AAEM MINIMONOGRAPH #16:
INSTRUMENTATION AND MEASUREMENT IN ELECTRODIAGNOSTIC MEDICINE

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AMERICAN ASSOCIATION
OF
ELECTRODIAGNOSTIC MEDICINE

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Reviewed and accepted by the:
1993-1994 Education Committee of the
American Association of Electrodiagnostic Medicine

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The AAEM has determined that the estimated study time for completing this CME activity is four hours and designates this CME activity for four credit hours in Category 1 of the Physician's Recognition Award of the American Medical Association.

EDUCATIONAL OBJECTIVES

The main objective of this minimonograph is to review issues in instrumentation and measurement that affect the accurate recording of nerve and muscle potentials and the interpretation of electrodiagnostic studies. Specific educational goals are to provide a review and to increase understanding of: (1) the basic nature of neurophysiologic action potentials and the concept of waveform content; (2) the operation of high- and low-cutoff filters and the distortion in waveforms that occur with improper filter selection; (3) the electrical characteristics of surface and needle electrodes and the influence of these properties on recorded potentials with special reference to the effect of monopolar and concentric needle electrodes on motor unit parameters; (4) digital instrumentation including analog-to-digital conversion, sampling rates, and waveform display; (5) sources of noise and techniques to minimize the problem during clinical studies; (6) electrical stimulation of nerve, sources of error resulting from incorrect use of stimulators, and techniques to control stimulus artifact; and (7) effect of uncertainties in measurement on calculated nerve conduction velocities.

INSTRUCTIONS

1. The reader should carefully and thoroughly study this minimonograph. If further clarification is needed, the references should be consulted. Do not neglect illustrative material.

2. Read the CME questions at the end of the minimonograph. Choose the correct answer to each question and record it on the CME Registration form on the last page. Retain a copy of your answers for your records.

3. Complete the Evaluation form on the reverse side of the CME Registration form.

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5. Correct answers to the CME questions and a certificate of CME credit earned will be mailed to you.

6. Review those parts of the minimonograph dealing with the question(s) you answered incorrectly, and read the supplemental materials on this aspect of the subject listed in the references.

AAEM MINIMONOGRAPH #16: INSTRUMENTATION AND MEASUREMENT IN ELECTRODIAGNOSTIC MEDICINE—PART I

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Reprinted from *Muscle & Nerve* 1995; 18:799-811

Electrodiagnostic medicine involves the recording, display, measurement, and interpretation of action potentials arising from the central nervous system, peripheral nerve, and muscle. Disease and injury alter neuromuscular architecture and physiology, leading to changes in the time-course (onset and duration), amplitude, and configuration of these potentials. These changes are used for diagnosis, to monitor disease progression, and to assess therapeutic interventions. Technical limitations and instrumentation factors can distort neurophysiologic waveforms during their acquisition, resulting in changes that are essentially identical to those seen with disease or injury. The electrodiagnostic physician needs to be aware of these sources of error and understand techniques that can be used to minimize distortion. Part I of this monograph reviews fundamental electrical principles of action potential generation, the concept of the frequency content of waveforms, analog and digital instrumentation basics, and the influence of filters and electrodes on recorded waveforms. Part II continues the discussion of instrumentation issues and covers amplifier basics, waveform display, noise, and electrical stimulation of nerve. A review of patient safety issues related to nerve stimulation and leakage current follows in the appendices to Part II.

BASIC BIOPHYSICS

Normal nerve and muscle cell membranes have a stable resting transmembrane potential as a result of a uniform distribution of positively charged ions on the outside of the cell membrane and negatively charged ions on the inside. While these charge distributions create a potential difference (voltage) across the membrane, the net potential measured between two points outside of the cell is zero be-

cause the electrical field contributions from the positive and negative charge layers cancel each other. During depolarization of a nerve or muscle cell, current flows along the electrical potential gradients across the membrane, along the interior of the cell, and finally through the extracellular fluid surrounding the nerve or muscle fiber. With depolarization, positive and negative charge distributions along the outside and inside of the membrane are no longer uniformly equal, resulting in the appearance of potential differences between two points outside of the cell. This potential difference varies in time as the membrane undergoes a depolarization/repolarization cycle, creating the extracellularly recorded action potential.

Action potential currents primarily result from the movement of Na^+ and K^+ ions. The tissue between any two points will resist or impede (hence "impedance") the movement of current. The driving force or voltage potential (V), the current (I) produced by the driving force, and the impedance (Z) are related by the equation:

$$\text{Voltage } (V) = \text{Current } (I) \times \text{Impedance } (Z) \quad (1)$$

In electrodiagnostic applications, it is particularly important to understand eq. (1). The potential (voltage) is actually a magnitude that represents a difference of voltage between two points, rather than an absolute value. Thus, current flow will be identical in a particular medium for a difference of 10mV whether that difference exists between two points whose absolute magnitudes are 100 and 90, or 2000 and 1990, or +5 and −5mV. The value of the voltage is positive or negative relative to the ground lead which is taken as the zero reference. In electrodiagnostic studies, action potential magnitudes are expressed in millivolts (mV, 10^{-3} volts) and microvolts (μV, 10^{-6} volts), currents in milliamperes (mA, 10^{-3} amperes), and impedances in kilo-ohms (kΩ, 10^3 ohms) or megaohms (MΩ, 10^6 ohms). Time measurements are of the order of milliseconds (ms, 10^{-3} seconds) and microseconds (μs, 10^{-6} seconds). Although the term Hertz (Hz) is in common usage, cycles per second (cps) (note: 1 cps = 1 Hz) will be used throughout the monograph for frequency measurements, because it is becoming the preferred term.

The action potentials which are ultimately measured during clinical studies result from current flowing through tissues in the vicinity of the recording electrodes. The magnitude and polarity of the recorded potential depends on the charge density near the membrane, the distance from the site of depolarization, the properties of the medium, and the location at which the potential is measured relative to depolarized and normally polarized segments of the membrane. The resultant current flows are complex and are influenced by the volume conductor characteristics of soft tissues. Volume conductor effects are complicated and have recently been reviewed by Dumitru and Delisa.[8]

FREQUENCY CHARACTERISTICS OF NEUROPHYSIOLOGIC WAVEFORMS

During clinical studies, neurophysiologic action potentials originating from nerve and muscle are displayed and interpreted as waveforms in which the amplitude varies with time. Some segments of these waveforms change rapidly with time while others change more slowly. The rapidly changing components indicate that the waveform contains high-frequency elements while the slowly changing components result from low-frequency signal elements. The frequency relationships present in a neurophysiologic waveform (or any waveform in general) can be understood by representing the signal as the summation of a series of sine waves of varying frequency, amplitude, and phase. Characterizing a waveform by the various frequency components it contains is known as frequency or Fourier analysis (Fig. 1).

The frequency content of a waveform refers to the number of terms (frequencies) needed to accurately depict the waveform. Frequency analysis of the compound motor action potential (CMAP) and the sensory nerve action potential (SNAP) are shown in Figure 2. The CMAP signal contains primarily low-frequency components below 1000 cps, while the SNAP has a broader range of frequency components which are shifted toward higher frequencies. The different peak frequencies reflect the differences in the fundamental frequency of the CMAP and SNAP, which can be inferred from their respective waveform durations. For the CMAP this occurs between 60 and 100 cps, which is consistent with the long duration (12–15 ms) of the CMAP; for the SNAP the peak is approximately 300 cps, which is consistent with the shorter duration (2–3 ms) of the sensory potential.

To prevent distortion of a neurophysiologic signal, the recording system, including the electrode, must be able to respond to all of the frequency components within the potential and suppress those frequencies that are unrelated to the potential. Action potentials with rapidly changing, high-frequency components include fibrillation potentials, polyphasic motor unit action potentials

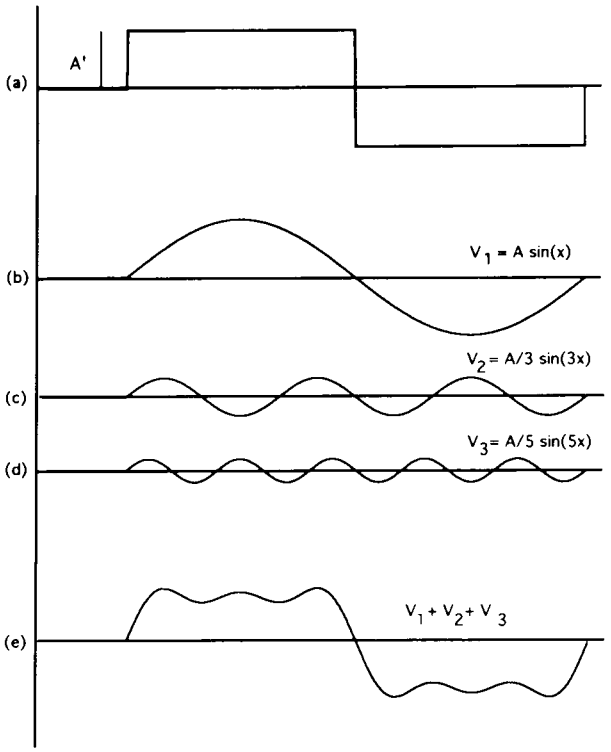

FIGURE 1. An example of the principle of Fourier analysis. Any waveform, the square wave in this example (a) can be decomposed into a series of sine waves of varying frequency, phase, and amplitude. In this example, the first three elements of the series (b, c, d) and their sum (e) are displayed. The addition of more frequency terms would result in a better approximation of the original waveform. ($x = 2\pi ft$ where f is frequency and t time, $A = 4A'/\pi$)

(MUAPs) and the potentials produced by single muscle fibers (single fiber EMG). Potentials with slowly changing components include the CMAP and the SNAP, especially close to the peaks of the positive and negative deflections and during the slow terminal return to baseline, the positive component of positive sharp waves, and somatosensory evoked potentials (SEPs).

BASIC APPARATUS AND INSTRUMENTATION CONCEPTS

During electrodiagnostic testing, the recording system must accurately reproduce the neurophysiologic potentials generated by muscle and nerve fibers. Two electrodes are introduced into the electric field produced by the depolarizing nerve or muscle and the potential is measured as the difference between the two electrode locations. By convention, the electrodes have been named the "active," or G_1, and "reference," or G_2, electrodes. The potentials measured are not those that actually occur, but those that exist as distorted by the presence of the electrode. Electrode size is generally governed by the application and chosen to be both practical and to limit the distortion to an acceptable level.

The basic components of a typical instrument used in EMG, nerve conduction velocity (NCV), and evoked potential studies are illustrated in Figure 3. Neurophysiologic signals range in amplitude from less than 1μV for SEPs to many microvolts for SNAPs to millivolts for MUAPs and CMAPs. These signals exist within an environment filled with electrical noise. Electrical noise from sources such as radio, 60 cps power lines, and other nearby electrical equipment pervade the environment. Even under the best conditions, the noise is typically several orders of magnitude larger than the waveform under study. Successful recording depends on the ability to selectively amplify the signal of interest approximately 50–250,000 times while rejecting the unwanted extraneous noise. After amplification the signals are generally 1–10 V in magnitude, which are amplitudes needed by the electronic circuits which perform the analog-to-digital conversion and control the audio and video output displays. Gain refers to the increase in amplitude of the action potential as a result of electronic amplification. Sensitivity refers to the display resolution of the amplified signal and is expressed in microvolts or millivolts per vertical screen division.

To achieve selective amplification of the neurophysiologic signal while rejecting environmental noise, a special amplifier known as a differential or instrumentation amplifier is used (Fig. 4). A differential amplifier measures the difference in the

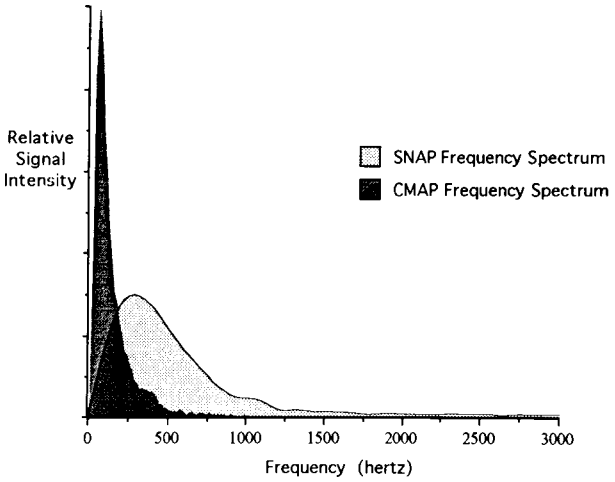

FIGURE 2. Average frequency power spectrum of the SNAP and CMAP from 9 normal subjects using filter settings of: CMAP 2–10,000 cps and SNAP 2–3000 cps.

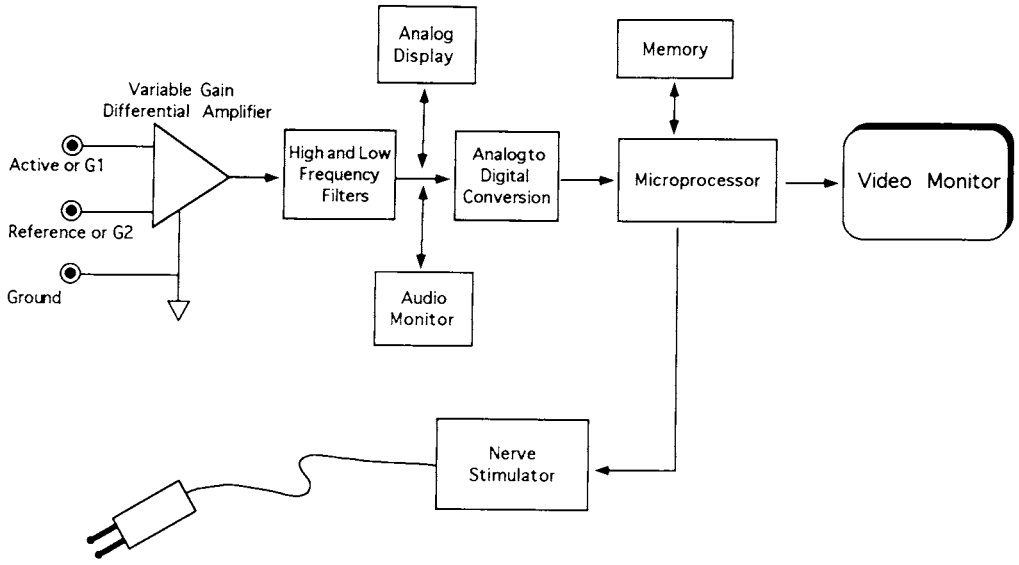

FIGURE 3. Most contemporary EMG/NCV/EP instruments use computer controlled data acquisition and display. Recording an action potential requires a series of steps consisting of amplification of the low level neurophysiologic signal, filtering out unwanted noise, conversion to digital values, scaling and analysis by the microprocessor, and finally display of the waveform on a video monitor. Using modern electronics, this process can be accomplished very rapidly (in milliseconds) allowing for "real time" display of signals.

electrical potential between the active and reference electrode inputs. If the recording electrodes are spaced reasonably close to each other and have similar characteristics, environmental electrical noise such as 60 cps power line interference will appear at both inputs with nearly the same amplitude. A signal that appears with equal amplitude at both inputs is considered "common" to the inputs and is termed a common mode signal. Since the differential amplifier only responds to the difference between the signals, most of this common but undesired signal is not amplified and is eliminated. With the same electrode setup, the active electrode is intentionally placed close to the action potential generator while the reference electrode is placed in an "electrically silent" area relative to the action potential generator. Thus at the same time that the amplifier is rejecting the common mode noise, the action potential itself will be amplified since the difference in this signal as seen by the active and reference inputs is large.

The ability to selectively amplify only the difference signal while rejecting the common mode signal is expressed by the common mode rejection ratio (CMRR). The CMRR is the ratio between the gain of the neurophysiologic potential (i.e., the difference signal) and the gain of the 60-cycle noise (i.e., the common mode signal). There are practical limits to the amount of common mode rejection which can be designed into amplifiers. Variability

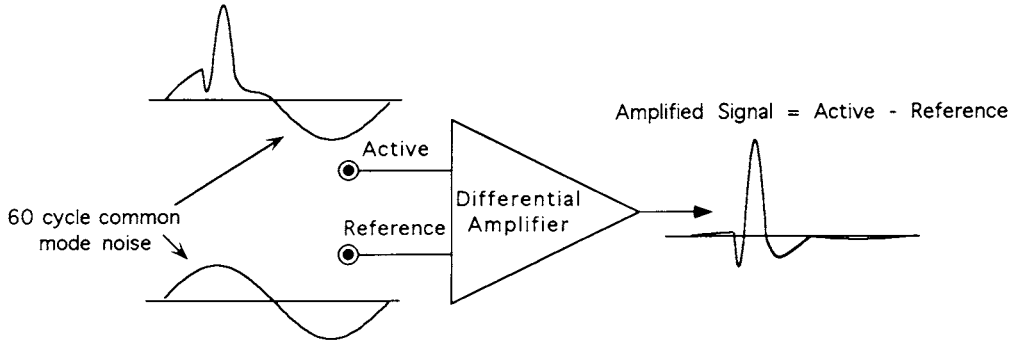

FIGURE 4. Differential amplification. The common mode 50/60-cps interference is present at both the active and reference inputs. The waveform of interest is seen by the active lead but not the reference. The differential amplifier only amplifies the difference in the signals present at the two inputs, thus eliminating most of the 60-cycle noise while retaining the waveform.

in electronic component tolerances and the presence of stray capacitance in circuits typically limit CMRR to the range of 300,000 or less.

CMRR is usually specified in units of decibels (dB) where:

$$\text{CMRR (dB)} = 20 \log \left(\frac{\text{Gain of the Difference Signal}}{\text{Gain of the Common Mode Signal}} \right) \quad (2)$$

Contemporary EMG amplifiers used for clinical studies should have a CMRR of 90 dB or greater.

During and following amplification, the neurophysiologic signal is filtered to remove extraneous and unwanted noise components. The filters will determine how the instrument responds to both rapidly changing (high-frequency) and slowly changing (low-frequency) signal components. The filtered and amplified signal is then displayed. Two general approaches are used for waveform display: analog and digital. In analog displays, the signal is displayed on a cathode ray tube as a continuously changing waveform. With digital display, the neurophysiologic potentials must be converted into a digital or numeric format before these signals can be analyzed. The conversion process is called analog-to-digital conversion (ADC) and is accomplished with special integrated circuit devices.

FREQUENCY RESPONSE AND FILTER SETTINGS

General Filter Considerations. Filtering is an integral step in the processing of neurophysiologic potentials. The most important function of filters is noise attenuation. Noise (which is always present) can be removed if the noise frequency is sufficiently different from the frequency components that make up the neurophysiologic waveform itself. Successful noise reduction improves the quality of a waveform and makes clinical interpretation easier and more accurate. A second role for filtering is to reveal information in a signal that may not be intuitive or even measurable under ordinary recording conditions. An example of this is single fiber EMG (SFEMG), in which the low-frequency filter cutoff is raised to 500 cps to remove the low frequency, slowly changing the waveform components. This decreases the pickup volume of the electrode, allowing the SFEMG needle to selectively record from the few muscle fibers immediately adjacent to the exposed needle tip to obtain jitter and density measurements.

The range of frequency components contained in a neurophysiologic potential can be defined by its frequency bandwidth, i.e., the range of frequencies from the low-frequency limit to the high-frequency limit. Filters are described by their effect on various frequency components. Low-frequency filters (LFF) remove low-frequency, slowly changing components of a waveform. Since they allow the higher frequencies to pass through and be included in the waveform they are also known as high-pass filters. High-frequency filters (HFF) remove the high-frequency, rapidly changing components and allow the low frequencies to pass through and hence are also known as low-pass filters. The cutoff frequency of a filter defines the frequency at which half of the signal energy has been removed (Fig. 5). As an example, an LFF with a cutoff of 10 cps will attenuate the amplitude of a 10 cps signal by approximately 30% (which corresponds to an energy loss of one half).

Filters are generally not perfect in that the re-

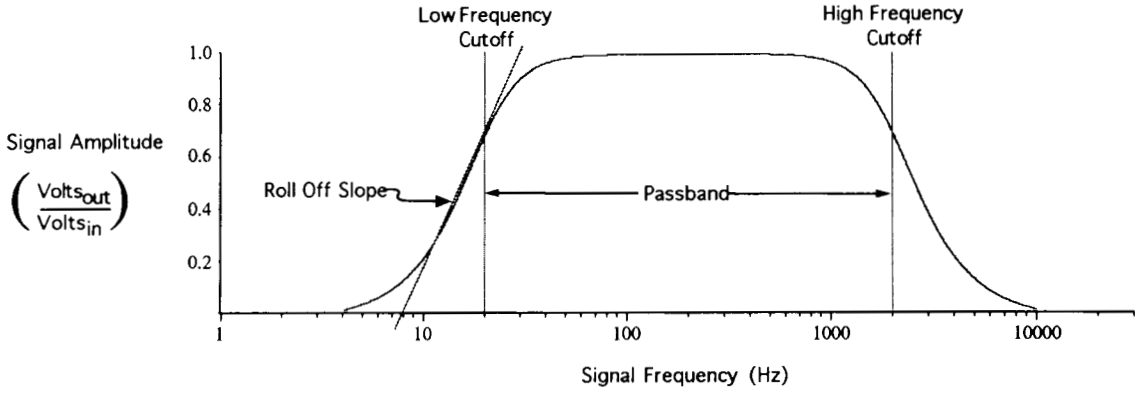

FIGURE 5. Frequency response of a 20-cps LFF in combination with a 2000-cps HFF. The region between the filters' cutoffs is known as the pass band or bandwidth. Near the upper and lower cutoffs, the attenuation occurs gradually. For the HFF, this means that some signal amplitude below the high-frequency cutoff is lost and also that considerable frequency components above the cutoff are allowed through the filter. A similar effect occurs with the LFF.

moval of unwanted frequencies occurs in a gradual fashion. The rate of rolloff of a filter reflects how rapidly it attenuates frequencies beyond its cutoff value. This rolloff rate is expressed by the order of the filter, with higher order filters having a more rapid rolloff and hence better selectivity. In addition to their effect on attenuating the amplitude of different frequencies components, a filter will change the phase of a signal passing through it. Phase shift is not particularly intuitive but can be thought of as a time delay occurring in a particular frequency component as it passes through the filter. If all the frequencies were delayed equally the effect would be unimportant, but filters can, and indeed most analog filters do, cause different amounts of phase shift in the frequency components near the cutoff value. This can result in waveform distortion. Further discussion of phase shift and rolloff are beyond the scope of this review and can be found in most general electronic texts including Barry[1] in AAEM Minimonograph #36.

Filters used in electrodiagnostic studies can be implemented as analog electronic circuits using combinations of resistors, capacitors, and amplifiers or as mathematical algorithms (digital filters) applied to the signal after conversion into a digital form.[14] If the characteristics of the filters, i.e., cutoff, rolloff rate, and phase response, are the same, then analog and digital filters will have essentially the same effect on the waveform.

All filters distort waveforms to varying degrees because of their frequency attenuation and phase shift characteristics. Because filters are used routinely, however, electromyographers become accustomed to these "distorted" waveforms and focus on detecting differences which indicate abnormality. As the use of digital signal processing in electrodiagnostic instruments continues to increase, digital filtering will become more routine. Because digital filters can be designed to have characteristics that are extremely difficult to implement in circuit form, these filters can extract additional waveform features that may not be immediately intuitive or obvious. Potential dangers also exist with digital filters if distortions created by the filtering process are misinterpreted as having a physiologic basis.[13,14]

Recommended settings for routine clinical studies are summarized in Table 1.

Table 1. Recommended filter settings for clinical studies.

Test	Low-freq setting (cps)	High-freq setting (cps)
Motor NCV	2–5	10k
Sensory NCV	5–10	2–3k
EMG—insertional activity	10–20	10k
EMG—quantitative analysis	2–5	10k
SSEP	5–20	1–2k

High-Frequency Response. The length of time during which a rapid change in potential occurs, usually from a baseline value, is the rise time of the potential. The high-frequency response of the amplifier will determine the minimum rise time for which it is able to detect the full potential change. Figure 6 demonstrates the distortion in an MUAP which occurs if the high-frequency response is inadequate to capture the rising slope of the main negative spike of the potential. As the high-frequency filter cutoff is sequentially reduced from 10,000 cps to 500 cps, there is a progressive decline in the amplitude of the waveform. Amplitude loss and elimination of rapid changes in the MUAP can also reduce polyphasia. Since rapid rise times occur when recording MUAPs with a concentric needle, an 8000–10,000-cps high-frequency limit is needed for quantitative analysis.[5]

Similarly, a 16–25% reduction in the main spike amplitude of the SNAP[5,9,19] occurs when the high-frequency response is lowered from 10,000 cps to the 500 cps. This effect is shown in Figure 7, where the inability to follow the rapidly rising and falling components of a SNAP can be easily observed. This results in a lower amplitude but also a longer latency to the peak of the waveform. Long-duration potentials, such as the CMAP and SEP,

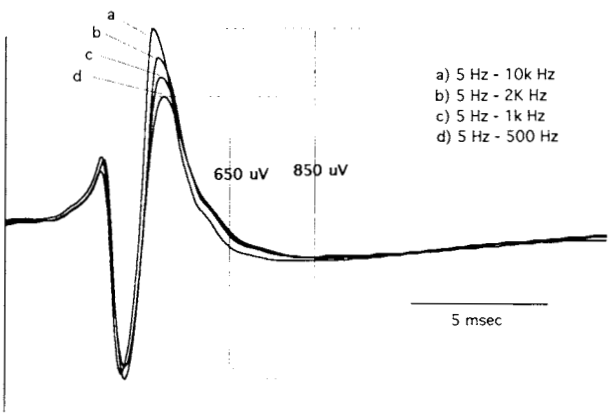

FIGURE 6. The effect of lowering the HFF cutoff on MUAP amplitude. As high-frequency components are removed by lowering the high-frequency cutoff from 10,000 cps to 500 cps, the rising slope of the main negative spike is attenuated, resulting in a 24% reduction in peak-to-peak amplitude.

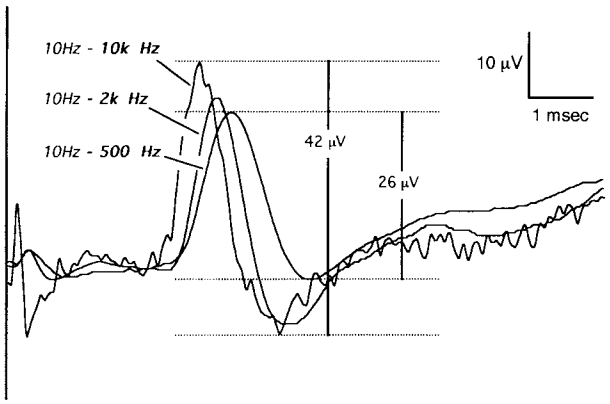

FIGURE 7. Changes in the median SNAP recorded with high-frequency filter settings ranging from 10,000 cps to 500 cps. The loss of the higher-frequency components of the waveform results in a slower risetime and a reduced amplitude. A progressive delay in the peak latency of 0.5 ms also occurs as the filter cutoff is lowered.

contain mostly low-frequency components. For these waveforms, the amplitude remains stable over a wide range of HFF settings ranging from 10,000 cps down to 500 cps.[9]

Latency to the onset of a waveform can also be affected by an inadequate high-frequency response. In this situation, the measurement system cannot follow the normal abrupt departure from baseline and the subsequent rising slope of the negative spike of the waveform.[7,19] Because of the slower rise of the waveform caused by the inadequate HFF, more time elapses before the signal reaches a voltage level that can be visually identified as the departure from baseline (Fig. 8).

Maintaining the high-frequency response at 10,000 cps would always ensure that the fast changing events in the waveform are preserved; however, significant high-frequency noise originates in the amplifier electronics and can obscure sensory evoked potentials. Decreasing the high-frequency response to 2000 cps removes most of this noise, allowing for better visualization of the SNAP (Fig. 9). The tradeoff that must be accepted for the noise reduction in sensory studies, however, is a 10–15% loss of SNAP amplitude.[5] In most clinical settings this is a reasonable compromise. If extremely low-noise amplifiers are available or averaging is used to remove noise, a higher frequency cutoff of 5000 cps can be used.

The body itself, namely the soft tissue volume conductor between the nerve or muscle fiber and the electrode, functions as a high-frequency filter. As the signal passes through body tissue, high-frequency features of the waveform, i.e., rising and falling edges and rapidly changing spikes, will

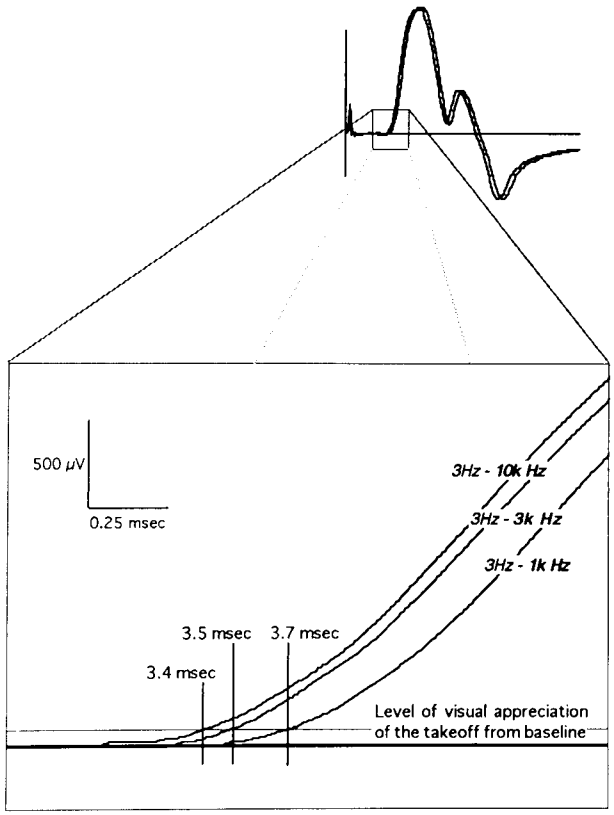

FIGURE 8. Onset of a median CMAP recorded with high-frequency filter settings ranging from 10,000 cps to 1000 cps. The loss of the high-frequency response results in a slower departure from the baseline and a less steep rise of the negative spike. The waveform takes longer to reach a voltage level that can be distinguished from the baseline, resulting in a latency delay of 0.3 ms between recordings performed at 10,000 cps and 1000 cps.

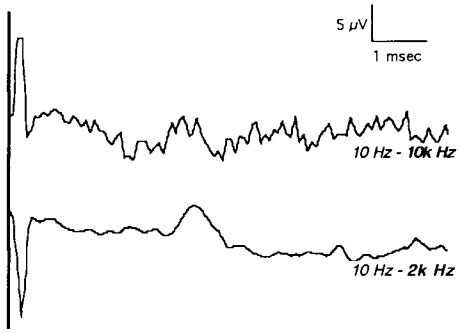

FIGURE 9. Electronic noise is generated and amplified along with any neurophysiologic potentials. The SNAP illustrated is small and requires high gain for adequate visualization. In the top tracing, using a HFF cutoff of 10,000 cps, the action potential is present but obscured by baseline noise. Decreasing the cutoff to 2000 cps removes most of the noise, allowing for easier waveform recognition and measurement.

be attenuated. A comparable way of viewing the tissue-filtering effect is to realize that low-frequency components pass through tissues with less attenuation than occurs with the high-frequency components. As a result, low-frequency action potential components can be recorded even with the electrode at a distance from the nerve or muscle cell. This effect is most obvious as one observes the blunting in the shape of the motor unit potential as an EMG needle is moved away from the motor unit.

Low-Frequency Response. The low-frequency response determines the ability to accurately record the slowly changing components of action potentials. CMAPs, evoked potentials, and the slow return to baseline of the MUAP are affected by the low-frequency response characteristics of the instrument. Inadequate low-frequency response can lead to errors in waveform amplitude, latency, duration, and morphology.

Progressively raising the low-frequency cutoff while recording a CMAP potential will cause initial amplitude loss and finally waveform distortion (Fig. 10). The recording system can no longer track the slow waveform changes near the negative peak and begins to decay back to the baseline, causing a reduction in the amplitude and duration of the main negative phase. Likewise, the slow return to baseline cannot be tracked and the waveform begins to decay sooner, distorting the terminal components of the waveform. Similar effects occur during sensory studies[9,19]; however, the LFF cutoff must be raised further for a comparable reduction in amplitude. This can be explained by the shorter duration of the SNAP which is the result of its signal content being contained in higher-frequency components than occur in the CMAP potential.

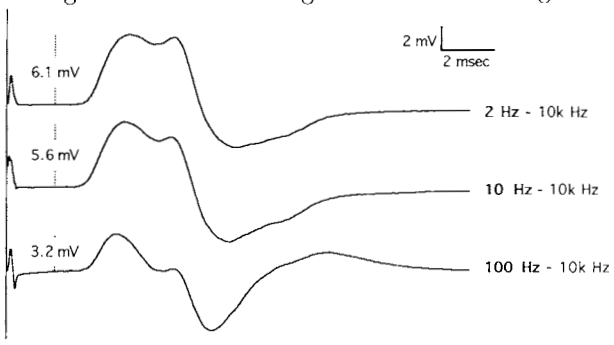

FIGURE 10. The long-duration CMAP contains predominantly low-frequency components. Raising the LFF filter cutoff from 2 cps to 10 cps initially causes a small (8%) reduction in the negative spike amplitude. With further raising of the LFF cutoff to 100 cps, the amplitude is markedly reduced, the duration of the negative spike decreases, and an extra phase is added, distorting the waveform configuration.

Raising the low-frequency cutoff also causes a decrease in the latency to the peak of a waveform (Fig. 11). The peak of the CMAP, and to a lesser extent the SNAP, are relatively slowly changing events. By removing the slowly changing components of the waveform, the instrument is unable to track the waveform peaks and begin an early return to the baseline, shifting the peak latency to a shorter value.

Motor unit potentials are also affected by changes in the low-frequency response. From a clinical standpoint, the low-frequency response is both a potential source of error and a useful way of extracting additional MUAP information. Similar to the terminal distortion seen in CMAP recordings (Fig. 10), the MUAP duration increases and an extra phase at the end of the potential can occur as the low-frequency cutoff is raised from 2 to 20 cps (Fig. 12). This type of distortion led to the recommendation by Buchthal[4] of a low-frequency limit of 2 cps for quantitative MUAP analysis.

If the LFF is further raised to exclude frequencies below 300–500 cps, there is a marked reduction in waveform amplitude and duration (Fig. 13). This effect can be used clinically to assess motor unit stability and has been termed "pseudo SFEMG." By raising the LFF cutoff, the electrical contribution from muscle fibers of the motor unit distant from the electrode is lost. This occurs because of the soft tissue filter effects previously described. Clinical applications of this technique are discussed further in the delay line and trigger section.

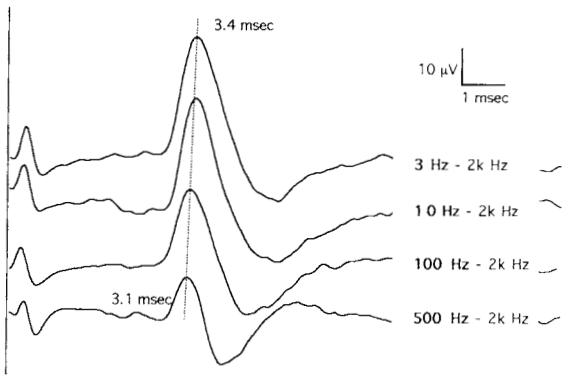

FIGURE 11. Effect of the low-frequency response on the SNAP. Amplitude and latency remain stable as the low-frequency cutoff is changed from 3 cps to 10 cps. With further increases in the LFF settings to 100 cps and 500 cps. there is a progressive shift to shorter peak latencies (0.2 and 0.3 ms, respectively) and reduced amplitudes.

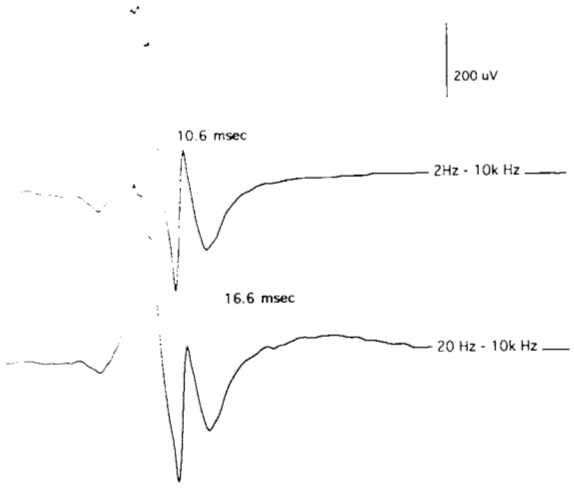

FIGURE 12. Effect of the low-frequency response on the MUAP amplitude. As the low-frequency response cutoff is raised from 2 cps to 20 cps, terminal distortion of the MUAP is seen. The addition of a trailing negative hump results from phase distortion by the filter and erroneously increases the duration of the potential from 10.6 to 16.6 ms.

Studying insertion activity with a low-frequency response of 2 cps can be difficult because movement artifacts (slowly changing polarization potentials) receive too much amplification, causing the baseline to wander. Switching the low-frequency cutoff to a higher value (10–20 cps) allows better visualization of spike potentials during insertional activity analysis. Shifting back to a lower value preserves waveform morphology when examining voluntary MUAPs.

Most potentials of clinical interest contain significant frequency components near 60 cps. Although it is tempting to use the 60 cps notch filters which are available on many instruments to decrease noise, this results in an unacceptably high degree of waveform distortion.

ELECTRODE FACTORS

In all electrodiagnostic applications, three electrodes are applied to the patient. Two electrodes, the active and reference, serve as locations between which the action potential is measured and amplified. The third, a ground electrode, serves as a zero-voltage reference point for the amplifiers.

Both surface and needle electrodes are made from a variety of metals and alloys including stainless steel, platinum, silver–silver chloride, nickel–chromium, and silver and gold alloys and plating. Whenever a metal electrode interacts with an electrolyte (i.e., electrode paste, sweat, or extracellular fluid) an electrochemical reaction occurs, resulting in a separation of negative and positive charges at the electrode–electrolyte interface.[10,22] The charge separation gives rise to an electrode polarization potential which acts like a small battery and can range in amplitude from 100 to 600 mV[21,22] depending on metal type, the electrolyte composition, and ionic concentration. With high-quality electrodes in good condition, polarization potentials are relatively stable and do not appear since the amplifier is not responsive to D.C. (unchanging) signals. Following movement of a needle or surface electrode, the polarization potential will dissipate and fluctuate for a short period of time, giving rise to spike potentials and baseline movement artifacts. When electrodes are dirty, corroded, or in the case of monopolar needle electrodes, the Teflon coating is frayed or excessively thinned, polarization potentials are unstable. Unstable polarization potentials can sometimes give rise to artifacts which resemble abnormal membrane potentials. Silver–silver chloride electrodes earned their reputation because their electrode polarization potentials are quite stable and hence relatively noise-free.

The electrode surface has an electrical capacitance and a resistance component. Both contribute to the total impedance of the electrode.[20] This impedance is frequency-dependent and increases as electrode area decreases. The importance of this variation of impedance with frequency and the interaction with the amplifier will be discussed later.

Surface Electrodes. Most NCV and SEP techniques utilize small surface disk electrodes or ring electrodes for the digits as the two electrodes between which the action potential is measured. In motor nerve conduction studies, the active electrode is placed over the motor point (generally midway between the origin and insertion of the

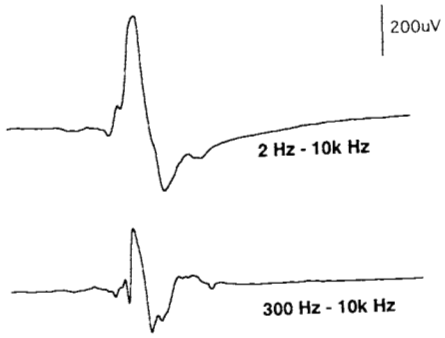

FIGURE 13. "Pseudo" SFEMG using a standard monopolar needle. Raising the low-frequency filter to 300–500 cps decreases the pickup area of the electrode, causing a marked reduction in MUAP amplitude and duration, and alters waveform shape.

muscle) and the reference electrode over the distal tendon or perhaps even further away. By placing the electrode over the motor point (motor endplate region) of the muscle, depolarization and action potential (AP) generation will occur directly beneath the active electrode. The volume conduction of this AP to the electrodes will give rise to a biphasic potential with an initial negative deflection[8] and a relatively easily determined takeoff. In contrast, when the active electrode is moved away from the motor endplate region, a small positive deflection precedes the main negative waveform spike. Accurate assessment of onset is more problematic in this situation but remains best measured to the initial departure from baseline.

The location of the reference electrode will also influence the shape and amplitude of the CMAP. Ideally, the reference is placed "far enough away" so that its potential will remain close to zero. Because of the large size of muscle compound action potentials, this is frequently not the case. When both the active and reference electrodes "see" the potential at the same time, the recorded waveform is the difference in voltage between the electrodes and has effects on amplitude and morphology (Fig. 14).[12]

In recording SNAPs, it is customary to place both electrodes over the nerve. The spacing between the active and reference electrode will affect the SNAP amplitude. As the interelectrode spacing is reduced to less than approximately 4 cm, the amplitude of the SNAP progressively declines. The 4-cm spacing tends to maximize the peak-to-peak amplitude because at this distance the traveling wave of depolarization in the sensory nerve will have completely passed the active electrode before arriving at the reference electrode. This assumes a conduction velocity in a sensory nerve of 50 m/s (50 mm/ms) and an action potential duration at the nerve membrane surface of 0.8 ms giving a length for the depolarized region of 4 cm. Hence, to use normal reference amplitude values, standardization of the distance between the two surface electrodes is necessary. The volume conduction principles which underlie this effect are discussed more fully by Dumitru.[8] Techniques for needle electrode sampling of nerve action potentials are well described by Oh.[17]

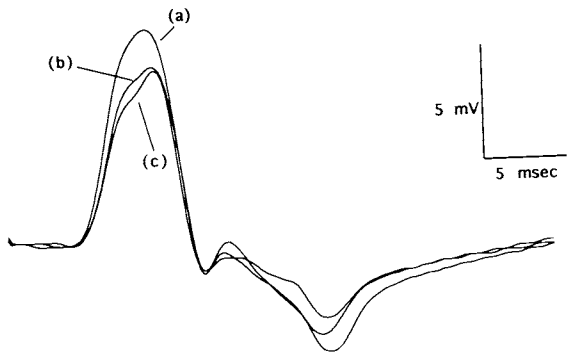

FIGURE 14. Variability in the median CMAP recorded with reference electrode placement: (a) at the metacarpal phalangeal joint; (b) at the midproximal phalanx; and (c) at the middistal phalanx. Placements (b) and (c) yield nearly identical responses while (a) has a 24% larger baseline-to-peak amplitude.

Needle Electrodes and Techniques. *Monopolar Needle Electrodes.* Monopolar EMG needle electrodes consist of a Teflon-coated stainless steel wire that tapers to a sharp tip (Fig. 15). When using a monopolar needle, the MUAP potential is measured as the voltage difference between the active needle tip which is placed into the muscle belly and a reference skin-surface electrode. In an effort to maintain a zero potential, the surface (reference) electrode is placed over a nearby tendinous or bony site. Even at this location, however, the surface electrode is seldom completely electrically silent. Depending upon the degree of muscle contraction originating from the muscle being studied or other muscles being tensed by the patient, background potentials will be recorded from the surface electrode and displayed along with those originating near the needle tip. Placing the surface reference electrode over the skin close to the needle can be helpful in reducing the influence of distant muscle contraction when searching for spontaneous activity.

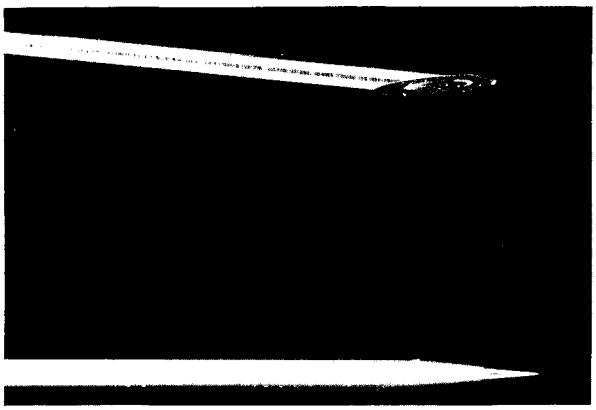

FIGURE 15. Comparison of a standard monopolar EMG needle (bottom) and concentric needle (top). The active recording surface consists of the exposed tip of the monopolar needle as compared to the central beveled core of the concentric needle.

The average exposed tip area is approximately 0.14–0.20 mm^2. With repeated sterilization, the exposed area can increase up to 20% even without obvious Teflon fraying. Since a monopolar needle records the average potential over its exposed tip, cracked, chipped, or frayed Teflon coating will cause a reduction in the amplitude of motor unit potentials. If breaks or cracks in the Teflon coating along the shaft of the electrode exist, the signal received at the tip will be "short-circuited" back into the tissue, further reducing the amplitude. Since the Teflon does not extend fully to the hub in many monopolar needles, the needle should not be inserted its entire length.

The reference electrode in the monopolar technique has approximately a tenfold lower impedance than the needle electrode. This impedance difference interferes with the amplifier's ability to reject common mode signals, often resulting in 50/60 cps interference.

Concentric Needle Electrode. The concentric needle consists of a fine wire electrode, insulated and housed inside an outer cannula or hypodermic needle (Fig. 15). Action potentials are measured as the difference between the tip of the central isolated core or active electrode and the bare metal cannula, which serves as the reference electrode. Because of the physical proximity of the active and reference recording surfaces, the reference cannula will be near the MUAP and will record an attenuated version of it.

The exposed area of the central core of a concentric electrode is determined by the wire diameter and the angle of the bevel. The active electrode tip area varies between 0.2 and 0.9 mm^2. The recording surface is usually smaller but more constant in size than the exposed tip of the monopolar needle. Because it is smaller, the core electrode impedance can be up to ten times larger than the monopolar electrode. As in the monopolar technique, the core and the cannula impedance change with frequency, decreasing at the higher frequencies. Any difference in impedance between the central core and the cannula will reduce common mode signal rejection. Because they are close together, the common mode signal impressed upon the two electrodes in the concentric application is more likely to be the same as compared to the monopolar needle/surface disk electrode pair. For this reason, common mode noise origination from power lines tends to be somewhat less problematic with concentric needle recordings.

The Influence of Needle Characteristics on MUAP Parameters. The type, size, orientation, and configuration of the needle electrode will influence the characteristics of the MUAP recorded. The recording surfaces of both monopolar and concentric needles are large relative to the diameter of a typical muscle fiber (60 μm).[2] As a result the active recording surface is in contact with many muscle fibers, some of which belong to a depolarizing motor unit, while others are electrically silent. The resulting MUAP recorded represents a summation and average of the electrical activity occurring along the entire exposed needle tip.

Clinical studies[4,6,7,11,15,18] comparing the differences in MUAP parameters between monopolar and concentric needles have shown variable results but in general have indicated that monopolar needles produce larger amplitudes and greater phasicity but comparable durations (Table 2). The lack of standardization in recording techniques and measurement criteria makes comparisons of absolute amplitude, duration, and phasicity between these studies difficult.

The reported MUAP amplitude increase of 0–100% when using monopolar needles as compared to concentric needles results from several factors. The main MUAP spike amplitude is determined by the small number of muscle fibers immediately adjacent to the electrode tip.[3,15,16] Since the active recording surface of the concentric needle is shielded on one side by the reference cannula, the active electrode does not record action potentials from muscle fibers behind the cannula.

Table 2. Value of MUAP parameters recorded with monopolar needles compared to concentric needles.

Study	#MUPs evaluated	Amplitude	Duration	Phases/turns*
Buchthal, Guld, and Rosenfalck[4]		↑	ns	
Nandedkar and Sanders[15]	200	↑	ns	↑ †
Howard, McGill, and Dorfman[11]	7270	↑	ns	↑
Chu, Chan, and Bruyninckx[7]	52	↑	↑	↑
Pease and Bowyer[18]	120	ns	ns	
Chan and Hsu[6]	202	↑	↑	ns

*Number of turns.

In contrast, the multidirectional recording nature of the monopolar needle likely places more muscle fibers from the MUAP under study close to the active electrode tip, resulting in an increase in the MUAP amplitude. A second factor relates to the previously discussed concept of differential amplification. When using concentric needles, the reference cannula will be in the physical vicinity of the actively depolarizing motor unit and will record an attenuated version of the MUAP. Since the recorded potential is the difference between the active tip and the reference cannula, the attenuated version of the MUAP will subtract from the MUAP seen by the active tip and result in a decrease in the amplitude of the potential.

Motor unit complexity and phasicity are related to the degree of temporal synchrony of the discharges of the individual muscle fibers contributing to the MUAP. Variability in muscle fiber conduction velocity, endplate dispersion, and conduction time in terminal axon branches all contribute to this synchrony. Monopolar needles record activity from more muscle fibers than occurs with concentric needles because of their generally larger exposed tip and the distant location of the reference electrode. This larger population of discharging muscle fibers results in less temporal synchrony and a corresponding greater degree of polyphasicity.

Lastly, MUAP duration reflects the total number of muscle fibers within the motor unit or at least within a recording volume that is large relative to the region responsible for determining motor unit amplitude. Duration measurements are greatly influenced by the slowly changing leading and terminal portions of the potential. Because body tissue acts as a high-frequency filter, the amplitude of rapidly changing waveform components diminishes rapidly while slowly changing components pass relatively unattenuated. This allows many muscle fibers to contribute to the slowly changing leading and terminal portions of the MUAP. As a result, the size and orientation of the needle tip play a less important role in duration measurements than for other parameters. Although the majority of studies have not demonstrated any significant differences in MUAP duration between needle types, two reports have found monopolar durations to be longer. Chu et al.[7] found monopolar durations to be statistically prolonged though the magnitude of the increased duration was only 10–20%. Chan and Hsu,[6] using a technique in which simultaneous recordings of the same MUAP were made with a concentric and monopolar needle, found significantly longer durations with the monopolar recording. This was attributed to the larger area of the monopolar needle and the distant reference location which limits the effect differential amplification has on canceling muscle fiber potentials discharging at a distance from the needle electrode.

SUMMARY

This portion of the monograph has explored action potential generation, the frequency content of waveforms, basic principles of signal amplification, distortions in waveforms that can result from inadequate high- and low-frequency filter settings, and the influence of electrode characteristics on waveform amplitude and morphology. An appreciation of the sources and magnitudes of the errors due to technical recording limitations, and importantly, techniques which can be used to minimize their effects, will improve the electrodiagnostic physician's ability to acquire accurate data.

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AAEM MINIMONOGRAPH #16: INSTRUMENTATION AND MEASUREMENT IN ELECTRODIAGNOSTIC MEDICINE—PART II

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Reprinted from *Muscle & Nerve* 1995; 18:812-824

This part of minimonograph #16 continues a discussion of basic electrodiagnostic instrumentation and addresses amplifier basics, waveform display issues, noise, electrical stimulation of nerve, and experimental error in performing nerve conduction velocity (NCV) studies. Two appendices cover patient safety concerns related to nerve stimulation and leakage current.

AMPLIFIER AND DISPLAY CONSIDERATIONS

Electrode–Amplifier Impedance Relationships. A simplified circuit representing the action potential generator (axon or muscle fiber), electrode, and amplifier is shown in Figure 1. The electrode impedance as illustrated includes both the intrinsic impedance of the electrode and the electrode–skin (or electrode–muscle) interface. To measure an action potential, current generated by the depolarizing axon or muscle fiber must flow through the electrode into the amplifier and return to the patient through the ground lead. Because of electrode impedance (resistance) to current flow, there will be a drop in the voltage of the action potential signal across the electrode. As a result, the action potential seen by the amplifier will be attenuated (point B on Fig. 1) relative to its actual amplitude at the electrode tip. To minimize this attenuation, the impedance of the amplifier must be much greater than the electrode impedance. The amount of signal attenuation is determined by the amplifier-to-electrode impedance ratio.

$$\text{Volts}_{\text{amplifier}} = \text{Volts}_{\text{electrode}} \times \left(\frac{\text{Impedance}_{\text{amplifier}}}{\text{Impedance}_{\text{amplifier}} + \text{Impedance}_{\text{electrode}}} \right) \quad (1)$$

To limit the drop in signal amplitude to 1%, the

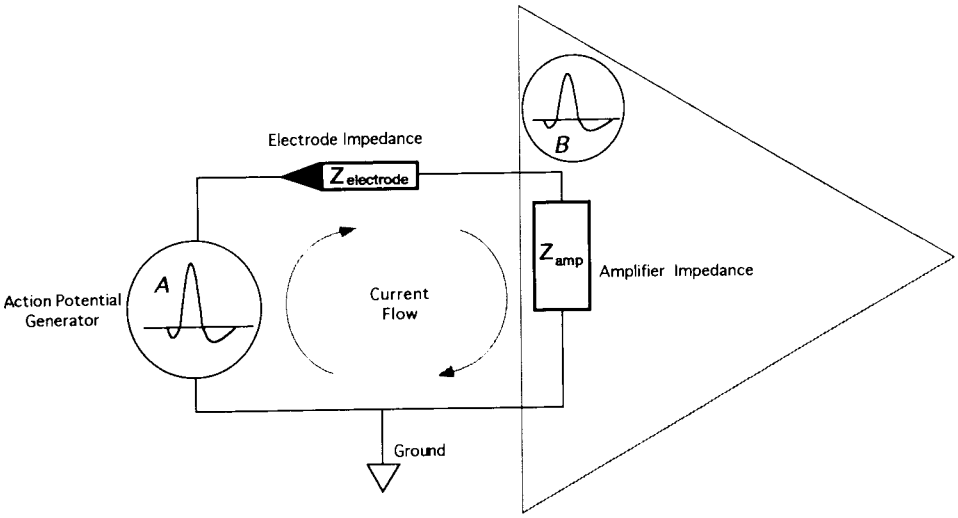


FIGURE 1. Equivalent circuit of a single electrode and the input of an electrophysiologic amplifier. The voltage measured between any two points is equal to the current multiplied by the impedance between the two sites ($V = I \times Z$). Current generated by the depolarization of a nerve or muscle will flow through both the electrode ($Z_{electrode}$) and the input stage of the amplifier ($Z_{amplifier}$). The voltage (V) seen at the amplifier input (site B) is equal to the current (I) flow times the input impedance (Z) of the amplifier ($V = I \times Z_{amplifier}$). The magnitude of this signal is smaller than the actual action potential generated (site A) because of a voltage drop which occurs as the current flows through the recording electrode ($V_{drop} = I \times Z_{electrode}$). By making the amplifier impedance much larger than the electrode impedance this voltage drop can be made negligible.

impedance of the amplifier must be maintained 100 times larger than the impedance of the electrode.

Electrode impedance will vary with frequency and between different types of electrodes. In general, concentric needles will have a higher impedance than monopolar needles which in turn have a higher impedance than surface electrodes. The impedance of the electrode and the amplifier both decrease as frequency increases. To minimize waveform distortion and improve noise rejection, the 100:1 ratio of electrode-to-amplifier impedance should be maintained across the range of frequencies contained in the waveform under study. Impedance as high as 1 MΩ[27] at low frequencies can occur with concentric needles. Thus, to prevent signal degradation, amplifiers should have a minimum input impedance of at least 100–200 MΩ. With modern amplifiers, input impedance is less of a problem. An input impedance of 200–1000 MΩ in parallel with 5–10 pF of capacitance is common. Because of this capacitive component, the input impedance will decrease as signal frequency increases. As an example of this effect, 5 pF of capacitance will decrease the input impedance to a maximum of 320 MΩ at 100 cps, 32 MΩ at 1000 cps, and only 3.2 MΩ at 10,000 cps. Amplifiers designed exclusively for low-impedance surface electrodes used in somatosensory evoked potential (SEP) studies may not be suitable for needle EMG applications.

The ability to reject undesired common mode noise also depends on both electrodes (active and reference) being electrically matched. Unequal electrode impedance unbalances the electrode–amplifier input divider networks, converting some of the common mode noise into a difference signal that is amplified to the same extent as the neurophysiologic signal. Regardless of the technique used, surface, ground, and reference electrodes must not be corroded or dirty and there should be adequate skin preparation to keep impedance low.

Digital Instrumentation Principles. Two techniques of waveform display are commonly employed in electrodiagnostic equipment: analog oscilloscope displays and computer-based digital video display systems. When analog display techniques are used, the neurophysiologic signals can be directly displayed on an oscilloscope screen following amplification and filtering. This method of display results in a continuously varying waveform that is viewed in real time. This approach offers the advantage of a high-quality waveform image (i.e., smooth appearing) but is limited by difficulty in making measurements of various waveform parameters.

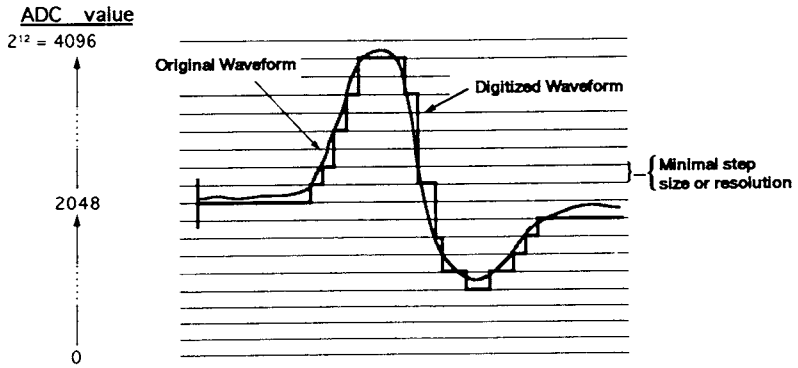
An alternative approach uses analog-to-digital conversion (ADC) and digital signal processing (DSP) techniques. Most contemporary electrodiagnostic equipment uses this method. Following amplification and filtering, the continuously varying

neurophysiologic signal is sampled at discrete time intervals and the amplitude of the signal is converted to a number (digital value). This process is called analog-to-digital conversion. ADC serves as the link or interface between the analog amplifiers and filters and the digital microprocessor-based portions of the instrument. If the sampling rate is fast enough, a series of digital values is generated which provides a close representation of the original analog signal. The advantage of using ADC and DSP is the increased ability to measure various features of the signal after it has been acquired. Changing the sweep and display sensitivity after waveform capture, automatic cursor placement, amplitude and area measurements, averaging, frequency analysis, and interference pattern analysis are examples of the features that are possible with digital waveform manipulation.

The resolution of the measurements made with ADC techniques is determined by the input voltage range and the number of bits of resolution of the analog-to-digital converter. A bit is a single binary digit, i.e., a 0 or 1. Groups of binary digits or bits are used to represent numerical values such as the changes in voltage of a neurophysiologic potential. Groups of binary bits are called bytes (8 bits) or words (16 or 32 bits). Typically, an ADC with 12 bits of resolution is used which can resolve 2^{12} or 4096 discrete amplitude levels. The actual voltage that corresponds to each of these discrete levels will vary depending on the amplifier gain. Effective resolutions will range from less than 1 μV to hundreds of microvolts as the gain changes from its highest to lowest level (Fig. 2).

Following the acquisition of a waveform, the numerical values representing the waveform amplitude over time are stored in electronic memory. In order to actually display the acquired waveform, the data values must be scaled so that an image of the waveform can be represented by a series of closely spaced dots or pixels on a video monitor (Fig. 3). In current electrodiagnostic instruments, the resolution of data acquisition is usually better than video monitor display resolution. The video monitor will typically have a full-scale display of 10 divisions with 30–50 vertical pixels per division. Thus the monitor is only capable of displaying 300–500 steps of amplitude resolution. If a 12-bit ADC was used to digitize the full-scale waveform, the 4096 available steps of digitized resolution must be scaled down to be displayed at the more limited screen resolution. Subtle changes in waveform features and morphology may be lost because of the limited number of pixels used to display the image; however, the advantage of this approach is that a captured waveform can be redisplayed at a greater sensitivity (i.e., 1 mV/div rather than 5 mV/div) without any loss in waveform accuracy. Electrodiagnostic instruments that allow this type of waveform manipulation most commonly use 12–16-bit analog-to-digital converters.

The series of digital numeric values which result from ADC must accurately reflect the ampli-



Amplitude step size for a 12 bit A to D Convertor

Gain s etting	10μV / div	20mV / div
Max. waveform (10 div)	100μV	200mV
ADC step size(resolution)	$\frac{100\ \mu V}{4096\ steps} = 0.0244\mu V$	$\frac{200\ mV}{4096\ steps} = 48.8\mu V$

FIGURE 2. Resolution of the analog-to-digital conversion process. The resolution or stepsize of A–D conversion depends on the full scale amplitude range of the input and the number of bits of the ADC. Since the number of bits is fixed, increasing the gain will allow smaller changes in the waveform to be resolved.

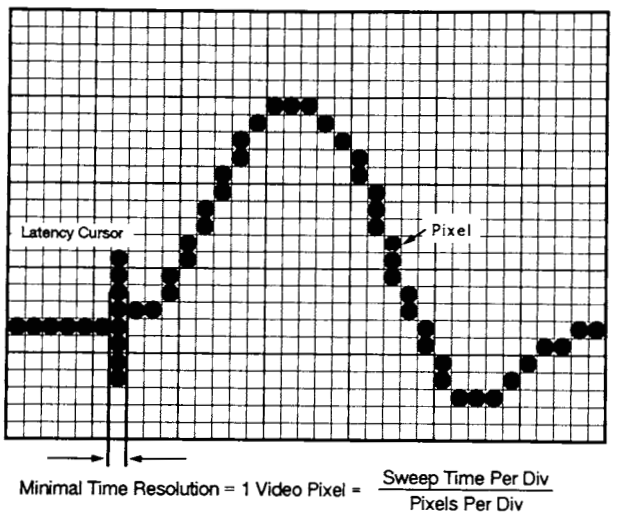


FIGURE 3. Waveform display resolution. Waveforms are displayed as a series of pixels or dots on a video monitor with the visual appearance depending on the number of pixels used to generate the image. The minimal latency or duration interval that can be appreciated is a single horizontal pixel whereas a single vertical pixel is the minimal amplitude change that can be seen.

tude, morphology, and variability in the actual neurophysiologic waveform. This requires that the sampling rate (rate of analog-to-digital conversions) be fast enough to capture the high-frequency, rapidly changing features of the waveform. A minimum sampling rate of more than twice the highest frequency component present in the waveform is needed to preserve all of the waveform information (Nyquist criterion). This sets the lower limit for the analog-to-digital conversion rate. If the sampling rate is insufficient, a form of signal distortion called "aliasing" occurs. This is illustrated in Figure 4. When a signal is sampled at too slow a rate, the signal components which occur at high frequencies will appear in the digitized data folded back at a lower frequency, that is, they appear under an "alias" at a lower frequency.

Simply achieving the Nyquist sampling rate may still not be sufficient to ensure that the morphology of the waveform is accurately reproduced after it is reconstructed on the video monitor as a pattern of pixels or connected points. Although digital data sampled at the Nyquist rate can in theory be used to reconstruct the waveform, the reconstruction process must be appropriately frequency limited. One way of doing this is to transform the series of digital values obtained by ADC to the frequency domain using the Fourier transform and then to reconstruct the waveform with the inverse Fourier transform. This approach is mathematically complex and does not work well for continuous signals such as free-running EMG, so instead, a much simpler method of creating the displayed waveform image is used. A "first order data reconstruction" method is used which consists of drawing a picture of the waveform by connecting the dots that correspond to the individual data samples. This process is conceptually simple and can be done very quickly as demonstrated by the rapid continuous waveform display of contemporary EMG machines. The trade-off is that the signal must be sampled at a more rapid rate than is dictated by the Nyquist criterion. To reconstruct accurately the high-frequency components of the waveform using this technique (connect-the-dots) requires a sampling rate of 5–10 times the highest frequency component.

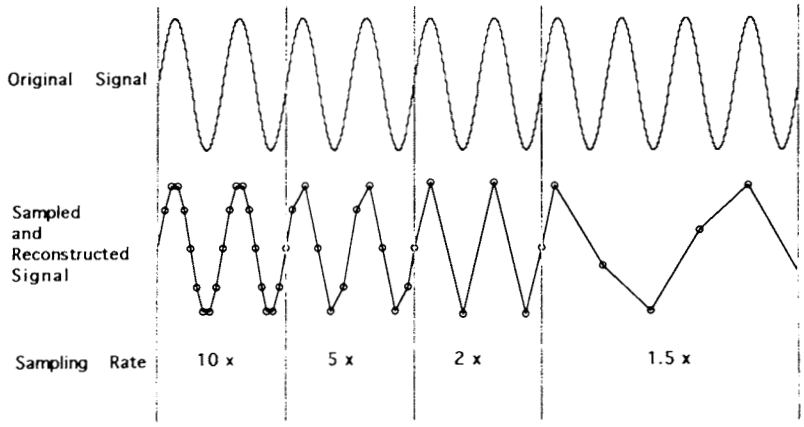


FIGURE 4. The effect of sampling rate. Sampling rates of 5–10 times the signal frequency result in visual reproductions that closely resemble the actual signal. When sampling at the Nyquist rate (twice the signal frequency), frequency information is preserved, but when the waveform is reconstructed by linear interpolation (i.e., drawing a straight line between the individual samples) significant distortion occurs, making the signal look like a triangle wave. Sampling rates less than twice the signal frequency can result in severe distortion and unrecognizable waveforms.

Gain and Sweep Settings Effect. Both the horizontal (sweep) and vertical (sensitivity) setting can influence latency and duration measurements of action potential waveforms.[5,8,14]

Visual appreciation of the point of departure of an action potential from the baseline is influenced by display sensitivity. High sensitivity or gain settings permits visualization of small deflections that are not seen at lower sensitivities. As demonstrated in Figure 5, an increase in the display sensitivity will allow smaller takeoffs from the baseline to be detected, resulting in a shorter onset latency.[8,14] Latencies obtained with a display sensitivity of 5 mV/cm may be as much as 0.5 ms longer than the same potentials measured at a sensitivity of 1 mV/cm.[14] Changes in peak latency are less affected by gain changes because the peak of the waveform is generally more easily identified over a range of settings.

A similar effect of display sensitivity occurs during the measurement of motor unit potential durations and is potentially a major source of error. As the waveform image is made larger by increasing the display sensitivity, the measured duration of the potential will increase as smaller and smaller deflections from the baseline are visualized (Fig. 6). This further complicates the quantitative measurement of motor unit action potential (MUAP) parameters and highlights the need to use standardized procedures for these measurements.

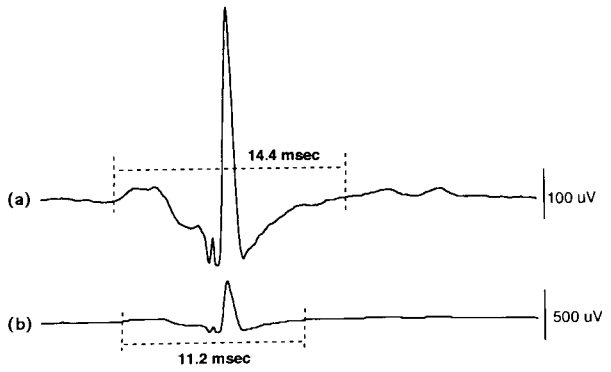


FIGURE 6. The effect of display sensitivity on MUAP duration. With an increase in the display sensitivity, smaller deviations from the baseline are apparent, resulting in a longer visually measured duration.

Changing the sweep setting has a less consistent effect on onset and peak latencies. Although typically, the onset latency will decrease as the sweep speed is increased[8] (i.e., changing the sweep from 5 ms/div to 2 ms/div), this effect is variable and generally of small magnitude.

When using digital EMG instruments, the horizontal display resolution will be limited by the number of horizontal pixels in each grid division. Typically, the video monitor will be capable of displaying 50–100 horizontal pixels per division. The waveform undergoes ADC at a rate that corresponds to the maximal horizontal pixel display requirements. As an example, if the fastest allowed

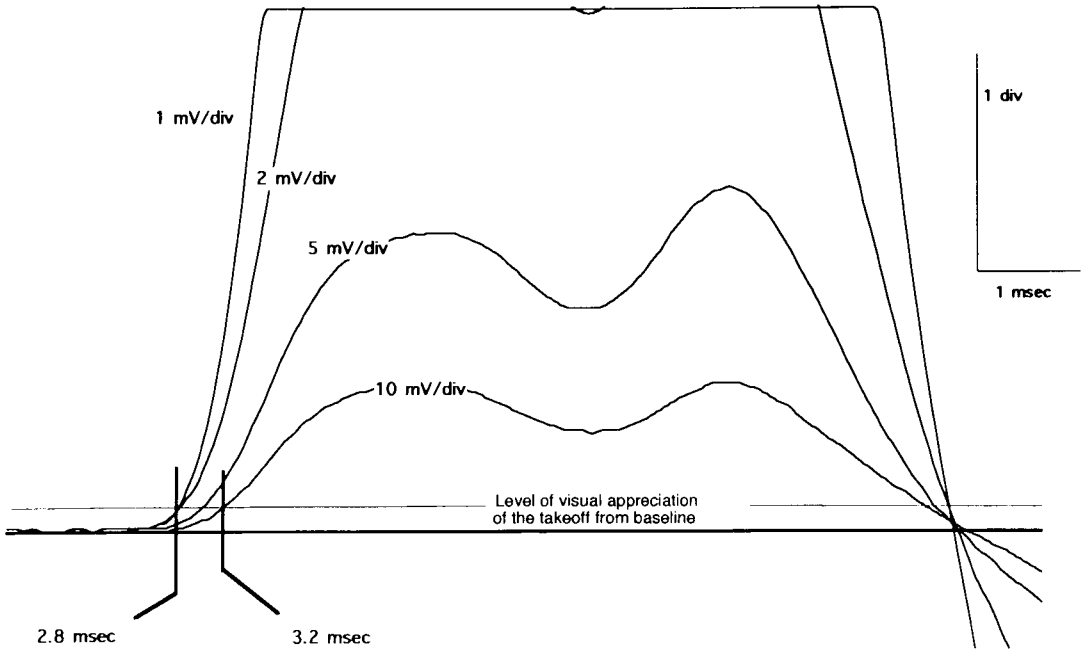


FIGURE 5. Effect of the display gain on the onset latency of a normal median CMAP. As display sensitivity is increased, a smaller deflection from the baseline of the waveform can be observed shortening the visual determined onset latency. In this example, a change in sensitivity from 10 mV/div to 1 mV/div shortens latency by 0.4 ms.

sweep is 1 ms/div and 50 horizontal pixels are displayed, a sampling rate of 50 samples per millisecond or 50,000 cps is needed to acquire 1 data sample for every horizontal screen pixel. When the sweep setting is changed to a slower sweep (i.e., from 1 ms/div to 2 ms/div), just 25 samples are needed for each 1-ms portion of the waveform. To redisplay the waveform at the 2 ms/div setting, only every other digitized waveform value needs to be displayed.

To make latency and duration measurements, a cursor is superimposed onto the waveform image displayed on the video monitor. The smallest cursor adjustment that can be visualized is movement of a single pixel forward or backward. Thus the minimal time interval or change that can be measured is the time associated with a single horizontal pixel. As before, if the display contains 50 pixels per division and the sweep is set at 2 ms/div, each pixel corresponds to 0.04 ms (2 ms/50 pixels = 0.04 ms). When making a measurement, the resolution of the measuring device should be at least twice that of the smallest change that is important to detect. Thus, to report a latency value with a clinical accuracy of 0.1 ms, a cursor resolution of 0.05 ms or less is needed. For most instruments this criterion can be met using a sweep setting of 2 ms/div or less. At slower sweeps (i.e., 5–10 ms/div), less latency accuracy is to be expected.

Many electrodiagnostic instruments are capable of making automated measurements of waveform latency, amplitude, and duration. When the algorithms that make these measurements use internal unscaled waveform data, which typically has better time and amplitude resolution than the displayed waveform image, the automatic measurements of latency and amplitude are relatively insensitive to the display settings. Excessive noise, large and prolonged stimulus potentials, or physiologic artifacts [e.g., contamination of antidromic sensory nerve action potential (SNAP) recording by motor unit potentials] can confuse the automated measured software, resulting in incorrect placement of markers. Visual verification of marker placement is always needed.

Triggers and Delay Lines. Triggers and delay lines are powerful tools used to isolate and display individual MUAPs. Using these techniques, changes in amplitude, duration, phasicity, and stability can be detected much more reliably than can be done from free-running EMG displays. Delay lines work by continuously sampling and storing into memory the ongoing EMG activity. When an MUAP occurs and exceeds a trigger value, typically set by the electromyographer to detect a peak in the potential, the MUAP is extracted from memory and displayed. The MUAP is displayed so that the trigger event is fixed or "time-locked" relative to the start of the sweep. The stable position of the potential makes averaging and qualitative and quantitative assessments of amplitude, duration, and phasicity accurate and reproducible. Delay lines also are useful in detecting motor unit potentials with late or early components (satellite potentials), instability, and blocking. Changes in waveform shape or blocking of individual spikes may be seen in a variety of neuropathic and neuromuscular junction disorders.[28]

More advanced signal processing techniques have been developed to automatically extract multiple motor units from a low-to-moderate force recruitment pattern.[7] A variety of MUAP identification algorithms are used that rely on both time and frequency domain features of the potential. This trend in quantitative motor unit analysis will become more commonplace as clinical electrodiagnostic equipment offers these features and as their clinical utility is documented. At the current time, there has not been adequate comparison studies of the various techniques to support one technique over another.

Signal Averaging. In certain clinical situations, such as the recording of SNAPs, particularly in patients with neuropathies or during the analysis of somatosensory evoked potentials (SEPs), the signal amplitude may be very small compared to background noise. Waveform averaging can be used in these situations to extract the neurophysiologic signal from the larger noise signal.

To average a neurophysiologic event, it must occur at a fixed time relative to a stimulus pulse or another trigger event which can be detected by the instrument. Following the stimulus pulse or trigger event, the waveform is digitized and stored in memory in the instrument. Sequential responses are mathematically summed, averaged, and subsequently displayed. When the undesired signal (noise) is randomly occurring, signal averaging will attenuate the noise, effectively increasing the size of the signal relative to the background noise. This signal-to-noise ratio (SNR) is related to the number of individual responses averaged:

$$\text{SNR} = \frac{\text{Signal Amplitude} \times \sqrt{\text{Number of Sweeps}}}{\text{Noise Amplitude}} \quad (2)$$

Not all undesired signals are randomly occurring. Episodic noise bursts often occur during SEP studies when the patient moves or tenses muscles in the vicinity of the recording electrodes or when the electrocardiogram signal is picked up by the electrodes. This type of noise can be eliminated by using a sweep rejection criterion. Rejection of a sweep occurs if the incoming signal exceeds a predetermined amplitude usually expressed as a percentage of the input range of the amplifier. Typically, sweeps should be rejected when the signal exceeds 95% of the input range.

Motor and motion artifacts can obscure antidromic SNAP or SEP recordings. These artifacts occur at a consistent time following the stimulus and are not removed by averaging. Sixty-cycle interference can usually be considered as a random signal but can become time-locked when it is an exact multiple of the stimulus frequency (i.e., 2.0, 3.0, 4.0, 5.0 cps). The use of a stimulus frequency such as 2.92 or 4.11 cps or random rate stimuli can prevent this effect.

The dependency of the SNR on the square root of the number of sweeps averaged means that the number of sweeps must increase by a factor of 4 in order to double the SNR. During an evoked potential study, a signal that cannot be identified with 500 stimuli because of background noise will require an additional 1500 stimuli to double the SNR. Time and patient tolerance are practical limits on the number of stimuli that can be used during clinical studies. Therefore, to obtain quality recordings, the use of good technique to maximize the signal amplitude (i.e., skin preparation to lower and match electrode impedance and accurate electrode placement) and to minimize noise (i.e., use of correct filter settings and adequate patient relaxation) is essential.

NOISE AND UNDESIRED SIGNALS

Noise can be considered as any signal that interferes with the recording of the desired waveform. During clinical studies, there are three major sources of noise: (1) electrical interference from power-line sources; (2) electronic amplifier noise; and (3) unwanted background electrophysiologic activity.

Power-line interference (50/60-cps sine wave noise) is frequently the most troublesome source of noise encountered in routine clinical studies. This noise is introduced into the recording system by two mechanisms: capacitive and electromagnetic coupling. Small capacitances in the 3–300 pF range exist between the patient, recording leads, power lines, and earth ground.[16] This capacitive coupling allows small interference currents to flow into the patient, the amplifier circuits, and/or the recording leads. Power-line noise can also be magnetically coupled into the amplifier if a loop is formed by the orientation of the recording electrode wire leads or by a combination of the input leads and the patient. Fortunately, most of the 60-cps interference appears as a common mode signal and is rejected. Some of this interference current will actually flow through the soft tissues between the active and reference electrodes, however, generating a difference voltage that will be amplified and displayed. This component of the power-line interference cannot be eliminated by increasing the common mode rejection ratio (CMRR) or other amplifier characteristics and must be reduced by decreasing the amount of noise that is coupled into the patient. Any active-reference electrode impedance mismatch will also contribute to the 50/60-cps interference signal.

The high-input impedance and CMRR of contemporary electrophysiologic amplifiers eliminate the majority of power-line noise. Additionally there are a number of techniques which can be used by the electrodiagnostic medical consultant to further reduce power-line interference. The most easily applied of these techniques include: (1) reducing skin–electrode impedance at *both* recording electrodes and the ground by abrasive skin preparation; (2) using short recording leads and/or twisted pair leads to reduce magnetically coupled interference; and (3) turning off unnecessary electrical equipment and empirically moving the patient to different locations in the room to find the "quietest" area since 50/60-cycle interference is not uniform within an examining room and is highest near electrical equipment (often including the EMG instrument itself) and power lines within walls. In addition, changing the physical orientation of the amplifier and recording leads can reduce or better match the coupled 60-cycle noise, minimizing its effect on the recorded signal.

Persistent power-line noise may be helped with the following measures, though some are not applicable in every situation: (1) checking for breaks in the electrode wire, which are usually hidden by the wire insulation, swapping leads or using an ohmmeter to identify a bad lead; (2) matching the impedance of the reference and active electrodes, which is most problematic during needle studies because the active recording electrode usually has a different impedance than the reference electrode (when using monopolar needle electrodes, substitution of a second needle inserted subcutaneously as a reference usually improves impedance

match); and (3) electrically shielding the examination room, although this is costly and rarely if ever required with modern electrodiagnostic equipment.

The second noise source, internal electronic amplifier noise, can also be a problem. This noise typically has a broad frequency spectrum and increases in amplitude at higher frequencies. It appears as baseline fluctuations (fuzziness) and an audible hissing when sensitivities of 10–20 μV/div or greater are used (see Fig. 9 of Part I). When recording small-amplitude sensory responses, particularly in diseased nerves, this noise can overwhelm and obscure the potential. Amplifier noise specifications refer to the magnitude of this internally produced noise with typical values ranging from 3 to 10 μV (peak-to-peak). This source of noise can be reduced by using low-noise amplifiers, lowering the high-frequency filter cutoff during sensory and SEP studies, and averaging to improve the signal-to-noise ratio.

The third type of noise is background electrophysiologic activity. It is important to recognize that the terminology "active" and "reference" are conventions which allow for uniformity in recordings between laboratories. From an amplifier standpoint, both inputs are equally active and can contribute to both the signal and noise which are recorded and displayed. Background activity from incomplete muscle relaxation beneath the reference electrode will be amplified and can contaminate and mask the signal being generated at the site of the active electrode. Patient positioning to ensure good relaxation, locating the reference electrode over a tendon or bony area away from potential muscle contraction, and using the audio to monitor for unwanted activity during sensory NCV and SEP studies can be used to minimize this type of noise.

NERVE STIMULATION

Following a stimulus pulse, electrical currents will be distributed in the volume conductor (extracellular space) around the peripheral nerve.[4] At low intensities most of the current flow is in the superficial soft tissues. As stimulus intensity increases, more of the current will enter the axon at the cathode, flow longitudinally, and exit at the anode. Depolarization of the membrane occurs beneath the cathode while hyperpolarization is occurring at the anode. If the stimulus intensity is large enough and its duration is long enough, threshold is reached at the cathode and an action potential is generated. Standard stimulation techniques require the placement of two small surface electrodes (cathode and anode) on the nerve 2–3 cm apart. The cathode is placed closest to the recording electrode. Traditionally, nerve stimulation has been accomplished using electrical stimulators. More recently, magnetic stimulators have been used to allow stimulation of both peripheral nerves and central nervous system neurons.

Two basic types of electrical stimulators are routinely used, i.e., constant-current and constant-voltage stimulators. Electrical stimulators are typically capable of generating voltages between 0 and 300 V, currents ranging from 0 to 60–100 mA, and stimulus pulse durations of 0.1–1 ms. Constant-voltage stimulators deliver a fixed voltage between the anode and the cathode. Because skin impedance varies from stimulus to stimulus and even during the short interval that the voltage is applied, the resultant current flow may not be constant. Constant-current stimulators deliver a constant current (milliamperes) to the patient for a broad range of stimulating electrode impedances. For this reason, the magnitude of the current strength delivered by a constant-current stimulator is more stable than that delivered by a constant-voltage stimulator and is more usable for stimulus-threshold measurements.

Technical Considerations in the Use of Electrical Stimulators. In vitro, following electrical stimulation, nerve fibers are depolarized in order of their size: large-diameter having lower thresholds than small-diameter axons. This relationship, however, is less apparent in vivo because of variability in anatomy, the position of nerve fibers within the nerve, volume conduction spread of stimulus currents, and possibly changes caused by pathology.[18] During clinical studies, complete depolarization of the nerve is needed to insure that the maximal conduction velocity will be measured. This is typically achieved by using supramaximal stimulation. The stimulus strength is increased until the evoked amplitude is maximized and, then by convention, further increased an additional 10–30%.[4,18]

Caution must be used when employing supramaximal stimulus strengths because the spread of excessive stimulus current can shift the actual site of nerve activation. The presumed site of nerve depolarization is beneath the stimulator cathode but when large currents are needed because of nerve pathology the stimulus site may be displaced by up to 12 mm. Using a current strength twice that needed to achieve a just maximum response can reduce the measured latency by 0.2 ms due to

distal stimulation of the nerve (Fig. 7). The problem of current spread and displacement of the actual site of nerve depolarization is particularly problematic when performing short segment "inching" studies where changes of several tenths of a millisecond in latency may be clinically important. Placement of the stimulator directly over the nerve and the use of just maximal intensities can help maintain accuracy.

Large or excessive stimulus currents can also cause inadvertent activation of neighboring nerves. Spread of excitation to neighboring nerves causes changes in waveform shape, amplitude, and latency that are confusing and easily misinterpreted. Though sometimes obvious, this can be quite subtle. Overflow can usually be detected by carefully observing the waveform shape. As stimulus strength is increased, a plateau in the waveform amplitude will be seen with small increments of stimulus intensity. This is followed by a secondary increase in amplitude or a change in waveform shape as the intensity is further increased and overflow begins. Also the appearance of an initial positive deflection in the evoked response not present at lower stimulus intensities may represent a volume-conducted response originating from distant muscles innervated by the inadvertently depolarized nerve. Regions where nerves are in close anatomic proximity such as the wrist, axilla, Erb's point, and the popliteal fossa are sites where overflow stimulation most frequently occurs.

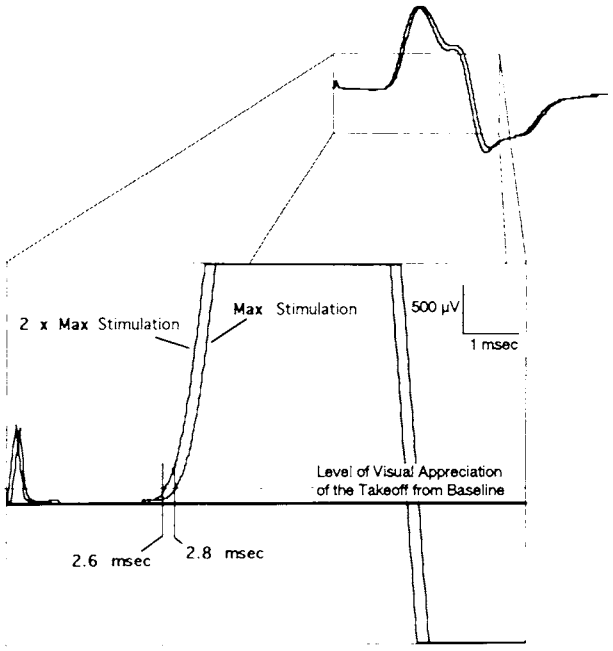


FIGURE 7. Effect of excessive stimulus intensity on the onset latency of a normal ulnar CMAP. Changing the stimulus intensity from just supramaximal to twice this value shortens the onset latency by 0.2 ms. Further increases in intensity will cause further reductions in onset latency.

An unintentional reversal of polarity of the stimulating electrodes (i.e., the anode placed closest to the recording electrode) will effect latency and conduction velocity values.[3,8] For example, assuming a 2-cm distance between the cathode and anode and a 50-m/s conduction velocity, an unsuspected stimulus polarity reversal will result in a latency 0.4 ms too long ($latency_{change}$ = anode–cathode distance/NCV). For a 20-cm segment of nerve, an error of this magnitude will result in a 5.5-m/s velocity error, if the polarity reversal occurred only at one of the stimulation sites.

During routine motor and sensory NCV studies, short duration stimulus pulses are preferred to decrease stimulus artifact, minimize discomfort, and reduce the uncertainty in the actual time of nerve depolarization. H-reflex studies are the exception as they are best elicited with stimulus durations between 0.5 and 1 ms.[13,20] The large diameter group Ia sensory afferents within the tibial nerve appear to be preferentially stimulated because of increased sensitivity to longer-duration stimulus pulses.[3,20,25] Using a through-the-knee stimulation technique, H-reflexes can be reliably obtained at relatively low stimulus intensities even in large individuals. By placing the cathode over the tibial nerve in the popliteal fossa and the anode on the medial side of the patella, current flows through the limb, easily depolarizing the tibial nerve.

Stimulus Artifacts. The stimulus artifact is a (large-amplitude) voltage spike with an exponentially decaying tail that is generated as a result of stimulus current flow in the volume conductor near the recording electrodes. When recording small amplitude responses such as the SNAP, the appearance of the SNAP "riding" on the stimulus artifact tail often makes determination of amplitude difficult and may alter latency values. It is generally hard to predict the actual shape, polarity, or amplitude of the stimulus artifact because multiple factors contribute to its generation.[10,15] The first, and an important, mechanism is the flow of stimulus current between the active and reference recording electrodes. This generates a voltage difference between the active and reference input that will be amplified and displayed. Moving the stimulator closer to the recording electrodes increases the magnitude of the stimulus current flow near the recording electrodes, resulting in a more

pronounced artifact. Second, any recording electrode impedance imbalance will convert common mode stimulus signal into a stimulus artifact proportional to the impedance mismatch. Third, capacitive coupling between the stimulating and recording leads contributes to artifact. The artifact due to these three mechanisms tends to be brief in duration (less than 1 ms) and is the initial spike seen at the start of the sweep. The more prolonged, slowly decaying stimulus tail is usually the result of amplifier saturation and slow recovery caused by the storage of charge in capacitive components of the amplifier. A number of techniques that can be used clinically to decrease the stimulus artifact are listed in Table 1.

Substituting a needle electrode for the stimulator cathode reduces the voltage and current required for nerve depolarization, usually decreases artifact, and improves localization of the actual site of nerve stimulation especially for deep nerve segments.[18] Some degree of caution is required when using needles for stimulation because excessive current densities at the tip of the stimulating needle cause hydrolysis and by inference potential tissue damage (Appendix B). There is, however, little direct in vivo experimental evidence to support any recommendations on the safety of using monopolar EMG needles as stimulating electrodes. Empirically, monopolar needles are used frequently without any apparent adverse effect and specific techniques exist which rely on their use as a stimulating electrode.[6,21] The margin of electrical safety can be improved by stripping the Teflon insulation back 3–4 mm to distribute the current over a larger area. Concentric EMG electrodes should not be used because the small area of the central electrode core leads to much higher current densities.

ERROR SOURCES AND NORMAL VALUES

Normative values for NCV (and MUAP) parameters will vary between studies, in part because of differences in sweep, gain, and filter settings. One setup is not necessarily better than another; standards of normal and abnormal are simply dependent on the particular settings used. Using normal data from other laboratories or from published studies, without adopting similar techniques and corresponding instrument settings, can result in misleading results. Even within a single laboratory, normal values will vary if amplifier sensitivity is altered. Since it is unrealistic for all laboratories to establish their own normal values, individual electromyographers should assure themselves that normal values obtained with their usual NCV and EMG techniques conform to published values before such "normal" values are blindly adopted. Recently, even the use of the mean ± 2 SD values for NCV latency and amplitude measurement has been questioned.[22] Waveform parameters for many commonly performed nerve conduction studies are not normally distributed and use of raw data from control subjects in determining normal limits may result in a high rate of misclassification of patient results.

For conduction velocity determinations there

Table 1. Techniques used to reduce stimulus artifact during NCV studies.

Minimize the current needed for nerve depolarization.	Use short duration and just supramaximal stimuli.
	Decrease stimulator–skin impedance by abrasive cleaning of the stimulation site.[15]
	Substitute a needle electrode for the stimulus cathode.*
Decrease the stimulus current and voltage gradient at the recording electrodes.	Use electrically isolated stimulators (standard on most modern instruments).
	Decrease and match recording electrode impedance with abrasive skin preparation.[15]
	Position the ground between the stimulator and recording leads.
	Remove perspiration and excess electrode paste to prevent stimulus current conduction along the skin surface.[24]
	Rotate the stimulator anode to adjust the orientation of the stimulus equipotential lines until the artifact is minimized.[10,12,24]
	Use short stimulator and recording leads, widely separated from each other, to minimize capacitive coupling.[10]
Stimulator design techniques.	Constant current stimulators generally have smaller artifacts compared to constant voltage stimulators.[10]
	Use amplifier blanking circuits,[1,9,23] direct coupled amplifiers,[26] biphasic stimulus pulses,[12,17] and/or digital stimulus artifact subtraction.[2]

**See text for caution regarding use of needle stimulation.*

are additional sources of error due to inaccuracies in distance (D) and latency (t) measurements.[14] The importance of these errors is under-appreciated and can lead to misinterpretation of clinical studies particularly when "abnormalities" are mild or borderline. The experimental error in a conduction velocity (CV) measurement is given by the expression:

$$\Delta CV = \left(\frac{\Delta D}{D} + \frac{\Delta t}{t}\right) \times CV \qquad (3)$$

where ΔCV = conduction velocity error, ΔD = distance error, and Δt = latency error.

Errors in time (latency) measurements between two stimulation sites result from the factors previously discussed: uncertainty over the actual waveform departure from baseline, gain and sensitivity effects, filter settings, stimulus artifacts, and excessive stimulus intensity. Distance measurements have fewer sources of error and generally contribute less to the CV error than do time errors. Distance measurement errors, however, become especially problematic in several situations: (1) when very short interstimulus distances are used such as occurs in inching studies ($\Delta t/t$ errors also increase in this situation); (2) when the anatomic course of the nerve is uncertain as occurs following nerve transpositions; and (3) when the nerve segment under study crosses the extensor surface of a joint (i.e., ulnar nerve at the elbow). In this situation redundancy in the nerve makes measurements of distance dependent on the joint angle.

Errors of 0.5 ms in time difference and 3–4 mm in the distance are reasonable estimates of measurement error.[14] As can be inferred from eq. (3), the clinical significance of these errors depends on the magnitude of the latency and distance error *relative* to the total segment latency change or distance. For example, a 4-mm error over a 2-cm distance (as used in inching studies) represents a 20% error, whereas the same error over a 20-cm nerve segment is only a 2% error. Figure 8 illustrates the effect these errors will have on CV calculations. These error considerations also imply that a single latency measurement at a fixed distance would yield a more accurate measure of disease progress over time than would a conduction velocity determination.

Not discussed in this monograph are additional sources of variability in NCV and EMG studies including temperature and age effects.[11,19] The many sources of potential error and variability when acquiring and measuring neurophysiologic

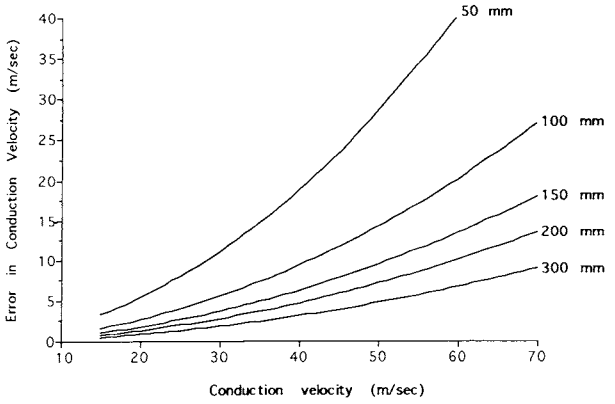


FIGURE 8. Error in conduction velocity resulting from inaccuracies in the measurement of interstimulus latency difference and distance. The family of curves reflect the CV error for a conduction time error of 0.5 ms and a distance error of 3.6 mm for interstimulus distances ranging from 50 mm to 300 mm.

potentials dictates that a reasonable degree of caution be used in evaluating mild or borderline changes, especially when mild abnormalities occur in neurologically normal patients or an abnormality exists in isolation (i.e., mild slowing without other signs of demyelination such as temporal dispersion or conduction block).

SUMMARY

This monograph has explored the influence of instrumentation and measurement issues on the detection, acquisition, display, and characterization of nerve and muscle action potentials encountered during clinical studies. The major areas covered included:

1. Distortions in waveforms that result from inadequate high- and low-frequency response.
2. The influence of electrode characteristics on waveform parameters.
3. Effect of amplifier gain, sensitivity, and sweep on waveform measurements.
4. Noise sources and techniques used to minimize this source of error.
5. Basics of nerve stimulation.
6. Technical and measurement factors affecting the accuracy of conduction velocity studies.

An appreciation of the sources and magnitudes of the various errors and techniques which can be used to minimize their effects will improve the electrodiagnostic physician's ability to acquire accuracy data. Knowledge of these technical limita-

tions will prevent overinterpretation of subtle and borderline abnormalities that may merely represent artifact or "experimental" error associated with performing a clinical study of a nerve or muscle.

APPENDIX A

Leakage Current and Patient Safety. An electrodiagnostic instrument can be a potential source of electrical shocks to both patients and equipment users. Electrical equipment is typically powered by 110-V, 60-cps (220-V, 50-cps in Europe) line current. The electrical connections at a wall outlet consist of a "hot" wire at 110 V, a neutral wire connected to earth ground, and a third safety ground also connected to earth ground. Within the EMG machine a power supply converts this line voltage to lower DC voltages, typically between 5 and 15 V, which are used to power the amplifiers, filters, and computer circuits. Direct physical contact with the high voltage points in the power supply sources is extremely unlikely because these circuits are physically isolated from any part of the equipment that the user or patient contacts. Because insulators are not perfect and alternating currents can be capacitively coupled from the power-line cords or power-supply transformers, however, small "leakage currents" can flow to the machine chassis and low voltage portions of the instrument.

This leakage current is potentially dangerous. Should a grounded patient touch the chassis, the leakage current will flow through the patient to ground, possibly resulting in harm. This potentially dangerous situation is prevented by using the safety ground which is electrically connected to the instrument chassis and serves as a low-resistance pathway to ground for leakage currents. There are several not uncommon situations where leakage current dangers can still exist. The first of these occurs when the chassis ground wire is broken or disconnected. Now the patient becomes the only pathway to ground for the leakage current and a shock hazard exists. Another potentially dangerous situation can occur when the patient is in contact with other electrical equipment such as an electric bed. Should the safety ground on any of these pieces of equipment fail, leakage current from them can flow through the patient via the ground electrode.

To preserve patient safety, standards have been established for the maximum allowable leakage currents. The ANSI/AAMI standard permits a maximum of 100 μA of leakage current from chassis to ground, and 50 μA of leakage current from patient input lead receptacles to ground (nonisolated inputs). These standards have evolved from experimental studies which show that a 20 μA current at 60 cps, if applied directly to dog heart through a small contact area (such as provided by a cardiac catheter or pacemaker), can induce ventricular fibrillation. All electrodiagnostic equipment should be periodically inspected by a qualified biomedical equipment technician to verify the integrity of safety grounds and to ensure that leakage current limits are not being exceeded.

The preceding discussion is of particular importance to older electrodiagnostic equipment in which the patient ground electrode is actually connected to earth ground. Most contemporary instruments use isolated amplifiers and patient connections. In these machines, all patient leads are electrically isolated from the power lines and earth grounds. This isolation commonly takes the form of optical isolation in which the neurophysiologic signal voltages are converted to an optical signal, transmitted across an electrical barrier, and then reconverted to voltage. Other isolation schemes using capacitive or magnetic coupling are equally effective. The use of isolated connections is a major improvement in patient safety, and makes it extremely unlikely that a dangerous level of leakage current can flow through the patient to ground. Standards for isolated amplifier equipment limit current flow from patient leads to ground to 10 μA, even in the event the patient leads come into direct contact with the hot 110-V 60-cps power line.

APPENDIX B

Hydrolysis Danger during Stimulation with Needle Recording Electrodes. Single randomly selected monopolar and concentric needle electrodes were studied in the author's (WCS) laboratory. The needles were mounted with half the shaft of the needle submerged in 0.9% saline.

In the monopolar experiment the needle was connected to the cathode of a standard constant voltage stimulator. A 1-cm disk electrode immersed in the bath was connected to the anode. Using a constant 1-cps stimulation rate, the voltage and then the pulse duration were increased while measuring current flow. At a voltage strength of 25 V and a pulse duration of 0.2 ms, hydrolysis was visually noted. The peak current was measured at 23 mA. From tip area measurements the peak current density was calculated to be 33 A/cm^2. The energy released with each pulse was calculated to

be 52 cal/cm^3. Visible sparking (ionization) of the hydrogen appeared at a voltage strength of 270 V with a 0.5-ms pulse duration.

In the concentric electrode experiment, the core was connected to the cathode and the cannula to the anode of the stimulator. Hydrolysis was noted at 160 V for a 0.1-ms duration and at 75 V for a 0.2-ms pulse. Ionization appeared at 280 V for a 0.1-ms pulse, and at 125 V when pulse duration was 0.2 ms.

The stimulus intensity at which hydrolysis or ionization occurs for a given needle will vary with its actual tip surface area. The greater the area, the greater the voltage and pulse duration required. While it is not known what tissue damage may result from hydrolysis and/or ionization, monopolar EMG needles can be used with a greater margin of safety if the insulation is stripped 3–4 mm back from the tip. Concentric EMG electrodes cannot be so modified and hence should not be used for stimulation.

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