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Implantable Cardiac Pacemakers

77.1 Indications .............................................................................. 77-1
77.2 Pulse Generators ..................................................................... 77-2
Sensing Circuit • Output Circuit • Timing Circuit •
Telemetry Circuit • Power Source
77.3 Leads .................................................................................. 77-7
77.4 Programmers ......................................................................... 77-8
77.5 System Operation .................................................................... 77-9
77.6 Clinical Outcomes and Cost Implications .............................. 77-10
77.7 Conclusion ............................................................................ 77-11

The practical use of an implantable device for delivering a controlled, rhythmic electric stimulus to maintain the heartbeat is relatively recent. Cardiac pacemakers have been in clinical use only slightly more than 30 years. Although devices have gotten steadily smaller over this period (from 250 grams in 1960 to 25 grams today), the technological evolution goes far beyond size alone. Early devices provided only single-chamber, asynchronous, nonprogrammable pacing coupled with questionable reliability and longevity. Today, advanced electronics afford dual-chamber multiprogrammability, diagnostic functions, rate response, data collection, and exceptional reliability, and lithium-iodine power sources extend longevity to upward of 10 years. Continual advances in a number of clinical, scientific, and engineering disciplines have so expanded the use of pacing that it now provides cost-effective benefits to an estimated 350,000 patients worldwide each year.

The modern pacing system is comprised of three distinct components: pulse generator, lead, and programmer (Fig. 77.1). The pulse generator houses the battery and the circuitry which generates the stimulus and senses electrical activity. The lead is an insulated wire that carries the stimulus from the generator to the heart and relays intrinsic cardiac signals back to the generator. The programmer is a telemetry device used to provide two-way communications between the generator and the clinician. It can alter the therapy delivered by the pacemaker and retrieve diagnostic data that are essential for optimally titrating that therapy. Ultimately, the therapeutic success of the pacing prescription rests on the clinician's choice of an appropriate system, use of sound implant technique, and programming focused on patient outcomes.

This chapter discusses in further detail the components of the modern pacing system and the significant evolution that has occurred since its inception. Our focus is on system design and operations, but we also briefly overview issues critical to successful clinical performance.

77.1 Indications

The decision to implant a permanent pacemaker for bradyarrhythmias usually is based on the major goals of symptom relief (at rest and with physical activity), restoration of functional capacity and quality of life, and preservation of well-being. Additionally, the indication may be based on the patient's clinical condition, which includes factors such as comorbidities, age, and life expectancy.
of life, and reduced mortality. As with other healthcare technologies, appropriate use of pacing is the intent of indications guidelines established by Medicare and other third-party payors.

In 1984 and again in 1991, a joint commission of the American College of Cardiology and the American Heart Association established guidelines for pacemaker implantation [Committee on Pacemaker Implantation, 1991]. In general, pacing is indicated when there is a dramatic slowing of the heart rate or a failure in the connection between the atria and ventricles resulting in decreased cardiac output manifested by such symptoms as syncope, light-headedness, fatigue, and exercise intolerance. Failure of impulse formation and/or conduction is the overriding theme of all pacemaker indications. There are four categories of pacing indications:

1. Heart block (e.g., complete heart block, symptomatic 2° AV block)
2. Sick sinus syndrome (e.g., symptomatic bradycardia, sinus arrest, sinus exit block)
3. Myocardial infarction (e.g., conduction disturbance related to the site of infarction)
4. Hypersensitive carotid sinus syndrome (e.g., recurrent syncope)

Within each of these four categories the ACC/AHA provided criteria for classifying a condition as group I (pacing is considered necessary), group II (pacing may be necessary), or group III (pacing is considered inappropriate).

New indications for cardiac pacing are being evaluated under the jurisdiction of the Food and Drug Administration. For example, hypertrophic obstructive cardiomyopathy (HOCM) is one of these new potential indications, with researchers looking at dual-chamber pacing as a means of reducing left ventricular outflow obstruction. Though efforts in these areas are ongoing and expanding, for now they remain unapproved as standard indications for pacing.

### 77.2 Pulse Generators

The pulse generator contains a power source, output circuit, sensing circuit, and a timing circuit (Fig. 77.2). A telemetry coil is used to send and receive information between the generator and the programmer. Rate-adaptive pulse generators include the sensor components along with the circuit to process the information measured by the sensor.

Modern pacemakers use CMOS circuit technology. One to 2 kilobytes of read-only memory (ROM) are used to direct the output and sensing circuits; 16–512 bytes of random-access memory (RAM) are
used to store diagnostic data. Some manufacturers offer fully RAM-based pulse generators, providing greater storage of diagnostic data and the flexibility for changing feature sets after implantation.

All components of the pulse generator are housed in a hermetically sealed titanium case with a connector block that accepts the lead(s). Because pacing leads are available with a variety of different connector sites and configurations, the pulse generator is available with an equal variety of connectors. The outer casing is laser-etched with the manufacturer, name, type (e.g., single-versus dual-chamber), model number, serial number, and the lead connection diagram for each identification. Once implanted, it may be necessary to use an x-ray to reveal the identity of the generator. Some manufacturers use radiopaque symbols and ID codes for this purpose, whereas others give their generators characteristic shapes.

Sensing Circuit

Pulse generators have two basic functions, pacing and sensing. Sensing refers to the recognition of an appropriate signal by the pulse generator. This signal is the intrinsic cardiac depolarization from the chamber or chambers in which the leads are placed. It is imperative for the sensing circuit to discriminate between these intracardiac signals and unwanted electrical interference such as far-field cardiac events, diastolic potentials, skeletal muscle contraction, and pacing stimuli. An intracardiac electrogram (Fig. 77.3) shows the waveform as seen by the pacemaker; it is typically quite different from the corresponding event as shown on the surface ECG.

Sensing (and pacing) is accomplished with one of two configurations, bipolar and unipolar. In bipolar, the anode and cathode are close together, with the anode at the tip of the lead and the cathode a ring electrode about 2 cm proximal to the tip. In unipolar, the anode and cathode may be 5–10 cm apart. The anode is at the lead tip and the cathode is the pulse generator itself (usually located in the pectoral region).

FIGURE 77.2 Internal view of pulse generator.

FIGURE 77.3 The surface ECG (ECG LEAD II) represents the sum total of the electrical potentials of all depolarizing tissue. The intracardiac electrogram (V EGM) shows only the potentials measured between the lead electrodes. This allows the evaluation of signals that may be hidden within the surface ECG.
In general, bipolar and unipolar sensing configurations have equal performance. A drawback of the unipolar approach is the increased possibility of sensing noncardiac signals: The large electrode separation may, for example, sense myopotentials from skeletal muscle movement, leading to inappropriate inhibition of pacing. Many newer pacemakers can be programmed to sense or pace in either configuration.

Once the electrogram enters the sensing circuit, it is scrutinized by a bandpass filter (Fig. 77.4). The frequency of an R-wave is 10 to 30 Hz. The center frequency of most sensing amplifiers is 30 Hz. T-waves are slower, broad signals that are composed of lower frequencies (approximately 5 Hz or less). Far-field signals are also lower-frequency signals, whereas skeletal muscle falls in the range of 10–200 Hz.

At the implant, the voltage amplitude of the R-wave (and the P-wave, in the case of dual-chamber pacing) is measured to ensure the availability on an adequate signal. R-wave amplitudes are typically 5–25 mV, and P-wave amplitudes are 2–6 mV. The signals passing through the sense amplifier are compared to an adjustable reference voltage called the sensitivity. Any signal below the reference voltage is not sensed, and those above it are sensed. Higher-sensitivity settings (high-reference voltage) may lead to substandard sensing, and a lower reference voltage may result in oversensing. A minimum 2:1 safety margin should be maintained between the sensitivity setting and the amplitude of the intracardiac signal. The circuit is protected from extremely high voltages by a Zener diode.

The slope of the signal is also surveyed by the sensing circuit and is determined by the slew rate (the time rate of change in voltage). A slew rate that is too flat or too steep may be eliminated by the bandpass filter. On the average, the slew rate measured at implant should be between 0.75 and 2.50 V/s.

The last line of defense in an effort to remove undesirable signals is to “blind” the circuit at specific times during the cardiac cycle. This is accomplished with blanking and refractory periods. Some of these periods are programmable. During the blanking period the sensing circuit is turned off, and during the refractory period the circuit can see the signal but does not initiate any of the basic timing intervals. Virtually all paced and sensed events begin concurrent blanking and refractory periods, typically ranging from 10–400 ms. These are especially helpful in dual-chamber pacemakers where there exists the potential
for the pacing output of the atrial side to inhibit the ventricular pacing output, with dangerous consequences for patients in complete heart block.

Probably the most common question asked by the general public about pacing systems is the effect of electromagnetic interference (EMI) on their operation. EMI outside of the hospital is an infrequent problem, though patients are advised to avoid such sources of strong electromagnetic fields as arc welders, high-voltage generators, and radar antennae. Some clinicians suggest that patients avoid standing near antitheft devices used in retail stores. Airport screening devices are generally safe, though they may detect a pacemaker’s metal case. Microwave ovens, ham radio equipment, video games, computers, and office equipment rarely interfere with the operation of modern pacemakers. A number of medical devices and procedures may on occasion do so, however; electrocautery, cardioversion and defibrillation, MRI, lithotripsy, diathermy, TENS units, and radiation therapy.

Pacemakers affected by interference typically respond with temporary loss of output or temporary reversion to asynchronous pacing (pacing at a fixed rate, with no inhibition from intrinsic cardiac events). The usual consequence for the patient is a return of the symptoms that originally led to the pacemaker implant.

**Output Circuit**

Pacing is the most significant drain on the pulse generator power source. Therefore, current drain must be minimized while maintaining an adequate safety margin between the *stimulation threshold* and the programmed output stimulus: Modern permanent pulse generators use constant voltage. The voltage remains at the programmed value while current fluctuates in relation to the source impedance.

Output energy is controlled by two programmable parameters, pulse amplitude and pulse duration. Pulse amplitudes range from 0.8–5 V and, in some generators, can be as high as 10 V (used for troubleshooting or for pediatric patients). Pulse duration can range from 0.05–1.5 ms. The prudent selection of these parameters will greatly influence the longevity of the pulse generator.

The output pulse is generated from the discharge of a capacitor charged by the battery. Most modern pulse generators contain a 2.8 V battery. The higher voltages are achieved using voltage multipliers (smaller capacitors used to charge the large capacitor). The voltage can be doubled by charging two smaller capacitors in parallel, with the discharge delivered to the output capacitor in series. Output pulses are emitted at a rate controlled by the timing circuit; output is commonly inhibited by sensed cardiac signals.

**Timing Circuit**

The timing circuit regulates such parameters as the pacing cycle length, refractory and blanking periods, pulse duration, and specific timing intervals between atrial and ventricular events. A crystal oscillator generating frequencies in the kHz range sends a signal to a digital timing and logic control circuit, which in turn operates internally generated clocks at divisions of the oscillatory frequency.

A rate-limiting circuit is incorporated into the timing circuit to prevent the pacing rate from exceeding an upper limit should a random component failure occur (an extremely rare event). This is also referred to as “runaway” protection and is typically 180–200 ppm.

**Telemetry Circuit**

Today’s pulse generators are capable of both transmitting information from an RF antenna and receiving information with an RF decoder. This two-way communication occurs between the pulse generator and the programmer at approximately 300 Hz. Real-time telemetry is the term used to describe the ability of the pulse generator to provide information such as pulse amplitude, pulse duration, lead impedance, battery impedance, lead current, charge, and energy. The programmer, in turn, delivers coded messages to the pulse generator to alter any of the programmable features and to retrieve diagnostic data. Coding
requirements reduce the likelihood of inappropriate programming alterations by environmental sources of radiofrequency and magnetic fields. It is also prevents the improper use of programmers from other manufacturers.

**Power Source**

Over the years, a number of different battery technologies have been tried, including mercury-zinc, rechargeable silver-modified-mercuric-oxide-zinc, rechargeable nickel-cadmium, radioactive plutonium or promethium, and lithium with a variety of different cathodes. Lithium-cupric-sulfide and mercury-zinc batteries were associated with corrosion and early failure. Mercury-zinc produced hydrogen gas as a by-product of the battery reaction; the venting required made it impossible to hermetically seal the generator. This led to fluid infiltration followed by the risk of sudden failure.

The longevity of very early pulse generators was measured in hours. With the lithium-iodide technology now used, longevity has been reported as high as 15 years. The clinical desire to have a generator that is small and full-featured yet also long-lasting poses a formidable challenge to battery designers. One response by manufacturers has been to offer different models of generators, each offering a different balance between therapy, size, and longevity. Typical battery capacity is in the range of 0.8–3.0 amp-hours.

Many factors affect longevity, including pulse amplitude and duration, pacing rate, single- versus dual-chamber pacing, degree to which the patient uses the pacemaker, lead design, and static current drain from the sensing circuits. Improvements in lead design are often overlooked as a factor in improving longevity, but electrodes used in 1960 required a pulse generator output of 675 μJ for effective stimulation, whereas the electrodes of the 1990s need only 3–6 μJ.

Another important factor in battery design lies in the electrolyte that separates the anode and the cathode. The semisolid layer of lithium iodide that is used gradually thickens over the life of the cell, increasing the internal resistance of the battery. The voltage produced by lithium-iodine batteries is inversely related to this resistance and is linear from 2.8 V to approximately 2.4 V, representing about 90% of the usable battery life. It then declines exponentially to 1.8 V as the internal battery resistance increases from 10,000 Ω to 40,000 Ω (Fig. 77.5).

When the battery reaches between 2.0 and 2.4 V (depending on the manufacturer), certain functions of the pulse generator are altered so as to alert the clinician. These alterations are called the elective-replacement indicators (ERI). They vary from one pulse generator to another and include signature decreases in rate, a change to a specific pacing mode, pulse duration stretching, and the telemetered battery voltage. When the battery voltage reaches 1.8 V, the pulse generator may operate erratically or cease to function and is said to have reached "end of life." The time period between appearance of the ERI and end-of-life status averages about 3 to 4 months.

![FIGURE 77.5](image.png) The initial decline in battery voltage is slow and then more rapid after the battery reaches the ERI voltage. An important aspect of battery design is the predictability of this decline so that timely generator replacement is anticipated.
77.3 Leads

Implantable pacing leads must be designed not only for consistent performance within the hostile environment of the body but also for easy handling by the implanting physician. Every lead has four major components (Fig. 77.6): the electrode, the conductor, the insulation, and the connector pin(s).

The electrode is located at the tip of the lead and is in direct contact with the myocardium. Bipolar leads have a tip electrode and a ring electrode (located about 2 cm proximal to the tip); unipolar leads have tip electrodes only. A small-radius electrode provides increased current density resulting in lower stimulation thresholds. The electrode also increases resistance at the electrode-myocardial interface, thus lowering the current drain further and improving battery longevity. The radius of most electrodes is 6–8 mm², though there are clinical trials underway using a “high-impedance” lead with a tip radius as low as 1.5 mm².

Small electrodes, however, historically have been associated with inferior sensing performance. Lead designers were able to achieve both good pacing and good sensing by creating porous-tip electrodes containing thousands of pores in the 20–100 μm range. The pores allow the ingrowth of tissue, resulting in the necessary increase in effective sensing area while maintaining a small pacing area. Some commonly used electrode materials include platinum-iridium, Elgiloy (an alloy of cobalt, iron, chromium, molybdenum, nickel, and manganese), platinum coated with platinized titanium, and vitreous or pyrolytic carbon coating a titanium or graphite core.

Another major breakthrough in lead design is the steroid-eluting electrode. About 1 mg of a corticosteroid (dexamethasone sodium phosphate) is contained in a silicone core that is surrounded by the electrode material (Fig. 77.7). The "leaking" of the steroid into the myocardium occurs slowly over several
years and reduces the inflammation that results from the lead placement. It also retards the growth of the fibrous sack that forms around the electrode which separates it from viable myocardium. As a result, the dramatic rise in acute thresholds that is seen with nonsteroid leads over the 8 to 16 weeks postimplant is nearly eliminated. This makes it possible to program a lower pacing output, further extending longevity.

Once a lead has been implanted, it must remain stable (or fixed). The fixation device is either active or passive. The active fixation leads incorporate corkscrew mechanisms, barbs, or hooks to attach themselves to the myocardium. The passive fixation leads are held into place with tines that become entangled into the netlike lining (trabeculae) of the heart. Passive leads generally have better acute pacing and sensing performance but are difficult to remove chronically. Active leads are easier to remove chronically and have the advantage of unlimited placement sites. Some implanters prefer to use active-fixation leads in the atrium and passive-fixation leads in the ventricle.

The conductor carries electric signals to the pulse generator and delivers the pacing pulses to the heart. It must be strong and flexible to withstand the repeated flexing stress placed on it by the beating heart. The early conductors were a single, straight wire that was vulnerable to fracturing. They have evolved into coiled (for increased flexibility) multifilar (to prevent complete failure with partial fractures) conductors. The conductor material is a nickel alloy called MP35N. Because of the need for two conductors, bipolar leads are usually larger in diameter than unipolar leads. Current bipolar leads have a coaxial design that has significantly reduced the diameter of bipolar leads.

Insulation materials (typically silicone and polyurethane) are used to isolate the conductor. Silicone has a longer history and the exclusive advantage of being repairable. Because of low tear strength, however, silicone leads tend to be thicker than polyurethane leads. Another relative disadvantage of silicone is its high coefficient of friction in blood, which makes it difficult for two leads to pass through the same vein. A coating applied to silicone leads during manufacturing has diminished this problem. A variety of generator-lead connector configurations and adapters are available. Because incompatibility can result in disturbed (or even lost) pacing and sensing, an international standards (IS-1) has been developed in an attempt to minimize incompatibility.

Leads can be implanted epicardially and endocardially. Epicardial leads are placed on the outer surface of the heart and require the surgical exposure of a small portion of the heart. They are used when venous occlusion makes it impossible to pass a lead transvenously, when abdominal placement of the pulse generator is needed (as in the case of radiation therapy to the pectoral area), or in children (to allow for growth). Endocardial leads are more common and perform better in the long term. These leads are passed through the venous system and into the right side of the heart. The subclavian or cephalic veins in the pectoral region are common entry sites. Positioning is facilitated by a thin, firm wire stylet that passes through the central lumen of the lead, stiffening it. Fluoroscopy is used to visualize lead positioning and to confirm the desired location.

Manufacturers are very sensitive to the performance reliability of the leads. Steady improvements in materials, design, manufacturing, and implant technique have led to reliability well in excess of 99% over 3-year periods.

### 77.4 Programmers

Noninvasive reversible alteration of the functional parameters of the pacemaker is critical to ongoing clinical management. For a pacing system to remain effective throughout its lifetime, it must be able to adjust to the patient's changing needs. The programmer is the primary clinical tool for changing settings, for retrieving diagnostic data, and for conducting noninvasive tests.

The pacing rate for programmable pacemakers of the early 1960s was adjusted via a Keith needle manipulated percutaneously into a knob on the side of the pacemaker; rotating the needle changed the pacing rate. Through the late 1960s and early 1970s, magnetically attuned reed switches in the pulse generator made it possible to noninvasively change certain parameters such as rate, output, sensitivity, and polarity. The application of a magnet could alter the parameters which were usually limited to only one of two choices. It wasn't until the late 1970s, when radiofrequency energy was incorporated as the
Implantable Cardiac Pacemakers

transmitter of information, that programmability began to realize its full potential. Radiofrequency transmission is faster, provides bidirectional telemetry, and decreases the possibility of unintended programming from inappropriate sources.

Most manufacturers today are moving away from a dedicated proprietary instrument and toward a PC-based design. The newer designs are generally more flexible, more intuitive to use, and more easily updated when new devices are released. Manufacturers and clinicians alike are becoming more sensitive to the role that time-efficient programming can play in the productivity of pacing clinics, which may provide follow-up for as many as 500–1000 patients a year.

77.5 System Operation

Pacemakers have gotten steadily more powerful over the last three decades, but at the cost of steadily greater complexity. Manufacturers have come to realize the challenge that this poses for busy clinicians and have responded with a variety of interpretive aids (Fig. 77.8).

Much of the apparent complexity of the timing rules that determine pacemaker operation is due to a design goal of mimicking normal cardiac function without interfering with it. One example is the dual-chamber feature that provides sequential stimulation of the atrium before the ventricle.

Another example is rate response, designed for patients who lack the normal ability to increase their heart rate in response to a variety of physical conditions (e.g., exercise). Introduced in the mid-1980s, rate-responsive systems use some sort of sensor to measure the change in a physical variable correlated to heart rate. The sensor output is signal-processed and then used by the output circuit to specify a target pacing rate. The clinician controls the aggressiveness of the rate increase through a variety of parameters (including a choice of transfer function); pacemaker-resident diagnostics provide data helpful in titrating the rate-response therapy.

The most common sensor is the activity sensor, which uses piezoelectric materials to detect vibrations caused by body movement. Systems using a transthoracic-impedance sensor to estimate pulmonary minute ventilation are also commercially available. Numerous other sensors (e.g., stroke volume, blood

![Marker Channel Diagram]

FIGURE 77.8 The Marker Channel Diagram is just one tool that makes interpretation of the ECG strip faster and more reliable for the clinician. It allows quick checking of the timing operations of the system.
TABLE 77.1 The NASPE/NPEG Code

<table>
<thead>
<tr>
<th>Position</th>
<th>I</th>
<th>II</th>
<th>III</th>
<th>IV</th>
<th>V</th>
</tr>
</thead>
<tbody>
<tr>
<td>Category</td>
<td>Chamber(s) paced</td>
<td>Chamber(s) sensed</td>
<td>Response to sensing</td>
<td>Programmability rate modulation</td>
<td>Antitachycardia arrhythmia function(s)</td>
</tr>
<tr>
<td>O = None</td>
<td>O = None</td>
<td>O = None</td>
<td>O = None</td>
<td>O = None</td>
<td>O = None</td>
</tr>
<tr>
<td>A = Atrium</td>
<td>A = Atrium</td>
<td>T = Triggered</td>
<td>P = Simple programmable</td>
<td>M = Multiprogrammable</td>
<td>S = Shock</td>
</tr>
<tr>
<td>V = Ventricle</td>
<td>V = Ventricle</td>
<td>I = Inhibited</td>
<td>C = Communicating</td>
<td>R = Rate modulation</td>
<td>D = Dual (P+S)</td>
</tr>
<tr>
<td>D = Dual (A+V)</td>
<td>D = Dual (A+V)</td>
<td>D = Dual (T+I)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Manufacturers' designation only

S = Single (A or V) S = Single (A or V)

Note: Positions I through III are used exclusively for antibradyarrhythmia function. (From Bernstein AD, et al., PACE, Vol. 10, July-Aug. 1987.)

temperature or pH, oxygen saturation, preejection interval, right ventricular pressure) are in various stages of clinical research or have been market released outside the United States. Some of these systems are dual-sensor, combining the best features of each sensor in a single pacing system.

To make it easier to understand the gross-level system operation of modern pacemakers, a five-letter code has been developed by the North American Society of Pacing and Electrophysiology and the British Pacing and Electrophysiology Group [Bernstein et al., 1987]. The first letter indicates the chamber (or chambers) that are paced. The second letter reveals those chambers in which sensing takes place, and the third letter describes how the pacemaker will respond to a sensed event. The pacemaker will "inhibit" the pacing output when intrinsic activity is sensed or will "trigger" a pacing output based on a specific previously sensed event. For example, in DDD mode:

D: Pacing takes place in the atrium and the ventricle.
D: Sensing takes place in the atrium and the ventricle.
D: Both inhibition and triggering are the response to a sensed event. An atrial output is inhibited with an atrial-sensed event, whereas a ventricular output is inhibited with a ventricular-sensed event; a ventricular pacing output is triggered by an atrial-sensed event (assuming no ventricular event occurs during the A-V interval).

The fourth letter in the code is intended to reflect the degree of programmability of the pacemaker but is typically used to indicate that the device can provide rate response. For example, a DDDR device is one that is programmed to pace and sense in both chambers and is capable of sensor-driven rate variability. The fifth letter is reserved specifically for antitachycardia functions (Table 77.1).

Readers interested in the intriguing details of pacemaker timing operations are referred to the works listed at the end of this chapter.

77.6 Clinical Outcomes and Cost Implications

The demonstrable hemodynamic and symptomatic benefits provided by rate-responsive and dual-chamber pacing have led U.S. physicians to include at least one of these features in over three-fourths of implants in recent years. Also, new prospective data [Andersen et al., 1993] support a hypothesis investigated retrospectively since the mid-1980s: namely, that pacing the atrium in patients with sinus node dysfunction can dramatically reduce the incidence of such life-threatening complications as congestive heart failure and stroke associated with chronic atrial fibrillation. Preliminary analysis of the cost implications suggest that dual-chamber pacing is significantly cheaper to the U.S. healthcare system than is single-chamber pacing over the full course of therapy, despite the somewhat higher initial cost of implanting the dual-chamber system.
77.7 Conclusion

Permanent cardiac pacing is the beneficiary of three decades of advances in a variety of key technologies: biomaterials, electrical stimulation, sensing of bioelectrical events, power sources, microelectronics, transducers, signal analysis, and software development. These advances, informed and guided by a wealth of clinical experience acquired during that time, have made pacing a cost-effective cornerstone of cardiac arrhythmia management.

Defining Terms

Atrial fibrillation: An atrial arrhythmia resulting in chaotic current flow within the atria. The effective contraction of the atria is lost, allowing blood to pool and clot, leading to stroke if untreated.

Battery capacity: Given by the voltage and the current delivery. The voltage is a result of the battery chemistry, and current delivery (current × time) is measured in ampere hours and is related to battery size.

CMOS circuit: Abbreviation for complementary metal-oxide semiconductor, which is a form of semiconductor often used in pacemaker technology.

Congestive heart failure: The pathophysiologic state in which an abnormality of cardiac function is responsible for the failure of the heart to pump blood at a rate commensurate with the requirements of the body.

Endocardium: The inner lining of the heart.

Epicardium: The outer lining of the heart.

Hermeticity: The term, as used in the pacemaker industry, refers to a very low rate of helium gas leakage from the sealed pacemaker container. This reduces the chance of fluid intruding into the pacemaker generator and causing damage.

Hypertrophic obstructive cardiomyopathy: A disease of the myocardium characterized by thickening (hypertrophy) of the interventricular septum, resulting in the partial obstruction of blood from the left ventricle.

Minute ventilation: Respiratory rate × tidal volume (the amount of air taken in with each breath) = minute ventilation. This parameter is used as a biologic indicator for rate-adaptive pacing.

Mode: The type of pacemaker response to the patient’s intrinsic heartbeat. The three commonly used modes are asynchronous, demand, and triggered.

Programmable: The ability to alter the pacemaker settings noninvasively. A variety of selections exist, each with its own designation.

Rate-adaptive: The ability to change the pacemaker stimulation interval caused by sensing a physiologic function other than the intrinsic atrial rhythm.

Sensitivity: A programmable setting that adjusts the reference voltage to which signals entering the sensing circuit are compared for filtering.

Stimulation threshold: The minimum output energy required to consistently “capture” (cause depolarization) of the heart.

References


Further Information

A good basic introduction to pacing from a clinical perspective is the third edition of *A Practical Guide to Cardiac Pacing* by H. Weston Moses, Joel Schneider, Brain Miller, and George Taylor (Little, Brown, 1991).

*Cardiac Pacing* (Blackwell Scientific, 1992), edited by Kenneth Ellenbogen, is an excellent intermediate treatment of pacing. The treatments of timing cycles and troubleshooting are especially good.

In-depth discussion of a wide range of pacing topics is provided by the third edition of *A Practice of Cardiac Pacing* by Seymour Furman, David Hayes, and David Holmes (Futura, 1993), and by *New Perspectives in Cardiac Pacing* 3, edited by Serge Barold and Jacques Mugica (Futura, 1993).

Detailed treatment of rate-responsive pacing is given in *Rate-Adaptive Cardiac Pacing: Single and Dual Chamber* by Chu-Pak Lau (Futura, 1993), and in *Rate-Adaptive Pacing*, edited by David Benditt (Blackwell Scientific, 1993).


Readers seeking a historical perspective may wish to consult "Pacemakers, Pastmasters, and the Paced: An Informal History from A to Z," by Dwight Harken in the July/August 1991 issue of *Biomedical Instrumentation and Technology*.

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