J. Phys. Ther. Sci. 33: 63–68, 2021

The Journal of Physical Therapy Science

Original Article

Fundamental research on surface electromyography analysis using discrete wavelet transform—an analysis of the central nervous system factors affecting muscle strength

Kazunori Morozumi, RPT, PhD[1)]*, Hironori Ohsugi, RPT, PhD[1)],
Katsuyuki Morishita, RPT, PhD[1)], Yuka Yokoi, RPT, PhD[1)]

[1)] *Department of Physical Therapy, Faculty of Social Work Studies, Josai International University: 1 Gumyo, Togane, Chiba 283-8555, Japan*

Abstract. [Purpose] We aimed to investigate the central nervous system factors that affect muscle strength based on the differences in load and time using the discrete wavelet transform, which is capable of a time-frequency-potential analysis. [Participants and Methods] Surface electromyography (EMG) of the right upper bicep muscle in 16 healthy adult males were measured at 10% MVC (maximum voluntary isometric contraction), 30%, 50%, 70%, and 80% to 100% MVC. We used a discrete wavelet transform for the electromyographic analysis and calculated the median instantaneous frequency spectrum (MDF) and frequency band component content rate (FCR) at 1-ms intervals as well as their spectrum integrated values (I-EMG). [Results] MDF and FCR tended to be high throughout the measurements. Specifically, the high-frequency band component content rate was high at the time of low muscle strength; fast-twitch muscle fibers may be involved during these muscle contractions. We found significant changes in the I-EMG as the muscle strength increased from 10% MVC to 100% MVC. [Conclusion] Analyzing the surface electromyograph using discrete wavelet transform enabled us to assess the central nervous system factors that increase in the EMG amplitude integrated values and change in the median instantaneous frequency spectrum and in the frequency band component content rate.

Key words: Surface electromyography, Central nervous system factors, Discrete wavelet transform

(This article was submitted Aug. 19, 2020, and was accepted Oct. 25, 2020)

INTRODUCTION

Muscle strength can be improved as a motor function, and the inherent factors that influence muscle strength values may include central nervous system factors[1–4)] and muscle hypertrophy factors[5, 6)]. Changes in muscle strength values related to strengthening exercises are associated with central nervous system factors[1–4)] in the early stages for approximately 2 weeks from exercise commencement, whereas from weeks 6 to 12, they are affected by muscle hypertrophy factors[5, 6)].

Central nervous system factors that influence muscle strength include the motor units involved in muscle contraction, total number of muscle fibers controlled by them (recruitment), changes in the rate of firing of alpha (α) motor neurons (rate coding)[1–4)], and changes in the type of muscle fibers (size principle)[7)]. Interactions between these factors can change the strength of voluntary muscle contraction.

Central nervous system factors are affected by the excitation level of the spinal cord and recurrent inhibition of the Renshaw cells[8)], which influence muscle contraction[9, 10)] and can change the maximum voluntary contraction (MVC) value. Thus, observing the changes in the central nervous system factors may enable us to assess motor function reserves which may

---

*Corresponding author. Kazunori Morozumi (E-mail: moroz@nifty.com)

be useful indicators of exercise goals during muscle strengthening exercises, etc.

The conventional methods for assessing the central nervous system factors during voluntary contraction include invasive methods such as generating muscle twitches using an electric stimulator and using needle electromyography (EMG), as well as eliciting the Hoffmann reflex11) which mainly assesses the spinal cord excitation level, among others. Among these methods, there are only a few in the field that can be conducted in a simple manner. As a result, many researchers have been working on devising non-invasive methods such as surface EMG analysis12–14).

The fast Fourier transform and maximum entropy methods have been used in various analyses to measure the surface EMG during MVC and to assess the central nervous system factors. Obtaining accurate measurements during the analyses could be difficult due to the overlap of the EMG waveforms during MVC12–14). Therefore, the decomposition analysis method, which measures the surface EMG using multipoint electrodes and analyzes the total number and frequency of EMG waveforms arising from the time difference between the electrodes3, 12, 13), is being investigated. However, this method also has some problems such as the appearance of irregular EMG waveforms, and as such, there are few methods of analysis that have been established as the gold standard.

In this study, we conducted an analysis of the surface EMG using wavelet transform15, 16) as a method to assess the central nervous system factors that may be related to muscle strength. We observed the factors that influenced the differences in load during muscle contraction and the changes in muscle strength over time. From these results, we investigated the possibility of assessing the central nervous system factors during MVC.

PARTICIPANTS AND METHODS

The inclusion criteria were males between the ages of 18 and 39 with no history of trauma or illness that has left movement sequelae in the limbs. This helped to minimize any errors caused by differences in gender and age distribution. The participants were 16 healthy adult males (mean age 19.8 [19–20] years, mean height 170.6 ± 6.0 cm, and mean weight 62 ± 8.1 kg).

A BIODEX 4 (Sakai Medical, KK, Japan) was used to measure the maximum voluntary isometric contraction of the right elbow joint during flexion (bicep). The measurement positions were as follows: shoulder 45° flexion, 30° abduction, inner and outer rotation in the intermediate position in the BIODEX chair in a sitting position (back angle 85°) as well as 90° flexion of the elbow, 90° supination of the forearm, intermediate flexion and extension of the wrist, extension of the metacarpophalangeal (MP) joint, and maximum flexion of the proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints of the fingers. The measurement position of the BIODEX attachment was the palm side of the line connecting the wrist joint at the radius and the styloid process at the end of the ulna. The muscle strength was measured for 5 seconds 3 times when exhibiting MVC. The maximum value obtained among them was taken as the MVC, and the 10% MVC, 30% MVC, 50% MVC, 70% MVC, and 80% MVC values of the biceps were calculated for each participant. These were used as target values for exerted muscle strength, and the target line was displayed on the BIODEX monitor screen (Fig. 1).

The surface EMG of the short head of the right bicep was measured. The measurement site was based on the line connecting the midpoint of the anterior axillary line and the midpoint of the cubital fossa cross striation. The midpoint of the reference line was marked, and a line was drawn perpendicular to the reference line from the mark to the inner edge of the upper arm. Around the midpoint of this line, points with a distance of 1 cm between the electrodes in the craniocaudal position were taken as the EMG measurement sites. At the site of measurement, the skin impedance was pretreated to 5 kΩ or less and the electrodes were attached. The EMG amplifier (BA1104, Nihon Santeku, KK, Japan) used for the measurement utilized a high-pass filter of 1 kHz, time constant of 0.03 seconds, and common mode rejection ratio (CMRR) of 90 dB or more. First, the surface EMG during MVC was measured, and the EMG measurement sequence during 10% MVC, 30% MVC, 50% MVC, 70% MVC, and 80% MVC was randomized using a random number table, and subsequently 3 sets of measurements were performed for 5 seconds each. The measurements were taken with a 3-minute break between the MVC and each % MVC measurement and a 20-second break between each measurement that lasted for 5 seconds and was performed 3 times as described earlier.

The EMG data were recorded at a sampling frequency of 2 kHz using the biosignal recording program VitalRecordar2 (Kissei Comtec, KK, Japan).

The EMG analysis in this study involved a three-dimensional analysis in the time-frequency-potential domain, and the wavelet function of the Daubechies N=10 discrete wavelet transform, which is highly localized in the frequency domain and has excellent frequency characteristics, was applied. The waveform obtained by the discrete wavelet transform was classified into the median power frequency of the instantaneous frequency spectrum every 1 ms and the frequency band component content rate (low-frequency band: less than 45 Hz, medium-frequency band: 45–95 Hz, high-frequency band: 95 Hz or more)15), the spectrum integrated value was calculated, and the three sets of data were averaged.

To compare the data obtained for each condition of measurement, one-way analysis of variance (ANOVA) and multiple comparison (Tukey HSD) were used for the statistical processing (SPSS for Windows Ver. 24) at a significance level of 5%.

We explained the purpose and contents of the study, benefits and risks, protection of personal data, refusal, and withdrawal of participation, etc., to all the participants, and written consent was obtained from each participant by hand. Moreover, this study was submitted to Josai International University's Institutional Review Board for assessment and was granted approval (Approval No.: 04W170030).

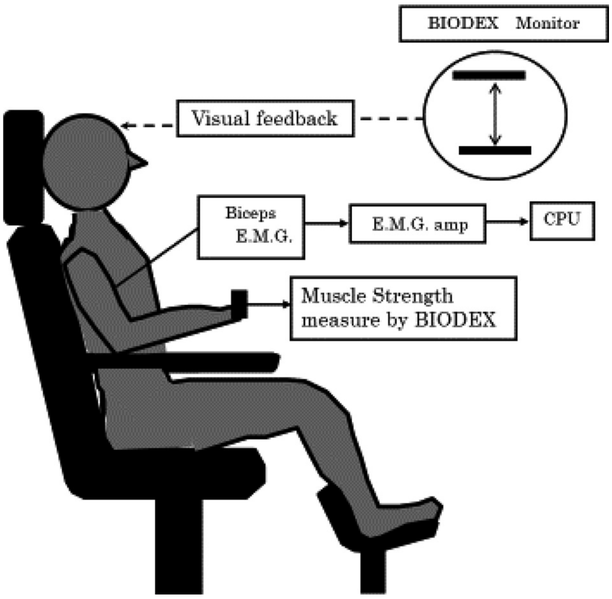

Fig. 1. BIODEX 4 (Sakai Medical, KK) was used to measure the maximum voluntary isometric contraction (MVC) of the right elbow joint flexion (bicep). The 10% MVC, 30% MVC, 50% MVC, 70% MVC, and 80% MVC of the bicep for each subject were calculated, which were used as target values for exerted muscle strength, and the target line was displayed on the BIODEX monitor screen.

Table 1. Mean measurement values

	% Maximum voluntary contraction					
	10%	30%	50%	70%	80%	100%
Mean torque value (Nm)	5.0 ± 1.1	15.0 ± 3.4	25.1 ± 5.6	35.0 ± 7.8	40.1 ± 9.0	50.2 ± 11.1
Median power frequency (Hz)	94.0 ± 10.3	105.2 ± 14.6	101.3 ± 13.1	93.3 ± 16.8	90.6 ± 18.2	85.9 ± 16.6
Low-frequency band content rate (%)	19.3 ± 3.3	15.4 ± 2.1	15.7 ± 2.2	17.7 ± 3.7	18.1 ± 4.0	19.2 ± 4.2
Medium-frequency band content rate (%)	30.8 ± 3.6	29.7 ± 4.6	30.9 ± 4.1	33.8 ± 5.6	35.4 ± 6.7	37.3 ± 7.2
High-frequency band content rate (%)	49.9 ± 5.3	54.9 ± 6.2	53.4 ± 5.9	48.5 ± 8.5	46.5 ± 10.0	43.5 ± 10.6
Amplitude integrated value (mv)	405 ± 297	1,363 ± 812	2,552 ± 1,484	4,192 ± 2,589	5,019 ± 3,770	5,556 ± 3,431

The median power frequency shows high values of 85–105 Hz throughout the measurements from 10% MVC to 100% MVC, which represent significant changes (ANOVA $p<0.01$). As for the frequency band component content rates, the low-frequency band component (less than 45 Hz) and the medium-frequency band component (45–95 Hz) content rate decreased as the muscle strength changed from 10% MVC to 30% MVC, and increased significantly (Tukey $p<0.01$) from 50% MVC through 100% MVC. The high-frequency band component (95 Hz or more) content rate decreased significantly (Tukey $p<0.01$) from 50% MVC through 100% MVC.
EMG amplitude integrated values increased significantly (ANOVA $p<0.01$) from 10% MVC to 100% MVC.

RESULTS

Table 1 shows the median power frequency results. The median power frequency revealed values of 85–105 Hz throughout the measurements from 10% MVC to 100% MVC, which represented significant changes (ANOVA $p<0.01$). In the multiple comparison results, the frequency significantly increased from 10% to 30% MVC (Tukey $p<0.01$) and decreased significantly from 50% to 100% MVC (Tukey $p<0.01$).

The mean frequency band component content rates are shown in Table 1.

With respect to the frequency band component content rates, the low-frequency band component (less than 45 Hz) content rate ranged from 15% to 19% throughout the measurements, which represented significant changes (ANOVA $p<0.01$). These changes decreased from 30% MVC (15.4 ± 2.1%) through 10% MVC (19.3 ± 3.3%) and then increased from 50% MVC (15.7 ± 2.2%) through 100% MVC (19.2 ± 4.2%). In the multiple comparison results, there was a significant difference (Tukey $p<0.01$) between 10% MVC versus 30% MVC and 50% MVC, and a significant difference was observed between 50% MVC versus 80% MVC (Tukey $p<0.05$) and 100% MVC (Tukey $p<0.01$).

The medium-frequency band component (45–95 Hz) content rate ranged from 29% to 37% throughout the measurements, which represented significant changes (ANOVA $p<0.01$). These changes decreased from 30% MVC (29.7 ± 4.6%) through

10% MVC (30.8 ± 3.6%) and subsequently increased from 100% MVC (37.3 ± 7.2%) through50% MVC (30.9 ± 4.1%). In the multiple comparison results, a significant difference was observed between 30% MVC versus ≥70% MVC and between 50% MVC versus 80% MVC and 100% MVC (Tukey $p<0.01$).

The high-frequency band component (95 Hz or more) content rate ranged from 43–54%, which was higher than for the other components, and represented significant changes (ANOVA $p<0.01$). These changes increased from 10% MVC (49.9 ± 5.3%) through 30% MVC (54.9 ± 6.2%) and decreased from 50% MVC (53.4 ± 5.9%) through 100% MVC (43.5 ± 10.6%). In the multiple comparison results, a significant difference was observed between 50% MVC versus 80% MVC and 100% MVC (Tukey $p<0.01$).

Table 1 shows the mean values. In the EMG amplitude integrated values, there were significant increases from 10% MVC to 100% MVC (ANOVA $p<0.01$). In the multiple comparison results, there were significant differences between the integrated values during voluntary contractions of 10% MVC versus ≥50% MVC as well as the integrated values during 50% MVC versus 80% MVC and 100% MVC (Tukey $p<0.01$), and the EMG integrated values increased alongside increased muscle strength values.

DISCUSSION

In the analysis of the surface EMG after applying Daubechies N=10 wavelet function of the discrete wavelet transform in this study, we found that central nervous system factors could be assessed from the differences in load during muscle contraction, changes over time, and maximum muscle strength. Data analysis in biosignal processing can be classified into time domain analysis, frequency domain analysis, and time-frequency domain analysis[15]. In the frequency domain analysis, the Fourier transform and the fast Fourier transform, which can significantly reduce the calculation time, are commonly used[15]. The Fourier transform is a method of transforming the signal from the time domain to the frequency domain using the Fourier function, and it analyzes the EMG signal in two dimensions: frequency and potential domains (power spectrum=frequency-square of potential) which are subsequently used for identifying muscle fatigue, muscle fiber type, etc. However, it is considered to be unsuitable for a three-dimensional analysis in the time-frequency-potential domain because the conditions of use are limited given the presumption that muscle activity waveforms are stationary[15]. Wavelet transform enables three-dimensional analysis in this time-frequency-potential domain[16]. Wavelet transform can elicit local frequency distribution at each time point, which is not possible with Fourier transform, making three-dimensional analysis in this time-frequency-potential domain possible, and the analysis is said to have a higher temporal resolution compared to that of fast Fourier transform[15, 16]. Wavelet transform includes continuous wavelet transform and discrete wavelet transform. Continuous wavelet transform (CWT) is an analysis method that calculates the inner product of a signal using a continuous wavelet function, and these methods include Mexican Hat, Gabor, Morlet, and Shannon. In an EMG analysis performed using these methods, the transform results are visualized, but the amount of data is too large to perform quantification, which is the next process, and the inverse transform process is so complicated that it is rarely used for actual quantification. Discrete wavelet transform (DWT) was devised by the Belgian mathematician Ingrid Daubechies in 1988 as a typical configuration method used for multiresolution analysis. There are different types of Daubechies, such as N=2, N=4, N=5, etc., and each has a set of scaling functions numbered by a natural number N and corresponding wavelets. In the EMG analysis in this study, a three-dimensional analysis in the time-frequency-potential domain was possible, and the wavelet function of the discrete wavelet transform Daubechies N=10, which is highly localized in the frequency domain and has excellent frequency characteristics, was applied. With this method, it was possible to decompose the original signal into high-frequency and low-frequency components by performing inverse transform using the transformed wavelet coefficient. Furthermore, it was possible to advance decomposition processing into high-frequency and low-frequency components. However, the frequencies of the decomposed waveforms were not evenly spaced. Hence, processing was performed to decompose the waveforms into evenly spaced waveforms.

Changes in the EMG amplitude integrated values increased significantly from 10% MVC to MVC alongside the increase in muscle strength values. This increase may be due to strategic changes in the central nervous system factors from 10% MVC through 30% MVC, or from 30% MVC to 50% MVC, or MVC for each muscle strength value. As voluntary muscle contractions increase during MVC, interference waveforms from the surface EMG become dense and the amplitude increases. Recruitment and rate coding[1–4] are involved in this phenomenon, and their interaction increases the voluntary muscle contraction strength. It has been reported that the EMG integrated values increase linearly with this increase in muscle contraction[17], and there are reports that relative increases are observed between the EMG integrated values and muscle strength from around 70–80% MVC[18–20]. The size principle is also considered to be a factor that can change muscle strength[7]. Henneman et al.[7] reported that Type I muscle fibers that are dominated by small motor neurons mainly contract at low loads, the involvement of Type IIa fibers increases with increasing muscle strength at high loads, and more strength further increases the involvement of Type IIb fibers (the size principle). Milner-Brown et al.[21] found that recruitment is the main activity during weak voluntary contractions and increases in rate coding are broadly associated with the transition from 70% MVC to strong muscle contractions of 80% MVC or more. During MVC, adjustments are made to obtain efficient and high voluntary contractions by rate coding and synchronization of the firing time of motor units involved in contraction[22], and it is thought that these cause the overlap of EMG waveforms during MVC by making the interference waveforms of the surface EMG dense. From the above, it is postulated that the median frequency gradually increases as the load increases, and the frequency content rate also increases from the low-frequency band component showing the activity of Type I fibers to the high-frequency band component showing the activity of Type II fibers. Moreover, when the load exceeds 70–80% MVC, the size principle and recruitment cease to apply and rate coding rises, thus leading to the inference that the EMG integrated values reach their maximum during MVC. However, the results of the median power frequency and frequency band component content rate showed changes that were different to those that were expected.

The high-frequency band component content rate in the median power frequency and frequency band component content rate tended to be high across all measurements. Nagata1) reported that in a surface EMG of the biceps, the components below 40 Hz were more prominent than those above 60 Hz over time during muscle contraction. Furthermore, Nagata1) reported that the EMG frequency becomes a slow wave when the muscle is fatigued and transitions to a low-frequency band. In the results of our study, the median power frequency ranged from 85–105 Hz throughout the measurements, which is higher than has been previously reported. As a result, it was considered that the effect of muscle fatigue could be ruled out as the reason for the results of this study showing changes different to what was expected.

Changes in median power frequency for each load measurement increased from 10% MVC to 30% MVC and decreased from 50% MVC through 100% MVC. With respect to the changes in the frequency band component content rate, the low- and medium-frequency component content rates decreased from 10% MVC to 30% MVC and increased from 50% MVC through 100% MVC. The high-frequency band component content rate increased between 10% MVC and 30% MVC and decreased from 50% MVC through 100% MVC. This change shows that as the load increases from 10% MVC to 30% MVC, there are increased numbers of Type IIb fibers that become involved in muscle contraction, with the involvement decreasing from 50% MVC to 100% MVC. Conversely, it is possible that Type I fibers and Type IIa fibers may show increased involvement in muscle contraction from 50% MVC to 100% MVC. Regarding the activity ratio between recruitment and rate coding during muscle contraction, Basmajian et al.2) reported that voluntary contractions are regulated by rate coding in small muscles first and by recruitment in large muscles. These muscle contractions may end mobilization when the recruitment is 50% MVC or less depending on the contracting muscles and style of activity, or may continue to 70% MVC or 80% MVC, and the rate coding also changes relatively due to the differences in the muscles. Solomonow et al.23) found that the relationship between muscle strength and EMG was influenced by the ratio of recruitment to rate coding for each muscle and it depends on recruitment from a low MVC% to 50% MVC and from 50% MVC to 100%, whereas from 50% MVC to 100% MVC, strength is gained by rate coding. It was also reported that recruitment showed non-linear changes when performed above 60% MVC23). Freund et al.24) found that rate coding has the highest firing rate in small motor units with a low threshold potential (Type I fibers) and the lowest firing rate in large motor units with a high threshold potential (Type IIb fibers). Regarding the relationship between recruitment and rate coding during voluntary muscle contractions, results that motor units with low threshold potential have the highest firing rate, and motor units with high threshold potential have the lowest firing rate24–26) have been reported, and studies with the exact opposite results have also been reported27, 28). When only looking at the results of our study, as a result of analyzing the changes in median power frequency and frequency band component content rate, we believe that the effect of the size principle was seen from 10% MVC to 30% MVC. Moreover, changes from 30% MVC to 70% MVC reflect the effect of recruitment, and at this time point, low-frequency band component (Type I fibers) and medium-frequency band component (Type IIa fibers) are mostly mobilized for muscle contraction, possibly leading to an increase in the EMG integrated values. It was inferred that changes in the median power frequency and frequency band component content rate and increases in the EMG integrated values from 80% MVC to 100% MVC were affected by rate coding. Freund et al.24) reported that rate coding showed the highest firing rate in small motor units with a low threshold potential, and the increase in the firing rate of Type I fibers was considered to be a factor involved in this phenomenon.

Hamamoto et al.29) reported on the analysis of surface EMG in 14 healthy adult females during MVC in isometric muscle contraction of the right bicep and showed that it became impossible to maintain 90° elbow flexion from the start of the incremental load. After the MVC measurement, the data for 1 second before and after were selected for each time point when the load was increased by 5% MVC during incremental isometric muscle contraction, and the analysis was performed using discrete wavelet transform Daubechies 5. In this report, the involvement of high-frequency band components increased from 15% MVC through 20% MVC when the load was incrementally changed during muscle contraction. Subsequently, it was reported that the involvement of high-frequency band components decreased as the load increased. This report on incremental loads is similar to the results of our research and shows trends that are contrary to already established theories. The results obtained by Hamamoto et al.29) and the changes in the EMG frequencies seen in our study were due to similar effects.

We concluded that increases in the EMG amplitude integrated values, changes in the median instantaneous frequency spectrum, and changes in the frequency band component content rate may represent strategic changes in central nervous system factors from 10% MVC to 100% MVC, and the increases were significant alongside the increase in muscle strength values. In our analysis of the surface EMG after applying Daubechies N=10 wavelet function of the discrete wavelet transform in this study, central nervous system factors could be assessed from the differences in load during muscle contraction, changes over time, and maximum muscle strength.

Conflicts of interest

None declared.

ACKNOWLEDGEMENT

We would like to express our gratitude to all the first-year students of the Department of Physical Therapy and the faculty of Social Work Studies, Josai International University, for their assistance in the experiment and cooperation in this study.

REFERENCES

1) Nagata A: The science of muscles and muscle strength. Tokyo: Fumaido Publishing, 1984, pp 19–164.
2) Basmajian JV, De Luca CJ: Muscles alive: their functions revealed by electromyography, 5th ed. Baltimore: Williams & Wilkins, 1985, pp 19–251.
3) Hourigan ML, McKinnon NB, Johnson M, et al.: Increased motor unit potential shape variability across consecutive motor unit discharges in the tibialis anterior and vastus medialis muscles of healthy older subjects. Clin Neurophysiol, 2015, 126: 2381–2389. [Medline] [CrossRef]
4) Clamann HP: Motor unit recruitment and the gradation of muscle force. Phys Ther, 1993, 73: 830–843. [Medline] [CrossRef]
5) Housh DJ, Housh TJ, Johnson GO, et al.: Hypertrophic response to unilateral concentric isokinetic resistance training. J Appl Physiol 1985, 1992, 73: 65–70. [Medline]
6) Abe T, DeHoyos DV, Pollock ML, et al.: Time course for strength and muscle thickness changes following upper and lower body resistance training in men and women. Eur J Appl Physiol, 2000, 81: 174–180. [Medline] [CrossRef]
7) Henneman E, Somjen G, Carpenter DO: Functional significance of cell size in spinal motoneurons. J Neurophysiol, 1965, 28: 560–580. [Medline] [CrossRef]
8) Uchiyama T: Recurrent inhibition and muscle tension control by Renshaw cells. J Soc Biomechanisms, 2003, 27: 276–282. [CrossRef]
9) Chen YS, Crowley Z, Zhou S, et al.: Effects of 12-week Tai Chi training on soleus H-reflex and muscle strength in older adults: a pilot study. Eur J Appl Physiol, 2012, 112: 2363–2368. [Medline] [CrossRef]
10) Morozumi K, Nishiya T, Akimasa Y, et al.: Effect of bite with and without dentures on soleus H-reflex in the elderly. J Phys Ther Sci, 2002, 14: 51–55. [CrossRef]
11) Burke D: Clinical uses of H reflexes of upper and lower limb muscles. Clin Neurophysiol Pract, 2016, 1: 9–17. [Medline] [CrossRef]
12) De Luca CJ, Adam A, Wotiz R, et al.: Decomposition of surface EMG signals. J Neurophysiol, 2006, 96: 1646–1657. [Medline] [CrossRef]
13) Nawab SH, Wotiz RP, De Luca CJ: Decomposition of indwelling EMG signals. J Appl Physiol 1985, 2008, 105: 700–710. [Medline]
14) Chowdhury SK, Nimbarte AD: Comparison of Fourier and wavelet analysis for fatigue assessment during repetitive dynamic exertion. J Electromyogr Kinesiol, 2015, 25: 205–213. [Medline] [CrossRef]
15) Hayashi R, Mita K, Hanaoka M, et al.: Surface electromyography and mechanomyography: principles and practice. Tokyo: Toyo Publishing, 2018, pp 432–552.
16) Phinyomark A, Nuidod A, Phukpattaranont P, et al.: Feature extraction and reduction of wavelet transform coefficients for EMG pattern classification.Elektronika ir Elektrotechnika, 2012, 122: 27–32.
17) Lippold OC: The relation between integrated action potentials in a human muscle and its isometric tension. J Physiol, 1952, 117: 492–499. [Medline] [CrossRef]
18) Kuroda E, Klissouras V, Milsum JH: Electrical and metabolic activities and fatigue in human isometric contraction. J Appl Physiol, 1970, 29: 358–367. [Medline] [CrossRef]
19) Gerdle B, Edström M, Rahm M: Fatigue in the shoulder muscles during static work at two different torque levels. Clin Physiol, 1993, 13: 469–482. [Medline] [CrossRef]
20) Badier M, Guillot C, Lagier-Tessonnier F, et al.: EMG power spectrum of respiratory and skeletal muscles during static contraction in healthy man. Muscle Nerve, 1993, 16: 601–609. [Medline] [CrossRef]
21) Milner-Brown HS, Stein RB, Yemm R: Changes in firing rate of human motor units during linearly changing voluntary contractions. J Physiol, 1973, 230: 371–390. [Medline] [CrossRef]
22) Yao W, Fuglevand RJ, Enoka RM: Motor-unit synchronization increases EMG amplitude and decreases force steadiness of simulated contractions. J Neurophysiol, 2000, 83: 441–452. [Medline] [CrossRef]
23) Solomonow M, Baratta R, Shoji H, et al.: The EMG-force relationships of skeletal muscle; dependence on contraction rate, and motor units control strategy. Electromyogr Clin Neurophysiol, 1990, 30: 141–152. [Medline]
24) Freund HJ, Büdingen HJ, Dietz V: Activity of single motor units from human forearm muscles during voluntary isometric contractions. J Neurophysiol, 1975, 38: 933–946. [Medline] [CrossRef]
25) Freund HJ: Motor unit and muscle activity in voluntary motor control. Physiol Rev, 1983, 63: 387–436. [Medline] [CrossRef]
26) De Luca CJ, Erim Z: Common drive of motor units in regulation of muscle force. Trends Neurosci, 1994, 17: 299–305. [Medline] [CrossRef]
27) Gydikov A, Kosarov D: Some features of different motor units in human biceps brachii. Pflugers Arch, 1974, 347: 75–88. [Medline] [CrossRef]
28) Grimby L, Hannerz J, Hedman B: Contraction time and voluntary discharge properties of individual short toe extensor motor units in man. J Physiol, 1979, 289: 191–201. [Medline] [CrossRef]
29) Hamamoto T, Nagase S, Hirata K, et al.: Proposal of surface EMG analysis parameters using discrete wavelet transform. J Investig Phys Ther, 2005, 8: 15–19.