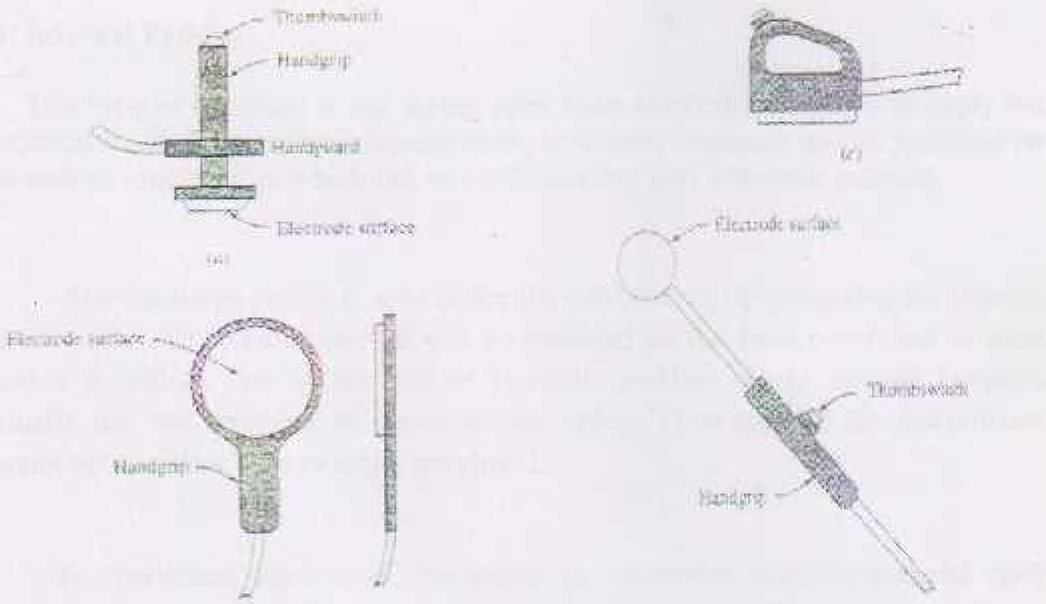
Where:

- Red line(first wave) is input pulse on P-type MOSFT from.
- Blue line (second wave) is input pulse on N-type MOSFT from.
- Green line (third wave) is the output current applied to the patient .

9.6 Defibrillator electrodes:

Electrode and their cables attach to the defibrillator to actually apply current through the body and heart to depolarize myocardial cells. Thoracic electrodes and the direct heart electrodes usually consist of bare metal disks made of noncorrosive material. Size varies, with electrodes for transchest application being in the range of 8-13 cm in diameter. Electrodes for direct defibrillation are usually 4-8 cm in diameter.

There are many styles of defibrillator and here I will illustrate several popular styles as in this figure:



a) Anterior Paddle:

In this design the insulted handgrip is perpendicular to the metal electrodes surface. The high-voltage cable enters from the side. A thumb switch to control the discharge is mounted at the top of the grip.

b) Posterior Paddle:

Constructed flat and is designed so that the patient can lie on it. This style of paddles always paired with one anterior paddle to form an anterior-posterior pair.

c) D-Ring Anterior Paddle:

Modern anterior paddles. This style of paddle is used in most current model defibrillators and has been popular on portable models for some times.

d) Internal Paddle:

This type of electrode is use during open heart surgical procedures to apply the electrical shock directly to the myocardium, so usually resemble spoons mounted on the ends of long handles which can be easily inserted into a thoracic incision.

The discharge switch is used to fire the defibrillator by energizing the charger transfer relay. In some models, S will be mounted on the front panel, but in most models it will be one of the patient electrode paddles. Some manual facturers actually use two switches in series for the safety. Discharge of the defibrillator cannot occur unless both switches are closed.

For transchest application, low-resistivity electrodes interface material (gel) must be applied between the skin and the thoracic electrodes in order to establish a low-resistivity pathway for electric current flow, as we note this gel is used to ensure an inefficient transfer of charge and reduce any burning of the patient's skin.

Chapter Ten

Testing and implementation

1- Power supply:

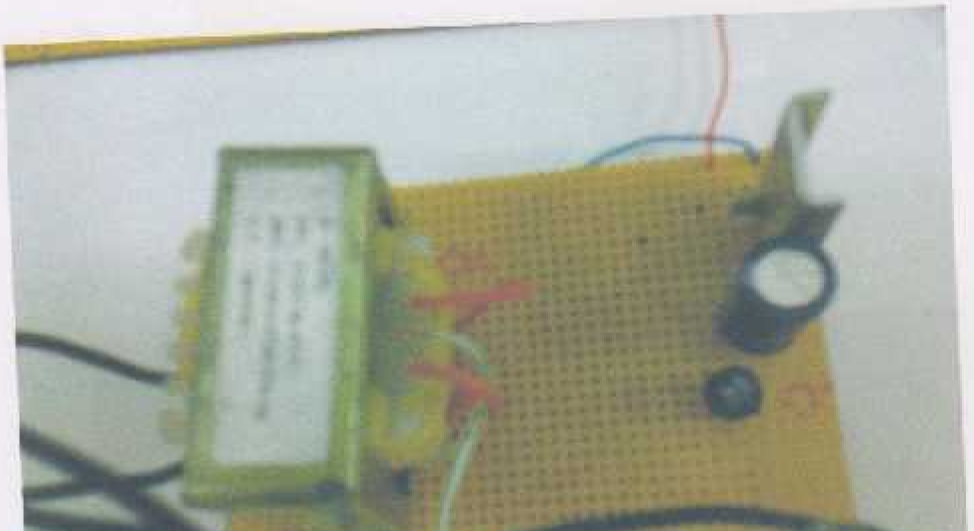


Fig 10.1:Power supply circuit

2- Comparator circuit :

Fig 10.2: Comparator circuit

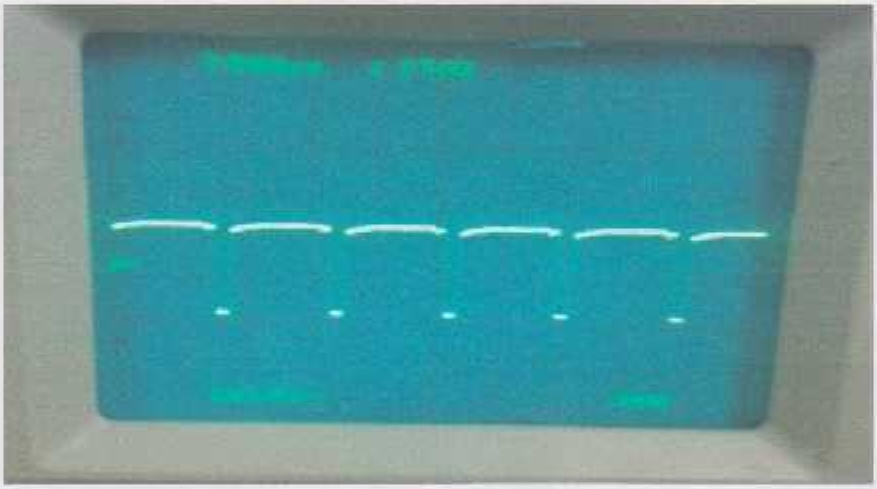


Fig 10.3: Comparator circuit output

3- Monostable multivibrator :

Fig 10.4 : Monostable multivibrator circuit

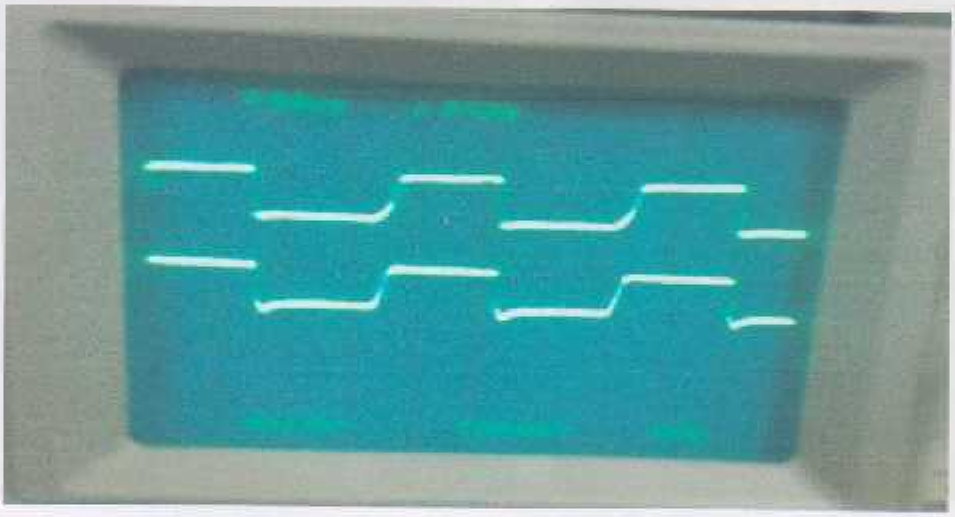


Fig 10.5: Monostable multivibrator output

4- **Charging and discharging circuit:**

Fig 10.6 : Charging and discharging circuit

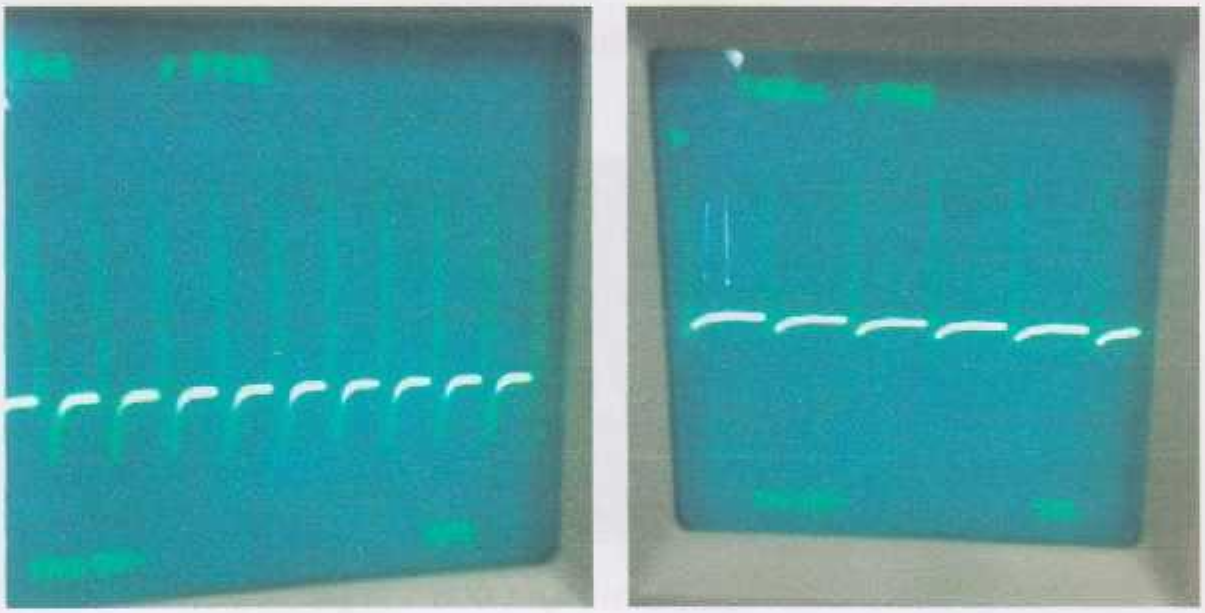


Fig 10.7 : Charging and discharging output

5- ECG Amplifier circuit:

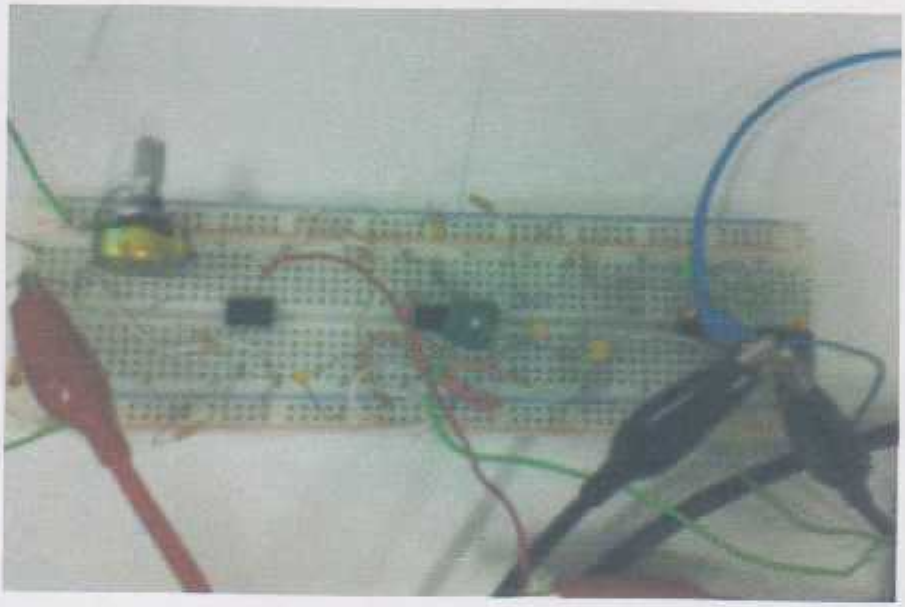


Fig 10.8: ECG Amplifier circuit

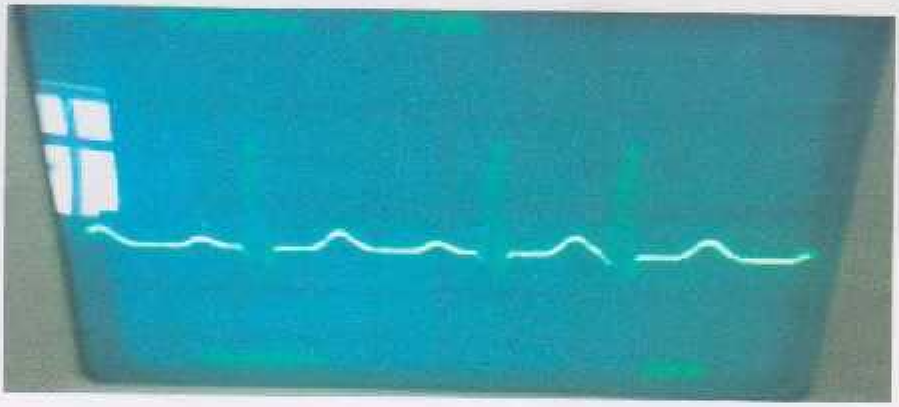


Fig 10.9: ECG Amplifier output put

Fig 10.10 : Complete defibrillator circuit

Chapter Eleven

Conclusion and Future Work

10.1 Conclusion:

1- defibrillator is a Device that used to apply a strong electrical shock to the heart muscle undergoing a fatal arrhythmia, the shocks can restore normal heart rhythms before the malfunctioning heart suffers sudden cardiac arrest, a seizure than can lead to death within minutes.

2- Synchronized Cardioversion refers to an electrical energy discharge that is synchronized with the large R or S wave of the QRS complex Synchronization in the early part of the QRS complex avoids energy delivery in the early phase of repolarization when ventricular fibrillation (VF) can be easily induced.

3- In project you can controlled for the amount of energy delivering to the patient by vary the potentiometer(RV1) in monostable multivibrator circuit.

4- Project circuit will give amount of energy to the patient until detection heart arrhythmia , if not detection it will not give any chock .

10.2 Future Work:

1- Adding LCD screen to view defibrillator wave before Appling to the patient.

2- Adding other circuits (such as seven segment circuit) that use to display the amount of energy delivering to the patient.

3- Design other circuits using to generate other waveform using in defibrillator such as tapered dc delay, and trapezoidal waveforms.

4- Adding biphasic modes in project .

5- Design defibrillator tester to ensure that the amount of energy delivering patient is correct value .

6- Adding unsynchronizing mode in project.

References :

1- American Heart Association journal , Print ISSN: 0009-7322, Copyright © 2003.

2- Implantable Cardioverter Defibrillator Stored ECGs,first edition, 2006 by Luc J. Jordaens and Dominic A.M.J. Theuns.

3- Encyclopedia of Medical Devices and Instrumentation Second Edition ,Volume 2, Editor-in-Chief John G. Webster.

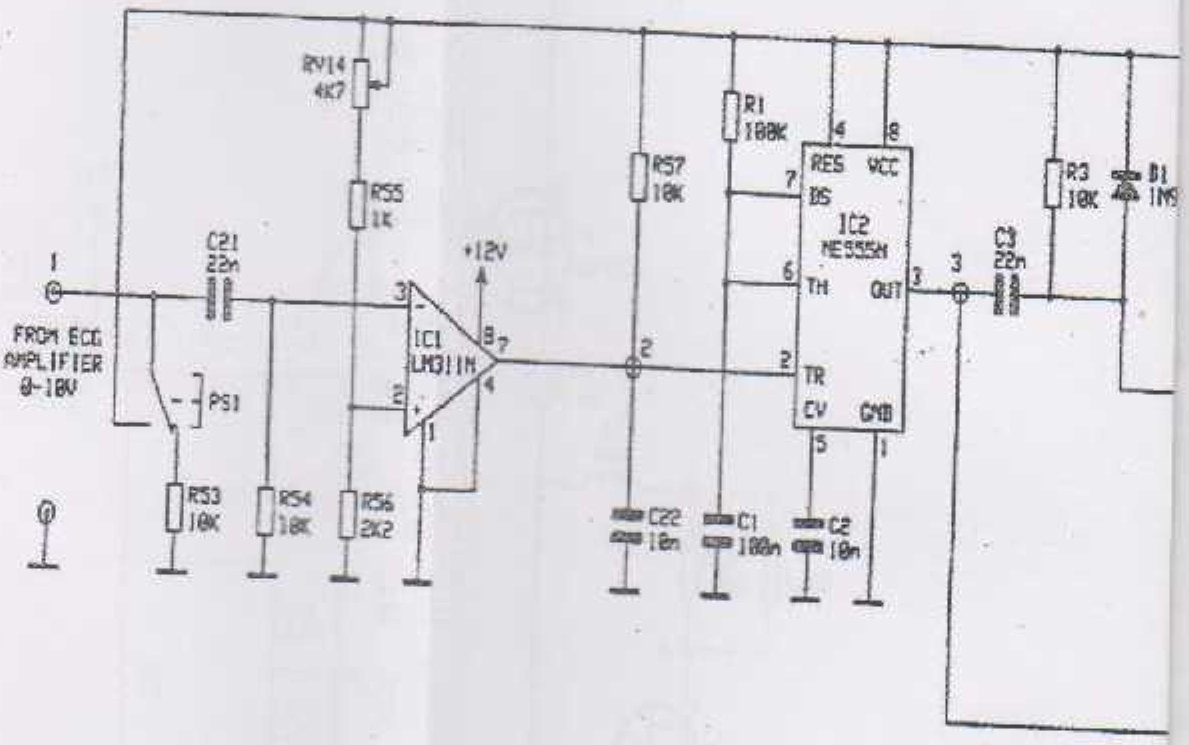
4- The Biomedical Engineering Handbook: Second Edition Ed. Joseph D. Bronzino , chapter 79 and 80.

5- Introduction to Biomedical Equipment Technology (4th Edition): Joseph J. Carr, John M. Brown. Chapter 9.

6- British Heart Journal, I9703 32, 209.by Paul Szekely, N. A. Wynne, D. T. Pearson, G. A. Batson, and D. A. Sideris.

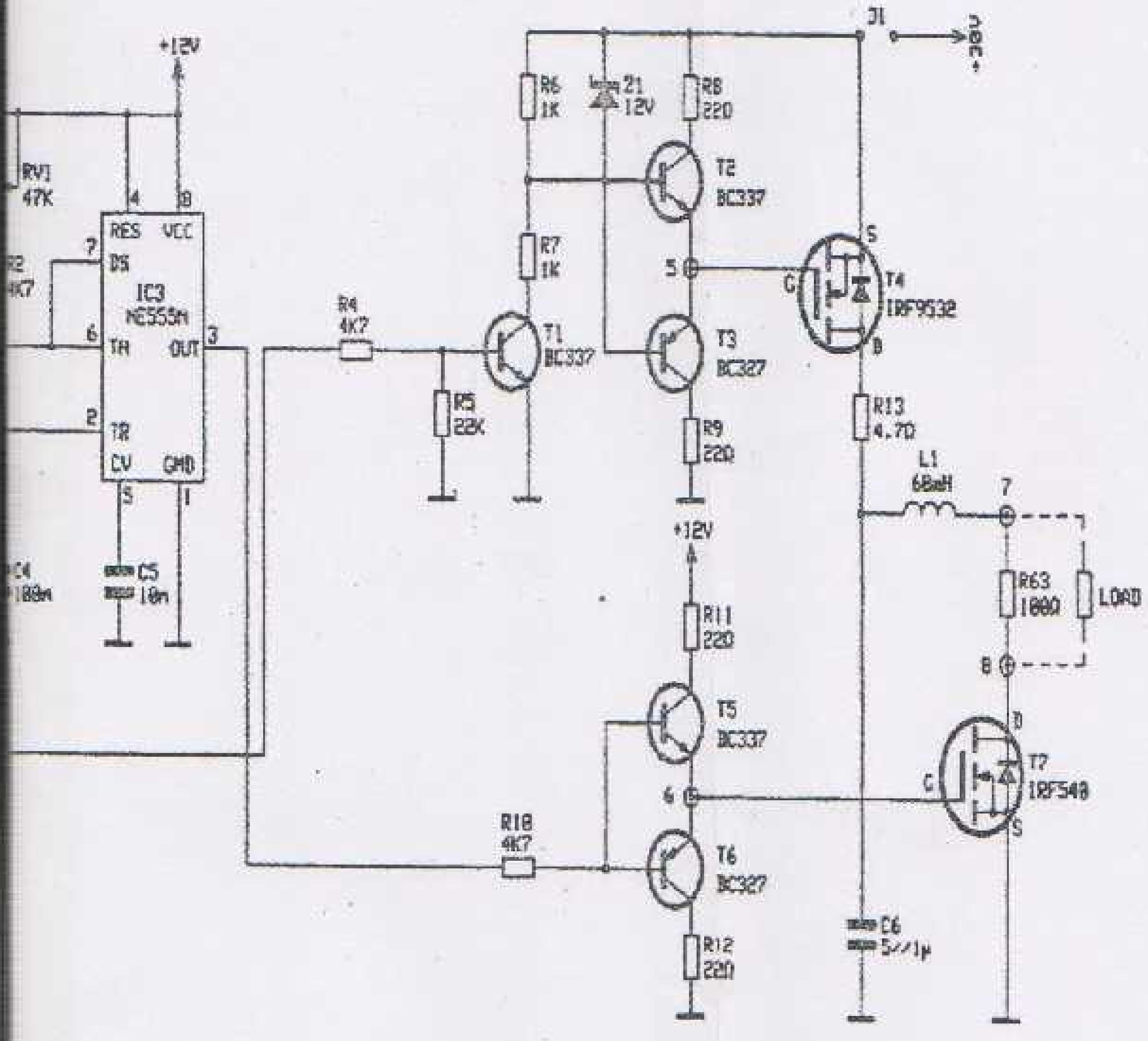
7- Standard handbook of biomedical engineering and design ,first edition, 2003, by myer kutz.

8- Measurement, Instrumentation, and Sensors Handbook CRCnetBase 1999 ,by John G. Webster , chapter 79.



RV14
4K7
R55
1K
C21
22n
1
FROM ECG
AMPLIFIER
0-10V
PS1
R53
10K
R54
10K
R56
2K2
+12V
IC1
LM311N
R57
10K
C22
10n
R1
100K
C1
100n
IC2
NE555N
RES
VCC
DS
TH
OUT
TR
CV
GND
C2
10n
R3
10K
C3
22n
D1

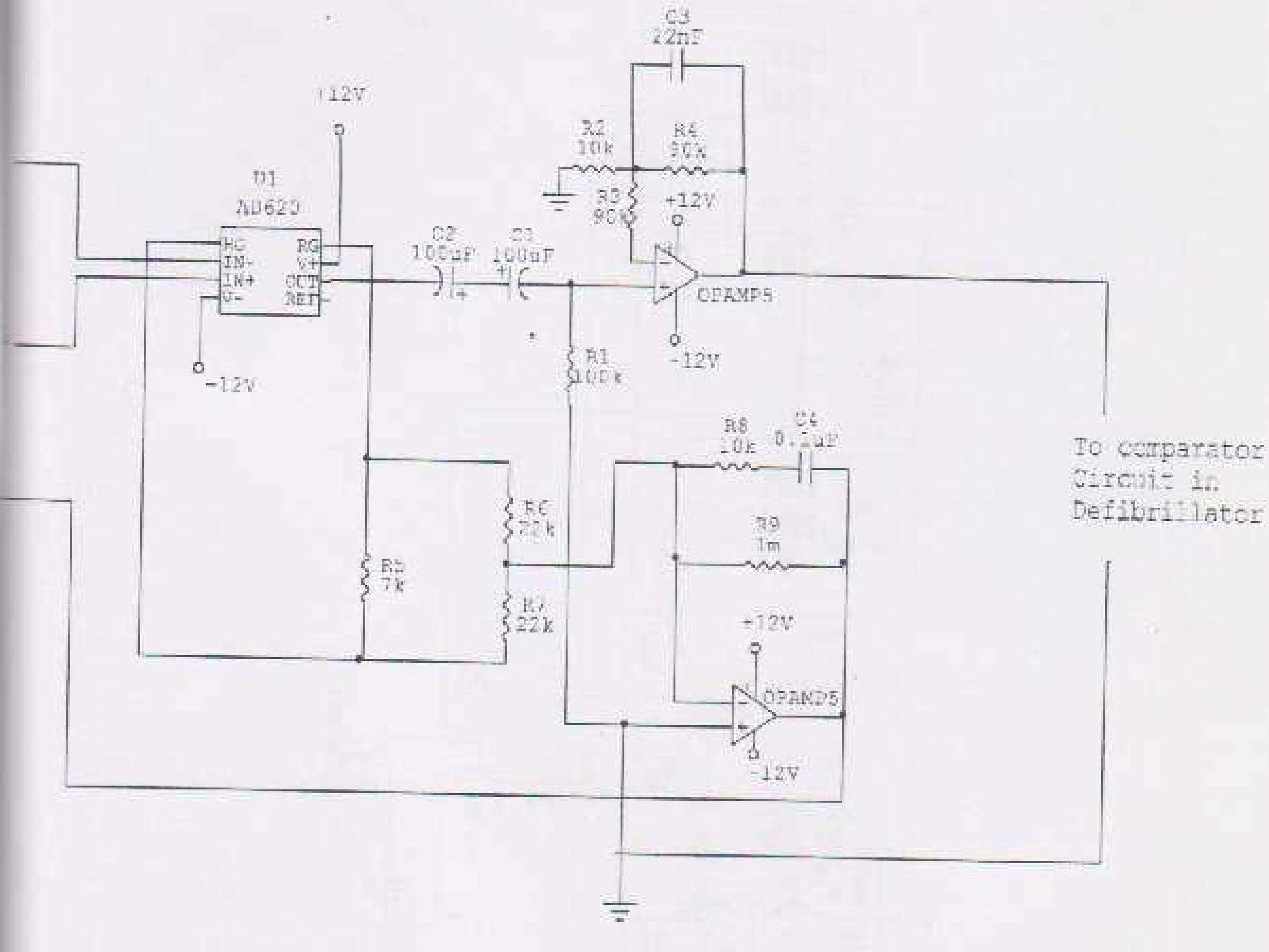
DEFIBRI



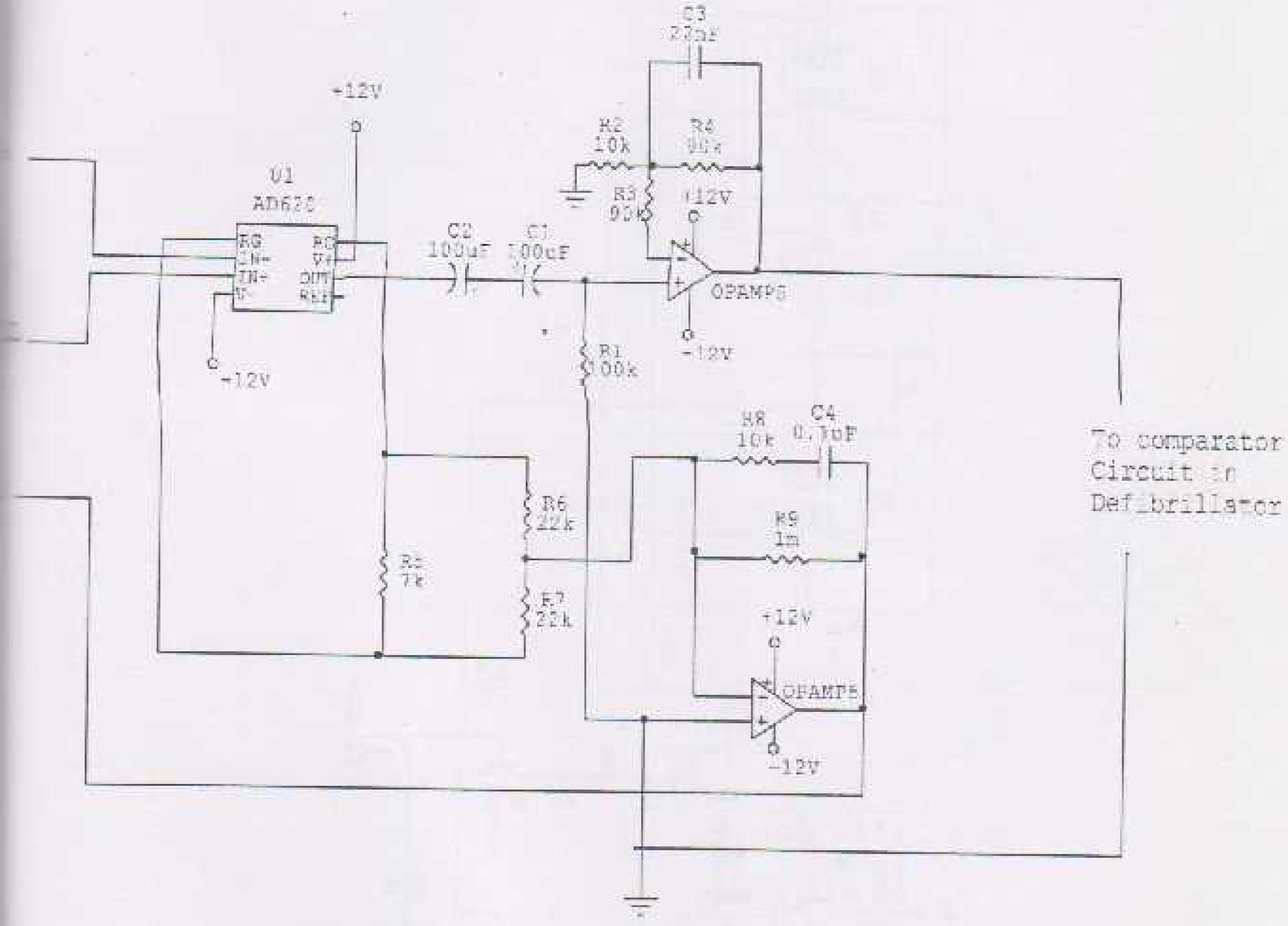
+12V
RV1
47K
R2
4K7
C4
100n
IC3
NE555N
RES
VCC
DS
TH
OUT
TR
CV
GND
C5
10n
R4
4K7
R5
22K
T1
BC337
R6
1K
R7
1K
Z1
12V
R8
22Ω
T2
BC337
T3
BC327
R9
22Ω
+12V
R11
22Ω
T5
BC337
T6
BC327
R18
4K7
R12
22Ω
J1
+Vcc
T4
IRF9532
R13
4.7Ω
L1
68mH
R63
100Ω
LOAD
T7
IRF540
C6
5//1µ

…TOR

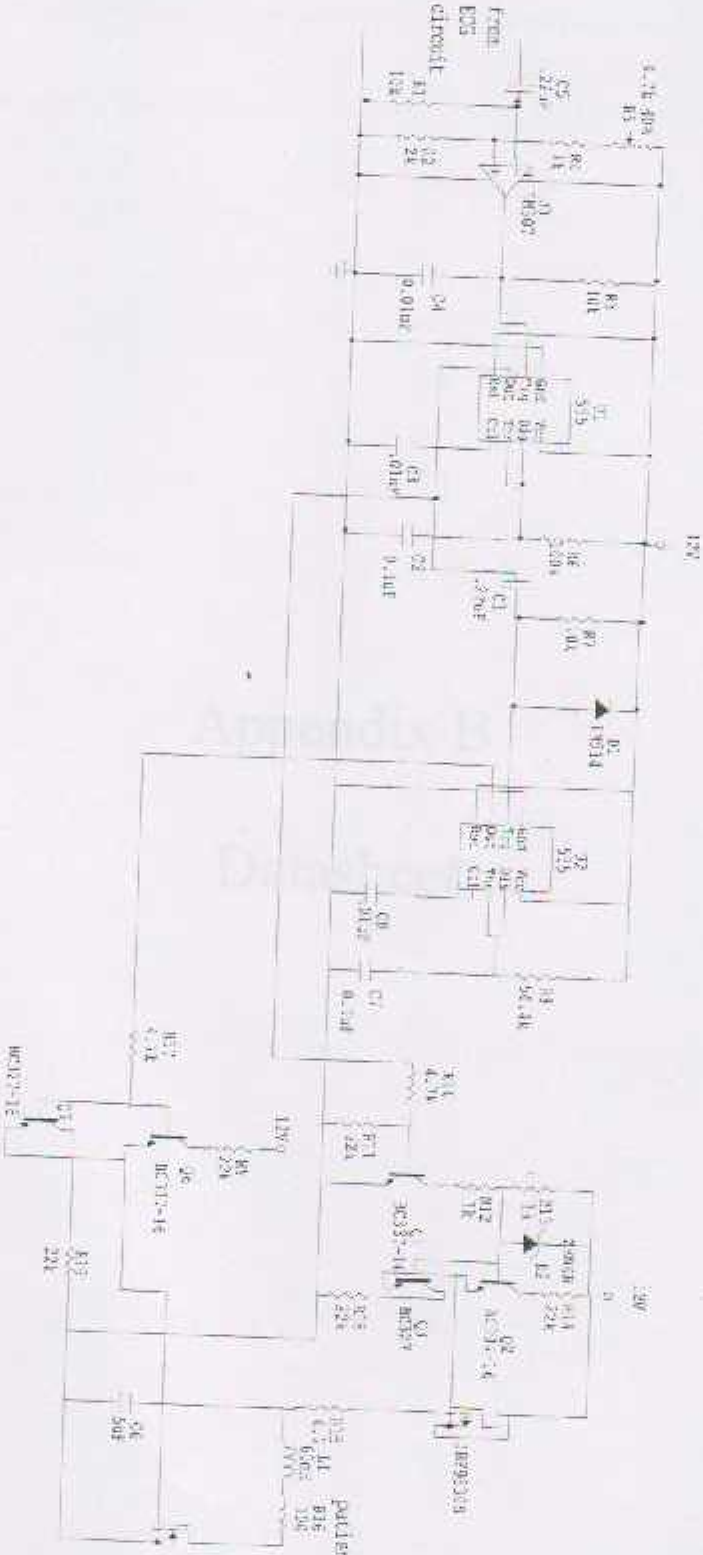
ECG circuit:


C3
22nF
+12V
R2
10k
R4
90k
U1
AD620
R3
90k
+12V
RG
IN-
IN+
V-
RG
V+
OUT
REF
C2
100uF
C1
100uF
OPAMP5
-12V
R1
100k
-12V
R8
10k
C4
0.1uF
To comparator
Circuit in
Defibrillator
R6
22k
R9
1m
R5
7k
R7
22k
+12V
OPAMP5
-12V

ECG circuit:


C3
R2
10k
R4
+12V
U1
R3
C2
100uF
C1
100uF
OPAMP5
-12V
R1
100k
R8
10k
C4
R6
22k
R9
1m
R7
22k
+12V
OPAMP6
-12V
To comparator
Circuit in
Defibrillator

Defibrillator circuit:



Appendix B

Datasheet

DESCRIPTION

The 555 monolithic timing circuit is a highly stable controller capable of producing accurate time delays, or oscillation. In the time delay mode of operation, the time is precisely controlled by one external resistor and capacitor. For a stable operation as an oscillator, the free running frequency and the duty cycle are both accurately controlled with two external resistors and one capacitor. The circuit may be triggered and reset on falling waveforms, and the output structure can source or sink up to 200mA.

PIN CONFIGURATIONS

D, N, FF Packages

F Package

TOP VIEW

FEATURES

- Turn-off time less than 2μs
- Max. operating frequency greater than 500kHz
- Timing from microseconds to hours
- Operates in both astable and monostable modes
- High output current
- Adjustable duty cycle
- TTL compatible
- Temperature stability of 0.005% per °C

APPLICATIONS

- Precision timing
- Pulse generation
- Sequential timing
- Time delay generation
- Pulse width modulation

ORDERING INFORMATION

DESCRIPTION	TEMPERATURE RANGE	ORDER CODE	DWG #
8-Pin Plastic Small Outline (SO) Package	0 to +70°C	NE555D	0174C
8-Pin Plastic Dual In-Line Package (DIP)	0 to +70°C	NE555N	0404B
8-Pin Plastic Dual In-Line Package (DIP)	-40°C to +85°C	SA555N	0404B
8-Pin Plastic Small Outline (SO) Package	-40°C to +85°C	SA555D	0174C
8-Pin Hermetic Ceramic Dual In-Line Package (CERDIP)	-55°C to +125°C	SE555CFE	
8-Pin Plastic Dual In-Line Package (DIP)	-55°C to +125°C	SE555CN	0404B
14-Pin Plastic Dual In-Line Package (DIP)	-55°C to +125°C	SE555N	0405B
8-Pin Hermetic Cerdip	-55°C to +125°C	SE555FE	
14-Pin Ceramic Dual In-Line Package (CERDIP)	0 to +70°C	NE555F	0581B
14-Pin Ceramic Dual In-Line Package (CERDIP)	-55°C to +125°C	SE555F	0581B
14-Pin Ceramic Dual In-Line Package (CERDIP)	-55°C to +125°C	SE555CF	0581B

DC AND AC ELECTRICAL CHARACTERISTICS

T_A = 25°C, V_{CC} = +5V to +15 unless otherwise specified

SYMBOL	PARAMETER	TEST CONDITIONS	SE555 Min	SE555 Typ	SE555 Max	NE555/SE555C Min	NE555/SE555C Typ	NE555/SE555C Max	UNIT
V_{CC}	Supply voltage		4.5		18	4.5		16	V
I_{CC}	Supply current (low state)[1]	V_{CC}=5V R_L=∞		3	5		3	5	mA
		V_{CC}=15V R_L=∞		10	12		10	15	mA
	Timing error (monostable)	R_A=2kΩ to 100kΩ							
t_M	Initial accuracy[2]	C=0.1μF		0.5	2.0		1.0	3.0	%
$\Delta t_M/\Delta T$	Drift with temperature			30	100		50	150	ppm/°C
$\Delta t_M/\Delta V_S$	Drift with supply voltage			0.05	0.2		0.1	0.5	%/V
	Timing error (astable)	R_A, R_B=1kΩ to 100kΩ							
t_A	Initial accuracy[2]	C=0.1μF		4	6		5	13	%
$\Delta t_A/\Delta T$	Drift with temperature	V_{CC}=15V			500			500	ppm/°C
$\Delta t_A/\Delta V_S$	Drift with supply voltage			0.15	0.6		0.3	1	%/V
V_C	Control voltage level	V_{CC}=15V	9.6	10.0	10.4	9.0	10.0	11.0	V
		V_{CC}=5V	2.9	3.33	3.8	2.6	3.33	4.0	V
V_{TH}	Threshold voltage	V_{CC}=15V	9.4	10.0	10.6	8.8	10.0	11.2	V
		V_{CC}=5V	2.7	3.33	4.0	2.4	3.33	4.2	V
I_{TH}	Threshold current[3]			0.1	0.25		0.1	0.25	μA
V_{TRIG}	Trigger voltage	V_{CC}=15V	4.8	5.0	5.2	4.5	5.0	5.6	V
		V_{CC}=5V	1.45	1.67	1.9	1.1	1.67	2.2	V
I_{TRIG}	Trigger current	V_{TRIG}=0V		0.5	0.9		0.5	2.0	μA
V_{RESET}	Reset voltage[4]	V_{CC}=15V V_{TH}=10.5V	0.3		1.0	0.3		1.0	V
I_{RESET}	Reset current	V_{RESET}=0.4V		0.1	0.4		0.1	0.4	mA
	Reset current	V_{RESET}=0V		0.4	1.0		0.4	1.5	mA
V_{OL}	Output voltage (low)	V_{CC}=15V							
		I_{SINK}=10mA		0.1	0.15		0.1	0.25	V
		I_{SINK}=50mA		0.4	0.5		0.4	0.75	V
		I_{SINK}=100mA		2.0	2.2		2.0	2.5	V
		I_{SINK}=200mA		2.5			2.5		V
		V_{CC}=5V							
		I_{SINK}=8mA		0.1	0.25		0.3	0.4	V
		I_{SINK}=5mA		0.05	0.2		0.25	0.35	V
V_{OH}	Output voltage (high)	V_{CC}=15V							
		I_{SOURCE}=200mA		12.5			12.5		V
		I_{SOURCE}=100mA	13.0	13.3		12.75	13.3		V
		V_{CC}=5V							
		I_{SOURCE}=100mA	3.0	3.3		2.75	3.3		V
t_{OFF}	Turn-off time[5]	V_{RESET}=V_{CC}		0.5	2.0		0.5	2.0	μs
t_R	Rise time of output			100	200		100	300	ns
t_F	Fall time of output			100	200		100	300	ns
	Discharge leakage current			20	100		20	100	nA

Voltage comparator

LM111/211/311/ LM311B

DESCRIPTION

The LM111 series are voltage comparators that have input currents approximately a hundred times lower than devices like the μA710. They are designed to operate over a wider range of supply voltages; from standard ±15V op amp supplies down to a single 3V supply. Their output is compatible with RTL, DTL and TTL as well as MOS circuits. Further, they can drive lamps or relays, switching voltages up to 50V at currents as high as 50mA.

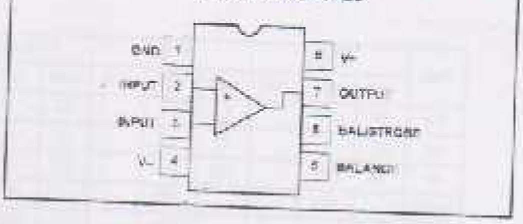
Both the inputs and the outputs of the LM111 series can be isolated from system ground, and the output can drive loads referred to ground, the positive supply or the negative supply. Offset balancing and strobe capability are provided and outputs can be wire-ORed.

Although slower than the μA710 (200ns response time vs 40ns), the devices are also much less prone to spurious oscillations. The LM111 series has the same pin configuration as the μA710 series.

FEATURES

- Operates from single 3V supply (LM311B)
- Maximum input bias current: 150nA (LM311—250nA
- Maximum offset current: 20nA (LM311—50nA)
- Differential input voltage range: ±30V
- Power consumption: 135mW at ±15V
- High sensitivity—200V/mV
- Zero crossing detector

PIN CONFIGURATION



APPLICATIONS

- Precision squarer
- Positive/negative peak detector
- Low voltage adjustable reference supply
- Switching power amplifier

ORDERING INFORMATION

DESCRIPTION	TEMPERATURE RANGE	ORDER CODE	DWG #
8-Pin Plastic Dual In-Line Package (DIP)	-55°C to +125°C	LM111N	0404B
8-Pin Plastic Dual In-Line Package (DIP)	-25°C to +85°C	LM211N	0404B
8-Pin Plastic Small Outline Package (SO)	0 to +70°C	LM311D	0174C
8-Pin Plastic Dual In-Line Package (DIP)	0 to +70°C	LM311N	0404B
8-Pin Plastic Small Outline Package (SO)	-25°C to +85°C	LM211D	0174C
8-Pin Ceramic Dual In-Line Package (CERDIP)	-55°C to +125°C	LM111FE	0580A
8-Pin Plastic Small Outline Package (SO)	0 to +70°C	LM311BD	0174C
8-Pin Plastic Dual In-Line Package (DIP)	0 to +70°C	LM311BN	0404B

DC ELECTRICAL CHARACTERISTICS[1, 2, 3, 5]

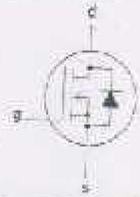
Over temperature range unless otherwise specified.

SYMBOL	PARAMETER	TEST CONDITIONS	LM111/LM211			LM311			LM311B			UNIT
			Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	
V_{OS}	Input offset voltage[3]	T_A=25°C, R_S≤50kΩ		0.7	3.0		2.0	7.5		2.0	7.5	mV
I_{OS}	Input offset current[3]	T_A=25°C		4.0	10		6.0	50		5	25	nA
I_{BIAS}	Input bias current	T_A=25°C		60	100		100	250		100	200	nA
A_V	Voltage gain	T_A=25°C		200			200			200		V/mV
	Response time[4]	T_A=25°C		200			200			500		ns
V_{SAT}	Saturation voltage	LM111/211 V_{IN}≤-5mV, I_{OUT}=50mA; LM311/B V_{IN}≤-10mV, I_{OUT}=50mA; T_A=25°C		0.75	1.5		0.75	1.5		0.75	1.5	V
I_{BALSTR}	Strobe on current	T_A=25°C		3.0			3.0			2.0		mA
$I_{LEAKAGE}$	Output leakage current[6]	LM111/211 V_{IN}≥5mV, V_{OUT}=35V; LM311/B V_{IN}≥10mV, V_{OUT}=35V; T_A=25°C, I_{STROBE}=3mA ($V-$ = V_{GND} = -5V)		0.2	10		0.2	50		0.2	50	nA
V_{OS}	Input offset voltage[3]	R_S≤50kΩ			4.0			10			10	mV
I_{OS}	Input offset current[3]				20			70			50	nA
I_{BIAS}	Input bias current				150			300			250	nA
V_{IN}	Input voltage range	V=±15V (Pin 7 may go to 5V)	-14.5	13.8 to -14.7	13.0	-14.5	13.8 to -14.7	13.0	V− +0.5		V+ −1.5	V
V_{OL}	Saturation voltage[6]	V+ ≥ 4.5V, V- = 0; LM111/211 V_{IN}≤-6mV, I_{SINK}≤8mA; LM311/B V_{IN}≤-10mV, I_{SINK}≤8mA		0.23	0.4		0.23	0.4		0.23	0.4	V
I_{OH}	Output leakage current	V_{IN}≥5mV, V_{OUT}=35V		0.1	0.5							µA
I_{CC}	Positive supply current	T_A=25°C		5.1	6.0		5.1	7.5		1.6	3.5	mA
I_{EE}	Negative supply voltage	T_A=25°C		4.1	5.0		4.1	5.0				mA

FEATURES

- 'Trench' technology
- Low on-state resistance
- Fast switching
- Low thermal resistance

SYMBOL



QUICK REFERENCE DATA

V_{DSS} = 100 V

I_D = 23 A

$R_{DS(ON)} \leq 77$ mΩ

GENERAL DESCRIPTION

N-channel enhancement mode field-effect power transistor in a plastic envelope using 'trench' technology.

Applications:-

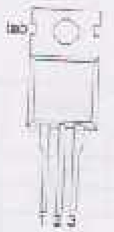
- d.c. to d.c. converters
- switched mode power supplies
- T.V. and computer monitor power supplies

The IRF540 is supplied in the SOT78 (TO220AB) conventional leaded package.
The IRF540S is supplied in the SOT404 (D^2PAK) surface mounting package.

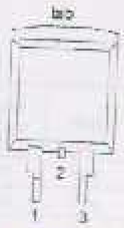
PINNING

PIN	DESCRIPTION
1	gate
2	drain[1]
3	source
tab	drain

SOT78 (TO220AB)



SOT404 (D^2PAK)



LIMITING VALUES

Limiting values in accordance with the Absolute Maximum System (IEC 134)

SYMBOL	PARAMETER	CONDITIONS	MIN.	MAX.	UNIT
V_{DSS}	Drain-source voltage	T_j = 25 °C to 175°C	-	100	V
V_{DGR}	Drain-gate voltage	T_j = 25 °C to 175°C; R_{GS} = 20 kΩ	-	100	V
V_{GS}	Gate-source voltage		-	± 20	V
I_D	Continuous drain current	T_{mb} = 25 °C; V_{GS} = 10 V	-	23	A
		T_{mb} = 100 °C; V_{GS} = 10 V	-	16	A
I_{DM}	Pulsed drain current	T_{mb} = 25 °C	-	92	A
P_D	Total power dissipation	T_{mb} = 25 °C	-	100	W
T_j, T_{stg}	Operating junction and storage temperature		- 55	175	°C

IRF9530/9531/9532/9533
IRFP9130/9131/9132/9133

P-CHANNEL POWER MOSFETS

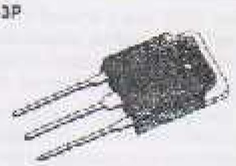
FEATURES

- Lower $R_{DS(ON)}$
- Improved inductive ruggedness
- Fast switching times
- Rugged polysilicon gate cell structure
- Lower input capacitance
- Extended safe operating area
- Improved high temperature reliability

TO-220

IRF9530/9531/9532/9533

TO-3P



IRFP9130/9131/9132/9133

PRODUCT SUMMARY

Part Number	V_{DS}	$R_{DS(on)}$	I_D
IRF9530/IRFP9130	−100V	0.30Ω	−12A
IRF9531/IRFP9131	−60V	0.30Ω	−12A
IRF9532/IRFP9132	−100V	0.40Ω	−10A
IRF9533/IRFP9133	−60V	0.40Ω	−10A

MAXIMUM RATINGS

Characteristic	Symbol	IRF9530 IRFP9130	IRF9531 IRFP9131	IRF9532 IRFP9132	IRF9533 IRFP9133	Unit
Drain-Source Voltage (1)	V_{DSS}	−100	−60	−100	−60	Vdc
Drain-Gate Voltage (R_{GS}=1.0MΩ)(1)	V_{DGR}	−100	−60	−100	−60	Vdc
Gate-Source Voltage	V_{GS}	±20				Vdc
Continuous Drain Current T_C=25°C	I_D	−12	−12	−10	−10	Adc
Continuous Drain Current T_C=100°C	I_D	−7.5	−7.5	−6.5	−6.5	Adc
Drain Current—Pulsed (3)	I_{DM}	−48	−48	−40	−40	Adc
Gate Current—Pulsed	I_{GM}	±1.5				Adc
Single Pulsed Avalanche Energy (4)	E_{AS}	550				mJ
Avalanche Current	I_{AS}	−12				A
Total Power Dissipation @ T_C=25°C Derate above 25°C	P_D	75 0.6				Watts W/°C
Operating and Storage Junction Temperature Range	T_J, Tstg	−55 to 150				°C
Maximum Lead Temp. for Soldering Purposes, 1/8" from case for 5 seconds	T_L	300				°C

ANALOG DEVICES

Low Cost, Low Power Instrumentation Amplifier

AD620

FEATURES

EASY TO USE

- Gain Set with One External Resistor (Gain Range 1 to 1000)
- Wide Power Supply Range (±2.3 V to ±18 V)
- Higher Performance than Three Op Amp IA Designs
- Available in 8-Lead DIP and SOIC Packaging
- Low Power, 1.3 mA max Supply Current

EXCELLENT DC PERFORMANCE ("B GRADE")

- 50 µV max, Input Offset Voltage
- 0.6 µV/°C max, Input Offset Drift
- 1.0 nA max, Input Bias Current
- 100 dB min Common-Mode Rejection Ratio (G = 10)

LOW NOISE

- 9 nV/√Hz, @ 1 kHz, Input Voltage Noise
- 0.28 µV p-p Noise (0.1 Hz to 10 Hz)

EXCELLENT AC SPECIFICATIONS

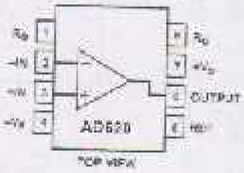
- 120 kHz Bandwidth (G = 100)
- 15 µs Settling Time to 0.01%

APPLICATIONS

- Weigh Scales
- ECG and Medical Instrumentation
- Transducer Interface
- Data Acquisition Systems
- Industrial Process Controls
- Battery Powered and Portable Equipment

CONNECTION DIAGRAM

8-Lead Plastic Mini-DIP (N), Cerdip (Q) and SOIC (R) Packages



PRODUCT DESCRIPTION

The AD620 is a low cost, high accuracy instrumentation amplifier that requires only one external resistor to set gains of 1 to 1000. Furthermore, the AD620 features 8-lead SOIC and DIP packaging that is smaller than discrete designs, and offers lower power (only 1.3 mA max supply current), making it a good fit for battery powered, portable (or remote) applications.

The AD620, with its high accuracy of 40 ppm maximum nonlinearity, low offset voltage of 50 µV max and offset drift of 0.6 µV/°C max, is ideal for use in precision data acquisition systems, such as weigh scales and transducer interfaces. Furthermore, the low noise, low input bias current, and low power of the AD620 make it well suited for medical applications such as ECG and noninvasive blood pressure monitors.

The low input bias current of 1.0 nA max is made possible with the use of Superβeta processing in the input stage. The AD620 works well as a preamplifier due to its low input voltage noise of 9 nV/√Hz at 1 kHz, 0.28 µV p-p in the 0.1 Hz to 10 Hz band, 0.1 pA/√Hz input current noise. Also, the AD620 is well suited for multiplexed applications with its settling time of 15 µs to 0.01% and its cost is low enough to enable designs with one in-amp per channel.

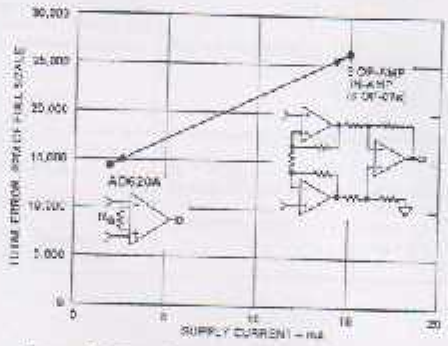


Figure 1. Three Op Amp IA Designs vs. AD620

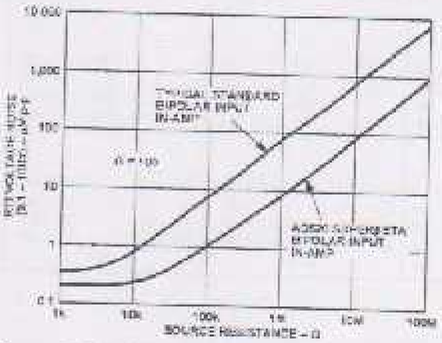


Figure 2. Total Voltage Noise vs. Source Resistance

AD620–SPECIFICATIONS

(Typical @ +25°C, V_S = ±15 V, and R_L = 2 kΩ, unless otherwise noted)

Model	Conditions	AD620A Min	AD620A Typ	AD620A Max	AD620B Min	AD620B Typ	AD620B Max	AD620S[1] Min	AD620S[1] Typ	AD620S[1] Max	Units
GAIN	G = 1 + (49.4 kΩ/R_G)										
Gain Range		1		10,000	1		10,000	1		10,000	
Gain Error[2]	V_{OUT} = ±10 V										
G = 1			0.03	0.10		0.01	0.02		0.03	0.10	%
G = 10			0.15	0.30		0.10	0.15		0.15	0.30	%
G = 100			0.15	0.30		0.10	0.15		0.15	0.30	%
G = 1000			0.40	0.70		0.35	0.50		0.40	0.70	%
Nonlinearity,	V_{OUT} = −10 V to +10 V,										
G = 1–1000	R_L = 10 kΩ		10	40		10	40		10	40	ppm
G = 1–100	R_L = 2 kΩ		10	95		10	95		10	95	ppm
Gain vs. Temperature											
	G = 1			10			10			10	ppm/°C
	Gain > 1[2]			−50			−50			−50	ppm/°C
VOLTAGE OFFSET	(Total RTI Error = V_{OSI} + V_{OSO}/G)										
Input Offset, V_{OSI}	V_S = ±5 V to ±15 V		30	125		15	50		30	125	μV
Over Temperature	V_S = ±5 V to ±15 V			185			85			225	μV
Average TC	V_S = ±5 V to ±15 V		0.3	1.0		0.1	0.6		0.3	1.0	μV/°C
Output Offset, V_{OSO}	V_S = ±15 V		400	1000		200	500		400	1000	μV
	V_S = ±5 V			1500			750			1500	μV
Over Temperature	V_S = ±5 V to ±15 V			2000			1000			2000	μV
Average TC	V_S = ±5 V to ±15 V		5.0	15		2.5	7.0		5.0	15	μV/°C
Offset Referred to the Input vs.											
Supply (PSR)	V_S = ±2.3 V to ±18 V										
G = 1		80	100		80	100		80	100		dB
G = 10		95	120		100	120		95	120		dB
G = 100		110	140		120	140		110	140		dB
G = 1000		110	140		120	140		110	140		dB
INPUT CURRENT											
Input Bias Current			0.5	2.0		0.5	1.0		0.5	2	nA
Over Temperature				2.5			1.5			4	nA
Average TC			3.0			3.0			8.0		pA/°C
Input Offset Current			0.3	1.0		0.3	0.5		0.3	1.0	nA
Over Temperature				1.5			0.75			2.0	nA
Average TC			1.5			1.5			8.0		pA/°C
INPUT											
Input Impedance											
Differential			10‖2			10‖2			10‖2		GΩ‖pF
Common-Mode			10‖2			10‖2			10‖2		GΩ‖pF
Input Voltage Range[3]	V_S = ±2.3 V to ±5 V	$-V_S + 1.9$		$+V_S - 1.2$	$-V_S + 1.9$		$+V_S - 1.2$	$-V_S + 1.9$		$+V_S - 1.2$	V
Over Temperature		$-V_S + 2.1$		$+V_S - 1.3$	$-V_S + 2.1$		$+V_S - 1.3$	$-V_S + 2.1$		$+V_S - 1.3$	V
	V_S = ±5 V to ±18 V	$-V_S + 1.9$		$+V_S - 1.4$	$-V_S + 1.9$		$+V_S - 1.4$	$-V_S + 1.9$		$+V_S - 1.4$	V
Over Temperature		$-V_S + 2.1$		$+V_S - 1.4$	$-V_S + 2.1$		$+V_S - 1.4$	$-V_S + 2.3$		$+V_S - 1.4$	V
Common-Mode Rejection Ratio DC to 60 Hz with 1 kΩ Source Imbalance	V_{CM} = 0 V to ±10 V										
G = 1		73	90		80	90		73	90		dB
G = 10		93	110		100	110		93	110		dB
G = 100		110	130		120	130		110	130		dB
G = 1000		110	130		120	130		110	130		dB
OUTPUT											
Output Swing	R_L = 10 kΩ, V_S = ±2.3 V to ±5 V	$-V_S + 1.1$		$+V_S - 1.2$	$-V_S + 1.1$		$+V_S - 1.2$	$-V_S + 1.1$		$+V_S - 1.2$	V
Over Temperature		$-V_S + 1.4$		$+V_S - 1.3$	$-V_S + 1.4$		$+V_S - 1.3$	$-V_S + 1.6$		$+V_S - 1.3$	V
	V_S = ±5 V to ±18 V	$-V_S + 1.2$		$+V_S - 1.4$	$-V_S + 1.2$		$+V_S - 1.4$	$-V_S + 1.2$		$+V_S - 1.4$	V
Over Temperature		$-V_S + 1.6$		$+V_S - 1.5$	$-V_S + 1.6$		$+V_S - 1.5$	$-V_S + 2.3$		$+V_S - 1.5$	V
Short Current Circuit			±18			±18			±18		mA

Order this document by LM358/D

MOTOROLA

LM358, LM258, LM2904, LM2904V

Dual Low Power Operational Amplifiers

DUAL DIFFERENTIAL INPUT OPERATIONAL AMPLIFIERS

SEMICONDUCTOR TECHNICAL DATA

Utilizing the circuit designs perfected for recently introduced Quad Operational Amplifiers, these dual operational amplifiers feature 1) low power drain, 2) a common mode input voltage range extending to ground/V_{EE}, 3) single supply or split supply operation and 4) pinouts compatible with the popular MC1558 dual operational amplifier. The LM158 series is equivalent to one–half of an LM124.

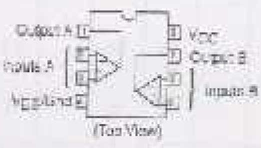
These amplifiers have several distinct advantages over standard operational amplifier types in single supply applications. They can operate at supply voltages as low as 3.0 V or as high as 32 V, with quiescent currents about one–fifth of those associated with the MC1741 (on a per amplifier basis). The common mode input range includes the negative supply, thereby eliminating the necessity for external biasing components in many applications. The output voltage range also includes the negative power supply voltage.

- Short Circuit Protected Outputs
- True Differential Input Stage
- Single Supply Operation: 3.0 V to 32 V
- Low Input Bias Currents
- Internally Compensated
- Common Mode Range Extends to Negative Supply
- Single and Split Supply Operation
- Similar Performance to the Popular MC1558
- ESD Clamps on the Inputs Increase Ruggedness of the Device without Affecting Operation

N SUFFIX
PLASTIC PACKAGE
CASE 626

D SUFFIX
PLASTIC PACKAGE
CASE 751
(SO–8)

PIN CONNECTIONS



MAXIMUM RATINGS (T_A = +25°C, unless otherwise noted.)

Rating	Symbol	LM258 LM358	LM2904 LM2904V	Unit
Power Supply Voltages Single Supply Split Supplies	 V_{CC} V_{CC}, V_{EE}	 32 ±16	 26 ±13	Vdc
Input Differential Voltage Range (Note 1)	V_{IDR}	±32	±26	Vdc
Input Common Mode Voltage Range (Note 2)	V_{ICR}	–0.3 to 32	–0.3 to 26	Vdc
Output Short Circuit Duration	t_{SC}	Continuous		
Junction Temperature	T_J	150		°C
Storage Temperature Range	T_{stg}	–55 to +125		°C
Operating Ambient Temperature Range LM258 LM358 LM2904 LM2904V	T_A	 –25 to +85 0 to +70 – –	 – – –40 to +105 –40 to +125	°C

NOTES: 1. Split Power Supplies.
2. For Supply Voltages less than 32 V for the LM258/358 and 26 V for the LM2904, the absolute maximum input voltage is equal to the supply voltage.

ORDERING INFORMATION

Device	Operating Temperature Range	Package
LM2904D	T_A = –40° to +105°C	SO–8
LM2904N		Plastic DIP
LM2904VD	T_A = –40° to +125°C	SO–8
LM2904VN		Plastic DIP
LM258D	T_A = –25° to +85°C	SO–8
LM258N		Plastic DIP
LM358D	T_A = 0° to +70°C	SO–8
LM358N		Plastic DIP

LM358, LM258, LM2904, LM2904V

ELECTRICAL CHARACTERISTICS (V_{CC} = 5.0 V, V_{EE} = Gnd, T_A = 25°C, unless otherwise noted.)

Characteristic	Symbol	LM258			LM358			LM2904			LM2904V			Unit
		Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	
Input Offset Voltage V_{CC} = 5.0 V to 30 V (26 V for LM2904, V), V_{IC} = 0 V to V_{CC} −1.7 V, V_O ≈ 1.4 V, R_S = 0 Ω T_A = 25°C T_A = T_{high} (Note 1) T_A = T_{low} (Note 1)	V_{IO}	– – –	2.0 – –	5.0 7.0 2.0	– – –	2.0 – –	7.0 9.0 9.0	– – –	2.0 – –	7.0 10 10	– – –	– – –	– 13 10	mV
Average Temperature Coefficient of Input Offset Voltage T_A = T_{high} to T_{low} (Note 1)	$\Delta V_{IO}/\Delta T$	–	7.0	–	–	7.0	–	–	7.0	–	–	7.0	–	μV/°C
Input Offset Current T_A = T_{high} to T_{low} (Note 1)	I_{IO}	–	3.0 –	30 100	–	5.0 –	50 150	–	5.0 45	50 200	–	5.0 45	50 200	nA
Input Bias Current T_A = T_{high} to T_{low} (Note 1)	I_{IB}	–	−45 −50	−150 −300	–	−45 −50	−250 −500	–	−45 −50	−250 −500	–	−45 −50	−250 −500	
Average Temperature Coefficient of Input Offset Current T_A = T_{high} to T_{low} (Note 1)	$\Delta I_{IO}/\Delta T$	–	10	–	–	10	–	–	10	–	–	10	–	pA/°C
Input Common Mode Voltage Range (Note 2) V_{CC} = 30 V (26 V for LM2904, V) V_{CC} = 30 V (26 V for LM2904, V), T_A = T_{high} to T_{low}	V_{ICR}	0 0	– –	28.3 28	0 0	– –	28.3 28	0 0	– –	24.3 24	0 0	– –	24.3 24	V
Differential Input Voltage Range	V_{IDR}	–	–	V_{CC}	–	–	V_{CC}	–	–	V_{CC}	–	–	V_{CC}	V
Large Signal Open Loop Voltage Gain R_L = 2.0 kΩ, V_{CC} = 15 V, For Large V_O Swing, T_A = T_{high} to T_{low} (Note 1)	A_{VOL}	50 25	100 –	–	25 15	100 –	–	25 15	100 –	–	25 15	100 –	–	V/mV
Channel Separation 1.0 kHz ≤ f ≤ 20 kHz, Input Referenced	CS	–	−120	–	–	−120	–	–	−120	–	–	−120	–	dB
Common Mode Rejection R_S ≤ 10 kΩ	CMR	70	85	–	65	70	–	50	70	–	50	70	–	dB
Power Supply Rejection	PSR	65	100	–	65	100	–	50	100	–	50	100	–	dB
Output Voltage-High Limit (T_A = T_{high} to T_{low}) (Note 1) V_{CC} = 5.0 V, R_L = 2.0 kΩ, T_A = 25°C V_{CC} = 30 V (26 V for LM2904, V), R_L = 2.0 kΩ V_{CC} = 30 V (26 V for LM2904, V), R_L = 10 kΩ	V_{OH}	3.3 26 27	3.5 – 28	– – –	3.3 26 27	3.5 – 28	– – –	3.3 22 23	3.5 – 24	– – –	3.3 22 23	3.5 – 24	– – –	V
Output Voltage-Low Limit V_{CC} = 5.0 V, R_L = 10 kΩ, T_A = T_{high} to T_{low} (Note 1)	V_{OL}	–	5.0	20	–	5.0	20	–	5.0	20	–	5.0	20	mV
Output Source Current V_{ID} = +1.0 V, V_{CC} = 15 V	I_{O+}	20	40	–	20	40	–	20	40	–	20	40	–	mA
Output Sink Current V_{ID} = −1.0 V, V_{CC} = 15 V V_{ID} = −1.0 V, V_O = 200 mV	I_{O-}	10 12	20 50	– –	10 12	20 50	– –	10 –	20 –	– –	10 –	20 –	– –	mA μA
Output Short Circuit to Ground (Note 3)	I_{SC}	–	40	60	–	40	60	–	40	60	–	40	60	mA
Power Supply Current (T_A = T_{high} to T_{low}) (Note 1) V_{CC} = 30 V (26 V for LM2904, V), V_O = 0 V, R_L = ∞ V_{CC} = 5 V, V_O = 0 V, R_L = ∞	I_{CC}	– –	1.5 0.7	3.0 1.2	– –	1.5 0.7	3.0 1.2	– –	1.5 0.7	3.0 1.2	– –	1.5 0.7	3.0 1.2	mA