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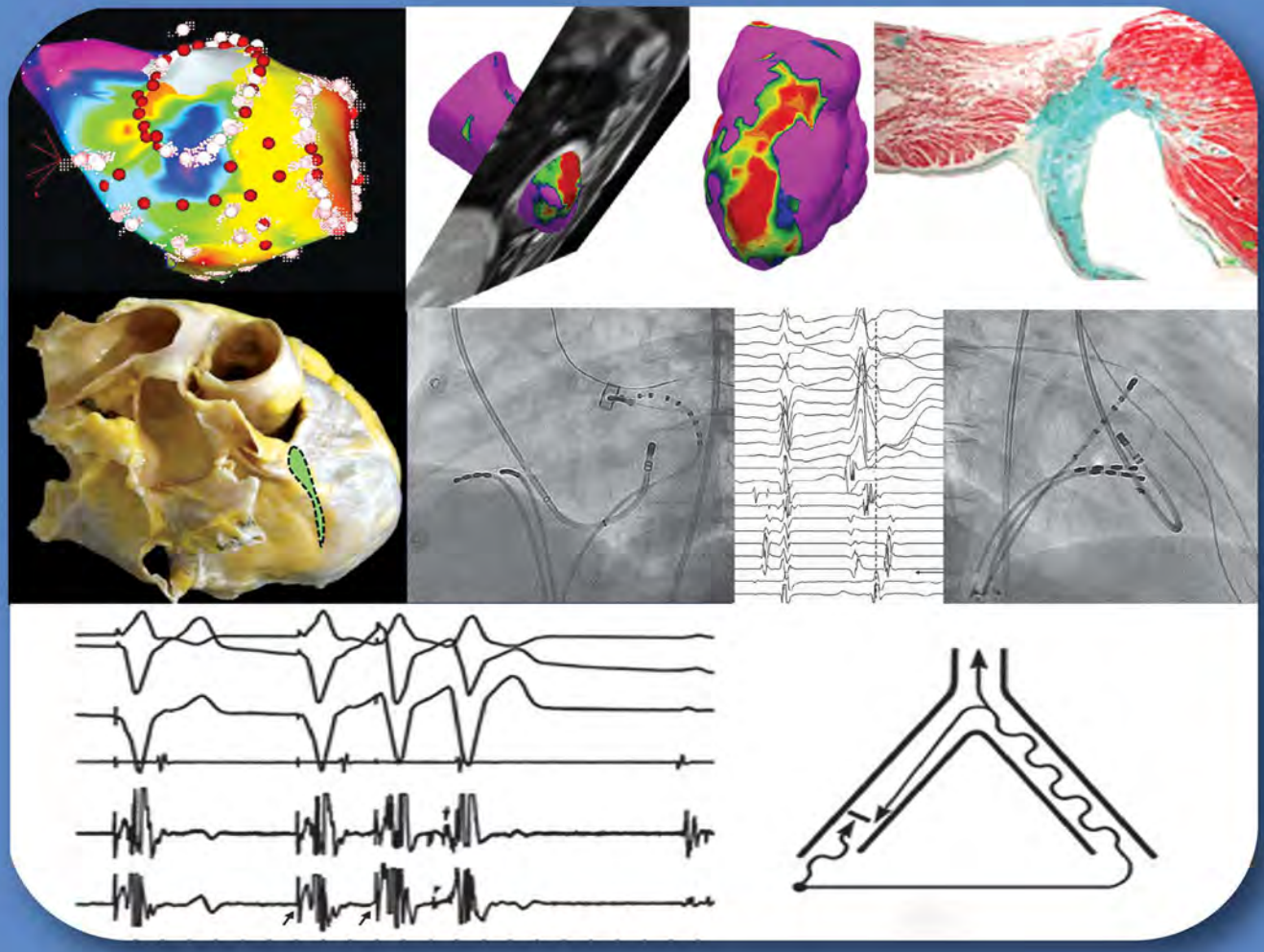
Practical Cardiac Electrophysiology



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CHAPTER 2

Katherine Duello, KL Venkatachalam

Electrophysiology Study: Technical Details, Electrograms, Noise and Filtering

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LIST OF ABBREVIATIONS

CMRR	Common-mode Rejection Ratio	HPF	High Pass Filter
DC	Direct Current	LPF	Low Pass Filter
DSP	Digital Signal Processing	RF	Radiofrequency
ECG	Electrocardiogram	RMS	Room Mean Squared
EP	Electrophysiology	WCT	Wilson Central Terminal

INTRODUCTION

Successful mapping and ablation in the electrophysiology (EP) laboratory is dependent on the acquisition of multiple, low amplitude intracardiac signals obtained amidst numerous sources of electrical noise and interference, presenting a significant engineering challenge. Knowledge of the challenges and ability to optimize the electronic data acquisition system can ensure that signals are presented in an uncomplicated and clinically relevant fashion with minimal artifact.

This chapter will present information on approaches to amplification and filtering of the electrocardiogram (ECG) and intracardiac signals in a typical EP lab, minimization of noise and interference, and clinical relevance of unipolar and bipolar signals. With this background we will discuss possible errors in mapping from artifact or inappropriate filtering.

In all of the electrogram illustrations, the following abbreviations are used:
P1 ART: arterial pressure tracing, II and V1: surface ECG leads, RVa and RVa d: proximal and distal right ventricular apex signals, HRA and HRA d: high right atrial catheter proximal and distal signals, His 4 to His 1: His bundle signals from an octapolar catheter with His 1 being most distal, Lasso 1,2 to Lasso 10,1: Circular catheter signals, ABL and ABL d: proximal and distal ablation catheter signals, CS 19,20 to CS1,2: coronary sinus catheter signals with CS1,2 being most distal.

DIFFICULTIES WITH SIGNAL ACQUISITION

The patient in the EP lab is connected to many pieces of equipment which can interfere with intracardiac signals; these may include an ECG machine, pulse oximeter, external defibrillator, intracardiac catheters and an electro-anatomic mapping system. Safety standards require the total current from patient connections via the patient to earth (the reference point), or "leakage current,"[1] to be less than 10 microamperes for each attached device. With multiple devices connected, there is a total leakage current of several tens of microamperes flowing through

the patient at a fundamental frequency of 50 or 60 Hz, with significant harmonics extending to several thousand hertz (Hz). While most patients safely tolerate this leakage current without significant risk of ventricular fibrillation, it can produce artifact on intracardiac signals. This artifact is exacerbated as the patient is coupled capacitively and inductively to the line voltage wiring in the EP lab (120–240V) and also picks up radiofrequency interference from wireless headsets, mobile phones and wireless monitors near the lab. The patient's body may have interference voltages of 1–3 V RMS (root mean squared, a mathematical approach to calculate the effective heating capacity of a complex signal), encompassing a wide frequency spectrum from 50 Hz to several tens of megahertz (MHz).

Signal to noise ratio (SNR) is an extremely important parameter in signal processing, which defines the clarity with which signals can be measured or visualized. In general, a minimum SNR of a factor of 10 (20 decibels) is required to distinguish small changes in signal in the presence of noise. Ideally, this number should be a factor of 1000 (60 decibels) or more, but this is very hard to achieve in a real-life EP lab. Intracardiac electrograms, typically ranging in amplitude from 25 microvolts to 5 millivolts, need to be accurately amplified and displayed ignoring any previously described interference, while delivering RF ablation energy (at about 500 KHz) with amplitudes of up to 100 V RMS (100 W maximum into a 100 ohm impedance). In this setting, a high SNR is not always possible. With smaller signals, such as 25 microvolts, a SNR of 10 may be sufficient, which corresponds to a total noise at the input of the amplification system of 2.5 microvolts in the bandwidth of interest.

Attention to detail in electronic design can overcome many of these challenges. **Figure 2.1** shows typical connections to the patient in the EP lab. "Noise" is used to describe sources intrinsic to the electronic system and "interference" to describe external sources. The capacitance (which is the ability of an object to store electrical charge) is caused by the proximity of the patient to grounded objects in the environment, and is a path for leakage current from the devices attached to the patient.

POSSIBLE APPROACHES TO ARTIFACT-FREE SIGNAL ACQUISITION

Ideally, interference in the EP lab is minimized by eliminating sources before electronic processing, as additional signal processing may add to a data-acquisition system's intrinsic artifact. Some methods used to decrease interference include shielding of cables, appropriate grounding of equipment, balancing signal inputs and outputs, filtering, electrical isolation, physical separation and perpendicular orientation, lowering circuit impedances and frequency- or time-domain-based cancellation or enhancement techniques.[1] The following discussion will address some of these methods.[2]

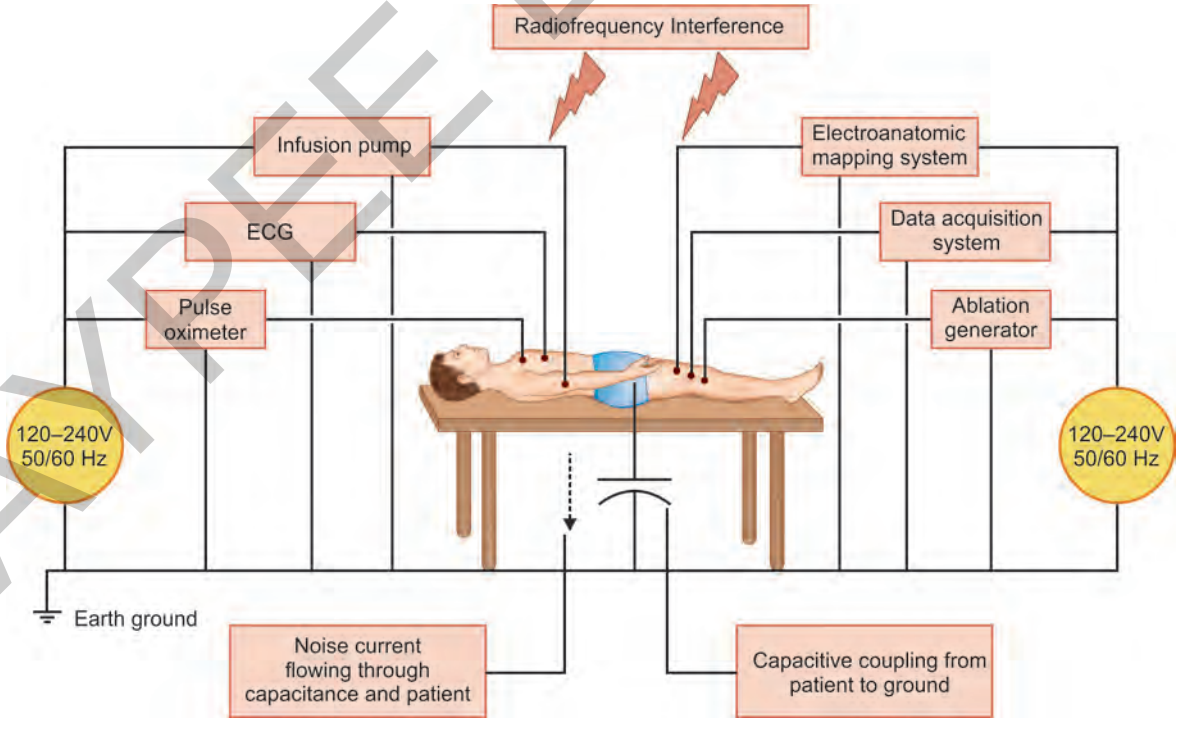

 Figure 2.1: Typical sources of electrical interference in EP lab

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Shielding and Grounding

Through electrical or magnetic coupling, line frequency interference may affect the signal processing system. Extensive shielding surrounding the EP lab prevents X-rays from exiting the lab and potential outside electromagnetic fields from entering the lab. Lab power sources should be routed away from the patient and kept as far away as possible from the cables carrying sensitive intracardiac signals. If in close proximity, minimizing the length of cables running parallel (which increases coupling and crosstalk), running groups of cables perpendicular to each other (optimizing rejection of coupled signals by the amplifiers) and minimizing coiling of cables (decreasing coupling) will decrease interference. Bundling cables together, as is frequently done for esthetic purposes, will increase crosstalk and interference. High-density braiding provides cable shielding that 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. To meet the leakage current requirements for patient safety, the electronic circuit ground reference (to which the cable shields are usually attached) needs to "float" with respect to the earth, which may also reduce the efficacy of cable shielding. Faulty shield connections are a common cause of interference, and replacing the interconnect cable between the catheter and connector block (junction box) may improve the signal. To minimize cable length, preamplifiers and amplifiers should be placed close to the patient, and may be replaced with fiber optic cables with subsequent conversion to electrical signals if needed.

Isolation

It is a mandatory safety requirement to electrically isolate patient connected equipment from the wall outlet. This has the additional effect of reducing circulating current at 50/60 Hz, which may cause interference. Most safety standards committees have a 10 microamperes "leakage current" limit that results in an effective coupling capacitance at 120 V/60 Hz of 220 pF (picofarads, a measure of capacitance) to earth ground. Equipment is usually designed to isolate instruments connected to the patient with coupling capacitances of less than 100 pF, reducing circulating currents at line frequencies.

To minimize the effects of interference between equipment, the EP lab staff can try to plug each of them into wall sockets as far apart as possible, but even this can cause interference occasionally. The signal quality may be improved with empiric trial of various sockets in the room. Keeping unused equipment off and unplugged from wall sockets also can minimize interference.

Balancing and Differential Amplification

Both balancing and differential amplification are additional approaches to minimizing interference.[3] Differential amplifiers possess the ability to reject signals common to both their inputs while also amplifying the difference. This property is called common-mode rejection ratio (CMRR). Typically, two adjacent electrodes on an intracardiac catheter are connected to the two inputs of the differential amplifier. Most amplifiers have CMRR around 100 dB (decibels) at low frequencies which decreases at higher frequencies. This means that in a data acquisition system with 1 V of interference at 60 Hz, the interference will be attenuated to 10 microvolts (100 dB or 100,000 times smaller than the original signal). At the same time, the required intracardiac signal is amplified by a factor of 100 or more depending on the circuit design. This will enhance the signal-to interference ratio. Using a tightly twisted pair of wires to deliver a bipolar signal as input to such an amplifier increases the chances of coupling equal amounts of interference to both inputs. This allows the amplifier's CMRR to effectively reject the interference.

"Balancing" the inputs requires maintaining similar impedances between each input and the circuit ground. Imbalances between the two inputs will affect the CMRR causing increased interference. This can be seen in the EP lab during ablation with RF energy while trying to observe electrograms on the distal pair of electrodes. If the pacing function on the distal pair of electrodes is enabled during ablation, noise on the distal electrogram is often noted due to the minor imbalance produced by attaching the pacing current source to the two electrodes, even if no pacing energy is being delivered. In this case, the electrogram appearance will be restored by turning off the pacing function. By lowering the input impedance of the differential amplifier, the interference may be reduced by decreasing coupled signals. However, very low input impedances will also reduce the amplitude of required signals, lowering the overall SNR. To ensure adequate transfer of signals, the input impedance of any voltage amplification system should be generally at least 10 times higher than the impedance of the signal source it is attached to.

Filtering

Filtering enhances portions of the frequency spectrum and rejects unwanted portions. It can be utilized at several stages in the electronic system including combinations of passive (only resistors, capacitors and inductors) and active (operational amplifiers) filters along with digital filters using digital signal processing (DSP) hardware and software algorithms.

ECG signals of diagnostic quality require a processing bandwidth of 0.05 Hz–100 Hz, while monitor quality ECGs

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are limited to 0.5 Hz–40 Hz. The lower frequency cutoff of 0.05 Hz results in better reproduction of the ST segments, but also results in increased baseline drift. With an upper cutoff frequency of 100 Hz, any rapid changes in QRS complexes are well represented with the tradeoff of increased susceptibility to interference at the power line frequency of 50/60 Hz. Using a combination of high- and low-pass filters (HPF and LPF), most filters in electronic systems in the EP lab achieve a band-pass response that allows certain bands of frequencies through the filter and rejects frequencies on either side. For example, a HPF with a cut off frequency of 0.05 Hz will block DC voltages (which have a frequency of 0 Hz), but pass any frequencies above 0.05 Hz. An LPF with a cutoff of 100 Hz will reject all frequencies above 100 Hz and pass all from DC up to 100 Hz. Adjusting the cutoff frequencies of the filters in the EP lab is analogous to using a tone control in a home stereo system to accentuate bass or treble frequencies. In the EP lab, data acquisition systems may also utilize digital filtering to improve deficiencies in traditional hardware filters.

By passing the amplified ECG voltage through both the HPF and LPF sequentially, a band-pass response from 0.05 Hz to 100 Hz is achieved **(Figure 2.2)**. If a diagnostic quality ECG is to be obtained, interference at 50 to 60 Hz could be a problem in this passband and most systems provide an optional notch filter to reject such interference. This notch (or band-reject) filter suppresses line-related noise by passing all frequencies on either side of a narrow band of frequencies centered on 50 or 60 Hz.

Depending on their design, these filters may also add noise into the system (usually at high frequencies). However, another significant disadvantage of the notch filter is that certain EP signals, such as near-field pulmonary vein deflections or rapidly varying fractionated potentials, can also be attenuated making it difficult to locate a good signal even with appropriate catheter position or effective ablation **(Figures 2.3A and B)**.

Due to differences in frequency content, intracardiac electrograms require a different passband from ECG signals. A lower cutoff frequency of 30 Hz is commonly used to ensure good rejection of baseline drift **(Figures 2.4A and B)**.

In such a system the fundamental frequency being amplified is 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 with more energy content at higher frequencies (also implying near-field information). An upper cutoff frequency of 500 Hz will typically reproduce signals with minimum distortion **(Figures 2.5A and B)**.

While a higher cutoff frequency (1000–2000 Hz) may improve accuracy by minimizing phase shifts (different signal delays in various parts of the frequency spectrum), it may also increase the overall noise and interference since the intrinsic noise of an electronic system increases by the square root of bandwidth (frequency range of the system). Distinguishing near-field signals with higher frequency energy content from far-field signals with low frequency content can be compromised with lower cutoff frequencies for the LPF. Selection of appropriate filters is further complicated when analyzing unipolar versus bipolar signals.

As most catheter signals are routed through the mapping system, filter selection also must be performed within the

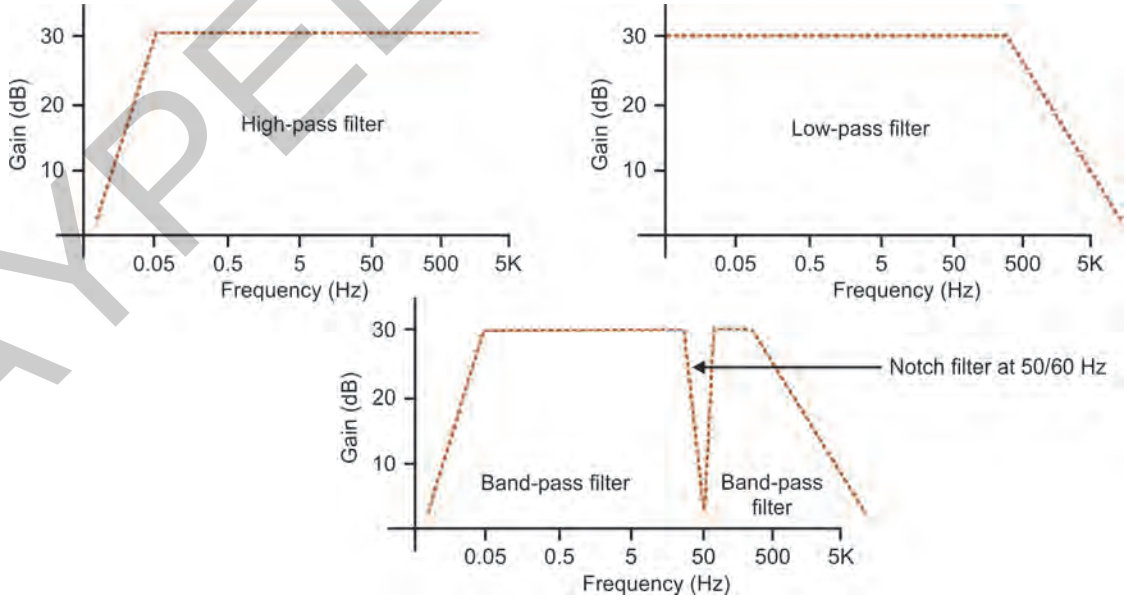

Figure 2.2: Combining a high-pass filter, low-pass filter and a notch filter to produce the composite frequency response

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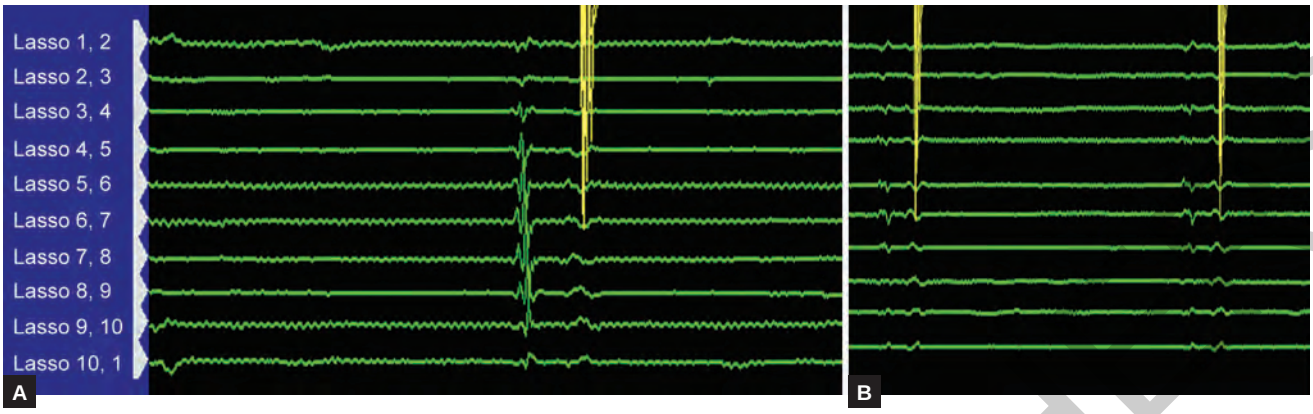
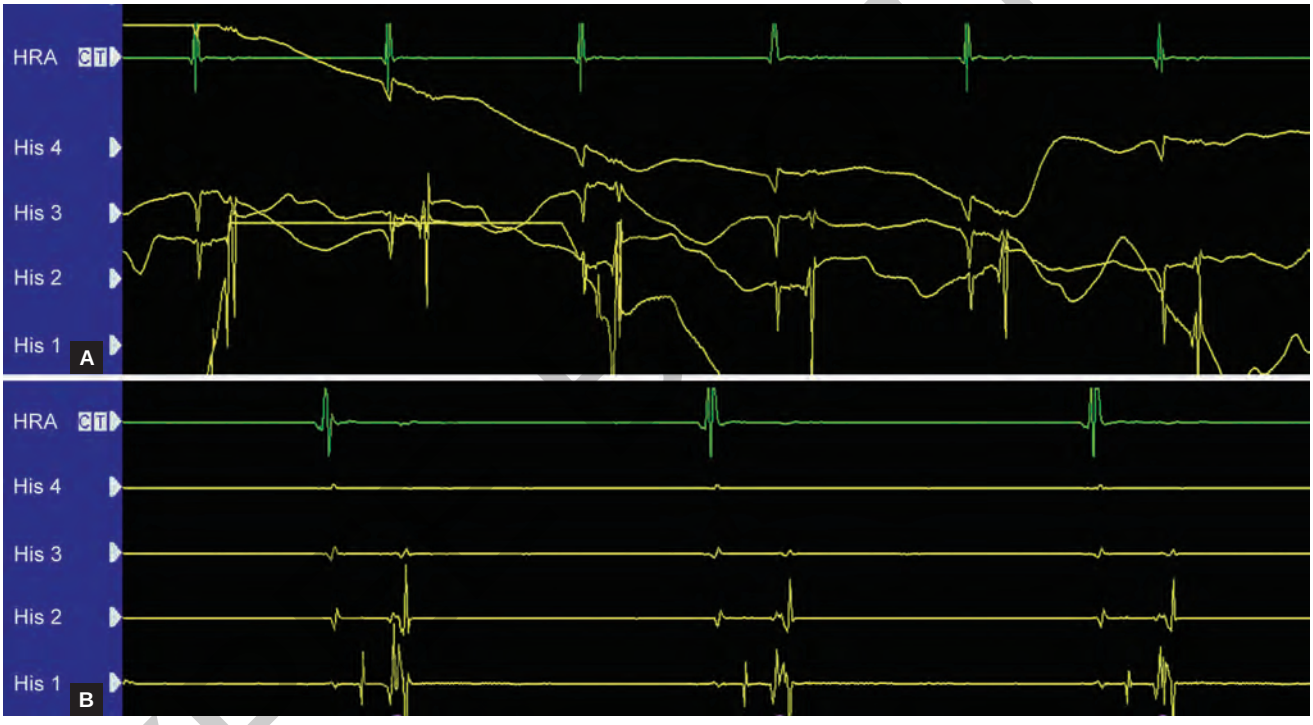


Figures 2.3A and B: Pulmonary vein potentials on a lasso catheter (A) are greatly attenuated when the 50/60 Hz notch filter is turned ON (B)



Figures 2.4A and B: Changing the high-pass filter cutoff frequency from 0.05 Hz (A) to 100 Hz (B) substantially reduce baseline drift in the His catheter signals

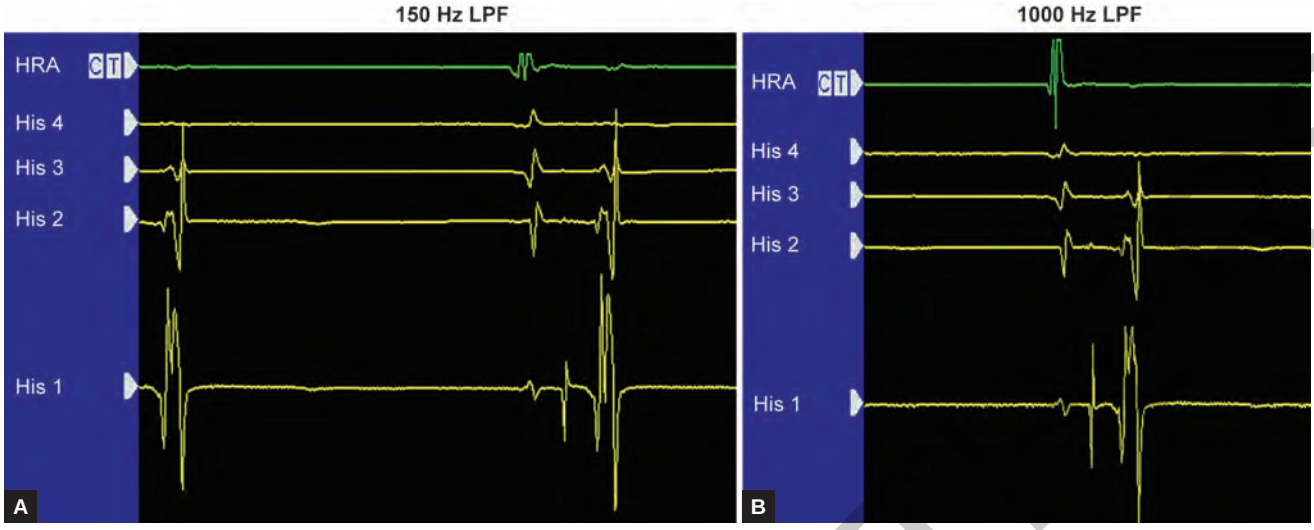
mapping system. This is in addition to the filter setting in the signal acquisition system. For optimization of signal acquisition with appropriate noise rejection, most mapping systems suggest specific frequency ranges for intracardiac electrograms.

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 has to be taken during set-up and interpretation of the EP study to avoid being misled. Several sources of artifact will be discussed next.

ARTIFACT

Saturation Artifact

Amplifier saturation can occur when an input signal is amplified at such a high gain that the amplified signal exceeds the power supply voltage of the electronic system. If a 2 mV peak signal is being amplified at a gain of 5000, the final output will be 10,000 mV or 10 V. However, if the system supply voltage is limited to 5 V, the overall signal amplitude would be limited to about 4.5 V (slightly below the



Figures 2.5A and B: The amplitude of the His deflection in His 1 is increased by increasing the low-pass filter cutoff frequency from 150 Hz (A) to 1000 Hz (B)

supply voltage). The signal would simply be cutoff above 4.5 V, although the amplifiers would attempt to drive themselves up to the required 10 V output unsuccessfully. The amplifiers remain "saturated" during this time and the peak of the signal remains unresponsive to changes in amplitude (flat top) until the signal amplitude decreases to below 4.5 V, when it can be tracked with fidelity once again. This is a form of distortion and can only be avoided by reducing the gain of the system (in this numerical example, to less than about 2000). When amplifiers saturate, they can take several milliseconds to recover, further delaying the time to faithful reproduction of signals. This is exacerbated with extremely large signals, such as pacing signals, which are typically 3 orders of magnitude larger than the electrograms. In this case, the signal recovery time for the paced electrodes as well as adjacent electrodes can be long enough that the artifact envelops a substantial portion of electrogram information immediately following the pacing spike. This can be a vexing problem when attempting to measure post-pacing intervals after entraining a tachycardia. Despite this artifact, the operator must make every effort to use the paced electrode signals for this measurement, as using more distant electrodes may produce artifactually longer postpacing intervals.

Clipping Artifact

One problem with using high-gain settings on multiple electrodes is that signals can overlap on the display. This is often circumvented by artificially "clipping" the signals of interest. The advantage of clipping (usually done at the software level although it can be done in hardware also) is that it can limit the numerical voltage displayed on-screen without the prolonged signal recovery times associated with amplifier saturation **(Figure 2.6)**. This allows the electrophysiologist to see a small electrogram that immediately follows a very large clipped electrogram without overlapping the displayed signals on adjacent channels.

However, the significant disadvantage of clipping is that one could get easily misled when looking at fused waveforms (pathway potentials during an accessory pathway ablation, for instance) if the selected clipping level by the software prevents visualization of two discrete, fused signals, and instead represents them as a single, wide signal **(Figure 2.7)**.

The operator usually cannot control the clipping level as this is programmed into the software in most systems. In this setting, one is almost always better off deliberately running the amplifiers at a low gain in order to prevent amplifier saturation and not require clipping (to avoid adjacent channel signal overlap). The only problem with this approach is that great care has to be taken to look at every nuance of the signals to not miss small potentials (such as pulmonary vein potentials or accessory pathway potentials), which are difficult to see at low gain settings, especially when the operator is several feet away from the display screen. In general, it is preferable to observe signals closely at low gain to minimize the deleterious effects of saturation and clipping.

Phantom Signals

Signals that appear real but are artifactual can also be seen due to the less than perfect properties of the input amplifiers, especially at higher frequencies.[1,3] An ideal differential amplifier would be able to reject equally coupled interference at both its inputs (such as from the distal and 2nd electrodes of a mapping catheter). At high-gain settings, even if a catheter is only partially in a sheath or is not even in the body, unequal coupling of large signals such as pacing spikes may occur. This can mimic real signals at

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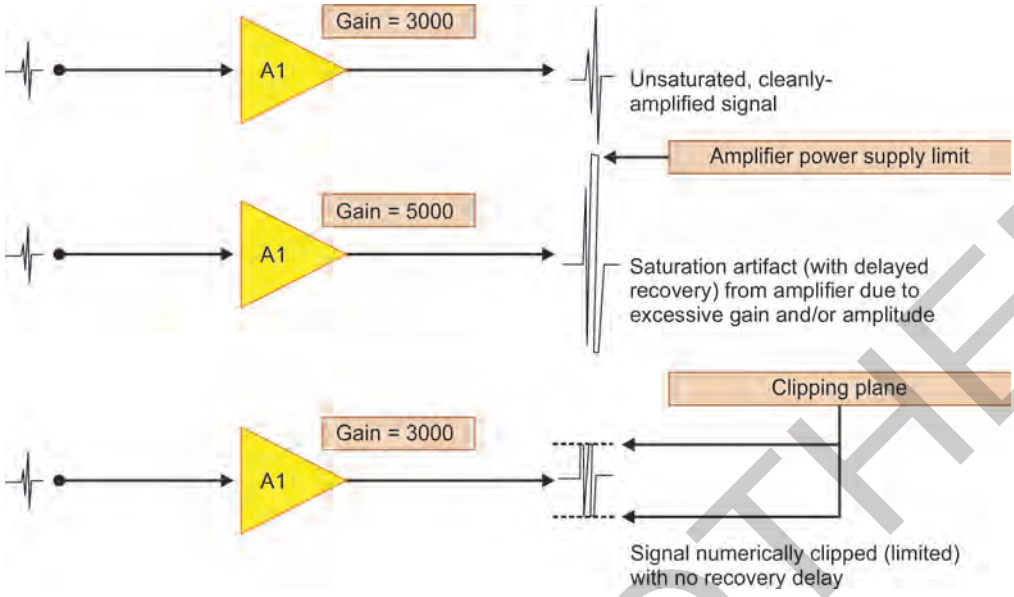

Figure 2.6: Saturation and clipping artifact

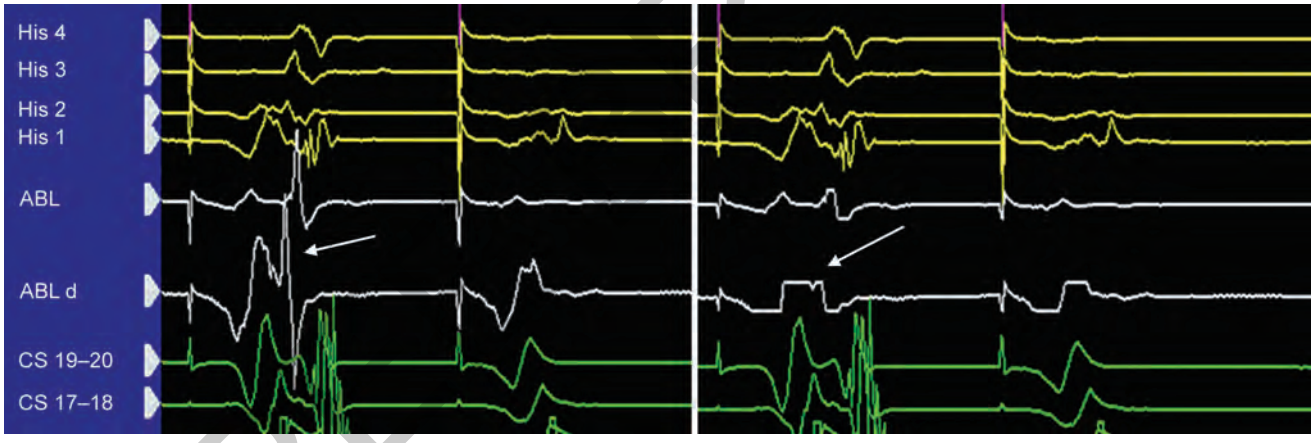

Figure 2.7: Clipping causing fusion of accessory pathway potential (on ABL d on right), compared to unclipped signal on left (arrow)

the pacing rate, especially after filtering smooths out the interference and makes it look like a far-field signal. This problem can be minimized by programming the output currents of the pacing stimulator to twice the diastolic threshold at a fairly narrow pulse width, confirming that all catheters are well extended out of the sheath tips and making good contact between tissue and tip. RF ablation noise can also be coupled into the system this way.

Electrode Contact Artifact

Intermittent contact between electrodes on different catheters or electrodes on the same catheter (such as a variable diameter lasso catheter with overlapping electrodes or contact between an ablation catheter and mapping catheter) can produce misleading signals. In the case of a lasso catheter, one could be misled into diagnosing pulmonary vein exit block due to noise on several overlapping electrodes that would produce sharp, near field signals reminiscent of pulmonary vein potentials, which do not get transmitted to the atrium. This may also be seen with a fixed diameter lasso catheter that has been pushed into a pulmonary vein ostium with consequent overlap of electrodes. Correlation of the location of these "potentials" with fluoroscopic electrode information reveals the true nature of these signals **(Figure 2.8)**.

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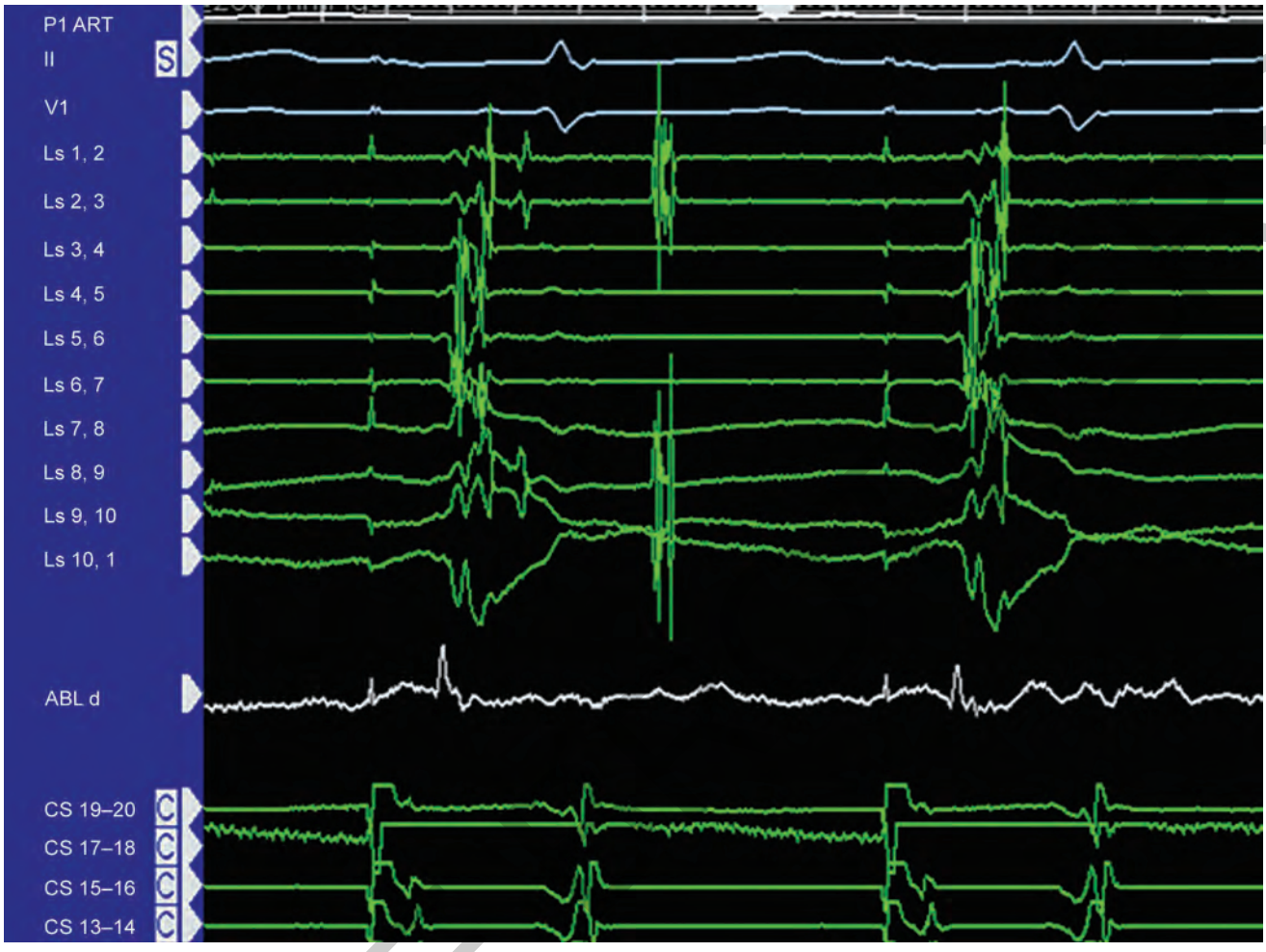

Figure 2.8: Mechanical artifact on lasso catheter mimicking pulmonary vein firing with exit block

UNIPOLAR AND BIPOLAR ELECTROGRAMS

Unipolar Electrograms

All recorded signals in the EP lab are bipolar since they are measured as a voltage difference between two electrodes. However, when one of the electrodes is some distance from the cardiac structures (ideally several inches away), the voltage difference between the exploring electrode and the distant electrode (indifferent electrode) is referred to as a unipolar signal.[4] On the 12-lead ECG, precordial leads V_1 through V_6 record voltages between each of the precordial wires and the indifferent electrode, which is the Wilson Central Terminal (WCT). WCT is assumed to be at the center of a triangle with the three limb wires at the apices **(Figure 2.9)**. The vector sum of currents entering WCT is theoretically zero, and this is considered to be a simplified electrical center of the heart.

When intracardiac electrograms need to be displayed in unipolar fashion, WCT may be used as the reference with the exploring electrode being the distal electrode of a mapping catheter. The mapping system can be configured to display these unipolar signals directly. Electroanatomic mapping systems generate their own WCT and display an independent unipolar electrogram on their dedicated displays.

A wavefront of depolarization that approaches a unipolar exploring electrode generates a positive deflection on the display. As this wavefront moves away from the electrode, a negative deflection is displayed. The steepest negative slope of the signal occurs when the wavefront is below the exploring electrode. This approach works best when the low-frequency cutoff for the amplifier channel is kept suitably low (DC or 0.5 Hz maximum instead of the usual 30 Hz cutoff used for bipolar channels). The instantaneous amplitude of the voltage is directly proportional to the area of the wavefront of depolarization and inversely proportional to the square of its distance from the exploring electrode.[5, 6] Directionality

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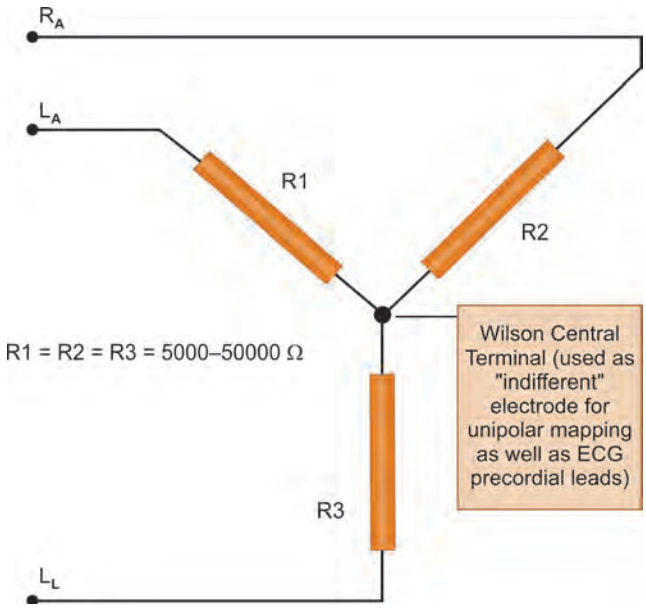

Figure 2.9: Wilson Central Terminal, used as reference electrode for unipolar mapping and ECG pre-cordial leads
Abbreviations: R_A, right arm; L_A, left arm; L_L, left leg; $R_{1,2,3}$, summing resistors

(superior, inferior, lateral, medial) is lost with this mapping approach.

If the wavefront of depolarization originates immediately below the exploring electrode, it spreads away from this electrode in all directions simultaneously, producing a negative deflection (QS waveform). This is usually a site of successful ablation of a focal tachycardia. A very small R wave may be present in this location since the origin of the ectopic focus could be intramyocardial, some distance away from the electrode. Poor electrode contact (or no electrode contact, as is seen if the electrode is floating in the cardiac chamber) manifests as a slow (low rate-of-change) signal instead of the expected rapid positive-to-negative polarity change. Using the unipolar signal to locate accessory pathways and ablate successfully has also been described, with the best results being obtained with a P-QS pattern (the atrial signal fused to the unipolar QS signal) on an "unfiltered" (no high pass filtering) signal.[7]

When mapping diseased tissue, far-field high-amplitude signals from healthy tissue may obscure the effects of low-amplitude local signals. Increasing the cutoff frequency of the high-pass filter to 10–30 Hz minimizes the far-field contribution.[4] A disadvantage of unipolar signals is the increased susceptibility to noise due to the differential coupling of interfering signals to the two widely spaced electrodes. This can be particularly troublesome when using WCT as the indifferent electrode due to surface electrode polarization and contact noise. Using an internal, indifferent electrode in the IVC instead may ameliorate this problem.

Bipolar Electrograms

When two narrowly spaced electrodes (less than 2–3 mm apart) in contact with myocardium are attached to the inputs of an instrumentation amplifier, the difference between the two signals produces a bipolar electrogram. Successive pairs of closely spaced electrodes on a multipolar catheter (1–2, 3–4, 5–6, etc.) produce reasonable bipolar signals as long as the electrode spacing is small. The same bipolar information may be obtained by subtracting one unipolar signal from another unipolar signal as long as the two unipolar electrodes are close to each other and in contact with myocardium. These signals are a good indicator of local electrical activity immediately beneath the two closely spaced electrodes, and so can provide information on timing of local activation.[8] Local activation is denoted by the time of maximum deflection of the electrogram. Reducing the distance between the two electrodes of the bipole can further reduce the sensitivity to distant activation events.[9] The amplitude of these signals in healthy ventricular myocardium (measured with a 10 mm electrode spacing) ranges from 3 mV to 10 mV with durations of less than 70 msec.[10] These amplitudes are reduced substantially with smaller spacing. Also, a lot of the far-field information present in a unipolar signal is removed from the bipolar signal because each electrode of the bipolar pair "sees" a similar (or identical) far-field voltage, and these similar signals are canceled out by the amplifier. Thus, it is reasonable to use a high pass filter with a higher cutoff frequency, such as 30 Hz, for bipolar signals since this will help minimize baseline drifts/shifts with no loss of information.

With close electrode spacing, changes in polarity on the bipolar signals correlate with changes in direction of activation, such as may occur during creation of conduction block during radiofrequency ablation. In the extreme case, this could be a disadvantage since a wave of depolarization that is perpendicular to the two electrodes would be canceled.[11] Conversely, a wave of depolarization that is parallel to the two electrodes would be picked up sequentially by both, resulting in maximum amplitude. The time of peak amplitude in a bipolar signal is the equivalent of the fastest negative downslope of the unipolar signal. Furthermore, if both minimally-filtered unipolar signals and standard-filtered (30–500 Hz) bipolar signals are obtained simultaneously, the time between onset of unipolar electrogram and first peak of bipolar electrogram is usually very short (less than 15 msec) at the site of successful ablation of an automatic tachycardia.[11]

Given the difference in size between the tip electrode and the more proximal electrode of a bipole pair, and the fact that the tip is usually in contact with the tissue with perpendicular orientation (leaving the proximal electrode offset from the tissue by several millimeters), it is not clear if the voltage on the bipolar electrogram is truly representative of the "quality of tissue" beneath the catheter tip.

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This uncertainty becomes particularly significant when creating voltage maps of ventricular myocardium, and using absolute amplitude cutoffs to differentiate scar from viable tissue. This is of somewhat lower concern during epicardial mapping since both electrodes are usually making contact with the tissue due to the more parallel orientation of the catheter "trapped" between epicardium and pericardium. During epicardial mapping, however, areas with more than 5 mm of epicardial fat thickness (near the base of the heart) tend to have lower bipolar voltages (less than 1.5 mV, possibly being misclassified as scar) compared to other areas.[12]

Another significant point is that bipolar electrograms on the distal pair of electrodes of the ablation catheter do not always decrease during ablation, which is the usual indicator of ablation success. Since the amplifier is calculating the difference between the distal electrode and proximal electrode potentials to generate the electrogram, as the tissue below the distal electrode is ablated this difference could decrease (as we would expect) or even increase with reverse polarity, since the proximal electrode may have a larger signal now due to its sampling of more healthy, unablated tissue compared to the desiccated tissue beneath the distal electrode. However, a true unipolar electrogram of the distal electrode would decrease unequivocally as the tissue is ablated effectively.

CONCLUSION

With a combination of high-quality hardware and software design, careful management of lab wiring, appropriate filter settings and knowledge of the limitations of the system, it is possible to effectively map and ablate cardiac arrhythmias in the electrically noisy environment of the EP lab.

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EDITORS' SUMMARY

In this chapter, Duello and Venkatachalam provide an insightful and practical approach to maximize the value of the recorded electrograms during our ablation procedures. Through the remainder of this textbook, several key maneuvers and techniques that define mechanism of arrhythmia and identify the appropriate substrate for ablation will be explained. However, the value of such maneuvers is greatly minimized when we cannot trust or do not know whether we can trust the electrograms we are interpreting.

Duello and Venkatachalam identify for us the fundamental issue of signal-to-noise ratio that defines our ability to record actual but sometimes small and fragmented signals that may be the key to successful ablation.

While there includes in this chapter an excellent discussion of filtering, clipping, gaining, and postprocessing signals to better identify the signals of interest, the authors emphasize the importance of excluding noise at the outset. For this, we make simple and practical suggestions that should form a checklist for every electrophysiology laboratory—identifying and varying the external output, avoiding coiling of cables from electronic equipment, and when necessary, braiding or placing cable perpendicular to each other.

With outstanding illustrative examples, the authors demonstrate the effect of clipping of both analog and digital signals that can exclude, as well as newly place apparent true signals that can obscure the true electrical phenomena occurring in a complex case. The use and limitations of a notch-filter, including an example where pulmonary vein potentials that were obscured because of the frequency of these signals falling within the filtered range.

While several of the limitations involved with optimizing the signal-to-noise ratio are theoretical and difficult to eliminate even in future iterations of mapping systems, the authors point out that improved sampling frequency and the dynamic range for processing signals are likely probable with improved computing power available and the recognized need for find mapping, specifically for supravalvar arrhythmia and identifying fragmented signals and late potentials for substrate-based ventricular tachycardia mapping.

Finally, we revisit the sum of the principles outlined by Duello and Venkatachalam in the subsequent sections that look at optimal techniques for 3-dimensional mapping and integrating analysis of recorded electrograms with complex maneuvers and the underlying pathological substrate.

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Practical Cardiac Electrophysiology

Practical Cardiac Electrophysiology is a clinical and practical textbook that emphasizes the fundamentals of clinical electrophysiology in order to best understand the complexities encountered in the invasive electrophysiology laboratory. The points of focus include cardiac anatomy, biophysics, electrocardiography, and the physiology behind commonly performed maneuvers. After establishing these cornerstones of our field, the complexities of even the most difficult arrhythmias are presented in a correlative and straightforward manner.

A specific attempt has been made to keep the content understandable and useable at the same time comprehensive so as to benefit the spectrum of trainees, allied staff, practicing electrophysiologists, and senior cardiologists. Such a balance would not have been possible without the contributions from masters and international experts in their respective fields for the 47 chapters that make this book. Each chapter is extensively illustrated, and many are based on case examples to aid the reader in relating the information provided to their own practice. Brief editorial summaries at the end of each chapter help in putting the context in perspective, and extensive referencing should guide the reader with their further reading.

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