Motor Evoked Potentials

7

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Introduction

Iatrogenic injuries are an undesired consequence of surgery, yet iatrogenic injuries to the motor system are much more devastating to a patient's quality of life than most injuries to the sensory system. In many cases, intraoperative injuries to the spinal cord will be detected by sensory evoked potentials (SSEPs), yet a focal injury to the anterior spinal artery (ASA) may be missed [1]. There is a lot of evidence in the literature describing selective injury to the anterolateral columns sparing dorsal columns with preserved SSEPs [2–5]. The inclusions of motor evoked potentials (MEPs) to the intraoperative monitoring *toolbox* can help to confirm/prevent selective lesions to the anterolateral columns of the spinal cord. Additionally, MEPs, compared to SSEPs, can more quickly detect an ischemic injury to the spinal cord [6]. Yet, MEPs are not without their limitations. Even given these limitations, proper application and interpretation of MEP data can be a significant adjunct in reducing iatrogenic injury during surgery.

History

Artificial stimulation of the motor system dates to 1664 when Swammerdam removed the heart of a frog and demonstrated that by gently stroking the severed nerve ends of the open wound the muscles would contract [7]. The most well-known experiment comes from Luigi Galvani when in 1771 he observed that electrical sparks applied to the nerves in the leg of a frog would cause twitches in the leg muscles [8]. In the 1860s, Hitzig and Fritsch stimulated the exposed brains of soldiers using direct cortical stimulation (DCS) and found that they could cause *crude* movements [9]. They continued their work on live dogs and found that not only could they cause these *crude* movements, but they also observed that specific areas, when stimulated, caused specific movements [10]. In the late 1930s, the neurosurgeon Wilder Penfield published his mapping studies of the human brain performed during epilepsy and tumor resection surgeries [11]. Penfield not only localized the motor and sensory areas of the brain but also defined the cortical somatotopy or motor and sensory homunculi of these two cortical areas. Penfield's basic stimulation technique, 60 Hz

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trains of stimuli lasting for one to a few seconds, is still practiced for cortical mapping of language and sensory areas. In the 1950s, Patton and Amassian were the first to record direct traveling waves from corticospinal tracts (CST) when stimulating the motor cortex/subcortex in both cats and primates [12]. They observed two types of waves: the first was a short-latency triphasic response termed the D-wave (direct wave), interpreted as a result of the direct activation of the CSTt, and the second set of waves were termed I-waves (indirect waves), interpreted as trans-synaptic activation of motor neurons of the CST within the motor cortex [13].

Research on the motor system continued, yet there existed no direct method to deliver stimuli to a subject's brain without accessing the brain directly given the extremely high impedance of the skull. In order to electrically cross this high impedance barrier, high stimulus currents are needed to activate the underlying neural tissue. In 1980, Merton and Morton developed a high-voltage single-pulse technique for the delivery of transcranial electrical stimulation (TES) to the intact human subject [14] (it should be noted that they discuss that this stimulation was "without undue discomfort" to the subjects).

One interesting study using this method was published after the work of Merton and Morton, by Levy et al., that delivered TES via an anodal electrode placed over the motor cortex and a cathodal electrode placed on the hard palate to record D-waves, via either electrodes placed over the thoracic spinal canal or by inserting electrodes into the level of the bony laminae or directly in the epidural space during surgery [15]. They claimed that these recordings represented descending activity of the motor system. In addition to demonstrating the recording, Levy et al. discuss using multiple pulses to help produce motor activation at lower stimulation levels, yet this idea was not pursued [15] until much later as will be seen below. In the late 1980s, Katayama and Tsubokawa recorded D-waves from the epidural space of the spinal cord stimulating surgically exposed motor cortex [16]. Epidural spinal electrodes were inserted percutaneously into the upper thoracic epidural space under X-ray control and pushed cranially to the lower cervical epidural space. During surgery the motor cortex and other cortical areas were then directly stimulated using both monopolar and bipolar stimulation. They demonstrated that direct application of monopolar anodal current to the motor cortex required lower stimulation intensities as compared to cathodal bipolar stimulation [16]. In order to better refine the most optimal stimulation configuration and also to understand the phenomena of latency changes with increasing stimulation current, Burke et al. proposed the discrete jumps in latency to be due to bends in the CST as the stimulation moved deeper in the brain [17]. A set of papers by Deletis, Rodi, and Amassian described the neurophysiologic mechanism underlying MEPs in anesthetized humans which is of importance in understanding the pitfalls during the routine use of MEP monitoring in the operating room [18, 19].

Physiological Background for Monitoring the Motor System

Depending on the type, location, and intensity of stimulation, the MEPs recorded during intraoperative neuromonitoring (IONM) are generated and transmitted from a limited subset of neural elements. These responses are, for the most part, transmitted by the largest fibers of the CST, and in deeply anesthetized patients, this electrical stimulus activates these largest fibers directly. It is important to note that it is the axons that are being activated and not the cell bodies. Additionally, as the stimulus intensity increases, the depth of stimulation also increases. The exception is in the awake subjects/patients or use of transcranial magnetic stimulation (TMS) where the pyramidal cell body is activated by interneurons ending up on the pyramidal cells in the cortical gray matter. Yet even given that we are testing a limited subset of the motor system, the data obtained with this method can still be useful for patient protection, and the physiology behind these responses needs to be properly understood in order to make proper data interpretations in the operating room.

Anatomy and Physiology of the Motor System

The motor system is a complex combination of neural subsystems existing in both the central and peripheral parts of the nervous system. It is important to realize that artificial stimulation most likely activates many different cortical fibers, while MEP monitoring techniques only record responses from a small portion of them (Fig. 7.1). The primary anatomic structures activated by transcranial MEPs (TcMEPs) are the axons of the Betz cell in layer 5 of the motor cortex. These axons are part of the CST and corticobulbar tract (CBT) which are upper motor neurons. These axons decussate at the level of the medulla and travel down the spinal cord to the α-motor neuron (αMN) whose cell body is located in the ventral gray of the spinal cord. The (αMN) is the lower motor neuron. Upon synapsing on the αMN, its axon travels out through the ventral root of the spinal cord to the peripheral nerves

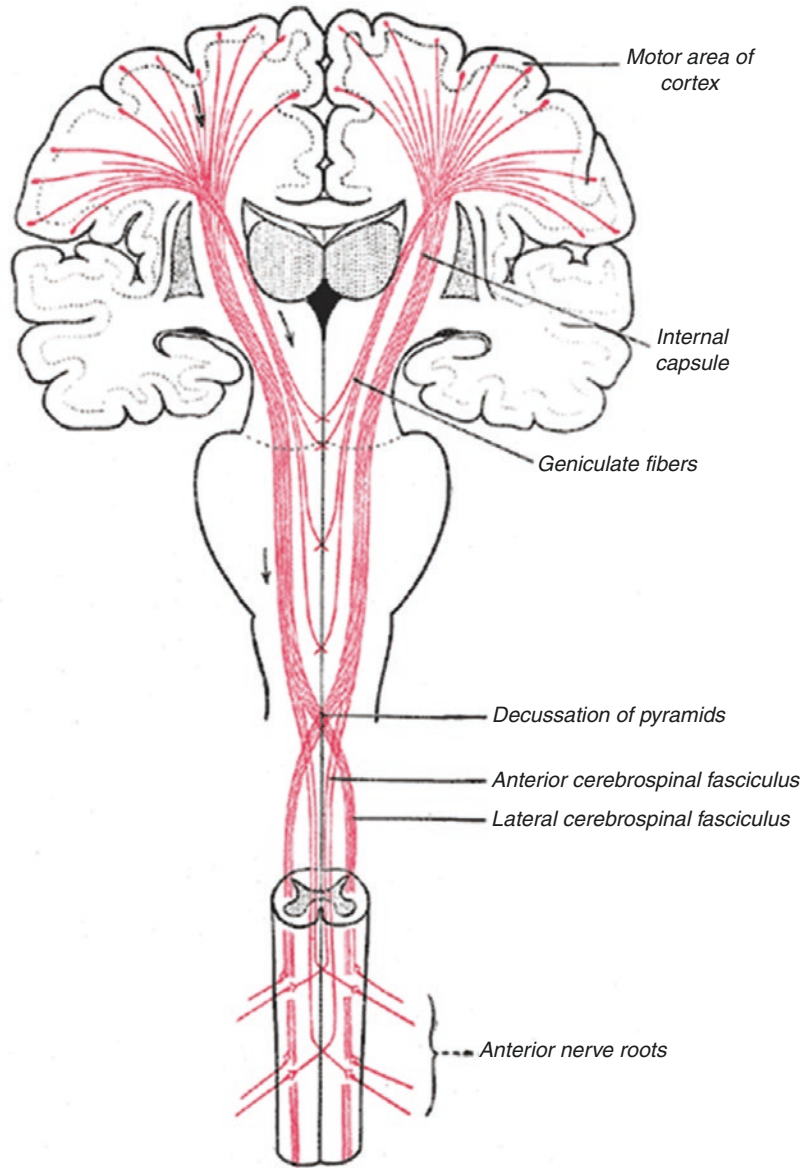

Fig. 7.1 Multiple areas of the cortex are involved in motor movements. In addition to the cortex, there also exist multiple subcortical areas. During artificial simulation under anesthetics, the corticospinal fibers are the main carriers of that stimulation information to the alpha motor neurons in the spine. Even though the stimulation will activate fibers from other areas, this information is usually not passed due to the synaptic junctions between the other areas and the corticospinal tract which under anesthesia are shut down for the most part. (From http://thebrain.mcgill.ca/flash/a/a_06/a_06_cr/a_06_cr_mou/a_06_cr_mou.html (copy left) and with permission from the GNU free documentation license)

and then finally to the muscle. The primary motor cortex, where the CST fibers originate, is located in the precentral gyrus and is primarily responsible for fine voluntary movement. This area of the cortex receives information from multiple cortical areas which include the extrapyramidal systems (areas such as the basal ganglia and cerebellum) and sensory areas including somatosensory, visual, auditory, both parietal, and frontal cortices. The primary motor cortex has a map of the body, or homunculus, with the head located laterally on the cortex and the leg located centrally. This topography illustrates the cortical surface area dedicated to innervation of parts of the body. See Chap. 2 for more information on the homunculus. The corticospinal and corticobulbar pathways are shown in Fig. 7.2.

At the surface of the cortex are six layers of gray matter. Each functional area of the brain has different proportions of each of these six layers, yet the basic six-layer structure is the same throughout the cortex. Each area of the cortex is defined based on its specific cytoarchitecture and neural organization. The nomenclature used for this differentiation is known as a Brodmann area [20]. Interestingly each Brodmann area generally corresponds to a specific functional area, even though the original differentiation was purely based on its cytoarchitecture (http://www.fmriconsulting.com/brodmann/Introduction.html). Generally layer 5 is the output while layer 4 is the input layer. The primary motor area (PMA), or Brodmann area 4, is located in the posterior portion of the frontal lobe just anterior to the central sulcus (Fig. 7.3). Layer 5 of the primary motor cortex contains large pyramidal cells known as Betz cells that send long axons directly to motor neurons located in the spinal cord or brainstem via the CST or CBT (the combination of these two tracts is known as the pyramidal tract). About 60% of the human CST arises from the primary motor cortex and area 6 (premotor area and supplementary motor area); the other 40% arises from the somatosensory cortex (areas 1, 2, and 3) and cingulate cortex (areas 23 and 24) [21, 22]. Even though all areas of the body are represented within the primary motor cortex, it appears that more proximal and axial muscle fibers in the CST have their origins in the premotor area (area 6), while the distal musculature tends to have its origin in the premotor areas (area 4) [23]. Since both sets of fibers are contained in the CST, stimulation used during IOM will activate both of them. From the cortex the CST funnels into the anterior half of the posterior limb of the internal capsule and then travels between the thalamus and parts of the basal ganglia (striatum and globus pallidus) to the ventral portion of the cerebral peduncles (in the middle two-fifths of the cerebral crus—anterior portion of the cerebral peduncles). At this level, the fibers that will eventually synapse on αMNs in the spinal cord gray matter innervate leg muscles and are lateral to fibers eventually innervating hand muscles. From the midbrain, the CST fibers enter the pons and pass through the pontine nuclei where fibers going to the leg muscles are now located ventrolateral relative to the fibers going to the hand muscles. The CST enters the ventral part of the medulla forming part of the medullary pyramids where fibers innervating the lower limbs are located ventrolateral compared to the fibers innervating the upper limbs. At the lower level of the medulla, 80–90% of the CT

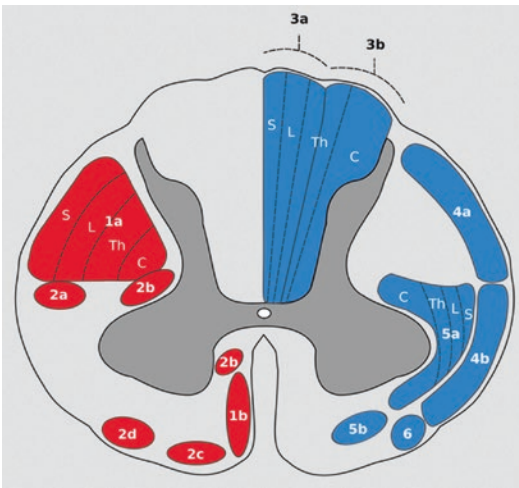

Fig. 7.2 The lateral corticospinal tract (1a in the figure) shows a lateral to medial homunculus with the sacral region being most lateral and the cervical region being the most medial. Region 1b is the anterior corticospinal tract in the spinal cord. (With permission from Wikimedia Commons—public domain)

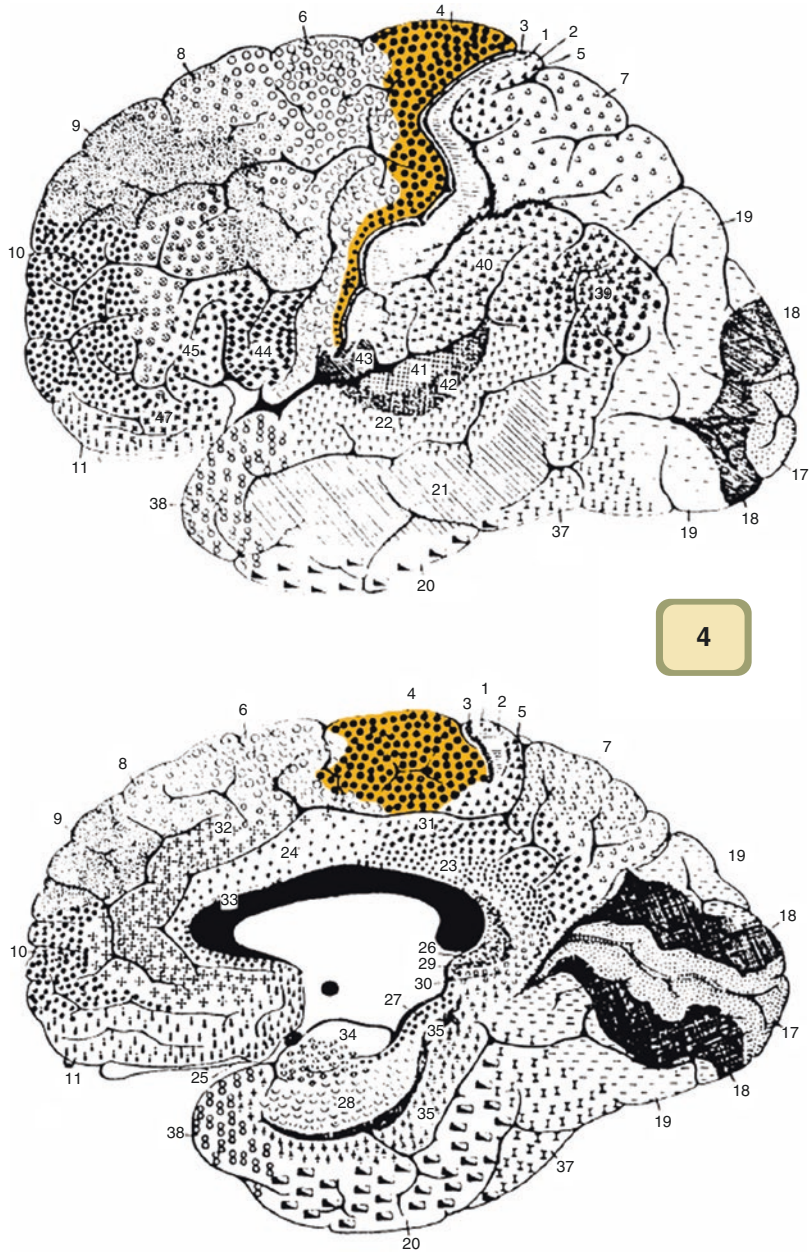

Fig. 7.3 Map of the cortex with all of Brodmann areas depicted. Area 4, the primary motor cortex, is *highlighted*. Area 4 is just anterior to the central sulcus. (With permission from Wikimedia Commons—public domain)

decussates with most fibers entering lateral CST of the spinal cord. Fibers going to the lower limb muscles tend to cross more rostrally than for the upper limbs. The 10–20% of uncrossed fibers in the anterior CST innervate αMN ending on more proximal and trunk musculature [24].

There are about one million fibers in each CST with around 2% of these fibers being large (11–20 μm) which are known as fast-conducting corticospinal fibers (conduction around 50 m/s). CST fibers for the upper limb are more medial than lower limb fibers. The rest of the CST fibers synapse on other interneurons within the gray matter of the spinal cord. The large CST fibers are essential for eliciting MEPs. About 55% of all CST fibers end in the cervical region with 25% innervating the lower limbs. The rest of the fibers innervate the thoracic region. It

is interesting to note that the CST is not symmetric, and it appears that CST fibers that cross more anterior tend to form the larger proportion of CST fibers in the cord whether it is the right or the left [25]. A single αMN has over 1000 synapses with over 50 direct inputs [26]; thus in the awake animal, generation of an action potential in the αMN is a complex process of competing systems. In the anesthetized animal, this complex system is shut down due to anesthetics. In addition to αMN CST inputs, there are inputs from interneurons driven by other CST fibers, inhibitory interneurons, Renshaw cells (which are inhibitory), sensory Ia and Ib fibers, and other descending tracts including the rubrospinal tract, vestibulospinal tract, reticulospinal tract, and tectospinal tract. Many of these presynaptic fibers synapse at multiple locations on the αMN, instead of one point. Due to the large number of synapses, it appears that the control of the αMN is multifactorial. In the non-neurologically compromised awake human, all the synaptic inputs to a specific αMN modulate the membrane potential; thus appropriate supratentorial modulation appropriately depolarizes the cell.

The CST enters the gray matter of the spinal cord in the ventral horn and fans out terminating in laminae IV through IX [27]. Yet the largest CST fibers appear to make monosynaptic connections to the αMN in laminae IX [28]. Most of the CST tends to synapse on interneurons, some of which being part of circuits that modulate the αMN, while others influence motor circuits such as the γ-motor system. Axons from the αMN innervate muscle fibers of a single muscle. The αMN and its axon are known as the lower motor neurons. The combination of the αMN, the terminal branches of the αMN, and the muscle fibers they innervate is known as the motor unit. Each motor unit is innervated by one axon and thus only one αMN. See Chap. 8 for more information on the motor unit.

Damage to either the upper motor neurons or the lower motor neurons will cause paralysis. Damage to the lower motor neuron will result in what is known as a flaccid paralysis—no muscle tone and no movement. Damage to the upper motor neuron demonstrates a more complex set of symptoms but generally includes no voluntary movement and a range of muscle tone from minimal tone to severe spasticity.

Indirect damage to the motor system can arise from reducing the blood supply to the critical structures. The cortex is supplied primarily by four main vessels, the two carotids and the two vertebral arteries. These four vessels supply the circle of Willis (COW) presenting connection between the carotid and vertebral arteries. The middle cerebral artery (MCA) coming off of the carotid artery supplies the lateral frontal and central cortex and its descending axons as well as much of the temporal lobes. The anterior cerebral artery (ACA) supplies the medial parasagittal frontal and central cortex and its descending axons originating from the motor cortex. Axons of the CST within the internal capsule are supplied by lenticulostriate branches originating from the MCA and the anterior choroidal arteries. At the level of the brainstem, the CST is supplied by branches of the vertebral and basilar arteries. The spinal cord is supplied by one anterior spinal artery (ASA), two posterior spinal arteries (PSA), and a varying number of radicular arteries. The ASA supplies the anterior 2/3 of the spinal cord including the lateral and anterior CST and the ventral horn. In the adult, the ASA is formed via fusion of the anterior spinal branches of the vertebral arteries, while the PSA originates from the posterior inferior cerebellar arteries (PICA) [29]. In the thoracic spinal cord, there is usually one large supply vessel coming from the aorta known as the artery of Adamkiewicz and two or three smaller vessels. Interestingly, in about 10% of patients, this vessel enters the spine at the L1–L2 level [30]. This variability in supply demonstrates one of the critical needs for neuromonitoring. Normally watershed zones are most commonly seen at levels T1, T5, and T8–T9 where reductions of blood flow in any of the feeder vessels can cause significant ischemia at these regions [29].

Electrophysiology

Electrical stimulation is used to generate APs in multiple points along the motor pathways. There are, in general, two types of recorded response in the anesthetized patient: (1) conducted volleys traveling along the spinal cord and peripheral nerves and (2) compound muscle action potentials (CMAPs). The latter one records the muscle response activated from excitation of the αMN. In general stimulation is applied at the level of the motor cortex or subcortical part of the CST, the spinal cord, or peripheral nerve. Each of these areas requires different stimulation parameters that will be described in the next section.

The descending volleys along the CST, initiated via stimulation, originate from separate but not independent circuits and are differentiable by their responses. The first response, defined as the D-wave (Fig. 7.4), or direct response, results from the direct stimulation of the CST fibers in the cortex. This response can come from either stimulating the axons directly or also stimulating the gray matter in turn generating the axonal response. During IONM procedures, it is the axon that is being activated. The second set of responses are defined as I-waves, or indirect waves (see Fig. 7.4), resulting from local circuits in the cortex being activated by the stimulus. It has been shown that the amplitude of the D-wave is proportional to the intensity of the stimulation of the subcortical white matter up to a certain point, which most likely represents the activation of the entire CST [32]. It was also observed that the latency does not increase linearly as stimulation intensity is increased. Late in the 1980s, Rattay demonstrated that the point of action potential initiation on an axon most likely occurs at bends or curves [33]. These sudden *jumps* in latency correspond to the location of CST bends which occur at the fan out of the fibers in the cortex, at the level of the genu of the internal capsule and at the level of the brainstem. This fact is important since if surgery is targeted at a specific area in the brain, you need to make sure that the stimulation does not directly activate CST fibers more caudal in the brain or brainstem from the point of surgical intervention.

Given that D-waves result from a direct activation of the cell body or axon, and the fact that the response is recorded from the axon, these responses are unaffected by anesthetics. I-waves, on the other hand, will usually not be present during certain forms of anesthetics given that they are generated via circuit pathways and contain multiple synapses. For many surgical procedures, such as during spinal instrumentation procedures, recording descending volleys along

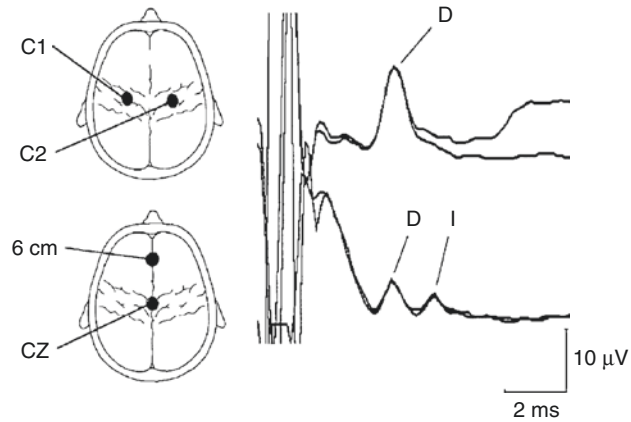

Fig. 7.4 Upper thoracic epidural recordings of D- and I-waves in a 14-year-old female during surgery for a low cervical intramedullary tumor. The upper trace was obtained after transcranial electrical stimulation over C1 (anode) and C2 (cathode) using 140 mA stimulus intensity and a stimulus duration of 500 µs. The lower trace was obtained after anodic stimulation at Cz and cathodal stimulation at 6 cm anterior to Cz, using the same stimulus duration but at 200 mA. Note the appearance of the D- and I-waves with this electrode arrangement (An upward deflection is negative). (Reprinted with permission from Deletis [31])

the CST is considered an invasive procedure and is not used. During most surgical procedures, muscle responses (known as CMAPs), resulting from driving the αMN via the largest fibers in the CT, are the monitored waves. Given that the αMN is a highly modulated cell, the effects of anesthesia are important in understanding the behavior of the CMAP response. As anesthesia starts to shut down synaptic transmission, it becomes increasingly difficult for a single pulse on one CT fiber to be able to generate a CMAP (although in some cases high-intensity long pulse stimulus durations will generate a CMAP). In order to compensate for the effect of anesthesia, it was found that a multi-pulse technique was necessary. It is important to remember that the multi-pulse technique is needed for generation of a CMAP, but the D-wave can be generated with a single pulse. It is interesting to note that when under deeper anesthetic states, I-waves are lost, thus giving credence to the synaptic nature of I-waves. When looking at the different anesthetic agents, it has been demonstrated that inhalational anesthetics are the most effective in abolishing the muscle MEP response [34–37]. It should also be noted that the blocking effects of inhalational anesthetics are not linear at the αMN; thus the interstimulus interval (ISI) between train of stimuli will need to change as concentrations change [35, 36]. Sloan et al. demonstrated that for low to moderate doses of isoflurane and N_2O, an ISI between 3 and 6 ms was optimal for producing CMAPs, yet for high concentrations only 1 ms ISI produced a recordable CMAP. Given that they studied using N_2O alone and found no major difference between the concentration and ISI, they concluded that isoflurane was the primary cause of the reduced CMAP amplitudes. At our institution we find that shorter ISIs help elicit MEPS when higher doses of inhalational agents are used and make less of a difference with a pure TIVA regime. A common technique to minimize the anesthetic effect at the αMN is to use an infusion of opioids with propofol. Even though this technique can still affect the CMAP response, its effect is much smaller than that of inhalational agents. Scheufler et al. investigated varying doses of propofol (combined with a constant remifentanil infusion) with different ISIs and stimulus intensities and found that an ISI of 1 ms produced the largest MEP response for a given dose of propofol [38]. It is important to note that in some cases, for patient safety, a specific anesthetic may be needed that is not optimally compatible when eliciting MEPs, and it is critical that the IOM technologist and neurophysiologist have a good line of communication with the anesthesiologist and surgeon.

There are additional pulse parameters that can help overcome the anesthetic effect at the αMN. The multi-pulse technique consisting of a train of 5–9 pulses and an ISI of 1–4 ms is the primary method. Some groups [39] describe using a 500 μs pulse width, while some IONM equipment does not allow for stimulation pulse widths above 75 μs.[1] Recently, Abalkail et al. investigated pulses with optimization via strength duration curve analysis and found that a pulse width of 200 μs is optimal when using a 4 ms ISI [40]. In 1993, Taniguchi et al. studied multiple stimulation parameters during craniotomies [41]. Using both cathodal and anodal monopolar stimulation, Taniguchi et al. looked at stimulation pulse width, train length, and ISI. The study found that an ISI of 2 ms was optimal (i.e., minimal stimulation intensity to obtain a maximum MEP response) yet varied with age, anesthetic regime, and functional integrity. It is important to note that they did not do a systematic strength duration analysis though. Using these results some groups have demonstrated that when a non-optimal MEP is obtained, one should try varying the ISI (Journee et al. personal communication). Varying the ISI is important since for the αMN to reach firing threshold, the temporal relationship between the D-wave volleys is critical. Deletis et al. found that short trains with an ISI of 4 ms were optimal in eliciting responses in the tibialis anterior muscle (TA) based on complete recovery of the D-wave amplitude [18]. It is important to note that Taniguchi et al. used a 200 μs pulse width, while Deletis et al. used a 500 μs pulse width which may affect the optimal ISI. Deletis

[1] At the time of this writing.

et al. demonstrate that if the ISI is a harmonic of the regular I-wave intervals, it will require less stimuli to be able to generate I-waves and in turn a CMAP, even though this may not be easy to determine in the OR other than by trying differing pulse widths if the response is difficult to obtain [18]. Szelenyi et al. found that an ISI of 4 ms always produced MEPs at the lowest stimulation threshold, yet the difference between the different ISIs was not statistically significant [42].

Another technique used to improve the efficacy of the CMAP is to use a conditioning pulse train [43]. The motor response recorded in the operating room is a combination of responses from multiple motor neuron pools. Each pool is directly activated by a single corticospinal axon. If all motor neuron pools are activated simultaneously, it would be easy to just modify the number of pulses in a train in order to elicit the maximum MEP amplitude. In most cases the motor neuron pools do not activate simultaneously when either giving a single pulse or single train of pulses due to dispersion (i.e., uneven conduction along the different fibers) between the fibers, even at supramaximal stimulation while under anesthesia. This dispersion has the effect of increasing the time difference between pulses arriving at the motor neuron pool. When there is a lesion in the fiber pathway, this dispersion effect increases. Thus, in many cases during surgery, the optimal MEP amplitude is not met due to abnormalities in the spinal cord fibers' conductivity or impaired spinal cord function. The purpose of the conditioning pulse train is to raise the αMN membrane excitability. This pre-pulse train (conditioning pulse train) facilitates the generation of the CMAP via the actual test pulse train by making it easier for the test pulse train to depolarize αMN. This technique is based on the the following two properties: (1) increaseing the membrane excitability (depolarizing the alpha motor neuron membrane) via direct activation of the corticospinal tract; and (2) via secondary neurons activated by temporal summation. In order to optimize the facilitation, the test stimuli need to be applied just when the αMN membrane is maximally depolarized from the conditioning train. Journée et al. developed such a methodology whereby a pre-train is applied prior to the test train to raise the excitability of the αMN [43].

It is known that the motor threshold of a muscle during a voluntary contraction is lower than when that muscle is at rest and that this difference is modulated by both cortical and spinal mechanisms [44]. These voluntary mechanisms used to reduce motor threshold cannot be used when the patient is anesthetized. By using homonymous conditioning (stimulating the same pool at the same site for both the conditioning and test pulse train), there is the potential for a large overlap between the motor pool stimulated with the conditioning pulse and the test pulse. Journée et al. describe two windows for facilitation: (1) with an intertrain interval (ITI) between 10 and 40 ms and (2) with an ITI >100 ms. It is recommend trying the shorter ITI first and then the longer ITI [43].

Transcranial Motor Evoked Potentials

TES and TMS are both used to activate the motor system and elicit MEPs. The two techniques differ in their location of action on the neuron. With electrical stimulation the electrical current flows from the anode to the cathode, and the predominant direction of flow is in the radial direction, while for magnetic stimulation, the magnetic field passes perpendicular to the plane of the coil which is placed tangential to the scalp (see Fig. 7.2). The electric field produced by TMS is perpendicular to the magnetic field and thus tangential to the cortex. Thus for each type of stimulation, the electric field is oriented 90° from each other. When the electric field is parallel to the neural element, activation is a function of distance and also changes in orientation (i.e., not exactly parallel) of that element with bends being the most likely sites of activation [45, 46]. The TES response is at a slightly shorter latency than the TMS response [47]. The latency difference is a function of the trans-synaptic nature of TMS activation versus the direct activation of CST fibers when TES is utilized. As described above,

when TES is applied in the awake animal, there is both a direct response (D-wave) from direct activation of the CST axons and also I-waves from indirect synaptic activation of the CST axons. Differing orientations of the coil will generate a response at differing latencies with respect to the D-wave produced by TES [48].

In the 1830s, Michael Faraday found that when a pulse of current is passed through a coil of wire, a magnetic field is generated. If a secondary conductor is nearby (within the induced magnetic field), a current is induced in this conductor that is related to the rate of change of the magnetic field [46]. When stimulating the brain using TMS, a coil is placed over the subject's head, and a brief pulse (usually around 100 μs) is passed through that coil generating a magnetic field that is large enough to pass through the subject's skull inducing a current within the brain. It is critical to point out that it is not the magnetic field that is directly stimulating the neural elements, but the secondary currents in the neural elements via induction. TMS has been tried during some surgical procedures [49], yet from a practical point of view due to the trans-synaptic nature of CT neuron activation, and the overall size of the stimulating element, TMS is not a suitable tool.

Electrical Elicited MEPs

The most common technique to elicit MEPs in the OR is via electrical stimulation applied to the scalp and/or exposed cerebral cortex and then to record the CMAP from the muscles. Using this technique, the functional integrity of both the CST and CBT can be continually monitored. The stimulus is applied over the motor cortex and recorded from the muscle or directly over the spinal cord. The montage and polarity used to apply the stimulation dictate the focalized nature of the stimulus, the laterality, and the extent of the artifact. For transcranial stimulation the montage can be categorized into bilateral (interhemispheric and midline) or unilateral (intrahemispheric). Using the international 10–20 EEG system, the standard MEP-stimulating electrodes are placed over the motor strip, and these are approximated with electrodes at positions C_1, C_2, C_3, C_4, and C_z, while for midline stimulation, having the cathode 6 cm anterior to C_z is also one possibility especially when muscle motor twitches disturb surgery. It should be noted that MacDonald recommends placing the leads a little more anterior to the standard central 10–20 locations and designates these as "M" locations [50]. The most common montages are the interhemispheric $C_1/C_2(C_2/C_1)$ and $C_3/C_4(C_4/C_3)$ montages. Making either C_1 or C_3, the anode will preferentially stimulate the CST fibers originating from the left hemisphere, while making either C_2 or C_4, the anode will preferentially stimulate the CST fibers originating from the right hemisphere. The $C_3/C_4(C_4/C_3)$ montage is able to elicit muscle responses in all four limbs but is preferential for monitoring upper limb MEPs, while the $C_1/C_2(C_2/C_1)$ shows a preference for the lower limbs, yet once again is able to elicit responses in all four limbs. The C_3 and C_4 montages have demonstrated the muscle activations with the lowest motor threshold in all four limbs [42] which might make it appear to be the most optimal for most MEP monitoring. An alternative is C_1/C_2 or C_2/C_1. C_3/C_4 and C_4/C_3 montages are known to cause large movement artifact. Instead of a focal stimulation, the stimulus is spread over a much larger area, in turn potentially activating many more fibers. Starting with the $C_3(C_4)/C_4(C_3)$ montage, due to it having the lowest motor threshold, is a good solution. Yet, it needs to be kept in mind that this montage has the potential of deeper current penetration, and thus in supratentorial surgeries, such as aneurysm surgery, the stimulation point may be caudal to the site of the surgery and therefore can miss a lesion to the CST. In this case using the $C_1(C_2)/C_2(C_1)$ montage may be more appropriate. Generally, in brain surgeries, direct stimulation of the exposed cortex via strip electrode is the method of choice. There are also other more focal montages such as the unilateral intrahemispheric C_3/C_z and C_4/C_z or the midline C_z/6 cm anterior to C_z. The $C_3(C_4)/C_z$ montage was shown to be appropriate for eliciting upper limb responses but was very poor in eliciting lower limb muscle responses. The $C_3(C_4)/C_z$ montage is the method of choice when eliciting corticobulbar responses such as those recorded from the vocal muscles

[51] or the facial muscles [52]. The focal montage is superior to that of the interhemispheric montages since direct stimulation of the facial nerve itself can occur with the larger spreading montage, without actually stimulating the CBT. This response may also give a false sense of security since the stimulation location may be distal to surgery and thus give false-negative results if the injury occurs proximal to the stimulation point. To exclude the possibility that the current spreads distally and directly activates cranial nerves, and not corticobulbar fibers, the use of single stimulus versus train stimuli is needed [51, 53]. Finally in a rare set of patients, using the midline montage of C_z/+6 cm to C_z may be beneficial for eliciting muscle responses from the lower limbs, yet the stimulus intensity needs to be high.

Stimulation intensity varies along with the MEP technique used. A theoretical calculation by MacDonald et al. showed that using pulse widths between 50 and 800 μs should allow for safe stimulation (below the level of damage to neural tissue) and that using a pulse width of around 200 μs is optimal for energy minimization based on the rheobase and chronaxie of the stimulated neural elements [40, 50]. It should be noted that each patient is somewhat different, and patient-specific physiology, disease state, and the patient's own response to anesthetic will affect the optimal stimulus parameters although the above ranges are good starting points.

At present there is no generally accepted ISI or train length as a standard for eliciting MEPs. Increasing the overall number of pulses within the train can reduce the stimulation threshold. It is also known that in some patients under light anesthesia, MEPs recorded from the muscles may be elicited by using one or two pulses, but in general the use of five pulses appears to be a good starting point [31]. Yet the use of more pulses (6–9) [54] or less pulses (3–4) [55, 56] is reasonable. Dong et al. [53] reported using three pulses when eliciting CBT MEPs. ISI starting points are also variable with the starting point ranging from 1 ms up to 4 ms. Szelényi et al. showed that using an ISI of 4 ms can minimize limb MEP thresholds [42], although using ISIs of 1 and 2 ms has shown to be best for both upper limb and CBT MEPs [50, 53, 57]. It is also worth mentioning that using an ISI of 2 ms is recommended for eliciting CBT MEPs because of their rather short latencies.

In the authors' experience, a pulse train of seven pulses with an ISI of 2 ms and a pulse width of 75 μs is a reasonable starting point for generating limb and CBT MEPs. Yet as discussed by MacDonald et al. [50], individual patient characteristics and anesthetic conditions may require altering of the parameters to get an optimal MEP response.

Direct Cortical Stimulation

In addition to transcranial stimulation for eliciting muscle MEPs, one can also stimulate the cortical surface [58–61] or subcortical space [62] directly. In order to help localize the motor strip, mapping the location of the SSEP phase reversal is recommended and described elsewhere in this book. For direct cortical stimulation (DCS), it is highly recommended to use a four- to eight-contact strip electrode placed over the specific region of interest. In cases where the motor strip is exposed, using a stimulation probe to localize the motor cortex is recommended since this will help guide the placement of strip electrode. In some cases, such as during aneurysm surgery, it may not be possible to directly test the cortex since the strip is usually placed under the skull due to that region not being included in the exposure. Thus for MCA aneurysm procedure, the strip would be placed over the lateral motor strip, while for ACA procedures, the strip is placed more medially. The cathode is placed at FPz (or as close as possible) with the anode (active, stimulating electrode) being one of the electrodes on the strip. Similar stimulation parameters to TceMEPs are used for DCS stimulation except that stimulation intensity should not exceed 25 mA [58]. In this study published by Szelenyi et al. [58], they were able to record MEPs from DCS in 84% of cases. Reasons for not being able to elicit MEPs with this method include seizure, brain swelling, premature aneurysm rupture, subdural scars, and patients with an aneurysm in the posterior circulations (it should be noted that in a small subsection of patients with anterior circu-

lation aneurysms, they did not place electrodes). Dislodgement of the electrode is an issue, yet we have found that once the electrode is in place, and by securing the lead wire with a staple, dislodgement of electrode was not an issue. One of the most frequent problems is the fact that the electrode contacts may not be over the motor strip. In some cases the surgeon might try to reposition the electrode, while in other cases, this has not occurred and we were not able to elicit MEPs. Szelenyi et al. have recommended that the surgeon uses the electrodes on the scalp when the exposure does not include the motor strip—the same ones used for TceMEPs—as a guide for placing the strip [58].

Once the electrode is placed, we start testing using a stimulation intensity of 10 mA if extradural or 3 mA if intradural and slowly increasing stimulation by 1–5 mA after five trials separated by 0.5–1 s. If no MEP response appears up to 25 mA for an extradural placement or 10 mA for intradural, then we switch the stimulating anode to the next electrode. We continue this until all electrodes are tested. The electrode with the lowest threshold is the one that is chosen to be used during monitoring. If no response is noted, we let the surgeon know this. The surgeon will then either reposition the electrode strip or continue without DCS MEPs.

In addition to stimulating the surface of the cortex, subcortical structures can also be mapped [62]. The primary reason for mapping these subcortical structures is to determine how close the resection is to the internal capsule. For these cases a monopolar stimulation probe is used. Using a pulse width of 75–500 μs (note that in the United States at the time of publication, a 500 μs was not available on all IONM devices) and a train of five to seven pulses with an ISI of 4 ms, stimulation is applied through a small ball tip probe of 1–2 mm. Stimulation intensity was increased to a maximum of 22 mA or until a muscle response was noted [62]. As the resection approaches the internal capsule, the threshold for CMAP activation reduces. It has been reported in the literature [63, 64] that the response to distance is 1 mA ≈ 1 mm. Thus for every decrease in stimulation intensity by 1 mA, the resection edge is 1 mm closer to the internal capsule. For distances greater than 5 mm between the resection cavity edge and the internal capsule, this ratio is acceptable. Yet, as the resection cavity becomes less than 5 mm away, the distance to threshold values becomes more nonlinear eventually approaching an asymptote (i.e., a minimal stimulation current needed to generate a response even if the probe is directly on the nerve). Seidel et al. demonstrate this effect by showing that as the threshold decreases to less than 3 mA, there is a significantly greater number of patients with permanent postoperative neurologic deficits compared to when the threshold stimulation amplitude is greater than 3 mA [62].

MEPs Recorded from the Muscles

Standard MEP monitoring uses the application of a stimulus at the head and the recording of potentials either from the spinal cord or muscle(s). The stimulus is applied via electrodes placed on the scalp for transcranial stimulation, overlying the dura, directly on the surface of the brain, or in the subcortical space. For transcranial stimulation, the subject matter of this chapter, gold cup electrodes, needle electrodes, or "corkscrew"-shaped electrodes could be used. Presently needle electrodes are the most commonly used, yet historically corkscrew and gold cup electrodes were preferred. Modern needle electrodes are of low impedance (around 400 Ω) which Journée et al. demonstrated [65]. MEP threshold is linearly related to impedance above 460 Ω, while below that MEP thresholds are constant [65]. Both the standard gold disk and the older needle electrode impedances are 800 and 1200 Ω, respectively. These electrodes are applied using the standard international 10–20 EEG system.[2] MacDonald et al. recommended

[2]Stimulation directly on the brain surface or dura uses other specially designed or modified electrodes and significantly lowers stimulus levels; otherwise the parameters and montage for stimulation are very similar. The technique of direct subcortical white matter stimulation is somewhat different in that the cathode is the stimulating (active) electrode which is different than for eliciting MEPs from the cerebral cortex.

placing the central stimulating electrode 1 cm in front of the standard 10–20 system placement of C_1, C_2, C_3, and C_4. This location better corresponds to the motor strip. The FPz electrode is at the standard 10–20 system location [57].

αMN innervated distal muscles receive the highest number of the large CST fibers and should be the matter of choice for recording limb MEPs. The most common muscles monitored are the abductor pollicis brevis, abductor hallucis, anterior tibialis, and forearm flexor and extensor carpi ulnaris. There are some situations where recording MEP responses from segmental muscle may be warranted. Such situations may include far lateral decompressions and foraminotomies [66]. In those cases the surgeon should be informed that MEPs may be less reliable due to the smaller numbers of large CST fibers innervating those muscles and also due to the potential overlapping between spinal roots [67]. Either surface or subdermal needles may be used to record muscle MEPs. The authors have found needles to be more stable and secure during long cases, yet care still needs to be taken due to the sharp nature of the needles and the fact they are not always visible to the surgical team. When using needles they should be placed in the muscle bellies about 2–3 cm apart. Table 7.1 lists a set of recommended muscles for MEP monitoring with the most likely innervation from the spinal root (the highlighted muscles are the best for monitoring general CT continuity).

When monitoring muscles innervated by the CBT, the electrode placement varies according to the muscle monitored. For muscles innervated by cranial nerves III, IV, and VI, it is preferable to use hook wire electrodes. Small needle electrodes placed in the skin parallel to each muscle are also an option, yet the selectivity and recorded EMG response are not optimal. The needles are placed in the muscle at about a 30° angle to the skin. The length of the needle should be around 1 cm. For cricothyroid (CRT) muscle recordings, either short needle electrodes or hook wire electrodes can be used. Hook wires are the recommended recording electrode since the large surface area of needle electrodes can give a false-positive or a false-negative result due to the large surface area of the electrodes recording far-field potentials from the neck muscles [51]. This *false data* may indicate that the functional integrity of the nuclei or CBT is intact when in reality that is not the case. Thus we recommend using the hook wire electrodes for recording. Hook wire electrodes placed in the vocal muscle require expertise of either an ENT specialist or anesthesiologist who is trained in this technique. For cranial nerve XII, we also recommend using hook wires in the tongue to minimize any damage from the needle due to movement which may lacerate the tongue. For cranial nerve IX, we have used both needles and hook wire electrodes in the soft pallet with equally good results. When placing needles or hook wire electrodes in the mouth for monitoring cranial nerves IX and XII, they should be placed after bite blocks are in place to minimize the chances of dislodgement of the electrodes during bite block placement.

Table 7.1 Common muscle-nerve root mappings

	Muscle	Cranial nerve or root
Corticobulbar	Orbicularis oculi	VII
	Orbicularis oris	VII
	Mentalis	VII
	Cricothyroid	X
	Vocalis	X
	Stylopharyngeus	IX
	Tongue	XII
Upper extremity	Trapezius	C4
	Deltoid	C5, C6
	Biceps brachii	C5, C6
	Triceps brachii	C6, C7
	Flexor carpi radialis	C6, C7
	Flexor carpi ulnaris	C8, T1
	Extensor digitorum communis	C7, C8
	Extensor carpi ulnaris	C7, C8
	Abductor pollicis brevis	C8, T1
	Abductor digiti minimi	C8, T1
Lower extremity	Iliopsoas	L1, L2
	Adductor longus	L2, L3, L4
	Vastus medialis	L2, L3, L4
	Biceps femoris	L5, S1
	Tibialis anterior	L4, L5
	Gastrocnemius	S1, S2
	Abductor hallucis	S1, S2
	External Anal Sphincter	S2, S3, S4

Stimulation intensity and the selected recording montage are dependent on where the stimulation is being applied. The actual stimulation current activating the neuron is the same no matter what montages we used; it is the intervening tissue that determines the actual stimulation current reaching neurons. When the stimulation has to penetrate the scalp and the skull, one needs a much higher stimulation intensity. About 80% of the stimulation energy is lost in TES. On the other hand, stimulation at the surface of the brain or at the white matter will require much lower stimulator delivered intensities due to no high impedances for passing current. Continuous MEPs elicited from the cortex are performed using a strip electrode placed over the motor area. The strip contact utilized is the anode, while a contact placed at FPz is the cathode.[3] For subcortical MEP mapping, the stimulating probe tip (the active electrode) is the cathode.

Recording of MEPs is done with a filter setting of 100–3000 Hz. We choose the 100 Hz high-pass filter to reduce the low-frequency artifact from the stimulator and flatten the response curve on the display. The low-pass filter can range from 750 to 3000 Hz depending upon the noise, yet as the filter is lowered, the high-frequency components can be lost. It is recommended not to change the filter settings during the procedure so the shape and amplitude of the waves are not modified by the filter. In some cases it will be necessary to adjust filter settings if new artifact is introduced during the procedure.

MEP Monitoring Using D-Wave

It is possible to record the traveling volley along the CST in the spinal cord during surgical procedures. This is performed by placing a disposable catheter recording electrode either sub- or epidurally both cranially and caudally to the site of the surgical intervention (Fig. 7.5). A commonly used electrode is the model CEDL-2PDINX-100 (Ad-Tech, Racine, WI) which has three 15 mm spaced electrodes. If it is physically possible, it is better to use for recording contacts 1 versus 3, but in some cases, it may not be possible to get all three electrodes to sit on the dura or the spinal cord, or in some cases, one of the contacts may fail and then another contact has to be used. D-waves are recorded with a 1–1.5 ms/Div time base, a high pass of 50–100 Hz, and a low pass of 1000–3000 Hz. Minimizing stimulation artifacts can be achieved by performing ten averages while switching the polarity during each average. The amplitude and latency of the D-wave vary depending upon the level of the spinal cord being recorded. In the cervical region, the amplitude is greatest with the shortest latency. As the electrode moves caudally down the cord, the amplitude reduces and the latency increases. The reduction of amplitude is due to the reduction of the number of large CST fibers contributing to the D-wave amplitude. The latency increase is related to the conduction speed in the spinal cord and the distance from the stimulating to the

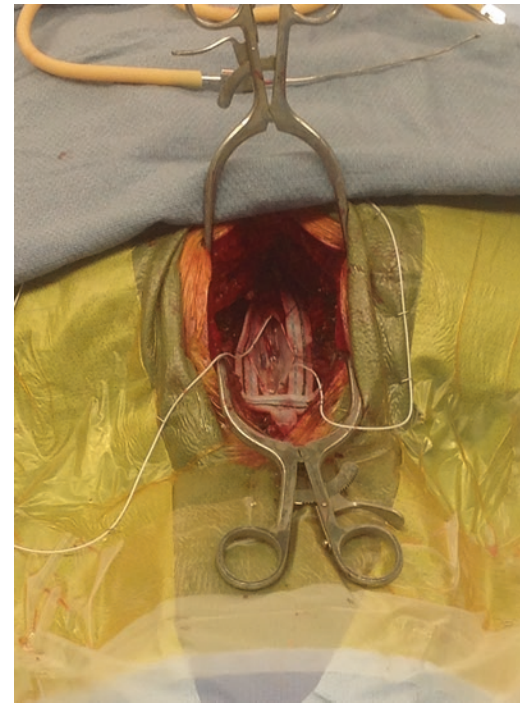

Fig. 7.5 Placement of both cranial and caudal D-wave electrodes during an intramedullary spinal cord tumor

[3] One may also use stimulation to map the cortex. This is not the subject matter for this chapter, although the techniques are similar.

recording electrodes. Other factors affecting the D-wave amplitude are related to the distance of the electrode from the spinal cord, the amount of damage to the CST, and the absolute level of the spinal cord where recording is done. Ulkatan et al. demonstrated that spinal cord anatomic position changes after correction of scoliosis can generate a false-negative D-wave amplitude change due to changes in the relative position of the epidural electrode to the CT. They also showed that no changes in the muscle MEPs occurred during epidural recorded changes indicating no injury to the spinal cord [68].

Neurogenic Response (Stimulation of the Spinal Cord with Recording from Peripheral Nerve)

Neurogenic MEPs were widely used in the 1990s but have since fallen out of favor due to the fact that there is no evidence that elicited recorded responses are generated by selective stimulation of the CST within the spinal cord [2, 69] and actually evidence proving that the response is mediated via the dorsal columns [2, 70, 71]. This technique requires the placement of stimulating needle electrodes between the spinous processes above the level of surgery (or in cases where the spinal cord or spine is exposed, one can use electrodes placed within the ligamenta flava or directly on the cord itself). For stimulation the cathode is placed caudal to the anode. Recording electrodes are applied over the sciatic nerve (or tibial nerve) in the popliteal fossa. Compound nerve action potentials (CNAPs) are then recorded. This method is based on hypothesis that CST fibers ending at αMN will be activated via the stimulation; therefore CNAPs represent activity from motor tracts [72]. It is known that antidromic stimulation of the dorsal columns [73] also activates the αMN, via branches of sensory rootlets ending up at the αMN using similar anatomic pathways that convey the H-reflex [71]. Furthermore, other motor tracts beside the CSST (e.g., the rubrospinal or vestibulospinal tracts) could activate the αMN. In fact the literature describes patients waking up with pure motor paraplegia who were monitored with neurogenic MEPs with no change in the neurogenic MEP during the procedure [2].

Indications and Contraindications for MEP Monitoring

Any surgery where there is risk of damage to the motor tracts or primary motor cortex should consider utilizing MEP monitoring. These surgeries include neurosurgical procedures in or near the motor cortex or CST and in the brain or brainstem, aneurysm clipping, or other vascular procedures that may affect the flow of blood to the motor system and also neurosurgical procedures of the spine, spinal cord, and cauda equina region. Orthopedic surgical procedures including spinal instrumentation for correction of spinal deformities, bony tumors, spinal cord decompression, and trauma and peripheral nerve entrapment correction procedures are possible procedures where MEPs are required. Vascular procedures such as carotid endarterectomy, aortic stenting, aneurysm repair, or spinal AVMs may require MEP monitoring as well. It is important to note that even with the general list mentioned above, there may be other procedures where potential damage to the motor system may warrant MEP monitoring, yet it is critical to note that given the pathology of the patient, the disease, and the goals of surgery, MEP monitoring may not be warranted, and thus every patient should be evaluated prior to surgery to determine if MEP monitoring is warranted.

Even though there is a large group of procedures where MEP monitoring may be warranted, MEP monitoring is not without its complications. Seizures are considered the second highest complication from MEP monitoring [53, 74, 75]. In 2002 MacDonald reviewed the literature and found the seizure rate for TceMEPs to be 0.03% [74]. When performing direct cortical MEP monitoring during aneurysm surgery, Szelényi et al. found a 1% seizure rate [53]. The risk for MEP-induced seizures in patients with symptomatic epilepsy was 1.5% using the high-frequency short-train mapping technique compared to the low-frequency long-train mapping technique

which was 9.5% [76]. Thus, in general, the rate of seizures is rather small, yet in those patients with a history of seizures, or pathology that may enhance its generation, immediate cessation of seizure could be achieved with irrigation of the cerebral cortex (if exposed) with ice-cold saline. This can usually halt the seizure within 5–10 s [77]. In addition antiseizure medication can be given, yet this alone can inhibit the generation of MEPs. Also, a detailed discussion should be with the surgeon so they can understand possible risks of monitoring MEPs as well as the risks of iatrogenic injury if MEPs are not used.

For both open cranial and spinal procedures (where direct access to the brain is not possible), Ativan (lorazepam), diazepam, midazolam, all benzodiazepines (barbiturates), or bolus of propofol [78] can help in halting the seizure. Yet, once a medication is given, it becomes rather difficult to record MEPs due to the cortical inhibition caused by the drug.

Lip and tongue lacerations are the most common complications of MEP monitoring and have a reported incidence rate of 0.2% [74]. Their most likely explanation is due to the contraction of the jaw musculature triggered through the motor part of the trigeminal nerve or even the CBT pathways [50]. To minimize this complication, it is highly recommended that dual bite blocks be used and placed in between the upper and lower jaw on both sides of the mouth (Fig. 7.6).

Other complications include burns under the stimulating electrodes, movement-induced injuries, transient cardiac arrhythmias, and potential damage to vascular structures with the use of electrode placed over the cortex. Burns are due to a buildup of heat between the stimulating electrode or even the recording electrodes and the skin in most cases due to the faulty cautery [74]. In cases where the electrodes are screwed into the scalp too tightly, there may be a cutoff of blood flow and thus no way for heat to be removed causing burns. The more common cause for burns is with equipment failures. If the return current, of the cautery system, or the ground of the IONM system fails, the electrodes, both stimulating and recording, may become those returns causing excessive current to pass through the small stimulating and/or recording electrodes generating burns. Any time a burn is noted during electrode removal, it is recommended that every piece of electrical equipment that comes into contact (either directly or indirectly) with the patient be checked by the hospital's biomedical engineering department/personnel. Once again this discussion should include the benefits and negatives of MEP monitoring during the procedure. Szelenyi et al. [58] stated that 2 of the 100 patients in whom DCS with strip electrodes was used had bleeding from bridging veins damaged during electrode placement. This bleeding caused no neurologic sequel in either patient.

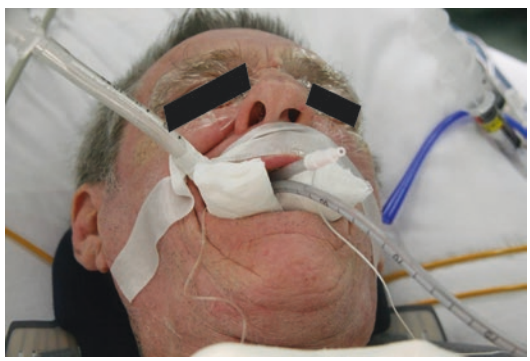

Fig. 7.6 Example of a double bite block to protect against lateral tongue lacerations

Interpretation and Alarm Criteria

Interpretation of MEP data is dependent on the location of surgery and type of surgery being monitored. What this means is that interpreting changes in MEPs during the monitoring of surgery for cerebral aneurysm appears to be different than monitoring during a scoliosis or other spinal procedures. Yet, there are some key principles when interpreting MEP changes and deciding whether criteria for an alarm have been reached. The primary alarm marker for MEPs is a change in amplitude. One of the first questions to answer is the time course of the change. Was the change gradual or was it over a very short period of time? Gradual changes tend to indicate something systemic is going on, i.e., changes in

the depth of anesthesia or blood pressure. Yet, fast changes may also be related to anesthetic effects, i.e., bolus applications of anesthetics. Thus, anesthetic and technical issues need to be evaluated very quickly during the troubleshooting. This is why it is highly recommended to continually review anesthetic concentrations and work closely with the anesthetic team to assure that any application of anesthetic is passed to the IOM team. In addition to the time course of the change, the focality of the change is also important. In general, focal changes are likely due to iatrogenic injury if all other technical factors can be ruled out. Although if working at the cervical spinal cord a systemic loss of MEPs would be due to a iatrogenic injury, an alarm should be issued immediately to evaluate the situation.

Effects of anesthesia on the MEP have been described earlier in this chapter. Yet, muscle relaxant has a significant effect on the MEP response. It is important to perform a train-of-four (TOF) test when using muscle MEPs in order to assure no muscle relaxant is in the patient's system. Some authors have described acceptable muscle MEPs when using a 2/4 TOF response. The authors find this to be an unacceptable state to monitor MEPs in. Obviously, if there is no response in any of the four twitches, then there will be no muscle MEP; the literature describes muscle MEP responses with at least two out of four twitches (see Sloan and Jantti [79]), yet given the variable nature of MEP amplitudes when not stimulating supermaximally, even in the cases when no muscle relaxant is being used, the authors recommend that no muscle relaxant be administered after intubation. It is also important to note the expected length of the procedure, since some relaxants have longer half-lives than others. This means that some relaxant such as non-depolarizing agents such as vecuronium and rocuronium will take a longer time to wash out and thus make it more difficult to record MEPs early in the procedure, where a depolarizing muscle relaxant, such as succinylcholine, will wash out much faster. In some instances the surgeons may want to have the muscles relaxed during back exposure or no movements during other exposures. Succinylcholine (SCh) is a common example of a short-acting depolarizing neuromuscular blocking agent that allows for quick recovery and monitoring of muscle MEPs. Yet, it is important to note that in cases of trauma, potassium abnormalities, malignant hypothermia, or other skeletal muscle issues, SCh should not be used [80], as well as other issues where a preoperative discussion with the anesthesiology team can be beneficial. As the muscle relaxant is wearing off, MEPs from the upper limbs will tend to return to full TOF 4/4 sooner than the lower limbs. In addition atrophied muscle or muscles innervated from damaged nerve roots may return at a slower rate than the "normal" tissue.

Basic alarm criteria for MEPs are mostly concerned with amplitude reductions. Criteria range from 100% loss to a 50% loss for spinal procedures [4, 5, 39, 80–83] and 50–60% loss for cranial procedures [58, 84] for muscle MEPs. For cranial procedures the alarm criteria of 50–60% reduction appear consistent and appropriate, yet for spinal procedures the alarm criteria are less concrete. Anesthesia primarily affects the synaptic transmission at the αMN. In addition, each TcMEP trial does not activate the complete pool of αMNs; thus for each trial, the number of excited αMNs is different which is another reason for the variable amplitude. For long cases there is a phenomenon known as anesthetic fade where the MEP amplitudes decrease over time with the stimulus level. This phenomenon is exacerbated by myelopathies. It is important to realize that this is a very slow change and not abrupt.

For epidural recordings (D-wave), the alarm criteria are more reliable. The D-wave is a function of stimulation at one point on the CST and recording at another point. The D-wave is less susceptible to anesthetic effects, and its amplitude is directly related to the stimulus amplitude (for the most part). The D-wave amplitude is proportional to the number of fast-conducting corticospinal fibers. In addition it has been shown to be very stable over time [12]. With this in mind, it appears that a 50% reduction in D-wave amplitude is indicative of cord injury and an alarm should be given to the surgical team [31, 39, 50, 85]. Yet as described by Yamamoto et al., during brain tumor surgery, a decrement of <30% is cor-

related with recovery, while there was a persistent motor deficit when greater than 30% [86]. This is in concordance with the 50% alarm criteria used in spinal surgery, given that in cranial surgery, only one hemisphere is being affected, and thus one CT is being manipulated. Thus, the surgical region location is important in choosing the appropriate MEP alarm criteria.

In addition to amplitude reduction criteria, there are also stimulation threshold elevation changes and morphology changes. Calancie et al. describe a technique that uses stimulation threshold changes to make predictions and generate alarms intraoperatively [55, 56]. Using this technique the MEP stimulation threshold is determined at the beginning of the case. A >100 V increase in stimulator delivered intensity for greater than 1 h is predictive of a poor motor outcome. Quiñones-Hinojosa et al. looked at morphology changes in the muscle MEP as an indicator or damage to the spinal cord [84]. Using this method the authors investigated the complexity (the number of peaks and troughs in the waveform) as an indicator of outcome. One of the most reliable alarm measures uses the combination of the D-wave with muscle MEPs to predict outcome while in addition offering a very stringent alarm criteria [39]. Using this technique a complete loss of the MEP and a >50% decrement in the D-wave result in a complete paraplegia, while a loss of the muscle MEP with no change in the D-wave amplitude or a less than 50% decrement will result in a temporary motor deficit [31, 39]. The shortcoming of this technique is that it requires the invasive placement of an epidural electrode which the other techniques do not. When looking at all of the factors that can affect interpretation of the MEP, it is important to note that each technique is not truly independent. Amplitude and morphology tend to be related. Thus, when the morphology changes, i.e., going from a complex polyphasic wave to a biphasic or monophasic wave, the amplitude of the peak tends to reduce as does the total energy in the wave. Another factor is the highly likely possibility of incomplete motor pool activation. Repetitive trials can help overcome this incomplete activation (Fig. 7.7). Using the paired-pulse technique of Journée et al. can help to minimize the false-negative rate experienced in the OR [43].

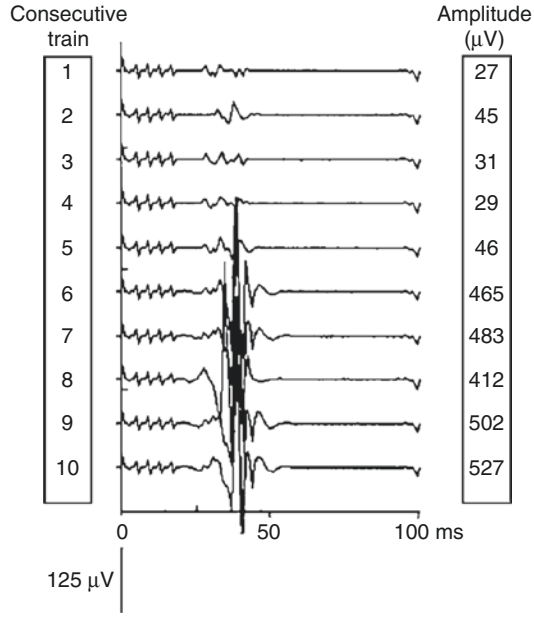

Fig. 7.7 Multi-MEP trials demonstrating the buildup effect. Note by the sixth trial the amplitude of the MEP has increased tenfold. (Reprinted with permission from Deletis [31])

Conclusion

Monitoring of the motor system, as in all IOM, is not simply looking at waveforms. Each modality, including MEP monitoring, includes special conditions that can confound the interpretation. The physiology of the motor system adds complexity to MEP monitoring by adding variability to each trial. Understanding this physiology is critical to properly performing and interpreting the MEP intraoperatively.

Review Questions

1. Describe how D-waves and I-waves work together to create activation of the alpha motor neuron.
2. How does anesthesia inhibit activation of the alpha motor neuron and what stimulation techniques can overcome this effect?
3. How can you assist a surgeon who is concerned that his tumor resection margins may be too close to the CST as it passes through the internal capsule?

4. What would you tell a surgeon if you were monitoring both TcMEPs and D-waves and saw a loss of MEP responses but a less than 50% change in D-wave amplitude?
5. Why are the most distal muscles preferred recording sites for TcMEP monitoring?

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