CHAPTER 1

History of cardiac pacing and defibrillation in the young

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The earliest years of cardiac pacing predate the birth of many current pediatric cardiac electrophysiologists. An old saying states that "failure to understand history dooms one to repeat it." In contrast, understanding this history of successful collaboration between pioneering physicians and engineering partners allows us to marvel at the developments that were to follow rapidly over the next 50 years, and potentially repeat this formula in years to come.

Benjamin Franklin harnessed electricity from lightning using a kite in 1752. An early "medical" use of electricity was not to augment life but to document the end of it with patients receiving an electrical shock to prove they were dead. In 1774, electrical energy was applied to resuscitate a child using a transthoracic approach.[1] As early as 1899, the *British Medical Journal* published a report of experiments demonstrating that application of electrical impulses to the human heart would lead to ventricular contractions.[2] In 1926, Dr. Mark C. Lidwell and physicist Edgar H. Booth of Sydney developed a device with pacing rates of 80–120 bpm and outputs varying from 1.5 to 120 V.[3] This "pacer" was described as being a portable device

"plugged into a lighting point." One pole was connected to a pad soaked in strong salt solution and applied to the skin and the other, "a needle insulated except at its point, was plunged into the appropriate cardiac chamber." In 1928, this apparatus was used to revive a stillborn infant whose heart continued to beat after 10 minutes of stimulation.[4]

During the 1930s, Dr. Albert Hyman noted that the success of intracardiac delivery of medications for cardiac arrest was likely independent of the medication but was instead related to the needle stick leading to alteration in electrical potentials and myocardial contraction. Knowing that multiple needle sticks would be impractical and dangerous, he developed a generator to deliver electrical impulses via needle electrodes.[5]

Following World War II there was a significant interest in pacemakers generated by investigations in the use of general hypothermia for cardiac surgery. Cardiac arrest was noted during hypothermia and adequate heart rate was required to maintain adequate hemodynamics during rewarming. John A. Hopps, an engineer at the National Research Council of Canada developed a pacemaker that produced impulses at a desired rate

through an electrode placed in the area of the sinus node.[6]

In 1952 Dr. Paul M. Zoll used an external pacemaker coupled with transcutaneous needle electrodes to rescue a patient suffering from Stokes-Adams attacks following a myocardial infarction.[7,8] The patient continued to experience ventricular asystole despite being administered 34 intracardiac injections of adrenaline over a 4-hour period. Dr. Zoll applied "external electrical stimulation" and successfully paced this patient's heart over the next 25 minutes.[8] The patient developed cardiac tamponade secondary to perforation of a cardiac vein during the intracardiac injections. Dr. Zoll then successfully paced a 65-year-old man with episodes of ventricular standstill for 5 days by external electrical stimulation at which time he developed an idioventricular rhythm at 44 bpm and was discharged.[9]

In the mid-1950s, open heart surgery was becoming a reality. Although for the first time in history, intracardiac palliation of structural heart disease was possible, the complication of surgical heart block was a significant morbidity. Dr. W. Lillehei, Dr. W. Weirich, and others at the University of Minnesota demonstrated that pacing could be performed by connecting a pulse generator to a wire electrode attached directly to the heart of a dog.[10,11] In January 1957, Lillehei used this pacing system in the first human patient, a child with post-operative heart block following repair of a ventricular septal defect. The pacer was programmed to a pulse width of 2 ms and a voltage ranging from 1.5 to 4.5 V (Figure 1.1).[12]

The generators used by both Zoll and Lillehei were devices which transformed alternating current into direct current to pace the heart. In 1957, following a power failure in Minneapolis in which patients could not be paced, Dr. Lillehei enlisted the help of Earl Bakken and Medtronic for battery backup for AC pacemakers. Silicon transistors had become commercially available in 1956 leading to the potential for development of smaller and more practical pacemakers. The original transistorized, zinc oxide battery-powered external pacemaker was developed by Mr. Bakken in 1957; the device was smaller and thus applicable for pediatric patients.[13,14] This, the first wearable external pacemaker, was housed in a small plastic

Figure 1.1 *Patient pushing pacemaker cart (1958). (Source: Reproduced with permission of Medtronic, Inc.)*
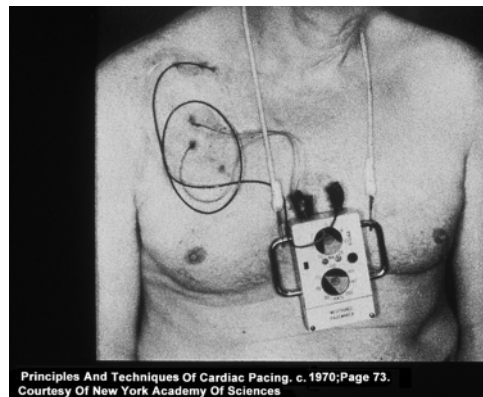

Figure 1.2 *Wearable pulse generator (1958). (Source: Reproduced with permission of Medtronic, Inc.)*

box, with controls to allow adjustment of pacing rate and voltage (Figure 1.2).

Although novel and potentially lifesaving, the advances described here were not a long term solution in that there was a significant risk of infection and external pacing was uncomfortable and impractical. There was a definite need for implantable pacing systems. Ake Senning, a Swedish surgeon, in collaboration with engineer Rune Elmqvist, developed a permanent implantable pulse generator with the first clinical implantation in 1958.[15] This device failed after three hours. A second device was implanted and lasted 2 days. The patient, Arne Larsson, went on to receive 26 different pacemakers until his death in 2001 at the age of 86 (Figure 1.3).[16]
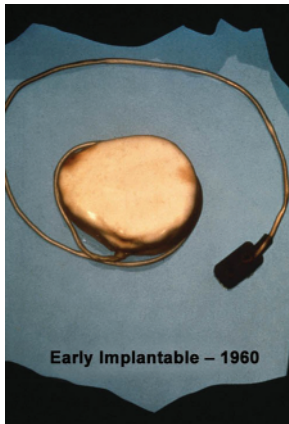
Figure 1.3 *History – First "permanent" implantable pacemaker and bipolar Hunter–Roth lead (1958). (Source: Reproduced with permission of Medtronic, Inc.)*

During that same year, Seymour Furman introduced temporary transvenous pacing using the recently described Seldinger technique.[17] In 1962 Ekestrom, Johannson, and Lagergren reported the first non-thoracotomy pacemaker implantation by introducing the electrode transvenously into the right ventricle.[18]

By the end of 1960 virtually all pacemakers used mercury-zinc cells as the power supply, but battery life expectancy was generally less than 2 years on average. A greater problem was that because the batteries emitted hydrogen gas, the pulse generator could not be sealed to protect from contamination with body fluids.[19] Dissatisfaction with this power source generated interest in alternatives that included, but were not limited to, bioenergy sources (using piezoelectric transducers that generated electricity based on the expansion and contraction of the abdominal aorta, or motion of the diaphragm), nuclear generators, and, by the mid-1970s, lithium batteries.[20] There was significant interest in the use of nuclear powered pacers because they offered a remarkable lifespan (10–20 years) and reliability. A number of drawbacks related to radiation exposure in case of a capsule leak and disposal hindered their acceptance.

Lithium-iodide power sources persist as the battery of choice today. Voltage output of the lithium-iodine cell showed gradual decline rather than the abrupt drop associated with the mercury zinc during battery depletion. This new battery

generated no gas byproduct allowing the entire pulse generator to be hermetically sealed in a titanium case, which was initially accomplished in 1969 by Telectronics and then by Cardiac Pacemakers, Inc., (Minneapolis, MN), in 1972. Battery life was significantly increased to greater than 5 years on average.

Leads

In the early 1960s it became routine practice to manage patients with temporary transvenous leads and an external pulse generator to relieve congestive heart failure. These served as a bridge to a thoracotomy for placement of a permanent pacemaker and lead system. Permanent transvenous pacing, which first appeared in the early 1960s, gained widespread acceptance by the end of the decade.[21,22] Initial leads were unipolar in design, but gradually gave way to a bipolar preference. Coaxial leads allowed for smaller lead diameter and greater durability. Smaller surface electrodes were designed to reduce energy consumption. Greater surface areas were achieved allowing improved lead function. Steroid-eluting leads were designed as a mechanism to reduce fibrosis at the epicardium-electrode interface, thus avoiding chronic rise in stimulation thresholds. Lead fixation, using passive or active mechanisms, were designed to prevent the previously high incidence of lead dislodgement. Silicone insulation gradually gave way to a preference for polyurethane. These newer leads had a generally smaller diameter than previous silicone leads, which facilitated the introduction of the implantation of two leads through a single vein, associated with the implementation of dual chamber pacing.

Pacing modes

The first implanted pacemakers were fixed rate ventricular systems, which competed with intrinsic ventricular activation. Unfortunately, the theoretical risk of inadvertent induction of ventricular fibrillation was in fact documented electrocardiographically.[23] Additionally, studies determined that fixed rate asynchronous pacing at times had an adverse hemodynamic impact on patients with myocardial dysfunction. Thus, the

impetus for development of a demand pacemaker which could sense intrinsic ventricular activity was heralded. Virtually simultaneously, two companies debuted demand ventricular pacemakers. In 1966 the Medtronic system functioned in a true demand mode, with inhibition of ventricular pacing during sensed intrinsic ventricular activity.[24] The Cordis "standby" pacemaker functioned in the ventricular triggered mode, such that a sensed R-wave triggered the pacer stimulus with no AV delay so that it fell within the refractory period of the intrinsic QRS complex.[25] These modes were, respectively, termed VVI and VVT, both non-competitive modes.

The first pacemakers to permit atrial synchronization with the ventricle depended upon new sensing technology to detect intrinsic atrial activity,[26] and the triggering of a paced ventricular response after a programmed AV interval. These devices were bulky because of the complexity of the circuitry, and also demonstrated a significant reduction in battery life. Problems with erratic sensing of the intrinsic atrial activity and abrupt drops in pacing rates that occurred when upper rate limits were reached also limited the acceptance of these early dual chamber systems.

In the early 1980s a third generation of dual chamber pacemakers was introduced. These generators had long-lived lithium batteries and generally incorporated new dual endocardial leads. Pacemaker systems were able to both sense and pace in both the atrium and ventricle allowing physiologic rates and AV synchrony. The development of leads which could be used for atrial stimulation, as well as atrial sensing, enhanced the functionality of these early dual chamber systems.

Rate adaptive pacing, for patients with chronotropic incompetence, permitted rate responsive pacing that augmented heart rate response when intrinsic sinus node function was inadequate.[27] A more recent breakthrough mode was antitachycardia pacing, applicable especially for postoperative congenital heart disease patients with recurrent medically-refractory intraatrial reentry tachycardias.[28]

Non-invasive programmability

Seymour Furman and associates reported in 1969 the first techniques for routine transtelephonic monitoring of pacemaker function.[29,30] Subsequent advances included the ability of the system to estimate battery longevity. Continuous advancement in these non-invasive technologies has finally led to the ability to provide non-invasive electrogram analysis for tachycardia detection, tachycardia termination, and antitachycardia defibrillation systems.

Multiprogrammability

By the mid-1960s, the early non-invasively programmable pacemakers had advanced to multiprogrammable units dependent upon bidirectional telemetry. In 1978, Intermedics introduced a pacemaker for whom pacing rates, pulse width, and sensitivity could be programmed; this system was a result of collaboration between engineer Robert Brownlee and physician G. Frank Tyers.[31] Dual chamber pacemakers also permitted programmability of pacing mode, in the event of recovery of intrinsic AV nodal function (allowing atrial pacing alone) or ventricular pacing only in the event of failure of the atrial lead (pacing and/or sensing capabilities).

Miniaturization

Initial external pacemaker systems required portable carts (Figure 1.1). By the early 1960s when permanent implantable systems were in place, the pulse generators were still bulky. Advanced pacemaker and software technologies allowed further miniaturization, but often at the expense of battery life. Smaller generators had a unique implant role for the smallest of neonates and pediatric patients, but required frequent generator changes due to battery depletion. Further decrease in lead size allowed implantation of multiple leads within a single vein, even in the smallest patients, but electrodes were still relatively large. Even these small lead systems were associated with a high incidence of venous obstruction/occlusion.

Pacemaker codes

The Inter-Society Commission for Heart Disease Resources (ICHD)[32] proposed a three-position

"conversational" pacemaker code in 1974 to distinguish pacemakers according to three fundamental attributes:

Position 1. Chamber or chambers paced:

   V – ventricle paced

   A – atrium paced

   D (dual) – both atrium and ventricle paced

   O – neither atrium or ventricle paced

Position 2. Chamber or chambers in which native cardiac events were sensed:

   V – ventricle sensed

   A – atrium sensed

   D (dual) – both atrium and ventricle sensed

   O – neither atrium or ventricle sensed

Position 3. Pacemaker response to sensing a spontaneous chamber depolarization:

   T – triggered

   I – inhibited

   D (dual) – both triggered and inhibited

   O – none

Subsequent revisions paralleled development of pacemaker capabilities. The most recent revision of this original three-position code was published in 2000 incorporating a five-position code.[33] Position 4 is used only to indicate the presence (R) or absence (O) of a rate adaptive mechanism, used to compensate for patients with chronotropic incompetence. Position 5 indicates whether multi-site pacing is present in none of the cardiac chambers (O); in one or both of the atria (A) with stimulation sites in each atrium or more than one stimulation site in either atrium; in one or both of the ventricles (V), the stimulation sites in both ventricles or more than one stimulation in either ventricle; or in dual chambers (D), in one or both of the atria and in one or both of the ventricles. This most recent coding was endorsed by both the North American Society for Pacing and Electrophysiology (NASPE), (now known as the Heart Rhythm Society: HRS), and the British Pacing and Electrophysiology Group (BPEG).

Guidelines for implantation of cardiac pacemakers and antiarrhythmia devices

The first guidelines were introduced in 1984 through a joint subcommittee of the American College of Cardiology and the American Heart Association.[34] Pediatric cardiac pacing was represented by Dr. Paul Gillette. Guidelines were grouped according to the following classifications – class 1: conditions for which there is general agreement that permanent pacemakers and antitachycardia devices should be implanted; class 2: conditions for which permanent pacemakers and antitachycardia devices are frequently used but there is a divergence of opinion with respect to the necessity of their insertion; class 3: conditions for which there is general agreement that permanent pacemakers and antitachycardia devices are unnecessary. Multiple revisions have occurred, coincident with advances in technologies of these devices. The most recent guidelines were issued in 2008 through the American College of Cardiology, American Heart Association, and Heart Rhythm Society, and were developed in collaboration with the American Association for Thoracic Surgery and the Society of Thoracic Surgeons.[35] Guidelines for pediatric pacing were generated with the input of Dr. Mike Silka. In addition to the class 1, 2, and 3 indications for pacemaker implantation, guidelines were also rated according to evidence to support the guidelines. Levels of evidence A: data derived from multiple randomized clinical trials or meta-analysis; B: data derived from a single randomized trial or nonrandomized studies; and C: only consensus opinion of experts, case studies, or standard of care. These guidelines continue to dynamically evolve, but are widely regarded by consensus as detailing the appropriate use of devices in both adult and pediatric patients.

North American Society of Pacing and Electropysiology (NASPE)

Senior pacing physicians during the 1970s founded the *Journal of Pacing and Clinical Electrophysiology* (*PACE*) and organized a supporting professional society, NASPE.[36,37] NASPE arose out of a concern for the growing complexity of pacemaker systems and implantation techniques, the maintenance of quality control and good manufacturing practices by companies, and the proper post-implantation care of an ever expanding patient population. As lead technology advanced, the non-invasive transmission of intracardiac electrograms allowed increased patient diagnostic surveillance and

treatment, and paralleled the explosive development of intracardiac electrophysiologic testing in advance of cardiac ablative therapies. NASPE (HRS since 2004) currently has over 5400 cardiac pacing and electrophysiology professionals worldwide and is the international leader in science, education and advocacy for cardiac arrhythmia professionals and patients, and the primary information resource on heart rhythm disorders. Its mission is to improve the care of patients by promoting research, education, and optimal health care policies and standards.

The implantable cardioverter defibrillator (ICD)

Prior to discussing the history of implantable cardioverter defibrillators, a brief review of defibrillation is in order. At the turn of the twentieth century, Prevost and Batelli researched ventricular fibrillation in dogs describing methods to fibrillate the heart using alternating (AC) and direct (DC) electrical currents. They noted it took stronger currents to defibrillate than to fibrillate the heart.[38] In 1947, Dr. Claude Beck performed the first successful human defibrillation using internal cardiac paddles on a 14-year-old boy who developed VF during elective chest surgery.[39] The device used on this patient, made by James Rand, had silver paddles the size of large tablespoons that could be directly applied to the heart. In 1956, Paul Zoll used a more powerful unit to perform the first closed-chest defibrillation of a human.[40]

The remarkable technical advances that occurred in clinical electrophysiology and pacemaker technologies through the 1960s and 1970s established the groundwork for the development of the implantable cardioverter defibrillator (ICD). External cardiac defibrillation was proven to be an effective method for terminating potentially life-threatening cardiac rhythm disturbances, including unstable ventricular tachycardia and ventricular fibrillation. In contrast to the pioneering collaborative efforts of multiple teams of physicians and engineers responsible for pacemaker development through the 1960s and 1970s, the development of the ICD is attributed almost single-handedly to the unwavering determination of Dr. Michael Mirowski in Baltimore, and his

engineering collaborator, Dr. Morton Mower. In a 1970 publication, Mirowski and Morton described the elements of an early ICD device, which would be required to quickly diagnose and treat ventricular fibrillation using a unit small enough for subcutaneous implantation.[41] Extended battery life would be a key component given the high output demands anticipated. Ventricular fibrillation detection techniques were initially dependent upon right ventricular pressure transducers, with a drop in blood pressure in post myocardial infarction patients triggering the device.[42] This unreliable sensing method was upgraded to the use of an intracardiac electrogram feature and a complex probability density algorithm distinguishing ventricular fibrillation from sinus rhythm. Initial device design used a hybrid endocardial and epicardial lead system with a single right ventricular transvenous lead and a subcutaneous defibrillation patch in the anterior chest wall. Subsequent iterations included a shock vector from a superior vena cava coil to apical patch. A completely transvenous system ultimately consisted of a right ventricular apical coil electrode with a second electrode in the superior vena cava or right atrium.

Initial animal studies demonstrated the efficacy of the device to terminate electrophysiologic induced ventricular fibrillation. Despite initial encouraging published results, there were vigorous dissenters who disqualified the device and the concept of the approach. The first human implantation occurred in 1980 at Johns Hopkins Hospital.[43] The device was non-programmable, committed, and had no telemetry capabilities. There was also no antitachycardia pacing option for patients with unstable ventricular tachycardia. Second generation defibrillators incorporated an epicardial right ventricular electrode for ventricular tachycardia detection.

Generator device and battery advancements have continuously developed. A significant design modification resulted in a new lead design in 1988, allowing for the first complete transvenous implantation[44] consisting of proximal and distal shocking coils.

The concept of tiered therapy was introduced in the early 1990s. A progressive therapy for ventricular tachycardia allowed for initial programmed bursts of antitachycardia pacing, followed by a low

Figure 1.4 *Early implantable single chamber device to current dual chamber Kappa. (Source: Reproduced with permission of Medtronic, Inc.)*

energy shock for unstable VT, culminating in a high energy shock for unstable VT not terminated using step 2 or for tachycardia that had degenerated to ventricular fibrillation.

Advancement in devices and patches has allowed the successful implantation of ICD therapy in even young patients, and those with complex congenital heart disease anatomy limiting ICD lead placement (endocardial and/or epicardial) and generator positioning (thoracic or abdominal). Current guidelines for ICD implantation are likewise detailed in the 2008 ACC/AHA/HRS Guidelines for Device Based Therapy of Cardiac Rhythm Abnormalities.[28]

Summary

It is difficult to find a better example in medicine where the development of technology was driven in large part by the needs of children than that seen in pacer therapy. This occurred secondary to the fact that symptomatic bradycardia frequently presents in childhood as congenital heart block or a consequence of congenital heart disease. The primary motivation for successful cardiac pacing paralleled the development of open heart procedures for patients with congenital heart disease. This new era of palliation of children previously doomed to a life of disability could not to be derailed by a heart rate that did not maintain an adequate cardiac output. The commitment of Dr. C. Walter Lillehei and other pioneers at the University of Minnesota in the 1950s to continue with their heroic efforts to offer these children the potential for a normal life led to Earl Balken developing what is now Medtronics in a small garage in Minnesota. Throughout the last 60 years, the needs of children relative to size and anatomy have led to the development of smaller pacers and leads that have, in turn, continued to advance the field for patients of all sizes and ages (see Figure 1.4).

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