Successful mapping and ablation in the electrophysiology (EP) laboratory is critically dependent on acquiring multiple, low-amplitude, intracardiac signals in the presence of numerous sources of electric noise and interference and displaying these signals in an uncomplicated and clinically relevant fashion, with minimal artifact. This represents a significant engineering challenge and, in a real-life EP laboratory, is not always successful. Understanding the challenges and reasons for failure and knowing how to optimize the electronic data acquisition system in the EP laboratory can go a long way toward ensuring smooth procedures and good outcomes.

In this 2-part review, we will present information on the layout of a typical EP laboratory, with emphasis on common approaches to amplification and filtering of ECG and intracardiac signals, sources of noise and interference with techniques to minimize them, and also discuss in detail the subtleties of unipolar/bipolar signals and their clinical relevance.

In all of the electrogram illustrations, the following abbreviations are used:

- P1 ART indicates arterial pressure tracing; II and V1, surface ECG leads; RVA and RVAd, proximal and distal right ventricular apex signals; HRA and HRAd, high right atrial catheter proximal and distal signals; His4 to His1, His bundle signals from an octapolar catheter, with His1 being most distal; Lasso 1,2 to Lasso 10,1, circular catheter signals; ABL and ABLd, proximal and distal ablation catheter signals; and CS 19,20 to CS 1,2, coronary sinus catheter signals, with CS 1,2 being most distal.

**High-Yield Electronics**

The small voltages and currents that constitute biopotentials must be manipulated and presented in an easily understandable manner to the electrophysiologist. This involves amplification and filtering of the signals. Signals of interest may be manipulated in the form of voltage or current. A limited but effective analogy is to consider voltage as being the equivalent of water pressure. If a tank containing water (tank 1) is placed a few feet above another tank (tank 2), tank 1 is considered to be at a higher potential (have a higher voltage) than tank 2. If the 2 tanks are not connected to each other, this situation will not change. However, if the 2 tanks are connected to each other through a pipe, water will begin to flow from tank 1 to tank 2. This flow of water constitutes the current in this “circuit.” The rate at which the water flows (the magnitude of the current) will depend on several factors, including the dimensions of the connecting pipe. The rate-limiting components of this circuit correspond to the resistance of the circuit or the opposition that the circuit offers to the flow of water (or current). Ohm’s law states that for a given temperature, the ratio of voltage to current in such a circuit is denoted by the resistance of the circuit. Capacitance is the ability of a component to store energy in an electric field (like having a rubber diaphragm inside the connecting pipe that flexes with change in pressure). Inductance is also the ability of a component to store energy, but in a magnetic field (like having a heavy paddle wheel inside the pipe that spins with the flow of water).

Amplification involves increasing the amplitude of a signal. Although we think of amplifiers as being electronic only, there are many other examples of amplification all around us. One is the internal combustion engine (a force amplifier), which amplifies the small force applied to a gas pedal to a large enough force to move a few tons of automobile. In electrophysiology, the small potential differences between electrodes on the chest wall or on a catheter are amplified by electronic amplifiers (usually voltage amplifiers) to a large enough voltage to be displayed on a monitor. The voltage gain (or amplification) of these amplifiers must be very large. For example, at a total gain of 5000, a 100-μV His bundle signal would be increased in amplitude to 500 mV. The amplification is usually achieved by using several stages of electronic circuits, which gradually increase the signal to the final amplitude. Because this signal is often contaminated with noise, some form of filtering is required. Electronic filters work very much like coffee filters, allowing the required information to pass through while restricting the flow of unwanted information.

Analog signals are continuous in time and can have any of an infinite number of values within a certain range of voltage or current. Digital signals, on the other hand, are discrete with respect to time and are a coded (numbered) representation of an analog signal. Once an analog signal is converted into a
digital signal (using an analog-to-digital converter), the numbers that constitute that digital signal can be manipulated by using digital circuits in a dizzying variety of ways to achieve amplification and filtering also.

The Problem

From an electromagnetic (EM) standpoint, the EP laboratory is an extraordinarily noisy environment. The patient is connected to multiple pieces of equipment including an ECG machine (and sometimes 2 independent ECG machines), pulse oximeter, external defibrillator, electroanatomic mapping system, and half-a-dozen intracardiac catheters. Each of these devices must meet patient safety standards for “leakage current,” which is defined as the total current from patient connections through the patient to earth.1 This current is required to be less than $10^{-9}$ A for each device, resulting in a total current of several tens of microamperes flowing through the patient to earth ground when multiple devices are connected, at a fundamental mains frequency of 50 or 60 Hz, with significant harmonics extending to several thousand Hertz. In general, most patients can tolerate this magnitude of leakage current without significant risk of inducing ventricular fibrillation (which depends on electrode contact surface area). However, this leakage current can interfere substantially with our ability to process extracardiac and intracardiac signals with minimal artifact. Furthermore, the patient acts as an antenna, being both capacitively and inductively coupled to the extensive mains voltage wiring in the EP laboratory (120–240 V AC) and also picks up a substantial amount of radiofrequency (RF) noise from wireless headsets, mobile phones, and wireless monitors in proximity to the EP laboratory. As a result, there can be measured voltages of 1–3 V RMS (root mean squared) on the patient’s body, encompassing a wide frequency spectrum from 50 Hz to several tens of megahertz. In this setting, the required intracardiac electrograms, which usually range in amplitude from 25 µV (as measured in infarcted regions during ventricular tachycardia mapping) to 5 mV (from a surface ECG lead), must be amplified and displayed, with no loss of detail and minimal added noise, while delivering RF ablation energy (at about 500 KHz) with amplitudes of up to 70 V RMS (50 W maximum into a 100 Ω impedance). Although a very high signal-to-noise ratio (SNR) is always desirable in an electronic system (with the signal being 1000, 10 000, or even 100 000 times bigger than the noise), this is not always achievable. For the smallest signals (25 µV), an SNR of 10 may be sufficient to distinguish signal from noise. This corresponds to a total noise at the input of the amplification system of 2.5 µV in the bandwidth of interest.

This is a sizable challenge, but one that is solved with attention to detail and good electronic design. Figure 1 is a block diagram of typical connections to the patient in the EP laboratory and the various sources of interference. Although the words “noise” and “interference” are used interchangeably in many discussions and refer to unwanted signals, we will use “noise” to describe sources intrinsic to the electronic system and “interference” to describe extrinsic sources. The capacitance connecting the patient to ground (caused by the proximity of the patient to various grounded objects in the environment) offers a path for leakage current to flow through the patient to ground from the mains-connected devices attached to the patient.

The Solution

Clearly, the best approach to minimizing interference in the EP laboratory is to minimize or eliminate sources of electric interference before attempting to deal with them with elec-
tronic processing. In general, any data acquisition system is always better off with the least amount of signal processing because this tends to add its own artifact. This will be discussed in detail later. Important methods used to combat interference in the EP laboratory include shielding of cables, appropriate grounding of equipment, balancing signal inputs and outputs, filtering, electric isolation, physical separation and perpendicular orientation, lowering circuit impedances, and frequency- or time-domain–based cancellation or enhancement techniques. The following section provides more information on these approaches.

Practical Points for the Electrophysiologist
It is always better to decrease interference rather than use signal processing to remove artifacts.

Shielding and Grounding
Line frequency interference may make its way into the signal processing system through electric or magnetic coupling. The EP laboratory itself is usually equipped with extensive shielding in the walls, floor, and ceiling to minimize leakage of x-rays from the EP laboratory to the outside. This, in turn, goes a long way toward minimizing leakage of RF interference into the EP laboratory. The power sources to the laboratory (120–240 V AC at 50/60 Hz) must be routed carefully away from the patient table and, most importantly, must be maintained with maximum possible separation from the cables carrying sensitive intracardiac signals. Where the 2 types of signals do come unavoidably close to each other, interference can be reduced by minimizing the length of cables running parallel to each other (as this increases coupling and cross-talk) and making an effort to have such groups of cables running perpendicular to each other (to optimize rejection of coupled signals by the amplifiers). Minimizing coiling of cables reduces magnetic interference by reducing coupling between cables. Unfortunately, for esthetic reasons, cables are usually carefully bundled together and tightly wrapped, and this essentially guarantees some cross-talk and interference in the EP laboratory. Cable shielding (with high-density braiding) is also critical because it helps to shunt capacitively and inductively coupled interference to the ground reference of the electronic system. Special nickel-iron alloys such as Mumetal may also be used to shield sensitive circuits exposed to magnetic interference. One disadvantage of biomedical systems with patient connections is that the electronic circuit ground reference (to which the cable shields are usually attached) must “float” with respect to earth ground to meet the leakage current requirements for patient safety. Although this “floating ground” reduces the efficacy of cable shielding, shielding is still extremely important. One quick and easy troubleshooting tip when faced with interference on a catheter is to replace the interconnect cable between catheter and connector block (junction box) because faulty shield connections are a common cause of noise/interference.

Practical Points for the Electrophysiologist
When troubleshooting interference, consider replacing the interconnect cable between the catheter and the junction box. Preamplifiers and amplifiers must also be placed very close to the signal source (the patient) to minimize cable length. If cables need to be long, optical transmission of signals (through fiberoptic cables) with subsequent conversion to electric signals may also be used successfully.

Isolation
Electric isolation of patient connections is a mandatory safety requirement and also helps reduce circulating currents at 50/60 Hz, which are a major source of interference in the EP laboratory. The 10-µA “leakage current” limit usually specified by most safety standards committees results in an effective coupling capacitance (at 120 V/60 Hz) of 220 picofarads (pF) to earth ground. Equipment designers attempt to isolate patient-connected instruments with coupling capacitances of less than 100 pF to provide adequate safety margins, further reducing circulating currents at line-frequencies. The EP laboratory staff does not have control over these parameters but can minimize the effects of interference between pieces of sensitive equipment by deliberately plugging them into wall sockets spaced far apart, although this approach can backfire sometimes and actually increase interference, requiring an empirical trial of various wall sockets in the room. Detaching and turning off unused equipment is also a worthwhile approach to keeping signals clean in the EP laboratory.

Balancing and Differential Amplification
Another important approach to minimizing coupled interference into the sensitive amplifier circuits of the signal acquisition system is balancing and differential amplification. A differential amplifier or an instrumentation amplifier (a more sophisticated version of a differential amplifier) is a pivotal component of any analog signal processing circuit (Figure 2). Its design allows amplification of the difference between the signals presented to its 2 inputs (usually connected to 2 adjacent electrodes on an intracardiac catheter or to 2 wires constituting a “lead” attached to surface ECG electrodes), while not amplifying any signal common to both inputs. An important property of such an amplifier is its common-mode rejection ratio (CMRR), which is a measure of its ability to reject signals common to both inputs while simultaneously amplifying the difference. Modern instrumentation amplifiers exhibit excellent CMRR (about 100 dB in practice) at low frequencies (Figure 2, graph), although this number decreases substantially at higher frequencies. What this CMRR of 100 dB implies in a data acquisition system is that 1 V of interference at 60 Hz, which is common to both inputs of the signal acquisition system, will be attenuated to 10 µV (100 dB or 100 000 times smaller than the original signal), whereas the required ECG or intracardiac signal is simultaneously amplified by a factor of 100 or more (as selected by the circuit design engineers). This greatly enhances the signal-to-interference ratio. Using a tightly twisted pair of wires to deliver a bipolar signal to the input of such an amplifier maximizes the chances of coupling equal amounts of interference to both inputs, allowing the amplifier’s CMRR to reject the interference more effectively. “Balancing” the inputs involves maintaining similar impedances between each
input and circuit ground. Any imbalance on the 2 inputs will have a significantly deleterious effect on the CMRR, increasing perceived interference. This is seen readily in the EP laboratory when ablating with RF energy, while attempting to observe electrograms on the distal pair of electrodes on the ablation catheter. If the pacing function is enabled on the distal pair of electrodes while ablating, the noise on the distal electrogram is substantial due to the slight imbalance produced by attaching the pacing current source to the 2 electrodes (even if no pacing current is being delivered). Turning off the pace function restores electrogram appearance (after a short delay caused by settling of filters) during RF delivery by restoring the balance to the 2 inputs (Figure 3).

This is especially important when trying to reject higher frequency interference. Lowering the input impedance of the instrumentation amplifier also helps to reduce interference by reducing coupled signals. There is a lower limit to this approach because very low input impedances will also reduce the amplitude of required signals, lowering the overall SNR. In general, the input impedance of any voltage amplification system should be at least 10 times higher than the impedance of the signal source that is attached to it to ensure adequate transfer of signals.

**Filtering**

Filtering is an effective way to enhance certain portions of the frequency spectrum while rejecting unwanted portions of the

---

**Figure 2.** Typical instrumentation amplifier with twisted-shielded input cables that may be connected to catheter terminals or ECG electrodes. Amplifier A3 amplifies the difference between the 2 signals, A and B. When amplifying ECG signals, A and B may also be summed and inverted with a separate amplifier and used to drive the right leg electrode to minimize interference (not shown). Also shown is a graph for common mode rejection ratio (CMRR). Note the decrease in common mode rejection at higher frequencies resulting in reduced interference at low frequencies, but worsening noise rejection at radiofrequencies used for ablation.

**Figure 3.** Noise on the ABLd electrode during radiofrequency ablation with (A) and without (B) the pacing function enabled on ABLd. The noise goes down substantially when the pacing function is disabled, eliminating the imbalance on the ABLd electrodes and improving noise rejection. The baseline noise on the CS 3,4 electrode in the lower image (C) is also eliminated when the pacing function on that electrode pair is disabled.
spectrum. Filtering may be performed at several stages in the electronic system and includes various combinations of passive (using only resistors, capacitors, and inductors) and active (also using operational amplifiers, which are the building blocks of most analog signal processing systems) filters in conjunction with digital filters (which use a combination of digital signal processing hardware and software algorithms to accomplish the same action).

Diagnostic-quality ECG signals typically require a processing bandwidth of 0.05–100 Hz, whereas monitor-quality ECGs may be limited to 0.5–40 Hz. The advantage of the lower frequency cutoff of 0.05 Hz is better reproduction of the ST segments, with one tradeoff being more baseline drift. At an upper cutoff frequency of 100 Hz, rapid changes in the QRS complex are reproduced well even at high heart rates, but there is increased susceptibility to interference at the power line frequency of 50/60 Hz. Almost all filters in electronic systems in the EP laboratory achieve a band-pass response (allowing a certain band of frequencies through the filter while rejecting frequencies on either side), by using a combination of high-pass filters (HPF) and low-pass filters (LPF). For example, an HPF with a cutoff frequency of 0.05 Hz will reject or attenuate signals with frequencies below 0.05 Hz and so will block DC voltages (which, by definition, have a frequency of 0 Hz) but pass all frequencies above 0.05 Hz. Conversely, an LPF with a cutoff frequency of 100 Hz will reject or attenuate all frequencies above 100 Hz while passing all frequencies from DC up to 100 Hz. This system of filters is analogous to the filters used in a “crossover network” in a home-stereo system, where the output of an LPF is directed to the bass loudspeaker and the output of the HPF feeds the tweeter (high-frequency loudspeaker). In sound systems with a separate midrange speaker, a band-pass filter may be used to selectively drive this speaker. Adjusting the upper and lower cutoff frequencies of the filters in the EP laboratory is also analogous to using a tone control in a home stereo system to emphasize bass or treble frequencies. Current data acquisition systems also using digital filtering (performed mathematically using digital signal processing techniques) to compensate for deficiencies in traditional hardware filters and so can use cutoff frequencies different than mentioned above, with no loss of signal quality.

By cascading the HPF with an LPF (sequentially passing the amplified ECG voltage through both filters), one achieves a band-pass response from 0.05–Hz (Figure 4). In a diagnostic-quality ECG system, 50- or 60-Hz interference could be a significant problem in the pass-band, and most systems also provide an optional notch filter to reject 50 or 60 Hz. Such a notch (or band-reject) filter passes all frequencies on either side of a narrow band of frequencies centered on 50 or 60 Hz, effectively suppressing line-related noise. Depending on their design, these filters may also add noise into the system (usually at high frequencies). Another significant disadvantage of turning on the notch filter is that certain EP signals, such as near-field pulmonary vein deflections, or rapidly varying fractionated potentials, can be attenuated substan-
tially, resulting in wasted effort at locating a good signal, despite appropriate catheter position or effective ablation (Figure 5).

**Practical Points for the Electrophysiologist**

High-frequency signals including pulmonary vein potentials and near-field fractionated signals may be inadvertently filtered out with the notch filter. Intracardiac electrograms require a completely different pass-band from ECG signals because of differences in frequency content. Good rejection of baseline drift is important, and a lower cutoff frequency of 30 Hz is commonly used (Figure 6). Although the fundamental frequency being amplified in such a system is quite low (only 5 Hz at a heart rate of 300 beats per minute), individual electrograms (such as His bundle electrograms, pulmonary vein potentials, and Purkinje potentials) can have components with very rapid deflection, indicating more energy content at higher frequencies (also usually implying near-field information). In most cases, an upper cutoff frequency of 500 Hz will suffice to process the signals with minimum distortion (Figure 7). Using a higher cutoff frequency than this (1000–2000 Hz) will improve

---

**Figure 5.** (A) Pulmonary vein potentials on Lasso catheter with and without 50/60 Hz notch. Note the significant reduction in signal amplitude with “smearing” of the signals, mimicking far-field signals, when the notch filter is active. In (B), fractionated potentials on ABLd lose a lot of the fractionation slowly as the notch filter is turned on.

**Figure 6.** Baseline drift reduction on the His electrodes with high-pass filter cutoff increased from 0.05 Hz (A) to 100 Hz (B) (shown to deliberately emphasize effect of filter cutoff).
accuracy by minimizing phase shifts (different signal delays in different parts of the frequency spectrum) but will do so at the expense of an overall increase in noise and interference because the intrinsic noise of an electronic system increases by the square root of bandwidth. Also, distinguishing near-field signals, which have more high-frequency content, from far-field signals, characterized by low rate-of-change information content, can be compromised with lower cutoff frequencies for the LPF. Selecting appropriate filters is further complicated by whether one is analyzing bipolar or unipolar signals. This will be covered in more detail in a subsequent section.

When using an electroanatomic mapping system such as Carto 3 (Johnson & Johnson, New Brunswick, NJ), most catheter signals are routed through the mapping system, and filter selection also must be performed in the mapping system (in addition to the filter setting selection in the signal acquisition system). To optimize signal acquisition while rejecting noise, the manufacturer of this system suggests a range of 16–500 Hz for intracardiac electrograms. These cutoffs may be adjusted in the Carto 3 system individually for each set of electrograms.

Overview of a Typical Data Acquisition System
A block diagram of a typical data acquisition system in the EP laboratory is shown in Figure 8. The individual ECG (chest) wires and catheters are connected to the terminal block at the end of the patient table. From here, they are routed to the various input amplifiers of the data acquisition system. A few particulars of ECG processing will be described next.

Figure 7. Amplitude change with different low-pass filter settings on His1 (150 Hz vs 1000 Hz). Note the significant change in near-field amplitude of the His and ventricular deflections at the 2 filter settings, with no appreciable change in the far-field (low-frequency) atrial signal amplitude on the same catheter.

The 3 limb connections (right arm [RA], left arm [LA], and left leg [LL]) are connected to a switching network through protection circuitry to prevent damage to the electronics during defibrillation or caused by static electricity. Also entering this circuit are the 6 precordial connections (V1 through V6). Each of these 9 wires is incapable of producing meaningful signals by themselves but can only do so when analyzed with reference to another wire (the voltage difference between 2 wires is a “lead”). For the 3 limb connections, true bipolar leads are obtained by referencing each of them to one of the other 2 (RA to LA is lead I, RA to LL is lead II, and LA to LL is lead III). The vector sum of lead I and lead III equals lead II (Einthoven’s law). Frank Wilson introduced a further 3 leads called Vr, Vl, and Vf to provide 3 intermediate vectors. Because of reduced amplitudes from the Wilson leads, Goldberger modified the vectors to provide additional voltage, producing the augmented leads aVR, aVL, and aVF, used in all modern 12-lead ECG systems. Both the Wilson and Goldberger lead systems need a reference electrode for the second wire and use the vector sum of the 3 limb wires to approximate a zero potential point, one that would ideally be seen at infinite distance from the whole system.
This reference, obtained originally by Wilson by summing the 3 signals $R_A$, $L_A$, and $L_L$ through 5000-$\Omega$ resistors, is referred to as Wilson Central Terminal (WCT) (Figure 9). The resistors were used to minimize the contribution of skin resistance changes to the final potential at the junction of the 3 wires (WCT). Modern instrumentation amplifiers have very high input impedances, with better shielding and noise rejection, and so the summing resistors in current systems can be much higher, with no detrimental effect. The theoretical summed current at WCT is zero. Estimates of the actual potential at WCT have been made with respect to an infinite reference using body surface potential data, and they are close enough to each other to be considered equivalent.$^7$ WCT is also used as the reference for unipolar intracardiac signals, and this will be discussed in detail later. Several approximations exist when acquiring unipolar signals. In practice, they represent widely spaced bipolar signals, using a theoretical reference electrode—the WCT.

The 12 leads thus derived are sequentially switched rapidly to produce a snapshot of each lead, to be sent for final display. The electric difference of each wire pair (lead) is amplified by an instrumentation amplifier with excellent noise and common-mode rejection characteristics (see section on balancing and differential amplification). A very important processing step at this stage is the right leg driver ($R_L$). Because the output of the instrumentation amplifier (the difference between the outputs of each wire pair) still has some noise at line-related frequencies, the outputs of each wire pair (points A and B in Figure 2) are also summed, inverted, and fed back to the patient through a right leg electrode. This, in effect, cancels out the original interference components present on the patient’s body. The efficacy of this approach can be tested easily by disconnecting the $R_L$ wire and seeing the significant increase in baseline drift and low-frequency interference (Figure 9, ECG). This general concept of cancellation is very similar to the one used in active noise-canceling headphones. Pacing spikes are then separated from the main signal by using a slew detector circuit that sends the information on pacing timing to the system microprocessor, which then reinserts an artificial pacer spike at the correct time on to the composite ECG signal on the display. The main signal is filtered from 0.05 Hz to 100 Hz (diagnostic-quality ECG). An optional notch filter is provided at 50/60 Hz to further reject line frequency interference. This cleaned-up analog signal is converted to a digital signal, using a high-resolution analog-to-digital converter (ADC) with 12-to 16-bit resolution. Resolution is a measure of the ability of an electronic system to distinguish small differences in signal amplitude. It also allows calculation of the dynamic range of the system. A 12-bit ADC has $2^{12}$ (4096) discrete levels that it can assign to an instantaneous voltage, constituting a dynamic range of 72 dB (12 bits×6 decibels/bit). This means that if the maximum input signal to the complete system (at the ECG wires attached to the patient) is 10 mV, the ADC can theoretically codify information from that maximum value down to (10 mV÷4096) or 2.44 $\mu$V, which is adequate for most EP applications. In comparison, a standard CD player for music
reproduction uses a 16-bit ADC, producing a dynamic range of 96 dB and an ability to theoretically codify signals over a range from 1 V down to 15.25 μV. The actual numbers may be higher or lower than this for several reasons, and those details have been omitted in this discussion. The digital values 0 and 1, which constitute the coded equivalent of each signal, may be manipulated further and then stored or displayed.

An important parameter during analog-to-digital conversion is the sampling rate of the ADC. Claude Shannon and Harry Nyquist of Bell Laboratories formalized the development of information theory and have a theorem named in their honor.6,7 They and others discovered that analog signals must be sampled at least twice as fast as the fastest signal in the system. In other words, to completely capture the information content unequivocally at a particular frequency, one must be able to “look at the signal” at least twice per cycle. In the case of ECG signals filtered to 100 Hz, this suggests that sampling at 200 Hz would suffice. In practice, the ECG channels are sampled at 400 Hz or above. One problem that can occur at this stage is attempted conversion by the ADC of an analog signal that has a rate greater than one-half the sampling rate. This would result in aliasing, producing spurious information in the frequency spectrum. For example, because the filters in the data acquisition system do not completely reject signals outside their pass-band but only attenuate them substantially, one could conceive of a high-frequency signal (at 360 Hz, for instance, which is the 6th harmonic of the 60-Hz power line frequency) of sufficient amplitude, reaching the ADC. If this signal were sampled at 400 Hz, 2 new signals would be created (called sum-and-difference frequencies) at 760 Hz (400 + 360) and 40 Hz (400 – 360). The difference frequency of 40 Hz would be within the pass-band of the system and be displayed as noise. This is why great efforts must be made up front to prevent these sources of noise and interference from reaching the ADC in the first place. The aliasing described here is identical to that seen during pulsed wave Doppler echocardiography when trying to “ping” at rates below twice the rate of flow of a high-velocity regurgitant jet. This causes portions of the displayed signal to “fold back” on themselves and provide misleading information. A similar phenomenon is noted when watching wagon wheel spokes in a Western film rotate slowly backward, although the wagon is moving rapidly forward, because the rotation rate of the spokes on the wagon wheels exceeds half the sampling rate of the film (24 frames/s).

The electrogram signal processing works in very similar fashion to the ECG system described above. The main differences are the need for more voltage gain (up to 10000 times amplification), more channels (64–128 channels), different filter frequencies (usually 30–500 Hz, although the upper cutoff can be extended to 2000 Hz), and a much higher sampling frequency. Purkinje potentials, for example, are barely visible if sampled at rates lower than 625 Hz.10

Given the significant sources of noise and interference, the minuteness of the signals being amplified, and the inherent imprecision of localizing signals in the moving heart, great care must be taken during setup and interpretation of the EP study to avoid being mislead. Several sources of artifact will be discussed in part II of this series.

Additional Detail of Definitions
EM indicates electromagnetic, consisting of static and dynamic electric fields, usually capacitively coupled to the object under consideration, and magnetic fields, which are inductively coupled; RF, radiofrequency, referring to radiation frequencies in the electromagnetic spectrum ranging in wave length from 1 mm to 100 km, corresponding to frequencies from 300 GHz to 3 kHz; RMS, root-mean-squared, is a mathematical approach to calculate the effective heating capacity of a complex signal; pF, picofarads, a measure of capacitance, which is the ability of an object to store electrical charge; common-mode rejection ratio, CMRR, is the ability of an amplifier to reject any signal that is common to both its input terminals while amplifying the difference between the signals on its input terminals; signal-to-noise ratio, SNR, is an extremely important parameter in signal processing that defines the clarity with which signals can be measured or visualized. In general, a minimum SNR of a factor of 10 (20 dB) is required to distinguish small changes in signals in the presence of noise. Ideally, this number should be a factor of 1000 (60 dB) or more, but this is very hard to achieve in a real-life electrophysiology lab; high-pass filters, HPF, are selective signal processing circuits that allow signals above a certain frequency (the cutoff frequency) to be transmitted unchanged while simultaneously rejecting (or at least significantly reducing the amplitude of) signals below that cutoff frequency; low-pass filters, LPF, are selective signal processing circuits that allow signals below a certain frequency (the cutoff frequency) to be transmitted unchanged while rejecting (or significantly reducing the amplitude of) signals above that cutoff frequency.

Disclosures
None.

References

Key Words: ablation ■ electrophysiology ■ mapping ■ artifact ■ filter