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EMG-force relationship during static contraction: Effects on sensor placement locations on biceps brachii muscle

Nizam Uddin Ahamed[a,*], Kenneth Sundaraj[b], Mahdi Alqahtani[c], Omar Altwijri[c], Md. Asraf Ali[b] and Md. Anamul Islam[b]
[a]*Innovative, Manufacturing, Mechatronics and Sports Lab (iMAMS), Faculty of Manufacturing Engineering, Universiti Malaysia Pahang, UMP Pekan, Pahang, Malaysia*
[b]*AI-Rehab Research Group, Universiti Malaysia Perlis (UniMAP), Kampus Pauh Putra, Arau, Perlis, Malaysia*
[c]*Biomedical Technology Department, College of Applied Medical Sciences, King Saud University, Riyadh, Kingdom of Saudi Arabia*

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Abstract.
BACKGROUND: The relationship between surface electromyography (EMG) and force have been the subject of ongoing investigations and remain a subject of controversy. Even under static conditions, the relationships at different sensor placement locations in the biceps brachii (BB) muscle are complex.
OBJECTIVE: The aim of this study was to compare the activity and relationship between surface EMG and static force from the BB muscle in terms of three sensor placement locations.
METHODS: Twenty-one right hand dominant male subjects (age 25.3 $\pm$ 1.2 years) participated in the study. Surface EMG signals were detected from the subject's right BB muscle. The muscle activation during force was determined as the root mean square (RMS) electromyographic signal normalized to the peak RMS EMG signal of isometric contraction for 10 s. The statistical analysis included linear regression to examine the relationship between EMG amplitude and force of contraction [40–100% of maximal voluntary contraction (MVC)], repeated measures ANOVA to assess differences among the sensor placement locations, and coefficient of variation (CoV) for muscle activity variation.
RESULTS: The results demonstrated that when the sensor was placed on the muscle belly, the linear slope coefficient was significantly greater for EMG versus force testing ($r^2 = 0.62$, $P < 0.05$) than when placed on the lower part ($r^2 = 0.31$, $P > 0.05$) and upper part of the muscle belly ($r^2 = 0.29$, $P < 0.05$). In addition, the EMG signal activity on the muscle belly had less variability than the upper and lower parts (8.55% vs. 15.12% and 12.86%, respectively).
CONCLUSION: These findings indicate the importance of applying the surface EMG sensor at the appropriate locations that follow muscle fiber orientation of the BB muscle during static contraction. As a result, EMG signals of three different placements may help to understand the difference in the amplitude of the signals due to placement.

Keywords: Biceps brachii, electromyography, EMG, force, sensor placement, relationship

*Corresponding author: Nizam Uddin Ahamed, Innovative, Manufacturing, Mechatronics and Sports Lab (iMAMS), Faculty of Manufacturing Engineering, Universiti Malaysia Pahang, 26600 UMP Pekan, Pahang, Malaysia. Tel.: +60 942 459 36; E-mail: ahamed1557@hotmail.com.

1. Introduction

The analysis of the electromyographic (EMG) signals generated by living individuals is an established area of research that was first initiated by Francesco in 1666 [1]. Surface EMG has since been used as a non-invasive technique to assess muscle function and is usually defined as a measure of muscle activation that focuses the algebraic summation of muscle action potentials passing under the recording electrodes or sensors [2,3]. In general, EMG signals are detected and recorded by placing the minimum number of single-channel connected sensors over the skin with a negative-positive conductor. A number of studies on EMG have been published, and researchers have demonstrated that the amplitude and frequency components of surface EMG are observably influenced by the sensor conditions, such as the diameter of the surface electrodes, the inter-electrode distance, the relative position of the bipolar electrode to the muscle belly, and the angle of the sensors to the direction of the muscle fibers [4–6].

Most research of surface EMG to date has been restricted to the BB muscle belly; however, a detailed understanding of the exact signal characteristics of other parts of the BB muscle is essential. For example, limited information is available on the lower part of the muscle belly, which is located between the BB muscle endplate region and the distal tendon insertion, and over the medial belly of each head (long and short head), which is parallel to the muscle fibers and below the proximal tendon [7,8], because EMG exhibits stochastic features during movement. However, great care is usually taken when placing the sensors to avoid the innervations zone (the location where the nerve terminals and muscle fibers are connected) and the tendons (end of muscle fiber) [9,10]. Some reports in the literature have discussed EMG activity in terms of sensor placement locations on BB muscle (except muscle belly) [11–13]. The EMG activity in these sensor placement areas was then measured using different analysis techniques and protocols.

Regression analysis is an important diagnostic criterion for examining the relationship between EMG amplitude and force under various sensor placement locations on the BB muscle during contraction [14]. The relationship between EMG activity and muscle tension is affected by multiple factors, such as sensor placement, anthropometric parameters, body movement and so on [15]. Moreover, the relationship between EMG and muscle activation remains unpredictable during contraction and at different locations on the muscle [16]. A number of studies have used regression analysis to determine the relationship and accuracy of the prediction between EMG and force. For example, Rantalainen et al. examined the EMG-force/torque relationship from the BB muscle. The authors found that the disruption of the physiological signal (EMG) caused by the innervations zones changed the reliability of the EMG-force relationship at a single bipolar channel level [17]. Doheny et al. examined the effects of joint angle on the relationship between EMG amplitude and median frequency and force in the BB, brachioradialis, and triceps brachii muscles. Based on that study, it was recommended that force should be normalized with respect to its maximum value at each angle when using surface EMG to estimate muscle excitation [18].

But, as whole, it is not clear how sensor placement at different locations affects the relationship between surface EMG and force on the BB muscle of the upper limb during static contraction. Thus, this study was designed to assess the following: (i) is the EMG (%MVC)-force relationship better in the muscle belly than the other two locations, and (ii) can the muscle belly generate maximum signals and show less signal variability than the other two sensor placement locations during static contraction. The rationale for our hypothesis is that previous studies identified muscle spindles (sensory receptors within the belly of a muscle that mainly identify the changes in the muscle length) within this area and embedded in extra fusal muscle fibers [19,20]. As a result, when the muscle is contracted, this area is able to recruit maximum motor units. Thus, the manuscript concerns the placement of EMG electrodes on the brachial biceps muscle, which would be optimal with respect to the EMG-force relationship.

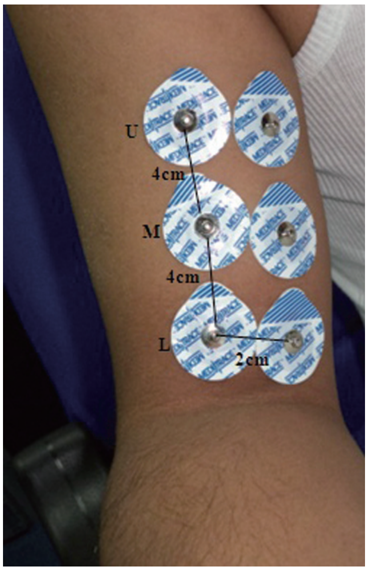

Fig. 1. Three sensor placement sites on the biceps brachii muscle. (Colours are visible in the online version of the article; http://dx.doi.org/10.3233/THC-140842)

2. Methods

2.1. Subjects and procedures

Twenty-one healthy male subjects (age 25.3 ± 1.2 years, weight 68 ± 2.1 kg and height 166.7 ± 3.6 cm; mean ± SD, respectively) participated in the study. None of the subjects had a history of musculoskeletal pain or injury in their dominant arm (right) or BB muscle. All subjects provided informed and written consent prior to the beginning of the experiment and completed a questionnaire describing their pretest health condition. All experimental procedures conformed to the Declaration of Helsinki and were approved by the Human Research Ethics Committee at the University.

Subjects were comfortably seated in a specially designed test chair (fixed in position by a belt) with the right elbow on a padded support and the elbow joint at a right angle. The subject performed static contraction using a digital hand grip dynamometer (Camry, China). An initial adjustment period allowed the subject to perform static contractions with the hand dynamometer. The main task was to apply an upward-directed force with the wrist using the elbow flexor muscles, primarily the BB muscle. The angle of the elbow joint during data recording was 90°. This angle measurement was from the shoulder to elbow and the elbow to palm, and was calculated by a digital inclinometer. The exerted force (kg) by the subject was visible on the dynamometer. EMG was recorded for three force levels; 16, 20, and 25 kg. To determine the generated maximal voluntary contraction (MVC) force, subjects were asked to enhance the force to maximum at a constant rate over 10 s and to hold the maximum force for a period of 10 s. Signals were recorded from each subject three times (three trials) for each force from each sensor location with a 5 min interval between the trials.

2.2. Data collection

A wireless Shimmer^{TM}sensor (Real-time Technologies Ltd., Ireland) was used to record the signal [EMG activity in microvolts (mV)]. The two inputs (negative and positive) were used for the EMG recording and a third was used as the reference channel. Three silver-silver chloride (Ag-AgCL) surface electrodes were used to record EMG signals from the muscle. The entire protocol was designed to minimize movement artifact and make sure a tolerable level of electrode impedance (inter-electrode impedance was less than 2000 Ω). Three sensor placement locations were chosen to evaluate the EMG variability: over the muscle belly ('M'); the lower part of the muscle belly ('L') (i.e., between the BB muscle endplate region and the distal tendon insertion); and the medial belly of each long and short head ('U') (i.e., parallel to the muscle fibers and below the proximal BB tendon) [7,8]. The belly of the BB muscle was recognized by manual palpation. The distance between the centers of the sensors located at 'U' (upper) and 'M' (middle) and between the sensors at 'M' and 'L' (lower) was 4 cm; the distance between the center of the sensors at 'U' and 'L' was 8 cm, with 2 cm between each pair (Fig. 1). The reference electrode was attached to the lateral epicondyle of the humerus of the right arm (approximately 1 inch on the olecranon of the elbow). However, excessive care was taken when placing the sensors to avoid the innervations zone and tendons (end of muscle fiber), because the length of this part of BB is relatively shorter than other muscles [9,10]. Figure 1 depicts the three sensor placement areas on the BB muscle.

2.3. Data analysis

The raw EMG signals were recorded at a sampling rate of 1 KHz before their A-D conversion and stored on a compatible computer. The fourth-order bandpass Butterworth filter was used to remove any skin movement artifacts as well as high frequency noises (cutoff frequency between 10 to 500 Hz). The recorded digitized EMG data sets were processed offline by filtering, windowing, and extracting signal. Signal processing was performed with Matlab (MathWorks, USA) software.

A normalization procedure was used to analyze the EMG amplitude. Thus, all data were normalized in terms of the root mean square (RMS) values (i.e., the individual RMS values of the tests during the contraction were taken as 100% MVC). The filtered EMG activity was normalized within each subject by dividing the observed EMG value for BB muscle by the maximum value recorded during the three maximal tests. The mean (RMS) normalized EMG activity was then calculated as the mean of the sum of the normalized EMG percentages from all subjects in each sensor location. The EMG associated with 100% MVC was designated as 100% and fractions thereof. The maximum peak-to-peak value of the EMG was considered as a relative measure of motor activity (the positive portion of the peak is defined as the peak-maximum). It should be noted that this normalization procedure only presents information of the level of muscle activity with regard to this peak value (i.e., shape of the EMG pattern).

2.4. Statistics

Statistical analysis was performed using Minitab^{TM} software (version 13.32). Significant differences in three sensor placement locations between the force and EMG amplitudes (RMS) were detected using repeated measures with analysis of variance (ANOVA), and post-hoc tests were applied to test differences with the significance level set at $\alpha = 0.05$, 95% ($P < 0.05$) confidence intervals for all variables. EMG signal steadiness or muscle activity variation was characterized by the coefficient of variation (i.e., the ratio of their standard deviations divided by their means, coefficient of variation (CoV) $= \sigma/\mu$). Linear regression (r^2) analysis was used to analyze the relationship between force and the EMG variables. In addition, the null hypothesis of linearity of each regression was tested using the F-test. Finally, each individual data set from the trial was calculated to fit with a linear regression line in the logarithmic curve fitting form of $y = a + bx$.

Table 1
Analysis of average EMG RMS (mV) values from three sensor locations from corresponding forces

SPL	NEA (mean ± SD)			Total (mean ± SD)	CoV (%)	r^2	F-ration
	Force 16 kg	Force 20 kg	Force 25 kg				
'L'	3.43 ± 0.59	2.47 ± 0.36	3.67 ± 0.25	3.19 ± 0.41	12.86	0.31	0.62
'M'	3.15 ± 0.46	3.79 ± 0.34	4.26 ± 0.21	3.74 ± 0.32	8.55	0.62	30.39
'U'	2.88 ± 0.83	3.12 ± 0.34	3.71 ± 0.32	3.24 ± 0.49	15.12	0.29	8.09

SPL: sensor placement location, NEA: normalized EMG amplitude (RMS), CoV: coefficient of variation.

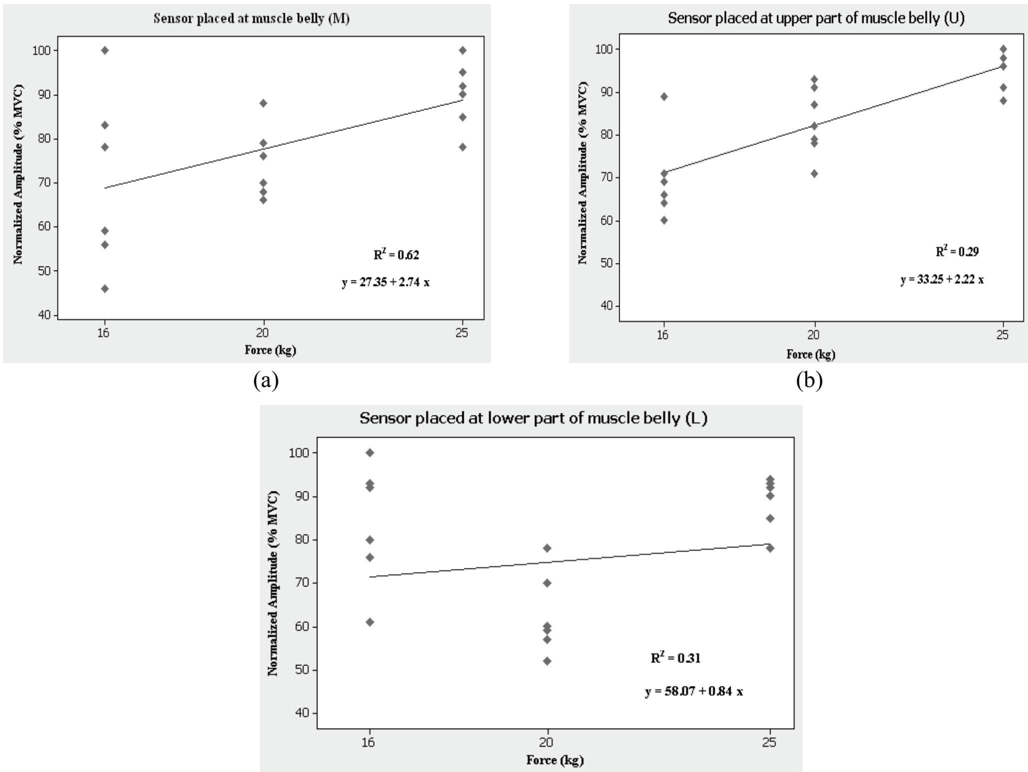

Fig. 2. Relationship between EMG (%MVC) and force at three sensor placement locations.

3. Results

Table 1 shows the summarized results from the subject's BB muscle at three sensor locations. The average EMG RMS amplitude data (mean ± SD) of three sensor locations are shown from each generated force in Table 1. As expected, the muscle belly ('M') showed considerably higher amplitudes (3.74 ± 0.32 mV) than the other two sensor locations ('L' and 'U': 3.19 ± 0.41 mV and 3.24 ± 0.49 mV, respectively) during the contraction. However, the average EMG was slightly less during low force production (16 kg), which argued against our hypothesis regarding the 'M' location, but continuously rose as the force increased. The EMG analyses showed that the lowest activity was in the lower part of the BB muscle belly (according to total results). Overall, most of the data indicated that the EMG amplitude of the muscle increased as the force increased. The CoV data from the EMG are included in Table 1 to assist in the comparison of the EMG and steadiness results. Taken together, these results showed that the generated signal was more variable at the locations ‘U’ (15.12%) and ‘L’ (12.86%) compared to ‘M’ (8.55%).

The linear relationships (r^2) between normalized EMG (%MVC) and the corresponding force are presented in Table 1. These values indicate the goodness of fit of a linear model applied to the data. The EMG slope coefficient (F-ratio) was also analyzed and indicated the difference was significant or it was not a significant between the sensor placement locations. Figure 2 shows the normalized EMG RMS amplitude data (in %MVC) plotted as a function of force for the static contraction in each sensor location whereby the regression lines have been drawn through the data points. The regression lines in the figures show the linear positive relationship between EMG and force in muscle belly and upper sensor placement sites (40–100% MVC). Whereas, results obtained from the lower part of the muscle, was no significant difference found between the three force levels. As shown in Figs 2(a)–(c), we found that the linear mode provides the best fit for the normalized EMG amplitude (%MVC) versus force relationship for the muscle belly ($r^2 = 0.62$) than the upper ($r^2 = 0.29$) and lower sites ($r^2 = 0.31$). We also observed that the high correlation between EMG and force was significant ($P < 0.05$). Three ratios (F-value) were also identified for the three sensor placement sites, whereby the ratio between EMG and force at the muscle belly (M) and upper part (U) was statistically significant (30.39 and 8.09; $P = 0.001$ and $P = 0.010$ respectively). But, the EMG amplitude did not vary significantly in the lower part of muscle with force ($P = 0.43$), which is what the F-ratio of 0.62 tells us anyway. In the regression lines shown in Fig. 2 with $y = a + bx, x$ refers to the value in the force and y to the value in the normalized EMG (%MVC).

4. Discussion

Understanding the impact of the EMG-force relationship and activity on the BB muscle is required to establish proper biomedical assessments and clinical evaluations. However, little work has been published to address this issue to date, and some evidence suggests that sensor placement has an effect on the BB muscle during different levels of static contraction (MVC). Thus, the main focus of this study was to examine the differences in activity and linearity of the EMG (amplitude) versus static force relationships for the BB muscle in terms of three sensor placement locations. Our findings demonstrated that the muscle belly generates highest electrical signal and has less signal variability during contraction. In addition, there was a strong linear relationship between EMG and force in this muscle region compared to the other two locations.

The present findings are in accordance with the observations of previous studies [21,22] that have assessed linear EMG amplitude (from the BB muscle) versus force relationships during contractions. This study extends the outcomes of those previous reports regarding the use of sensors on the BB muscle, which suggested that EMG and force has a strong relationship with electrode placement on the muscle belly. For example, electrical stimulation effectively isolated the force-length relationships of the BB during isometric contraction when surface electrodes of 8 mm diameter were placed 20 mm apart longitudinally on the belly of the BB muscle [23]. Another study showed that EMG signal on BB muscle has some effects (e.g., index sensitivity, fatigue) under appropriate inter-electrode distance and longitudinal electrode position (positioned 5 mm apart along the direction of the fibers). The results were determined under isometric (static) contraction with 20, 40, 60, 80, and 100% MVC [24]. In addition, it has been shown that the EMG signal caused by the innervations zone affects the consistency EMG-force relationship sub-maximal voluntary isometric contractions [17]. In that study, the electrode grid was placed over the mid-line of the short head of the BB, which was 2 cm proximal to the distal tendon.

Our results also indicate that the EMG signals from the lower part of the muscle belly (near the tendon and innervations zone) is inconsistent, similar to previous findings. Thus, it is important to standardize the electrode orientation on BB muscle in order to compare the pattern of responses for EMG amplitude values and to normalize EMG amplitude data for evaluating the mean values during contraction [25]. One of the unique findings of our study was the difference between the shapes of the EMG-force based on normalized EMG data. These assumptions are further supported by the statistically significant EMG increase observed when the contraction level increased from 40% to 100% MVC (Figs 2(a) and (b)).

One question that arises from this study is if the signal variability was due to inter-electrode distance or by the level of contraction. During contraction, the inter-muscular variability of the signal depends mainly on the sensor placement locations or inter-electrode distance [26,27]. Based on our EMG data, EMG amplitude was steady on the muscle belly compared to the lower and upper parts (4 cm apart from belly to other two locations; Fig. 1). Muscles are generally assumed to have consistent activations across their bellies [28], and there is strong evidence to support the findings with respect to other sensor placement areas compared to the muscle belly. Indeed, Cerqueira et al. found low-to-moderate coefficients of variation (CoV; 14.6–16.3% for RMS) during both the dynamic and static contractions when the EMG sensor was located on the greatest portion of the BB tendon [29].

No study to date has compared the results and shown a relationship for the EMG-force obtained from these three sensor placement locations on the BB muscle during isometric contraction mode. Our results demonstrated that the EMG amplitude increases or decreases as the distance between the sensors and the muscle length increases or decreases, respectively. Together, our findings suggest that the specific placement of EMG sensors for recording muscle strength should be decided with utmost care. Sensor placement on an exact muscle location is essential, since the EMG amplitude is reduced when the sensors are placed over the myotendonous junction or innervations zone. In this study, these locations were located below the muscle belly ('L') and on the upper part of the belly ('U') [30]. A number of previous studies have suggested that EMG amplitude is maximized when the sensors are placed over the muscle belly, but not between the innervations zone and tendonous region [7,21,30]. Our results also show similarities to previous findings whereby the muscle belly provides a good relationship in terms of EMG (%MVC) versus force compared to the other two locations. However, previous studies have not simultaneously compared this relationship among these three sensor locations on BB muscle. This is the first study that clearly shows EMG activity (relationship) in the top to bottom (muscle fibers) locations of the BB muscle.

The findings from this study may be useful to researchers who are interested in EMG analysis in terms of force, sensor placement, and contraction. In addition, our results can be used for different practical applications for the non-invasive evaluation of BB muscle. Our findings may also help to develop EMG-assisted biomechanical modeling techniques as well as prosthetic devices. Surface EMG is commonly used to control prosthetic devices, while in most industrial devices the utilized force is estimated proportionally to muscle activity [31]. Further studies are needed to define the EMG-time relationship in terms of these three sensor placement locations during static and dynamic contractions. Such studies must include different anthropometric parameters variations (e.g., age, sex, and muscle thickness) during movements of the upper limbs at different angles and various external conditions that influence the signal amplitude (e.g., heat versus cold).

There were a number of limitations in this study with regard to the tests performed. First, the strategies used to normalize the EMG data and sensors throughout the study were not capable of recording the activity from the deep muscle tissues of the BB. Second, all subjects were given verbal instruction during the MVC trials and seated in identical testing positions in order to ensure maximal effort. However, without the use of methods, such as the superimposed burst technique (short or long period), there is no way to determine whether a subject afforded maximal effort during the procedure [32,33].

5. Conclusions

The relationship between EMG signal measured at different anatomical locations on the biceps brachii and force in a cohort of healthy volunteers is examined. This study has defined the situation under which EMG sensor position on the BB muscle is active, less variable, and has a good relationship with force. The results demonstrated that the EMG (amplitude) versus force relationships for the muscle belly ('M') were more linear compared to the other two locations ('L' and 'U'). The study also found that amplitude increases and has less variability in this sensor placement area than the tender points during static contraction. Together, these results suggest that proper surface electrode placement should follow the muscle fiber orientation on the BB muscle, and utmost care must be taken when placing them in the direction of muscle fibers. Our findings can also be applied to biomedical engineering methods for the analysis and control of the neuromuscular system, ergonomics field of research, rehabilitation engineering, and movement biomechanics.

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References

[1] Reaz M, Hussain M, Mohd-Yasin F: Techniques of EMG signal analysis: Detection, processing, classification and applications. Biol. Proced. Online. 2006; 8(1):11–35.
[2] Cooper MA, Herda TJ, Vardiman JP, Gallagher PM, Fry AC: Relationships between skinfold thickness and electromyographic and mechanomyographic amplitude recorded during voluntary and non-voluntary muscle actions. J Electromyogr Kinesiol. 2014; 24(2):207–213.
[3] Behm D, Power K, Drinkwater E: Comparison of interpolation and central activation ratios as measures of muscle inactivation. Muscle Nerve. 2001; 24(7):925–934.
[4] Li W, Sakamoto K: The influence of location of electrode on muscle fiber conduction velocity and emg power spectrum during voluntary isometric contraction measured with surface array electrodes. Applied Human Science. 1996; 15(1):25–32.
[5] Wu X, Hou W, Zheng X, Wang H, Zha M: Relationship between surface EMG and angle of elbow joint. Space Medicine and Medical Engineering. 2007; 20(4):259.
[6] Bouisset S, Maton B: Quantitative relationship between surface EMG and intramuscular electromyographic activity in voluntary movement. Am J Phys Med Rehabil. 1972; 51(6):285–295.
[7] Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G: Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol. 2000; 10(5):361–374.
[8] Sargon MF, Tuncali D, Çelik H: An unusual origin for the accessory head of biceps brachii muscle. Clinic Anat. 1996; 9(3):160–162.
[9] Saitou K, Masuda T, Michikami D, Kojima R, Okada M: Innervation zones of the upper and lower limb muscles estimated by using multichannel surface EMG. Journal of Human Ergology. 2000; 29:35–52.
[10] Farina D, Merletti R, Nazzaro M, Caruso I: Effect of joint angle on EMG variables in leg and thigh muscles. IEEE Eng Med Biol Mag. 2001; 20(6):62–71.
[11] Graves AE, Kornatz KW, Enoka RM: Older adults use a unique strategy to lift inertial loads with the elbow flexor muscles. J Neurophysiol. 2000; 83(4):2030–2039.
[12] Naito A, Yajima M, Fukamachi H, Ushikoshi K, Sun Y, Shimizu Y: Electromyographic (EMG) study of the elbow flexors during supination and pronation of the forearm. The Tohoku Journal of Experimental Medicine. 1995; 175(4):285.
[13] Rainoldi A, Galardi G, Maderna L, Comi G, Lo Conte L, Merletti R: Repeatability of surface EMG variables during voluntary isometric contractions of the biceps brachii muscle. J Electromyogr Kinesiol. 1999; 9(2):105–119.
[14] Karlsson S, Gerdle B: Mean frequency and signal amplitude of the surface EMG of the quadriceps muscles increase with increasing torque – a study using the continuous wavelet transform. J Electromyogr Kinesiol. 2001; 11(2):131–140.

[15] Dolan P, Adams M: The relationship between EMG activity and extensor moment generation in the erector spinae muscles during bending and lifting activities. J Biomech. 1993; 26(4):513–522.

[16] Wang L, Buchanan TS: Prediction of joint moments using a neural network model of muscle activations from EMG signals. IEEE Trans Neural Syst Rehabil Eng. 2002; 10(1):30–37.

[17] Rantalainen T, Kłodowski A, Piitulainen H: Effect of innervation zones in estimating biceps brachii force–EMG relationship during isometric contraction. J Electromyogr Kinesiol. 2012; 22(1):80–87.

[18] Doheny EP, Lowery MM, FitzPatrick DP, O'Malley MJ: Effect of elbow joint angle on force–EMG relationships in human elbow flexor and extensor muscles. J Electromyogr Kinesiol. 2008; 18(5):760–770.

[19] Roll J, Vedel J: Kinaesthetic role of muscle afferents in man, studied by tendon vibration and microneurography. Exp Brain Res. 1982; 47(2):177–190.

[20] Schuurmans J, de Vlugt E, Schouten AC, Meskers CG, de Groot JH, van der Helm FC: The monosynaptic Ia afferent pathway can largely explain the stretch duration effect of the long latency M2 response. Exp Brain Res. 2009; 193(4):491–500.

[21] Beck TW, Housh TJ, Johnson GO, Weir JP, Cramer JT, Coburn JW, Malek MH: The effects of interelectrode distance on electromyographic amplitude and mean power frequency during isokinetic and isometric muscle actions of the biceps brachii. J Electromyogr Kinesiol. 2005; 15(5):482–495.

[22] Dalton P, Stokes M: Acoustic myography reflects force changes during dynamic concentric and eccentric contractions of the human biceps brachii muscle. European Journal of Applied Physiology and Occupational Physiology. 1991; 63(6):412–416.

[23] Leedham J, Dowling J: Force-length, torque-angle and EMG-joint angle relationships of the human in vivo biceps brachii. European Journal of Applied Physiology and Occupational Physiology. 1995; 70(5):421–426.

[24] Dimitrova N, Arabadzhiev T, Hogrel J, Dimitrov G: Fatigue analysis of interference EMG signals obtained from biceps brachii during isometric voluntary contraction at various force levels. J Electromyogr Kinesiol. 2009; 19(2):25.

[25] Zuniga J, Housh T, Camic C, Hendrix C, Mielke M, Schmidt R, Johnson G: The effects of parallel versus perpendicular electrode orientations on EMG amplitude and mean power frequency from the biceps brachii. Electromyography and Clinical Neurophysiology. 2010; 50(2):87–96.

[26] Troiano A, Naddeo F, Sosso E, Camarota G, Merletti R, Mesin L: Assessment of force and fatigue in isometric contractions of the upper trapezius muscle by surface EMG signal and perceived exertion scale. Gait Posture. 2008; 28(2):179–186.

[27] De Luca CJ, Kuznetsov M, Gilmore LD, Roy SH: Inter-electrode spacing of surface EMG sensors: Reduction of crosstalk contamination during voluntary contractions. J Biomech. 2012; 45(3):555–561.

[28] Wakeling JM: The recruitment of different compartments within a muscle depends on the mechanics of the movement. Biol Lett. 2009; 5(1):30–34.

[29] Cerqueira LS, Carvalho JF, Pompeu F: Electromyography of the biceps brachii during dynamic contractions. Rev Bras Med Esporte. 2013; 19(6):457–461.

[30] De Luca CJ, Hostage EC: Relationship between firing rate and recruitment threshold of motoneurons in voluntary isometric contractions. J Neurophysiol. 2010; 104(2):1034–1046.

[31] Kamavuako EN, Rosenvang JC, Bøg MF, Smidstrup A, Erkocevic E, Niemeier MJ, Jensen W, Farina D: Influence of the feature space on the estimation of hand grasping force from intramuscular EMG. Biomed Signal Process Control. 2013; 8(1):1–5.

[32] Kent-Braun JA, Le Blanc R: Quantitation of central activation failure during maximal voluntary contractions in humans. Muscle Nerve. 1996; 19(7):861–869.

[33] Norcross MF, Troy Blackburn J, Goerger BM: Reliability and interpretation of single leg stance and maximum voluntary isometric contraction methods of electromyography normalization. J Electromyogr Kinesiol. 2010; 20(3):420–425.