Preface

As early as the late 1700s, Physicians speculated that electrical current could be used to stimulate the heart. In 1882, von Ziemssen used electrical current to directly stimulate the heart of a woman whose anterior chest wall had been removed after resection of a chest tumor. In 1952, Zoll used transthoracic current to pace the heart, and in 1958 the first implantable pacemaker was placed by Ake Senning and Rune Elmquist. At the same time, Furman and Robinson demonstrated the feasibility of transvenous cardiac pacing. In the late 1960s, Mirowski and colleagues pioneered the concept of an implantable device that could be used to defibrillate the heart. Over the last 50 years, implantable cardiac devices have become the primary treatment for bradyarrhythmias and ventricular tachyarrhythmias and have emerged as an important adjunctive therapy for patients with heart failure. It is currently estimated that almost 400,000 pacemakers and defibrillators are implanted annually in the United States.

With exponential expansion of the use of implanted cardiac devices, it has become critical for all physicians to become knowledgeable about them, as they continue to increase in complexity. The second edition of Cardiac Pacing for the Clinician has the same goal as the first edition: To provide a succinct yet comprehensive reference for the implantation and follow-up of implantable cardiac rhythm devices. The book is intended to be a practical guide for the day-to-day management of these increasingly complex devices, and is intended for all physicians caring for patients with devices. We also hope that the emphasis on clinical care will be useful for implanting surgeons, nonphysician medical associates, and clinical members of industry.

The book is divided into four sections. The first section describes pacing leads and pacemaker function. The second section focuses on device implantation. New to this edition is a chapter on implantation of left ventricular leads, used in the biventricular pacing systems intended to treat patients with heart failure. Purposely we have asked two experienced implanters to discuss their personal methods for placing leads in the cardiac venous systems to illustrate the diversity of techniques and “tricks of the trade.” The third section reviews the use of implantable cardiac devices in particular clinical situations. All of the chapters from the first edition have been extensively revised; new to this edition are chapters on device use for patients with atrial fibrillation, heart failure, and syncope, providing further evidence for the expanding indications

for implantable devices. The final section is devoted to device follow-up. It is our belief that the greatest impact for device therapy on patient outcomes is in follow-up. Important topics such as avoiding inappropriate therapies in patients with defibrillators, ensuring optimal and individualized device function for patients, and techniques for minimizing the risks of environmental electromagnetic interference are extensively reviewed.

The last 6 years have shown rapid evolution of cardiac implanted devices that can provide not only therapy but also information on the clinical status of a patient. While beneficial, this increasing complexity also means that all clinicians must be knowledgeable about device function, indications for device use, and device follow-up. We believe that the second edition of Cardiac Pacing for the Clinician will provide the essential clinical information necessary for treating patients with implanted cardiac devices.

2

The Pulse Generator

Richard S. Sanders

Basic Pacing Concepts and Terminology

A cardiac pulse generator is a device having a power source and electronic circuitry that produce output stimuli. Functionally, at its simplest, current sourced by the device's battery travels through a connecting pathway to stimulate the heart and then flows back into the pacemaker to complete the circuit.

Although numerous and varied designs of cardiac pacemakers are available, all have the same basic components:

- A power source in the form of a battery
- Circuitry (output, sensing, telemetry, microprocessor or microsequencer, memory)
- A metal casing (can) welded shut to keep out fluids
- A feedthrough (a piece of wire surrounded by glass or sapphire) that maintains a hermetic seal to provide an electrical connection through the can
- A means of connecting a pacing lead (wire to the heart) to the header of the pacemaker
- Sensors (e.g., acceleration, vibration, impedance)

Modern pacemakers are extremely sophisticated and highly programmable, capable of storing a rather impressive array of diagonstic data. Weighing about 25 g, they can pace and sense in one, two, or three chambers and adjust their rate by tracking intrinsic atrial activity or by responding to input from a sensor.

Because the pacemaker is an electronic device, the clinician may be unfamiliar with the engineering nomenclature associated with this technology. The more common terms are listed in Table 2.1. How the pacemaker works and factors to take into consideration when programming a pacemaker are discussed in the following sections.

Table 2.1 Engineering nomenclature.

Ampere	A unit of electric current equaling one coulomb per second
Amplifier	Device or circuit that amplifies, enlarges, or extends an electrical signal
Anode	Positive pole electrode
Capacitor	Device that can store an electrical charge
Cathode	Negative pole electrode
Circuit	A closed path followed by an electric current
Coulomb	A measure of charge equal to the amount of electricity transported by 1 A of current for 1 s
Current (I)	Flow of electrons through a conductor
Current drain	Current drawn from a battery
Impedance	The total opposition to the flow of current (including that through the conductor and across the interface to the stimulation site). Often used interchangeably with the term "resistance"
Insulation	Material that offers high resistance to the flow of electric current
Joule	Unit of measurement for energy or work (one joule = one coulomb flowing across a potential of 1 V for 1 s)
Ohm (Ω)	Basic unit of electrical resistance and/or impedance (one ohm = resistance produced when one ampere of current produces a voltage of 1 V across a conductor)
Ohm's Law	Voltage = current times resistance ($V = IR$)
Resistance	The electrical property of a material that resists the flow of electric current
Resistor (R)	Electronic circuit component that produces a known resistance
Volt (V)	Basic unit of measurement of electrical potential difference
Watt	Unit measure for power

Power Source (Battery)

Pacemakers directly benefited from advances in battery technology; thus, a variety of power sources have been employed in cardiac pacemakers over the last four decades. The first chemical cell to achieve wide-scale use was composed of mercury–zinc (HgZn) also known as the Rueben or Mallory cell (1,2). Unfortunately, mercury–zinc had some undesirable characteristics. The cell voltage was 1.35 V, so most pacemakers incorporated four to five cells in series to provide the 5.0–6.0 V deemed necessary to produce consistent capture or depolarization of the heart. In addition, as the cell is depleted, the voltage decreases to almost zero precipitously with little or no warning. As a consequence, pacemaker patients had to be followed frequently to assure that the battery voltage did not decrease to a point where capture was lost. Several manufacturers flirted with nuclear-powered cells but this much-touted battery failed to achieve much popularity because of the government restrictions and bulky radiation shielding, which made them rather large (3). One manufacturer produced a rechargeable cell (nickel–cadmium); however, these devices required constant recharging and the memory effect of these rechargeable batteries had an adverse effect on the overall longevity.

Prior to 1975, pacemaker longevity in excess of 3–4 years was the exception (4,5). This changed in the mid-1970s as the industry migrated almost exclusively to a lithium iodide-based cell. This cell had many of the characteristics that were considered ideal for a pacemaker power source. It was highly reliable and had a very long shelf-life. Its energy density was better than previous cells, enabling manufacturers to provide many ampere-hours in a small volume. This allowed for significant size–volume reductions in the pacemaker. In contrast to the mercury–zinc cell, which exhibited a sudden decrease in voltage just before depletion, the lithium iodide cell had a more predictable and gradual decrease of voltage as it approached depletion. The chemical reaction between lithium and iodine produces lithium iodide, a resistive barrier.

$$2L_i \rightarrow 2L_i^+ + 2e^- \quad \text{(anode reaction)}$$

$$2L_i^+ + 2e^- + I_2 \rightarrow 2L_iI \quad \text{(cathode reaction)}$$

$$2L_i + I_2 \rightarrow 2L_iI$$

At the interface where the anode (positively charged electrode) and the cathode (negatively charged electrode) interact, a lithium iodide barrier grows, building up internal resistance in the cell. This has two distinct advantages: first, it allows the manufacturer to select a predefined depletion level known as the *elective replacement point*, after which the device can be expected to function another 3–6 months; second, it creates a system that effectively prevents any internal shorts because the lithium iodide layer forms wherever the anode and cathode come in contact (6). Yet another advantage of lithium iodide was the absence of outgasing (any gaseous result of the chemical reaction). Mercury–zinc batteries produced hydrogen gas as a byproduct and therefore had to be encapsulated in gas-permeable epoxy or a chemical "getter" was necessary for sealed devices. In the case of the epoxy-encapsulated device, the epoxy allowed the hydrogen to diffuse into body tissues. Unfortunately this allowed water to enter and contact the pacemaker circuit. With some models, the water dissolved ionic contaminants left on the circuit, which produced shorts or dendrites (essentially mineral deposits) and resulted in premature battery depletion and several pacemaker recalls (7). Lithium iodide batteries produced no gases and allowed engineers to hermetically seal the battery and circuit inside a metal can, eliminating a significant mode of failure while better protecting the electronic circuit from spurious electrical interference.

Lithium iodide chemistry, which has a favorable energy density, has a low power density and can only supply currents in the very low milliampere range. This limited power density makes lithium iodine batteries less suitable for some of today's high current requirements, such as long-range telemetry. Additional chemistries are being introduced into today's pacemakers, including carbon monofluoroide (CFx) and manganese dioxide (MnO2) chemistry. These new cell formulations achieve many of the benefits of the lithium iodide chemistry and provide significantly better power delivery. These new chemistries may well totally supplant lithium iodide as the power source of choice for pacemakers.

Gradual Decay of Battery Voltage (Elective Replacement Indicator)

All cells experience self-discharge; the decrease in capacity occurs as a result of the reaction of cathodal and anodal materials, even when the cell

is not connected to a circuit. Lithium iodide cells have a relatively low self-discharge. Once a cell is connected to the circuit, power is supplied to keep the pacemaker awake in what is called a quiescent state, still executing logic commands and looking for sensed inputs. Thus, even when a pulse generator is sitting in its packaging, not connected to a lead, its battery capacity is being slowly depleted as a result of the quiescent current requirements of the circuitry and the self-discharge of the cell. This factor, along with considerations regarding the ability to maintain sterility of the device in its packaging, has led manufacturers to provide device-specific "use-before" or "shelf-life" dates. By specifying the maximum time between device cell connection and implantation, manufacturers assure that clinicians can expect the device to meet its published postimplant longevity specifications despite the effects of preimplant depletion of the power source.

At beginning-of-life (BOL) a lithium iodide cell will have a voltage output of around 2.8 V and an internal impedance < 1,000 Ω (Fig. 2.1). As it ages, cell impedance rises because of the progressive buildup of lithium iodide. This generally continues until the elective replacement point is reached, and the cell impedance rises above 8,000 Ω. The cell should not be allowed to deplete to lower levels because the pacemaker's circuitry requires a minimum voltage level to operate. Manufacturers designate a specific point that triggers the elective replacement indicator (ERI) significantly in advance of the point where circuit operation is threatened. It is at this point that replacement is advisable (or at least follow-up monitoring of the patient should be intensified).

One should always consult the physicians' manual to determine the specific behavior that indicates the need for elective replacement. A mode conversion from dual-chamber to single-chamber (e.g., DDD to VVI), rate response to nonrate response or a change in magnet rate behavior may be observed when the ERI is triggered. Some pulse generators also provide an additional pre-ERI alert known as an intensified follow-up indicator (IFI), a safety feature that

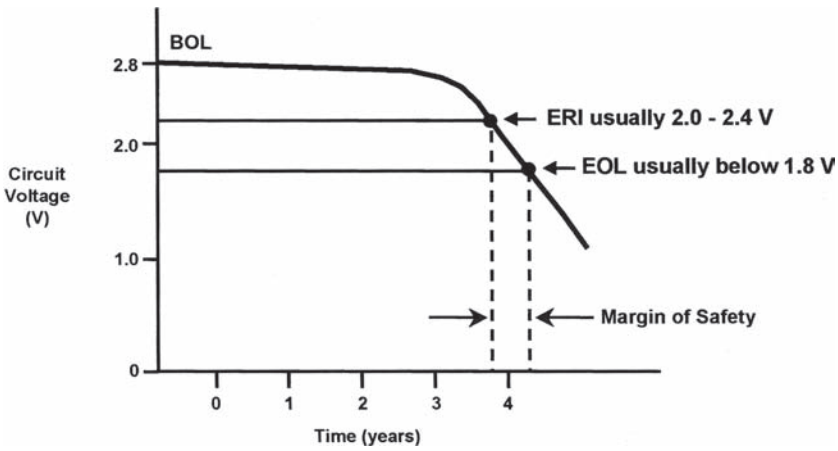

Fig. 2.1 At beginning-of-life (BOL), the lithium iodide battery exhibits an output voltage of 2.8 V, which slowly decays until it nearly depletes, whereupon the voltage decays more rapidly. At around 2.0–2.4 V, the elective replacement indicator is triggered, which usually leaves 6 months before the pulse generator will begin to behave abnormally as it reaches its end of useful life (EOL).

signifies that the frequency of follow-up visits should be increased because the unit is approaching ERI. This is particularly useful if the pacemaker is programmed to settings that require higher battery current drains (e.g., high voltage outputs or pacing rates) (8).

Longevity

At the simplest level, pacemaker longevity is determined by the battery capacity and the current drain. In reviewing longevity specifications for a device as published in physicians' manuals and promotional materials, one should pay particular attention to the associated conditions and assumptions. Pacemaker manuals usually describe longevity using an assumption of nominal outputs and pacing rate (Table 2.2). Nominal values adequately pace the heart in the majority of patients, and are preset in the device before it is shipped by the manufacturer. The longevity of the device is affected by the output settings, as is the margin between the triggering of the ERI and the onset of compromised behavior (9).

Physicians' manuals sometimes provide two different battery ratings: the stoichiometric capacity, or total amount of energy stored in the battery and the usable battery capacity. The latter is significantly lower than the stoichiometric capacity and is the value most relevant to the clinician, because the battery effectively becomes useless once the voltage drops below the circuit-operating threshold. Battery capacity is expressed in ampere-hours or A h. A common rating for a pacemaker's battery capacity is in the range of 1.0–1.5 A h (10).

The total current drain on the battery is primarily composed of: (a) quiescent current (i.e., that required to keep the circuitry running (sensing circuitry, amplifiers, and central processing unit), and (b) pacing current that is required to produce the output pulse used to stimulate the heart. Total current drain of a dual-chamber pacemaker set to nominal outputs (quiescent current + pacing current) is typically 10–15 μA per pacing cycle. Fifty to sixty percent of the total current drain is quiescent, and is relatively constant. This type of current drain is wholly determined by the circuit design – there is little that a clinician can do to change it. In addition, as the pacing output is reduced the proportion attributed to the quiescent current increases. Thus, a pacemaker that is

Table 2.2 Effect of current drain on longevity for Intermedics Marathon DDDR Pulse Generator.

		Years of service life			
	Rate	**100% Pacing**	**100% Pacing**	**50% Pacing**	**50% Pacing**
Nominal output		500 Ω load	750 Ω load	500 Ω load	750 Ω load
DDDR mode, 3.5 V pulse amplitude	60	6.3	7.4	7.1	8.0
0.45 ms pulse width	70	5.8	6.9	6.8	7.7
Low output					
DDDR mode, 2.5 V pulse amplitude	60	7.4	8.4	8.0	8.7
0.45 ms pulse width	70	6.8	7.8	7.7	8.4

From Baker RGJ, Falkenberg EN. Bipolar versus unipolar issues in DDD pacing (Part II). *PACE* 1987;10:125–132, with permission.

Table 2.3 Effect on longevity of output voltage, output pulse width, pacing rate, and lead impedance.

Parameter	Effect on longevity	Factor
Increasing output voltage	Decrease in longevity	Exponential
Increasing output pulse width	Decrease in longevity	Linear
Increasing pacing rate	Decrease in longevity	Linear
Increasing lead impedance	Increase in longevity	Linear

not pacing or has its output near zero still drains some amount of energy from its battery. For this reason, reducing output does not always have a profound longevity benefit.

The pacing current drain during each paced cycle is a function of output voltage, pulse width, and lead impedance. The interrelationship between these factors can be expressed by the following equations:

$$E_P = V_P \cdot I_P \cdot T$$

Using Ohm's law we can substitute

$$I_P = V_P / R_L$$

Thus

$$E_P = V_P^2 T / R_L$$

where E_P, energy of each pacing pulse (μJ); V_P, output pulse voltage (V); I_P, output pulse current (mA); T, output pulse width (ms); and R_L, lead impedance (Ω).

The mathematical relationship indicates that the energy extracted from the battery per pulse is linearly related to the pulse width and lead impedance but varies by the square of the output voltage. It follows that, when programming output, the voltage will have a more pronounced effect on energy consumption than a proportional change in pulse width (Table 2.3).

Battery Impedance and Output Programming

Despite the fact that the battery generates 2.8 V, it is possible to program the device to higher outputs through the use of specialized circuits that step up the voltage to higher levels. The selection of output voltages varies by manufacturer but generally covers the range of 0.5–8.0 V.

If a pacemaker is close to its elective replacement point, programming from a low- to a high-output setting immediately increases current drain on the battery, causing the terminal voltage of the battery to fall. This is an effect of Ohm's law and is the result of high battery impedance in a partially depleted cell. This has been shown to cause premature activation of the ERI (11).

In rate-responsive devices, current drain will increase as the exercise rate increases, which will decrease the battery voltage. In rate-responsive devices with a mature battery and a high battery impedance, exercise could trigger the ERI. In the same device, a reasonably normal battery voltage may be observed during follow-up with the patient at rest. Generally, standard reprogramming disengages an ERI that has occurred in response to a transient increase in current drain.

Effects of Cold and Heat on the Battery

Lithium iodide batteries are also affected by extremes of cold or heat. Although these effects are not seen when a device is implanted, be aware that:

1. A pacemaker can be exposed to cold before implant such as during shipping. Low temperatures slow down the reaction between lithium and iodine, which causes the internal resistance to rise and voltage to drop. This condition is temporary and exists only during the exposure to low temperatures. The battery recovers immediately at room temperature, but the ERI could already be triggered. Occasionally, if a device is not interrogated after implant, the clinician might see the ERI at the first follow-up and falsely assume there is something wrong. If this occurs, it is possible that the unit has been exposed to cold prior to implant, activating the ERI, which at this point could usually be reset during normal programming (12).
2. Exposure to extreme heat can cause a pacemaker to explode. If a deceased patient is to be cremated, it is recommended that the unit be removed. Most funeral directors are now aware of this, particularly if they have experienced this unfortunate phenomenon.

Circuitry

Cardiac pacemakers incorporate some of the most advanced, high-reliability electronic circuitry available. The basic building block is the integrated circuit (IC), which starts as a silicon wafer and has a number of miniaturized circuit elements etched into its surface during the manufacturing process. Modern pulse generators incorporate custom-designed, very large-scale integrated (VLSI) circuits (13). ICs are built up layer by layer and can incorporate millions of electronic elements. The elements are so fine that they can barely be seen with an optical microscope.

Some components are not integrated into the IC, such as larger-value capacitors, diodes, inductors, transmission coil, and so on. These components that cannot be incorporated on the IC must be added as discrete components. It is the goal of the pacemaker designer to minimize the number of discrete components to save cost and space. The ICs and discrete components are mounted onto a layered substrate, which, like a matrix of major highway interchanges, provides pathways on its surface, and within and between its layers, by which the components are interconnected. Historically, this substrate was composed of a ceramic material. Today, these substrates are almost exclusively constructed of a flexible polymer. This combination of discrete components and ICs, mounted on a substrate, are known as hybrid circuits (Fig. 2.2). The hybrid circuit and the battery together comprise 80–90% of the space in the pacemaker's can (13).

It is often useful to think of a pacemaker's circuit not in terms of specific components, but rather according to their various functions. These functional sections include: logic and control, memory, timing, sensing, output, data transmission, and programming. Figure 2.3 illustrates the functional block diagram of a typical pulse generator, the Sulzer Intermedics Marathon. The circuitry associated with each of these functions is described in the following sections (14).

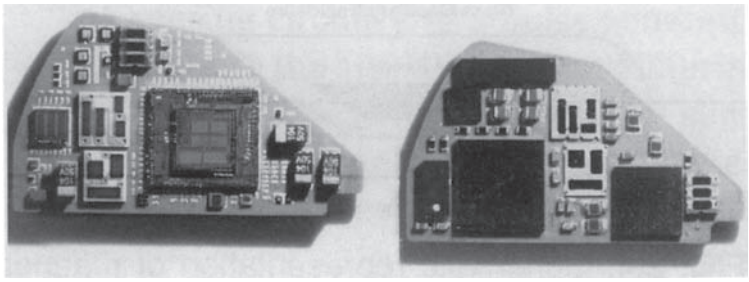

Fig. 2.2 Hybrid circuit (front and back), consisting of one or more integrated circuits and a number of discrete components mounted on a ceramic substrate.

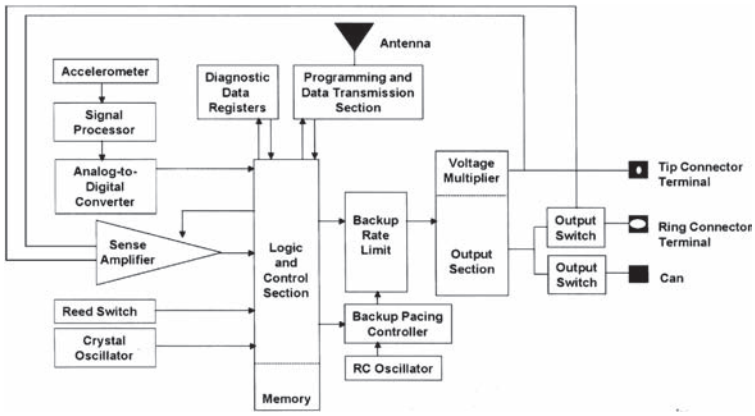

Fig. 2.3 Block diagram showing functional organization of Intermedics Marathon SR pulse generator.

The Microprocessor

In most modern pacemakers the logic functions are controlled by microprocessors. A pacemaker microprocessor is very similar to the central processing unit (CPU) in a desktop computer. A pacemaker microprocessor operate with currents millions of times less than low-power microprocessors in today's laptop computers. This low-power operation is necessary in order to achieve overall size and longevity and can limit its processing power. The pacemaker CPU is generally customized and integrated with other components such as memory on one IC.

The microprocessor is constantly accessing its memory for instructions on what to do next. Processor speed is somewhat dependent on a crystal oscillator or clock. The higher the speed, the more instructions that can be executed per second, but more current is required. Fortunately, because of the nature of pacing, the processor does not need to be on all the time. It can be in a sleep mode, a good deal of the pacing cycle, and only needs to be awakened when it receives an input or needs to execute a particular function. The percentage of time the processor is awake and performing tasks is called the duty cycle. The higher the duty cycle, the higher the current drain and the shorter the device longevity. Thus, even in the absence of pacing, frequent sense events can increase current drain. Actually, the microprocessor receives inputs from

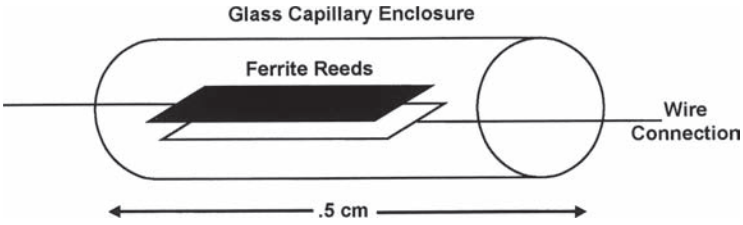

Fig. 2.4 Schematic representation of magnet-activated pacemaker reed switch.

several circuits other than sensing and crystal oscillator circuitry. It checks to see if the reed switch (a magnet-activated switch composed of two very small ferrite reeds placed close together inside a tiny glass tube) is open or closed (Fig. 2.4). It is used as a way for an ordinary magnet to activate or deactivate certain features or functions, such as temporarily suspending sensing.

The processor also receives inputs from the sensor(s) for rate-response and telemetry/programming commands. All these inputs are used by the microprocessor to determine whether when and to which chamber to deliver an output pulse. Some inputs, such as from the crystal oscillator, occur as fast as millions of times per second. Other inputs, such as from the reed switch, might not occur for months or years. In the case of a totally pacemaker-dependent patient, there may be no sensed inputs for the life of the patient.

Sensing Circuitry

Pacemaker leads connect the sensing and output circuitry to the outside world via feedthroughs. The feedthroughs function like a single-lane highway that handles two-way traffic, like a serial bus in computer lingo. The pacemaker spends much more of the pacing cycle looking for sense events than it does actually providing pacing output, which generally has a duration of half a millisecond or less. How a pacemaker senses depends on the particular device, but there are some similarities.

The sensing circuitry of a pulse generator is used for both the amplification and filtering of intracardiac signals. To prevent the sensing of noncardiac signals by the pacemaker, the intracardiac signal is processed by the circuitry to determine whether it has sufficient amplitude and the appropriate frequency content. This relationship between amplitude and frequency may be characterized by plotting a frequency–response curve. This curve is derived in the laboratory by inputting sine-squared ($\sin^2$) or some other test waveforms of varying amplitudes and frequencies into the circuit. Although intracardiac signals are far more complex than the $\sin^2$ signals, the latter are easy to generate and can be reproduced with precision (15).

The frequency–response curve shown in Fig. 2.5 illustrates several key points. First, everything above the curve is sensed by the pacemaker, while everything below is not. Second, the sensing circuit is designed and tuned to be most sensitive to signals within a specific frequency range. This point may differ somewhat between various manufacturers' sensing circuits, but is usually in the range of 20–30 Hz. Signals with frequencies at or near the center point are passed through the filters with little or no loss of amplitude, whereas signals far from the center frequency will be significantly attenuated. Once

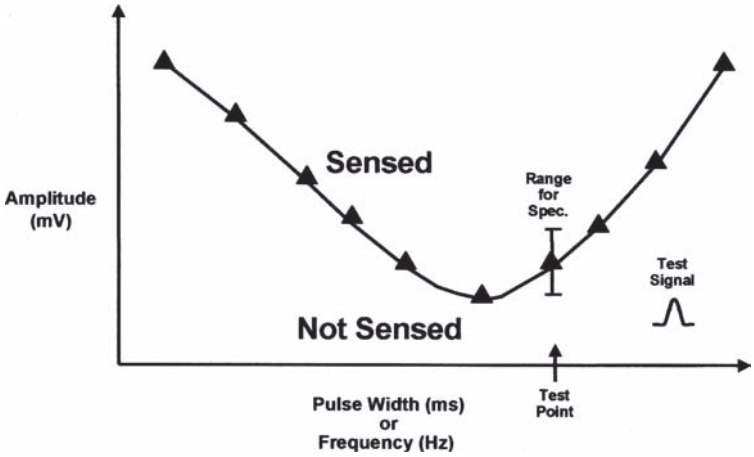

Fig. 2.5 Frequency–response curve of pacemaker sense amplifier exposed to $\sin^2$ input signals. Signals above the curve are sensed, whereas those below are not. Another characteristic of pacemaker sensing circuits is the preference given to frequencies within the range of 20–60 Hz, a significant frequency component of cardiac IEGMs.

the signal passes through the filters, if it has sufficient amplitude, it will be detected by the pacemaker.

These characteristics of the sensing circuit have many clinical implications. The programmable sensitivity settings of a pacemaker, usually expressed in millivolts, correspond to the device's response to a $\sin^2$ signal at or near the 0 dB frequency. Often, manufacturers will use different test signals for their atrial and ventricular sense amplifiers to better approximate the frequency content of the signals originating in each chamber called the intracardiac electrogram (IEGM). When a complex signal like an IEGM, made up of various frequencies, is encountered, those portions of the signal at or near the zero dB frequency are preferentially passed through, whereas other portions are attenuated (run-on). Thus, a large intracardiac signal with frequency components outside the central range will appear significantly smaller to the pacemaker. Moreover, owing to differences in this so-called bandpass filtering from one pacemaker model to another, a particular sensitivity setting (e.g., 2 mV) may not be equivalent in different devices. One further note: the body itself acts like a filter, and the morphology of the IEGM may differ significantly from the surface EKG. Unlike the surface EKG, the IEGM reflects more of a local electrical event near the electrode within the heart (Fig. 2.6).

It is impractical to analyze the frequency content of an intracardiac signal in the implant setting; therefore, clinicians often calculate a surrogate value, known as the slew rate, which is related to frequency (Fig. 2.7). The slew rate is the slope of a straight segment (the intrinsic deflection) of the electrogram; this corresponds to the change in voltage with respect to time ($\Delta V/\Delta T$), and is expressed in mV/ms. A signal with a slew rate of approximately 0.5 mV/ms is considered the minimum requirement to ensure an appropriate signal frequency for sensing.

Some portions of the signal may not have adequate amplitude and slew rate to be sensed because of the complex nature of the IEGM. Typically, the portion

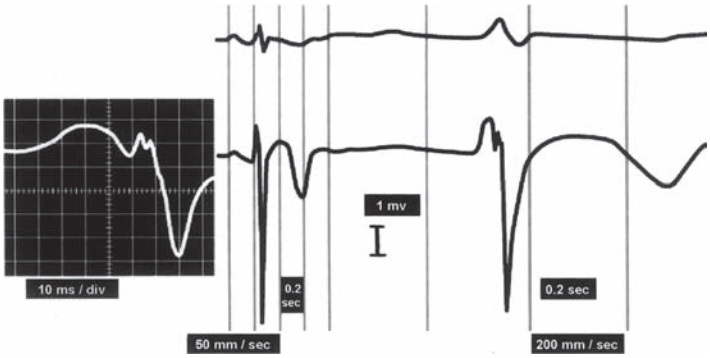

Fig. 2.6 Surface EKG (*Top*) and ventricular electrogram (*Bottom*) demonstrate the difference in amplitude and frequency content. Sweep speed has been increased from 50 to 200 mm/s.

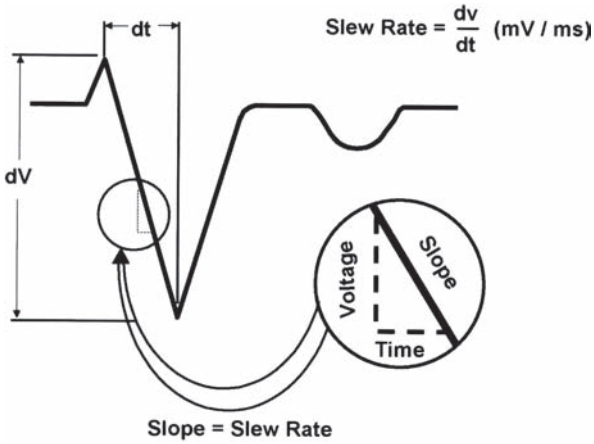

Fig. 2.7 Slew rate, or the change in voltage divided by the change in time of an intracardiac signal.

of the signal with the highest slew rate and amplitude is in the middle of the electrogram, representing the moment when the wave of depolarization passes directly under the sensing electrode. Thus, the earliest parts of the signal, which may be visible on the surface ECG, may not be sensed by the device. In this case, the pacemaker output is delivered into already depolarized tissue, rendering the pulse ineffective. This should not be considered a sensing malfunction, but is simply a delay in the timing of sensing the event by the generator. It is most commonly observed in dual-chamber devices in patients with right bundle branch block. Arrival of localized depolarization in the area of the right ventricular lead is significantly delayed, allowing the output pulse to be delivered during left ventricular depolarization and well into the surface QRS (Fig. 2.8). Clinically, this may manifest itself in the apparent undersensing of a P-wave or QRS, a surface ECG phenomenon known as pseudofusion. Either shortening or extending the A-V delay might alleviate the pseudofusion.

Clinicians should follow certain precautions to ensure appropriate sensing of a patient's intrinsic electrograms. The amplitude and slew rate of the

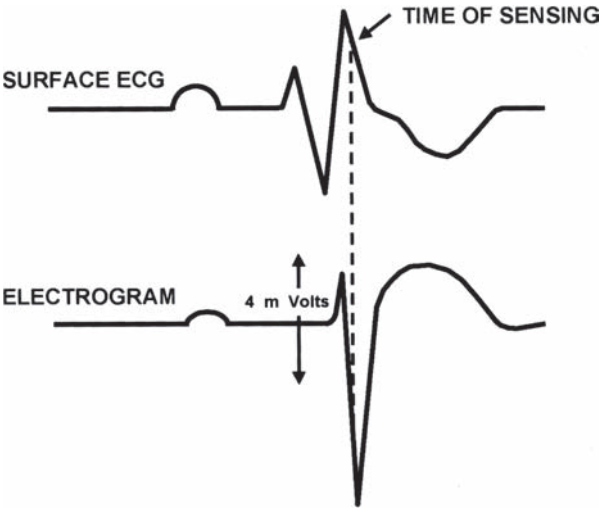

Fig. 2.8 Illustration of mechanism that may cause apparent undersensing. Pulse generator with a lead in the RV apex senses electrogram during the large second deflection. Thus, an output at any time prior to the corresponding point on the surface EKG is normal but may result in fusion or pseudofusion phenomenon.

electrogram should be evaluated at the time of implantation, and, if either or both appear inadequate or marginal, the lead should be repositioned. A specialized piece of equipment (pacing systems analyzer or PSA) which should be used during the implant procedure measures both of these characteristics and provides the required quantitative measure of the electrogram. Assuring a robust sensing signal at the time of implant provides the most programming flexibility to deal with problems that may arise later. Because the characteristics of the signal may change over time, owing to fibrotic encapsulation and/or inflammatory reactions following implantation, or owing to the activity of certain cardioactive drugs, a sensing threshold test should be performed prior to hospital discharge, during the acute-to-chronic phase (typically, the first 6–8 weeks postimplant), and periodically thereafter (16,17). During this test, the pacemaker's sensitivity is progressively programmed to less sensitive settings (i.e., higher mV values) to determine the setting at which reliable sensing is lost (18). The device is then programmed to a value that provides an adequate margin to ensure continued sensing, while minimizing the possibility of inappropriate sensing of extracardiac signals.

Many modern pacemakers incorporate a means to automatically adjust the sensing circuit to be more or less sensitive (autogain function). The auto gain function constantly monitors the intracardiac signal and maintains a preset safety margin to should any changes in the amplitude or slew rate occur over time. This minimizes the need to evaluate and adjust the sensing setting during follow-ups.

Sensing Circuitry and Electromagnetic Interference

Unwanted signals from external and endogenous sources can be sensed by the pacemaker. Endogenous sources include myopotentials generated by skeletal muscles and far field signals originating in the chamber opposite to which

sensing was intended (e.g., R-waves would be considered far field when sensed in the atrium) (19,20). External sources are numerous and include cellular phones, electronic article surveillance systems, diathermy equipment, electrocautery, some large electronic motors and, in fact, any equipment generating a large electromagnetic field. Their effects are usually temporary and may include inhibition, asynchronous pacing, and high-rate ventricular pacing in dual-chamber modes. In rare cases (e.g., electrocautery, external defibrillation) EMI can trigger the backup mode or ERI. Proximity is an important factor, because energy generally falls off by a factor related to the square of the distance. Lower sensitivities and bipolar sensing make devices less susceptible to electromagnetic interference (EMI); medical professionals are encouraged to follow warnings in physicians' manuals to avoid serious interaction with these sources of interference (21). In most cases, simple precautions are extremely effective, like not putting a cell phone in a breast pocket over the pacemaker (22).

Pacemaker circuitry also contains feed through filters and noise discriminating capabilities that provide an additional level of protection against sources of EMI. Feed-through filters help protect the circuit from damage owing to high-voltage EMI by shunting the energy away from the circuit. Lower energy EMI signals that pass through the protective circuitry are evaluated by a noise discrimination function. If the number of sense events in a given cardiac cycle is greater than what would normally be expected from anything cardiac in origin, the pacemaker may revert to a noise mode. Noise reversion tends to occur if the sensed rate is greater than 7 events/s. A pacemaker's noise response is usually fixed-rate pacing at a rate set by the manufacturer and lasts as long as the interference is present, although the specific behavior should be verified by referencing the physicians' manual included with the pacemaker.

Output Circuitry and Pacing Thresholds

Output circuitry is usually composed of capacitors and electrical switches controlled by the microprocessor or logic circuitry. Output circuitry can deliver voltage in excess of the battery voltage, generally through the use of a charge pump. A charge pump provides the flexibility to program many discrete voltages and also allows for voltage regulation. The charge pump uses a number of small capacitors to dump charge into a larger capacitor. The larger capacitor is then discharged through the lead and heart tissue for a controlled period corresponding to the pulse width.

As mentioned previously, the voltage the battery can supply decreases as it depletes. Older pulse generators used voltage doublers to provide programmable output voltage that would decrease output as the battery depleted. For example, when battery voltage was at BOL, output settings provided by the manufacturer were roughly multiples of the BOL battery voltage (e.g., 2.8, 5.4, or 8.1 V). At or near ERI the device might actually be delivering the same multiples of the now diminished capacity (e.g., 2.2, 4.4, or 6.6 V). Even if a patient's pacing threshold remained constant, the output of the device might thus require readjustment because of declining battery voltage (23,24).

In the preceding example, the voltage is considered unregulated (i.e., there is no guarantee that the delivered voltage will match the programmed value). Use of unregulated outputs can affect the value reported for the

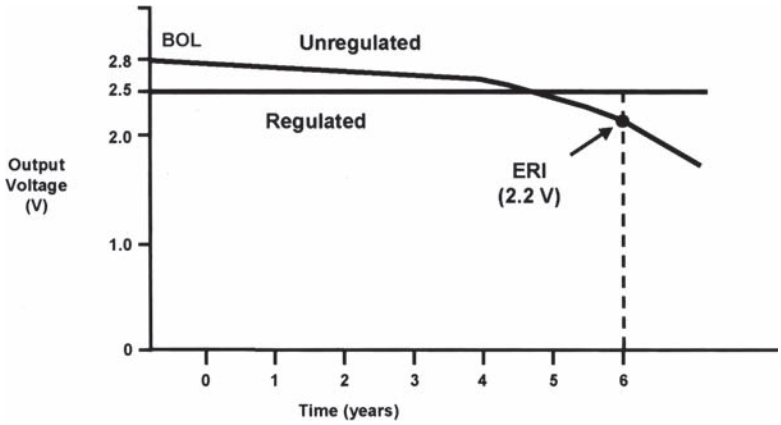

Fig. 2.9 Unregulated voltages decline as battery voltage declines, whereas regulated voltages are maintained at the programmed value.

pacing threshold, because the actual output does not necessarily match the programmed value (or, put another way, the programmed value may overstate the actual voltage being delivered). Toward ERI, it may appear as though the pacing threshold is increasing, requiring even more energy to stimulate the heart, whereas this may merely be the result of diminishing battery voltage. Therefore, it should not be mistaken as an impending lead or physiologic problem, especially when the lead impedances are relatively constant (25). Modern manufactured pacemakers with charge pumps have regulated output voltage so that even near the elective replacement point the device will deliver the programmed output (Fig. 2.9).

The output capacitor delivers its voltage, which decays exponentially throughout the pulse. The beginning voltage, called leading edge voltage (V_1), is nearest the programmed value, and the voltage at the end of the pulse, called the trailing edge (V_2), is dependent on the impedance of the lead system or load and the capacitance of the output capacitor. The amount of current delivered to the heart is related to the programmed voltage and the lead impedance, and is governed by Ohm's law ($V = IR$). An illustration of a typical pacemaker output is shown in Fig. 2.10.

Both the amplitude and width of the output pulse should be defined when expressing pacing threshold. As discussed in greater detail in Chapter 1, successful stimulation (or capture) of cardiac tissue follows a strength–duration relationship. At narrower pulse widths, it requires higher voltages to stimulate tissue, whereas at longer pulse widths the curve becomes asymptotic and flattens out (23,26).

Programming the output near the flat end of the strength–duration curve tends to be inefficient, as a point of diminishing returns is reached (prolonging pulse width does not enable further reductions in voltage). This minimum voltage at which the heart can be stimulated regardless of the pulse width is called the *rheobase* (Fig. 2.11). The pulse width that corresponds to two times the voltage at rheobase is known as the *chronaxie*; it closely approximates the point of minimum threshold stimulation energy (27,28).

The following formula describes the relationship of stimulus voltage, current, and pulse duration to stimulus energy:

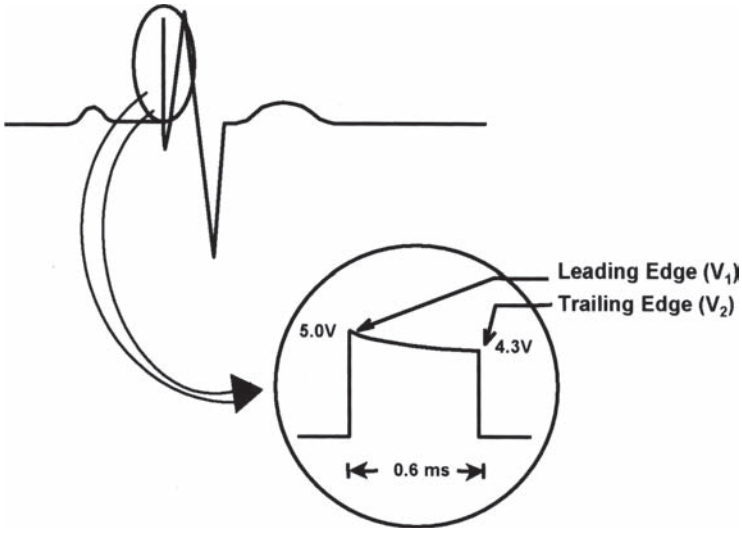

Fig. 2.10 Typical pacemaker output pulse showing leading and trailing edge voltage.

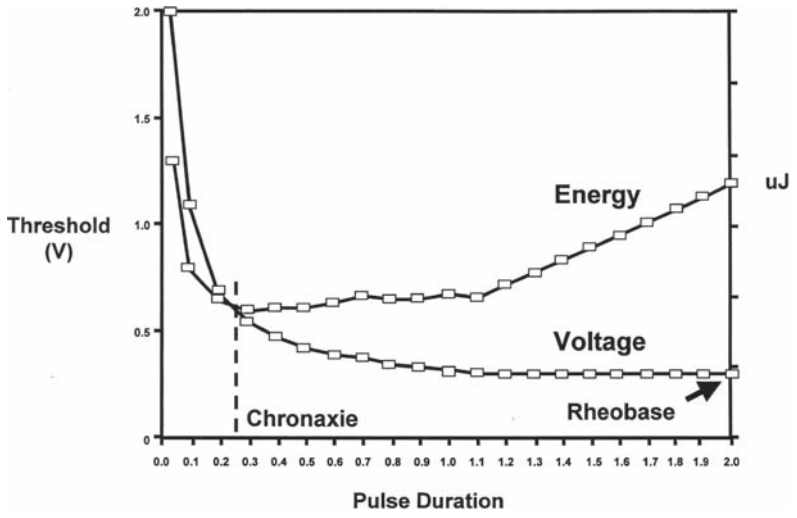

Fig. 2.11 Strength–duration curve shows relationship between stimulation voltage and pulse width. Points on and above the curve capture, whereas those below do not. At short pulse widths, a small change in pulse width is associated with significant change in threshold amplitude; however, this is not the case at longer pulse durations, where only a small change is seen. (Reprinted from Ellenbogen KA, ed. *Cardiac Pacing*, 2nd ed. Oxford: Blackwell Science, 1992, with permission.)

$$E = V^2 / R \times t$$

where E is the stimulus energy in microjoules; V, the stimulus voltage in volts; R, the total pacing impedance in KΩ; and t, the pulse width in ms. The chronaxie represents the point of minimum threshold energy on the strength–duration curve; at greater pulse widths only a slight reduction in threshold voltage is seen, whereas at lesser pulse widths threshold voltage and stimulation energy

steeply increase. Thus, chronaxie pulse width and the point of minimal stimulation energy are usually fairly close.

It is essential to understand the threshold strength–duration relationship in order to be able to appropriately program stimulus amplitude and pulse width. For the most part, pulse generators allow evaluation of stimulation threshold by automatically decrementing either stimulus voltage (at a constant pulse width), or of pulse width (at a constant stimulus voltage). Considered independently of pulse width, stimulation voltage is usually programmed to about twice the threshold value so as to provide an adequate safety margin; if only pulse width is considered, the pulse width is generally programmed to at least three or more times the threshold value. Such methods provide similar safety margins with a threshold pulse duration of 0.20 ms or less; however, because of the flattened right tail of the strength–duration curve, an adequate stimulation safety margin may not result from tripling a threshold pulse duration greater than 0.3 ms (24).

Most modern pulse generators can reliably detect if an output pulse has resulted in appropriate cardiac depolarization by sensing the evoked potential of repolarizing tissue. This allows the pulse generator to automatically vary the output "on the fly" while maintaining an adequate safety margin. The result is some increase in longevity and reduction in the burden of checking thresholds and output adjustments during follow-up.

Factors such as method of measurement, type of electrode, drugs, and duration of lead implantation affect the threshold and strength–duration curve (29–31). In addition, a factor known as the *Wedensky effect* should sometimes be taken into consideration (32). The Wedensky effect posits that, when stimulation thresholds are measured by decrementing the stimulus voltage until loss of capture, the threshold is usually lower by 0.1–0.2 V than when gradually increased from subthreshold until capture is achieved.

Residual voltage may reside on a capacitor following discharge. This is because the pulse width may terminate discharge before all stored energy has had a chance to dissipate. This is an advantage because, when it is time for the next pacing pulse, it does not need to be charged from ground zero. When doing manual thresholds, however, this can create some distortions in threshold measurement. When a device is programmed from very high to very low voltage, it may take several cycles for the output capacitor to discharge enough energy to reach a lower voltage. Some manufacturers design in a means to accelerate the draining of energy following pulses. If taking threshold measurements and jumping from 8 to 1 V, wait a few cycles (8–10) to make sure the lower voltage has been reached.

Telemetry and Communications Circuit

Telemetry is a term used to describe measurement at a distance. Pacing devices are capable of wireless bidirectional telemetry; that is, the pulse generator and the programmer are able to transmit from one to the other. Telemetry is essential for modern pacemakers, which have so many programmable parameters and unique functions that may need to be adjusted or turned on or off (18,33). Some of the programmable parameters in modern pacemakers can be found in Table 2.4.

Table 2.4 Programmable parameters in modern pacemakers.

Pacing mode	A-V delay after pace/sense
Polarity	Magnet response
Lower rate	ERI mode reversion
Maximum pacing rate	Mode switching adjustments
Atrial/ventricular pulse widths	Setup of diagnostic functions (e.g., storage of IEGMs)
Atrial/ventricular pulse amplitudes	
Atrial/ventricular sensitivity	Antitachycardia features
Atrial/ventricular refractory period	Noninvasive programmed stimulation
Atrial/ventricular blanking period	Sleeping rate
Postventricular atrial refractory period	Hysteresis rate
Atrial refractory extension	Telemetry ON/OFF
Adaptive A-V delay	Rate responsive or sensor settings

When devices were nonprogrammable or had one or two programmable parameters, the ability to interrogate was not that important. One of the first units capable of transmitting information was a rechargeable pacemaker sold in the 1970s; telemetry was used to confirm proper alignment of the recharging head (18).

The next evolution was confirmation of programming. This was important, especially for parameters that were not obvious or visible on the surface ECG, such as pulse width and sensitivity. When multiprogrammable devices were introduced, telemetry was broadened to include stored programmable settings so that the physician could interrogate and get the programmable rate, output, and sensitivity. In the late 1970s pacemakers still were not able to assess information about the battery or lead system through the pacemaker programmer. The decay of the output pulse is dependent on the lead impedance. The slope of the wave-form between the pulse leading and trailing edge has been used as a relative indicator of lead impedance, so it was possible to tell if lead impedance was increasing or decreasing by looking at the slope from one visit to the next. Similarly, amplitude was examined as a reflection of battery voltage, because pacemakers in those days did not have regulated outputs (Fig. 2.12).

By the early to mid-1980s most pacemakers could directly measure lead impedance and battery voltage via telemetry. Some could even transmit this over the telephone along with programming data. As devices became more sophisticated – with a dozen or more programmable parameters – faster communication schemes were required to keep total communication time down to acceptable levels. One way to minimize communication time involved single-parameter programming. For example, initially if you wanted to program rate and output, you would have to do so sequentially. Faster communication schemes allowed programming of multiple parameters at the same time known as batch programming. All modern pacemakers allow batch programming (34).

A diagnostic tool that has achieved popularity is the ability to transmit an intracardiac electrogram, or IEGM, which enables the clinician to assess what the pacemaker is seeing. Much of this information is analyzed and transmitted in digital format. IEGMs can be transmitted in real time or stored and

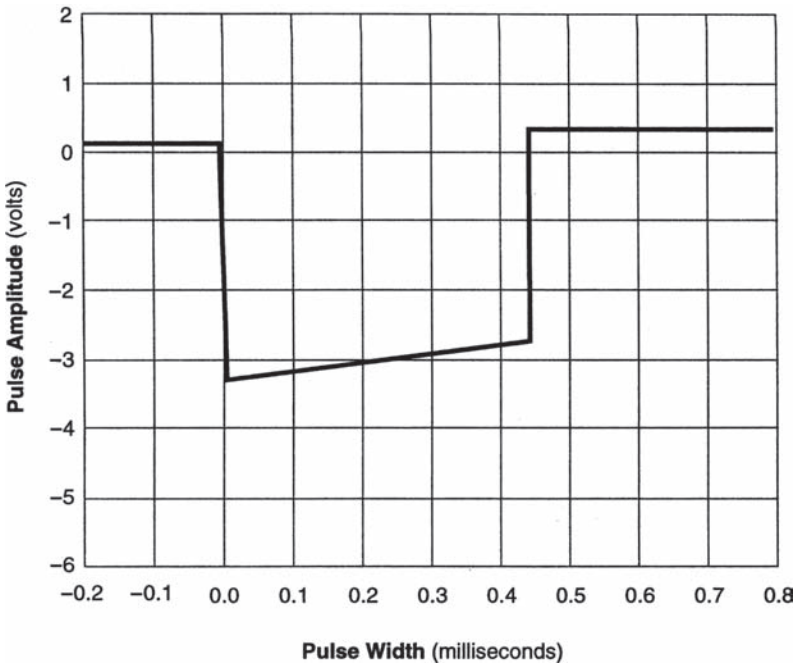

Fig. 2.12 Diagrammatic representation of an output pulse as it would appear on an oscilloscope.

retrieved for later use. Stored waveforms capability arrived later because it required considerable memory.

Pacemaker Interrogator

Programming Methods and Schemes

One of the simplest programming schemes involved the use of a magnet. Some early pacemakers could be programmed merely by holding a magnet directly over the pulse generator. When the reed switch closed, the pacemaker would step through a series of programmable rates; once the desired rate was achieved, the magnet was removed. In order to program a lower rate, the magnet would be held in place until the highest rate was achieved, after which the rate would jump to the lowest rate option, and the cycle would start again.

A more sophisticated magnetic programming scheme involves a pulsed electromagnetic field, usually generated by an electromagnet and picked up by a coil. Inductive coupling is used, whereby the magnetic field permeates the wire coil, inducing current flow in the coil. These currents are then picked up and decoded to form a programming command.

Better communication circuits involve the use of radiofrequency transmission, and incorporate a coil that acts as an antenna to receive incoming signals and encoded instructions. The same radiofrequency coil is used to transmit information back to the programmer. One relatively simple scheme is called *pulse position modulation*. This is a form of digital communication using what amounts to a Morse code, involving electromagnetic signals that are usually of a

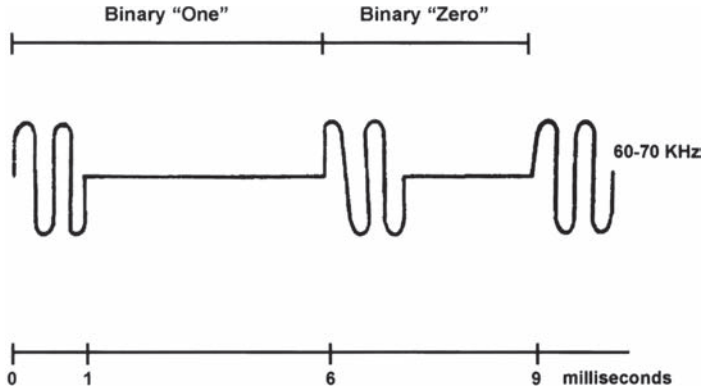

Fig. 2.13 Pulse position modulation scheme for radiofrequency communication between pacemaker and programmer.

specific frequency. The receiving coil is tuned to that frequency (called the *carrier frequency*). The carrier frequency is modulated to encode the information transmitted. There are two methods: amplitude modulation (AM) and frequency modulation (FM). In pacemakers, the common method is to turn the radiofrequency carrier on and off very rapidly to represent a binary code so that a "1" or a "0" would be represented by pulses (short versus long) (Fig. 2.13).

The higher the frequency of the radiofrequency transmission, the narrower the time interval between energy pulses and the more "1s" and "0s" that can fit into a unit of time. There are constraints with low-power radiofrequency transmission having to pass through the skin, air, body tissues, and metal shielding of the can; these become more pronounced at higher frequencies. All this serves to limit the transfer rate of information. In addition, because of the critical nature of the information, it needs to be encoded to provide security against error misprogramming. Therefore, for example, if a single parameter is programmed, signals will often echo back to the programmer to ensure that it was interpreted correctly. This takes additional time. Sometimes redundant information is sent just to ensure accuracy (or for security). Data encoding schemes for modern pacemakers are very sophisticated, so misprogramming owing to external sources of interference should be extremely rare. Encoded pulses generally occur with very precise timing. In addition, the transmission has to satisfy several conditions, one of which is that the electromagnetic energy detected by the pacemaker must be high enough to overcome the insensitivity of its receiving coil. This eliminates interference from other noise sources that have relatively low field strengths, and may also eliminate signals with frequencies outside the tuned range.

Sources of interference can prevent programming or communication, although misprogramming is unlikely as a result of these conditions. For example, a cathode-ray TV screen or similar medical equipment, if in close proximity to the programmer, might prevent programming from occurring. In addition, the programmer has to receive signals back from the pacemaker, very limited power source. Thus means that signals from the pacemaker are much weaker than those able to be sent by the AC line-powered programmer. Further, if the pacemaker is buried deep in a pocket, it usually requires that the wand of the programmer increase its receiving sensitivity, making it more susceptible to environmental noise.

Some programming schemes require dual interlock, where a reed switch must be closed before the device will be able to accept programming commands. Reed switch closure can also instruct the pacemaker to transmit a signal. This can be used as a locator beacon to properly align the programming wand.

Programmer software is constantly being updated by manufacturers to correct errors or malfunctions, add features, and expand or limit the range of programmable parameters. If there is a problem interrogating or programming a device, check with the manufacturer to ensure that the software level is appropriate for the model being programmed.

The latest improvement allows communication to occur over greater distances (20 ft.) and increased speeds without the aid of the programming wand or head. Specific frequency bands have been dedicated to medical devices so as to avoid interference with other types of equipment which rely on radio frequencies for their operation. This also enables pulse generators to be interrogated at home without active participation by the patient and has opened the door to remote patient follow-up.

Pacemaker Interrogation

Interrogation is important for modern pacemakers to ascertain the pacemaker's programmed parameters, and the state of the battery and lead; it also allows downloading of stored diagnostic data to evaluate the pacemaker–patient interaction. Interrogation often includes identification of the pacemaker model and serial number as well as determination of whether some special condition exists (e.g., replacement indicator, noise reversion, backup mode, etc). It is recommended that, prior to programming, the clinician routinely interrogate the system and, in fact, many programming protocols require interrogation to be the first operation (34,35).

There are a variety of interrogation data, generally falling into two categories: real-time measured data and stored event or programmed data. The former are immediate and instantaneous events as they occur, whereas the latter are previously recorded or programmed. Interrogation should include real-time measured data, which enable checking of lead integrity and battery status. A number of parameters can be measured for each pacing cycle and displayed as real-time data, including pulse amplitude, pulse width, sensitivity (if auto-gain is present in the device), current consumption, battery voltage, battery impedance, and lead impedance. Because these are measured values, they are only accurate within a specified tolerance. Repeated interrogation of measured parameters may seem very consistent and not vary by the tolerance specified, but the absolute value of that parameter may vary by the tolerance. For example, taking three readings of battery voltage might result in readings of, perhaps, 2.75, 2.74, and 2.75. Yet the actual values might be off by 10% from the real value of the voltage. This is why the battery voltage measured by telemetry is rarely the sole indicator of battery replacement. Instead, it is used for looking at trends in battery voltage. For example, an abnormal drop in battery voltage from one visit to the next might signify a problem that needs closer scrutiny. Knowing the total current drain from the battery allows for a rough calculation of longevity. Some of the newer devices have a gas gauge to estimate longevity. These are rough approximations because the exact battery capacity also varies from battery to battery and is dependent on many other variables.

Lead Impedance

The noninvasive measurement of lead impedance was one of the earliest uses of measured pacemaker telemetry. Following the deployment of lithium iodide it soon became clear that the pacing lead was more likely to fail than the pulse generator. In fact, lead failure has been responsible for most of the serious recalls and advisories in the pacing industry in the past 30 years (7,36,37). Therefore, it should come as no surprise that lead impedance telemetry became a valuable diagnostic tool to help with the assessment of lead integrity (38,39).

There are various techniques for measuring lead impedance. Voltage at the beginning and end of the output pulse is measured on the output capacitor in some pacemakers. This voltage decay is a function of the lead impedance. The higher the lead impedance, the slower the decay on the output capacitor. Because impedance changes throughout a pacing pulse, this method provides an average impedance and may differ slightly when compared to values obtained at implant by a pacing analyzer that generally measures impedance at the beginning of the pulse. Lead impedances generally do not change dramatically (>200 Ω) from one follow-up to the next unless there is a pending or immediate problem (40).

A drop in impedance is generally the result of failing insulation. In the case of a unipolar lead, this failure exposes the pacing coil to blood and allows current to leak directly back to the can from the site of the insulation breach. This results in less energy reaching the heart, which may cause loss of capture. In bipolar pacing systems an insulation breach sometimes occurs between the two conductor coils, which can result in a direct short, attenuating the energy being delivered to the heart and also causing sensing failure. When this occurs in a coaxial bipolar lead, a couple of things can happen. Inhibition or triggering can occur when the two conductors make and break contact (which will appear to be oversensing on the surface ECG); and/or loss of capture or inappropriate sensing may occur owing to the attenuation of pacing current and IEGM amplitude. Programming to a unipolar configuration may temporarily alleviate the loss of capture but does not prevent oversensing when the conductors make and break contact, so prompt lead replacement is recommended.

Both conductor fractures and connection problems can cause lead impedance to rise, often to high levels. This sets off a chain of events: the flow of current lessens so there is an attenuated output and capture is lost, and battery current drain decreases. Sensing can also be affected, with oversensing due to make-or-break connections or undersensing. Because the measurement of lead impedance occurs only during the pacing pulse, which is a fraction of the pacing cycle, intermittent fracture may not be reflected in the measured values; therefore, repeated measurements should be made if trouble is suspected. When troubleshooting, it may also be necessary to try other techniques for placing stress on the lead to induce the coils to separate (e.g., applying pressure or asking the patient to raise or move the arm on the side where the device is implanted). Some devices have the capability of taking beat-by-beat lead impedance measurements over a prolonged period and displaying them graphically, increasing the sensitivity/specificity of lead impedance as a diagnostic tool (Fig. 2.14) (41).

Once again, the cautious approach is to take repeated measurements if an unanticipated value is encountered. Repeat the measurement and correlate

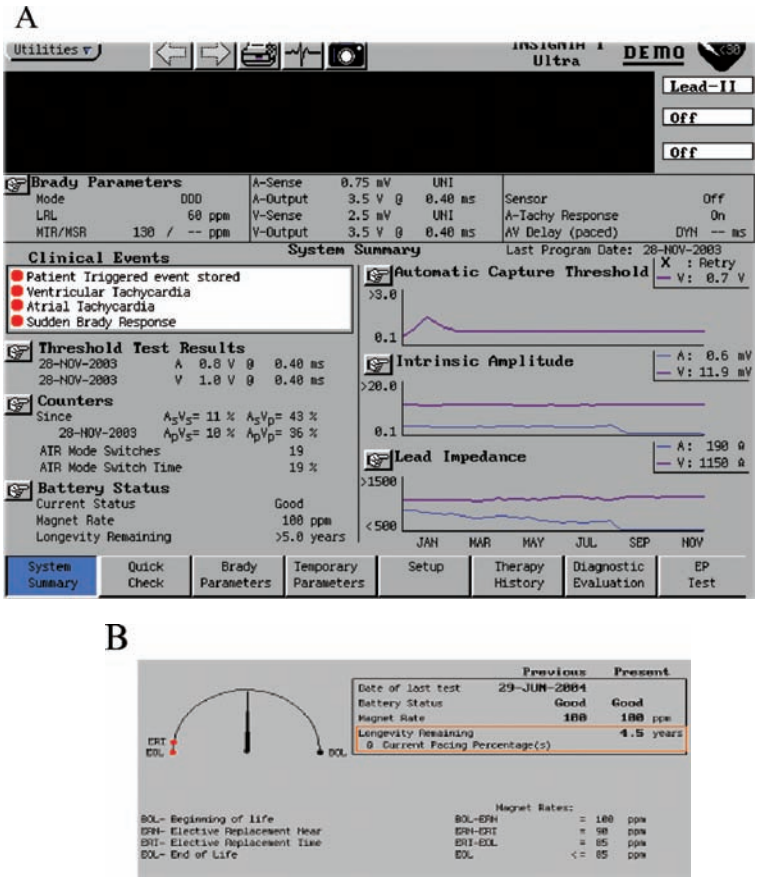

Fig. 2.14 (**a**). Status screen and modern pulse generator capable of storing sequential measurements of lead impedance and displaying the trend graphically. (**b**). Interrogator screen shows battery status in an easily understandable pattern.

with clinical symptoms (such as lack of sensing) if readings are high. It is an unusual case when a physician would take remedial action based on telemetry alone, with the possible exception of a device under an advisory that has a failure mechanism with a known footprint related to a dramatic change in lead impedance or other measured value.

Stored Data

Implantable devices are now capable of accumulating information on patient–device interaction. This is stored digitally and accessible via the programmer. Most devices report on the percentage of paced versus sensed events by chamber. This can be further subdivided by rate range. An example of one such display is shown in Fig. 2.15.

These data are useful for assessing device utilization as well as for providing information on tuning the lower rate and rate–response settings. For example, making a 5 ppm adjustment in the lower rate can dramatically lower the percentage of pacing during sleeping hours.

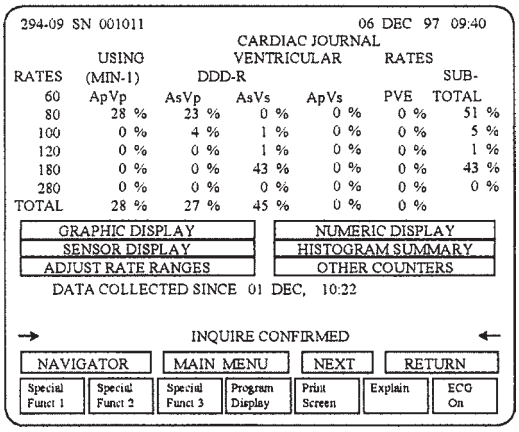

Fig. 2.15 Interrogation screen from Intermedics Marathon DR pacemaker showing percentages of cycles in five categories by rate range (ApVp, atrial pace followed by ventricular pace; AsVp, atrial sense followed by ventricular pace, etc.; PVE, premature ventricular event).

Stored data are particularly useful in helping to assess the frequency and duration of tachyarrhythmias. Digital stored data include mode switching episodes, tachycardia counters, and segments of stored IEGMs from either or both chambers. Most of the aforementioned are time stamped. The amount of stored data is constantly increasing because of the availability of low-power memory and may, in the very near future, have many of the capabilities of external cardiac event recorders (42). The physician is also given the option of clearing the diagnostic data registers, and should be aware that certain programming operations may clear them unintentionally. It is always prudent to print out any diagnostic data that you would like to access at the time of the initial interrogation, before any reprogramming takes place.

Clinical Issues

Although the historical evolution of cardiac pacing technology is interesting from an engineering perspective and bears directly on the operating characteristics of today's devices, it is the clinical observations that are of overriding interest. Knowing how the pacemaker works can help the clinician distinguish a true malfunction that requires intervention from seemingly "quirky" behavior that can arise in the presence of external interference or unusual cardiac rhythms. An understanding of the design considerations and operating characteristics can prevent an inappropriate intervention (e.g., explant) in response to a behavior that may appear wrong to the clinician, yet may in fact represent normal operation. Just like a computer, pacemakers generally "do what they are told."

All the major pacing companies have 24 h technical assistance help lines staffed with individuals trained to assist with troubleshooting their company's devices. The physicians' manual is an excellent reference for learning how the device works, although it rarely covers all the idiosyncrasies one might encounter in the clinic.

Without getting into the complexity of design, it should be noted that each manufacturer has its own way of implementing even basic functions. This chapter

focuses on some of the more common behaviors, and the reader should recognize that a specific device may be at variance with what has been presented here.

New circuit designs and battery chemistries in future devices will, in all likelihood, change the way devices work. The trends are already apparent, with increasing automatic functions and greater degrees of data storage and retrieval. Many of these efforts are focused on improving ease of use rather than changing the therapeutic benefits of pacing. For example, devices now have the ability to continuously adjust the pacing output, sensitivity, and AV delay. When combined with the capability of longer-distance telemetry the need for an in-office assessment may be reduced or eliminated. Design engineers are constantly trying to balance the need for clinical flexibility and ease of use. Future generations of microprocessors and improvements in packaging technology will enable pacemakers to take on tasks that normally require the intervention of the clinician. Although increasing automation and ease of use appear likely to decrease the amount of time and effort needed to administer pacemaker therapy, a basic understanding of the engineering considerations always benefits both physician and pacemaker recipient.

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