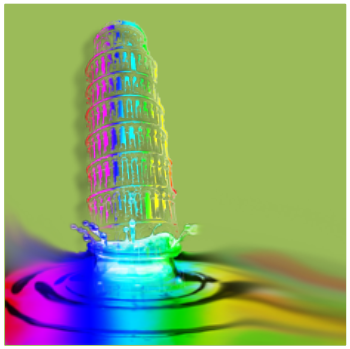
Pacemaker e defibrillatori



15/05/2020

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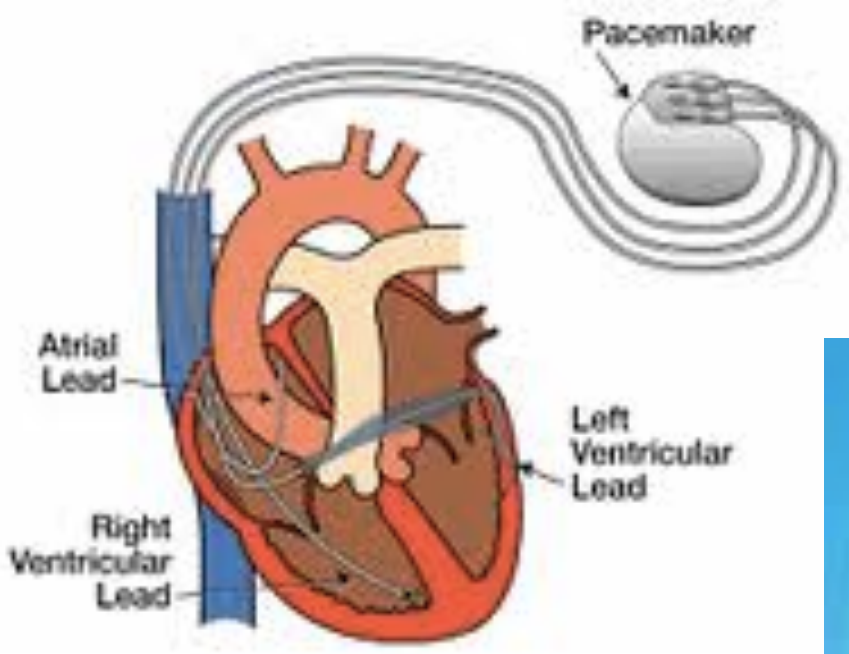
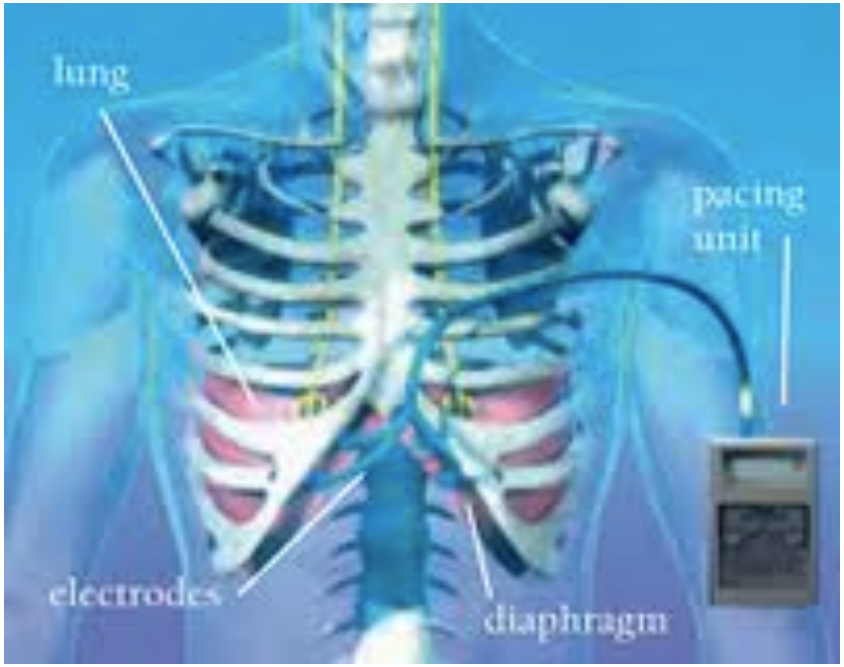
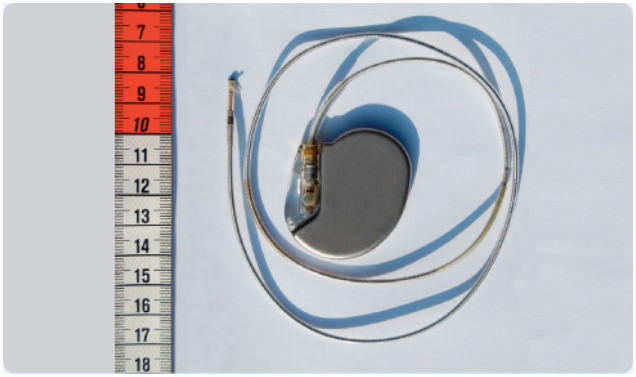
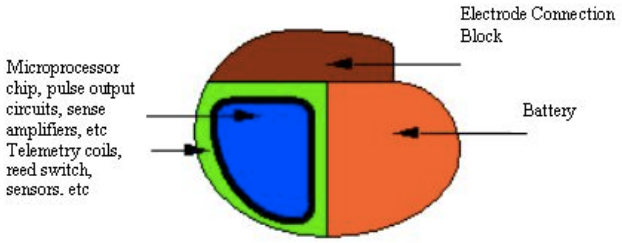
Pacing system

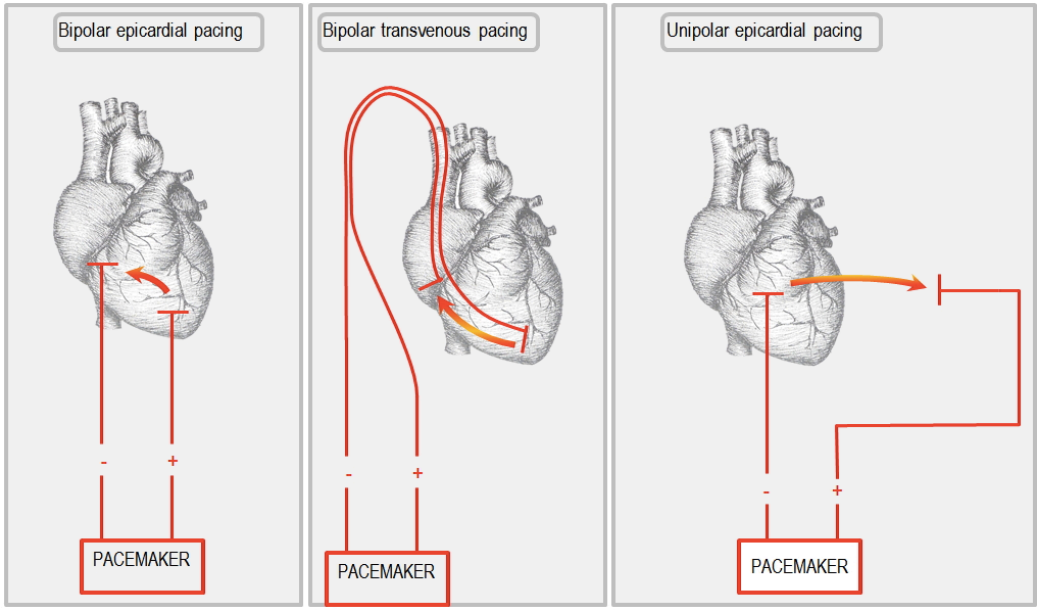




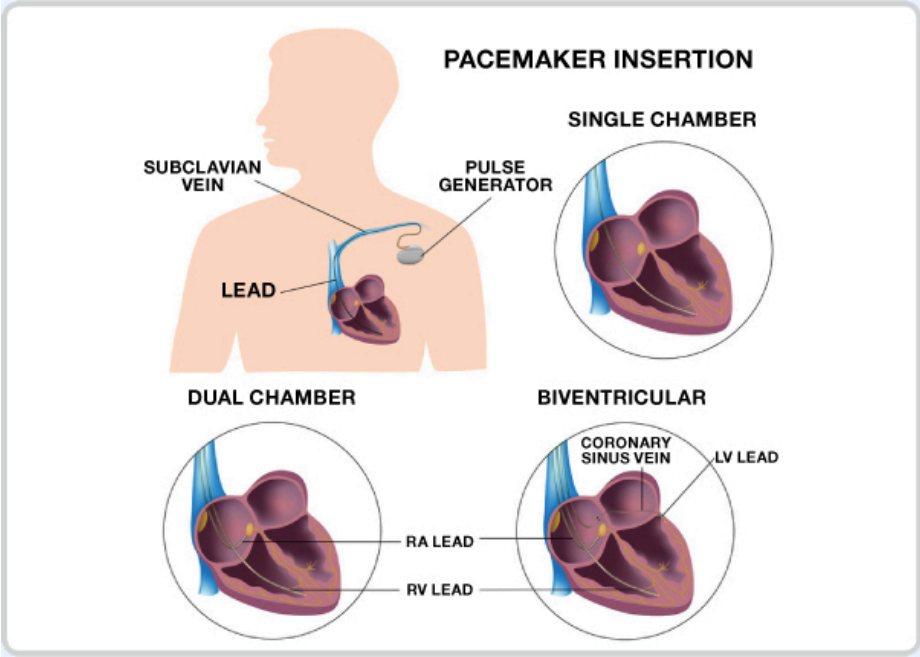
Pulse generator

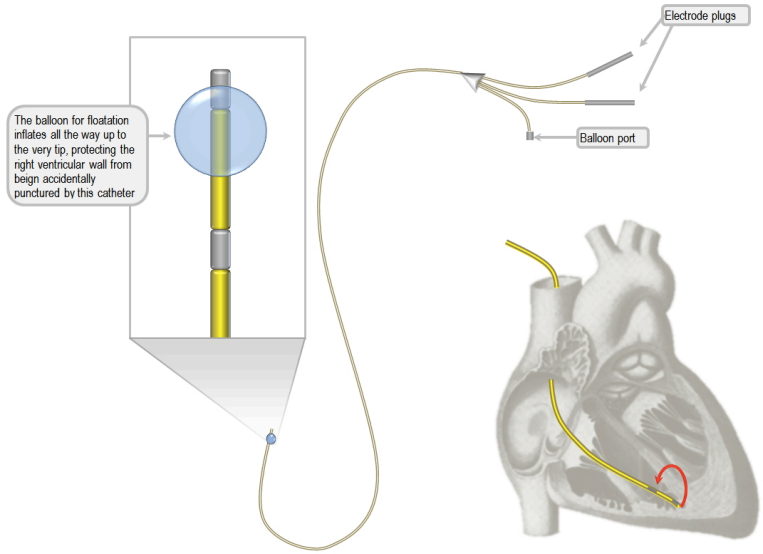


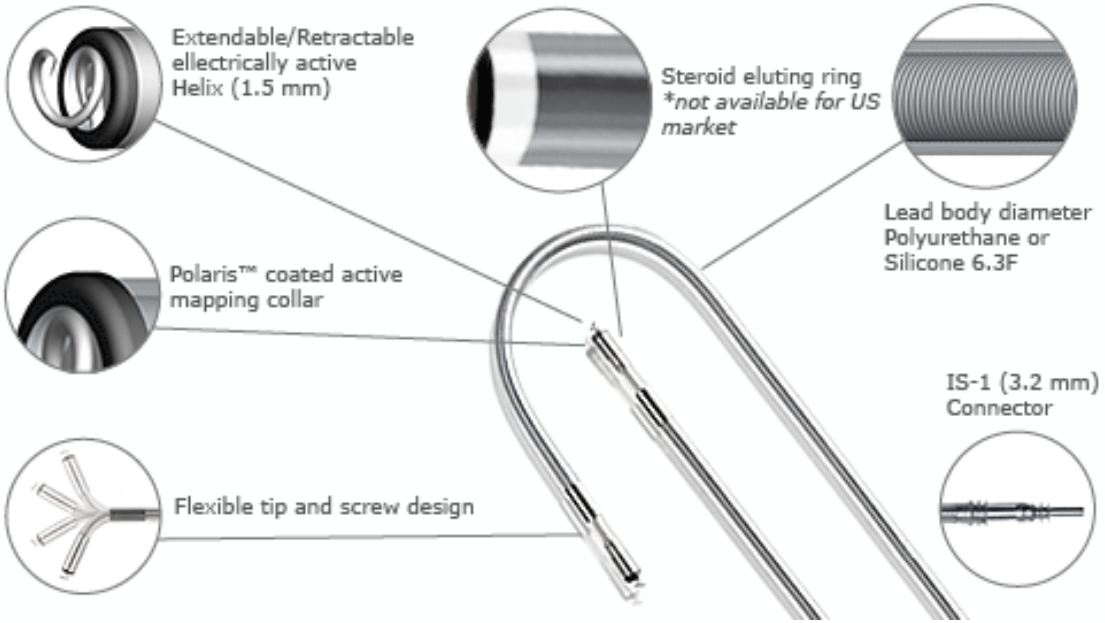


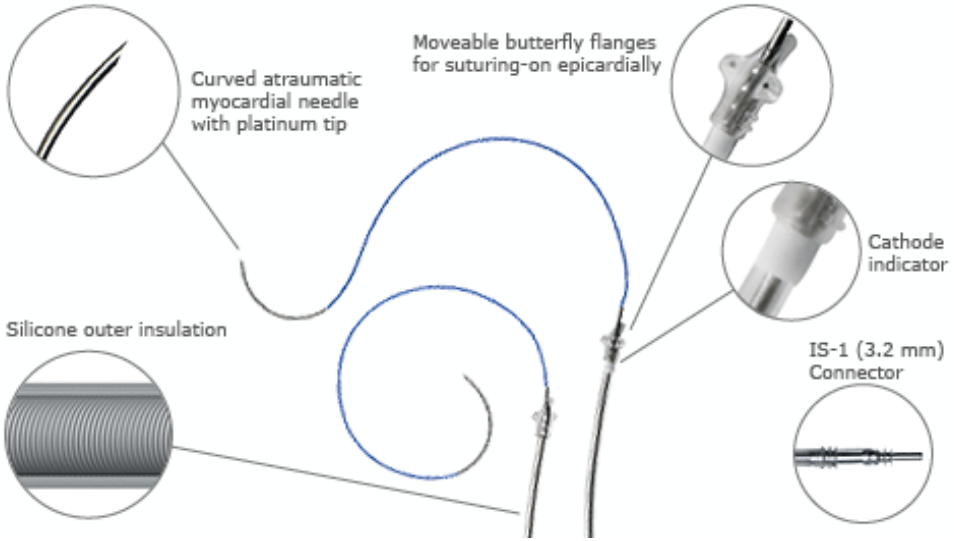


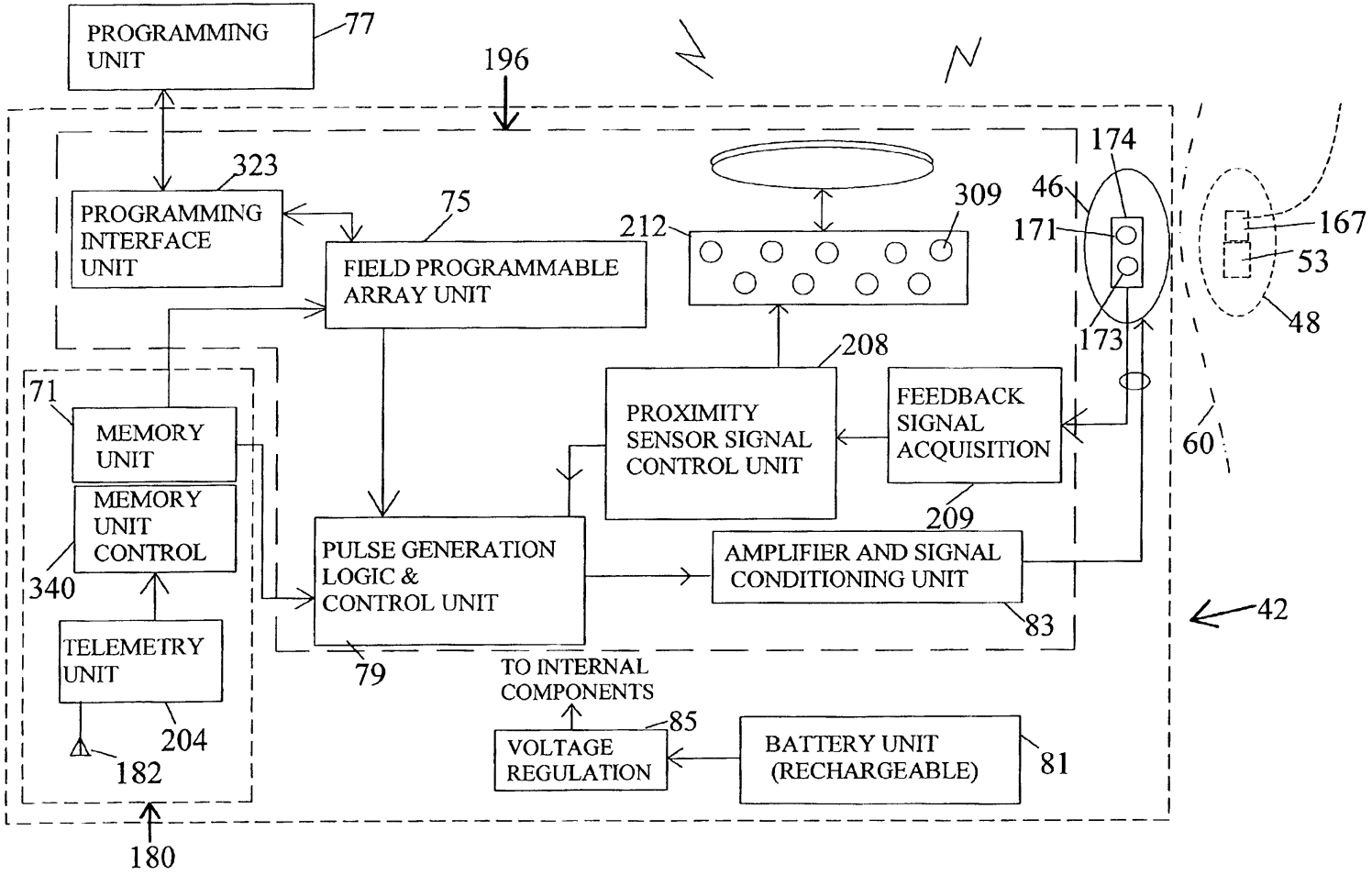
Sensing (and pacing) is accomplished with one of two configurations, bipolar and unipolar. In bipolar, the anode and cathode are close together, with the anode at the tip of the lead and the cathode a ring electrode about 2 cm proximal to the tip. In unipolar, the anode and cathode may be 5–10 cm apart.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


PACEMAKER INSERTION
SINGLE CHAMBER
SUBCLAVIAN VEIN
PULSE GENERATOR
LEAD
DUAL CHAMBER
BIVENTRICULAR
CORONARY SINUS VEIN
LV LEAD
RA LEAD
RV LEAD


Electrode plugs
Balloon port
The balloon for floatation inflates all the way up to the very tip, protecting the right ventricular wall from being accidentally punctured by this catheter


Extendable/Retractable
ellectrically active
Helix (1.5 mm)
Steroid eluting ring
*not available for US
market
Lead body diameter
Polyurethane or
Silicone 6.3F
Polaris™ coated active
mapping collar
IS-1 (3.2 mm)
Connector
Flexible tip and screw design


Curved atraumatic
myocardial needle
with platinum tip
Moveable butterfly flanges
for suturing-on epicardially
Cathode
indicator
Silicone outer insulation
IS-1 (3.2 mm)
Connector


PROGRAMMING UNIT
77
196
323
PROGRAMMING INTERFACE UNIT
75
FIELD PROGRAMMABLE ARRAY UNIT
212
309
174
46
171
167
53
48
173
208
71
MEMORY UNIT
MEMORY UNIT CONTROL
340
PROXIMITY SENSOR SIGNAL CONTROL UNIT
FEEDBACK SIGNAL ACQUISITION
60
209
PULSE GENERATION LOGIC & CONTROL UNIT
AMPLIFIER AND SIGNAL CONDITIONING UNIT
83
42
TELEMETRY UNIT
204
182
79
TO INTERNAL COMPONENTS
85
VOLTAGE REGULATION
BATTERY UNIT (RECHARGEABLE)
81
180

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 trouble-shooting 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.

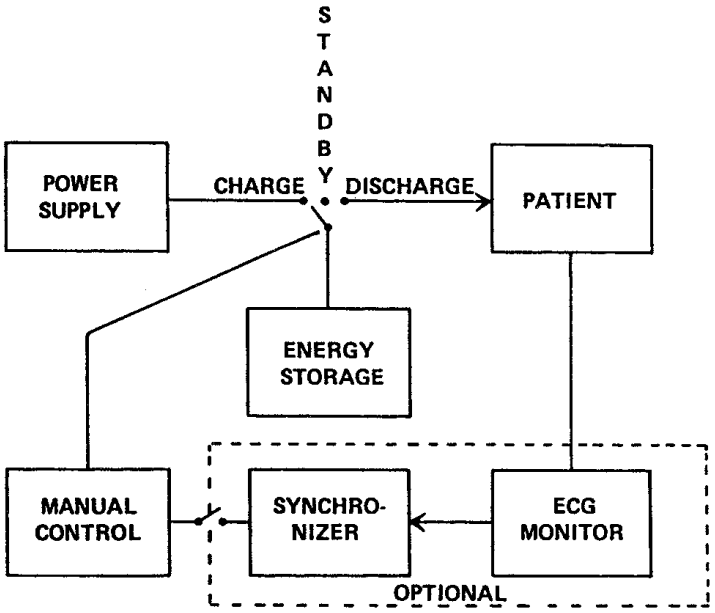
The longevity of very early pulse generators was measured in hours. With the lithium-iodide technology now used, longevity has been reported as high as 15 years. The clinical desire to have a generator that is small and full-featured yet also long-lasting poses a formidable challenge to battery designers. 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.

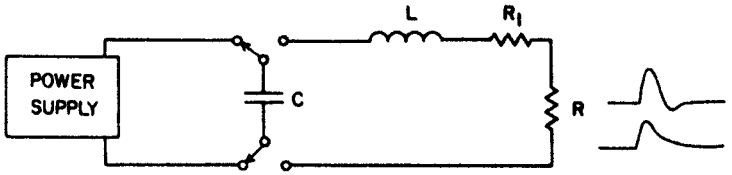
Another important factor in battery design lies in the electrolyte that separates the anode and the cathode. The semisolid layer of lithium iodide that is used gradually thickens over the life of the cell, increasing the internal resistance of the battery. The voltage produced by lithium-iodine batteries 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.

Defibrillatori Esterni



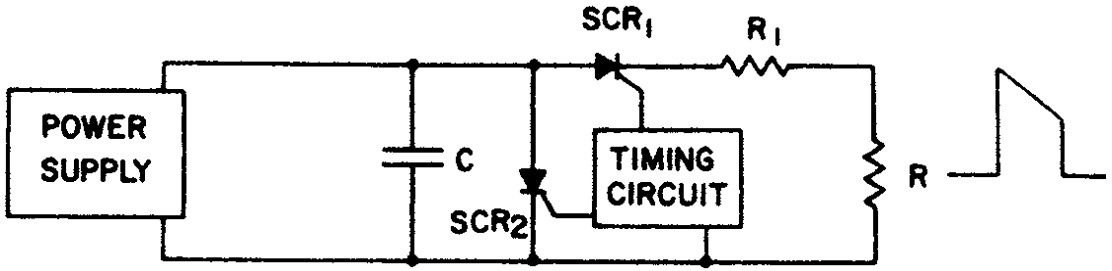
$$W_s = \frac{1}{2} CE^2$$



$$W_d = W_S \times \left(\frac{R}{R_i + R} \right)$$

Ws= energia immagazzinata
C= capacità
E= tensione

Wd= energia rilasciata
R= resistenza individuo
Ri= resistenza ingresso



$$L\frac{di}{dt}+\left(R_i+R\right)i+\frac{1}{C}\int idt=0$$

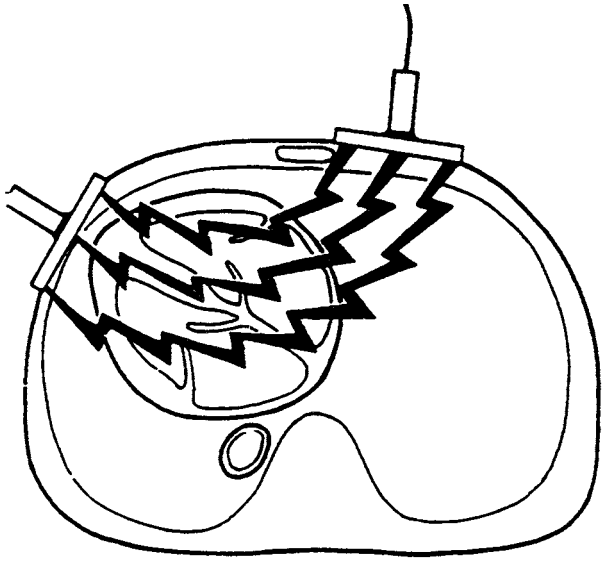
Derivando rispetto al tempo

$$L\frac{d^2i}{dt^2}+\left(R_i+R\right)\frac{di}{dt}+\frac{1}{C}i=0$$

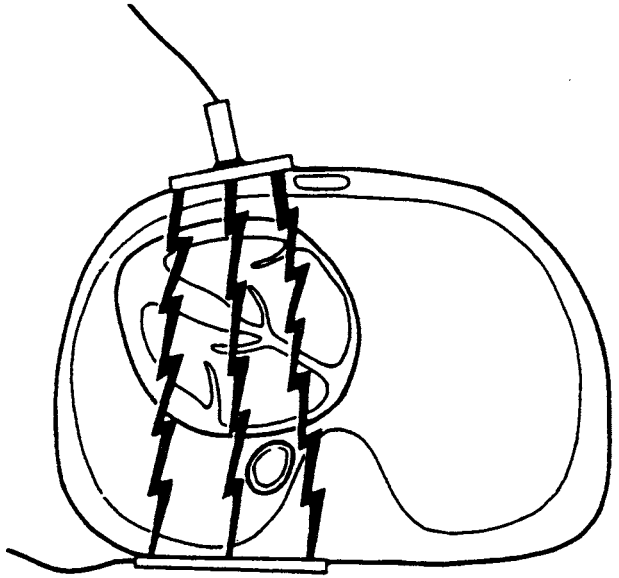
$$W_d=0.5\ I_i^2\ R\left[\frac{d}{\log_e\left(\frac{I_i}{I_f}\right)}\right]\left[1-\left(\frac{I_f}{I_i}\right)^2\right]$$

Elettrodi

Electrodes for external defibrillation are metal and from 70–100 cm2 in surface area. They must be coupled to the skin with an electrically conductive material to achieve low impedance across the electrode-patient interface. There are two types of electrodes: hand-held (to which a conductive liquid or solid gel is applied) and adhesive, for which an adhesive conducting material holds the electrode in place. Hand-held electrodes are reusable and are pressed against the patient's chest by the operator during shock delivery. Adhesive electrodes are disposable and are applied to the chest before the shock delivery and left in place for reuse if subsequent shocks are needed. Electrodes are usually applied with both electrodes on the anterior chest as shown in Fig. 1or in anterior-to-posterior (front-to-back) position, as shown in Fig. 2.



Anterior-Anterior



L-Anterior-Posterior

Sincronizzazione

Most defibrillators for trans-chest use have the feature of synchronization, which is an electronic sensing and triggering mechanism for application of the shock during the QRS complex of the ECG. This is required when treating arrhythmias other than ventricular fibrillation, because inadvertent application of a shock during the T wave of the ECG often produces ventricular fibrillation. Selection by the operator of the synchronized mode of defibrillator operation will cause the defibrillator to automatically sense the QRS complex and apply the shock during the QRS complex. Furthermore, on the ECG display, the timing of the shock on the QRS is graphically displayed so the operator can be certain that the shock will not fall during the T wave.