383. RESEARCH OF RELATION BETWEEN MUSCLE BIOSIGNAL AND SYSTOLIC BLOOD PRESSURE, AND APPLICATION OF ITS CHARACTERISTICS FOR EVALUATION OF EFFICIENCY

M. Mariūnas, K. Daunoravičienė, J. Griškevičius, J. Andrašiūtė
Vilnius Gediminas technical university
J. Basanavičius str. 28 a, Vilnius, LT - 03224, Lithuania
E-mail: mariunas@me.vgtu.lt, daunoraviciene@me.vgtu.lt,
julius.griskevicius@me.vgtu.lt, jurgaandrasiute@gmail.com
Tel.: +370 6 52745015, +370 6 52744748

(Received 28 June 2008, accepted 20 August 2008)

Abstract. Paper presents analytical relations and methodology for muscle efficiency evaluation in the case of limited information. The relation between systolic blood pressure and amplitude of muscle biosignal has been derived, and characteristics of this dependency have been analyzed. When the loading is large the muscle biosignal amplitude could not be unambiguously determined according to systolic blood pressure. Features of the relation between muscle biosignal and systolic blood pressure have demonstrated the capability of evaluation of human adaptability to set load and to watch for rates in rehabilitation process.

Keywords: biosignal, systolic blood pressure, efficiency, relation, load, loading duration, fatigue, muscle, adaptation period

Introduction

Efficiency is the quality of being efficient, which means performing in the best possible and the least wasteful manner. Reasons of decline in organism efficiency are caused by changes in other organism systems and are always concrete e.g. they depend on character of performed work, intensity, mode of muscle work, blood pressure, heard rate and other factors. Mostly, efficiency reduces during physical activity due to fatigue in the CNS, endocrines system, and vegetal systems and in working muscles. The main factor of diminishing human physical activity is muscular fatigue [1]. Muscular fatigue is defined as any exercise-indicated reduction in the capacity of muscles to generate maximal force or power output. Muscle soreness, stiffness, and pain are frequently associated with fatigue. Notable physiological and biomechanical changes take place during the fatigue period leading up to complete task failure. However, objective criteria to quantify fatigue-related phenomena are limited. Physiological fatigue is characterized by complex assortment and integration of numerous competing and parallel systems that interact nonlinearly [7, 9–24]. A lot of studies are dedicated to low-frequency fatigue. It was concluded that after exercises of the maximal intensity there was a smaller fatigue of low-frequency in the muscles but there was a greater decline in muscle endurance after exercises. This shows that metabolic fatigue caused by performing the exercise of maximal intensity can partially compensate for the decrease in the muscle contraction force at low-frequency fatigue but it is detrimental to muscle endurance [26, 27]. Several physiological studies, methods and techniques are applied for clarifying hypothesis of different muscles efficiency stages [25–27]. Other factors of person's efficiency decrease are blood pressure and heart rate during work or exercises. Recent studies have resulted in different experimental results obtained with healthy people and ones with diseases [2–6]. Blood pressure rates are usually caused by person's physical activity. Registered values of blood pressure after exercises increase and often are close to heart rate ascension [28]. Electromiographical (EMG) data collected during physical activity and blood pressure rates are very coherent and their values vary in a different manner due to experimental circumstances [2–6, 8, 28]. Unfortunately, scientists have not reached common agreement on relation between EMG and blood pressure rates during person's work. And the influence of physical activity on these rates is not sufficiently investigated.

Human efficiency could be evaluated by using many parameters and methods. Results of research of efficiency are very useful for variety of fields such as rehabilitation, sport biomechanics, ergonomics, etc., and have wide application.

The purpose of this work is to define the relation between muscle biosignal and systolic blood pressure as well as apply its characteristic features in human efficiency evaluation.

Methods

Electromyography system "*Viking Quest*" has been used for experimental research. Disposable Ag/AgCl electrodes were used in the present study to record EMG signals. The EMG signals were bandpass filtered and differentially amplified. All EMG signals were smoothed and analyzed by the *MATLAB 7.1* software [29–31].

Systolic blood pressure was measured by metrologically tested and accurately calibrated to 20 mm Hg–300 mm Hg gauge sphygmomanometer.

Main research have been made on brachioradialis muscle (*m. brachioradialis*), which lies in the lateral side of forearm. For experimental studies and for ensuring of their accuracy, ten health subjects – five females (from 24 to 30 years) and five males (from 28 to 32 years) have been invited. The investigation has been focused on five load groups: from 8 N to 24 N. The muscle has been subjected to loading with different duration times from 5 up to 14 min. EMG activity was recorded in the muscle as the subject was holding the static load in the extended arm. Systolic blood pressure and the pulse were measured at the same time.

Analysis and results

Muscle is getting tired while performing particular work in a time-span. Under action of constant load, the work done by the muscle can be expressed as:

$$A = \int_0^{T_i} \frac{U^2(F_i,t)}{R}\,dt\,, \tag{1}$$

where $U(F_i,t)$ is an amplitude of muscle biosignal, μV, which corresponds the force F_i in time, F_i is muscle loading force N, t is the time, s, R represents muscle resistance.

Neglecting the energy loss in the muscle during its work, it can be stated that muscle performs the same work when it is loaded with different magnitude loads. Thereby, following equalities can be written:

$$\int_0^{T_1} \frac{U^2(F_1,t)}{R}\,dt = \int_0^{T_2} \frac{U^2(F_2,t)}{R}\,dt = \ldots = \int_0^{T_i} \frac{U^2(F_i,t)}{R}\,dt = \ldots = \int_0^{T_n} \frac{U^2(F_n,t)}{R}\,dt, \tag{2}$$

where T_i stands for working duration of the muscle loaded with the i-th force until it gets tired, in other words, it is a reserve of muscle work for the i-th force of muscle loading or the value of T_i, that corresponds the point $U^2(F_i, t)$ of the function $\partial U^2(F_i,t)/\partial t = 0$.

Since function $U(F_i,t)$ is complicated and it is difficult to form its accurate mathematical expression, it is common to obtain it experimentally in the graphical manner. Therefore approximate calculation of the function integral can be expressed as follows:

$$\int_0^{T_i} \frac{U^2(F_i,t)}{R}\,dt \approx \frac{1}{2}\Delta t \sum_{i=2}^{m_i}\left[\frac{U^2(F_{i+1},t)}{R_{i+1}} + \frac{U^2(F_i,t)}{R_i}\right], \tag{3}$$

where Δt is a discretization step of the time coordinate, m_i represents a number of discretization steps for the i-th force of muscle loading, corresponding the beginning of muscle fatigue.

Multiplying and dividing right side of (3) by m_i and after simplifications we get:

$$\int_0^{T_i} \frac{U^2(F_i,t)}{R}\,dt \approx \frac{1}{2m_i} \sum_{i=2}^{m_i}\left[\frac{U^2(F_{i+1},t)}{R_{i+1}} + \frac{U^2(F_i,t)}{R_i}\right] T_i\,. \tag{4}$$

In case if T_k at the k-th moment is known and we can calculate the value of $\int_0^{T_k} \frac{U^2(F_k,t)}{R}\,dt$, then the reserve of T_i muscle efficiency, as muscle is loaded with F_i force, can be estimated from the expression:

$$\frac{1}{2m_i} \sum_{i=2}^{m_i}\left[\frac{U^2(F_{i+1},t)}{R_{i+1}} + \frac{U^2(F_i,t)}{R_i}\right] T_i = \\ = \frac{1}{2m_k} \sum_{k=2}^{m_k}\left[\frac{U^2(F_{k+1},t)}{R_{k+1}} + \frac{U^2(F_k,t)}{R_k}\right] T_k\,, \tag{5}$$

or

$$T_k = \frac{m_k \sum_{i=2}^{m_i}\left[\frac{U^2(F_{i+1},t)}{R_{i+1}} + \frac{U^2(F_i,t)}{R_i}\right]}{m_i \sum_{k=2}^{m_k}\left[\frac{U^2(F_{k+1},t)}{R_{k+1}} + \frac{U^2(F_k,t)}{R_k}\right]} \cdot T_i\,, \tag{6}$$

or, when R = const:

$$T_k = \frac{m_k \sum_{i=2}^{m_i}\left[U^2(F_{i+1},t) + U^2(F_i,t)\right]}{m_i \sum_{k=2}^{m_k}\left[U^2(F_{k+1},t) + U^2(F_k,t)\right]} \cdot T_i\,. \tag{7}$$

Equation (6) provides the relation of efficiency duration to magnitudes of the i–th and the k–th muscle loading forces. The latter expression could be solved with the predefined accuracy.

Analyzing the dependence of muscle biosignal *U(F, t)* in the male and female groups on the loading force magnitude F_i (Fig. 1 and Fig. 2), it was noted that when t □ T_i, i =1, 2, ..., n, the dependence is approximately linear.

In the female group it is distinctive that the amplitude of muscle biosignal is increasing more than in the male group (Fig. 1 and Fig. 2).

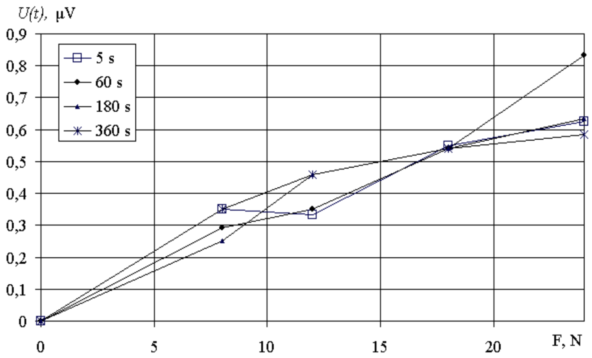

Fig. 1. Muscle biosignal dependence on a loading magnitude in male group

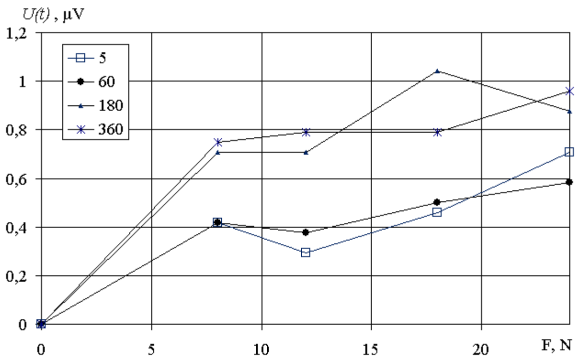

Fig. 2. Muscle biosignal dependence on a loading magnitude in female group

Relation between the muscle biosignal amplitude and the loading duration as the load is 12 N is presented in Fig. 3 and it corresponds to $U^2(F, t)$ given in Fig. 4. The time function of a squared muscle biosignal U^2 (Fig. 4) is complicated and it can be expressed as an exponential function. However, for approximate calculations, when $t \leq T_i$, the function $U(t)$ can be simplified and considered as linear. For example, the area under the $U^2(F, t)$ curve represents the required amount of energy or work for the muscle during the loading time (Fig. 4).

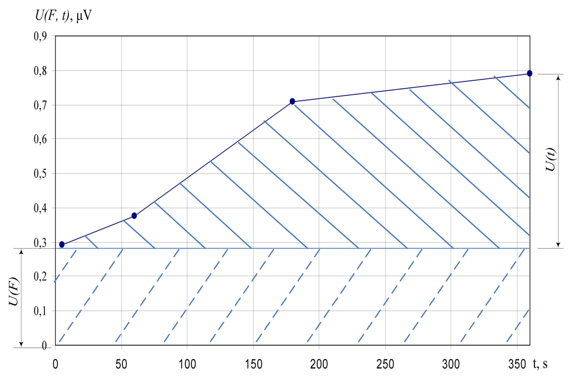

Fig. 3. Relation between muscle biosignal amplitude and loading duration as the load is 12 N

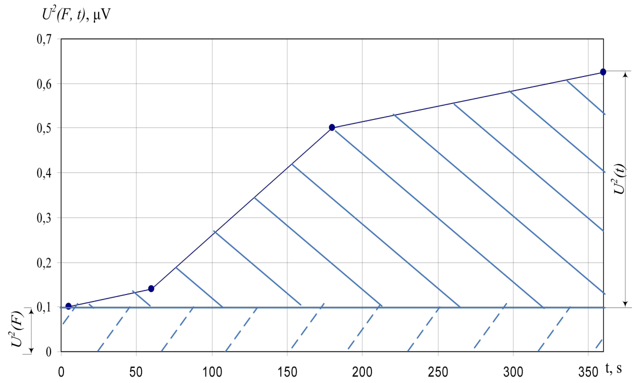

Fig. 4. Dependence between U^2 and loading duration as the load is 12 N

The value of integral $\int_0^{T_i} U^2(F_i,t)$ could be calculated approximately:

$$\frac{1}{2m_i}\sum_{i=2}^{m_i}\left[U^2(F_{i+1},t)+U^2(F_i,t)\right]T_i \approx$$

$$\approx U^2(F)\cdot t+\frac{1}{2}U^2(t)\cdot t \,. \qquad (8)$$

Analyzing results from Figs. 3-4 it can be noted that $U^2(t)$ could be expressed as follows:

$$U^2(t)=\left[U(F)+U(t)\right]^2-U^2\left(F\right)=$$

$$=2U(F)U(t)+U^2(t)\,. \qquad (9)$$

Inserting (9) into (8) equation, we obtain:

$$\int_0^{T_i} U^2(F_i,t)=U^2(F)\cdot t+U(F)U(t)\cdot t+\frac{1}{2}U^2(t)\cdot t\,, \qquad (10)$$

where $U(F)$ is a value of the biosignal amplitude, when $t = 0$, over different magnitudes of the loading force F_i, i=1, 2,…, n; $U(t)$ is a value of the biosignal amplitude, when $t = T_i$, during the action of F_i minus $U(F_i)$, that is $U(t)=U(T_i)-U(F_i)$.

Proposing that the dependence of the biosignal magnitude for male and female groups could be expressed linearly in the small interval of $0 \leq F \leq F_{\max}$:

$$U_v(F)=a_vF\,,\; U_m(F)=a_mF\,. \qquad (11)$$

Then $U^2(F)$ would be calculated as follows:

$$U_v^2(F)\approx\left(a_vF\right)^2,\; U_m^2(F)\approx\left(a_mF\right)^2. \qquad (12)$$

By expressing the biosignal dependence on time t for male and female groups analogically we would get following approximate dependencies:

$$U_v(t) = b_v t\ ,\ U_m(t) = b_m t\ , \tag{13}$$

where $U_v(F)$, $U_v(t)$, $U_m(F)$, $U_m(t)$ is a value of the biosignal amplitude for male and female accordingly.

Inserting (12) and (13) into (10) dependence for the male group, when $t = T_i$ we get:

$$\int_0^{T_i} U^2(F_i,t) = a_v^2 F^2 \cdot T_i + a_v F b_v T_i^2 + \frac{1}{2} b_v^2 T_i^3 = cT_k\ ,$$

or

$$T_i^3 + \frac{2a_v}{b_v} FT_i^2 + \frac{2a_v^2}{b_v^2} F^2 T_i - \frac{2c}{b_v^2} T_k = 0\ , \tag{14}$$

when $$c = \frac{1}{2m_k} \sum_{k=2}^{m_k} \left[U^2(F_{k+1},t) + U^2(F_k,t) \right]. \tag{15}$$

For the female group the dependence could be rewritten analogically to (14):

$$T_i^3 + \frac{2a_m}{b_m} FT_i^2 + \frac{2a_m^2}{b_m^2} F^2 T_i - \frac{2c}{b_m^2} T_k = 0\ . \tag{16}$$

Taking, that the values of parameter c and T_k are known, for the male and female groups from (14) and (16) expressions by solving the third order equations we get three roots.

Solutions will be good only for a real numbers and must satisfy the condition:

$$T_{i,l} > 0\ ,\ l = 1,\ 2,\ 3. \tag{17}$$

At all events, when $T_{i,l} > 0$ and real numbers, $\max\limits_{l}\left[T_{i,l} \right]$ would be the indicator of muscle working reserve.

Results presented in Figs. 1–2 for the male and female groups where processed using the least squares method and for the considered case were estimated following coefficients: $a_v \approx 29{,}17$, $a_m \approx 26{,}25$, $b_v \approx 0{,}36$ and $b_m \approx 0{,}83$.

Dependences of biosignal amplitude and the systolic blood pressure on the loading duration were experimentally determined, when the loads were F = 8, 12, 18 and 24 N (Fig. 5 and 6).

According to the Figs. 5–6 results, the dependence of muscle biosignal on the systolic blood pressure was derived and represented in the Fig. 7.

Analyzing together Figs. 5–7, it was noted, that when the loads are F = 8 and 24 N, in the dependence of biosignal amplitude and systolic blood pressure there exists a point T_k = 600 s, where $\partial U(F,t)/\partial t = 0$ and $\partial Hg(F,t)/\partial t = 0$. In Fig. 7 it is furthermost from the origin and touches an ordinate. First points of maximal values of broken curves represents the adaptation period.

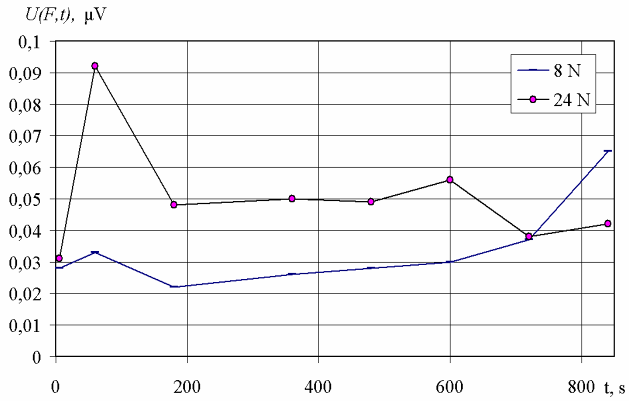

Fig. 5. Dependence of biosignal amplitude on loading duration

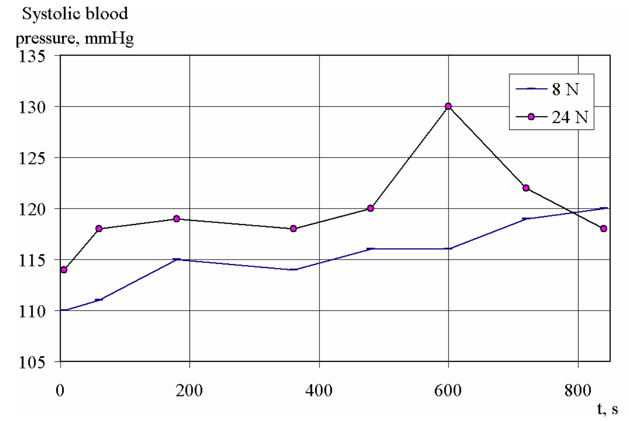

Fig. 6. Variation of systolic blood pressure values in loading duration

When the load is F = 8 N, neither the biosignal amplitude nor systolic blood pressure reaches maximum point T_i, where $\partial U(F,t)/\partial t = 0$ and $\partial Hg(F,t)/\partial t = 0$. Therefore curve 1 has not got the second maximum, and the first peak represents the adaptability of the muscle and the systolic blood pressure to a given load period. In present graphs time coordinates are not discriminate. Using expressions (16) and (15) T_i was calculated value, when F = 8 N and it was estimated that only the first root is real $T_{i,1} > 0$ and it equals to $T_{i,1} = 883{,}5$ s, while the other two roots are complex. Figs. 5–6 also indicate that when the loading time is approaching to $t \to 800$ s, values of both, biosignal amplitude and systolic blood pressure, approach to the point, where $\partial U(F,t)/\partial t = 0$ and $\partial Hg(F,t)/\partial t = 0$. Presented approximate calculation methodology for the efficiency reserve, in the case when there is a lack of information, can be applied practically. However, each point of the curves in presented figures is matched by particular time interval, which can be easily estimated using Figs 5–6. In Fig. 7 occurring small loops of curves or similar formations arise from the inaccuracy of characteristics during the experiments. There are zones in Fig. 7, where multiple values of systolic blood pressure correspond to one value of magnitude of biosignal. It can be observed also that when the loading force on the muscle is increasing, so does zones of the biosignal ambiguity dependence on systolic blood pressure (zone, where F = 8 N, 1 curve and zone, where F = 24 N, 2 curve).

Analyzing results in Fig. 7 as well as other obtained curves, we can point out the following features:

- it is characteristic that the less area is under the curve the better muscle adapts to the selected loading;
- when the muscle is loaded with large loads or the man has a specific pathology, there exists an ambiguity between systolic blood pressure and muscle biosignals, whereas when the loads are small, the dependence between the magnitude of biosignal and the systolic blood pressure is always unambiguous.

Performed research indicates that the beginning of human fatigue or his efficiency reserve can be estimated using features of muscle biosignal and systolic blood pressure, applying attribute $\partial U^2(F_i,t)/\partial t = 0$ or $\partial Hg(F_i,t)/\partial t = 0$. However, for the estimation of systolic blood pressure dependence on the loading force and time, integral quantity is measured, its magnitude at the specific time moment is expressed by one number and not the spectrum as in case of muscle biosignal. Therefore, in estimating magnitude of systolic blood pressure we obtain much less information about the phenomena in the muscle and organism.

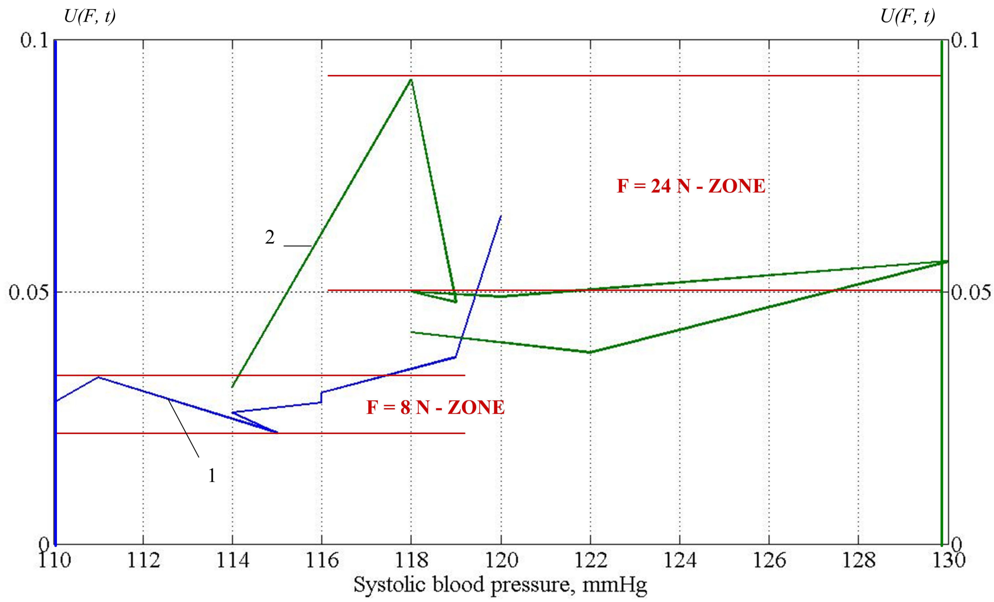

Fig. 7. Dependence of muscle biosignal $U(F_i,t)$ on the systolic blood pressure at different magnitudes of force

Conclusions

Both analytical and experimental research results were provided in this paper. Mathematical formulation and methodology was presented for the estimation of the efficiency in male and female groups, when there is a lack of information. Interdependences of muscle biosignal and systolic blood pressure were estimated and they demonstrated the following:

- when the human organism well adapts to selected loads, the magnitude of the muscle biosignal in male and female groups could be estimated unambiguously using the systolic blood pressure;
- ambiguity arises, when muscle loads are large and long lasting, that is the magnitude of biosignal could not be estimated definitely from the values of systolic blood pressure;
- when the dependence graph of muscle biosignal on the systolic blood pressure covers the bigger area, it indicates that the load is too large for a person or a person has a pathology. The less is the area of the presented dependence curve, the more man or woman is physically ready to perform an activity;

Proposed methodology can be applied in testing athletes or in process of rehabilitation and monitoring the change of health and efficiency indicators.

References

[1] **Liu Y., Kankaanpaa M., Zbilut J. Z., Webber Ch. L.** EMG recurrence quantifications in dynamic exercise. Jornal: Biological Cybernetics. 90, 2004, p. 337–348.

[2] **Gheorghiade M., William T. and others.** Systolic Blood Pressure at Admission, Clinical Characteristics and outcomes in Patients Hospitailized with Acute Heart Failure. The Journal of American Medical Association. Vol. 296, No 18, 2006, p. 2217–2226.

[3] **Dae Hyun Kim and Hyuseok Kang.** Systolic Blood Pressure and Outcomes in Patients Hospitailized with Acute Heart Failure. The journal of the American Medical Association. Vol. 297, No 20, 2007, p. 807–808.

[4] **Kurl S., Laukkanen J. A., Rauramaa R., Lakka T.A. and others.** Systolic Blood Pressure Response to Exercise Stress Test and Risk of Stroke. The Journal of American Heart Association. No 32, 2001, p. 129–139.

[5] **Michael F., Rourke O.** Vascular mechanins in the clinics. Journal of Biomechanics Engineering. 36, 2003, p. 623–630.

[6] **Finet G., Ohayen J., Roufal G.** Biomechanical Interaction between Cap Vulnerable Coronary Arery Disease. Journal of Biomechnaics. Vol. 15, No 1, 2004, p. 145–150.

[7] **Knafiltz M., Molineri F.** Assesment of Muscular Fatique During Biking. IEEE transactions on neural systems and rehabilitaion engineering. Vol. 11, No 1, 2003, p. 17–23.

[8] **Vince Stanford.** Biosignals Offer Potential for Direct Interfaces and Health Monitoring. IEEE Compiuter Society. Vol. 3, No 1, 2004, p. 99–103.

[9] **Ebersole Kyle T., O'Connor Kristian M.** Electro-mechanical Efficiency of the Superficial Quadriceps Femoris Muscles during Maximal Concentric Muscle Actions. Medicine & Science in Sports & Exercise: Vol. 37(5), 2005, p. S441–S442.

[10] **Ricard Mark D., Hills-Meyer Patrick, Miller Michael G. and Michael Timothy J.** The effects of bicycle frame geometry on muscle activation and power during a wingate anaerobic test. Journal of Sports Science and Medicine. Vol. 5, 2006, p. 25–32.

[11] **Firas Massaad, Thierry M. Lejeune and Christine Detrembleur.** The up and down bobbing of human walking: a compromise between muscle work and efficiency. Journal of Physiology. Vol. 582.2 2007. p. 789–799.

[12] **Tal'nov A. N., Cherkassky V. L., Kosyukov A. I.** Movement-related and steady-state electromyographic activity of human elbow flexors in slow transition movements between two equilibrium states. Neuroscience. Vol. 79, Issue 3, 26 May 1997, p. 923–933.

[13] **Howard J. D., Hoit J. D., Enoka R. M., Hasan Z.** Relative activation of two human elbow flexors under isometric conditions: a cautionary note concerning flexor equivalence. Experimental Brain Research. Vol. 62, Number 1, 1986, p. 199–202.

[14] **Rudroff Thorsten, Didier Staudenmann, Roger M. Enoka.** Electromyographic measures of muscle activation and changes in muscle architecture of human elbow flexors during fatiguing contractions. J. Appl. Physiology. Vol. 104, 20 March 2008, p. 1720–1726.

[15] **Serrao M.**, **Pierelli F., Don R., Ranavalo A., Cacchio A., Curra A., Sandrini G., Frascarelli M., Santilli V.** Kinematic and Electromyographic study of the nociceptive withdrawal reflex in upper limbs during rest and movement. The Journal of Neuroscience. Vol. 26(13), 29 March 2006, p. 3505–3513.

[16] **Naito A.**, Yajma M., Fukamachi H., Usnikoshi K., Sun Y., Shimizu Y. Electromyographic (EMG) study of the elbow flexors during supination and pronation of the forearm. J. Exp. Med. Vol. 176, 1995, p. 285–288.

[17] **Hubal Monica J., Rubinstein Scott R., Clarkson Priscilla M.** Mechanisms of Variability in Strength Loss after Muscle-Lengthening Actions. Medicine & Science in Sports & Exercise. Vol. 39(3), March 2007, p. 461–468.

[18] **Fabio Esposito, Emiliano Cè, Massimiliano Gobbo, Arsenio Veicsteinas and Claudio Orizio.** Surface EMG and mechanomyogram disclose isokinetic training effects on quadriceps muscle in elderly people. European Journal Applied Physiology. 2005, Vol. 94, p. 549–557.

[19] **Kyle T Ebersole, David M Malek.** Fatigue and the Electromechanical Efficiency of the Vastus Medialis and Vastus Lateralis Muscles. Journal of athletic training. 2008, 43(2). p. 152–156.

[20] **Alenka Maček Lebar, Tomaž Mencinger, Damijan Miklavčič.** Surface EMG as a method for following-up sports training efficiency. Acta Univ. Palacki. Olomuc., Gymn. 2005, Vol. 35, No. 1, p. 27–34.

[21] **Iridiastadi H., Nussbaum M.** Muscle fatigue and endurance during repetitive intermittent static efforts: development of prediction models. Ergonomics, Volume 49, Number 4, 15 March 2006, p. 344–360.

[22] **E. B. Swallow, H. R. Gosker, K. A. Ward, A. J. Moore, M. J. Dayer, N. S. Hopkinson, A. M. W. J. Schols, J. Moxham, and M. I. Polkey.** A novel technique for nonvolitional assessment of quadriceps muscle endurance in humans. Journal of Applied Physiology. Vol. 103, 2007, p. 739–746,

[23] **R. R. Neptune, M. L. Hull.** A theoretical analysis of preferred pedaling rate selection in endurance cycling. Journal of Biomechanics 32 (1999), p. 4009–415.

[24] **Prem. Bajaj, Thomas Graven-Nielsen, Lars Arendt-Nielsen.** A Psychophysical Study of Muscle Fatigue and post-exercise muscle soreness of first dorsal interosseous muscle of hand.

[25] **Mihai T. Tarata.** Mechanomyography versus Electromyography, in monitoring the muscular fatigue. Biomedical Engineering Online, 2003, p. 1–10.

[26] **Skurvydas A.** Low frequency fatigue and muscle endurance after performing intermittent eccentric exercise and continuous eccentric-concentric exercise. Ugdymas, Kūno kultūra, Sportas, No. 4(37), 2000, p. 46–50.

[27] **Linstedt S. L., LaStayo P. C., Reich T. E.** When Active Muscle Lengthen: properties and consequences of eccentric contractions. News of Physiology science, Vol. 16 (2001), p. 256–261.

[28] **Mariūnas M., Kuzborska Z.** Research into the heart rate and blood pressure dependence on loading and time. Journal of Vibroengineering. Vol. 9, No3, 2007, p. 51–54.

[29] **Mariūnas M., Daunoravičienė K.** Determination and analysis of muscular endurance parameters. Journal of Vibroengineering, Vol. 8, No. 2, 2006, ISSN 1392-8716, p. 1–6.

[30] **Mariūnas M., Kojelytė K.** Investigation the relationship of muscle mechanical characteristics with Biosignal energy. Journal: Solid State Phenomena. Vol. 113, 2006, p. 157–163.

[31] **Mariūnas M., Daunoravičienė K.** Research of muscle endurance characteristics. Mechanika w Medycinie 8, Rzeszow, 2006, p. 135–141.