Springer Handbook of Medical Technology

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38. Cardiac Pacemaker Systems

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Pacemaker technology is one of the largest milestones in Medical Technology. Over the past five decades millions of patients benefited from the advantages of this masterpieces of engineering.

Today all types of *bradycardia* and most *tachycardias* can be treated successfully. Pacemakers improve the quality of life significantly. They help to avoid syncopes and dizziness. Pacemakers also support the heart and improve the *cardiac* output. State of the art pacemakers are no longer simple pulse generators but cardiac rhythm management systems.

Pathological changes in the excitation formation or conduction within the heart are divided into bradycardiac (decelerating) and tachycardiac (accelerating) clinical characteristics. If there are bradycardiac changes, the cardiac output is usually too low, causing the patient to suffer from dizziness, a low maximum stress level, and even problems with consciousness. In these cases, cardiac pacemaker therapy can allow the sinus rhythm to be restored and the symptoms can either be relieved or eliminated altogether.

The idea of electrically stimulating the heart dates back to the beginning of the last century with the work of Burns and Aldini (a nephew of Galvani). But it was 1927 by the time Hyman built the first functioning external pacemaker, a small electric clockwork-driven generator. In 1948, Shockley, Bardeen, and Brattain invented the transistor and made it possible to drastically reduce the size of electric switching units, which also advanced the development of pacemaker design. The first implantable pacemaker was implanted in 1958 by a Swedish doctor named Elmquist. This device was made of 20 discrete components and weighed about 180 g. Pacemakers today weigh about 60 g and have the functionality of a small computer.

Pacemaker therapy is based on the delivery of current pulses, which lead to the artificial depolarization of some cardiac cells. They go across the conduction system of the heart, as well as across the *gap junctions* (intercellular ionic bonding channels, which serve to transmit the excitement directly from cell to cell), triggering a complete contraction of the heart. Based on this triggering effect of artificial pulses, this is referred to as the *all-or-nothing law*. In principle, the current pulses can be applied in four different ways:

Transcutaneous stimulation: In rare cases, the heart is stimulated via externally applied electrodes. In principle, this is possible, but this way is largely ineffective due to the relatively long distance to the heart. This requires high current intensity, which, in turn, triggers unwanted contractions in the skeletal muscles. External stimulation only makes sense in emergencies, e.g., when using an external defibrillator.

Esophagus stimulation: Esophagus stimulation is a minimally invasive procedure. Here, a catheter is advanced via the mouth or nose into the esophagus so that the electrodes come to rest near the atrium of the heart. This approach is mainly used for diagnostic purposes since the atrial position allows better differentiation of atrial and ventricular action. Stimulation within the esophagus does cause pain, however, which kept this method from gaining widespread practice.

Transient intracardial stimulation: In many cases, bradycardia is only temporary, requiring only temporary therapy (in general terms, this is referred to as transient stimulation). This is used for the following indications:

- Asystoles or external ventricular bradycardia with acute sinoatrial (SA) and atrioventricular (AV) blocks
- Acute conduction defect in the case of a recent anterior myocardial infarction
- Complicated pacemaker exchanges
- As a prophylactic during cardiological operations
- Acute poisoning, especially with pharmaceuticals, such as digitalis or antiarrhythmics

In these cases, a catheter is pushed into the right ventricle via a venous access. Alternatively, so-called heartwires (flexible, insulated wires with a needle at both ends) are fixed in the heart through the chest wall. This is especially suitable for open-heart surgery, since there is already direct access. Pulses are generated via an external stimulator.

Intracardial stimulation using implants: Especially common, however, is permanent stimulation for the rest of the patient's life, for example, when the patient suffers from the following conditions:

- Considerable defects in the auricular stimulus conduction system, such as SA or AV blocks of the second or third degree or fascular blocks
- Bradycardias and arrhythmias after cardiac infarction
- A sick sinus node (sick sinus syndrome)
- Carotis sinus syndrome (this clinical condition is characterized by an oversensitivity of the pressoreceptors in the carotis sinus, where, for example, heart activity is limited when one turns one's head)

Technically, all methods are very similar. Clinically, however, the use of implants is especially important, so in what follows, implant technology is discussed exclusively.

38.1 Structure of a Pacemaker System

A pacemaker system consists of an external programming device, the actual cardiac pacemaker, which contains a battery and electronic elements, as well as a pacemaker electrode (Fig. 38.1). The electrode is advanced almost exclusively via a venous access (e.g., the vena cava sup.) underneath the collarbone into the right half of the heart and anchored there. Afterward, the pacemaker is connected and implanted subcutaneously in a pocket of skin.

38.1.1 The Programming Device

When the pacemaker is programmed, the system should be optimally adapted to the needs of the patient, and the

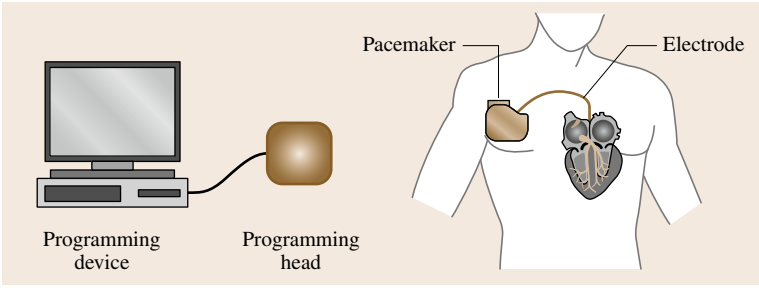

Fig. 38.1 Structure of a pacemaker system

lifetime of the battery should be maximized. With the long running times of today, it can't be assumed that the conditions on the day of implantation will remain constant over the entire lifetime of the pacemaker. It is therefore necessary to be able to adapt the pacemaker to changing conditions.

External devices that are usually based on a PC platform and drive manufacturer-specific short-range telemetric equipment are used for programming. This data transmission, usually based on inductive near-field coupling in a range below 100 kHz, allows the implant to be queried/reprogrammed. To do this, the programming head is first placed on the skin above the pacemaker housing. Afterward, the current pacemaker parameters, as well as diagnostic values (markers, event histograms, ECG, etc.), are transmitted out and evaluated. Finally, the updated parameters are reloaded into the implant. The most significant programmable parameters include the following:

- Stimulation frequency (lower and upper limits)
- Pulse amplitude and duration
- Input sensitivities
- Refractory periods
- Hysteresis (difference between the intervention and stimulation frequency)
- Av delay and hysteresis
- Pacemaker mode

The number of programmable parameters is meanwhile in the triple digits, and optimum pacemaker programming requires special knowledge, requiring supplemental education, even for cardiologists. For this reason, automatic functions are being increasingly implemented that allow the implant to adapt itself. In addition, Web-based expert systems are being considered, or are already in the test phase, that are supposed to ensure the constant availability of the most current knowledge. For more detailed information, one must refer to the respective pacemaker manuals or the corresponding special literature [38.1–4].

38.1.2 The Pacemaker

The pacemaker itself consists of a battery, the electronics, a hermetically enclosed titanium housing, and an epoxy resin connector for accommodating the electrode (Fig. 38.2). Nowadays, lithium-iodide batteries (open-circuit voltage about 2.8 V, capacitance about 1 Ah, inner resistance a few 100 and 50 kΩ) are usually used as a power source. With this type of battery, the anode is made of lithium and the cathode of iodine. In addition to its extraordinarily low self-discharge (< 1%/year), the lithium-iodide battery offers high stability of the inner resistance beyond the discharge time. Only at the end of the service life does the open-circuit voltage rapidly drop. Lithium-iodide batteries thus offer the greatest possible reliability with their long lifetime, small dimensions, and light weight. Today, they have running times ranging from 5 to 10 years with this technology [38.3].

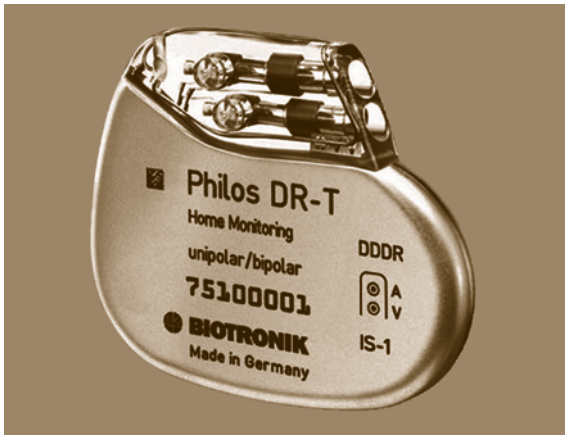

Fig. 38.2 Cross-section of a DDD pacemaker (source: Biotronik)

The trend toward primary cells, or rechargeable batteries, which are at the beginning of their technical development and are occasionally discussed even today, offer no major advantages. The supposed prolonged service life must be weighed against the drastic increase in aging-related failures after about 10 years, which makes an implant change recommendable after this time, anyway.

The electronics are usually designed as a hybrid, multichip module. The once-used thick film technique is being increasingly replaced by multilayer ceramics due to the increasing complexity of the circuits. Recently, brand new PCB technologies (e.g., Dycostrate) have been attracting interest, which, thanks to their flexibility, allow electric connection technology to be considerably simplified, thus improving the device quality.

38.1.3 The Pacemaker Electrode

The pacemaker electrode or probe establishes the connection between the pacemaker and the heart. It consists of a connector, i. e., the plug for connection to the

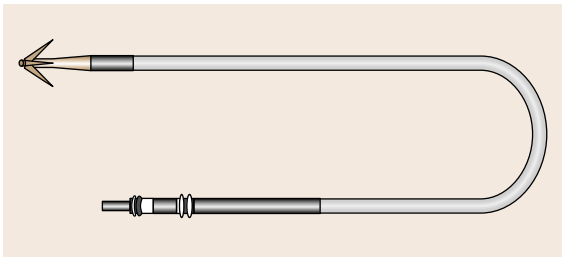

Fig. 38.3 Schematic structure of a pacemaker electrode

pacemaker, the electrode conductor, and the electrode tip (Fig. 38.3). The connector configuration has been standardized according to the IS-1 standard [38.4], so that all modern electrodes can be connected to any pacemaker.

One of the main technical challenges is the high bending stress of about 40 million load changes per year. For this reason, electrode conductors nowadays are made of four individual coiled wires, which provide the highest possible reliability with high flexibility. With a coiled wire, the bending radius is reduced, which reduces the alternating load stress on the feed line material (usually MP35N stainless steel). For this reason, electrode breakage is no longer a major problem.

The newest developments in the area of multifocal stimulation (Sects. 38.3.6 and 38.3.7) sometimes require very thin electrodes. In these cases, so-called DFT wires are used, which are special wires with a sheath made of MP35N and a silver core. These guarantee low electrical resistances while also having high mechanical strength.

Special demands are placed on the insulation material. It must be biocompatible and also be able to withstand the mechanical and chemical loads in the body. Right now, insulation made of silicone and polyurethane are being used [38.3, 4]. Polyurethane has a much lower friction coefficient, which is why it is especially preferred when implanting several electrodes via the same vessel. Clinical experience has shown, however, that when there is minor contamination, which can hardly be controlled in production, this can lead to hydrolysis, and thus to possible aging/brittleness, which ultimately leads to a life-threatening short-circuit. For this reason, several tens of thousands of electrodes from different manufacturers had to be removed/replaced a few years ago, which is why silicone is the preferred insulation material today.

Furthermore, a distinction is made between uni- and bipolar pacemaker electrodes (Fig. 38.4). Unipolar means that the electrode tip acts as the cathode and the pacemaker housing (or another counterelectrode with a large surface area) acts as the anode. Due to the considerably greater surface area of the pacemaker compared to the electrode tip, the housing is also called an *indifferent pole*. Bipolar probes also work with a cathodic tip. The anode, however, is also placed in the distal electrode area (about 2.5 cm away from the tip). Bipolar probes are somewhat thicker and more rigid than unipolar electrodes since they have two feed line coils. It is hardly possible to repair electrode breakage.

Important advantages of bipolar probe technology are the insensitivity to electric potentials whose origins lie outside the area where the electrode tip was implanted (potentials of the skeletal muscle, P-wave for ventricular probes, R-wave for atrial probes) and the lower risk of muscle contractions, even if higher energy is delivered [38.1, 3]. In addition, interference due to electromagnetic couplings in the feed lines cancel each

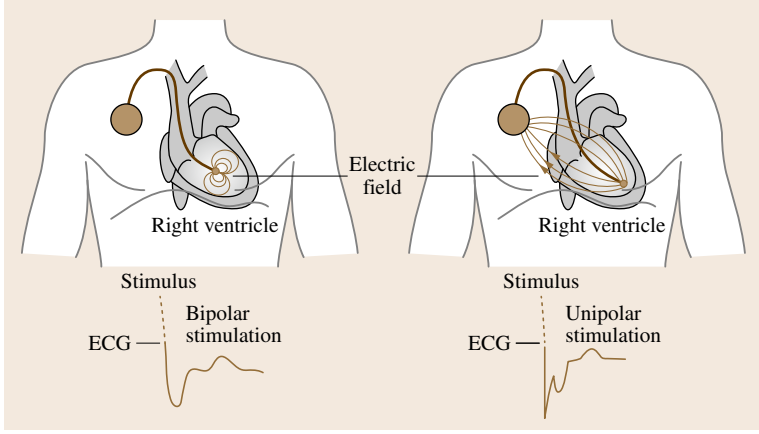

Fig. 38.4 Electric field distribution and ECG for uni- and bipolar stimulation

other out. A widespread compromise is the implantation of bipolar probes in the atrium of the heart (a smaller potential requires a higher sensitivity and the suppression of interference signals) and unipolar probes in the ventricle area.

The electrode head has direct contact with the endocardium. Its size, geometry, and surface structure are major determining factors for the stimulation and detection properties. A porous or fractal surface has proven advantageous [38.5]. The probe head is anchored using various fixation systems (Fig. 38.5). One distinguishes between electrodes with passive fixation (barblike anchor systems out of the insulation material of the electrode) and those with active fixation (system of screws on the electrode head). The former are preferred for electrode anchoring in the strongly divided surface of the right ventricle (trabeculae structure), and the latter is used for anchoring in the smoother right atrium. A hybrid type is the so-called J-electrodes, which ensure stable contact with the wall thanks to the J-shaped hook at the beginning of the electrode feed line. They are mainly used in the atrium. Probes without a fixation mechanism are used for transient pacemaker stimulation in emergency situations and can be easily removed again.

All described electrodes are fed transvenously to the right heart. They stimulate the inside of the heart and are therefore referred to as endocardial electrodes. Epicardial electrodes, which are applied to the outside of the heart, are no longer used for permanent pacemaker therapy. At most, they play a minor role in pediatric applications.

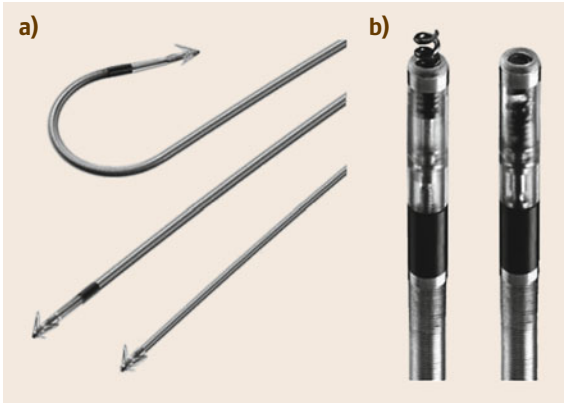

Fig. 38.5a,b Passive (a) and active (b) anchorable probe heads of cardiac pacemaker electrodes. The top passive electrode is a so-called J electrode. Below this follow a bipolar and a unipolar standard electrode. The screw electrode, which must be actively anchored, is shown once with an extended tip (ready for screwing into the myocardium) and, next to this, with a retracted tip (to avoid complications when inserting the electrode in the heart)

Pacemaker electrodes are hollow on the inside. This allows a reinforcement wire (mandrin) to be inserted during implantation, which makes it easier to push in the electrode. The mandrin is removed again before the electrode is connected to the pacemaker.

38.2 The Functionality of a Cardiac Pacemaker

A cardiac pacemaker not only blindly stimulates at a defined rate but attempts to synchronize itself to the still-existing heart rhythm and only to intervene when needed. Its structure can thus be divided into three basic modules: (1) the input stage for detecting intracardial electrical signals, and thus physiological events, (2) the output stage for stimulating the heart, and (3) a control unit, which takes the physiological demands, such as the heart rate, synchronicity, etc., into account (Fig. 38.6). The complexity of the control unit is determined to a great degree by the type of illness and, therefore, the stimulation mode to be used. A special section is dedicated to this aspect (Sect. 38.3).

The Sensing Function

The input sensitivity of a pacemaker system describes its capability to detect electric signals from the heart in the millivolt range and to interpret them as a contraction of the atrium/ventricle (perception, detection or sensing function). The fundamental structure of the input stage required for this is equivalent to that of an ECG amplifier [38.2], whereby the boundary conditions of an implant (size, supply voltage, energy requirement) sometimes demand special solution approaches.

To differentiate between physiological heart signals and interference signals, input filters are used in pacemakers. The filter characteristics have now been standardized (e.g., DIN 7505, Part 9) in order to establish binding safety standards. The greatest input amplification of a pacemaker after this stage lies between 30 and 70 Hz [38.6].

Furthermore, in clinical practice, an indirect measure for the frequency content of the electrical signal, the maximum slew rate, which is easy to determine by

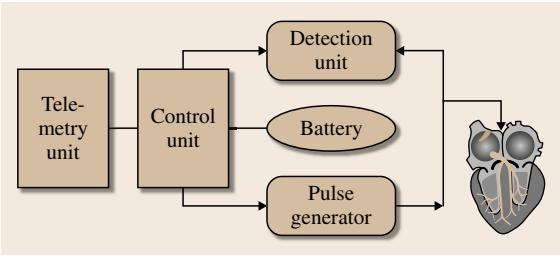

Fig. 38.6 Fundamental block diagram of a cardiac pacemaker

means of differentiation, is used. In order for a signal to be detected as such by the pacemaker, it must exceed a certain slew rate. The limit values of the slew rate lie at 0.5 V/s for the atrium area and at 1 V/s in the ventricle.

Another parameter for defining the perception function is the signal's amplitude. The so-called perception threshold or sensitivity of a pacemaker is usually programmable. In the atrium, one strives for an amplitude of 2.5 mV. In the ventricle, values up to 10 mV are to be registered. Often, real signals lie considerably below this, however [38.1].

Another special feature is so-called *blanking*. The detection channel is connected to the same electrode as the output stage. The amplifier would be overdriven by the stimulation impulse and be put out of operation over a longer period of time. For this reason, shortly before delivering a pulse, the inputs are decoupled by a transistor, which is generally referred to as *blanking*. A few milliseconds after the stimulation pulse, the blanking switches are closed again.

Logical blanking follows physical blanking in the sense of a switch. This means that no detection events within a certain window after the pulse are evaluated by the following circuit. There are several reasons for this. For one thing, vibration processes, which are unavoidable when an amplifier is switched on again, for example, due to small voltage differences, are not misinterpreted as physiological events. For another thing, refractory periods can be introduced this way, similar to cellular behavior, where an input amplifier is switched to become insensitive. The advantages of such a refractory period are explained in more detail in Sect. 38.3 with the introduction of various pacemaker modes.

38.2.1 The Stimulation Function

The behavior of a stimulation system is determined by the actual impulse generator, the electrodes required for coupling, and the tissue through which the excitation current flows. A very detailed description of this system can be found in [38.5].

The Pulse Generator

Nearly all pulse generators are based on the principle of capacitor discharge, i. e., a so-called reservoir capacitor C_{res} is charged to a certain voltage via the switch S_1 during the passive phase and discharges itself into the tissue to be excited in the active phase via switch S_2 and an additional coupling capacitor C_c (Fig. 38.7). If the required stimulation voltage lies above the battery voltage U_{bat}, a charge-pumping circuit downstream multiplies the voltage by the factor n.

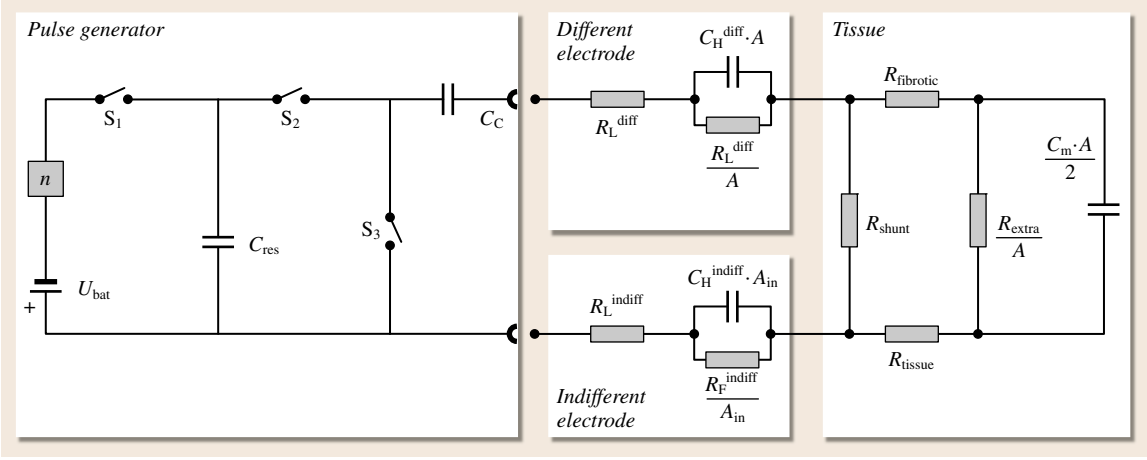

Fig. 38.7 Equivalent circuit of a stimulation system, consisting of a pulse generator, electrodes, and tissue

This setup prevents the battery from being directly coupled with the myocardium in the event of a single component defect (e.g., defective switch), which would lead to electrolysis and thus to damage to the surrounding tissue. This principle increases the safety of the stimulators and is referred to as *single failure safety*. The output voltage and current are thus in no way constant with respect to time, but drop nearly exponentially, depending on the impedance of the external components.

Since capacitive components (C_c as well as the limiting phase capacitances of the electrodes) are charged by the stimulation pulse, effective discharging between the individual stimuli must be ensured. This is done using a so-called autoshorting switch S_3, which is closed during the passive phase.

The Electrodes

From an electrical engineering point of view, pacemaker electrodes are solid-state electrolyte interfaces. A detailed description of such systems can be found in [38.2]. They therefore have capacitive properties due to the Helmholz layer and possibly oxidized top layers. C_H describes the specific Helmholz capacitance in what follows, which only delivers the capacitance that is interesting for the circuit after multiplying with the geometrical electrode surface A. This way, surface modifications, which lead to a roughening and thus to a larger electrochemical contact surface, can be easily considered in the model. R_F correspondingly describes the Faraday resistance and R_L stands for the lead resistance.

The Tissue

The excitation of at least one heart muscle cell (myozyte) is required for electrically stimulating the heart, i. e., a depolarization of its cell membrane. In its resting state, the transmembrane potential is about −90 mV. If this value is increased by a depolarization voltage U_{dep} of about 30 mV, automatic depolarization, and thus the contraction, is initiated.

For stimulation using extracellular electrodes, a displacement current is therefore necessary beyond the cell membrane. The first approximation of the electrical equivalent circuit for the target organ consequently consists of a specific cell membrane capacitance C_m and a parallel ohmic resistance R_{extra}, which takes the extracellular space into account. There are three other approximately ohmic contributions in addition to this. $R_{fibrotic}$ represents the resistance of the thin layer of fibrotic tissue, which develops after implantation between the electrode and the place of stimulation. R_{tissue} describes the resistance of the tissue between the place of stimulation and the counterelectrode. R_{shunt} stands for a leakage current, which is not coupled into the target organ via the blood or other tissues, but flows directly into the counterelectrode.

All named elements are at least known in terms of their order of magnitude, so that quantitative estimations are also possible (Table 38.1). Using the Laplace transformation and simplifications

$$C_1 = \left(\frac{1}{C_{res}} + \frac{1}{C_c} + \frac{1}{C_H^{diff} A} + \frac{1}{C_H^{indiff} A_{in}} \right) , \quad (38.1)$$

Table 38.1 Summary of the used parameters and their number values (after [38.7–10])

Parameter		Range	Average	Source
Reservoir capacitance C_{res}		5–20 µF	10 µF	10
Coupling capacitance C_c		5–20 µF	10 µF	10
Spec. Helmholtz capacity C_H	Polished surface	0.1–0.4 µF/mm²	0.2 µF	11
	Fractal surface	10–500 µF/mm²	40 µF	12
Lead resistance R_1		10–100 Ω	50 Ω	10
Geom. electrode surface A		1–20 mm²	10 mm²	10
Surface of the indifferent electrode A_{in}	Unipolar	5–15 cm²	10 cm²	10
	Bipolar	20–100 mm²	50 mm²	10
Pulse duration T		0.1–2 ms	0.5 ms	10
Membrane capacitance C_m			0.01 µF/mm²	13
Tissue resistance R		30–70 kΩ	40 kΩ	14
Shunt resistance R_{shunt}		0.1–2 kΩ	600 Ω	10
Depolarization voltage U_{dep}			30 mV	15

$$C_2 = \frac{C_m A}{2} , \tag{38.2}$$

$$R = R_{\text{fibrotic}} + R_{\text{tissue}} , \tag{38.3}$$

$$R_1 = R_F^{\text{diff}} + R_F^{\text{indiff}} , \tag{38.4}$$

$$R_2 = R_{\text{shunt}} , \tag{38.5}$$

$$a = (R_1 R_2 + R_1 R + R_2 R)\, C_1 C_2 , \tag{38.6}$$

$$b = (R_2 + R)\, C_2 + (R_1 + R_2)\, C_1 , \tag{38.7}$$

and the two poles p_1 and p_2

$$p_1 = -\frac{b - \sqrt{b^2 - 4a}}{2a} , \tag{38.8}$$

$$p_1 = -\frac{b + \sqrt{b^2 - 4a}}{2a} , \tag{38.9}$$

yield the minimum pulse voltage, referred to as the stimulation threshold, required for a successful stimulation

$$U_{\text{thr}} = \frac{2U_{\text{dep}}}{C_1 R_2} \frac{\sqrt{b^2 - 4a}}{\exp(p_1 T) - \exp(p_2 T)} . \tag{38.10}$$

Similarly, one also gets a relationship for the charge (important for calculating the service life), which must be taken from the battery for every stimulus [38.5]. The specific Helmholz capacitance has special meaning for the development of new systems. With increasing C_H, both the stimulation threshold and the charge threshold decrease. Clinically, this means that electrodes with fractal surfaces have the lowest energy consumption, and thus guarantee the longest implant service life. In order to maximize the ohmic parts of the tissue at the same time, the electrode head should be as small as possible. This design – a geometrically small head with a high specific interface capacitance – has become a standard under the term *highly ohmic electrodes* in the last 5 years.

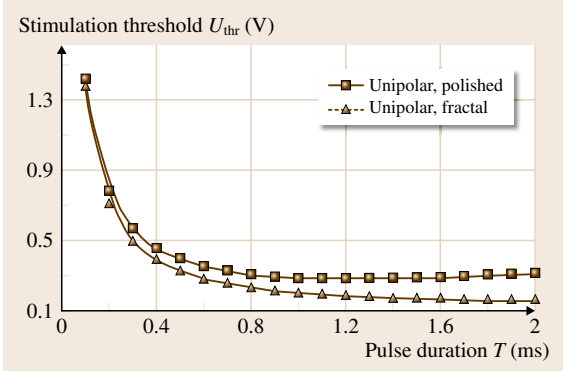

Fig. 38.8 The stimulation threshold as a function of pulse duration. An effective stimulation can be achieved for amplitude values lying above the curve

Figure 38.8 shows the dependence of the stimulation threshold on the pulse duration. From the output voltage perspective, it is recommended to choose a large pulse duration in order to avoid switching on a charging pump, which involves high conversion losses. At the same time, however, it has become evident that longer pulse durations increase the charge consumption nearly linearly. From a clinical aspect, then, it makes sense to select the shortest pulse duration that is still possible without using a charging pump. To be able to specify a simple rule of thumb for optimizing the pulse parameters, the two terms chronaxy and rheobase have been empirically introduced, which can be determined from the stimulation threshold curve. The rheobase is a theoretical value and describes the minimum voltage that triggers an electric cardiac response for an infinite pulse duration. The chronaxy is the pulse duration at the stimulation threshold for a voltage twice the strength of the rheobase. Based on experience, the energetic optimum lies near the chronaxy [38.1].

38.3 Stimulation Modes

From a physiological standpoint, the two atria/two ventricles are always excited together. For this reason, a maximum of two stimulation sites – one in the atrial or one in the ventricular plane – and, correspondingly, a maximum of only two detection sites are required. Based on this, the generally recognized NBG code was developed for defining stimulation modes.

The multifocal modes, which were recently introduced for special heart disorders (interatrial conductive

disturbance, cardiac insufficiency), allow biatrial/biventricular stimulation, but are not yet included in the NBG code.

38.3.1 International (NBG) Pacemaker Code

NBG code (NASPE/BPEG Generic Pacemaker Code), which has been applied since 1988, describes the global function of a pacemaker with the specification of a maximum of five letters, of which the first three are always used, and the fourth and fifth optionally [38.3].

The first letter refers to the stimulation site:

V: Ventricle: Stimulation in the ventricle only
A: Atrium: Stimulation in the atrium only
D: Dual: Stimulation in both the atrium and the ventricle
S: Single: Single-chamber stimulation in the atrium or ventricle
0: No stimulation

The second letter stands for the place of perception:

V: Ventricle: Detection in the ventricle only
A: Atrium: Detection in the atrium only
D: Dual: Detection in both the atrium and the ventricle
S: Single: Single-chamber perception in the atrium or ventricle
0: No perception

The third letter defines the operating mode, i. e., the pacemaker function, which is triggered by a perceived signal:

I: Inhibited: The pacemaker stimulation is suppressed
T: Triggered: A perceived signal leads to the pacemaker delivering the pulse
D: Dual: Inhibition and triggering
0: No inhibition and no triggering

The fourth letter describes the programmability, telemetry, and frequency adaptation:

0: Not programmable
P: Programmable: Up to two functions are programmable
M: Multi programmable: More than two functions are programmable
C: Communication: Data telemetry possible
R: Rate modulation: Adaptation of the pacemaker frequency to a load-induced signal

The fifth letter refers to the antitachyarrhythmia function:

0: No antitachyarrhythmia function
P: Pacing: Antitachyarrhythmia stimulation
S: Shock
D: Dual: Pacing and shock

A variety of possible pacemaker modes result from combining different detection/stimulation channels together with different operating modes. A detailed explanation can be found in [38.2]. In what follows, the two most common modes, the VVI and the DDD modes, will be described in more detail.

38.3.2 VVI Pacemaker

The VVI pacemaker is a single-chamber pacemaker, where one single electrode is anchored in the right ventricle. It detects the ventricular activity in addition to the fixed-rate ventricular stimulation of the V00 and inhibits its output when an intrinsic ventricular event occurs. Thus, a VVI system only stimulates when required, i. e., when the natural excitement fails, which is why it belongs to the group of so-called *demand pacemakers*. A schematic diagram is shown in Fig. 38.9.

The flow diagram in Fig. 38.10 shows the principal mode of operation. The basic interval and the refractory period of the ventricular detection channel are started at the same time. If an intrinsic event is detected within the refractory period, it is interpreted as interference and is discarded. If it is detected outside of the refractory period, the basic interval and refractory period counters are reset. If no R-wave is detected, the VVI pacemaker behaves like a V00 and stimulates after the basic interval is over.

In the functional diagram in Fig. 38.11, the ventricular stimuli are marked with "V". As compared to the V00 principle, R-events are added now, which indicate

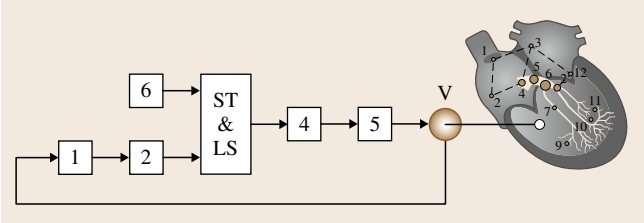

Fig. 38.9 Schematic diagram of the VVI pacemaker. The ventricular electrode (V) leads to a control and logistics circuit via an input amplifier **(1)** and a monoflop for checking the ventricular refractory period **(2)**, which also evaluates the output of the basic interval counter **(6)**. The stimulation channel is equivalent to that in the V00 pacemaker

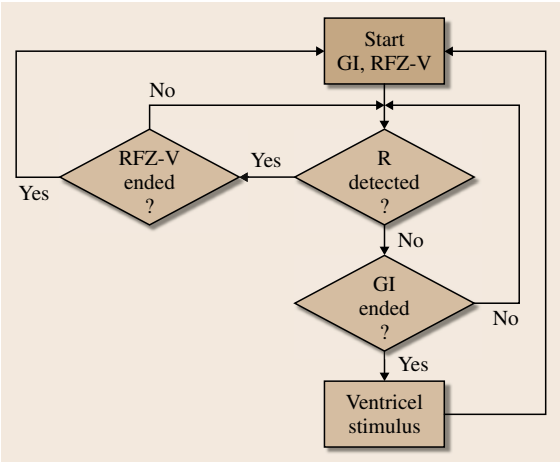

Fig. 38.10 Flow diagram of a VVI pacemaker

the detection of an R-wave. A third marking, plotted below the counter display, indicates the end of the artificial refractory period (RFZ-V) of the ventricular detection channel.

In the ECG, the first two stimulated events (V) are shown, which were triggered by the end of the basic interval. A natural event (R) follows this, which was detected outside of the refractory period and resets the counter. The pacemaker doesn't deliver a stimulation pulse, i. e., it is in inhibiting mode. Afterwards, there is an escape interval, since no intrinsic event is detected. This is followed by another normal basic interval. The ventricular extrasystoles (VES) that follow the stimulated event (V) are detected as an intrinsic cardiac excitement since they are outside of the refractory period, which causes the counter to be reset. Afterward, another escape and a basic interval follow with the associated ventricular stimuli.

Clinical Assessment of the VVI Pacemaker

The VVI pacemaker has an energy-saving mode. It only becomes active when there is no intrinsic cardiac activity (demand pacemaker). It also does not lead to parasystoles and avoids stimulation in the vulnerable phase. Like the V00 pacemaker, however, it is not able to adapt the heart rate to physical exertion. The main problem, however, is the loss of atrioventricular synchrony, which can trigger pacemaker syndrome.

VVI Pacemaker Application

Bradycardias can be generally treated with a VVI pacemaker. Extensive studies have proven in the last few years, however, that the mortality of patients with a VVI pacemaker are considerably higher than that of synchronously stimulated patients. Therefore, the use of VVI pacemakers only makes sense for patients with atrial flutter or fibrillation in connection with an AV block, as well as atrial paralysis (atrium which is not able to contract).

38.3.3 Two-Chamber Pacemaker

In the case of the two-chamber cardiac pacemakers, electrodes are applied to both the atrium and the ventricle to ensure synchronous stimulation. Two principal function options result from this – atrium- and ventricle-based actions. In the case of the atrium-controlled sequence, the basic interval counter and the newly added AV counter (which simulates the natural PR interval) always start at the same time. The ventricular-based sequence, on the other hand, uses a series connection of the two counters, whereby the AV counter must first run down before the timer can be started, and vice versa. The different approach can be especially well explained using the example of a DDD pacemaker.

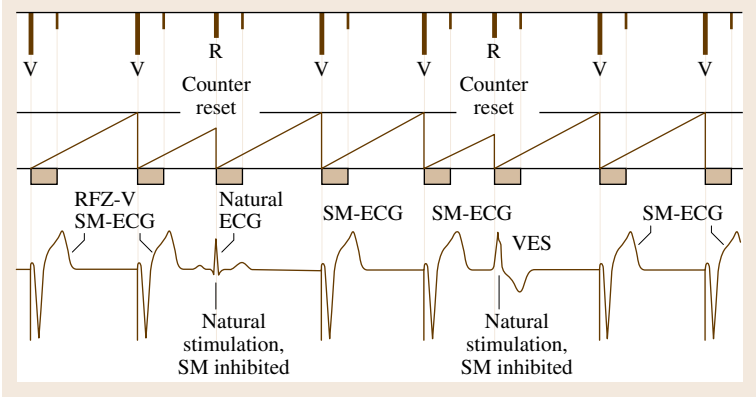

Fig. 38.11 Functional diagram of a VVI pacemaker

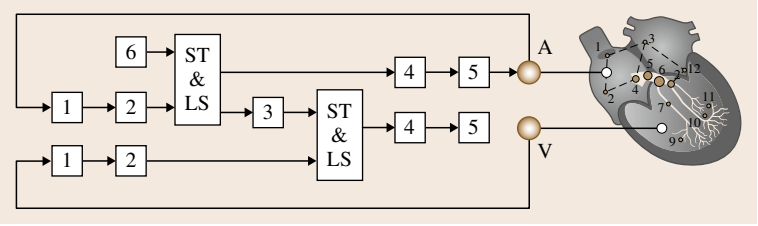

Fig. 38.12 Atrium-controlled DDD pacemaker. Both the atrium and the ventricle have a detection and stimulation channel. The basic interval counter **6** is connected to the atrium control unit, which influences the ventricular control unit via the AV counter

The Atrium-Controlled DDD Pacemaker

The disadvantages of the DVI pacemaker, especially the danger of stimulation of an intrinsic atrial event in the vulnerable phase, are remedied with the DDD pacemaker (Fig. 38.12), which, from the rhythmology point of view, is a kind of *all-rounder* among pacemakers. It combines the AAI, VAT, and VVI functional modes.

Of all the pacemakers, the DDD pacemaker comes closest to simulating the physiological function of the heart. DDD means that stimulation and detection are both possible in the ventricle and the atrium. The pacemaker also masters the trigger and inhibition mode.

The AV interval is started after every atrial excitation (natural or after the basic interval has been artificially induced). If no ventricular heart activity is detected, the ventricle is stimulated at the end of the AV interval. Ventricular extrasystoles (VES) reset the basic interval counter without starting an AV interval. This

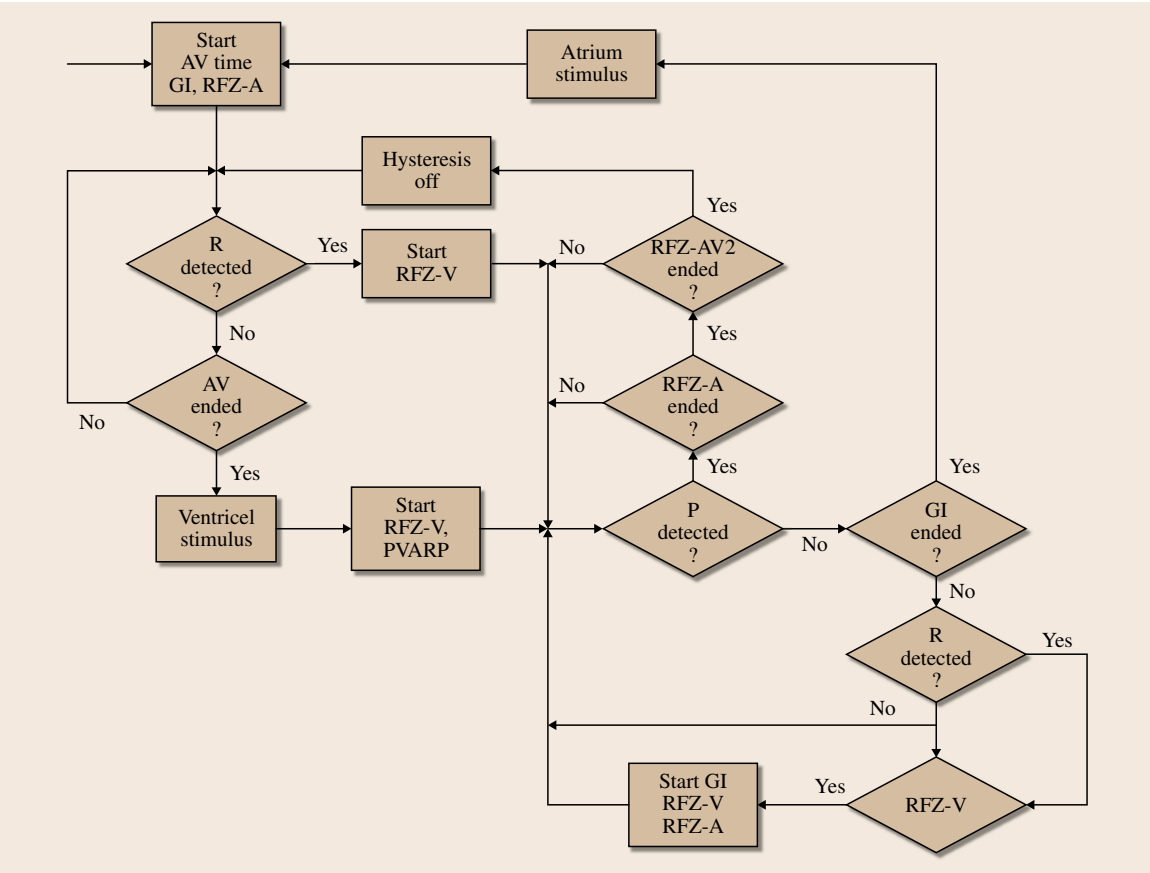

Fig. 38.13 Flow diagram of an atrium-controlled DDD pacemaker

automatically leads to a compensation pause. The basic interval counter is reset as a consequence when one of the following three events happens

- An atrial stimulus is delivered,
- A natural atrial excitation has been detected,
- A natural spontaneous ventricular excitation occurs outside of the AV interval (i. e., a ventricular extrasystole).

In addition, after each event, the respective refractory periods are started. As a special feature, an atrial refractory period should be mentioned here, which begins after ventricular events, in addition to those already described. The task of this so-called postventricular atrial refractory period (PVARP) is to detect ventricular stimuli or to suppress their retrograde feedback to the atrium. Without these protective measures, this could easily lead to one's own output signal being self-detected, and thus to pacemaker-induced tachycardia. Details are shown in Fig. 38.13.

Detection and stimulation in the atrium (P, A) and ventricle (R, V) are shown in the first line of the functional diagram (Fig. 38.14). The small upward dashes indicate the end points of the refractory periods of the atrium, and the downward ones those of the ventricle. The second line of the functional diagram contains the basic interval counter with the refractory periods of the ventricles and the atria.

The bottom line is the AV counter, which has so-called AV hysteresis. The AV interval can be prolonged by means of a second trigger level as soon as the preceding atrial event has been stimulated and was not of a physiological nature. This way, the transit-time differences are compensated, which ensures constant good AV synchronization.

The third line in the functional diagram (Fig. 38.14) contains the surface ECG. The ECG first indicates an atrial pacemaker stimulation (A) with a subsequent artificial P wave. After the AV interval is over, the ventricle is stimulated and the PVARP is begun. After the timer has expired, the above-described cycle repeats itself.

Now a detected natural atrial excitation (P) follows. The basic interval counter is reset and an AV interval is started. Since this is a natural P wave, no AV hysteresis is used. During the AV interval, no ventricular excitation is detected, and therefore, after it is over, a ventricular pacemaker stimulus is delivered (V) and the PVARP timer is started.

Then the basic interval timer runs out, after which a pacemaker pulse is generated in the atrium (A). Even before the AV interval is over, the R wave of the intrinsic ventricular event is detected. The intended ventricular stimulus is inhibited and the PVARP timer is not started. The ventricular extrasystole that now follows is detected by the pacemaker, which then resets the basic interval timer and starts the refractory period.

After the compensation pause, another atrial (P) and a ventricular pacemaker stimulus (V) follow. The P wave of the subsequent natural ECG is detected by the pacemaker (P) and the atrial stimulus is inhibited. The detected R wave prevents a ventricular stimulation since it occurs before the AV interval is over.

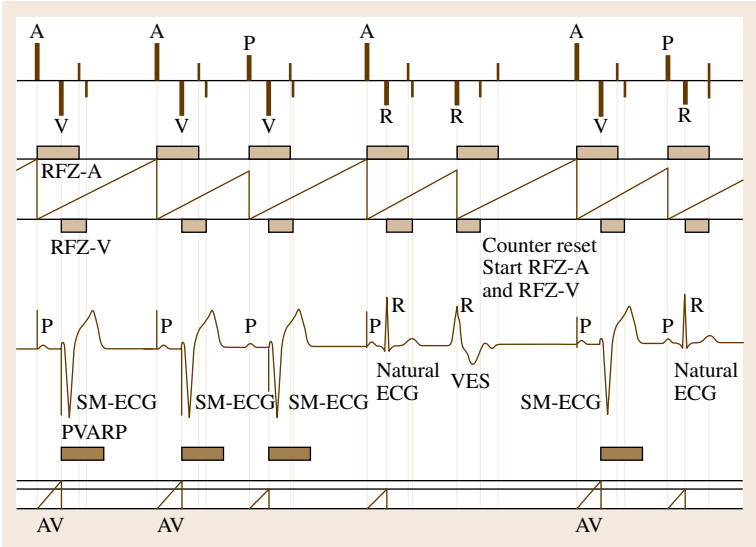

Fig. 38.14 Functional diagram of an atrium-controlled DDD pacemaker

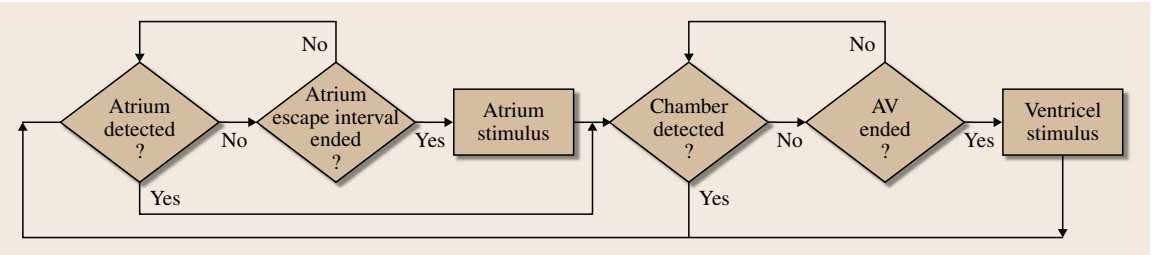

Fig. 38.15 Flow diagram of a ventricle-controlled DDD pacemaker

DDD Pacemaker (Ventricle-Controlled)

The decisive advantage of the DDD system – the complete synchronicity of both chambers – can also be achieved with time control on the ventricular level, of course. In the case of ventricular-based timing, the two counters, AV and the basic interval (here also referred to as the atrial escape interval), are in series and are never active at the same time (Fig. 38.15).

First, measurements are made in the atrium. If an intrinsic atrial event occurs, the AV counter is started and an attempt is made to detect signals in the chamber. If no atrial excitation occurs, an artificial atrial stimulation is delivered at the end of an atrial escape interval, and then measurement continues in the chamber. If a natural excitation occurs in the chamber, measurements are taken in the atrium and the associated timer is started. If this is not the case, another ventricular stimulus is delivered after an AV interval, the counter is reset, and measurements are taken from the atrium. For reasons of simplicity, refractory periods were omitted here.

Clinical Assessment of the DDD Pacemaker. This pacemaker comes closest to simulating the physiological function of the heart. With this pacemaker, no pacemaker syndrome is possible. It has an energy-saving mode of operation. Only in the case of chronotropic insufficiency is the system unable to adapt to the physical stress.

Detecting the atrial electric action is dangerous when the atria have tachycardiac arrhythmia. To prevent ventricular tachycardia from occurring here, the 1:1 atrium/ventricle conduction frequency is limited. If the atrial frequency exceeds this limit, referred to as the maximum synchronization frequency, then there is only partial conduction to the ventricle. For example, the AV delay increases from beat to beat until a beat is no longer conducted and finally the cycle starts over (Wenckebach mode). Another possibility is to just conduct every second atrial action to the ventricle. A retriggerable atrial refractory period was introduced under the term *dual demand*, which leads to a variable prolongment when tachyarrhythmias occur, and is therefore also considered to be a protective function. Most of these methods, however, have the disadvantage that they abruptly change the ventricle frequency when an atrial tachyarrhythmia occurs, which patients find unpleasant.

For this reason, more complex methods for reacting to tachycardiac atrial arrhythmia were introduced recently under the term *mode switching*. The precondition for this is an algorithm for automatically detecting atrial tachyarrhythmias. The state of the art nowadays is the so-called "x out of y" algorithm, where a programmable minimum number x of PP intervals out of a total number y of consecutive intervals must lie under a certain limit. In this way, even irregular atrial tachycardias are reliably detected. When such an arrhythmia sets in, the pacemaker switches over to a ventricular stimulation independent of the atrial action (e.g., VVI). After the end of the atrial tachyarrhythmia, there is another stimulation in DDD mode.

Application of the DDD Pacemaker. DDD pacemakers are suitable for:

- Patients with normal sinus node function and AV block,
- Rare atrial arrhythmia,
- Congenital long QT syndrome and torsade des pointes (special type of ventricular tachycardia).

A special indication for using two-chamber stimulation is hypertrophic obstructive cardiomyopathy. Due to the formation of muscle contractures in the outflow tract of the left ventricle, blood is prevented from flowing out. By choosing a very short AV delay and sometimes also a stimulation in the outflow tract, the muscular excitement can be modified such that the outflow tract obstruction is considerably reduced [38.1]. This pacemaker should not be used for supraventric-

ular tachyarrhythmias. In this case, a VVI system is indicated.

38.3.4 Frequency-Adaptive Pacemakers

Frequency-adaptive pacemakers allow the heart rate to be increased in stressful situations, even in patients with sinus node dysfunction. They are therefore indicated for physically fit patients who have an insufficient heart rate increase under stress (chronotropic insufficiency). The frequency adaptation is achieved via sensors, which determine the stress-related parameters and relate these to the stimulation frequency (so-called *interference coupling*). Up to now, a multitude of different parameters have been tested [38.3], e.g.:

- Physical activity (vibration or acceleration sensors)
- Respiratory minute volume
- Blood temperature
- Blood oxygen content

Other parameters were tested at the same time, but only the physical activity measurement found practical application. Thanks to miniaturized acceleration sensors, it is technically easy to execute and responds immediately. The disadvantage is the nonphysiological sensor information. Furthermore, passive movements (e.g., driving over cobblestones) is misinterpreted. The state-of-the-art technology is therefore made of so-called *dual-sensor systems*, where the movement information is linked with a respiratory sensor signal based on the transthoracic impedance. This way, the reaction times are adapted and nonphysiological effects can be avoided by comparing both signals. Emotional stress or fever, which can also cause the heart rate to increase, are also not detected by this, however.

The currently most promising solution approach is the development of regulating pacemaker systems. The healthy body has several control elements for stabilizing circulation. In addition to an affected sinus node when there is chronotropic insufficiency, the AV node, the myocardium, and the vascular system also act regulatively. Regulatory systems are based on the basic assumption that the control variables of intact control elements can be used to determine the physiologically correct heart rate. For example, an increased inotropism indicates that a greater cardiac output is required, which is used to increase the stimulation frequency. Figure 38.16 makes these relationships clear. The following are currently being tested as input variables:

- AV delay (dromotropism)
- QT interval of the intracardial ECG signal
- Contractility of the myocardium ($\mathrm{d}p/\mathrm{d}t$)
- Motion dynamics of the myocardium (acceleration, impedance)

The dromotropic pacemaker is impressive with its elegance, but it can only be used with limitations, since some people with chonotropic insufficiency also have A-V dysfunction. The QT interval for detecting myocardial contractility has already proven itself clinically but under certain boundary conditions can exhibit positive feedback since other parameters also influence the chamber action potential. The two latter approaches seem to be especially promising now and are already be-

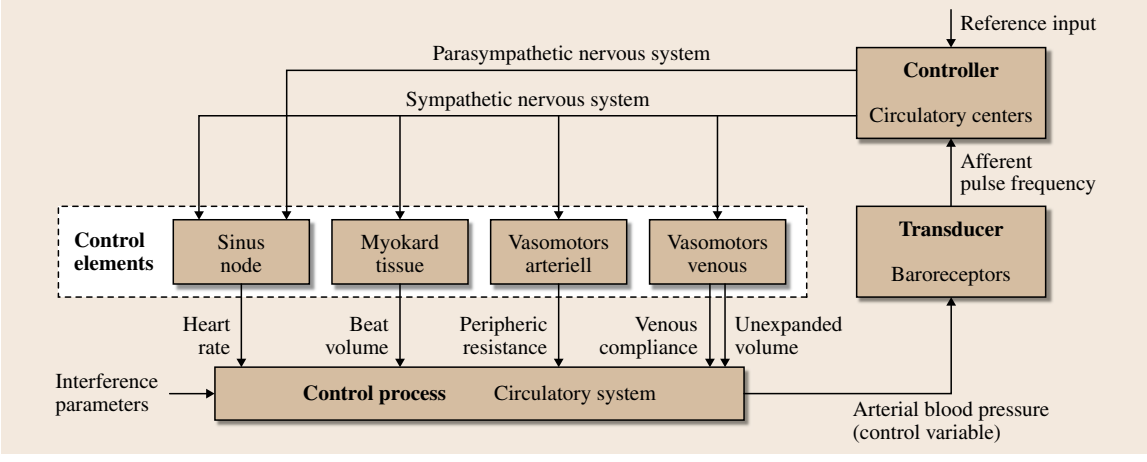

Fig. 38.16 Simplified control diagram of the baroreflex (after [38.7])

ing clinically tested in prototype implants. Their further clinical acceptance will depend on the degree to which they are able to guarantee long-term stability. Due to the sensors growing into the tissue, myocardial restructuring, or just simply aging, these systems are subject to drift, which requires recalibration.

38.3.5 Antitachyarrhythmia Pacemakers

One can attempt to end tachycardia both via a certain sequence of stimulation pulses as well as via electric shock. Since atrial tachycardia is being increasingly successfully treated with catheter-supported ablation methods, pacemaker treatment only plays a subordinate role here. The main indication for electric rhythmization treatment are dangerous tachycardiac ventricular arrhythmias. The corresponding systems are usually offered in connection with a defibrillation option (so-called ICDs).

38.3.6 Biatrial Stimulation Systems

It recently became clear that a pathological prolongment in the interatrial transition period can lead to atrial reentry tachycardia. For this reason, special pacemaker systems have been developed that allow several consecutive stimulations at short intervals in both atria. These so-called *biatrial stimulation systems* have meanwhile proven themselves in patients with prolonged P waves. Technically, they are like a DDD system, whereby, however, the second electrode comes to rest in the left atrium. The challenge here is the development of

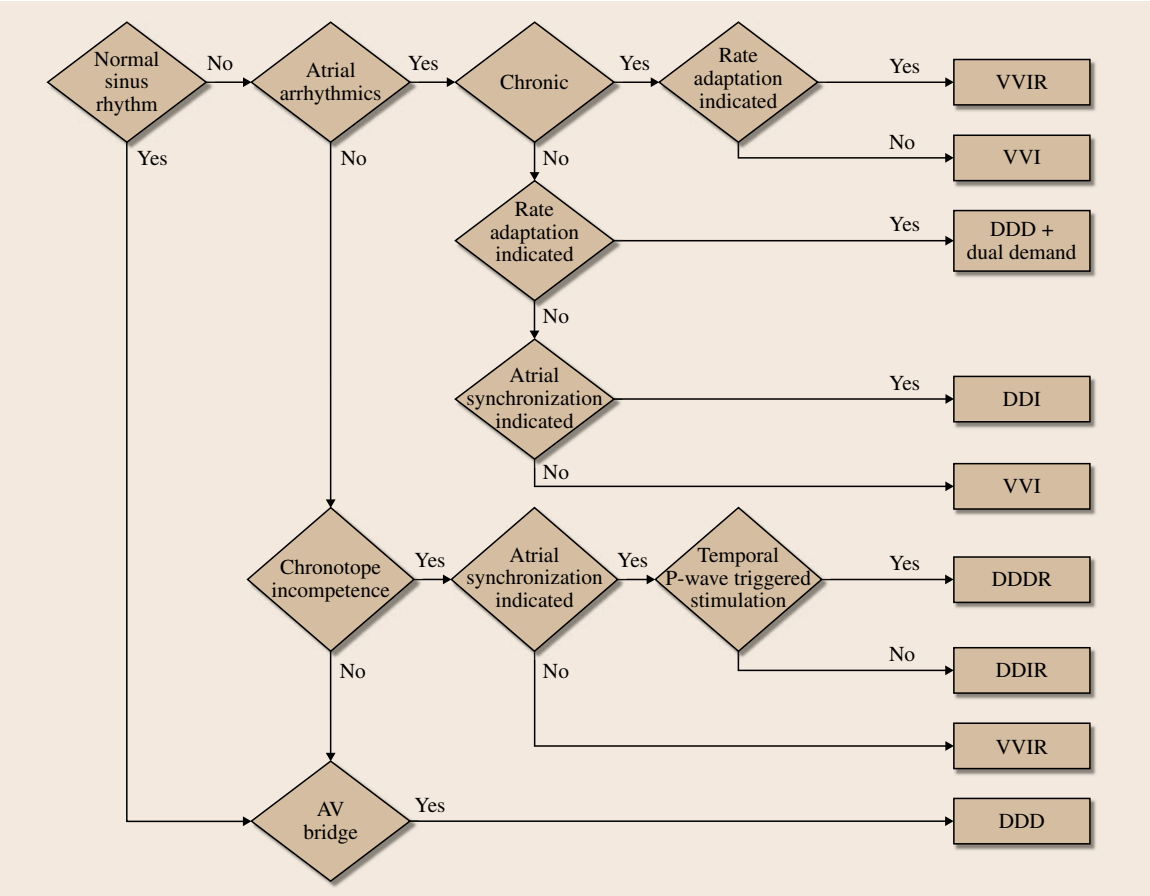

Fig. 38.17 Semischematic diagram of biventricular stimulation. Stimulation electrodes are located in the right atrium, right ventricle, and in a cardiac vein (V. cordis), which runs laterally over the left ventricle (LA left atrium, RA right atrium, LV left ventricle, RV right ventricle, V. cordis)

suitable electrode systems. The state of the art for stimulating the left atrium is access via the sinus coronarius. This way, the implant remains in the low-pressure system. This puts high demands on the flexibility of the electrodes and the skills of the implant surgeon, however.

38.3.7 Biventricular Stimulation Systems

The objective of biventricular stimulation is the treatment of serious myogenic cardiac insufficiency, independent of the cause that led to damage of the heart muscle. Although optimizing the heart rate and atrioventricular synchronization have already had favorable effects on patients with cardiac insufficiency these methods are often unsuccessful in achieving sufficient hemodynamic improvements. This is especially the case for patients with a pronounced left bundle-branch block, which leads to a desynchronization of the contraction of the right and left ventricles [38.11].

The technical basis of biventricular stimulation is an atrium-controlled ventricular pacemaker (DDD system), which has an additional second ventricular output for detection in the left ventricle. Here, the left ventricular electrode is placed in a cardiac vein in the left ventricle. The access path is the same as with biatrial stimulation, via the sinus coronarius (Fig. 38.17).

This method, which is still very new, has already been introduced in clinical practice. Its long-term success, however, can only be verified in broader studies [38.12]. Improvements in the hemodynamic situation are to be proven with this special stimulation method. In further development, improvement in the electrode technology is necessary, and suitable parameters for patient selection and for optimally programming the pacemaker system are to be selected. The combination with an implantable cardioverter/defibrillator is possible.

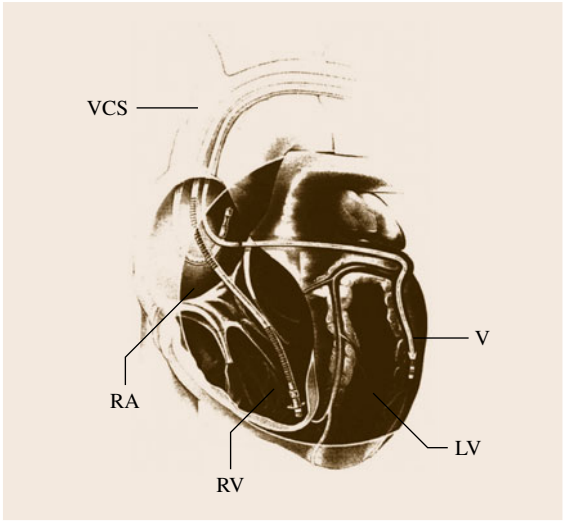

Fig. 38.18 Selecting a suitable pacemaker system (after 38.4)

38.3.8 Selection Criteria

The correct selection of a suitable pacemaker system still requires expert knowledge today, despite all the automatic functions of the pacemaker, since an exact diagnosis is often determined by essential details. Figure 38.18 attempts, however, to formalize the selection, to at least provide a rough starting point.

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