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Comparison of subdermal needle and surface adhesive electrodes for intraoperative neuromonitoring during spine surgeries

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Abstract

Background This study aimed to further compare subdermal needle electrodes (NE) and surface adhesive electrodes (SE) during intraoperative neuromonitoring (IONM) in spine surgeries.

Methods We analyzed data from forty consecutive patients undergoing spine surgery. The data mainly included impedance, Root Mean Square (RMS), Tc-MEP and free-run EMG characteristics of NE versus SE with (left foot) and without (right foot) conductive paste (CP) during IONM.

Results Results indicated that SE with CP exhibited significantly lower impedance than SE without CP and higher impedance than NE. The RMS of free-run EMG recorded by SE were found to be higher than the NE, but no significant differences were found between SE with CP and SE without CP. Furthermore, NE yielded higher MEP amplitudes and superior signal-to-noise ratios (SNR) than SE. The ratios of MEP amplitude and SNR recorded by SE to NE were not significantly different between the left and right foot. The success rate for MEP induction was significantly higher with NE than SE.

Conclusion This research advocates for prioritizing NE in IONM during spine surgeries due to their favorable performance characteristics.

Keywords Intraoperative neuromonitoring, Needle electrodes, Surface adhesive electrodes, Motor evoked potentials, Electromyography, Spine surgery

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Background

The intraoperative neuromonitoring (IONM) has been considered as a valuable technique for ensuring the functional integrity of neural structures (e.g., spinal cord and nerve roots) by providing real-time feedback of neural status to the operative team during spine surgeries [1, 2]. Historically, various modalities have been employed for IONM, including somatosensory evoked potentials (SSEPs), motor evoked potentials (MEPs), and electromyography (EMG). It is widely accepted to be the best choice for monitoring neural functions and can reduce neurologic adverse postoperative outcomes in spine surgeries [3].

During IOMN, needle electrodes are the most common choice due to the low impedance characteristics [4]. However, the invasive nature of needle electrodes undoubtedly can increase the risk of skin, blood vessels and nerves injuries [5–7]. The needle electrodes also lead to postoperative discomfort and the breaks of needles sometimes happen due to the body pressure during the long operation time of spine surgeries [5, 8]. This has led to the exploration of alternative methodologies, particularly the use of non-invasive surface adhesive electrodes. Recent studies suggest that the skin surface adhesive electrodes have been largely used in recording muscle MEP for assessment of the integrity of cortico-spinal tract transmission [9–13]. Moreover, a recent study also showed the surface electrodes appeared to be a promising alternative to needle electrodes for spinal cord intraoperative monitoring, as they are non-invasive, and could record signals with equivalent abilities compared to needle electrodes [14].

Despite the advantages offered by surface electrodes, their efficacy remains a topic of debate. The study by Skinner et al. showed that surface electrodes were not suggested to be used in electromyographic and MEP monitoring of spine surgery [15]. Because sensitive neurotonic detection requires near field recording, and intramuscular electrodes are preferred. In addition, higher Tc-MEP amplitude responses were seen with longer needles compared with shorter needles which can lead to a greater distance to the muscle generator [16]. Therefore, it is still questioned whether surface electrodes could replace needle electrodes in IONM. According to our experience and the above studies, although surface electrodes have advantages such as non-invasiveness and time-saving in operation, their high impedance and weakened signal recorded on the skin, sometimes can make them unable to record MEP and have large noise.

Therefore, the aim of this study was to further compare the performance of subdermal needle electrodes and surface adhesive electrodes (with and without conductive paste) in terms of the impedance, Root Mean Square (RMS), Tc-MEP and EMG characteristics between subdermal needle and surface adhesive electrodes during IONM. By analyzing these parameters, we seek to establish whether surface electrodes can serve as a viable alternative to needle electrodes in the context of IONM for spinal surgeries, thereby providing clinicians with evidence-based recommendations for electrode selection. This ultimately will contribute to the optimization of intraoperative monitoring practices.

Methods

Patients

This is a prospective observational study. A total of forty consecutive patients who underwent spine surgeries with full IONM data were included in this study. This study was approved by the local ethical committee of the hospital. All patients participated with the informed consent agreement signed before surgeries.

Anesthesia management

At first, general anesthesia was induced with intravenous etomidate (0.3 mg/kg), sufentanil (0.3 μg/kg) and rocuronium (0.6 mg/kg). After induction, a total intravenous anesthesia (TIVA) protocol was used during surgery. Anesthesia was maintained with continuous infusion of propofol (4–6 mg/kg/h), remifentanil (0.1–0.2 μg/kg/min) and dexmedetomidine (0.4 μg/kg/h) to achieve bispectral index (BIS) ranging from 40 to 60. Sufentanil 5 μg was added intravenously every hour. Muscle relaxant, i.e., rocuronium, was only used during the induction of general anesthesia, and when opening wound necessarily demanded by operators, otherwise no muscle relaxant was added after anesthesia induction.

Stimulation parameters

Transcranial electrical stimulation was delivered using subdermal corkscrew electrodes. The C3 anode and C4 cathode pairs according to the International 10–20 electrode placement system were used for stimulation of the left hemisphere, and the reverse arrangement was used for stimulation of the right hemisphere. A train of five to nine stimuli was delivered with an individual pulse width of 50 μs at an inter-stimulus interval of 2 ms. The stimulation intensity normally started from 150 V and would be increased and determined whether a MEP wave could be clearly recognized. The transcranial electrical stimulations were applied using a commercially available IONM system (Cadwell Cascade PRO IONM system; Cadwell Industries Inc., Kennewick, USA).

Recording methods

The following two kinds of recording electrodes were used simultaneously during IOMN in the present study: (1) Twisted pair subdermal needle electrodes:12 mm length×0.4 mm width (27G) noncoated stainless steel (Ambu® Neuroline™ A/S, Ballerup, DK). (2) Disposable surface adhesive electrodes:15 × 20 mm Ag/AgCl gelled electrode, i.e., combining nanometer Ag/AgCl conductive film and conductive gel together (Xi'an Friendship Medical Electronics Co., Ltd, China).

Tc-MEPs and free-run electromyogram (EMG) were recorded using subdermal needle and surface electrodes at bilateral abductor hallucis (AH) muscles. Skin preparation was performed before placing the electrodes. The skin was cleaned and disinfected with alcohol in order to eliminate any wetness, sweat or sebum on the skin. The insertion of needle electrodes into muscles was under surface electrodes with an angle of 45 degrees, as shown in Fig. 1. The distance between recording and reference electrode was 2 cm. A tape was placed over the surface and subdermal needle electrodes to avoid detachment [14]. In addition, the left surface electrodes used conductive paste (CP), but the right surface electrodes were put on the skin without CP. The positions of all electrodes were consistent and uniform throughout the study in all patients. All patients using the two kinds of electrodes were included in analysis in this study.

Neuro monitoring characteristics

We considered MEPs with any recognizable waveform as successful recordings. In patients with successful MEP monitoring, we used complete loss of MEP waveforms (all-or-none) as a warning criterion for the significant MEP changes [17]. The MEPs were recorded at the beginning, critical parts and end of surgery. In more details, the beginning stage occurred following the successful intubation of the anesthesia; the critical stages of surgery included osteotomy, deformity correction, spinal decompression, interbody fusion cage instrumentation, etc.; the end stage occurred following the suturing of the wound. Free-run electromyogram (EMG) monitoring for nerve root monitoring during operation was enrolled in this study. The following characteristics of MEPs and free-run electromyogram (EMG) successfully recorded with both needle and surface electrodes were compared:

1. The impedance of needle and surface electrodes were measured at the beginning and the end of the operation.
2. The Root Mean Square (RMS) value has been used to quantify the EMG signal because it reflects the physiological activity in the motor unit during contraction [18]. RMS value is normally low at the state of recording resting potentials without muscle contraction, which is called assessment of baseline noise. RMS values were calculated using free-run EMG recorded for 30s at the beginning and end of operations. The MATLAB program was utilized to calculate the square of the difference between each data point and the average value, and these squared differences were summed up. Subsequently, this sum was divided by the total number of data points to yield the average squared value. Lastly, the square root of this average squared value was taken, resulting in the RMS value for each set of data.
3. Amplitude of MEP was defined as the largest peak-to-peak deflection of the muscle response (from the tallest peak to the lowest trough).
4. Signal-to-noise ratios (SNR) of MEP were calculated as follows: each MEP amplitude was divided by the largest noise amplitude. The noise amplitude was calculated from the last part of the MEP waveform [14].
5. To calculate the SNR of free-run EMG action potentials, we first computed the root mean square (RMS) value of the baseline data using MATLAB. This RMS value serves as the noise reference. Subsequently, we calculate the RMS value of the signal using the same method, which signifies the effective value or amplitude of the signal. Finally, we determine the SNR using the formula $SNR = 20 * \log10(RMS_{signal} / RMS_{noise})$.
6. The time using transcranial electrical stimulation beginning to induce a recognizable MEP waveform

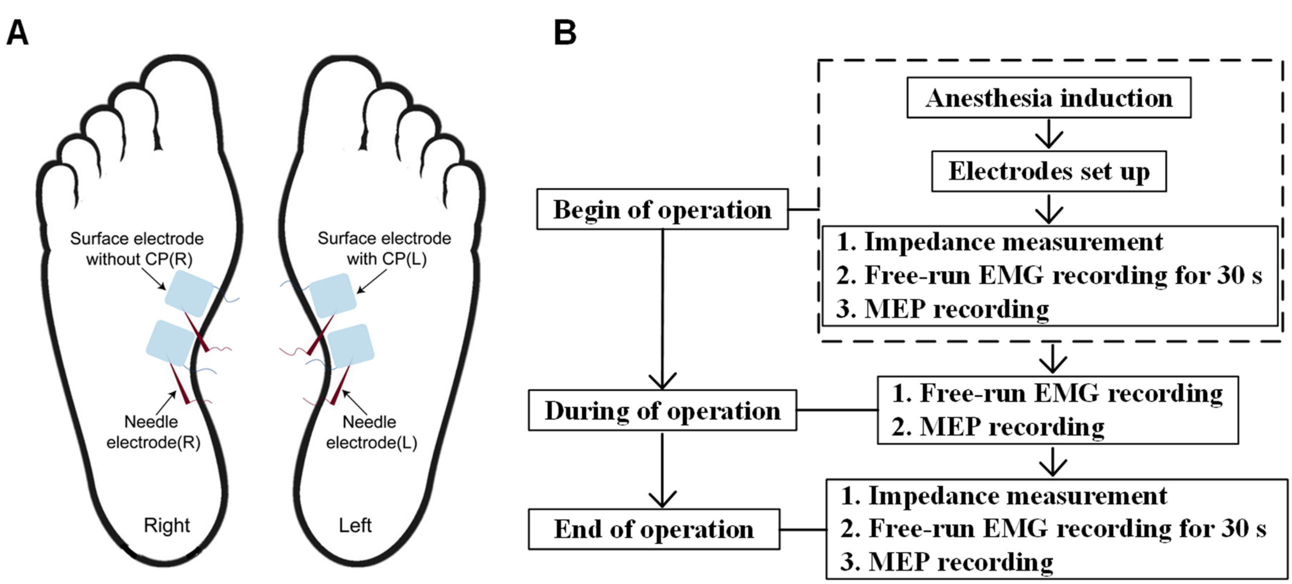

Fig. 1 Experiment setup and time-line of operation procedure. (**A**) Positions of subdermal needle electrodes (NE) and surface adhesive electrodes (SE) with (left foot) and without (right foot) conductive paste (CP). (**B**) Time-line and measured parameters of the whole operation

from beginning to the end of operations were recorded at all electrodes on both feet.

Statistical analysis

The MEP parameters, including latency, amplitude, SNR, and impedances were compared between surface and subdermal needle recordings. The impedance of needle electrodes with 0 kΩ was set to 0.01 kΩ. The Shapiro-Wilk test was conducted to assess the normality of distributions. The data of impedance and RMS were not normally distributed even though they were $\log_{10}$-transformed, therefore, they were analyzed using nonparametric Kruskal-Wallis test. The pairwise comparisons were done between beginning and end of operation and among all electrodes. Significance values were adjusted by the Bonferroni correction for multiple tests. Nonparametric Spearman correlation analysis was done between impedance and RMS. MEP amplitudes and SNR data were compared in each side between needle and surface electrode. Their data were normally distributed after $\log_{10}$-transformed. Compare the ratio of MEP amplitude of surface electrode to amplitude of needle electrode between two sides. They were analyzed using paired t-test using $\log_{10}$-transformed data to obtain normal distribution. $P < 0.05$ was considered to be significant.

Results

Patients

Forty consecutive patients including 23 women with full IONM data who met the requirements of the experimental design were included in this study (Table 1). The ages of patients were from 10 to 76 years (median age at 58 years). Body Mass Index (BMI) of all patients were from 15.24 to 33.98 (median value at 24.37). There were various kinds of spinal diseases and operation methods in this study. Twenty cervical spondylosis patients with fifteen patients underwent cervical anterior operation and five patients underwent posterior operation. Four patients included two thoracic spinal stenosis, one thoracic fracture, one spinal tumor, underwent thoracic posterior operation. Five patents with lumbar disc herniation underwent three spinal endoscopy and two lumbar posterior operation. Eleven patients underwent posterior spinal deformity correction surgery included two congenital scoliosis, five adolescent idiopathic scoliosis, two degenerative spinal deformities and two spinal kyphosis deformity patients after spinal fracture. The diseases and operation types were shown in Table 1.

Impedance and RMS

At the beginning of operation, the impedances of needle electrodes were 1.95 kΩ (median) with a range from 0.01 to 8.3 kΩ (left) and 1.95 KΩ (median) with a range from 0.01 to 8.4 kΩ (right); the impedances of surface adhesive electrodes were 36.15 kΩ (3.9–86 kΩ, min-max) (left, with CP) and 95.6 kΩ (15.7–100 kΩ, min-max) (right, without CP). The surface electrodes without CP had the highest impedance ($p < 0.05$). The impedances of surface electrodes with CP were higher than needle electrodes ($p < 0.05$). The needle electrodes had the lowest impedance ($p < 0.05$) (Fig. 2).

Table 1 Patients' characteristics

Characteristic	Value	
Total number of patients	40	
Gender (n)		
Man	17	
Woman	23	
Age (years)		
Median	58	
Range	10–76	
BMI		
Median	24.37	
Range	15.24–33.98	
Disease	Value	Operation types
Cervical spondylosis	20	Cervical anterior (15), Cervical posterior (5)
Thoracic spinal stenosis	2	Thoracic posterior (4)
Thoracic fracture	1	
Spinal tumors	1	
Lumbar disc herniation	5	Spinal endoscopy (3), Lumbar posterior (2)
Congenital scoliosis	2	Posterior spinal deformity correction surgery ($n = 11$)
Adolescent idiopathic scoliosis	5	
Degenerative spinal deformities	2	
Spinal kyphosis deformity after spinal fracture	2	

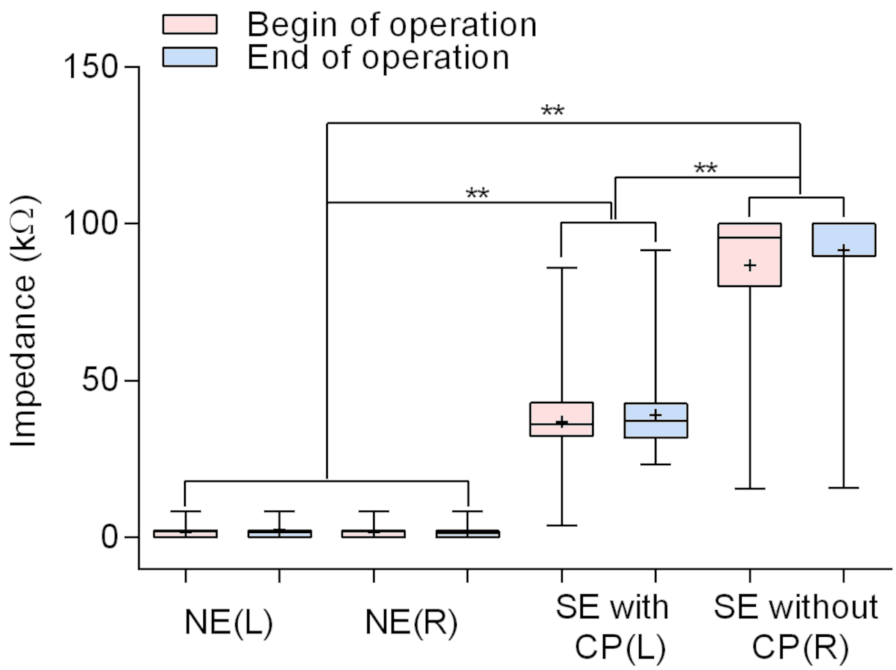

Fig. 2 Impedance of four electrodes positions during the operation time. SE with CP exhibited significantly lower impedance than SE without CP and higher impedance than NE. NE had the lowest impedance. Impedance of all electrodes did not change from the beginning to the end of operations. NE(L): Needle electrodes (Left foot); NE(R): Needle electrodes (Right foot); SE with CP (L): Surface electrodes with conductive paste (Left foot); SE without CP (R): Surface electrodes without conductive paste (Right foot)

At the end of operation, the impedances of needle electrodes were 1.65 kΩ (median, 0.01–8.5 kΩ, min-max) (left) and 1.4 kΩ (median, 0.01–8.5 kΩ, min-max) (right); the impedances of surface adhesive electrodes were 37.15 kΩ (median, 23.4–91.6 kΩ, min-max) (left, with CP) and 89.78 kΩ (median, 15.9-89.78 kΩ, min-max) (right, without CP). The surface electrodes without CP still had the highest impedance, whereas the needle electrodes had the lowest ($p < 0.05$). This indicated that conductive paste could reduce the impedance effectively when using surface electrode. The impedance of all kinds of electrodes did not change from the beginning to the end operations (Fig. 2).

At the beginning of operation, the RMS of needle electrodes were 0.55 mV(median, 0.53–1.43 mV, min-max) (left) and 0.54 mV(median, 0.50–1.23 mV, min-max) (right); the RMS of surface adhesive electrodes were 1.57 mV(median, 0.61–241.4 mV, min-max) (left, with CP) and 1.63 mV(median, 0.71–452.3 mV, min-max) (right, without CP). The RMS of surface electrodes were higher than the needle electrodes, but no significant differences were found between left (with CP) and right (without CP) surface electrodes. (Fig. 3).

At the end of operation, the RMS of needle electrodes were 0.56 mV(median, 0.54–8.06 mV, min-max) (left) and 0.55 mV(median, 0.53–2.48 mV, min-max) (right); the RMS of surface adhesive electrodes were 1.93 mV(median, 0.61–303 mV, min-max) (left, with CP) and 3.51 mV(median, 0.73–860.3 mV, min-max) (right, without CP). The RMS of surface electrodes were also found to be higher than the needle electrodes, but no significant differences were found between left (with CP) and right (without CP) surface electrodes. (Fig. 3). This indicated that conductive paste could reduce the RMS to some extent, but there was no significance when using surface electrode. The RMS of all kinds of electrodes were not found to have significant changes from the beginning to the end of operations (Fig. 3).

The impedance of all electrodes was found to be positively correlated with RMS values ($r = 0.73$, $p < 0.01$) (Fig. 4).

MEP and EMG parameters

The visible and clear MEPs were successfully induced using four electrodes in 30 (out of 40) patients. The clear MEPs recorded at the same time during operations were chosen for analysis. Amplitudes were $\log_{10}$-transformed to obtain normal distribution. Statistical analysis used $\log_{10}$-transformed data. The MEP amplitude of needle electrode (159.48 ± 182.64 mV, mean ± SD) was higher than the amplitude of surface electrode with CP (30.75 ± 29.85 mV, mean ± SD) at the left foot (t = 7.152,

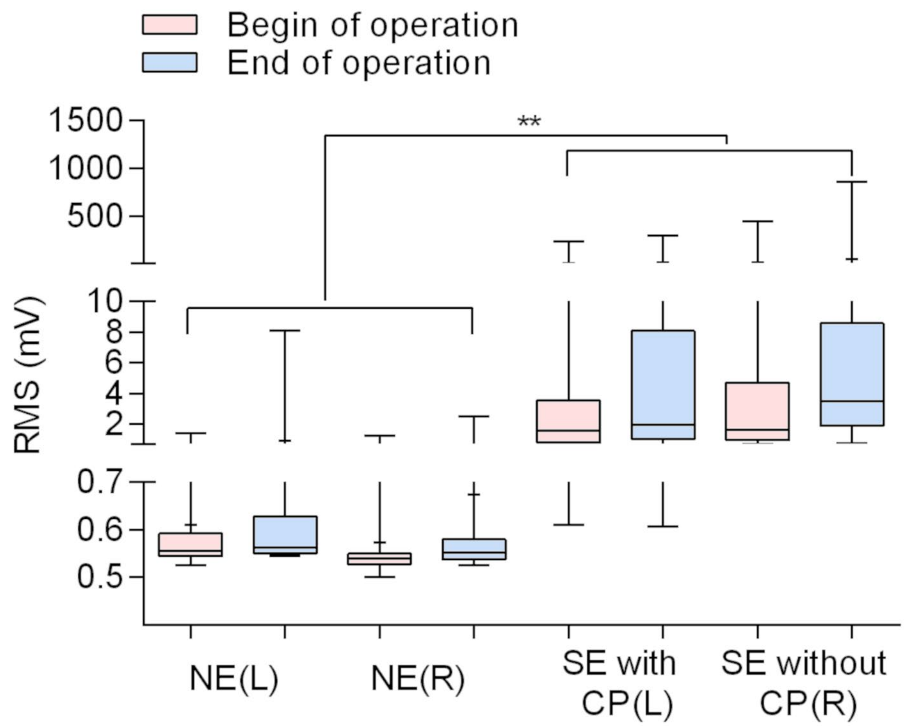

Fig. 3 RMS of free-run EMG at resting state of four electrodes positions during the operation time. The RMS of free-run EMG recorded by SE were found to be higher than the NE, but no significant differences were found between SE with CP and SE without CP. RMS of all electrodes did not change from the beginning to the end of operations. NE(L): Needle electrodes (Left foot); NE(R): Needle electrodes (Right foot); SE with CP (L): Surface electrodes with conductive paste (Left foot); SE without CP (R): Surface electrodes without conductive paste (Right foot)

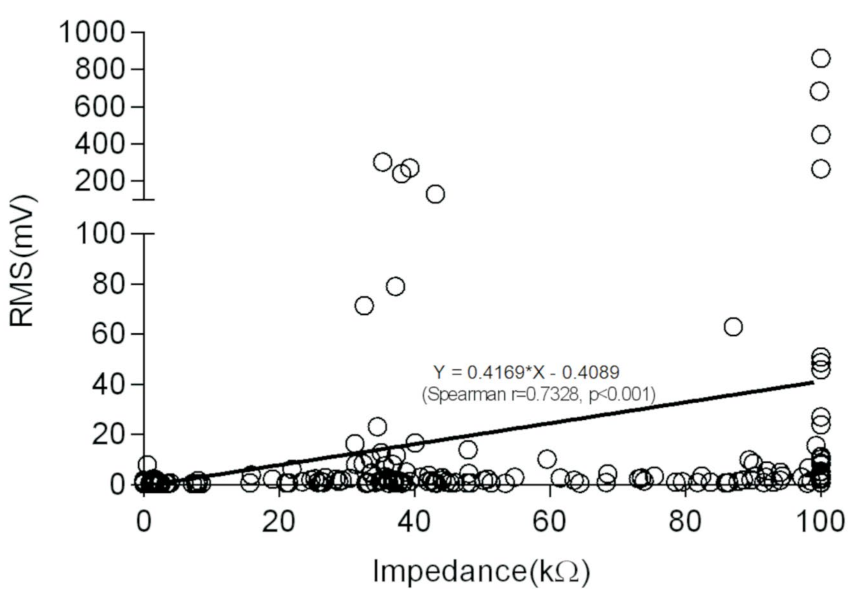

Fig. 4 Correlation between impedances of all electrodes and RMS of resting-state of free-run EMG. The RMS of free-run EMG at resting state (i.e., baseline noise) was found to be positively correlated with impedances of all electrodes

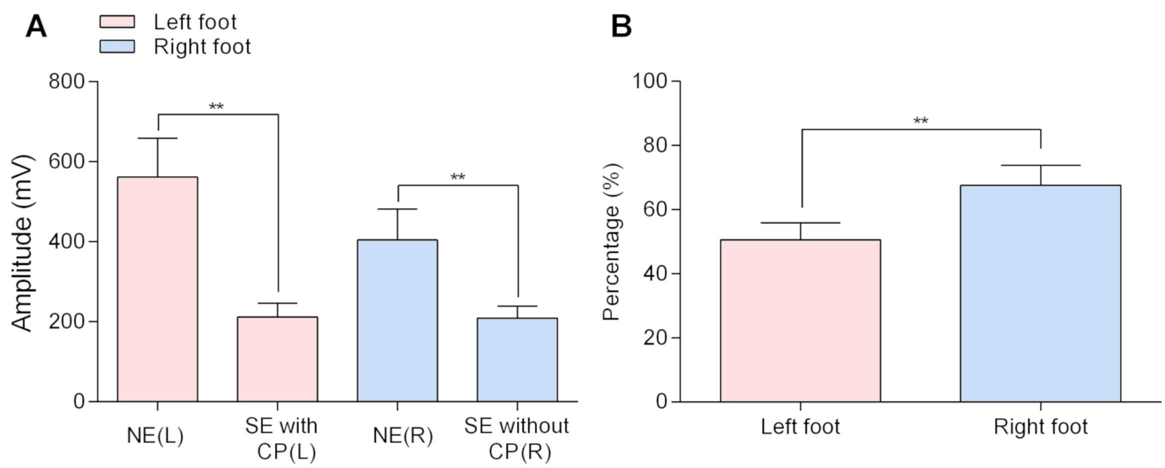

Fig. 5 MEP amplitudes of all electrodes. The MEP amplitudes of needle electrodes were higher than the amplitudes of surface electrodes with and without CP on both feet. The ratio (i.e., amplitude(SE)/amplitude(NE)) of the right foot was higher than that of the left foot

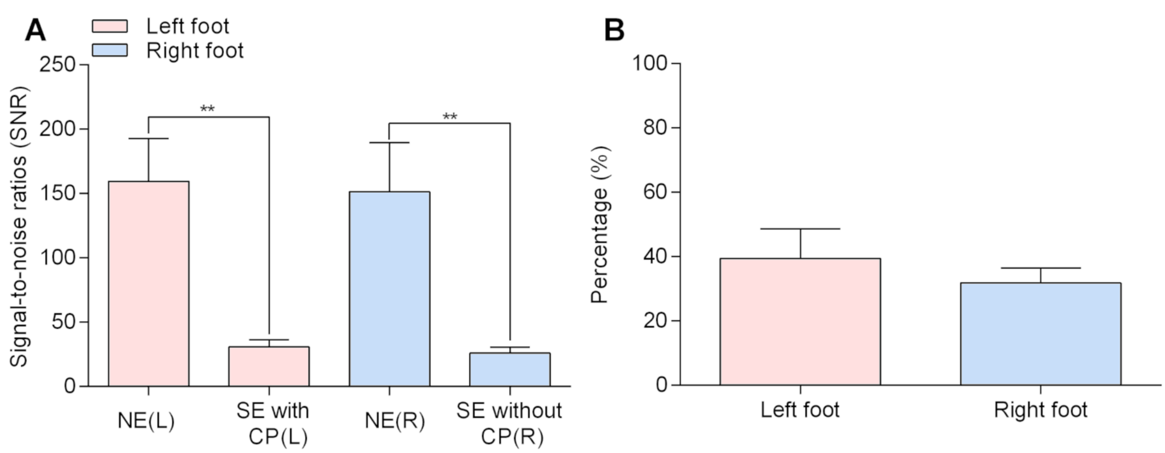

Fig. 6 SNR of all electrodes. SNR of needle electrodes were higher than SNR of surface electrodes with and without CP on both feet. The ratio (i.e., SNR(SE)/SNR(NE)) of both feet was not found to be significantly different from each other. SNR: Signal-to-noise ratio

$p < 0.01$). The MEP amplitude of needle electrode (151.32 ± 208.94 mV, mean ± SD) was also higher than the amplitude of surface electrode without CP (25.83 ± 25.95 mV, mean ± SD) at the right foot (t = 7.357, $p < 0.01$). This indicated that needle electrode could induce a higher MEP amplitude compared to surface electrode whether using CP or not (Fig. 5A). The MEP amplitude was not found to be significantly correlated with BMI at all electrodes (Pearson correlation, $p > 0.05$) (Supplementary Table 1).

The ratio of MEP amplitude of surface electrode with CP to amplitude of needle electrode in the left foot is 50.68 ± 28.84%. The ratio of MEP amplitude of surface electrode without CP to amplitude of needle electrode in the right foot is 67.65 ± 33.59%, which is higher than the left foot (t = 2.696, $p < 0.05$) (Fig. 5B). This indicated that using conductive paste did not increase MEP amplitude when using surface electrode.

The Signal-to-noise ratios (SNR) of MEPs of four electrodes were $\log_{10}$-transformed to obtain normal distribution. Statistical analysis used $\log_{10}$-transformed data. The SNR of the needle electrode (159.49 ± 182.64, mean ± SD) is higher than the surface electrode with CP (30.75 ± 29.85, mean ± SD) in the left foot (t = 7.152, $p < 0.01$). The SNR of the needle electrode (151.32 ± 208.94, mean ± SD) is higher than the surface electrode without CP (25.83 ± 25.95, mean ± SD) in the right foot (t = 7.357, $p < 0.01$) (Fig. 6A). The SNR was not

Table 2 Comparison of time inducing visible MEP using needle and surface electrodes throughout the operations

Time inducing MEP	Left foot	Right foot	Number	Percentage
	Needle electrode; Surface electrode with CP	Needle electrode; Surface electrode without CP		
Both at the same time	14	15	29	36.25%
Needle electrode first	26 (9 without inducing visible MEP using surface electrode)	25 (9 without inducing visible MEP using surface electrode)	51	63.75%
Surface electrode first	0	0	0	0

Left foot Vs. Right foot: Pearson Chi-Square test: $p = 0.816$

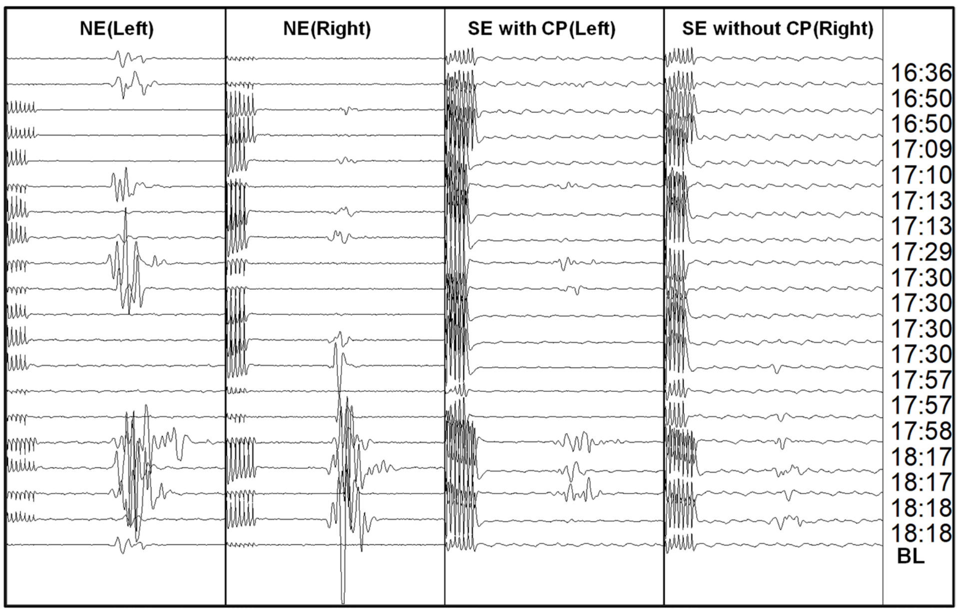

Fig. 7 MEP induction and recoding during the whole operation period in one patient. The visible and identical MEP waveforms appeared later using surface electrodes than needle electrodes

found to be significantly correlated with BMI at all electrodes (Pearson correlation, $p > 0.05$) (Supplementary Table 1).

The ratio of MEP SNR of surface electrode with CP to SNR of needle electrode in the left foot is 39.41 ± 50.20% (mean ± SD). The ratio of MEP SNR of surface electrode without CP to SNR of needle electrode in the right foot is 31.76 ± 25.59% (mean ± SD). The data of ratios were $\log_{10}$ transformed to obtain normal distribution. After statistical analysis, the ratios of the left foot were not found to be significantly different from the right foot. This indicated that using conductive paste did not increase SNR of MEP when using surface electrode (Fig. 6B).

Needle electrodes successfully induced visible and clear MEP waveforms on both feet in all 40 patients. On the left foot, there were 26 times that needle electrode induced visible and clear MEP earlier than surface electrode with CP. The 26 times included 9 times that MEP wave was not induced by surface electrode with CP from the beginning to the end of operations. There were 14 times that needle electrode and surface electrode began to induce visible and clear MEP wave at the same time. On the right foot, there were 25 times that needle electrode induced visible and clear MEP earlier than surface electrode with CP. The 25 times included 9 times that MEP wave was not induced by surface electrode without CP from the beginning to the end of operations. There were 15 times that needle electrode and surface electrode began to induce visible and clear MEP wave at the same time (Table 2). No significant difference was found between the MEP induction conditions between left and right foot (Pearson Chi-Square test: $p = 0.816$) (Table 2). This indicated that success rate inducing MEP using surface electrode was lower than needle electrode. Surface electrodes had higher RMS and could be susceptible to external interference. This resulted in large noises in when using surface electrodes, which could cover up MEP waveforms (Fig. 7 and Supplementary Fig. 1). Using conductive paste did not increase success rate when using surface electrode.

Three patients with spinal endoscopy surgeries recorded action potentials using free-run EMG caused by nerve root stimulations. SNR of EMG signals recorded by needle electrode were higher than surface electrodes (Supplementary Fig. 2).

Discussion

In this study, we compared the performance of subdermal needle electrodes and surface adhesive electrodes in the context of IONM during spinal surgeries. Our findings reveal that needle electrodes yield significantly higher MEP amplitudes and improved signal-to-noise ratios (SNR) compared to surface electrodes, suggesting their superiority in presenting signal quality. Furthermore, under the same transcranial electrical stimulation intensities, the higher success rate of MEP induction associated with needle electrodes reinforces their role in providing real-time feedback to the surgical team, which is essential for timely interventions that can prevent irreversible neural damage. This study not only contributes to the ongoing discourse regarding electrode selection for IONM but also emphasizes the importance of optimizing monitoring techniques to ensure patient safety and effective surgical outcomes.

One important thing to assess the functional stability of the electrode is to see whether the properties of the electrode itself are stable. Impedance and Root Mean Square (RMS) values serve as critical indicators of signal quality in electrode material evaluation, providing essential guidance for optimal electrode selection. Consistency in impedance of electrode is critical for the reliability of EMG measurements. The stability in impedance over time and the balance in impedance between electrode sites have a considerable effect on the signal to noise ratio (SNR) of the measured EMG signal, both in terms of noise levels and spatial resolution [19, 20]. In this study, the impedance and RMS of needle electrode and surface electrode were tested at the beginning and end of the operation to see if the nature of the electrode recording signal could be stable. We found that the impedance and RMS of the needle electrode and the surface electrode (with or without conductive paste) did not change significantly from the beginning to the end of the operation, proving that the nature of the types of electrodes applied in this study was very stable throughout the procedure. The impedance of the surface electrodes (with or without conductive paste) were significantly higher than that of the needle electrode, which is consistent with the previous study [21]. This is because myoelectric signals must pass through various connective tissues, subcutaneous layers and skin and reflect on surface electrodes. The skin has higher impedance given the strong relationship between tissue impedance and water [22]. Several biological factors can affect skin impedance, such as thickness of dead skin layer, subcutaneous fat, moisture, oil content and hair presence. These factors undoubtedly can increase the impedance of the surface electrode [21]. Therefore, previous study has pointed out that skin preparation or the use of conductive gels can reduce the impedance of surface electrodes, and in this study, conductive paste significantly indeed reduced the impedance of surface electrodes [23].

Studies have shown that the root mean square (RMS) of the EMG signal is a reliable parameter and displayed a positive correlation with muscle force and muscle tension, which represents muscle motor units [24, 25]. In this study, all patients were under general anesthesia during operations, and the muscles were completely relaxed without any contraction, so the RMS was used to evaluate the EMG signal at resting state (i.e., baseline noise) recorded by different types of electrodes. In this study, the RMS of the surface electrodes (with or without CP) were significantly higher than that of the needle electrode, suggesting that the resting-state EMG recorded by surface electrodes could be mixed with more noise or external interference [26]. The operating room normally has an extremely complex environment with a variety of electrical equipment, and the electrodes are susceptible to interference from the surrounding environment. The unwanted noise can be the ambient noise which is generated by electromagnetic radiation such as computers, ventilators, power lines, high frequency electrotome, ultrasonic bone cutter, the earth, etc [27]. Another important noise is transducer noise which is generated at the electrode-skin junction when using surface electrodes. D/C (Direct Current) voltage potential and A/C (Alternating Current) voltage potential are the main two noise sources, mainly caused by the impedance effect related with skin, conductive transducer and electrode sensor [20, 27]. These factors could undoubtedly increase the noise of the surface electrodes and be subject to external interference. In the present study, conductive paste could not only significantly reduce the impedance, but also reduced the RMS of the surface electrode to a certain extent, which is consistent with previous research findings [28]. We further found that there is a positive correlation between impedance and RMS, suggesting that reducing impedance might reduce the noise of EMG to a certain extent and improve the quality of EMG at resting state. Conductive paste may help decrease the impedance and noise when using adhesive gelled surface electrode [28].

The goal of choosing a proper type of electrode used in IONM is to maximize the amplitude of the signal while minimizing the noise. In this study, we found that the amplitude of MEPs recorded by needle electrodes were significantly higher than those of surface electrodes (with or without CP), which is consistent with previous research [14]. In addition, even though CP reduced the impedance of surface electrodes, it decreased the ratio of the MEP amplitude of surface electrode relative to needle electrode. One possible reason could be that the layer of conductive paste could increase the distance the distance between the MU and the detection point on the skin, which inversely decrease the amplitude of sEMG signal contributed by a motor unit [29]. Another reason is that the conductive paste itself has a certain impedance, causing some signal energy to be consumed within the layer of conductive paste, thereby resulting in a decrease in signal amplitude [30]. Additionally, poor conductivity or uneven distribution of the conductive paste may also lead to further signal attenuation [31]. This dual-effect phenomenon between conductivity enhancement and signal amplitude degradation when using CP was systematically documented for the first time, providing critical insights for optimizing electrodes selection and operational protocols in clinical electrophysiological monitoring.

Signal-to-noise ratios (SNR) of MEPs recorded by needle electrodes were higher than using surface electrodes (with or without CP), which is consistent with a previous study [14]. In addition, after the application of conductive paste, the proportion of SNR of surface electrode relative to needle electrode was not significantly increased. This indicated that the conductive paste did not significantly further improve the signal quality of MEP recorded by adhesive gelled surface electrodes. Therefore, could be proposed that the conductive material may significantly improve the signal quality of the dry surface electrodes, but it has little effect on the wet surface electrode with an adherent gel layer [32]. For wet surface electrodes, they inherently feature a relatively thick adhesive layer. The application of conductive paste to these electrodes results in only a slight reduction in the RMS of the baseline but unfortunately also led to a decrease in MEP amplitudes. Given the formula for calculating the signal-to-noise ratio (SNR), it is clear that this does not contribute positively to enhancing the SNR. Conversely, for dry electrodes, the addition of a very thin layer of conductive paste has the potential to improve the SNR. In our study, simply reducing the impedance for surface adhesive gelled electrodes using conductive paste (CP) could not significantly improve the EMG signal quality. Noise can be reduced to some extent by using CP, but it may affect the amplitude of the signal. In the present study, no correlation was found between BMI and MEP amplitudes and SNR. In our study, abductor hallucis (AH) muscles of feet were used for neuromonitoring. Fat with low water content had high resistivities [33]. It has been shown that nowadays a proportionally greater amount of fat was located in the abdomen or trunk compared with the lower extremities in more population of central obesity [34]. Therefore, foot fat in patients with different BMIs had little effect on the induction of MEPs.

In this study, under the same transcranial electrical stimulation intensities, surface electrodes had a higher probability of failing to elicit MEPs compared to needle electrodes. This was similar to the results of a previous animal experiment [26]. Moreover, the time when surface electrodes began to be able to record identifiable MEP waveforms sometimes was later than needle electrodes. These findings highlight the advantage of needle electrodes for intraoperative neuromonitoring. The earlier MEPs are detected during surgery, the more beneficial it is for monitoring surgical safety. However, a previous study showed that surface and needle electrodes induced MEP had similar elicitability [14]. The electrodes recording position in this study was the abductor hallucis (AH) muscle, which is different from the tibialis anterior (TA) muscle recorded in the previous study. TA muscle is relatively superficial whereas abductor hallucis (AH) muscle is deeper. Superficial muscles may make it easier to record EMG signals with surface electrodes [14]. In addition, intraoperative neuromonitoring usually includes MEP and free-EMG monitoring. The surface electrodes used in this study were often mixed with a large amount of noise and were easily disturbed by the external environment, which made it difficult to identify the waveforms of evoked EMG signal. This increased the difficulties of neuromonitoring during operations. Intraoperative neurophysiological monitoring has very high requirements for real-time performance. Once the electrodes selected cannot induce electromyography evoked potentials that can be visible, then it will be difficult to deal with, because the patient has completed the disinfection and clothing. Then it will be difficult to replace the electrodes in time, which undoubtedly loses the value of intraoperative neuromonitoring. Furthermore, the signal quality of surface electrodes remains vulnerable to unpredictable conditions of the skin. The ability to detect changes in neural function more accurately and timely with needle electrodes could lead to effective interventions during surgery, enhancing patient safety and potentially improving postoperative outcomes.

Despite the valuable insights gained from this study, several limitations must be acknowledged. The relatively small sample size and the lack of multicenter data may limit the generalizability of our findings. For example, there are differences in the operating habits of working staff in different institutions. In a study, the optimal MEP stimulation site in the scalp was determined preoperatively which could be directly used during the operation [23]. This may improve the success rate of MEP induction. Another is the choice of the brand of surface electrodes, which may lead to some differences in the quality of the electrodes. Future research should aim to include multicenter with a broader patient population. In addition, there are various types of surface electrodes, and some electrodes even have signal amplification functions, which may greatly increase the signal amplitude recorded by the surface electrodes and improve the success rate of induced visible evoked potential waveforms [32, 35]. Therefore, other types of surface electrodes may have different results compared to our study. Furthermore, the surface electrode of the left foot with conductive paste and the surface electrode of the right foot without conductive paste were fixed in our study. If there is an exchange group with no conductive paste on the left foot and with conductive paste on the right foot, the results would be better explained, however, we considered that it had nothing to do with which side of the conductive paste was placed. Future studies could use more rigorous experimental designs to substantiate the clinical implications of electrode performance in intraoperative neuromonitoring.

Conclusions

In conclusion, our findings underscore the superiority of needle electrodes in recording muscle MEPs compared to surface adhesive gelled electrodes during intraoperative neuromonitoring in spinal surgeries. The significant differences in MEP amplitudes, impedance, SNR, and success rate of MEP induction highlight the importance of electrode selection in optimizing surgical outcomes, thus providing new insights that may influence future practices in neuromonitoring for spinal surgeries. The choice of surface electrodes in IONM still needs to be cautious. Noise can be reduced to some extent by using CP, but it may affect the amplitude of the signal. These results advocate for the continued use of needle electrodes in clinical practice, while also suggesting avenues for future research to explore innovative surface electrode technologies and methodologies that could further enhance the neuromonitoring efficacy and patient safety.

List of abbreviations

IONM	Intraoperative neuromonitoring
MEP	Motor evoked potentials
CP	Conductive paste
SNR	Signal-to-noise ratios
AH	Abductor hallucis
TA	Tibialis anterior
RMS	Root Mean Square
EMG	Electromyography
NE	Needle electrode
SE	Surface electrode

Supplementary Information

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Supplementary Material 1

Supplementary Material 2

Supplementary Material 3

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Author contributions

WWX, DKS, HF, KFW and YZ conceived the ideas for this research and provided overall guidance. WWX contributed to the data collection. TL gave supervision for EMG analysis. WWX, DKS and HF contributed to analysis of data. WWX, DKS and HF prepared the first draft. KFW and YZ revised and commented on the manuscript.

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Data availability

No datasets were generated or analysed during the current study.

Declarations

Ethics approval and consent to participate

This study was approved by the local ethical committee of the hospital (2022PHB431). All patients participated with the informed consent agreement signed before surgeries.

Consent for publication

Not applicable.

Competing interests

The authors declare no competing interests.

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