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Cardiac pacemakers: a basic review of the history and current technology

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Abstract   In the 60 years since the first human implant of a cardiac pacemaker, tremendous improvements have been made to devices themselves as well as the lead systems. Improvement in battery materials has allowed for production of smaller devices with greater longevity and a vast array of technologies allowing for communication between the device and the operator. Lead wires, typically to as the weakest part of the pacing system, have also seen a metamorphosis as improvements in conductor materials and hybrid insulation have been shown to improve reliability. With the recent development of leadless pacing systems, the downfalls of implantable leads can be avoided. These improvements have allowed a more widespread use of cardiac pacing in veterinary applications since the first reported canine implant in 1967.
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Brief history of cardiac pacing

The first implant

In 1958, Arne Larsson, a 43-year-old Swedish man, was suffering from severe Stokes-Adams attacks. He required almost daily resuscitations and was on the verge of giving up. His wife had learned of a physician in Stockholm, Sweden, who had invented a device that could stimulate the heart. This was

being investigated in animals, but there was no human experience.

In the fall of 1958, in a secret operation, Dr. Rune Elmqvist, who developed the device, and Dr. Åke Senning, a cardiologist, implanted the device in Mr. Larsson [1]. Dr. Elmqvist developed the device while working as an engineer for Elema-Schönander (later Siemens-Elema). Even though the first device lasted only a few hours, a replacement device was available for implant. Mr. Larsson did remarkably well, receiving a total of 26 devices in his lifetime. He passed away in 2001 after a battle with melanoma at the age of 86 years. He outlived both Senning and

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Elmqvist. In 1994, Siemens-Elema sold its pacemaker line to St. Jude Medical, which became a part of Abbott Laboratories in 2017 (Fig. 1).

The first American implant

Even though the first implant was successful, albeit not without electrical complications, this did not quickly pave the way for widespread use. The first implantation of a cardiac pacemaker in the United States took place at Millard Fillmore Hospital in Buffalo, New York, in 1960.

While working in his laboratory at The University of Buffalo, Wilson Greatbatch made a mistake which led to the invention of an electrical circuit that would allow for delivering an electrical impulse at a certain rate per minute. By placing a large resistor into an oscillator circuit, he learned that this circuit would produce an electrical output at a regular rate. He then powered this circuit with mercury oxide batteries and encased the entire unit in silicone. He went on to produce about 50 of these devices in his garage [2].

On June 6, 1960, Dr. William Chardack implanted Greatbatch's pacemaker into a 77-year-old man suffering from complete heart block. This procedure extended the life of this patient by 18 months and heralded the beginning of many more implants to come. Since that first implant in 1960, the procedure is much more commonplace with more than 750,000 permanent pacemakers implanted annually in the United States alone [3].

The first veterinary implant

Seven years later in 1967, Buchanan et al. [4] report the first implant of a permanent cardiac pacemaker in a 10-year-old male Basenji dog. The dog had been suffering from numerous bouts of heart

Fig. 1 Depiction of the progression of pacemakers from the 1950's (left) to today (right). Longevity ranged from a few months on early devices to over ten years for current devices. (Courtesy Abbott/St. Jude Medical).

failure brought on by complete heart block. This was an epicardial system, requiring a thoracotomy for the fixed-rate system that was implanted.

The first reported implant of transvenous cardiac pacemaker in a dog was reported by Musselman et al., [5] a decade later in 1976. The transvenous approach is considered a safer and less traumatic approach than the epicardial implant that generally requires an abdominal approach and incision of the diaphragm for access to the left ventricular epicardial surface.

Pacemaker system components

The first pacemakers were barely more than a battery and a timer. They did not have sensing capabilities; the pacing rate was fixed, and there was no communication with device for programming or diagnostic information retrieval. The first pacemaker invented by Greatbatch included only eight components. The surgical procedure was much more involved than today's transvenous approach. A thoracotomy was required, and an epicardial lead needed to be sewn onto the surface of the heart. Today's pacemakers have enormous programmability, contain thousands of components, and are capable of complete wireless communication, allowing physicians to monitor devices when the patients are at home without any patient interaction to initiate the transmission. Progress has been made from the first simple pacemaker that delivered a pacing pulse to the heart, regardless of intrinsic activity to today's highly programmable devices, allowing the synchronization of the left and right ventricles while being triggered from an intrinsic atrial impulse.

The components required for the pacemaker to operate include a single or several pacing leads and the pacemaker generator (including battery, output, and sensing circuits).

Pacing leads

Pacemaker leads are thin wires used to connect the myocardium to the pacing output circuitry. The diameter of the typical pacemaker lead ranges from approximately 5 French to 8 French, a circumference of about 5.24—8.34 mm. Typical leads are either unipolar (using the device itself as the anode) or bipolar, which uses an anode and cathode on the lead wire itself.

The leads wires are affixed to the myocardium by different methods: active and passive fixation for endocardial placement and screwed on or

sutured directly on the myocardium for epicardial placement (Fig. 2).

Pacing thresholds have been shown to be lower in the acute setting, and chronic pacing thresholds typically rise over time. Most leads in use today also incorporate a steroid, typically dexamethasone, at their tip which elutes over time [6]. The use of steroids helps stabilize the pacing thresholds for a long term.

The conductor wires are typically MP-35N (SPS Technologies, Cleveland), an alloy of nickel, cobalt, chromium, and molybdenum. This material provides for high strength and resists corrosion. These conductor wires are wound either in a coradial or a coaxial configuration, which allows for greater flex caused by movement of the myocardium during contraction and relaxation of the heart. Each strand of the conductor wires is also individually coated with ethylene tetrafluoroethylene for insulation [7]. These individual strands are then encased in an outer insulation of either silicone, polyurethane [8], or a copolymer [9] type of insulation.

The lead wires have been referred to as the weakest link in the pacemaker system. At a heart rate of 80 beats per minute, these lead wires are flexed more than 100,000 times per day. It is the flexing that can eventually lead to fractures of the conductor cables, leading to lead failure. In addition, when two or more leads are implanted, interaction between the leads can cause abrasion of the insulation material, exposing the conductors, and cause lead failure. This generally appears with a low lead impedance measurement (short circuit) during pacemaker interrogation.

Energy from the pacemaker is delivered to the electrode tip that is in contact with the myocardium. Electrode tips are generally porous or helical and consist of various materials such as

- Platinum iridium
- Elgiloy — cobalt, iron, chromium, molybdenum, nickel, and manganese
- Platinized platinum — platinum coated with platinized titanium
- Vitreous or pyrolytic carbon
- Titanium
- Titanium oxide
- Iridium oxide—coated titanium
- Titanium nitride

Recently, a leadless pacemaker system has been developed. Medtronic has a market-released device, and Abbott (formerly St. Jude Medical) has a device in clinical trials awaiting approval by the Food and Drug Administration (FDA) (Fig. 3). The device itself is delivered via a catheter-mounted system and attached to the right ventricular myocardium. The devices are retrievable and are expected to last up to 8—10 years. The pacing electronics is contained within the tube-shaped device. Pacing occurs between the fixation mechanism and the device itself. These devices are currently indicated for pacing in the ventricle only. They do not have the capability of atrioventricular synchronous pacing [10].

Pacemaker generator

The design of pacemakers today incorporates a sensing circuit, a logic circuit, and an output circuit, with the ability to communicate with a pacemaker programmer and a remote-monitoring device (Fig. 4). Signals come into the pulse generator's
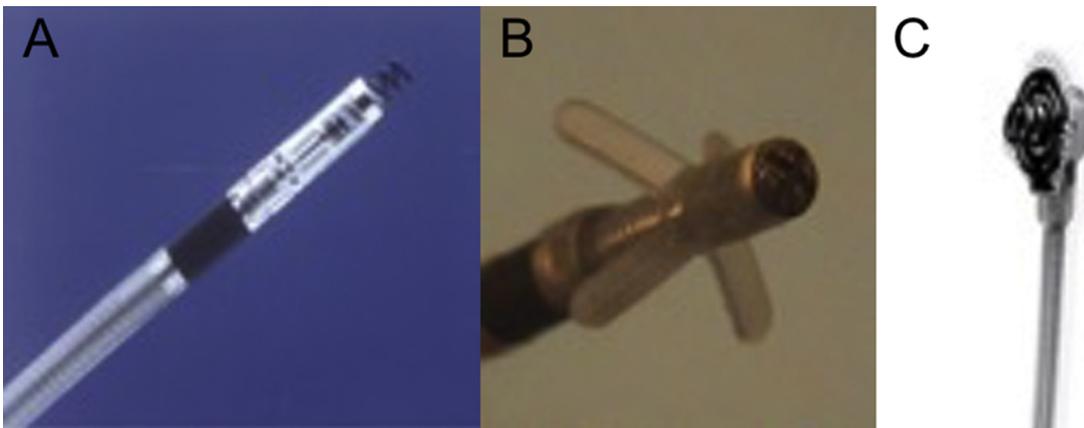

Fig. 2   The image the left shows an active fixation lead which uses a small helix which is screwed into the myocardium. The image in the center shows a passive fixation lead in which small fixation tines get tangled in the trabeculations of the heart muscle, holding it in place until endothelization takes place. These are for endocardial placement of leads. The image on the right displays a screw on lead used for epicardial placement.

Fig. 3 Leadless pacemakers developed by Medtronic (left) and Abbott Labs (right). (Medtronic®, Abbott®).

circuitry from the lead; the first area these signals enter is the sense amplifier. Like the name suggests, it magnifies or amplifies the small electrical signals into something the pulse generator can work with. The signal goes next to the logic circuit. This is the 'brains' of the pacemaker. Here, the pacemaker determines whether it ought to deliver an output pulse or inhibit an output pulse. If a pacemaker output pulse is needed, the output circuit follows through and paces the heart.

The communication circuit allows the various circuits to work together. Within this logic circuit, the device can determine if signals received in the sensing circuit are physiological signals that can be used to either trigger the device to pace or inhibit the output in the face of an intrinsic cardiac event. Extraneous environmental signals and noise from compromised components (leads and setscrew) can be analyzed within the logic circuit and force changing of the pacing mode from demand to asynchronous pacing. This allows for committed backup pacing and thus avoids periods of asystole in pacemaker-dependent patients (Fig. 5). This could be especially important for those pet owners who use an underground fence. Noise created by activation of the electric signal from the fence could then be rejected by the pacemaker circuitry yet
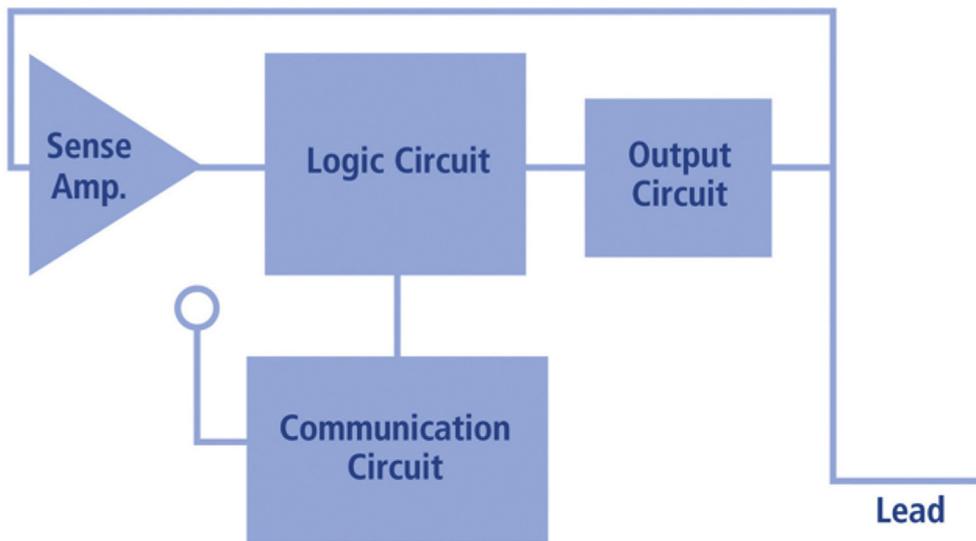

Fig. 4 Schematic of the pacemaker circuitry.

Fig. 5 Pacing spikes delivered at the pacing interval fail to capture (arrows) the myocardium.

still allow for backup pacing and avoid a no-output situation. Systems today have the capability to discern non-physiologic signals and switch their mode of inhibited pacing to committed pacing. This is important in the pacemaker-dependent patients. Older generations of devices lacked this ability and would inhibit pacing when these signals were sensed. Signals from airport security systems and microwave ovens are not an issue with today's devices. This noise reversion mode is programmable in all currently available devices.

The output circuit consists of two basic factors: voltage and duration. The former is the programmed voltage, and the latter is the pulse width of the delivered pacing pulse. Today's devices are constant-voltage. To get a true constant voltage, the device would need to be able to alter current to adjust for resistance. So, in reality, we achieve the above based on impedance/resistance. The largest component of the output circuit is the battery. The battery is typically about $2.75-3.0$ V, depending on the manufacturer. Various outputs are achieved by using a step-up or a step-down transformer to arrive at the desired voltage. The size and design of the battery has gone through numerous changes over the years, becoming much smaller and far more efficient.

The programmer allows the operator to make changes to various functions of the device such as rate, output energy, and sensing, among others. This communication with the implanted device allows one to review any stored data collected by the device such as rate histograms, lead and battery performance, and alert conditions that may have been stored to allow troubleshooting of issues.

Battery evolution

The design of the pacemaker battery over the years first began with mercury zinc power cells in the 1960s. Unfortunately, this type of battery emitted a gas that interfered with hermetic seal of pulse generator causing these devices to have a short lifespan from several months to a few years. The nuclear age arrived in the early 1970s allowing for the potential lifespan of the battery to exceed 40 years. Proper disposal of the radioactive material after patient death was a major issue, and the nuclear battery concept was scrapped. Introduction of nickel cadmium batteries in the 1980s allowed for rechargeable batteries to be introduced. These cells did not prove to be very reliable, and the use of nickel cadmium batteries was discontinued. Current generations of pacemaker batteries use lithium iodide. These

batteries are lightweight, have predictable discharge characteristics, are smaller which has allowed for smaller pacemakers to be designed, and can last ten or more years.

The battery longevity can roughly be calculated as follows (however, the pacemaker does this automatically):

Usable battery capacity = 1.2 amp hours.

Pacer current drain = 20 microamps.

Hours in 1 year = 8760.

[1,200,000 microamps (hrs)]/20 microamps = 60,000 h.

60,000 h/8760 = 6.849 years.

The energy delivered plays a significant role in longevity. The energy equation is

$(E = V^2 \times t)/R$.

where V = output voltage, t = pulse width of pacing pulse, and R = resistance. The energy (E) delivered is expressed in micro Joules. According to this equation, if we double the voltage, we quadruple the energy consumed. In the example given in the following, if the pacemaker of the dog in Case A is programmed to 2.5 V, the device longevity will greatly exceed that of the pacemaker of the dog in Case B in which the output was programmed to 5 V, given all other parameters are the same. The longevity calculated by the device is generally accurate but is based on certain assumptions such as consistent impedance over time and 100% pacing. A range of longevity is generally displayed, such as $2.4-4.8$ years for example. The former indicates that the device would last about 2.4 years if paced 100% of the time at the current settings and measured impedance.

Case A	Case B
2.5 V, 0.5 ms, 750 Ω	5.0 V, 0.5 ms, 750 Ω
$E = (2.5)^2/750 \times 0.5$	$E = (5)^2/750 \times 0.5$
$E = 6.25/750 \times 0.5$	$E = 25/750 \times 0.5$
$E = 4.2$ μJ	$E = 16.6$ μJ

Another factor affecting longevity is rate response. For those patients in need of varying rates to accommodate for increased activity in the face of chronotropic incompetence, devices can raise the rate based on several factors. Some devices use a piezoelectric crystal to detect movement; the greater the amount of deflection of the crystal, the greater the rate produced by the device. There are also sensors that use minute ventilation to increase the rate. This rate response can affect longevity as the device is pacing more and therefore increase the battery drain. The

battery replacement indicator displayed on the device interrogation report is usually expressed as elective replacement indicator or elective replacement time. When this message displays, there is generally 3—6 months of battery energy remaining. Some devices begin to turn off certain functions (rate response and automatic threshold testing) when this message is displayed to further conserve energy. When End of Service or End of Life is displayed, the battery is dangerously low in usable energy, and proper function of the device cannot be ensured.

The longevity equation is important in veterinary cardiology when the reuse of explanted pacemakers is considered. Devices explanted from humans for reuse need to be thoroughly tested to determine remaining battery longevity. Explanted devices should be hooked to a resistor load of 500 Ω and interrogated to obtain measured data including battery voltage, battery resistance, and remaining longevity. A pacemaker simulator is typically used to assess proper battery characteristics and functionality of the device. If adequate, the devices can then be cleaned and sterilized for use in the veterinary patient. Information regarding sterilization techniques can be found at https://www.mddionline.com/fda-pick-sterilization-method-not-just-any-method.

In a study by Selvaraj [15], 887 patients were followed up for device implants ranging from pacemakers to defibrillators to cardiac resynchronization devices. Of these implants, 260 of the devices were reused [11]. There were no postoperative infections or device malfunctions noted in the group with reused devices, showing that careful resterilization and screening of proper device function of these devices were safe. The reuse of pacemakers is especially attractive in veterinary cardiology as the new devices from manufacturers can prove to be cost prohibitive for client-owned pets in need of a device. The use of dual-chamber pacemaker can be used when a single-chamber device is indicated. It is important to use a pin plug to occupy the atrial port in this situation. If the port is not plugged, body fluids can enter the header port and cause damage to the internal circuitry. After the useable life of the pulse generator has been reached, the device should be returned to the manufacturer as companies break down the devices to analyze the components. This is an FDA requirement in the human population. For those devices not returned to the manufacturer, disposal of the device should follow the usual biohazard waste procedures of the institution. Device explant should be performed especially in the setting of cremation as the intense heat can cause explosion of the device.

Magnetic resonance imaging compatibility

Devices are typically labeled as Magnetic resonance imaging (MRI) compatible, conditional, or non-MRI conditional. Devices labeled as compatible are safe for whole-body scans without regard to the patient positioning. Devices labeled as conditional are safe for scans with proper programming and generally not for scanning the implant area. Typically, the recommendation is for scans at 1.5 T or less. Historically, non-MRI conditional devices would be contraindicated for scanning. However, several studies have shown that patients with these devices can safely have an MRI (non-thoracic) at 1.5 T [12]. Careful programming and post-MRI interrogations are essential in this situation. For patients with an intrinsic heart rate of >40 beats per minute, the device can be programmed to ignore the magnet response and programmed to a non-pacing mode. For those patients who do not have a reliable underlying rhythm (pacer dependent), the device should be programmed to ignore the magnet response and programmed to a non-sensing mode (committed pacing) [13]. Magnetic resonance imaging can cause elevated levels of interference, and a non-sensing mode will allow the device to continue to pace without sensing the extraneous noise from the imaging system. It is important to have the device interrogated before and after the procedure. All programmed parameters should be restored to those existing before the procedure. Newer models of compatible devices have a specific mode that can be programmed before the procedure and then returned to preprocedure values.

Component failure

Pacemakers have come under a recall from time to time for assorted reasons. Recalls and advisories information can be found at the FDA website: https://www.fda.gov/MedicalDevices/Safety/ListofRecalls/default.htm.

From leads to batteries to internal components, all have been found to be the cause of an advisory. Pacemaker leads are designed to flex and yet still hold up over the years. Implanted leads will flex over 400,000,000 times over a typical 10-year battery lifespan. Leads are not infallible; conductor wires can break, and insulation can degrade, exposing the conductors. The operator

can also be at fault for some of the failures. Tying down the suture sleeve on the lead too tight can cause a crush phenomenon, damaging the conductor wires and causing the lead to fail. Typical sutures used for securing the lead wires to the fascia are non-absorbable 2-0 Ticron or 2-0 Silk. Damaging the insulation at implant can expose the conductor wire, causing a short circuit. This results in very low impedance values and can cause rapid battery depletion because of a damaged lead [14]. In this example, a lead issue can cause a device issue. Physical damage to the setscrews that attach the leads to the header of the device can also cause failures. The stainless-steel setscrews are protected by a silicone septum that insulates the electrically active setscrew from the body fluids. If this septum is damaged, body fluids can enter the header and cause a short circuit leading to battery depletion or loss of output, putting the patient in danger. Overtightening the setscrew can damage the receptacle for the hex wrench. This may make a generator changeout very difficult. Conversely, not tightening the setscrew enough will cause high impedance (open circuit) and loss of pacing [15]. Internal shorting of components can cause rapid battery depletion and loss of function of the device, leading to no output. Other causes of sudden loss of output can come from electrocautery. In this case, energy from the cautery device can be coupled in the device, flooding the sense amplifiers or directly causing a component failure which leads to a no-output condition [16]. External defibrillation is generally safe, and the pads (or paddles) should be placed as far away from the device as possible but still give a good vector for defibrillation. Energy from the external defibrillator can be coupled into the device, causing major component failure. Sometimes, this results in a reset (or backup) mode within the pacemaker which can generally be reset with the programmer, and other times, it can be a terminal condition resulting in no output.

Pacemaker parameters

Managing electrical activity in the heart during periods of bradyarrhythmia requires the ability to deliver enough energy to depolarize (capture) the heart and to correctly sense intrinsic cardiac activity. These functions can be affected by a multitude of factors:

- Setting of output parameters (pulse amplitude and pulse width)
- Sensitivity parameter settings

- Impedance (component and system including patient interface)

Electrical concepts

When addressing the concept of pacing, we must consider the energy equation (Ohm's law). The basic equation is $V = IR$. where

V = the voltage or the force which drives the current
I = the current, described as the flow of electrons past a certain point measured in milliamperes
R = the resistance (impedance) or opposition to flow of electrical current expressed in Ohms (Ω).

Changes in any one of these factors cause a change in the entire picture. Implantable pacemakers are constant-voltage devices. If any part of the $V = IR$ equation changes, the device will adjust to keep the voltage the same. Impedance and current have an inverse relationship. Resistance and impedance are both expressed in Ohms; however, the two differ in that resistance is the opposition to current flow in a direct current circuit, whereas impedance describes the resistance to current flow in an alternating current circuit. The terms are often used interchangeably when describing current flow in pacing circuits. For the purposes of this article, the term impedance will be used. If impedance rises, the current falls, and vice versa. Increased impedance can lower current drain and prolong battery life. High-impedance pacing leads are designed to do just this. This is accomplished by adding a high-impedance metal, such as titanium nitride, to the lead and thin Multifilar wires and connecting the pin to the tip of the lead and the anode. A garden hose analogy can serve as a good illustration of the 'voltage, current, and impedance.' The water is voltage, and by restricting the end of the hose with a nozzle, impedance to flow increases and decreases the flow of water (current). Similarly, by removing the nozzle, a decrease or even elimination of impedance occurs, but when that happens, the flow of water will increase. Low impedance can lead to rapid battery depletion.

Impedance values typically range from 300 to 1200 Ω. Leads can vary in impedance for several reasons. The type of lead fixation can play a role in impedance, with active-fixation leads having a higher impedance than passive-fixation leads. The interface between the myocardium and the lead

tip can also contribute to impedance values. For example, if the lead does not contact the myocardium securely, it can result in a higher impedance measurement. Similarly, an epicardial lead placement can also result in a larger impedance value than those found in an endocardial placement. Pacing polarity can cause a change in impedance as well. When using unipolar pacing, which uses the device itself (the generator, also called 'can') as the anode, the impedance is lower than with a bipolar pacing configuration. This is because the surface area of the anode in the unipolar configuration is far greater than the surface area of the anodal ring of the bipolar pacing lead. A low impedance measurement ($<$200 Ω) indicates a short circuit, usually from an insulation breach. An unusually high impedance measurement ($>$2,000 Ω) indicates an open circuit, typically caused by a conductor fracture.

Capture threshold

The capture threshold, sometimes referred to as the pacing or stimulation threshold, is the minimum amount of energy delivered by the device to consistently depolarize (capture) the myocardium. The capture threshold is not constant and can change over time with changes in resistance, medications, or underlying pathophysiology [17]. Thresholds are typically low at the initial implant and generally rise over time as the leads become chronic [18]. Typically, the output voltage to the lead(s) can be decreased to twice the pacing threshold 90 days after implant. At this point, the leads are generally considered to be chronic. Acceptable pacing thresholds for canines are $<$1.5 V for an acute implant and $<$2.5 V for a chronically implanted system. A generally accepted tenet is that the device should be programmed to twice the pacing threshold. For example, for a capture threshold of 1.5 V, the pacing output would typically be programmed to 3.0 V. This is referred to as the capture safety margin or threshold safety margin. Loss of capture occurs when the output pulse from the pacemaker is of insufficient amplitude to capture the heart (loss of capture) (Fig. 6).

Once capture thresholds are determined, the clinician is now faced with the challenge of programming the settings for the two output parameters (pulse amplitude and pulse width). Doing this requires managing two seemingly contradictory requirements: The output should be large enough to assure pacing, even when the threshold changes (and the assumption must be that the measured
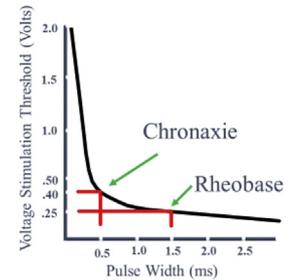

Strength Duration Curve

Fig. 6   The minimum energy consumption at threshold should be found at the chronaxie pulse width. How can strength duration curves be used clinically? For chronic leads, a determination of the chronaxie point establishes the most efficient pulse width setting.

capture threshold is the 'low point') but the output should not be unnecessarily high and consume too much battery energy. For example, if the ventricular threshold was 1.0 V at 0.4 ms, capture would be reliably maintained if the device was programmed to 7 V at 0.4 ms, but that would drain the battery prematurely and reduce the service life of the device.

The strength—duration curve is useful in determining proper programming of the device (Fig. 3). The black line on the graph shows the capture threshold. At higher voltages, very little pulse width is required. As voltage decreases, a longer pulse duration is required to capture the heart. However, at a given point, the extension of pulse width makes no difference (this is where the line flattens out). At this point, no further decrease to voltage can be made without sacrificing capture. This point is called the rheobase. Although each patient has an individual strength-duration curve (or pulse amplitude-pulse width curve), they all tend to have this same characteristic shape. The bend in curve is called the chronaxie point or simply chronaxie. It is defined as twice the rheobase. The strength-duration curve is particularly useful because the chronaxie shows the combination of pulse width and pulse amplitude (voltage and duration) that uses the least energy to reliably capture the heart.

Sensing

Sensing refers to the capability of the device to detect intrinsic cardiac activity. Pacemakers use an intracardiac electrogram allowing the device to 'see' the amplitude of the cardiac signal from inside the heart. Unipolar sensing refers to the use of the pacemaker itself and the tip of the pacing

lead as the antenna for sensing. This results in a larger signal and is more likely to sense myopotentials and could possibly cause oversensing. The possibility of pacemaker inhibition to the sensing of extraneous signals is more likely in a unipolar sensing circuit. A bipolar sensing circuit uses a much smaller antenna because the anode and cathode are incorporated into pacing lead and is less likely to oversense extraneous signals. The height, or amplitude, of the smaller signal is then amplified and measured by the device.

Sensitivity in pacemakers are programmed in such a way as to sense the intrinsic signal reliably, but not so sensitive as to detect extraneous noise from myopotentials or environmental sources. Signals being oversensed could cause the pacemaker to inhibit its output signal and not deliver a pacing pulse when it may be needed. Some types of oversensing can originate from environmental factors such as myopotentials or electrical interference from outside sources. This could prove to be detrimental to patients who are pacemaker dependent as these extraneous signals could cause the pacemaker to inhibit its output. On the other hand, a pacemaker not programmed sensitive enough to detect the intrinsic signal can lead to inappropriate pacing. An acceptable intrinsic measured atrial signal is >1 mV, whereas an acceptable measured ventricular signal would be >10 mV for the typical canine implant. The typical safety margin is 2:1. For example, an atrial signal of 1 mV would require the atrial sensitivity to be programmed to .5 mV. In contrast to the pacing threshold which is two times the capture thresholds, the sensing circuit is just the opposite. A smaller value is programmed to make the device more sensitive. In this case, the sensitivity should be programmed to at least ½ the measured intrinsic signal.

Pacing modes

Table 1 gives a cross-reference for pacing modes of operation. Position I indicates the chamber (or chambers) paced. If only one chamber is paced, the designation would be either A or V, and if both chambers are paced, the designation would be D. O indicates that no chamber is paced. A manufacturer code of 'S' in this position indicates single chamber is paced (atrial or ventricular).

Position II indicates the chamber (or chambers) sensed. If only one chamber is sensed, the designation would be A or V, and if both chambers are sensed, the designation would be D. O indicates that no chamber is sensed. A manufacturer code of 'S' in this position indicates that single chamber is sensed (atrial or ventricular). Position III indicates the mode of response to sensing. Triggered means the device will pace when an intrinsic signal is sensed. Inhibit means the pacing output will be withheld when an intrinsic signal is sensed. D indicates that the device can both trigger and inhibit in response to intrinsic signals. O indicates that there is no response (neither triggered nor inhibited) to sensed intrinsic signals. Essentially, this is committed pacing. Position IV indicates other programmable functions of the device. Rate modulation allows the device to change its pacing rate based on physiologic need such as increased activity. Communicating indicates that the device can transmit and receive information regarding diagnostics or programming. Multiprogrammable means that greater than three parameters are programmable. All pacemakers today are multiprogrammable. P indicates simple programmable, limiting the functions to three or less. It would be rare to see any of these types of devices in operation today. Position V is reserved for antitachycardia functions. This would indicate that the device can rapidly pace a tachycardic rhythm, shock a tachycardic rhythm, or both. Most cardiac defibrillators utilize the 'D' designation in this position.

Using the table previously given, a device labeled VDDRO would mean the device paces in the ventricle only and senses in both the atrium and the ventricle. It can either trigger or inhibit in the presence of a sensed intrinsic ventricular signal, has rate modulation capabilities, and does not have antitachycardia capabilities. Pacing in the ventricle is triggered from sensed atrial events.

Table 1 The Heart Rhythm Society (HRS) and British Pacing and Electrophysiology Group (BPEG) guide to pacemaker modes of operation.

I	II	III	IV	V
Chamber(s) Paced	Chamber(s) Sensed	Mode(s) of Response	Programmable Functions	Antitachycardia Functions
V=Ventricle	V=Ventricle	T=Triggered	R=Rate Modulated	O=None
A=Atrium	A=Atrium	I=Inhibited	C=Communicating	P=Paced
D=Dual (A&V)	D=Dual (A&V)	D=Dual Triggered/ Inhibited	M=Multiprogrammable	S=Shocks
O=None (S=Single)	O=None (S=Single)	O=None	P=Simple Programmable O=None	D=Dual (P&S)

This can be accomplished via a special lead (sometimes referred to as a single-pass lead) with floating electrodes that sense atrial signals and trigger the ventricular output based on the atrial signals. This technique of using a single ventricular pacing lead to accomplish atrioventricular synchrony is used frequently in both veterinary and human implants [19].

Conclusion

In the six decades since the first cardiac pacemaker was implanted, great strides have been made to make the device more efficient, easier to implant, highly programmable, and simpler to monitor. Wireless capabilities of the devices can allow for remote monitoring of the device when patients are at home, saving time and resources. These improvements have allowed a greater number of patients, both human and veterinary, to benefit from this technology.

Manufacturer/distributor websites

Abbott Labs (formerly St. Jude Medical): www.abbott.com/abbott-stjudemedical-en.htm.
Medtronic: www.medtronic.com/us-en/index.html.
Boston Scientific: https://bostonscientific.com/
Biotronik: https://www.biotronik.com/en-us.
Livanova (Sorin, Ela): www.livanova.sorin.com/customer-service/north-america.
Dextronix: https://www.dextronix.com/
Infiniti Medical: http://infinitimedical.com/products/lungavita-veterinary-cardiac-pacemaker-and-programmer/

Conflicts of Interest Statement

The authors do not have any conflicts of interest to disclose.

Acknowledgments

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