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Tutorials Articles

Exploring motor behavior through transcranial magnetic stimulation: a comprehensive tutorial from the neurobiomechanics and motor control perspectives

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ABBREVIATIONS

aMT	Active motor threshold
ANOVA	Analysis of Variance
CoG	Centre of gravity
DELMEP	Deep learning-based algorithm
ECG	Electrocardiogram
EMG	Electromyography
EOG	Electrooculogram
IFCN	International Federation of Clinical Neurophysiology
IZ	Innervation zone
M1	Primary motor cortex
MEP_{P-P}	MEP amplitude
MEPs	Motor evoked potentials
MRI	Magnetic resonance imaging
MUAPs	Motor unit action potentials
MVICs	Maximal voluntary isometric contractions
nTMS	Neuronavigated TMS
ppTMS	Paired-pulse TMS
CS	Conditioning stimulus
TS	Test stimulus
ISI	Interstimulus interval
TEPs	TMS-evoked potential
rMT	Resting motor threshold
rTMS	Repetitive TMS
sEMG	surface EMG
TMS	Transcranial magnetic stimulation
TMS-EEG	TMS and electroencephalography

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ABSTRACT
Transcranial Magnetic Stimulation (TMS) is a well-established method for probing corticospinal and intracortical processes, yet its application in biomechanics and motor control research often lacks methodological integration. This tutorial synthesizes fundamental and advanced procedures for applying TMS within neurobiomechanical frameworks, highlighting how neurophysiological measures relate to motor behavior. We detail practical considerations for recording motor evoked potentials (MEPs) using surface electromyography (sEMG), including electrode montage selection, coil positioning, neuronavigation, and data-processing approaches. We also outline protocols for single- and paired-pulse TMS, motor mapping, and TMS-EEG integration, emphasizing how each technique interfaces with biomechanical variables, including posture, muscle length, and task demands. By integrating foundational concepts and practical guidelines on TMS applications to movement-related mechanisms, this tutorial provides a structured guide for designing, executing, and interpreting studies at the intersection of neurophysiology, biomechanics, and motor control.

KEYWORDS: Transcranial Magnetic Stimulation | TMS | Cortical excitability | Motor mapping | Electromyography | Motor Evoked Potential

INTRODUCTION

Transcranial magnetic stimulation (TMS) is a non-invasive tool that enables the assessment and modulation of human cortical function [1,2]. Clinically, single and repetitive TMS (rTMS) pulses are approved for specific therapeutic indications – most notably treatment-resistant depression, migraine, and obsessive–compulsive disorder. In research, TMS is widely used to investigate sensory-motor integration, corticobulbar and corticospinal excitability, and brain-behavior relationships. Single-pulse TMS enables the evaluation of

corticobulbar and corticospinal excitability by eliciting motor evoked potentials (MEPs) recorded using electromyography (EMG) [3]. Single-pulse TMS can also be combined with electroencephalography (TMS-EEG) to probe cortical reactivity and functional connectivity [4]. In turn, rTMS can modulate neural circuits by engaging mechanisms related to synaptic plasticity, including frequency-dependent excitation or inhibition, and state-dependent interactions with ongoing brain activity [2].

Although comprehensive safety guidelines and methodological recommendations exist [1,2], they primarily address clinical or neurophysiological procedures and provide limited guidance for researchers working at the intersection of neuroscience, biomechanics, and motor control. As TMS becomes increasingly integrated with kinematic, kinetic, and EMG analyses, there is a growing need for tutorials that explicitly bridge neurophysiological measures with biomechanical and motor control paradigms.

Traditional biomechanical assessments – including muscle mechanics, movement kinematics, and joint kinetics – describe the external and internal forces underlying motor behavior but offer limited insight into the neurophysiological mechanisms that shape these outputs. It highlights the importance of methodological approaches that integrate central and peripheral measures. Conversely, TMS-based measures can reveal the excitability [5], connectivity [6], and dynamics of the motor system [7]. Yet these physiological signals are frequently interpreted without considering the biomechanical context in which behavior emerges. Integrating TMS with biomechanical and motor control assessments enhances the ability to interpret how neurophysiological processes shape observable movement patterns [8,9]. Although variables such as muscle mechanics, kinematics, and kinetics efficiently describe movement, the neural mechanisms governing these outputs are often overlooked [10]. A guided, standardized tutorial is therefore essential to help researchers navigate the complexity of TMS experimental setups and stimulation protocols.

This tutorial fills this gap by providing both introductory and advanced explanations of single- and paired-pulse TMS specifically tailored for biomechanics and motor control research. We outline conceptual and practical strategies for linking TMS-derived neurophysiological signals to movement-related variables, highlight common pitfalls, and discuss how electrode configurations, coil positioning, and analytical decisions in TMS experimental procedures influence MEP properties and their interpretation.

Physical Principles and Instrumentation in TMS

Physical Principles and TMS Coil Types

The physical principle governing TMS is the electromagnetic induction, described by Faraday in 1838. The TMS pulse consists of discharging a bank of capacitors, charged to approximately 1500-2000 V, generating an electrical current of 2,000 to 8,000 amperes in approximately 60-100 microseconds, which passes through a coil placed on a person's scalp. This time-varying current induces a magnetic field of around 1-2 Tesla. The magnetic field itself does not attenuate as it penetrates the head; rather, it depends primarily on the distance between the TMS coil and the head surface. Depending on factors such as the TMS coil type, pulse configuration, and head composition, the magnetic field is estimated to penetrate approximately 2-4 cm. Due to the time-varying magnetic field, an electric field is induced in the cortical tissue, potentially altering neuronal membrane potential. When it reaches the depolarizing threshold, it triggers an action potential [1,2,11].

The topography of the induced electric field is intrinsically related to the different geometries of TMS coils. Among the TMS coils available commercially, the round (Figure 1a) and figure-of-eight (also known as butterfly-shaped; Figure 1b) coils are the most widely used by researchers and clinicians. The focality of a figure-of-eight is typically measured as the full-width-at-half-maximum (FWHM) of its electric field distribution, ranging from 28 to 37 mm [12]. In turn, round coils produce a ring-shaped electric field without a specific focal region. The more specific focal area from the figure-of-eight coil makes it suitable for better spatial mapping of motor cortical representations [11]. Circular coils lack the same focality and are commonly applied for peripheral stimulation [13]. Figure-of-eight coils may also have angled wings, named double-cone coils (Figure 1c). Double-cone coils deliver stronger stimulation through improved inductive coupling with the cortical surface, enabling activation of deeper sulcal areas, such as the leg motor representation. However, double-cone coils induce a broader electric field distribution with an FWHM of about 80 to 90 mm [14]. H-coils have been reported to penetrate deeper, reaching up to 6 cm [15]. However, as in all TMS applications, regardless of coil configuration, the strongest electric field is always induced at the cortical surface, and the focality-depth trade-off must be considered. Namely, for a TMS pulse, an increased stimulation depth is always associated with a broader electrical field distribution (lower focality).

Depending on the stimulation frequency and intensity, the TMS coil may overheat, or the stimulus intensity may decay across trains of pulses. These occur because a high electrical current (~ 2-8 kA) passes through the coil windings, which have a corresponding resistance, dissipating energy as heat. The two most commonly used TMS pulse waveforms are the monophasic pulse, which induces a predominantly unidirectional electric field, and the biphasic pulse, generated by a rapidly reversing bidirectional current [16]. Modern equipment already supports the generation of customizable pulse shapes [17], providing additional flexibility for devising new stimulation protocols. Due to the different effects of each pulse waveform on the corticospinal system, it is recommended to carefully verify the exact pulse specifications and their intended neurophysiological mechanisms.

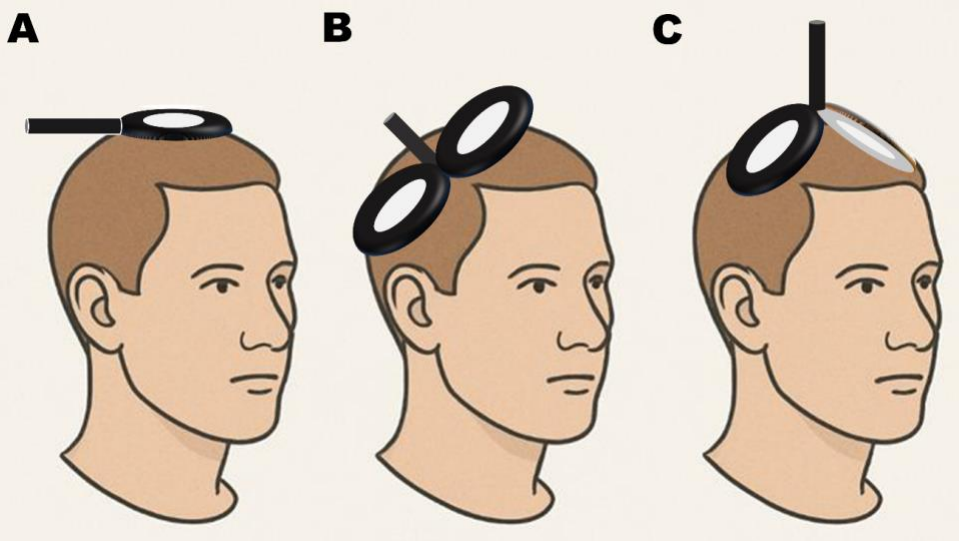

Figure 1. The most common types of TMS coils used in clinical and scientific practice are presented above: a. circular coil; b. figure-of-eight or butterfly-shaped coil; c. double-cone coil. The positioning of each coil may vary according to the objectives and the targeted cortical stimulation areas.

TMS Coil Positioning and Neuronavigation

The primary motor cortex (M1) is a common target, and locating TMS cortical targets is commonly performed manually due to their simplicity, cost-effectiveness, and accessibility. It is worth mentioning that the most common method for identifying M1 targets relies on anatomical landmarks and standardized measurement protocols. The "5 cm rule" is frequently applied: the coil is placed 5 cm lateral to the vertex (Cz in the 10-20 EEG system) along a parasagittal line, targeting the hand knob area [18]. This approach, however, assumes uniform head anatomy across individuals – a limitation given the variability in cortical folding and scalp-to-cortex distance.

Additionally, this process relies heavily on operator expertise and iterative adjustments, making it susceptible to inter-session variability and subjective error. While cost-effective and widely accessible, manual identification can be particularly demanding when targeting small or deep cortical regions, where even minor TMS coil displacements (e.g., > 3 mm) may significantly alter outcomes. Such limitations underscore the trade-off between practicality and precision in non-navigated TMS protocols. Considering that this methodological approach introduces notable and reliable challenges, a cap with a grid of 1 cm marks can ensure the stimulation site, as shown in green and red in Figure 2, allowing the operator to make marks with colored pens.

Regarding the orientation of the TMS coil, the literature suggests that – at least concerning the hand area representation in M1 – it should be positioned at about 45º [19] relative to the midline or parasagittal line of the body. This orientation results in an electric field generated by the TMS coil aligned with the predominant orientation of neuronal cortical columns along the central sulcus [19]. Therefore, if the TMS coil is kept at the same coordinate while being rotated along its axis, significant reductions in MEP amplitude (MEP_{P-P}) may be observed. Thus, positioning a figure-of-eight TMS coil, as shown in Figure 2, is recommended. Other TMS coils can be placed in different ways on the subject's head, depending on the cortical stimulation area.

In contrast, neuronavigation systems (e.g., InVesalius Navigator; https://github.com/invesalius/invesalius3) address these limitations by integrating magnetic resonance imaging (MRI) derived individual anatomy with real-time tracking of the TMS coil position and orientation [20]. These systems use infrared or electromagnetic tracking to align the TMS coil with the target cortex, accounting for sulcal morphology and ensuring consistent coil placement (Figure 3). Such precision is indispensable for studies that require repeated sessions (e.g., therapeutic interventions) or target non-motor regions (e.g., premotor or cerebellar areas involved in biomechanical coordination).

Neuronavigated TMS (nTMS) also enable retrospective analysis of stimulation sites, thereby enhancing reproducibility across laboratories. While resource-intensive, its use is increasingly justified in mechanistic studies and clinical trials where target accuracy directly impacts outcomes. Several practical considerations are critical for effective TMS coil positioning regardless of the methodological model used. First, maintaining TMS coil tangency to the scalp is essential to minimize the distance to the underlying cortex and ensure optimal stimulation intensity. Second, stabilizing the TMS coil orientation – such as positioning the handle posteriorly for M1 stimulation to induce posterior-anterior current flow – enhances the consistency of the induced electric fields [20].

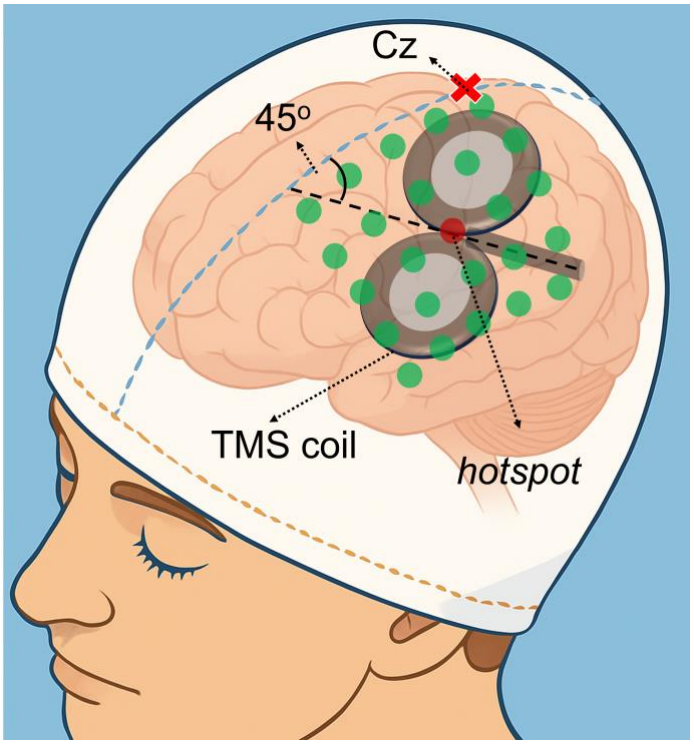

Figure 2. Assuming the use of a cap, it allows for the representation of a 5 x 5 grid of coordinates projected over different regions of the left cerebral cortex, intending to support TMS coil positioning. The figure-eight or butterfly TMS coil is positioned at a 45º angle to the midline and over the hotspot (indicated by the red circle). The red cross provides the Cz coordinate, a valuable head reference for the TMS coil positioning.

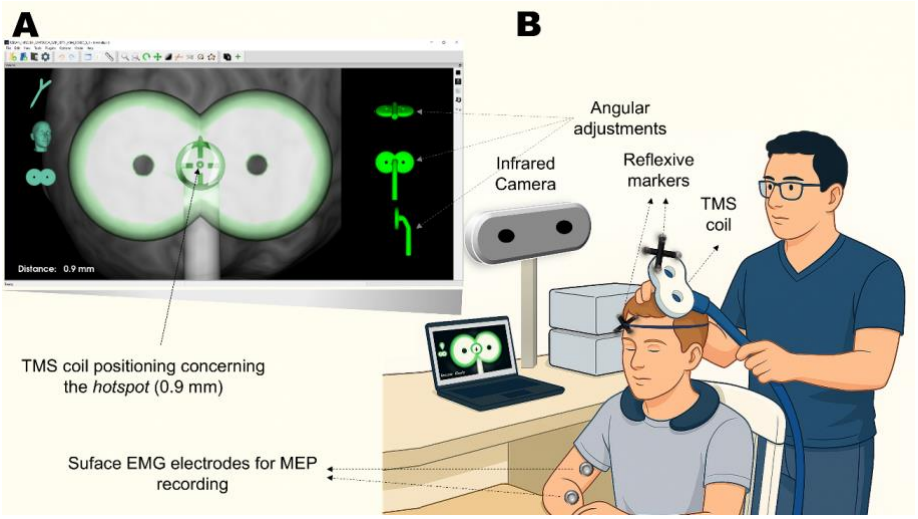

Figure 3. Schematic illustration of a neuronavigation system based on an infrared spatial tracking device and the InVesalius Navigator interface. In A, a screenshot of the InVesalius Navigator system is shown. The navigation interface provides an example of the TMS coil positioning relative to the hotspot (0.9 mm from the target) and with its three rotations aligned as intended, enabling consistent and accurate cortical stimulation. In B, an infrared camera-based spatial tracking system is shown. Reflective markers are placed on the subject's head and the TMS coil. Neuronavigation combines the head and TMS coil coordinates with a magnetic resonance image of the subject's brain anatomy to provide dynamic feedback to the operator during the TMS session.

TMS and Electromyography (EMG)

EMG is an electrodiagnostic technique widely used by researchers and clinicians to record muscle activity under various conditions [21]. Although invasive and surface electrodes can be used to record myoelectric activation, we aim to discuss the non-invasive approach, i.e., surface EMG (sEMG), for TMS applications.

For TMS-EMG integration, the recording system must meet basic technical requirements. Adequate triggering between the TMS device and the EMG system is highly recommended, preferably via a trigger signal, to allow precise identification of each TMS pulse for post-processing MEPs. This trigger system can be built into the system or implemented with external devices, such as home-made circuits using microcontrollers. It facilitates the extraction of stimulus-locked epochs from continuous EMG recordings, which are essential for reliable latency and MEP_{P-P} analysis. Since the artefact and the MEP appear as brief, isolated events in the sEMG signal, their spectral content tends to be concentrated in the 50-500 Hz frequency range. Thus, sampling frequencies higher than 2 kHz are desirable.

Electrode placement has a substantial impact on MEP_{P-P} and its reliability. Although there are recommendations for the use of surface electrodes in movement analysis, such as SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles; www.seniam.org) [22], the MEP_{P-P} can vary by up to ~6 times depending on the electrode montage [23]. SENIAM recommends placing two electrodes 1-2 cm apart, but not within the innervation zone (IZ). Garcia et al. [23] demonstrated that bipolar SENIAM-based placement may underestimate corticospinal excitability. In turn, guidelines [1,2] published by the International Federation of Clinical Neurophysiology (IFCN) suggest a pseudomonopolar montage based on the muscle belly-tendon or body prominence. For MEP recording, which relies on synchronous activation of multiple motor units, detecting a larger and more coherent portion of the muscle can be advantageous. Stålberg et al. [24] provide only limited support for the hypothesis of eventual localization of the IZ in the muscle belly. Garcia et al. [3,23] report that recording from the IZ would enable MEP detection with a greater MEP_{P-P} due to a higher probability of coherent summation of motor unit action potentials (MUAPs). The simplest way to locate the IZ is using an atlas [25], but the IZ can also be estimated by electrostimulation. Electrical devices such as acupuncture pens can help locate the motor point, which may be a few millimeters from the IZ. As Barbero et al. [25] adopted, it is worth noting that the gold standard for locating the IZ is electrode arrays. Figure 4 exemplifies how the bipolar SENIAM-based and the pseudomonopolar IZ montages are defined on a fusiform muscle.

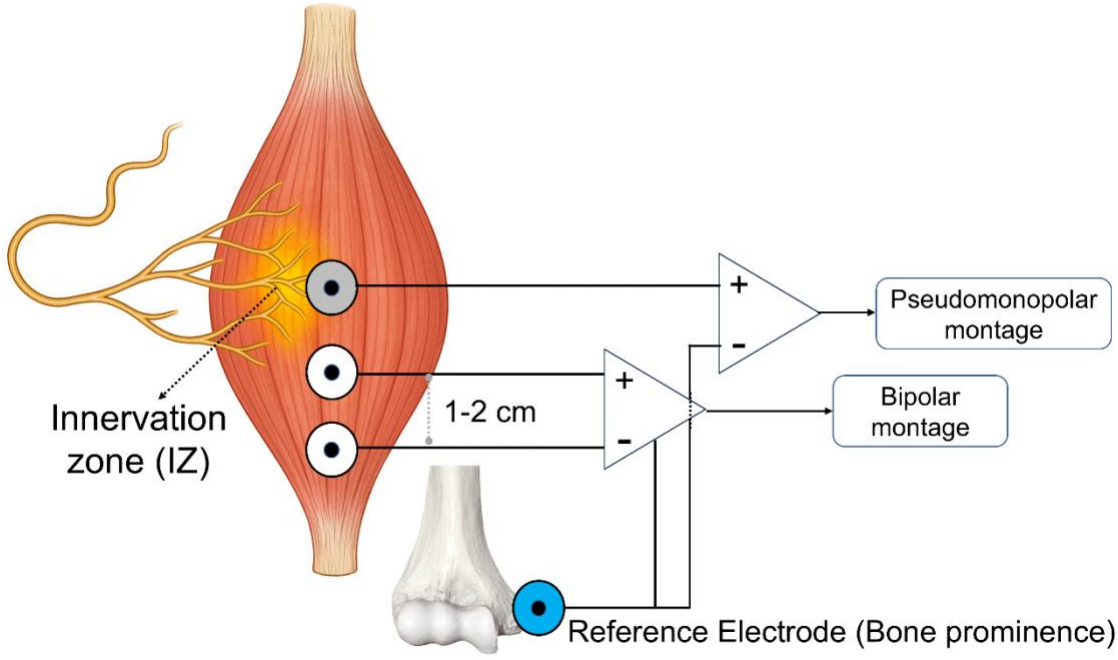

Figure 4. Two different electrode montages are presented for recording MEPs in a fusiform muscle: Bipolar versus Pseudomonopolar. It should be noted that, according to Hermens et al. [22], bipolar montage is far from the innervation zone (IZ), whereas pseudomonopolar montage requires an electrode over the IZ. According to Garcia et al. [2,23], the pseudomonopolar montage appears to offer advantages for MEP recording, as discussed above.

In relation to the MEP properties, two parameters are often taken into account, as can be seen in Figure 5: a. the peak-to-peak amplitude (MEP_{P-P}); and b. latency. To obtain both, after identifying the stimulation onset, the EMG signal is segmented into a post-stimulus window where the MEP typically occurs. A time window of 10-60 ms after the TMS pulse is generally used. To calculate the MEP_{P-P}, within this interval, the maximum positive and negative deflections of the waveform are identified, and the MEP_{P-P} is computed as the difference between them. Filters can also be applied, including band-pass filters in the EMG range to remove low-frequency drift and high-frequency noise and notch filters to minimize electrical noise. For latency estimation, it is even more critical to use a trigger or marker to indicate the moment of TMS pulse application, given the variability in latency, which ranges from a few to several tens of milliseconds, depending on the muscle and anthropometric characteristics of the subjects. Although one can manually extract latency from EMG marks or develop their own processing pipeline, Milardovich et al. [26] developed a deep learning-based algorithm (DELMEP) to maximize MEP latency estimation. According to the authors, DELMEP yields significantly lower mean absolute errors (~0.5 ms) and high accuracy.

Besides choosing a suitable surface-electrode montage for MEP recording, there are important considerations when analyzing the data. MEP data typically exhibit a positively skewed distribution, which can distort the analysis if a few very large values are present. Log transformation is necessary to make the distribution of repeated MEP_{P-P} measurements more symmetrical and to approximate it to a normal distribution [27]. It is crucial for many statistical analyses that assume normality, such as Analysis of Variance (ANOVA) and linear mixed-effects models. Notably, linear mixed-effects models provide a more transparent and statistically reliable approach to analyzing repeated-measures experimental designs, which benefit from the use of raw, non-normalized MEP measures [28,29].

Alternatively, MEP_{P-P} may be normalized using two possible methods [30]. The first is based on external references. This method typically references maximal voluntary isometric contractions (MVICs) based on the envelope or rectified MVIC signal. However, this method may increase between-subject variability without significantly enhancing stability. In turn, the second one uses the largest MEP_{P-P} as a reference and may reduce between-subject variability by up to ~ 64%. However, normalized data may require nonparametric methods or ratio-based statistical techniques, which can be less powerful or bias the interpretation of TMS studies in which MEPs exhibit intrinsic variability. Nonetheless, for repeated measurements of MEP_{P-P}, at least 20 TMS single pulses are recommended [31].

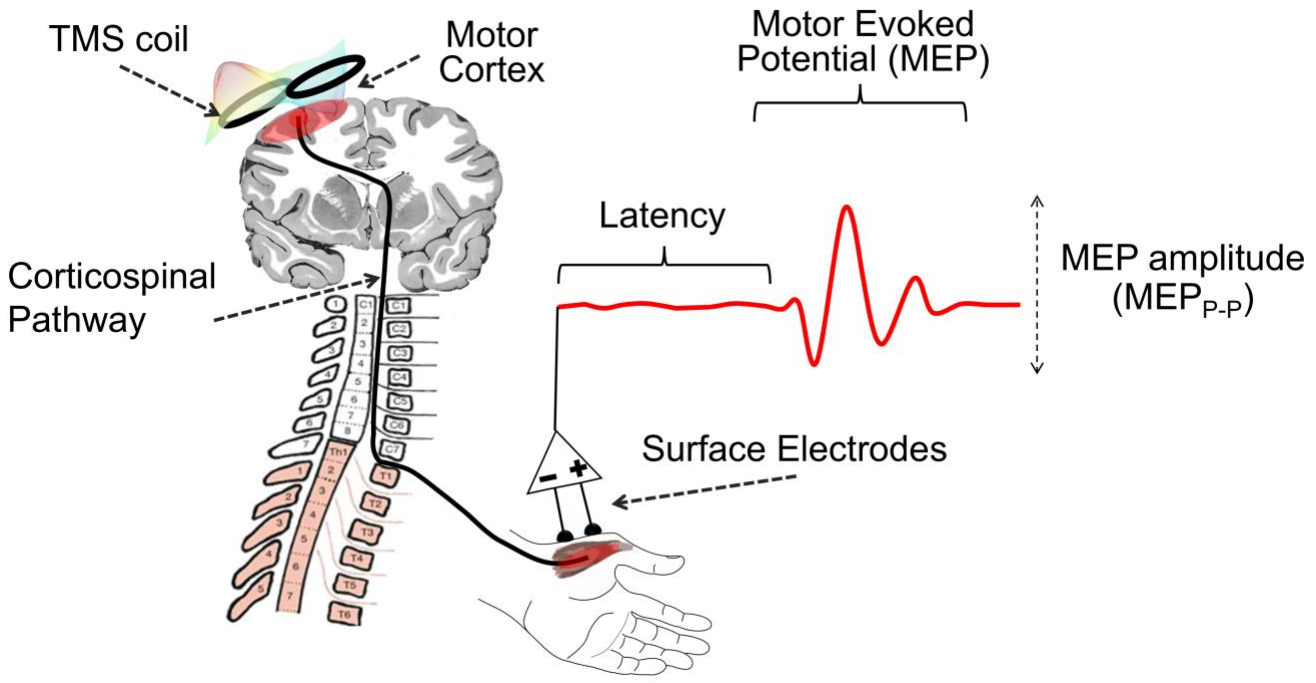

Figure 5. Schematic representation of the MEP profile recorded via surface EMG from the thenar region following single TMS. The parameters latency (time elapsed between the TMS pulse and MEP onset) and MEP_{P-P} exemplify the evaluation of corticospinal excitability.

TMS and Electroencephalography (EEG)

Integrating TMS and EEG (TMS-EEG) has become a valuable non-invasive method for investigating human brain dynamics. It enables the evaluation of key cortical features, including excitability and connectivity [4]. Over the last decade, this approach has been increasingly applied across different clinical populations, contributing to the identification of potential TMS-EEG biomarkers and predictors related to treatment response and the underlying mechanisms of brain disorders [32]. Furthermore, TMS-EEG enables individualized therapeutic interventions through closed-loop systems, allowing real-time modulation of brain activity based on ongoing neural responses [33].

Due to the electromagnetic interaction between TMS and EEG, simultaneous use of the techniques is challenging. Synchronized post-synaptic potential fields induced by TMS pulses must overcome multiple barriers before reaching the EEG electrodes, e.g., the brain, meninges, skull and skin, each with different electrical conductivities. Furthermore, the EEG signal records activity from sources other than the brain, such as cardiac, muscular, and auditory activity, as well as ambient electrical noise. As a result, recent advances by leading researchers in the field of TMS-EEG have synthesized essential preparatory steps and best-practice guidelines to ensure the acquisition of high-quality recordings while minimizing artefacts induced by magnetic stimulation [34]. The steps include skin preparation; EEG cap proper positioning; placement of reference and ground electrodes; monitoring physiological signals such as electrooculogram (EOG), electrocardiogram (ECG), and sEMG activities; and artefact minimization, including those from the mechanical impact of TMS pulse, electromagnetic coupling and the auditory perception of TMS-induced clicks.

In summary, implementing these standardized preparatory steps and best-practice recommendations is essential to optimize data quality and ensure methodological rigour in TMS-EEG experiments. By systematically minimizing artefacts and enhancing signal reliability, researchers can more accurately investigate the neural mechanisms underlying TMS, thereby advancing both experimental and clinical applications of this technique. Furthermore, TMS-EEG enables personalized therapeutic interventions through closed-loop systems that facilitate real-time modulation of brain activity based on ongoing neural responses [4,33,34,35].

Brain Function Assessment Using TMS

The Hotspot and Motor Thresholds (rMT and aMT)

The hotspot is the cortical coordinate at which the lowest single-pulse TMS intensity reliably elicits an MEP of a predefined magnitude in the target muscle (commonly 50-100 μV MEP_{P-P} under resting conditions). The hotspot should be determined systematically using a grid or neuronavigation; when neuronavigation is not available, a cap with grid marks (1 cm spacing) provides a pragmatic alternative. Apply an even number of pulses (e.g., 6-10) at candidate coordinates with inter-stimulus intervals of 5-10 s, and select the site with the most consistent responses.

Resting motor threshold (rMT) and active motor threshold (aMT) are widely used reference points for scaling stimulation intensities. Resting MT is commonly defined as the minimal intensity that evokes MEP_{P-P} of ~50 μV in ≥50% of trials at rest, though some studies use 100 μV [23]; investigators should state which criterion they adopted and why. Active MT is determined during a low-level voluntary contraction (often 5-10% MVC) and can reveal excitability features not visible at rest. Visual estimation of MT (based on observed twitching) overestimates intensity compared to EMG-based determination and is therefore discouraged. EMG-based methods remain the gold standard [36,37].

Modern threshold-hunting algorithms (e.g., QUEST, ML-PEST, Bayesian adaptive approaches such as MTAT 2.1 and variants) reduce determination time and provide robust estimates; authors should report the algorithm used and key parameters.

Motor Mapping

Motor mapping quantifies the spatial representation of muscles in M1 and is particularly informative when combined with neuronavigation and EMG. Use a sufficiently dense grid to cover the expected cortical representation and apply randomized stimulation across sites to minimize order effects. For each coordinate, apply multiple pulses (commonly ≥5; increase to ≥20 when mapping motor cortical excitability across different conditions to improve spatial reliability) and derive metrics such as map area (cm^2), centre of gravity (CoG), and map volume (MEP_{P-P} weighted spatial extent) [38].

Single and Paired-Pulse paradigms: Assessing Inhibition and Facilitation

Paired-pulse TMS (ppTMS) probes intracortical inhibitory and excitatory circuits by combining a conditioning stimulus (CS) and a test stimulus (TS) separated by an interstimulus interval (ISI). Well-established phenomena include short-interval intracortical inhibition (SICI; CS subthreshold, ISI 1-5 ms) mediated predominantly by GABAAergic circuits, and intracortical facilitation (ICF; ISI ~6-25 ms) associated with glutamatergic/NMDA mechanisms. The specific responses depend on CS/TS intensities, pulse waveform, and current direction – details that must be reported precisely. Paired-pulse measures are sensitive to intra- and inter-session variability, making it important to consider repeated assessments or larger samples to mitigate variability [39,40,41].

TMS-EEG: Probing Cortical Networks

Combining TMS with EEG (TMS-EEG) extends the reach of TMS beyond corticospinal readouts, allowing direct recording of TMS-evoked potentials (TEPs) as a proxy to cortical excitability and connectivity. Typical M1 TEP components are described around N15, P30, N45, P60 and N100 ms post-stimulus. TEP amplitudes and latencies are informative about excitatory/inhibitory balance and neuromodulatory states; for example, specific TEP components are sensitive to GABAergic modulation [4,35].

TMS-EEG is sensitive to numerous artefacts (electromagnetic decay, muscle and scalp artefacts, bone-conducted auditory potentials, and amplifier saturation) that require a combination of hardware and processing solutions. Best-practice preparatory steps include careful skin preparation, proper cap fitting with Cz using 10–20 landmarks, low electrode impedance (<5 kΩ), foam padding between the coil and scalp to reduce mechanical artefacts, adjustment of cable orientation to minimize electromagnetic loops, and auditory masking to reduce click-related potentials. Data-processing pipelines should transparently report the artefact-removal steps (e.g., trial rejection, independent component analysis, template subtraction) and verify that the TEP components of interest are preserved [4,33,34,35].

TMS-EEG also enables advanced closed-loop and brain-state-dependent paradigms that can guide stimulation timing for both mechanistic and therapeutic applications; however, these require rigorous control of artefacts and careful experimental design [33].

Biomechanical Perspectives on Cortical Excitability

TMS applied within a neurobiomechanical framework elucidates how cortical excitability adapts to mechanical constraints (posture, joint angle, muscle length), task demands (force, speed), and sensorimotor context. Posture and muscle length modulate spinal and supraspinal sensory inputs, altering MEP_{P-P} and latency; therefore, it is crucial to standardize joint angles, limb support, and muscle activation levels across experimental conditions [42,43].

Studies show that muscle shortening, specific limb postures, or preparatory phases of movement can increase MEP_{P-P} responses interpreted as cortical compensation for changes in mechanical efficiency or task-specific demand. During gait and other rhythmic tasks, time-locked TMS pulses reveal phase-dependent cortical contributions to locomotor control. These insights are particularly relevant for studying ageing, falls risk, and sport-specific adaptations [42,44].

Paired-pulse paradigms can be especially informative in clinical populations (e.g., post-stroke), where single-pulse responses may be reduced; facilitatory paired-pulse protocols sometimes uncover residual corticospinal pathways that are not evident with single-pulse protocols. Use of combined cortical and spinal stimulation can help dissociate cortical from spinal contributions.

Neural Adaptations in Sport

Athletes exhibit task-specific cortical reorganization that reflects the demands of their sport: larger cortical representation areas, altered motor thresholds, shorter latencies and enhanced MEP_{P-P} in muscles most relevant to trained skills. These adaptations underscore the importance of carefully matching control groups and accounting for training history when interpreting TMS findings in athletic cohorts [44,45,46].

Basic TMS protocol (recommended workflow)

Below is a pragmatic workflow for studies comparing corticospinal excitability between rest and contraction states:

1) Screening: administer a standard TMS safety questionnaire and record handedness and relevant medical history accounting for contraindications, such as epilepsy, metal implants, pacemakers, and pregnancy;

2) EMG preparation: abrase and clean skin, use conductive gel/abrasive paste, and position electrodes. For MEP recording, consider a pseudo-monopolar montage (innervation zone to inactive site) to maximize MEP_{P-P} when appropriate, or use high-density arrays when spatial resolution is required. Report electrode montage in detail [22,23,47];

3) Neuronavigation and targeting, when available, use individual MRI-based neuronavigation. If unavailable, use a cap/grid with predefined coordinates (e.g., relative to Cz/C3/C4), and systematically search for the hotspot. Document coil orientation (e.g., ~45° to midline for hand area) and tangency to the scalp [18,19,20];

4) Threshold determination: use EMG-based MT measures with an adaptive algorithm (e.g., QUEST, ML-PEST, MTAT/Bayesian) and report the method and criteria (e.g., 50 μV in ≥50% trials). Avoid visual estimation [36,37];

5) Data acquisition: randomize stimulus order when testing multiple sites/intensities, include inter-trial intervals that prevent carry-over effects, and record coil coordinates for each pulse (especially for longitudinal or multi-session designs). For motor mapping, collect multiple pulses per site (≥5-20, depending on specific goals) and use internal normalization strategies for amplitude comparisons [30,31,38].

6) Artefact control (TMS-EEG and EMG): apply foam padding between coil and scalp, orient cables, use auditory masking, monitor EOG/ECG/EMG for physiological artefacts, and document preprocessing steps for transparency [4,9,34].

Safety considerations

Follow established safety guidelines for single- and repetitive TMS protocols. Key points include auditory protection (TMS clicks can reach high sound pressure levels), thorough pre-screening, trained operators, emergency procedures, and avoidance of ferromagnetic objects near the coil. For rTMS, adhere to published safety tables and dose-limits [1,2,48].

CONCLUSION

TMS is a versatile, non-invasive method that, when combined with EMG and EEG and applied within a neurobiomechanical framework, provides unique insights into the neural control of movement. Achieving reproducible and interpretable results requires careful attention to electrode montage, coil positioning (ideally guided by neuronavigation), trial counts and normalization strategies, thresholding algorithms and rigorous artefact control for TMS-EEG. By integrating methodological rigor with biomechanical thinking, researchers can leverage TMS to advance understanding of motor control, rehabilitation and sports science.

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