

Neural Blockade Anesthesia Monitor

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The ability to monitor the effectiveness of an anesthetic is an area that has not kept pace with the rest of medical technology. Using traditional systems engineering methods, a prototype monitor for local anesthetics was successfully designed and validated. This article describes the clinical physiology of local anesthetics, the selection of technologies to measure the physiological responses, the monitor design process, and the results of clinical trials. The subject neural blockade monitor provides a noninvasive, real-time tool for the anesthesiologist. Synergistic blending of systems engineering and biomedical engineering resulted in a highly effective prototype that was developed in a relatively short period of time.

INTRODUCTION

Significant progress has been made in the use, efficacy, and safety of anesthesia. However, clinical tools are not currently available to specifically monitor or objectively evaluate the effectiveness of local anesthetics in humans or animals. This article describes a technology that has the ability to monitor the effects of local anesthetics on the patient and inform the clinician regarding patient condition from onset to termination of neural blockade.

The use of local anesthetics, which block the signals carried in a nerve from the site of pain to the brain (hence the term “neural blockade”), has greatly enhanced the safety and quality of anesthesia for more than a century. Improvements in technique (e.g., nerve stimulation guidance), devices (e.g., needles), and anesthetic agents have also been made. Nevertheless, no routinely used monitor enables the clinician to characterize the effect of the block in the patient.

The ability to collect and interpret data to characterize the effect of blocking agents would give the clinician an unparalleled means to limit toxicity, improve dosing regimens to enable care to better approximate analgesic need (both in location and quality), manage uncooperative patients (adult asleep, adult uncooperative [psychological or defiant], pediatric uncooperative [developmental or communication barrier], pediatric asleep), and better refine techniques (e.g., to improve discrimination between motor/sensory blockade [differential blockade]) for patients.

BACKGROUND

Imagine getting a dose of local anesthetic and the best way for the physician to know if you are numb is to pinch you! Unfortunately, this is pretty much the current “state of the art.” Despite this, modern anesthetic practice has vastly improved the safety of performing

complicated and challenging surgery, even in the sickest patients. This is the case, in part, because the anesthesiologist can effectively manage patient consciousness, blood pressure, heart rate, and temperature. However, in the domain of neural blockade, the ability to precisely monitor such effectiveness has not been available.

For example, the use of axial (i.e., epidural or spinal) anesthesia has become commonplace, yet no method exists to routinely know what part of the body is affected, and how “numb” it is, using acceptable, routine, objective measurements. Simple subjective tests such as needle sticks, pinches, or cold stimuli at various parts of the body can suggest where and to what degree a patient has an effective block, but the need for active patient participation presents confounding difficulties when a patient is unable or unwilling to respond to such tests. Moreover, currently available tests are qualitative and show large variation with respect to reliability, repeatability, and interpatient correlation.¹ Studies comparing anesthetic agents offer limited references to the effectiveness of using quantitative monitors as a measure of clinically relevant neural blockade.² Perception of pain or cold varies among people and is not readily correlated with an objective standard. It is also particularly difficult to assess using today’s relatively simple sensor and signal analysis technologies. We postulated that, as assessed on any level greater than single neurons and action potentials, the density or completeness of a block at a particular dermatomal level can be measurable and informative.

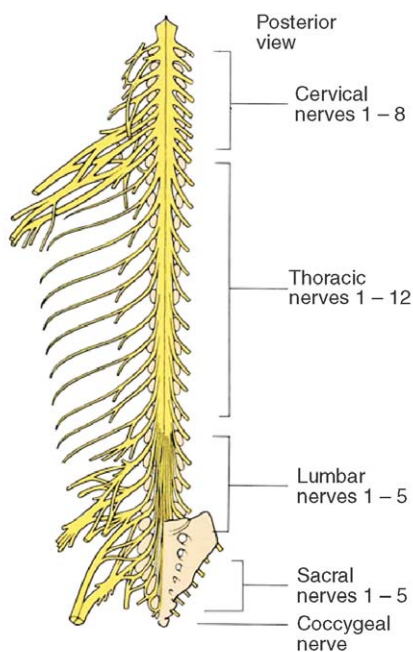


Figure 1. The spinal cord and associated spinal nerves.

Physiology 101

The nervous system is composed of the central nervous system (CNS), including the brain and spinal column, and the peripheral nervous system (PNS), including the cranial and spinal nerves. The spinal cord provides a transmission pathway for impulses to and from the brain while also serving as the integrating center for spinal cord reflexes. Thirty-one bilateral pairs of spinal nerves radiate from the spinal cord and service sensory and motor activation functions.

The spinal nerves are named according to the respective vertebrae of origin: there are 8 cervical, 12 thoracic, 5 lumbar, 5 sacral, and 1 coccygeal (Fig. 1) vertebrae. Cervical nerves service the head, neck, shoulders, arms, and diaphragm. Thoracic nerves service the arm (T1) and trunk of the body (T2–T12). The lumbar and sacral nerves service the hips, pelvic cavity, and legs. The coccygeal nerve, when present, serves no specific function.

The autonomic nervous system (ANS) is a specialized portion of the PNS that is further composed of two divisions: sympathetic and parasympathetic (Fig. 2). The sympathetic division is the “action” part of the ANS and is responsible for the physiologic responses apparent in stressful situations. Fear, anxiety, and, of course, pain cause sympathetic responses such as increased heart rate, vascular constriction (leading to high blood pressure), and shunting of blood away from structures

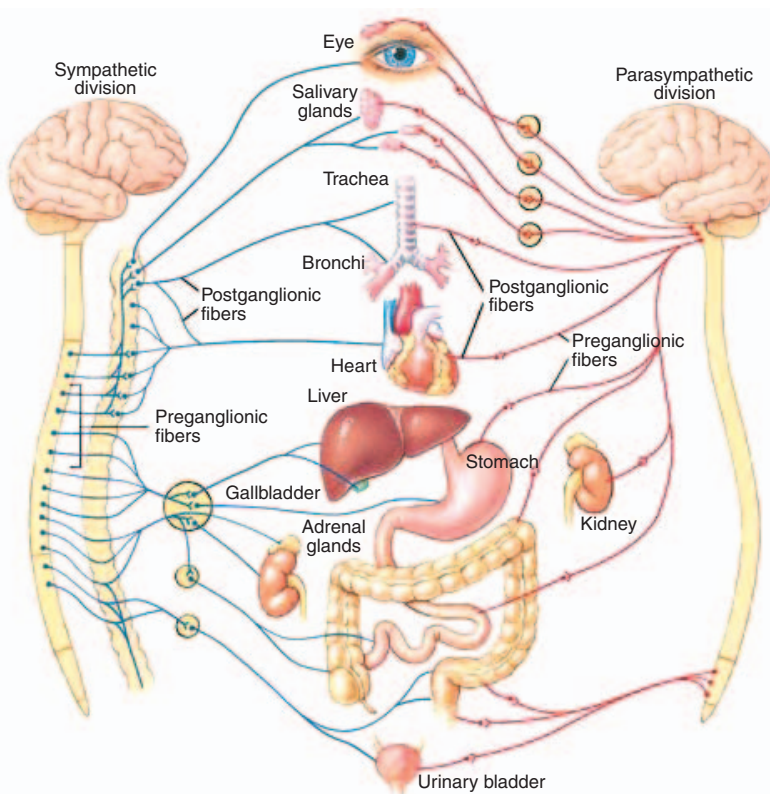


Figure 2. Functional diagram of the autonomic nervous system. (Reproduced, with permission, from <http://www.mhhe.com/socscience/intro/ibank/ibank0111.jpg>, © The McGraw-Hill Companies, Inc.)

nonessential for “fight or flight,” such as the intestines. The parasympathetic division is dominant under non-stressful conditions. Under normal conditions parameters such as heart rate and blood pressure decrease and digestion increases.

Anesthesiology 101

Sympathetic, sensory, and motor nerves (but not parasympathetic nerves) are affected by local anesthetics that are commonly injected into the epidural space (Fig. 3). Sensory nerves carry information from receptors in the periphery to the CNS regarding various stimuli: heat, light and deep touch, position sense, and pain (nociception). Motor nerves carry information from the brain to the muscles to control and move them.

The primary objective when using regional neural blockade such as epidural anesthesia is to prevent nociceptive impulses from reaching the brain or CNS.³ Anesthetics prevent or block impulses by directly influencing the sympathetic, sensory, and motor systems. Because of differences in nerve type, location, and size, sympathetics are most likely to be blocked first, sensory next, and finally, motor. Decreasing block density is realized as the drug diffuses and travels along the epidural space, away from the infusion/injection site⁴ (Fig. 4). In most cases, the effect realized farthest from the infusion site appears

to block only the sympathetic nervous system.⁵ Sensory block (anesthesia/analgesia) appears over an area closer to the injection site and progresses both toward the head (cephalad) and away from the head (caudal). Motor block usually develops in the area nearest the injection site and, depending on local anesthetic concentration, may have a shorter overall latency.⁶ Blockade at the higher thoracic spinal cord levels, which regulate cardiac and respiratory functions, may not be desirable. The balancing influences of the parasympathetic system are controlled from locations in the brain stem and the sacral segments of the spinal cord; therefore, axial neural blockade (as in an epidural or spinal block) normally has no influence on higher cardiac and respiratory functions.

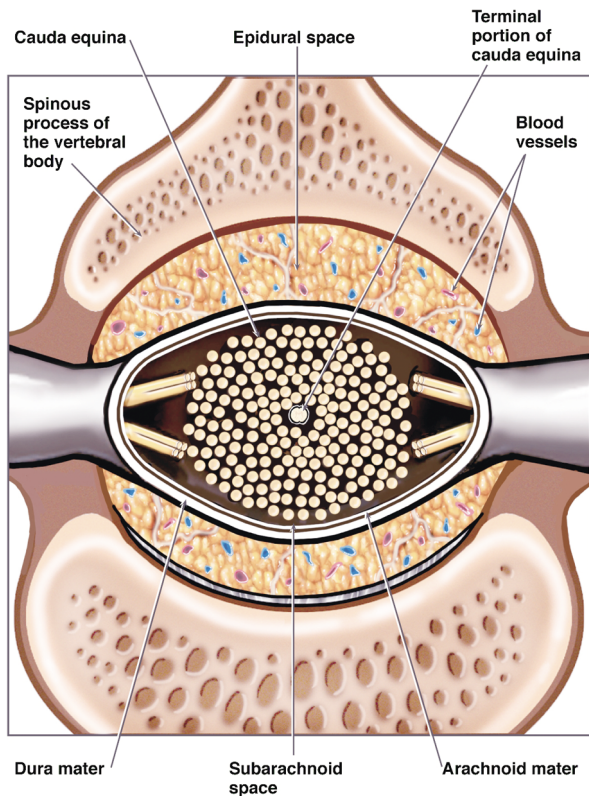


Figure 3. Transaxial (superior) view of a lumbar vertebra (L2–L5), showing the location of the epidural space. (Reproduced, with permission, from www.nucleusinc.com, © 2004 Nucleus Medical Art, Inc. All rights reserved.)

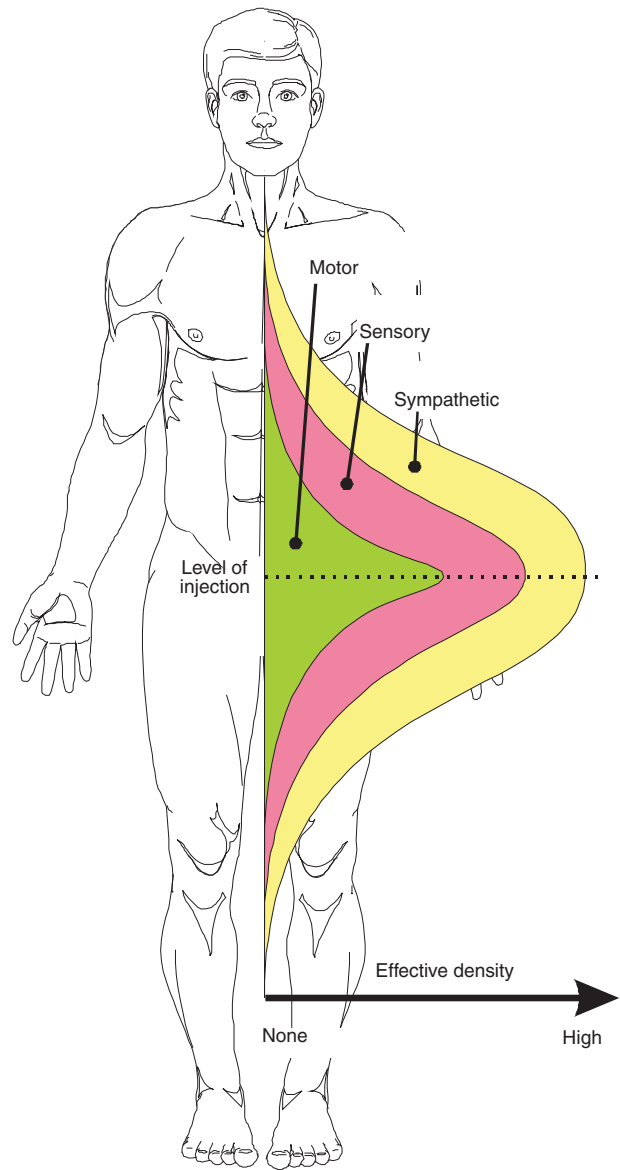


Figure 4. The anatomical level and density effects of neural blockades on sympathetic, sensory, and motor nerves. The effective block density decreases at levels farther away from the point of injection of the anesthetic.

CONCEPT DESIGN

Requirements

The starting point of the systems engineering process is to conduct a needs assessment and identify requirements. In this case, the medical need is to perform a procedure without adverse effects or complications resulting from inappropriate doses of anesthetics. In its simplest form, the clinician needs to know what nerves are blocked, when they are blocked, and to what extent they are blocked. In a clinically relevant context, this means knowing if the block is active in the anatomic location overlying the pain and that the density of the block is adequate to provide the necessary relief. The measurement system design requirements are derived from the clinical requirements. The challenge to the system designer is to measure the effect of the anesthesia as inferred from the physiological responses of the sympathetic, sensory, and motor nerve systems. This equates to understanding the cause and effect relationship associated with neural blockade anesthetics.

Measurable Physiological Responses to Neural Blockade

Physiological responses to neural blockades have been documented in detail.⁷ This section summarizes specific physiological responses that are measurable by direct and external methods.

Cardiovascular

The balance between the sympathetic and parasympathetic systems results in changes in relative heart rate and associated heart rate variability. Blockage of the sympathetic functions preclude direct neurogenic vasoconstriction, while other physiological responses produce an increase in blood volume in the blocked region. A relative increase in temperature and decrease in regional blood pressure are expected in blocked regions.⁸ The increase in blood volume associated with vasodilation increases the volumetric density of hemoglobin in a measured volume of tissue. Changes in the reflectance or absorption spectrum of the hemoglobin pigment may be measurable as an indirect indicator of vasodilation.⁹

Gastrointestinal

The absence of sympathetic tone (unbalanced parasympathetic efferents) would be expected to cause an increase in peristaltic activity and secretion as well as a relaxed pylorus.⁴ These changes may be measured by an electrogastrogram (EGG). A difference in the nature of digestive tract sounds can also be associated with this change.

Pulmonary

Independent of respiratory rate, the relative dimensions of airway passages and structures may be

influenced, and spasmodic effects have been shown to be reduced.¹⁰ Measurable results of the dimensional variation can include changes in the amplitude and spectral content of breathing sounds.

Neurological

Blockade of the motor nerves will change active and passive muscular activity, which can be monitored. An electromyogram (EMG) is a measurement of electrical activity associated with muscle actions. Rectified-integrated EMG (RIEMG) is shown to be related to the density of the blockade and has been used as a standard for comparative evaluation.¹¹ Active nerve conduction changes are manifest in changes in the skin conductance response.¹² The relationship is such that reduced conductance response, measured with a constant voltage monitor, is realized with increased density of the blockade. Skin resistance, measured with a constant current monitor, is similarly influenced, but in a less repeatable manner than skin conductance. Conductance and resistance changes are correlated with sweat production, which decreases with an increase in blockade density.¹³

Muscular

Reduced centrally mediated myoelectric stimulation, which is associated with increased axial blockade density, naturally reduces underlying muscular motion and baseline activity. Evoked muscle activity based on active peripheral nerve stimulation (e.g., train of four) is a common mode of monitoring neuromuscular block.¹⁴ Active stimulation has been omitted from consideration in this study in deference to passive monitoring methods.

Other

Extrapolating from the neurological and muscular outcomes, a change in the compliance, or "elasticity," of the skin and subcutaneous tissue can be expected.¹⁵ Such variance may be manifest as a measurable difference in the tissue modulus. Less notable changes include variation in secretion rates of the liver, pancreas, and other organs controlled at the mid-thoracic level; changes in endocrine production and metabolic rates; and adaptations in the condition of the bladder and sphincters controlled at the upper lumbar level.

Physiological Monitoring Methods

The next part of the systems engineering process is to determine how requirements can be met or, in this case, how the physiological measurements can be made, how the data can be analyzed, and how the information can be made available to the user. Identification of the physiological processes that accompany regional neural blockade enabled determination of those methods that

could be used in a potential technology to monitor the presence of such responses, or lack thereof, and do so noninvasively. This section considers the applicable classes of technologies and their associated processes that can be used to make the required measurements.

Table 1 provides a summary of the specific monitoring techniques that satisfy the various physiological responses, given the classes of technology. The table is the super-set of implementations from which designers can select. The information displayed is significant for a variety of reasons. First, it shows that a single physiological response can be measured by a number of different technologies. For example, muscle activity can be monitored with spontaneous EMG, spontaneous acoustic myograms (AMGs), accelerometers, strain gauges, magnetometers, or laser Doppler velocimetry. Second,

it shows that a single technology may be used successfully to monitor more than one physiological response. For example, a microphone may be used to monitor heart rate, blood flow, peristaltic sounds, and breathing sounds. The net result is that the designer is afforded a variety of monitoring methods, including those that are passive, noninvasive, and quantitative.

Passive Biopotential

Biopotential signals originate in the body and can be collected routinely and noninvasively by using skin surface electrodes interfaced to single-ended or differential amplifiers. Electrodes should maintain a low impedance interface with the skin and need to be placed strategically. In addition, the amplifiers should provide adequate gain and common mode rejection to maximize

Table 1. Summary monitoring matrix based on the application of technologies to physiological system responses.

| Monitoring methods | Technology (measurable response) | | | | |
|--------------------------------|--|------------------------------------|--|--|---|
| | Cardiovascular | Gastrointestinal | Pulmonary | Neurological | Muscular or tissue |
| Passive biopotential | EKG (heart rate, variability) | Spontaneous EGG (gastric activity) | | Ag-AgCl electrode (sweat), spontaneous EMG (muscle activity) | |
| Active biopotential | | | Volume impedance (respiration rate) | I_c or V_c circuit (skin resistance/conductance), stimulator (nerve conductance) | |
| Auscultation; passive acoustic | Microphone (heart rate, blood flow) | Microphone (peristaltic sounds) | Microphone (breathing sounds) | | Spontaneous AMG (muscle activity) |
| Displacement | Plethysmograph (blood pressure) | | Accelerometer or strain gauge (respiration rate) | | Accelerometer or strain gauge (muscle activity) |
| Active acoustic | | Ultrasound (gastric activity) | Ultrasound (respiration rate) | | Ultrasound (tissue compliance) |
| Thermometry | Thermistor or thermocouple (skin temperature) | | Thermistor or thermocouple (respiration rate) | | |
| Passive electromagnetic | Magnetometer (heart rate, variability), midband IR scanner (skin temperature) | | | Magnetometer (net nerve conduction) | Magnetometer (muscle activity) |
| Active electromagnetic | Photoplethysmograph (blood pressure), colorimeter (hemoglobin density), spectral reflection (skin temperature) | | | | Laser Doppler velocimeter (muscle activity) |

the signal-to-noise ratio. Electrocardiograms (EKGs) represent electrical currents that traverse the heart and initiate its contraction. EKGs, EMGs, and electropotential response (e.g., sweat production) are typical signals that may be monitored in a passive biopotential mode.

Contact Thermometry

Noninvasive surface temperature measurements rely on the use of a temperature-sensitive transducer (e.g., thermistor, thermocouple) with associated conditioning electronics. The sensor element is a passive measurement device that exhibits a repeatable and calibrated resistance versus a temperature relationship. Skin temperature measurements may be made in the contact thermometry mode. A thermal sensor placed in the stream of inspiratory/expiratory gases will monitor cyclic thermal activity that correlates to respiration rate.

Auscultation or Passive Acoustic Measurements

Auscultation is the process of listening to the sounds made by the body as a means of diagnosis. The detection and recording of longitudinal wave signals (e.g., sounds) requires a pressure-sensitive device, such as a microphone, with a matched amplifier. Such signatures i.e., AMGs, correlate with EMG signals. It has been shown that useful information exists in the signal below the frequency cutoff for normal human hearing.¹⁶ Thus the signal processing used for such signals should preserve the maximum possible bandwidth and amplitude fidelity. In addition to measuring muscle sounds, auscultation or passive acoustic measurements can be used to monitor peristaltic, respiratory, heart rate, and blood flow sounds.

Displacement

A device sensitive to changes in position can be used to detect muscle activity. Vibromyography has been shown to correlate mechanical muscle activity with its electrical stimulus¹⁷ through single- or multi-axis accelerometer measurements. Integration of the acceleration signal yields muscle velocity, while integration of the velocity signal yields muscle displacement. Strain gauges can also be applied to the direct measurement of muscle displacement; however, the methodology is more complex than is the case with accelerometers. When using a plethysmograph, strain gauges are effective at measuring changes in blood pressure, which may be associated with vascular dynamics.

Other Active Methods

Active biopotential measurements can be collected using electrodes and amplifiers, but require a DC or oscillatory signal source (e.g., a voltage or current driver) as well. Skin resistance or conductance and respiration

rate measurements, using thoracic impedance, are made in an active potential mode.

Active acoustic (ultrasound) measurements are made by detecting the delay time and amplitude of a transmitted signal that is reflected or refracted by structures with differing specific acoustic impedances. Ultrasound monitors may be applied to the measurement of motion artifacts in structures such as the heart, stomach, and diaphragm, or used to infer tissue properties such as compliance.

Active electromagnetics measure differential optical properties of tissues and structures. Body tissues reflect or absorb light in response to a variety of biomechanical (e.g., tissue density, position) or biochemical (e.g., oxygen saturation of hemoglobin) activity. The noted responses are measured by examining the signal intensity as a function of time, wavelength, or both. Colorimeters may function in absorption mode to infer hemoglobin density or in reflectance mode to infer temperature. In addition, laser Doppler velocimeters use modulated reflected light to measure structure motion.

Other Passive Methods

The electromagnetic spectrum contains a number of characteristics directly or indirectly related to biophysical activity. Midband infrared (IR) and induced magnetic fields are particularly important. Midband IR power levels are directly related to the temperature of a radiating part and can be measured using an optical scanner. Midband IR scanners can be applied to measuring skin surface and body core temperature, the latter being inferred from the tympanic membrane. Magnetic signatures of the order 10 to 100 pT in field strength result from the Faraday effect associated with electric biopotentials in the conducting medium of the body tissue. Sensitive magnetometers will measure magnetograms that result from the signatures of EMG and EKG activity.

MONITOR TEST BED

Requirements and Trade-Offs

It is neither practical nor cost-effective to use all of the monitoring means identified in Table 1. The systems engineer must optimize the system design by conducting a trade-off assessment in anticipation of defining a candidate system architecture. Sets of functional and operational requirements for the neural blockade monitor were developed as the basis for the weighting function used in the assessment process:

- The monitor will be noninvasive. This minimizes possible impact to the patient and the clinician (e.g., recovery time, risk of infection), and facilitates use of the device.

- The monitor will be passive. This reduces the risk that the monitors will have an adverse impact on the patient and that other surgical or clinical devices will be affected.
- The monitor will be responsive to normally occurring physiological responses without requiring the patient to be actively involved in generating responses to be measured.
- The monitor will discriminate both the anatomic level and conduction density of the block. Ideally, a continuum of level and density should be achieved; however, an indication of the existence of a block at a specific level will suffice.
- The monitor will present objective, quantitative, and unambiguous data. The data accuracy, resolution, and frequency response will be consistent with the highest degree of system performance reliability associated with life support systems.
- The monitor will be insensitive to phenomena that mimic the body's physiological response to regional anesthetics.
- The monitor will be able to determine the patient's condition regardless of when monitoring occurs relative to administration of the anesthetic.
- The monitor components must not interfere with the related medical procedure.

These requirements imposed additional clinical and logistical constraints that could be used to reduce the number of applicable technologies discussed earlier. The laser Doppler velocimeter was eliminated because of its potentially hazardous laser radiation source. Volume impedance, skin resistance/conductance, nerve stimulators, and all ultrasound devices were not considered viable because they require active stimulation for taking measurements. To be effective for the physiological responses of interest, a magnetometer must be very sensitive. The excessive electrical noise level in a typical clinical setting will likely impede such a level of sensitivity.

The final challenge was to select a set of monitors from those still considered viable that would provide specificity to both anatomical level and density of blockade, be small and unobtrusive, and be cost-effective. Cardiovascular, neurological, and muscular responses have historically produced meaningful level-related physiological responses to neural blockade and were therefore selected as primary monitor inputs. The appropriate technologies selected to monitor

these responses were a skin temperature thermocouple, spontaneous EMG, and spontaneous AMG, respectively. The ensemble of the primary sensors would be capable of monitoring blockade density at a single point on the body, with the primary sensor transducers collocated in a very small surface area. To discriminate blockade anatomical level, a number of similar monitors would need to be placed on other areas of the body. A monitor array including temperature, EMG, and AMG (Fig. 5) was designed for each site. A single-lead EKG was included as an independent means of corroborating high thoracic level blockade.

We have used a systems engineering approach to formulate the design of a neural blockade monitor. The design considers the body's expected physiological response to local anesthetic neural conduction blocking agents, the technologies that can be used to monitor the responses, and the limitations and constraints on the clinical use of the monitor.

Design

To discriminate the cephalo-caudal spatial effect of neural blockade on epidural patients, three sites were monitored: the fourth thoracic (nipple, T4) and tenth thoracic (umbilicus, T10) dermatomal levels on the anterior axillary line, and the anterior of the thigh, representing the second lumbar dermatome (L2).

Three channels of spontaneous EMG measurements were taken using surface disk electrodes placed approximately 3 cm apart and parallel to local surface muscle fibers. A data acquisition system performed signal amplification and digitization. Data were acquired in the spontaneous mode with two-pole bandpass filtering between 10 and 500 Hz. The per-channel data sampling rate was 2000 samples/s at a 12-bit resolution.

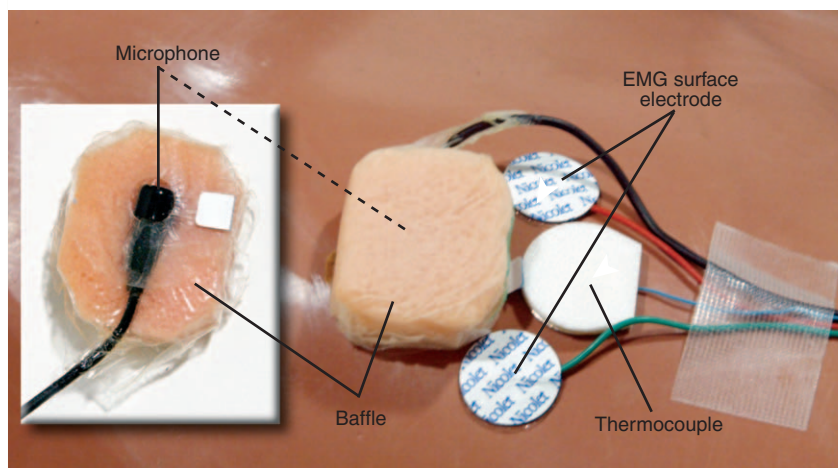


Figure 5. The monitor array used at each dermatomal level. The array monitors spontaneous EMG with surface electrodes, surface temperature with a skin thermocouple, and spontaneous AMG with a sound-baffled condenser microphone. The inset illustrates the patient-facing side of the spontaneous AMG sensor showing the microphone within the sound-baffling.

Three channels of spontaneous dermatomal AMG measurements were taken using condenser microphones applied to the skin with double-sided tape. Sound dampening backings were applied to the microphones to reduce the influence of external noises. The data acquisition system amplified and digitized the AMG signals with two-pole filtering between 0.5 and 100.0 Hz. The per-channel data sampling rate was 1000 samples/s at a 12-bit resolution.

A single EKG channel (lead II) was acquired using diaphoretic surface electrodes. The data acquisition system amplified and digitized the EKG signal with two-pole filtering between 2 and 100 Hz. The data sampling rate was 1000 samples/s at a 12-bit resolution.

A single tympanic probe and three skin thermocouple probes provided core and skin surface temperature measurements, respectively. The thermocouples were interfaced to signal conditioners and an independent digitizer. The per-channel data sampling rate of 0.2 samples/s was sufficient for the DC-0.02-Hz frequency response of the probes. The relative accuracy and repeatability of each thermocouple was $<0.1^{\circ}\text{C}$; absolute accuracy was approximately $\pm 1^{\circ}\text{C}$.

Two separate skin-type thermocouple probes measured ambient temperature. The first ambient thermocouple was clipped to the draw sheet next to the patient and under the sterile drapes. The second thermocouple was suspended in the free air of the operating room, in a low-traffic area approximately 2 m away from the patient. A single condenser microphone, placed under the operating table, monitored ambient room noises.

Published literature^{8,9,11,14,16,17} contains ample information about the signal characteristics of the signals of interest. It was this information that was used to guide the selection of the data acquisition system setup. Additional consideration was given to the system's available discrete filter and sampling rate set-points.

The EMG/AMG/EKG data acquisition system could not record continuous data for long durations, so the system was designed to collect 20-s data epochs at pre-defined intervals. The first epoch was immediately prior to the administration of any anesthetic (baseline) and at 2, 5, 10, 15, 20, 25, 30, 45, 60, 75, and 90 min after dosing. Temperature data records were time-continuous, starting immediately after the thermocouples were placed and ending approximately 2 min after the last EMG/AMG/EKG sampling epoch.

CLINICAL STUDY

Patient Preparation and Investigational Procedure

A clinical study was designed to evaluate and validate the performance of the neural blockade monitor. To minimize variability we evaluated a common

surgical procedure (radical retropubic prostatectomy), conducted by the same surgeon, using the same lumbar epidural blockade anesthetic. The Johns Hopkins University Institutional Review Board approved the study protocol, and all seven male subjects (47–65 years old, 78–110 kg) gave informed consent prior to their participation.

Mid-lumbar (L2–L5) epidural catheter placement was performed during very light sedation with the patient secured in the sitting or lateral position. After positioning the patient in the supine position, a 3-mL anesthetic test dose was administered to identify inadvertent intravascular or intrathecal placement of the catheter. In the absence of an inappropriate response to the test dose, temperature data acquisition was initiated. The main epidural anesthetic dose was then administered. The anesthesiologist used traditional monitoring procedures (pinch test) to assess and record the anatomic level of the block. Post-anesthetic datasets were collected on a not-to-interfere basis. All study monitor displays were positioned out of sight of the surgeon and the anesthesiologist, and neither was advised of real-time monitoring results. No requests were made for temporary suspension of surgical activity during data collection intervals, and the results of monitoring were not used in any way to influence patient care.

Data Processing and Analysis

All data files were converted from their native formats into a format directly compatible with the MATLAB computation and visualization tool. The data conversion process was validated by visualizing signal amplitude time histories. All signal amplitudes were transformed from digital bit representations to appropriate scientific units (e.g., $^{\circ}\text{C}$, mV).

EMG and AMG

Identification of a single derived value that would best describe the patient's instantaneous physical condition was a key objective of the processing undertaken in this study. Both EMG and AMG measurements are broadband signals that contain information in the time and frequency domains. Typically, the power of such signals provides a single value from the time domain data. Power is proportional to the square of the amplitude of the signal and, for a signal that has a nonconstant component (e.g., alternating current), it is necessary to average over a finite period of the signal to generate a useful value. The root-mean-square (rms) estimator was selected to process the signals and generate an averaged power indication:

$$\text{rms} = \sqrt{\sum_{i=1}^n x_i^2/n}$$

where n is the number of samples in the integration interval. An integration interval was selected that retained as much of the passband information as practical. EMG signals were integrated over 50 ms ($n = 100$) and AMG signals were integrated over 250 ms ($n = 250$). Incrementally deleting the earliest data point and adding the next point of the series implemented a running rms computation.

The absolute value of EMG signals can be influenced by conditions such as skin conductance, skin temperature, electrode displacement, and site preparation. To mitigate discrepancies, the EMG data were normalized such that the histogram of the baseline epoch contained 50% of all data values between ± 5 mV. Other epochs were scaled by the same gain factor calculated for the baseline epoch. AMG data were compensated, on a sample-by-sample basis, for the presence of acoustically transmitted noise by subtraction of the ambient signal level.

Temperature

We expected changes in both body core temperature and ambient air temperature to influence changes in skin temperature. To accommodate these conditions, each raw dermatomal ($t_{DL,SI}$) temperature value was standardized ($T_{DL,SI}$) by the tympanic and ambient temperatures as in the following:

$$T_{DL,SI} = t_{DL,SI} - [t_{Tympanic,SI} - t_{Tympanic,0}] - [t_{Ambient,SI} - t_{Ambient,0}] ,$$

where DL is the dermatomal level and SI is the sampling interval. $SI = 0$ correlates to the baseline data epoch. Standardized temperature data were then parsed into 20-s segments that correlated with the EMG/AMG/EKG sampling epochs. The average of each of the normalized data epochs was computed as a simple arithmetic mean of the data.

EKG

Each subject's average heart rate was determined from the lead-II EKG datasets. The average of each of the data epochs was computed as a simple arithmetic mean of the interval between successive heart beats.

Findings

Temporal changes in EMG, skin surface temperature, and heart rate were associated with adequate blockade and correlated to the clinical assessment of the level and density of the block. AMG showed primary correlation to external influences, and response to the block was not evident in the presence of noise.

The data revealed three notable aspects of the EMG signals. First, EMG signal amplitude decreased at a rate

inversely related to the number of dermatomal levels separating the monitor site and the epidural catheter. That is, the signals at the more cephalad dermatomal levels decreased later than the more caudal levels. Second, after about 15 to 30 min, all EMG signal amplitudes assumed a generally constant and lower level than at initiation of the primary dose. Third, the constant level at the L2 dermatomal level was approximately 30% lower than that of the T4 and T10 dermatomal levels, likely indicating that effective block was created at this discrete level simply by the test dose.

Figure 6 shows a set of curves of the rms EMG signal amplitudes for all seven subjects. The data are fifth-order polynomial curves fit to the normalized signal amplitudes for the seven study subjects at each of the three monitor levels. The plot also indicates the typical anatomic level of blockade achieved according to pinch tests administered by the attending anesthesiologist. Time is referenced to the administration of the main epidural dose ($t = 0$), and the baseline dataset is plotted at $t = -1$ min. The traces are normalized to the baseline by subtracting the respective baseline value from each successive data point. The time between the main dose and the surgical incision was 17 ± 6 min. Lower amplitude limits of the L2 and T10 signals are reached sooner ($t = 15$ min) than the T4 signal ($t = 30$ min), suggesting that the L2 and T10 levels achieve onset before the T4 level, as expected.

Noise artifacts (surgical manipulation) dominated AMG data from levels near the operative site (L2 and T10). The data did not display any correlation to the physiological condition of the patient. The AMG data from the T4 level exhibited little change in signal. The AMG data did not reflect the same consistency observed in the EMG data. In several instances, the data were

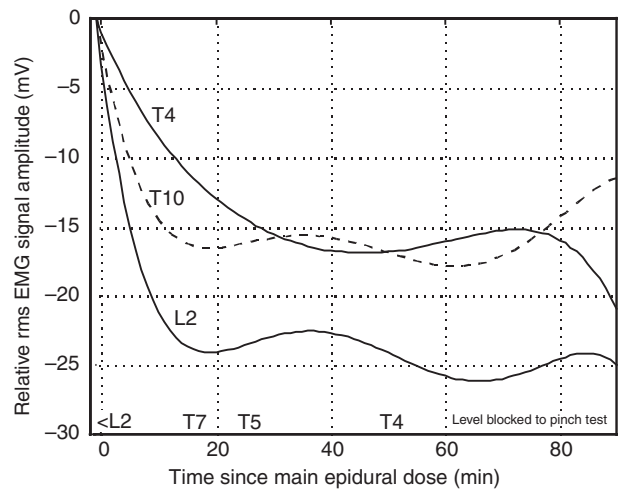


Figure 6. EMG signal amplitude time series measured for all seven subjects. The curves are fifth-order polynomial fits to the normalized rms data values. Also shown is an indication of the dermatomal level of blockade determined by pinch tests.

inexplicably saturated; in others, they were corrupted by external influences.

Average temperature trends for the tympanic- and ambient-compensated data all showed a gradual increase as a function of time; longer onset was noted at dermatomal levels farther from the epidural injection site. Figure 7 presents average temperature data for the anatomic levels for all seven subjects. All levels exhibit the same upward trend as a function of time; they approach a relative increase of approximately 3.5°C over the 90-min data collection period. The changes in temperature reflect the blockade of the sympathetic nervous system after administration of the local anesthetic. These changes also suggest differences in the contribution of blood volume changes and vasodilation at each dermatomal level.

Heart rate increased by an average of 7.8 ± 9.2 beats per minute (bpm) to 10 min following initiation of the primary dose, then decreased an average of 19.0 ± 10.8 bpm in the following 15 min. The average heart rate for all seven subjects is shown in Fig. 8. These data conform to the documented effect of local anesthetic on the cardiac accelerator fibers in the T1–T4 spinal segments.

It was determined that a universal neural blockade monitor had to be fully functional regardless of when it is applied, relative to the administration of the anesthetic agent, and that the rate of change, or gradient, of the absolute signals presents salient information. The most compelling indicator is the change in EMG signal level between zero and 10 min, as previously shown in Fig. 6. Level L2 amplitude decreases approximately 1.5 times that of T10 and 2.5 times that of T4. Using the monitored parameters in combination rather than singly will enhance the utility of a universal monitor.

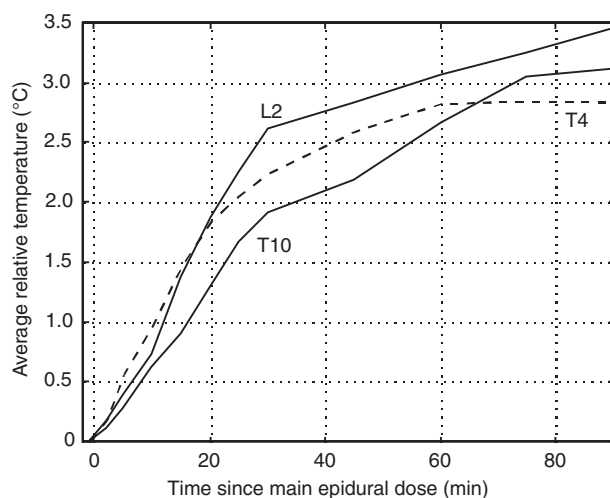


Figure 7. Average temperature time series values for all seven clinical subjects. The data are compensated for changes in each subject's core temperature and variability in the ambient temperature of the operating room.

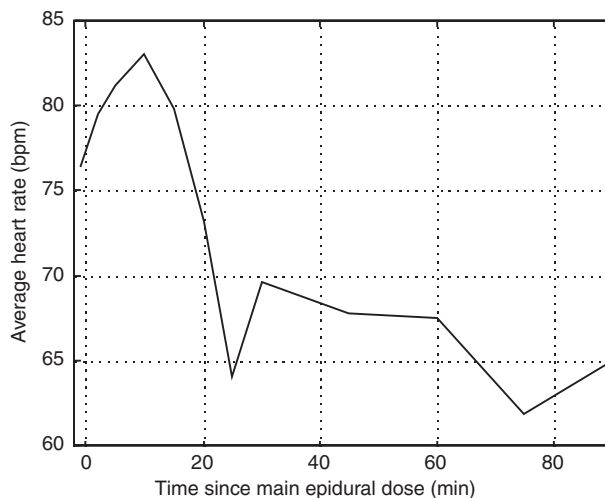


Figure 8. Average heart rate (beats per minute) time series values for all seven clinical subjects. The data are derived from a lead-II EKG measurement.

CONCLUSION

This investigation examined the use of passive and noninvasive monitors to objectively distinguish the level and density of neural blockade as a function of time and as related to the administration of local anesthetic. The application of a rigorous systems engineering process and methodology proved to be valuable in producing a rapid prototype that comprehensively and conclusively demonstrated the feasibility of an objective neural blockade monitor. Moreover, the use of a systems engineering process reduced the amount of time needed to develop and demonstrate the prototype.

The data from the EMG, temperature, and EKG monitors, combined with the clinical assessment provided by the anesthesiologist and the conditions of the operative procedure, confirm that an objective measure of block level and density can be performed in the clinical setting. The work presented here prepares a strong foundation and demonstrates proof of principle for this technology. Results thus far suggest that the monitor need only acquire EMG, temperature, and EKG parameters to discriminate the level and density of an axial block.

In summary, it has been shown that the onset of regional neural blockade can be objectively monitored by quantifying a decrease in signal amplitude of a surface EMG, an increase in skin temperature, and changes in heart rate. Moreover, the blockade density determined by these objective means correlates well with the traditional subjective method of pinch tests. It is reasonable to expect that an advanced version of the prototype monitor can give the anesthesiologist/clinician an objective tool for real-time monitoring of the level and density of regional neural blockade.

The results of this research have led to the issuance of U.S. and foreign patents for the monitoring

concept and method. The intellectual property rights have been optioned for further development and validation by direct licensing to industry and through technology transfer grant proposals to the National Institutes of Health.

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