

Chapter 2

The Evolution of Pacemakers: An Electronics Perspective

Since the first artificial pacemaker was introduced in 1932, much has changed and will continue to change in the future [1], [2], [3]. The complexity and reliability in modern pacemakers has increased significantly, mainly due to developments in integrated circuit design. Early pacemakers merely paced the ventricles asynchronously, not having the capability of electrogram sensing. Later devices, called demand mode pacemakers, included a sense amplifier measuring cardiac activity, thereby avoiding competition between paced and intrinsic rhythms. By the introduction of demand pacemakers, also the longevity increased since pacing stimuli were only delivered when needed. In 1963, pacemakers were introduced having the capability to synchronize ventricular stimuli to atrial activation. Since that time, clinical, surgical and technological developments have proceeded at a remarkable pace providing the highly reliable, extensive therapeutic and diagnostic devices that we know today.

Modern pacemaker topologies are extremely sophisticated and include an analog part (comprising the sense amplifier and a pacing output stage) as well as a digital part (consisting of a micro controller, and some memory), implementing diagnostic analysis of sensed electrograms, adaptive rate response and device programmability. Pacemakers have become smaller and lighter over the years. Early devices weighed more than 180 g, whereas today devices are available weighting no more than 25 g [4]. This weight reduction has occurred partly due to the development of high energy-density batteries. Finally, there have been remarkable advances in cardiac lead technology. Novel electrode tip materials and configurations have provided extremely low stimulation thresholds and low polarization properties [5]. In this chapter, we will concentrate on the evolution of analog circuit designs applied in cardiac pacemakers.

2.1 The heart

In order to better understand why some patients require pacemakers and how these devices work, it is worth briefly discussing how the heart and its electrical system work. In a global view one can think of the heart as a pumping station which pumps the blood through the body. In order to do so, the heart is divided into four chambers: two atria and two ventricles, as shown in Fig. 2.1. The two atria act as collecting reservoirs (primer pump) for blood returning to the heart while the two ventricles act as pumps to eject the blood to the body. Deoxygenated blood returning from the body via the superior and inferior vena cava, enters the right atrium and passes through the tricuspid valve to the right ventricle, which expels it through the pulmonary artery to the lungs. Oxygenated blood returning from the lungs enters the left atrium via the pulmonary veins, passes via the mitral valve to the left ventricle and is pumped out through the aorta back to the body. The tricuspid and the mitral valves are important to prevent the back flow of blood from the respective ventricle to the atrium [6].

The pumping action starts with the simultaneous contraction of the two atria, called “diastole”. This contraction serves to give an added push to get the blood into the ventricles. Shortly after that, the ventricles contract virtually in unison, making the beginning of “systole”. Therefore, from electrical point of view the heart can, in most instances, be treated as just two chambers.

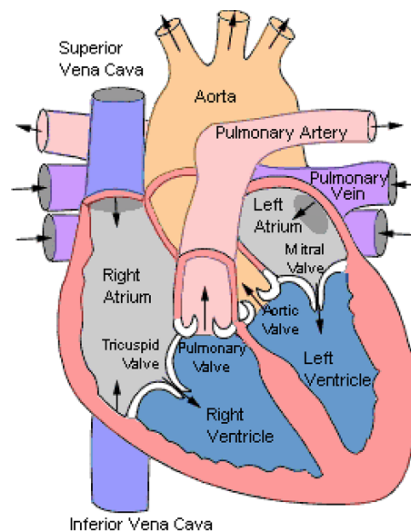


Fig. 2.1. The heart [6]

Excitation and conduction system

The heart is composed basically by three types of cardiac muscle: the atrial muscle, the ventricle muscle and the specialized fibers (which can be further subdivided into excitation and conduction fibers). Both the atrial and the ventricle muscles, which make up the myocardium, each have a similar form of contraction to other muscles of the body, but with a larger period of contraction time. On the other hand, the excitation and conduction fibers have very weak contractions, but do have rhythmicity and variable conduction speed. Once an electrical activation has occurred, contraction of the muscle follows. An orderly sequence of activation of the cardiac muscle in a regularly timed manner is critical for the normal functioning of the heart.

The excitation and conduction system of the heart, responsible for the control of the regular pumping of the heart, is presented in Fig. 2.2. It consists of the sinoatrial node (SA node), internodal tracks and Bachmann's bundle, the atrioventricular node (A-V node), the bundle of His, bundle branches and Purkinje fibers. A heart pacemaker is a device that exhibits automaticity, i.e., generates electrical impulses (known as action potentials) via automatic self-activation, and delivers them to the muscles of the heart in such a way that these muscles contract and the heart beats. Several cells in the heart are able to generate inherent impulses (inherent rate of the A-V node is about 50 beats per minute and of the Purkinje fibers about 40 beats per minute), but at a lower rate than the SA node (which produces about 60–80 beats per minute). The normal rhythm of the heart, between 60 and 100 beats per minute, is controlled by the discharges from the SA node, unless the SA node is non-functional. The Internodal tracks and Bachmann's bundle transmit this excitation throughout the atria and initiate a coordinated contraction of the atrial walls. Meanwhile, the impulse reaches the A-V node, which is the only electrical connection

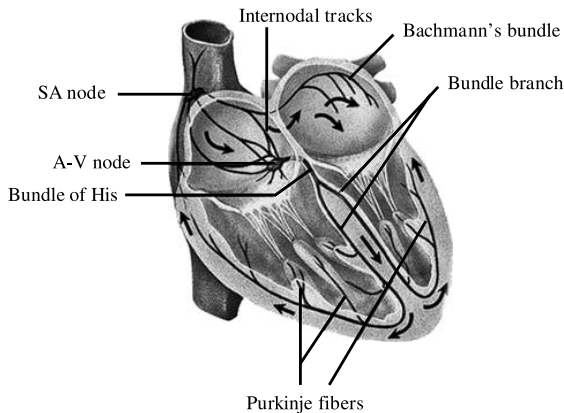


Fig. 2.2. The cardiac conduction system [8]

between atria and ventricles. The A-V node introduces an effective delay, allowing the contraction of the atria to finish before ventricular contraction begins. By this delay, an optimal ventricular filling is achieved. Subsequently, the electrical impulses are conducted very rapidly through the His–Purkinje system (comprising the bundle of His, bundle branches and Purkinje fibers). Once the bundle of His activates, its signal splits into the right bundle branch, which goes to the right ventricle, and the left bundle branch which leads to the left ventricle. Both bundle branches terminate in Purkinje fibers. The Purkinje fibers are responsible for spreading the excitation throughout the two ventricles and causing a coordinated ventricular contraction [6].

2.2 Cardiac signals

2.2.1 Surface electrocardiogram

The electrocardiogram (ECG) is the recording on the body surface of the electrical activity generated by heart. It was originally observed by Waller in 1899 [7]. In 1903, Einthoven introduced some concepts still in use today, including the labelling of the various waves. He chose the letters from P to U to label the waves and to avoid conflict with other physiologic waves being studied at that time [7]. Figure 2.3 depicts a typical ECG signal.

ECG signals are typically in the range of 2 mV peak-to-peak and occupy a bandwidth of 0.05–150 Hz. The character of the body surface waves depends on the amount of tissue activating at one time and the relative speed, direction of the activation waveform (action potentials) and the position of the electrodes. Therefore, the pacemaker potentials, i.e. the SA node potentials, that are generated by a small tissue mass are not seen on the ECG. The first ECG wave of the cardiac cycle is the P wave, and it represents the

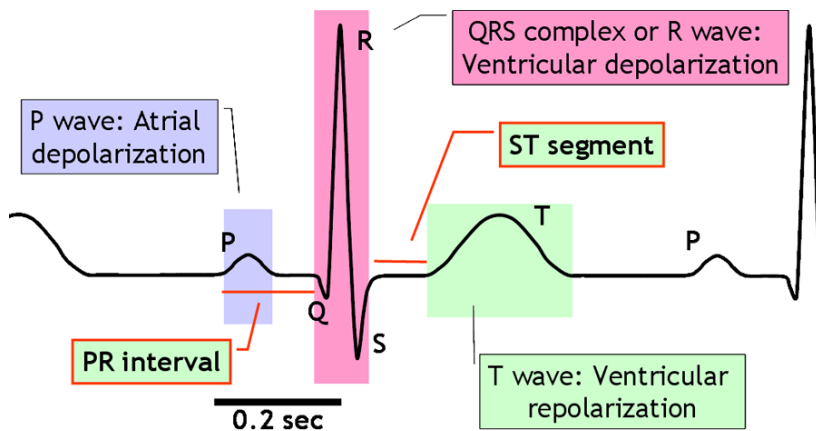


Fig. 2.3. Typical electrocardiogram

depolarization of the atria. Conduction of the cardiac impulse proceeds from the atria through a series of specialized cardiac cells (the A-V node and the His-Purkinje system) as explained in the previous section. Again the total mass is too small to generate a signal large enough to be seen on the ECG. There is a short relatively isoelectric segment following the P wave. This is the P-Q interval which is defined by the propagation delay time of the specialized cells and, usually, is 0.2 s. Once the large muscle mass of the ventricles is excited, a rapid and large deflection is seen on the body surface. This ventricles' depolarization waveform is generically called the QRS complex. Following the QRS complex is another isoelectric segment, the S-T interval. The S-T interval represents the duration of the action potential, normally about 0.25 s to 0.35 s. After this short segment, the ventricles return to their electrical resting state, and a wave of repolarization is seen as a low-frequency signal known as the T wave. In some individuals, a small peak occurs at the end or after the T wave and is called the U wave. Its origin has never been fully established, but it is believed to be a repolarization potential [8].

2.2.2 Intracardiac electrogram (IECG)

An intracardiac electrogram (IECG) is a recording of changes in electric potentials of specific cardiac locations measured by electrodes placed within or onto the heart by using cardiac catheters. The IECG can be recorded between one electrode and an indifferent electrode, usually more than 10 cm apart (unipolar electrogram) or between two more proximate electrodes (<15 mm) in contact with the heart (bipolar electrogram). Sensing the intrinsic activity of the heart depends on many factors related to the cardiac source and the electrode-tissue interface where complex electrochemical reactions take place. In most situations, it is desirable that the IECG does not contain signals from other more distant cardiac chambers. Bipolar lead systems are much less sensitive to far-field potentials and electromagnetic interference (EMI) sources obscuring the cardiac signal.

2.2.3 Cardiac diseases – arrhythmias

Arrhythmias (or dysrhythmias) are due to cardiac problems producing abnormal heart rhythms. In general arrhythmias reduce hemodynamic performance, including situations where the heart's natural pacemaker develops an abnormal rate or rhythm or when normal conduction pathways are interrupted and a different part of the heart takes over control of the rhythm. An arrhythmia can involve an abnormal rhythm increase (tachycardia; >100 bpm) or decrease (bradycardia; <60 bpm), or may be characterized by an irregular cardiac rhythm, e.g. due to asynchrony of the cardiac chambers. An "artificial pacemaker" can restore synchrony between the atria and ventricles.

2.3 The history and development of cardiac pacing

2.3.1 What is an artificial pacemaker?

An artificial pacemaker is a device that delivers a controlled, rhythmic electric stimulus to the heart muscle in order to maintain an effective heartbeat for long periods of time and thereby ensures the pumping capacity of the heart. Indication for permanent pacemaker implantation and the selection of the appropriate pacemaker mode are based mainly on the cardiac diseases such as failure of impulse formation (sick sinus syndrome) and/or conduction (A-V block). Functionally, a pacemaker comprises at least three parts: a electrical pulse generator, a power source (battery) and an electrode (lead) system, as we can see in Fig. 2.4 [9].

Different types of output pulses (monophasic, biphasic, etc.) can be used to stimulate the heart. The output stimulus provided by the pulse generator is the amount of electrical charge transferred during the stimulus (current). For effective pacing, the output pulse should have an appropriate width and sufficient energy to depolarize the myocardial cells close to the electrode. Generally, a pacemaker can provide a stimulus in both chambers of the heart. During A-V block, ventricular pacing is required because the seat of disease is in the A-V node or His–Purkinje system. However, in case of a sick sinus syndrome, the choice of pacemaker will be one that will stimulate the right atrium. A pacemaker utilizes the energy stored in batteries to stimulate the heart. Pacing is the most significant drain on the pulse generator power source. The battery capacity is commonly measured in units of charge (ampere-hours). Many factors will affect the longevity of the battery, including primary device settings like pulse amplitude and duration and pacing rate. An ideal pulse generator battery should have a high energy density, low self-discharge rate and sufficient energy reserve between early signs of depletion and full depletion to allow for safe replacement of the device. The electrical connection between the heart and the implanted pulse generator is provided by an implantable electrode catheter called ‘lead’. In an implantable pulse generator system, commonly two types of lead systems are used. A unipolar lead system has a single isolated conductor with an electrode located at the tip. A bipolar

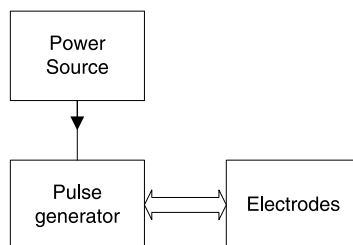


Fig. 2.4. Basic pacemaker functional block diagram

lead has two separate and isolated conductors connecting the two electrodes, i.e. the anode and cathode, usually not more than 15 mm apart. The cathode refers to the electrode serving as the negative pole for delivering the stimulation pulse and the anode to the positive pole. For unipolar pacing-sensing systems, the distance between anode and cathode easily exceeds 10 cm. Its cathode is typically located at the lead tip whereas the pulse generator housing, usually located in the pectoral region, is used as anode. Several types of bipolar leads exist, including the coaxial lead allowing a diameter in the range of 4–5 F (french = 0.33 mm), which is comparable to state-of-the-art unipolar leads. The sensing behavior of bipolar lead systems outperform their unipolar counterparts by providing a better signal to interference ratio. Especially for sensing atrial activation, bipolar electrodes are less sensitive to far-field potentials generated by the ventricles. Moreover, bipolar leads are less sensitive to electromagnetic interference (EMI) sources and skeletal muscle potentials. However, owing to their construction, bipolar leads are stiffer and more complex from a mechanical construction point of view.

2.3.2 Hyman’s pacemaker

In the early nineteenth century, many experiments such as drug therapy and electrical cardiac pacing had been conducted for stimulating the heart in cardiac arrest. Previous methods employed in electrically stimulating the heart were performed by applying the same current that would cause contraction of the muscle tissue of the heart. Later, Albert S. Hyman stated that “the introduced electric impulse serves no other purpose than to provide a controllable irritable point from which a wave of excitation may arise normally and sweep over the heart along its accustomed pathways.” Hyman designed the first experimental heart pacemaker in 1932 [10], shown in Fig. 2.5.

Hyman’s pacemaker was powered by a hand-wound, spring-driven generator that provided 6 minutes of pacemaking without rewinding. Its operation is as follows: The hand crank (F) winds the spring motor (D) which drives the magneto-generator (A) at a controlled speed (E and H) and causes the interrupter disc (not shown) to rotate. The magneto-generator supplies current to a surface contact on the interrupter disc. The companion magnet pieces (B’ and B”) provides the magnetic flux necessary to generate current in the magneto-generator. Subsequently, the interrupter disc produces a pulsed current at 30, 60 or 120 beats per minute, regulated by the impulse controller (G), which represents the periodic pacing waveform delivered to the electrode needle (L). The neon lamp (C) is illuminated when a stimulus is interrupted. In Fig. 2.6 a suitable block diagram of the Hyman’s pacemaker is given [11].

2.3.3 Dawn of a modern era – implantable pacemakers

The origin of modern cardiac pacing is defined as the time when the first pacemaker was implanted without the need for opening the chest. The first

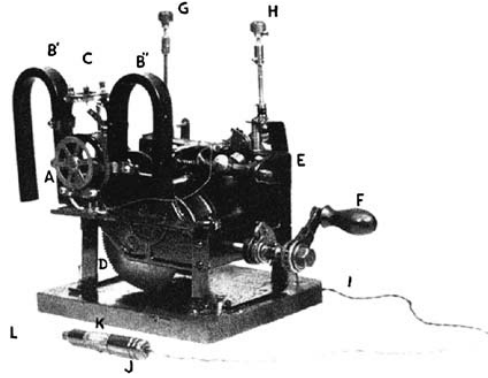


Fig. 2.5. The first artificial pacemaker [10]. A, magneto-generator; B' and B'', companion magnet pieces; C, neon lamps; D, spring motor; E, ballistic governor; F, handle; G, impulse control; H, speed control; I, flexible electric cord; J, insulated handle; K, handle switch, and L, electrode needle

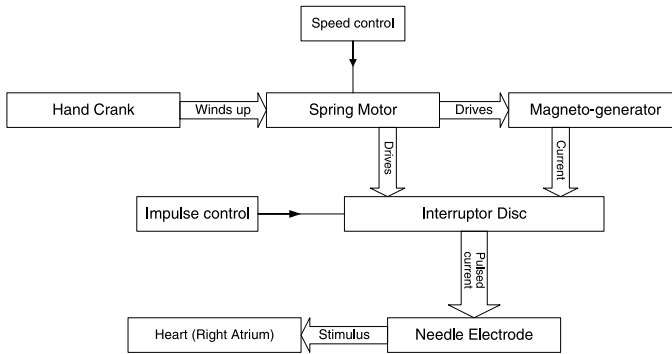


Fig. 2.6. Block diagram of Hyman's pacemaker [11]

pacemaker, developed by Dr. Rune Elmqvist, was used in a patient in 1958 by Dr. Ake Senning [12]. In 1959 the engineer Wilson Greatbatch and the cardiologist W. M. Chardack developed the first fully implantable pacemaker [13]. This device was essentially used to treat patients with complete A-V block caused by Stokes–Adams diseases and delivered single-chamber ventricular pacing. It measured 6 cm in diameter by 1.5 cm thick and the total weight of the pacemaker was approximately 180 g. The pacemaker circuit delivered to the electrode pulses 1ms wide, a pulse amplitude of 10 mA and a repetition rate of 60 beats per minute. The average current drain of the circuit under these conditions was about 12 μ A, which, energized by 10 mercury–zinc cells, gave a continuous operation life estimated at 5 years. The schematic of the

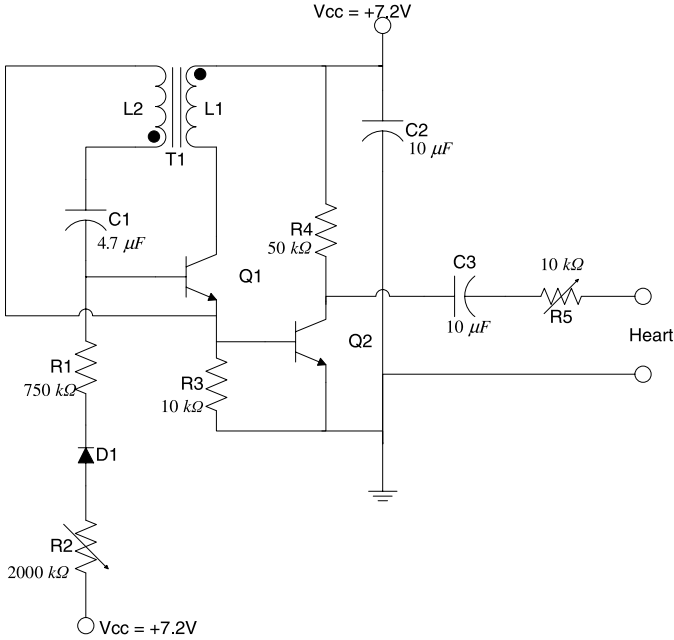


Fig. 2.7. Schematic of the first implanted pacemaker [13]

implanted pacemaker is shown in Fig. 2.7 and consists of a (square) pulse forming oscillator and an amplifier.

Basically, the cardiac pacemaker includes a blocking oscillator [14], which is a special type of wave generator used to produce a narrow pulse. The blocking oscillator is closely related to the two-transistor astable circuit, except that it uses only one amplifying device. The other is replaced by a pulse transformer, which provides inductive regenerative positive feedback. The transistor of the blocking oscillator is normally cut off between pulses and conducting during the time that a pulse is being generated. The operation of a blocking oscillator during a single cycle may be divided into three parts: the turn-on period, the pulse period and the time interval between adjacent pulses (relaxation period).

The turn-on period (t_0) occurs when V_{cc} is applied to the circuit, R1 and R2 provide forward bias and transistor Q1 conducts. Current flow through Q1 and the primary (L1) of T1 induces a current through the secondary (L2), increasing the voltage across C1 and thus across the base-emitter junction of Q1. The positive voltage of L2 is coupled to the base of the transistor through C1. This provides more collector current and consequently more current through L1. Very rapidly, sufficient voltage is applied to saturate the base of Q1. Once Q1 becomes saturated, the circuit can be defined as a series RL (resistance-inductance) circuit and the current increase in L1 is determined by the time constant of L1 and the total series resistance. From t_0 to t_1 (pulse

period) the voltage across L1 will be approximately constant as long as the current increase through L1 is linear. The pulse width depends mainly on the time constant $\tau_C = L1/R3$. At time t_1 , L1 saturates. At this time, C1, which has charged during the pulse period, will now discharge through R1 and cut off Q1. This causes its collector current to stop, and, as a consequence, the voltage across L1 returns to 0. The length of the time between t_1 and t_2 is the relaxation period.

Demand pacemaker

As was shown in the previous section, the early pacing devices simply delivered a fixe-rate pulse to the ventricle at a preset frequency, regardless of any spontaneous activity of the heart. These pacemakers, called asynchronous or fixed rate, compete with the natural heart activity and can sometimes induce arrhythmias or ventricular fibrillation. By adding a sensing amplifier to the asynchronous pacemaker in order to detect intrinsic heart activity and thus avoid this competition, one obtains a demand pacemaker, which provides electrical heart-stimulating impulses only in the absence of natural heartbeat. The other advantage of the demand pacemaker compared to the fixed rate system is that now the battery life of the system is prolonged because it is only activated when pacing stimuli are needed.

Berkovits introduced the demand concept in June 1964, which is the basis of all modern pacemakers. In Fig. 2.8 a suitable block diagram of a demand pacemaker is given. Intracardiac electrodes of conventional demand pacemakers serve two major functions, namely pacing and sensing. Pacing is achieved by the delivery of a short, intense electrical pulse to the myocardial wall where the distal end of the electrode is attached, similarly as in the early pacing devices. However the same electrode is used to detect the intrinsic activity of the heart (e.g., R-waves in the ventricle). The electrical pulse generator consists of the following components: a sense amplifier circuit, a timing control circuit and an output driver circuit (electrical impulse former).

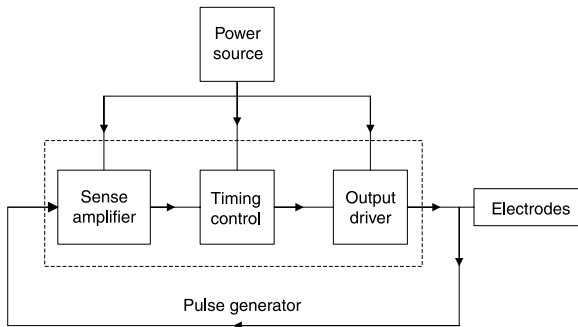


Fig. 2.8. Basic demand pacemaker functional block diagram

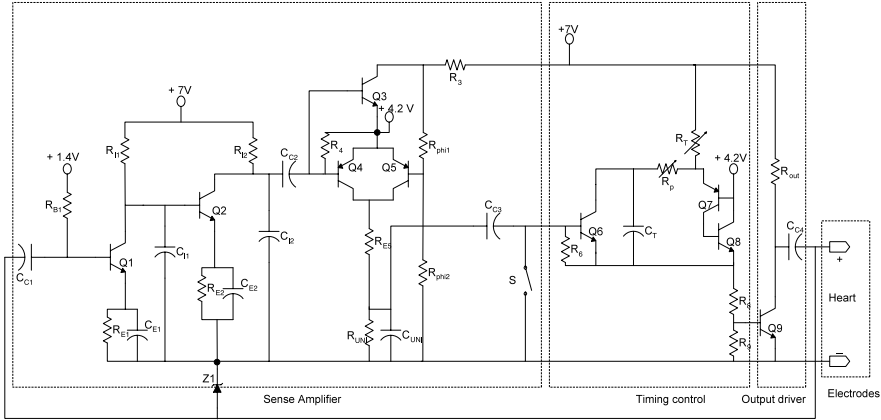


Fig. 2.9. Schematic of the pulse generator of the first demand pacemaker [15]

The schematic of the pulse generator designed by Berkovits is given in Fig. 2.9 [15]. The general function of this circuit was to make the timing circuit responsive to cardiac activity. This allowed inhibition of the pacing pulse from the pulse generator whenever the heart beats on its own. To achieve such function the sense amplifier played a fundamental role. It was designed to amplify and normalize the cardiac signal. Also, the sense amplifier was configured to filter out the undesired signals such as P and T wave and 60 Hz stray signals. The electrical signals picked up by the electrodes are coupled by capacitor C_{C1} into the input of the sense amplifier. The first two transistors, Q1 and Q2, are class A amplifiers. The maximum gain of this amplifier stage is above 50. AC signals at the collector of Q2 are coupled through capacitor C_{C2} to the bases of both transistors Q3 and Q4. The circuit is symmetrically responsive to negative or positive inputs, since signals of positive polarity turn on Q3 and signals of negative polarity turn on Q4. Either transistor momentarily turns on Q6 which is the timing capacitor (C_T) discharge switch.

A bandpass filter with a bandwidth of 20–30 Hz was incorporated in the sense amplifier. Three differentiators (R_{B1} and C_{C1} , R_{E1} and C_{E1} , and R_{E2} and C_{E2}) limited the low frequency response of the detecting circuit to discriminate against the P and T waves and any other frequencies well below 20 Hz. Two integrators (R_{I1} and C_{I1} , and R_{I2} and C_{I2}) were designed to reduce high frequency noise components well above 30 Hz. However, these filters were not totally effective in preventing the triggering of Q6 by 60 Hz signals. For this reason, a rate discrimination circuit (comprising transistors Q3, Q4 and Q5, resistors R_{E5} and R_{UN1} , and capacitors C_{UN1} and C_{C3}) was provided.

The rate discrimination stage had two functions. First, to provide unipolar current pulses (and thus act as a rectifier) of constant magnitude independent of the amplitude of the input signals above a threshold value (1 V at the

bases of Q3 and Q4). A phase inverter circuit (Q5, R_{phi1} and R_{phi2}) was provided to invert the polarity signal from transistor Q3. Second, to provide rate discrimination, which avoided triggering of Q6 by signals occurring at a rate greater than a minimum value. The 60 Hz signals have a rate of 120 pulses per second which is much greater than 72 pulses per minute. Each pulse fully charged C_{UNI} and the next pulse was delivered before the capacitor had an opportunity to discharge to any meaningful extent and the increase in the capacitor voltage was negligible. Consequently, steps of negligible magnitude were transmitted through capacitor C_{C3} to the base of transistor Q6.

Switch S was used only to define the operation mode of the system, free-running mode (switch closed) or demand mode (switch opened). In free-running mode, the switch was closed and, therefore, transistor Q6 remained in cut-off condition. When the switch was opened, i.e., in the case of the pacemaker required to operate in demand mode, each pulse transmitted through capacitor C_{C3} to the base of transistor Q6 caused the transistor to conduct. Capacitor C_{T} discharged through the collector-emitter circuit of the transistor. In this case, the timing cycle was interrupted and the junction of capacitor C_{T} and resistor R_{p} did not increase in potential to the point where transistors Q7 and Q8 were triggered to conduction. After capacitor C_{T} had discharged through transistor Q6, the transistor turned off. The capacitor then started charging once again and the new cycle began immediately after the occurrence of the last heartbeat. The free-running operation would take place were there no input to the base of transistor Q6. Transistor Q6 would remain non-conducting and would not affect the charging of capacitor C_{T} . The capacitor C_{T} would trigger, and discharge through transistors Q7 and Q8 to control the generation of a pulse. The timing control circuit which determines the pulse duration (1 ms) and the repetition rate (72 pulses per minute) of the pulse generator, is made up of transistors Q7 and Q8, capacitor C_{T} and the resistances R_{p} , R_{T} , R_8 and R_9 . The pulse duration is determined by the time constant $\tau_p = C_{\text{T}} \cdot R_p$ and the rate mainly by $\tau_T = C_{\text{T}} \cdot R_T$. The capacitor charge current flows through the resistances. During the charging period both transistors are off. As C_{T} charges, the emitter voltage of Q7 rises and eventually exceeds the 4.2 V reference sufficiently to forward bias the transistor causing collector current to flow. This turns on Q8, raising its emitter voltage, which in turn raises the potential on the lower plate of the capacitor. This creates a regenerative turn on of both Q7 and Q8 which is sustained as long as C_{T} can supply current, a time determined primarily by resistor R_p . During this discharge time, the output transistor Q9 is turned on, causing current to flow in the electrode circuit. The output driver comprises transistor Q9, resistor R_{out} and capacitor C_{C4} . After 1 ms C_{T} is discharged, transistors Q7, Q8 and Q9 turn off and the pulse is terminated.

Finally, to avoid damage to the circuit due to high voltage signals from the electrodes, a zener diode (Z1) was placed between the terminals of the electrode.

A variation of this concept is the demand-triggered pacemaker, which stimulates every time it senses intrinsic heart activity, i.e., the stimulus falls directly on the natural QRS.

Dual-chamber pacemaker

A dual-chamber pacemaker typically requires two pacing leads: one placed in the right atrium, and the other placed in the right ventricle. A dual-chamber pacemaker monitors (senses) electrical activity in the atrium and/or the ventricle to see if pacing is needed. When pacing is needed, the pacing pulses of the atrium and/or ventricle are timed so that they mimic the heart's natural way of pumping. Dual-chamber pacemakers were introduced in the 1970s. One of the first description of a dual-chamber pacemaker was given by Berkovits in 1971. Berkovits announced a “bifocal” (AV sequential) pacemaker that sensed only in the ventricle but paced both chambers. In the presence of atrial standstill or a sinus node syndrome plus A-V block, the bifocal pacemaker could deliver a stimulus to the atrium and then, after an appropriate interval, to the ventricle. Berkovits improved on his original design given in Fig. 2.9 with a dual-chamber demand pacemaker. An schematic of this design is given in Fig. 2.10 [16]. In accordance with the principles of the demand pacemaker de-

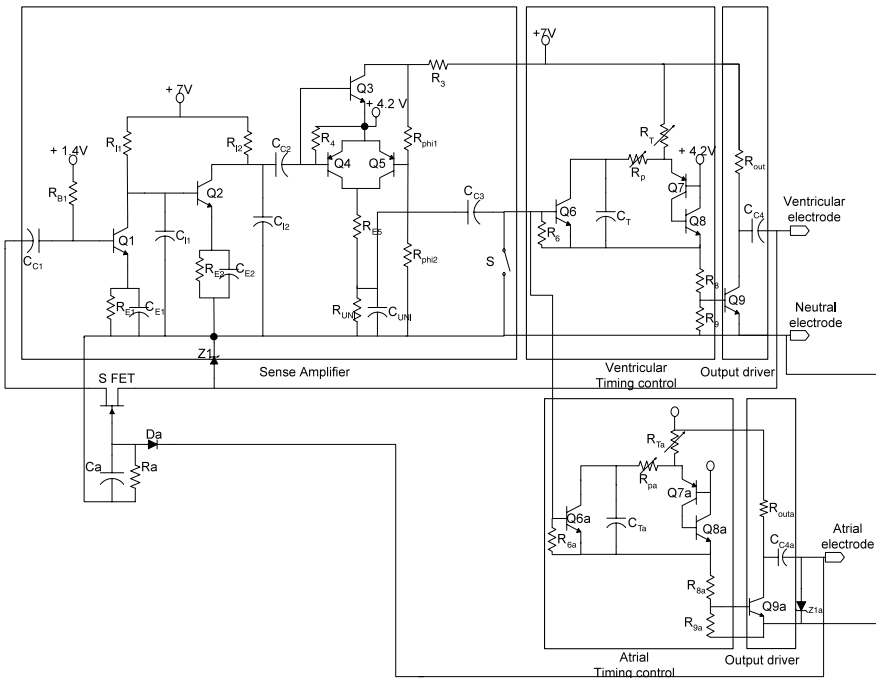


Fig. 2.10. Schematic of the dual-chamber demand pacemaker [16]

sign, a sense amplifier was provided to detect intrinsic ventricular activity. The timing control circuits determined both atrial and ventricular timeout stimulating period. However, the atrial-stimulating impulse was generated first, and, after a predetermined time interval (200 ms), the ventricular-stimulating impulse was generated. Three electrodes were provided, a neutral electrode, an electrode for atrial stimulation and an electrode for ventricular pacing and sensing. The FET switch (S FET) was inserted in the feedback path of the ventricular electrode in order to avoid erroneous detection because of the atrial contraction. The FET switch was normally conducting. The negative pulse generated at the atrial electrode was transmitted through the diode D_a , charging the capacitor C_a and turning off the switch. When the atrial-stimulating terminated, C_a discharged through resistor R_a and turned on the switch again. In this manner, the sense amplifier was disabled during each atrial stimulation and for a short interval thereafter.

More sophisticated dual-chamber pacemakers that sensed intrinsic activity and paced in both chambers were developed, with their first use in late 1977.

Rate-responsive pacemaker

The latest innovations include the development of “rate-responsive” pacemakers in the early eighties, which could regulate their pacing rate based upon the output of a sensor system incorporated in the pacemaker and/or lead. The sensor system consists of a device to measure some relevant parameter from the body (body motion, respiration rate, pH, blood pressure and so forth) and an algorithm in the pacemaker, which is able to adjust the pacemaker response in accordance with the measured quantity. Modern rate-responsive (also called frequency-response) pacemakers are capable of adapting to a wide range of sensor information relating to the physiological needs and/or the physical activity of the patient.

A block diagram of a rate-responsive pacemaker is given in Fig. 2.11. The system is based on a pacemaker having a demand pulse generator, which, in addition, is sensitive to the measured quantity. Many rate-responsive pacemakers currently implanted are used to alter the ventricular response in single-chamber ventricular systems. However, rate-responsive pacing can also be done with a dual-chamber pacing system.

2.4 New features in modern pacemakers

A modern pacemaker consists of a telemetry system, an analog sense amplifier, analog output circuitry, and a microprocessor acting as a controller, as one can see in Fig. 2.12 [4].

Nevertheless, the sense amplifier keeps on playing a fundamental role in providing information about the current state of the heart. State of the art implantable pulse generators or cardiac pacemakers include real-time sensing

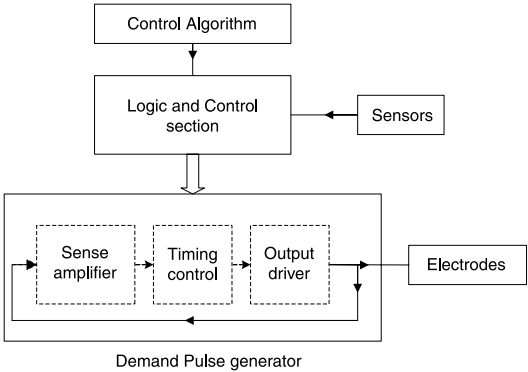


Fig. 2.11. Block diagram of a rate-responsive pacemaker

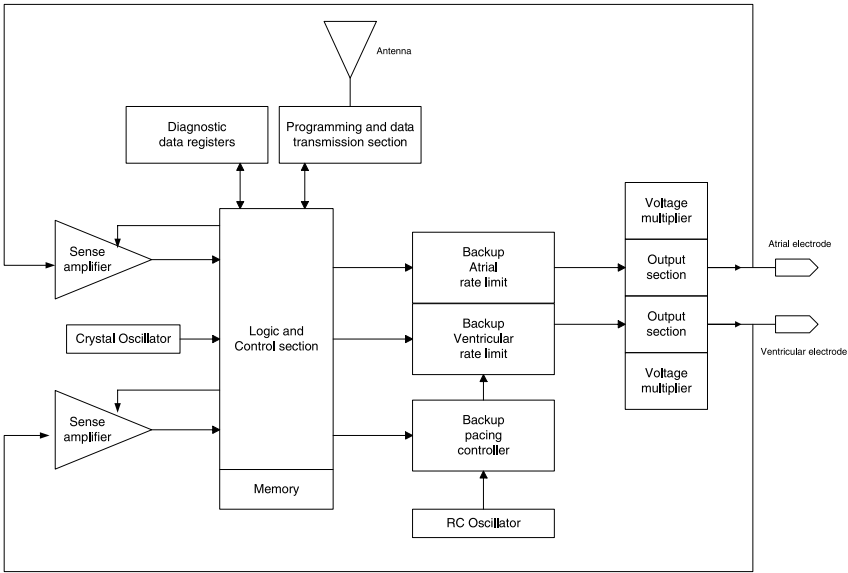


Fig. 2.12. Block diagram of a typical modern pulse generator [4]

capabilities that are designed to detect and monitor intracardiac signal events (e.g. R-waves in the ventricle). A sense amplifier and its subsequent detection circuitry, together called the front-end, derive only a single event (characterized by a binary pulse) and feed this to a micro-controller that decides upon the appropriate pacing therapy to be delivered by the stimulator. Over the years, huge effort is put into the improvement of sense amplifier and detection circuitry. The dynamic range of the atrial and ventricular electrograms sensed by an endocardial lead typically lies between 0.5–7 mV and 3–20 mV respectively. Slew-rates of the signals range between 0.1 and 4 V/s. For the

QRS complex, the spectral power concentrates in the band from 10 to 30 Hz. The T wave is a slower signal component with a reduced amount of power in a band not exceeding 10 Hz. Amplification of intrinsic cardiac signals requires circuitry that is robust against artifacts generated from non-cardiac electromagnetic sources located outside or inside the patient. Introduction of electronic article surveillance systems (EAS) has raised concerns with regard to the possible interaction between emitting field sources and the sense amplifiers of implantable medical devices like pacemakers [17], implantable cardioverter defibrillators and insertable loop recorders [18]. Other sources of electromagnetic interference (EMI) include cellular phones, airport metal detector gates, high voltage power lines [19], electro-cautery devices and MRI equipment [20]. Especially the more sensitive atrial-sensing channel of a brady-arrhythmia device is more prone to detection of EMI. Any type of EMI having sufficient amplitude could cause the pacemaker to react in a clinically undesirable way either inhibiting or triggering stimuli. Fortunately, noise reversion algorithms and circuits mostly provide reliable discrimination between EMI and intrinsic cardiac activity.

Morphological analysis

In pacemakers, one of the challenges is the reduction of unnecessary therapies delivered to the patient's heart when the heart rate dynamics becomes comparable to that of lethal tachyarrhythmias like ventricular tachycardia (VT) or ventricular fibrillation (VF). This situation includes supraventricular tachycardia (SVT) that may occur as a result of atrial fibrillation. As heart rate does not discriminate between lethal tachyarrhythmias like VT/VF and SVT or atrial tachyarrhythmias, the morphology of the QRS complex, or more specifically, the R-wave morphology can be used for a more accurate discrimination between SVT and VT.

In addition, to ensure efficient use of the memory available in an implantable device, the incidence of false positives, erroneously triggering automatic storage, should be minimized. For insertable loop recorders (ILRs), promoting factors include the low amplitude electrogram signal as a result of the limited vector available for pseudo ECG measurement and the presence of muscle EMG and mechanical disturbance of the electrode tissue interface. Therefore, signal analysis methods improving discrimination of signals from noise are of great importance.

Since the information retrieved by the above front-end circuit is reduced to a single event, morphological attributes of the electrogram are completely suppressed. Recent research and clinical studies report details on how morphological aspects of the electrogram relate to various pathological states of the heart and on how the wavelet transform can contribute efficiently to analysis.

Analyzing the structure of the electrogram over multiple scales allows discrimination of electrogram features pertaining over all scales from those only

seen at fine or coarse scales. Based on such observations, the presence or absence of electrogram features related to proximal or distal electrophysiological phenomena can be discriminated. The wavelet transform, being a multiscale analysis technique, offers the possibility of selective noise filtering and reliable parameter estimation. An algorithm based on wavelet analysis that compares morphologies of baseline and tachycardia electrograms based on differences between corresponding coefficients of their wavelet transforms has been found highly sensitive for VT detection [21]. Whereas smoothing attenuates spectral components in the stop band of the linear filter used, wavelet denoising attempts to remove noise and retain whatever signal is present in the electrogram.

Off-line ECG analysis, like Holter analysis, employs the discrete wavelet transform, implemented in the digital domain using multi-rate filter banks [22]. In these applications, the wavelet transform provides a means to reliably detect QRS-complexes. However, in patient worn external applications (e.g. intelligent Holter devices), it is not favourable to implement the WT by means of digital signal processing due to the high power consumption associated with A to D conversion and computation.

2.5 Summary and conclusions

A brief overview of the history and development of circuit designs applied in pacemakers has been presented. The advances in integrated circuit designs have resulted in increasingly sophisticated pacing circuitry, providing, for instance, diagnostic analysis, adaptive rate response and programmability. Also, based on future trends for pacemakers, some features and improvements for modern cardiac sensing systems have been described.

In the next chapters we will investigate a fully integrated implementation of the analog WT circuit to be used in pacemakers. But, before that, the advantages of the wavelet over the Fourier analysis will be the subject of the following chapter.

References

- [1] D. J. Woollons, *To beat or not to beat: the history and development of heart pacemakers*, Engineering Sci. Education J., vol. 4, pp. 259-268, 1995.
- [2] L. A. Geddes, *Historical highlights in cardiac pacing*, IEEE Engineering in Medicine and Biology, vol. 9, pp. 12-18, 1990.
- [3] W. Greatbatch and C. F. Holmes, *History of implantable devices*, IEEE Engineering in Medicine and Biology, vol. 10, pp. 38-49, 1991.

- [4] R. S. Sanders and M. T. Lee, *Implantable pacemakers*, Proceedings of the IEEE, vol. 84, no. 3, pp. 480-486, 1996.
- [5] H. G. Mond, *Recent advantages in pacemaker lead technology*, Cardiac Electrophysiology Rev., vol. 3, pp. 5-9, 1999.
- [6] A. C. Guyton and J. E. Hall, *Textbook of Medical Physiology*, 9th edn, W. B. Saunders, Toronto, January 1996.
- [7] J. D. Bronzino, *The Biomedical Engineering Handbook*, 2nd edn, vol. 1, CRC Press and IEEE Press, Boca Raton, FL, 2000.
- [8] R. Sutton and I. Bourgeois, *The Foundations of Cardiac Pacing. Part I*, Futura Publishing Company, Mount Kisco, NY, 1991.
- [9] M. Schaldach and S. Furman, *Advances in Pacemaker Technology*, Springer-Verlag, Berlin, 1975.
- [10] A. A. Hyman, *Resuscitation of the stopped heart by intracardial therapy*, Arch. Intern. Med., vol. 50, pp. 283-305, 1932.
- [11] www.ep-history.org.
- [12] R. Elmqvist and A. Senning, *An Implantable Pacemaker for the Heart*, Tliffe, London, pp. 253-254, 1959.
- [13] W. Greatbatch, *Medical cardiac pacemaker*, US Patent 3,057,356, October 1962.
- [14] www.tpub.com.
- [15] B. V. Berkovits, *Demand pacer*, US Patent 3,528,428, September 1970.
- [16] B. V. Berkovits, *Atrial and ventricular demand pacer*, US Patent 3,595,242, July 1971.
- [17] J. W. Harthorne, *Pacemakers and store security devices*, Cardiol. Rev., vol. 9, no. 1, pp. 10-17, 2001.
- [18] C. C. de Cock, H. J. Spruijt, L. M. van Campen, W. A. Plu, and C. A. Visser, *Electromagnetic interference of an implantable loop recorder by commonly encountered electronic devices*, Pacing Clin. Electrophysiol., vol. 23, no. 10, pp. 1516-1518, 2000.
- [19] T. W. Dawson, K. Caputa, M. A. Stuchly, and R. Kavet, *Pacemaker interference by 60-Hz contact currents*, IEEE Trans. Biomed. Eng., vol. 49, no. 8, pp. 878-886, 2002.

- [20] G. Lauck, A. von Smekal, S. Wolke, K. C. Seelos, W. Jung, M. Manz, and B. Luderitz, *Effects of nuclear magnetic resonance imaging on cardiac pacemakers*, Pacing Clin. Electrophysiol., vol. 18, no. 8, pp. 1549-1555, 1995.
- [21] C. D. Swerdlow, M. L. Brown, K. Lurie, J. Zhang, N. M. Wood, H. W. Olson, and J. M. Gillberg, *Discrimination of ventricular tachycardia from supraventricular tachycardia by a downloaded wavelet-transform morphology algorithm: a paradigm for development of implantable cardioverter defibrillator detection algorithms*, Pacing Clin. Electrophysiol., vol. 13, no. 5, pp. 432-441, 2002.
- [22] G. Strang and T. Nguyen, *Wavelets and Filter Banks*, Wellesley-Cambridge Press, Wellesley, MA, 1996.

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