Applied and Computational Mechanics 5 (2011) 67–76

Use of electromyography measurement in human body modeling

L. Valdmanová[a], H. Čechová[b,*]

[a]Department of Mechanics, Faculty of Applied Sciences, University of West Bohemia, Univerzitní 8, 306 14 Plzeň, Czech Republic
[b]New Technologies — Research Centre, University of West Bohemia, Univerzitní 8, 306 14 Plzeň, Czech Republic

Received 1 December 2010; received in revised form 23 May 2011

Abstract

The aim of this study is to test the use of the human body model for the muscle activity computation. This paper shows the comparison of measured and simulated muscle activities. Muscle active states of biceps brachia muscle are monitored by method called electromyography (EMG) in a given position and for given subsequently increasing loads. The same conditions are used for simulation using a human body model (Hynčík, L., Rigid Body Based Human Model for Crash Test Purposes, Engineering Mechanics, 5 (8) (2001) 1–6). This model consists of rigid body segments connected by kinematic joints and involves all major muscle bunches. Biceps brachia active states are evaluated by a special muscle balance solver. Obtained simulation results show the acceptable correlation with the experimental results. The analysis shows that the validation procedure of muscle activities determination is usable.

Keywords: Hill-type model, muscle activity, EMG measurement

1. Introduction

Currently safety studies play one of the main roles in the automobile industry development. The crash tests usually performed with standard crash test dummies are very expensive in general and moreover dummies are not able to simulate the "real" life issues as vein pressure, tissues ruptures, muscles pre-stress etc. However experiments with biological tissues or even full scale cadavers are governed by ethical issues. Hence human body models become a powerful tool for a human-friendly and safe vehicles computer aided design. Recently automobile industry focuses on human body models which become more biofidelic as the computing techniques develop.

Previous works published by many authors described the influence of the pre-crash muscular activity on the impact for various body regions, i.e. [6].

The main aim of the study is to present a procedure of validation of a specific human body model using an electromyography (EMG) measurement. In particular the study determines the biceps brachia muscle forces of a previously developed human body model [4] and compares them to experimental results. For this purpose the static test with isometrically contracting biceps brachia muscle is performed. Measured EMG responses are compared to the muscle activity determined by a muscular balance solver [5]. Simulations are done in the computational environment of the PAM software [13]. The comparison of the obtained results of the simulation to the experiment is discussed.

*Corresponding author. Tel.: +420 377 634 710, e-mail: hcechov@ntc.zcu.cz.

2. Experiment

2.1. EMG theoretical base

EMG is the clinical measuring method based on the analysis of the electrical signal closely associated with muscle activations. The EMG is related to tension in such a way that the amplitude of the signal grows with the force generated by the muscle [14]. Muscle contraction is a muscle response to the electrical potential changes. Muscle fibers contract when the action potentials of the motor nerve achieve a depolarization threshold. This action generates an electromagnetic field and the potential is measured as the voltage [7].

The EMG signal is usually measured with two kinds of electrodes — surface or fine wire. Measuring by surface electrodes enables global examination of muscular electrical activity while fine wire electrodes monitor the signal of only several muscle fibers close to the electrode. Advantages of the surface electrodes are painless, easy application and a global response of a monitored muscle. Hence surface electrodes are used for the presented experiment.

2.2. The experimental setup

The static test is performed by measuring the EMG activity on the biceps brachia muscle when carrying a given load. The biceps brachia muscle is the biggest muscle of the upper arm. It starts with two heads (caput longum and caput breve) on the scapula near by the limb and finish by the main tendon on the radius. The main function of this muscle is the elbow flexion and it also contributes to the elbow supination [10]. It is situated nearly under the skin so it could be very clearly touched. Therefore, it is suitable for the monitoring using surface electrodes.

Fig. 1 shows the test setup. The biceps brachia is not the main elbow flexor in this position, however it is a large surface muscle suitable for surface EMG monitoring without cross talks. The volunteer holds his right hand flexed in the elbow such that the upper arm and the forearm form the right angle. The palm is turned to the body. This very simple position is chosen to eliminate possible inaccuracies.

The weight is hung in the supposed palm center of gravity to eliminate voluntary muscle contractions caused by holding the load. Such load placement is chosen with regard to following simulation since each rigid body segment of the model can by loaded only in its gravity centre.

Fig. 1. EMG measurement, signal recording while carrying imposed load

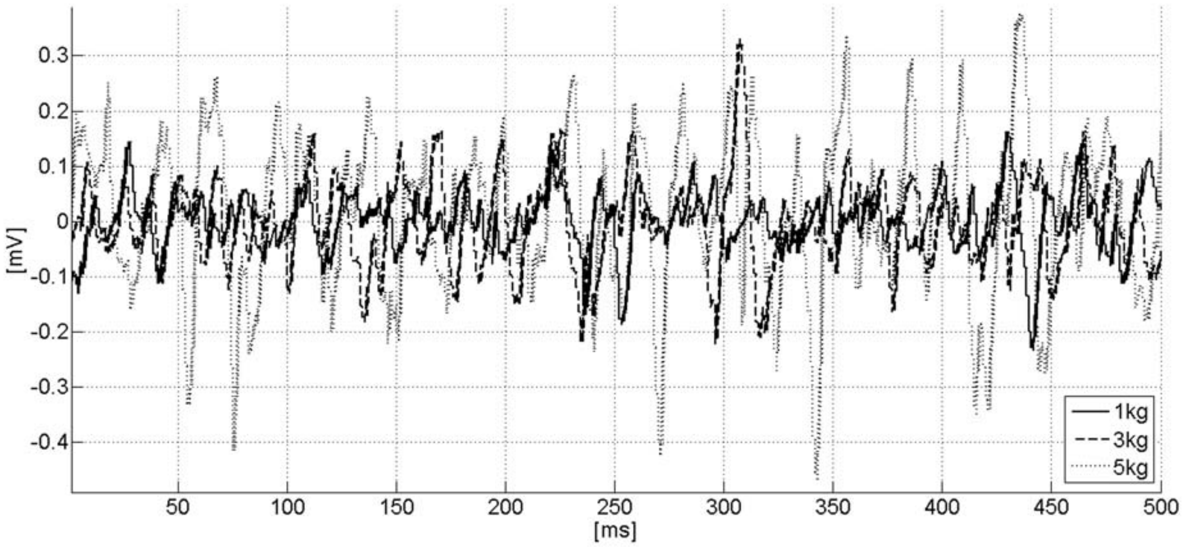

Fig. 2. Raw EMG signal; comparison of EMG biceps brachia response to different loads

The electrodes are stuck on the skin and the EMG signal of isometric muscle contraction for various loads is monitored. The active electrode is situated upon the muscle bunch, reference electrode upon the tendon. The third electrode supports the grounding. The resulting EMG signal is the difference between the active and reference electrode. The hand is subsequently loaded by the weights from 0.5 kg to 5 kg stepwise increased with the increment 0.5 kg. Each case is four times repeated with sufficient rest period.

2.3. *Signal processing*

The raw EMG output is the EMG activity-time relation as shown in Fig. 2. In general the EMG amplitude increases with the increasing contraction force and contraction velocity [3]. However there is no clear dependency between the force and EMG amplitude. Presented test analyzes isometric contractions hence the influence of contraction velocity is eliminated. Simultaneously the particular measurements are also isotonic thus the mean value of the EMG signal corresponds to the muscle force generated to keep the given loaded position [14].

Firstly, the raw EMG signal has to be processed. During recording all signals are filtered by the low pass filter with the cut-off frequency of 10 kHz and the high pass filter with the cut-off frequency of 20 Hz as recommended in [14]. To obtain the muscle activity generated during performed contraction the mean value of the full-wave rectified signal is calculated. The full-wave rectification equals the absolute value of EMG usually with a positive polarity. The original EMG has a mean value of zero but the full-wave rectified signal does not cross through zero. Hence, it has an average that fluctuates with the strength of the muscle contraction [14]. The full-wave rectified signals with their mean values are shown in Fig. 3. Described process is applied on each measurement and computed mean values for each load are averaged. The computed average activities of biceps brachia muscle are displayed in Table 1.

Usually the measured EMG signal is normalized by the EMG signal generated by the maximal isometric muscle contraction, see [1], [2] or [3]. The main question now is how to exactly determine the maximal isometric contraction. To avoid inaccuracies and to ensure the measurement reproducibility instead of maximal isometric contraction the quantities of the maximal applied loads are taken. Hence each obtained mean value of the EMG amplitude is normalized

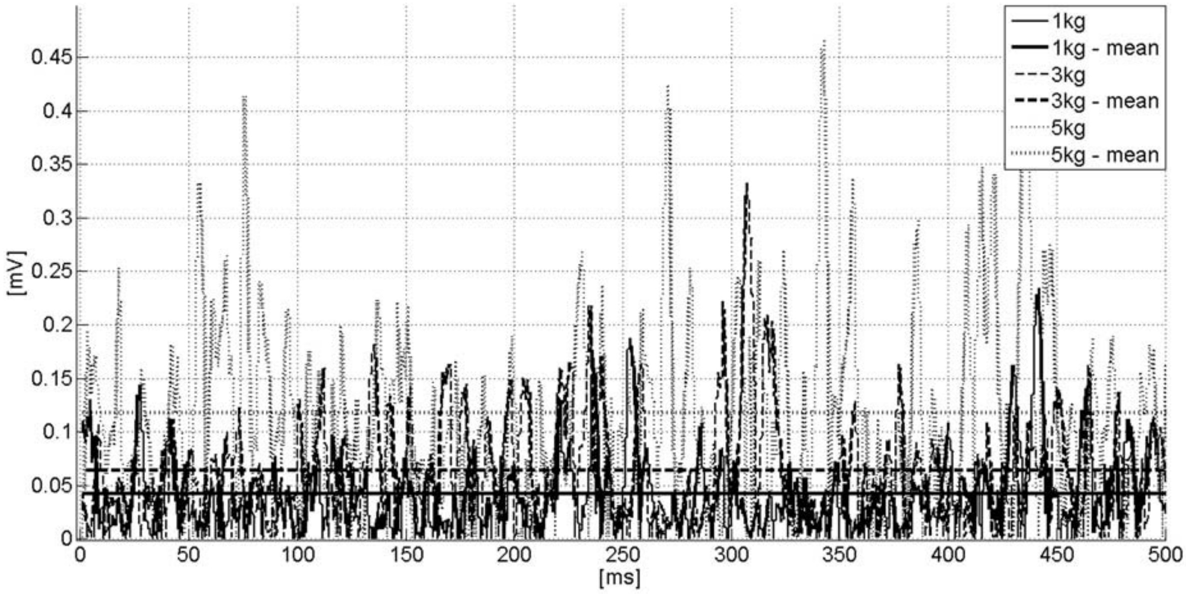

Fig. 3. Full–wave rectified EMG signal with its mean value; comparison for different loads

Table 1. The average EMG values for each load

No.	Load [kg]	Mean EMG values [mV] $mean\_r\_EMG$	
		Average value	**Standard deviation**
1	0.5	0.028 6	0.005 3
2	1.0	0.040 2	0.003 3
3	1.5	0.049 8	0.003 3
4	2.0	0.062 1	0.004 7
5	2.5	0.068 3	0.012 8
6	3.0	0.069 9	0.005 9
7	3.5	0.078 1	0.006 7
8	4.0	0.094 2	0.008 8
9	4.5	0.095 1	0.011 9
10	5.0	0.118 1	0.005 1

by the maximal measured mean value of the EMG amplitude which corresponds to the case of 5 kg load. The same normalization is then applied as well in simulation presented bellow.

3. Computer reconstruction

The experimental test is performed in the environment of the PAM software.

3.1. Human body model

The used rigid body based human model is called Robby and it has been developed during the long time cooperation with the ESI Group company. The Robby's development and validation

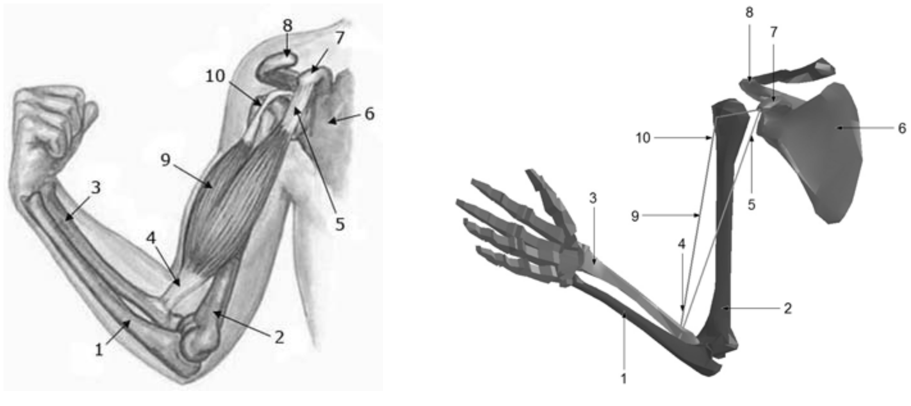

Fig. 4. The biceps brachia muscle structure: 1 – ulna, 2 – humerus, 3 – radius, 4 – tendon, 5 – caput breve, 6 – scapula, 7 – coracoid apex, 8 – coracoid process, 9 – biceps, 10 – caput longum

is described in [4]. The model is designed mainly for automotive safety application. This articulated rigid body model represents a 50^{th} percentile human male with anthropometry according to [8]. It is implemented in the PAM software. The rigid body segments are connected by kinematic joints. The human body model accommodates all major muscles. Each skeletal muscle is modeled by bar elements (one or a set) connected to bones in order to perform a correct muscle function. Muscle behavior is defined according to a Hill-type model [16]. Muscle physiological properties as pennation angle, maximal isometric force, and optimal length are set according to published sources [12] and [9]. Separated upper arm model with the biceps brachia muscle is shown in Fig. 4 in comparison with the real anatomical architecture.

3.2. *Skeletal muscle model*

The Hill-type muscle model consists of the active and passive parts (see Fig. 5). The active part is represented by the contractile element (CE), which substitutes the contractile mechanism. The parallel connected passive part is modeled by the visco-elastic element (VE) and substitutes the collagen and elastin network of the muscle tissue.

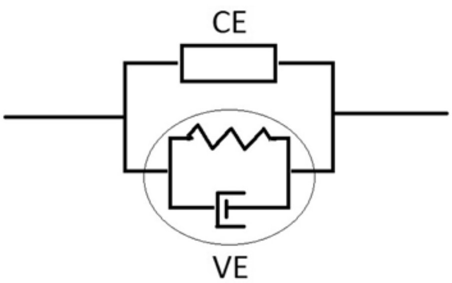

Fig. 5. Schema of Hill-type model

The resultant force of the whole muscle element is computed as the sum of forces of its elements

$$F_{mus} = F_{CE} + F_{VE}, \tag{1}$$

where F_{CE} is the active force of the muscle and F_{VE} is the passive force. The active force is defined as

$$F_{CE} = N_a(t) \cdot F_L(l) \cdot F_v(v), \tag{2}$$

where $N_a(t)$ is the muscle active state, $F_L(l)$ is the force-length characteristic and $F_v(v)$ is the force-velocity characteristic. The dependencies $N_a(t)$, $F_L(l)$, $F_v(v)$ are driven by relations mentioned below or can be directly defined by the user.

The muscle active state, $N_a(t)$, is interpreted as the ratio of a current value of muscle active force to the maximal force that can be exerted by a given muscle at a given length and elongation/shortening rate. Then $N_a(t)$ is dimensionless quantity ranging from its minimal possible value A_{min} to its maximum activation equal to 1. Based on the literature [13] it can be estimated that in vivo $A_{min} = 0.005$. The state where $N_a = 1$ corresponds to fully tetanized muscle. Active muscle state dynamics is driven by a set of two ordinary differential equations

$$T_{ne}\frac{\mathrm{d}N_e}{\mathrm{d}t} = u - N_e, \tag{3}$$

$$T_a\frac{\mathrm{d}Na}{\mathrm{d}t} = N_e - N_a, \tag{4}$$

where N_e is the neuromuscular excitation, T_{ne} is the time constant of excitation, and T_a is the time constant of activation and u is the neuro-controller output signal [13].

The active force-length characteristic function $F_L(l)$ depends on the instantaneous muscle length l and respects the relation

$$F_L(l) = F_{\max}\mathrm{exp}\left(-\left(\frac{\frac{l}{l_{opt}} - 1}{C_{sh}}\right)^2\right), \tag{5}$$

where C_{sh} is the shape parameter that determines the concavity of the muscle force-length characteristic [13].

The muscle force-velocity characteristic function has different forms for muscle elongation and shortening

$$F_v(v_n) = \begin{cases} 0 & v_n \le -1 \\ \frac{C_{short}(1+v_n)}{C_{short}-v_n} & -1 < v_n \le 0, \\ \frac{C_{leng}+C_{mvl}v_n}{C_{leng}+v_n} & v_n > 0 \end{cases} \tag{6}$$

where v_n is the muscle elongation/shortening rate normalized to the maximum shortening velocity v_{max}, $v_n = v/v_{max}$; C_{short} and C_{leng} are the Hill-type shape parameters for muscle shortening and lengthening, respectively, and C_{mvl} is the parameter which determines the ratio of ultimate force during active lengthening to the isometric force at full activation [13].

In the presented test the unknown muscle active force F_{CE} is searched. Since the test is a static isometric problem, $F_v = 1$, $F_L = \mathrm{const}$ and $F_{CE} = N_a \cdot \mathrm{const}$. Hence it can be said that the active muscle force is proportional to the muscle active state.

3.3. Muscular balance solver

The musculo-skeletal system is overdetermined. It means that from the mechanical point of view it is possible to get the actual position by activation of different muscles. Hence it is necessary to solve the problem by an optimization method. The question now is which cost function to choose. In [11] the summary of various cost functions is published. In the presented study a special muscular balance solver is used [5]. It includes the cost function that minimizes muscular discomfort while respecting the main muscle role, i.e. if the muscle belongs to a group of agonists or antagonists. Agonists are the main actors of a movement while antagonists oppose the movement. The resultant muscle active force is also influenced by the parameter involving a voluntary muscle contraction. A human subject can carry a given load in a given posture under more or less overall voluntary muscle contraction (0–100 %). So the voluntary contraction can be expressed as an ability of human being to willingly tense its muscles without carrying any load. The level of voluntary contraction is the input parameter defined by the user.

The muscular balance solver [5] evaluates the activations for all particular muscles to keep a given loaded position involving the cost function which can be represented as the least possible muscle energy that should be expended for the given task. Constrains are represented by force and momentum equilibrium equations of all body segments. As the output the muscle active forces (F_{CE}) are obtained. Further the active forces are using the Hill muscle model, described in the previous paragraph, converted to muscle active states (N_a).

3.4. Simulation setup

The human body model Robby is positioned in accordance to the experiment. The right upper arm is right angled in the elbow and the hand canter of gravity is loaded. The Robby's right upper arm including muscle elements is displayed in Fig. 6. All joint elements except the elbow and the wrist are locked to keep the same position during the loading of the hand as in the experiment. No voluntary contraction is assumed. Since the volunteer's physical proportions correspond to a 50$^{\text{th}}$ male as well as the Robby no model scaling is applied.

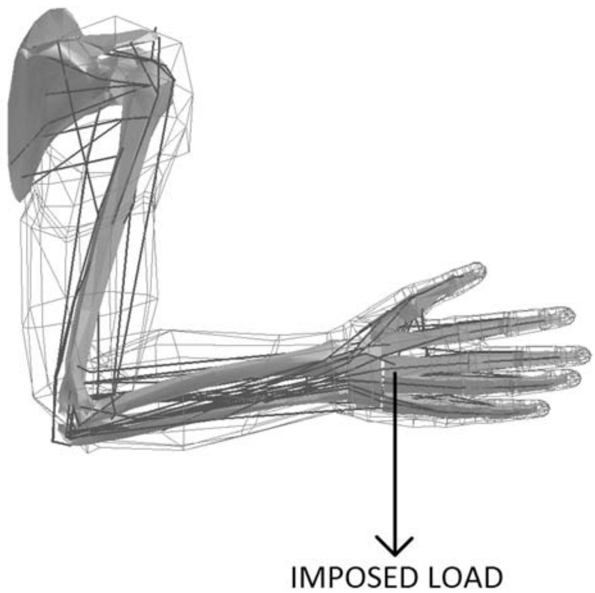

Fig. 6. Loaded Robby's right upper extremity

4. Results

Whether the relationship between EMG signal and muscle tension is linear, it has been the subject of many studies. Despite that, the EMG signal is often used as an estimate of the active state defined in the Hill-type muscle element [2]. In general, the relation $N_a(t)$ describes the muscle response to the neurocontroller input signal, and the output of this process is similar to a rectified, normalized, and slightly filtered EMG [15].

In presented case of isometric and isotonic contraction the muscle active state is constant in time and corresponds to the normalized mean value of the full-wave rectified EMG signal, $mean\_r\_EMG$. As described above all measured and processed EMG signals are normalized by the maximum measured case, $mean\_r\_EMG_{max}$, i.e. the one with the 5 kg load.

Further using the muscular balance solver the values of biceps brachia active muscle states, N_a, are computed for ten different loads. Obtained values determine muscle activities that are exerted by the biceps brachia to keep the hand loaded by the weights from 0.5 kg to 5 kg in the given position. All computed cases are normalized by N_{amax}, i.e. N_a computed for the load of 5 kg. The ratio of the computed normalized muscle activations then corresponds to measured normalized EMG responses and can be compared:

$$\frac{mean\_r\_EMG_i}{mean\_r\_EMG_{max}} \approx \frac{N_{ai}}{N_{amax}}, \tag{7}$$

where $i = 1, \ldots, 10$. The measured and computed results are summarized in Table 2.

Table 2. The comparison of the measured and computed active muscle states of biceps brachia muscle; all measured/computed cases are normalized by the measured/computed cases with 5 kg of load

No.	**Weight [kg]**	**Normalized active muscle state [–]**		
		Measurement $\frac{mean\_r\_EMG}{mean\_r\_EMG_{max}}$	**Simulation** $\frac{N_a}{N_{amax}}$	**Difference [%]**
1	0.5	0.242	0.266	9.0
2	1.0	0.340	0.300	11.8
3	1.5	0.421	0.358	14.9
4	2.0	0.526	0.450	14.4
5	2.5	0.578	0.559	3.3
6	3.0	0.591	0.671	11.9
7	3.5	0.661	0.783	15.6
8	4.0	0.798	0.893	10.6
9	4.5	0.805	1.000	19.6
10	5.0	1.000	1.000	0.0

5. Conclusion

The paper introduces the procedure of the human model Robby validation using the EMG measurement. The comparison of measured and simulated muscle activities of biceps brachia, the greatest muscle of the upper arm is shown. The paper presents as well the method of EMG application in human body modeling using Hill muscle model in general.

Firstly, muscle activations were monitored by clinical electromyography in a given unchanging position and for given loads. While keeping still the same upper arm position for the various loads the biceps brachia muscle responses were recorded. Each case was four times repeated to get relevant statistical data set. The EMG signal was monitored by the surface electrodes and processed. Then the same conditions were simulated using the human body model Robby. The Robby's right arm is positioned according to the experiment. The hand center of gravity is subsequently loaded by the forces corresponding to the used weights. Using the muscular balance solver the activations of the arm muscle are calculated for each case based on the optimization method.

Normalized measured and computed muscle activations of the biceps brachia muscle show an acceptable correlation. The procedure presents a possible way of validation of muscle activity determination in the Robby model. Robby model can be used for the muscle activity computation without bigger falsities.

The numerical human body models used this way bring the possibility to simulate the "real" life in contrast to dummies. Presented method enables to involve for example pre-crash muscle activities into computer simulations and contribute to development in passive safety.

Acknowledgements

The work is supported by the internal grant project SGS-2010-077. Special thanks belong to prof. MUDr. Zdeněk Ambler, DrSc., from FN Plzeň.

References

[1] Bogey, R. A., Perry, J., Gitter, A. J., An EMG-to-Force Processing Approach for Determining Ankle Muscle Forces During Gait. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 10 (3) (2005) 302–310.

[2] Craik, R. L., Oatis, C. A., Gait analysis, Mosby, 1994.

[3] Electromyography, Kompendium, (in Czech) http://biomech.ftvs.cuni.cz/pbpk/kompendium/biomechanika/experiment_metody_emg.php.

[4] Hynčík, L., Rigid Body Based Human Model for Crash Test Purposes, Engineering Mechanics, 5 (8) (2001) 1–6.

[5] PAM-Muscle Solution — Dedicated to Muscular Balance Calculation, Reference Manual, ESI Group, France, 2009.

[6] Pithioux, M., Chavet, P., St-Onge, N., Nicol, C., Influence of muscle preactivation of the lower limb on impact dynamics in case of frontal collision, International Journal of Crashworthiness, 10 (2005) 557–565.

[7] Rash, G. S., Electromyography Fundamentals, Western Washington University. http://myweb.wwu.edu/ chalmers/EMGfundamentals.pdf.

[8] Robbins, D. H., Anthropometric Specifications for Mid-sized Male Dummy, The University of Michigan, 1983.

[9] Seireg, A., Biomechanicaal Analysis of Musculoskeletal Structure for Medicin and Sports, Hemisphere Publishing Corporation, 1989.

[10] Sinělnikov, R. D., Atlas of human anatomy I., AVICENUM, Praha, 1980 (in Czech).

[11] Tsirakos, D., Baltzopoulos, V., Bartlett, R., Inverse optimization: functional and physiological considerations related to the force-sharing problem, Critical Reviews in Biomedical Engineering (25) (1997) 371–407.

[12] Valenta, J., Konvičková, S., Valerián, D., Biomechanics of human sceletal articulations, ČVUT, Praha, 1999 (in Czech).

[13] Virtual Performance Solution-Solver Notes Manual, ESI Group, France, 2009.

[14] Winter, D. A., Biomechanics and Motor Control of Human Movement, Willey & Sons, INC, Waterloo, 2005.

[15] Winters, J. M., Stark, L., Estimated mechanical properties of synergistic muscles involved in movements of a variety of human joints, Journal of Biomechanics 21 (12) (1988) 1 027–1 041.

[16] Wittek, A., Mathematical Modeling of the Muscle Effects on the Human Body Responses under Transient Loads, Chalmers University of Technology, Sweden, 2000.