NEUROMORPHIC MODELING OF REACHING ARM MOVEMENTS

Theses of the Ph.D dissertation

Róbert Tibold

Supervisor: József Laczkó Ph.D

Pázmány Péter Catholic University
Faculty of Information Technology
Interdisciplinary Technical Sciences Doctoral School

Budapest, 2012

Introduction and aims

In everyday running life there are many situations in which people can be injured seriously thus becoming spinal coord injured (SCI) patient or suffer from the symptoms of diseases effecting different movement functions like controlling or executing a given movement. Such movement disorders - without further details - are Parkinsons Disease (PD) [1] caused by the malfunctioning of the basal ganglia and/or the dopaminerg system; dystonia as a neurological movement disorder where oscillating muscle contractions result in twisting and uncontrolled repetitive movements with abnormal postures [2]; multiple sclerosis in which the nerves of the central nervous system (*brain and spinal cord*) degenerate as a result of inflammation of the nerves [3], [4]. Another serious disorder of the brain is the stroke in which the individual rapidly loses some of his brain functions due to disturbance or damage of the blood supply of the brain. In many cases of stroke the motor cortex of the brain is highly effected. In such cases the person can not move the limb(s) on one side of the body (*hemiplegia*). In more serious cases when the brainstem is involved in the stroke too the patient abilities for sensing and balancing may be reduced or totally lost. Therefore, for current science of motion it would be a great deal to help people overcoming their serious movement disorders whether it is caused by either accident or neural disorder.

In SCI patients in both paraplegics (*caused by the injury or the illness of the thoracic area of the spinal cord - normal movement functions of the neck, hand and thorax are usually not effected*) and tetraplegics (*caused by the injury or the disease of the cervical area of the spinal cord*) functional electrical stimulation (FES) has been recently used in many studies to recover lost functions and muscle density of the leg muscles partially [5-7]. For example cycling using FES offers paraplegics the possibility of muscle and cardiovascular training as well as the chance for independent locomotion [8]. It is desirable to find optimal stimulation patterns for generating smooth cycling movement of spinal cord injured patients by means of functional electrical stimulation. Pilissy *et al.*, [5] showed that the ankle might play an important role in FES cycling. Based on the measurements of cycling of healthy individuals (N=42) it was found that at higher gear the muscles spanning the ankle joint had to generate more torque so as to maintain a given cycling speed [5]. FES cycling has not been applied only in SCI patients but in patients with multiple sclerosis as well by Szecsi [9]. In the studies applying FES during cycling stimulation patterns were generated as a function of the pedal angle. Here, the most challenging task was to find the proper timing course of stimulating the right flexor or extensor muscle at the right instants.

However, FES has its own limits. Namely, personalized anthropometric parameters, neaural and biomechanical features (*muscle geometry, muscle action lines with acting muscle forces*) of the particular limb are not taken into account in generating stimulation patterns.

If the results of accurate 3D modeling of these properties were applied in the generation of movement patterns it could definitely increase the effectiveness of novel methods applied in the rehabilitation.

Many studies have revealed the applicability of FES in the cases of the hand [10] and the forearm [11] respectively. The Freehand system [10], [12] is an implanted FES device for restoration of lateral and palmar grasps following C5 or C6 tetraplegia. Certain shoulder movements are transformed into electrical signals and transmitted to an electrical stimulator which stimulates the muscles of the palm and the wrist through implanted electrodes to perform grasping. Naito [11] presented a method to stimulate the biceps brachii to perform supination of the forearm. However, no information has been found so far whether FES using 3D neuro-biomechanical modeling would be applied for the entire upper extremity. This is caused by the high complexity of the shoulder girdle which makes the number of degrees of freedom (DOF) increased resulting a more complex methodology.

Whole upper limb models [13-17] or partial studies [18-25] have been described over the past decades. These models were mostly developed to study the static and dynamic effects of muscles and joint torques giving insight into human limb coordination.

Unfortunately current rehabilitation methods are not capable of fully restoring the previously lost movement functions. According to statistics from the United States 50% of individuals suffering from stroke may have paralysis on one side of their body after rehabilitation procedure [26], [27]. Therefore, it is really desirable to either improve the efficiency of the rehabilitation techniques applied or to develop new methods based on both personal anthropometric and neuro-biomechanical parameters of the patient using 3D modeling approaches. Graphic based multidimensional computer models were developed to discern motor activity patterns of musculoskeletal systems [28], [29]. Inverse problems were studied in a neuro-mechanical transducer model that computes possible muscle forces and activities of flexor and extensor motoneuron pools during voluntary limb movements [30-32]. This model was applied for studying planar movements of the rat hindlimb.

In motor control when dealing with the 3D modeling of different limb movements on one hand a complex 3D inverse kinematic problem must be solved to obtain muscle forces while on the other hand the redundancy problem must be taken into account as well where an optimal solution has to be found for selecting a proper combination of muscles activated during movement execution. A human limb has generally more joints and around each joint the human body has much more spanning muscles than necessary [33], [34]. Therefore, it is still not an obvious question what is controlled by the central nervous system (CNS) and how it chooses control strategies. The muscle synergy problem has been recently investigated in different studies for multi-finger quick force production [35] and for different limb postures [36-40].

The most important aim of my Ph.D. studies is to investigate motor variances in essential motor control levels and the effect of external loads on these variances. Furthermore, to model how muscles of the upper limb may act while producing forces so that to be able to reproduce these activities artificially by applying e.g. electrical stimulation in the future although the generation of FES patterns and the development of new control methods for improving the efficacy of current upper limb rehabilitation methods are not the scope of the thesis.

In order to do that based on the measurement of reaching and grasping arm movement of healthy individuals a 3D musculoskeletal model of the entire upper limb was developed containing the geometry of 3D limb segments (with shoulder complex); the location of 3D muscle attachment sites and the anatomical/biomechanical structure of four muscles.

Methods of investigation

I investigated a daily executed point-to-point motor task performed by healthy individuals. Twenty persons (aged 21-25; mean±SD.: 21.1±1.9), 14 men and 6 women with no upper extremity complaints, voluntarily participated in the study at the National Institute for Medical Rehabilitation (NIMR). The participant sat in front of a 2-level-computer desk. The motor task was executed under three conditions corresponding to three objects with different masses: 1) a light CD case (0.06 kg); 2) a load (1 kg – referred as O1); 3) a load

(2 kg – referred as O2) ten times per load condition. The actual object was placed on the lower level of the desk. In the starting position the arm of the participant was hanging alongside the body and the palm was facing backward. The particular movement was divided into 3 phases:

a) **uplifting phase:** the individual was instructed to lift his arm from the initial position to reach and grasp the object on the lower level of the desk and had to lift it and place it onto the upper level and finally release the object and move the arm back to the initial hanging position;
b) **pause:** In this second phase, the arm remained in the hanging position taking a short pause (2-5 sec);
c) **putting down phase:** in this third phase, the participant had to lift his arm to reach the object on the upper level of the desk, put it back down to the lower level, release the object and move the arm back to the starting position again. The entire task was repeated ten times: 10 uplifting and 10 putting down trials were recorded under each condition.

To measure 3D coordinates of upper limb joints I used an ultrasonic-movement analyzer system (usMA) (ZEBRIS CMS 70P, Zebris Medical GMBH, Germany). I measured the muscle activites of 4 arm muscles - biceps (BI), triceps (TR), deltoid anterior (DA) and deltoid posterior (DP) - with the built in EMG recorder. Kinematic and muscle activity data processing was solved by using self developed

MATLAB (The MathWorks Inc.,Natick, MA, USA) based algorithms.

I developed a MATLAB based simulation method that is able to determine the direction and magnitude of muscle torques and muscle forces in the three dimensional space considering four muscles of the whole upper extremity containing the shoulder complex only using measured 3D coordinates of upper limb joints. In the model I elaborated a general mathematical method to determine 3D muscle geometry (*3D muscle insertion and origin areas*) based on cadaveric measurements [41] and personal anthropometric data of the participating individuals. Furthermore, biomechanical features taken from the literature are taken into account by considering the differing structures and strain capabilities of the investigated muscles.

By using the developed 3D musculoskeletal model I analized how the heavier object in the hand effected the motor variances of 4 different levels of the motor apparatus: endpoint, joint configuration, recorded EMG, computed muscle force. In the analysis of these levels of the motor apparatus I investigated two object conditions 1) *without load* (empty *CD* case) and 2) *with load* (2 kg object (*O2*)) and I calculated variances of joint configuration (JC), endpoint position (EP), EMG activities and muscle forces (FORCE) - as mentioned above - for the time interval when the object was held by the individual (HOLDING). I carried out my statistical investigations (repeated measures ANOVA at a $p<0.05$ significance level; correlation analysis) by applying STATISTICA (StatSoft, Inc., Tulsa, OK, USA) and MATLAB.

Novel scientific results

Thesis I. Three dimensional model for establishing the direction and magnitude of 3D muscle force vectors and joint torques in general upper limb movements using measured 3D kinematic and anthropometric data.

Related publications: [Tibold et al., 2011 , Tibold and Laczkó, 2011, Laczkó and Tibold, 2010, Laczkó et al., 2009]

In the case of 3D movement modeling the directions of torques and muscle forces acting in a given joint are not obvious especially not if one would like to consider how muscles (agonists and antagonists) are operating together at a given time instant. Another important issue is the determination of either proper 3D muscle geometry (containing 3D muscle insertions and origins) or the biomechanical characteristics of muscles (the active ($F_a(l)$) and passive ($F_p(l)$) force length relations). In the following groups of theses results related to geometric properties and variances of muscle forces and joint torques are summarized.

I.1. I have given a mathematical algorithm for determining the direction of three dimensional muscle force vectors to generate a desired joint torque during the execution of general, three dimensional point-to-point arm movements. As a part of the algorithm I have elaborated a novel general method to determine three dimensional muscle insertion and origin areas located on the surface of the bone by applying both cadaveric parameters and personal anthropometric values to have accurate muscle geometry.

If only one muscle is active at a time instant *t*, than the torque generated by the muscle in the spanned joint is computed as the difference of the total torque and the gravitational torque [30-32]:

$$\overrightarrow{F_m(t)} \times \overrightarrow{R_m(t)} = \overrightarrow{\beta(t)} \cdot \overset{(\mathbf{joint})}{I}(t) - \overrightarrow{T_{g(\mathbf{joint})}(t)} \qquad (1)$$

where $\overrightarrow{F_m(t)}$ is the force generated by the muscle, $\overrightarrow{R_m(t)}$ is the moment arm of the muscle, $\overrightarrow{\beta(t)}$ is the angular acceleration of the joint spanned by the muscle, $\overset{(\mathbf{joint})}{I}(t)$ is the moment of inertia, $\overrightarrow{T_{g(\mathbf{joint})}(t)}$ is the gravitational torque due to the rotated body part. Depending on the direction of the required torque either the flexor or the extensor muscle group should be activated at each instant for an artificial control of the joint torque. Such virtual muscle forces (Figure 1) were predicted for 4 arm muscles separately (BI, TR, DA, DP) at each time step during the desired movement. The novelty of my computation is that I considered that all parameters (except $I^{(\mathbf{joint})}(t)$) were three dimensional vectors and not only their magnitude but their direction was changing in time throughout the movement. Modeling approaches usually have been elaborated by restricting joint rotations around predefined rotational axes, although natural human movements do not comply with such restrictions [42]. I elaborated 3D computation for all of the vectors as function of time.

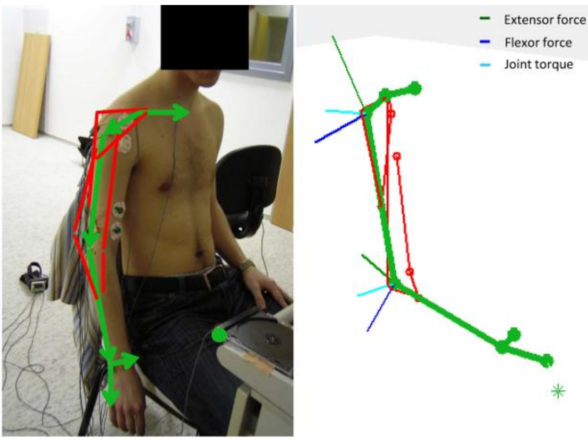

Figure 1. Virtual determination of 3D muscle forces for flexor (blue) and extensor (green) muscles neded to generate the 3D joint torque (light blue) in the spanned joint at a discrete time instant.

I determined personalized 3D coordinates of muscle attachments located on the surface of the bone. First, based on the cadaver study of Veeger [41] I put the questioned muscle insertion-origin points to the midline of the bone segment containing either the insertion or the origin. But since 1) muscles are located on the bone surface and not on the midline of the bone segment and since 2) Veeger provided data for only fully stretched elbow - which was only the initial posture of the measured movement - I applied the rotation formula of Rodrigues to predict 3D coordinates of muscle attachments (Figure 2) for the entire movement interval.

$$\begin{aligned}\left(\overline{uBII_{t+1}}\right) &= \overline{uBII_t}\cos\Theta_{(e)t} + \left(z_t \times \overline{uBII_t}\right)\sin\Theta_{(e)t} + \\ &+ \left(z_t z_t^T * \overline{uBII_t} * (1-\cos\Theta_{(e)t})\right)\end{aligned} \tag{2}$$

$$\Theta_{(e)t} = \alpha\left(E_{t+1}\right) - \alpha\left(E_t\right) \tag{3}$$

$$\overline{E_{t+1}BII_{t+1}} = \left\|E_{t+1} - BII_{t+1}\right\| * \overline{uBII_{t+1}} \tag{4}$$

$$OnBoneSurf\_BII_{t+1} = \overline{E_{t+1}BII_{t+1}} + E_{t+1} \tag{5}$$

Remark: A mathematical method for computing the biceps insertion. The same method as above was applied by replacing the rotation axis z_t ; the rotated muscle attachment unit vector; the angle of rotation θ_t to particular ones related to the specific muscle.

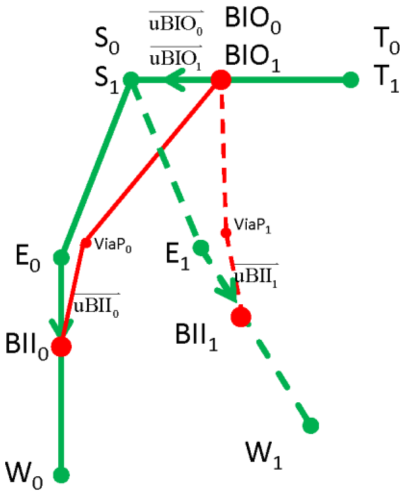

Figure 2. *Anatomical landmarks (T – thorax; S – shoulder; E – elbow; W - wrist); muscle attachments of the biceps (BIO - origin; BII - insertion); unit muscle attachment vectors (* $\overrightarrow{uBII}, \overrightarrow{uBIO},$ *) used in the rotation method of Rodrigues during the calculation of the biceps attachments located on the surface of the bone are presented at two successive time instants (t and t+1). The rotation method was applied to determine muscle insertions and origins for not only the initial position (t=0) but for the whole movement interval as well.*

I.2. I have proven that the elbow and shoulder joint torque profiles predicted by the 3D biomechanical model are invariant to changes of the mass of the object held in the hand. The range but not the shape of the torque-time curve depended on the object held in the hand.

Mean torque profiles of all trials were generated within each participant for the three object conditions during lifting and putting-down separately for the holding movement periods (*the actual object is in the hand of the participant*). Ten trials of each participant were averaged. The data of one representative individual are shown (Figure 3).

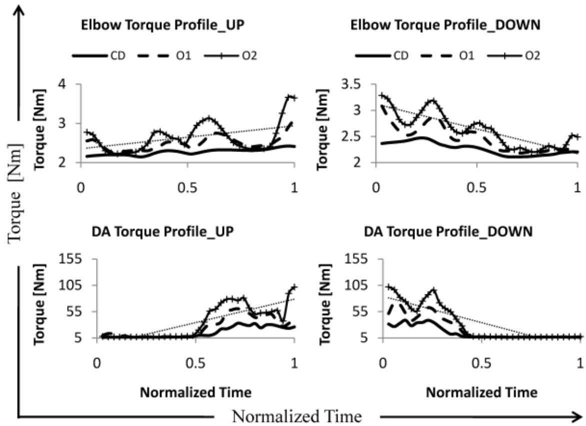

Figure 3. *Mean predicted torque profiles of ten trials in each object conditions (CD, O1, O2) for both movement directions (UP, DOWN) during holding time interval (the period during which the person was holding the object) for one representative individual). Dotted lines (linear regression fit) sign the linear increasing or decreasing tendency of the torque during lifting and putting-dowy. Only the range but not the shape of the torque-time curve depended on the mass of the object.. The magnitude of joint torques was larger for heavier objects than for the lightest ones. The invariance of the shape is shown by correlation coefficients for all conditions (regarding directions, objects) in both joints (Table 1).*

To present the effect of changing object conditions on mean torque profiles correlation analysis was performed for holding time intervals (Table 1). The strongest ($0.77 \leq r \leq 0.99$) Pearson coefficients were observed when comparing the linear relation between O1 and O2 torque curves. Weaker ($0.35 \leq r \leq 0.98$), but still strong linear correlation was present between CD and O1 torque profiles while the weakest ($0.12 \leq r \leq 0.96$) correlation was found between CD and O2 conditions. Because mean correlation coefficients are higher than 0.58 in all object conditions for both directions indicating high linear connection between separate conditions the torque profile is considered to be object invariant for both directions (Table 1). The magnitude but not the shape of the torque-time curve depended on the object held in the hand. The magnitude and the amplitude of joint torques were larger for heavier objects than for the lightest one in both the elbow and shoulder. The joint torques show increasing profiles for lifting and decreasing profiles for putting-down (Figure 3).

Table 1.

Pearson r-values of all individuals for the elbow and shoulder joint torques observed during holding considering all object conditions.

SUBJID	ELBOW TORQUE						SHOULDER TORQUE					
	UP			DOWN			UP			DOWN		
	CD-O1	CD-O2	O1-O2	CD-O1	CD-O2	O1-O2	CD-O1	CD-O2	O1-O2	CD-O1	CD-O2	O1-O2
1	0.71	0.58	**0.85**	0.89	0.87	**0.97**	0.69	0.79	**0.9**	**0.91**	0.77	0.8
2	0.69	0.51	**0.88**	0.9	0.82	**0.95**	0.75	0.87	**0.9**	0.92	0.87	**0.95**
3	0.89	0.74	**0.92**	0.89	0.87	**0.9**	0.98	0.97	**0.99**	0.95	0.9	**0.97**
4	0.63	0.55	**0.86**	0.87	0.83	**0.94**	0.51	0.71	**0.87**	0.9	0.84	**0.98**
5	0.83	0.66	**0.88**	0.66	0.76	**0.89**	0.79	**0.93**	0.73	**0.97**	0.86	0.87
6	0.86	0.72	**0.91**	0.55	0.72	**0.92**	0.88	0.81	**0.93**	0.92	0.91	**0.97**
7	0.67	0.62	**0.86**	0.97	0.96	**0.98**	0.45	0.5	**0.8**	0.95	0.94	**0.99**
8	0.85	0.74	**0.92**	0.95	0.94	**0.97**	0.82	0.76	**0.83**	0.9	0.75	**0.98**
9	-0.3	0	**0.87**	0.4	**0.72**	-0.1	0.35	0.12	**0.86**	**0.87**	0.75	0.79
10	0.44	0.37	**0.86**	0.9	0.87	**0.97**	0.75	0.74	**0.89**	0.79	0.72	**0.94**
11	0.64	0.58	**0.85**	0.81	0.7	**0.82**	0.9	0.8	**0.91**	0.9	0.89	**0.93**
12	0.33	0.63	**0.83**	0.9	0.69	**0.98**	0.22	0.2	**0.87**	0.79	0.64	**0.9**
13	0.79	0.78	**0.87**	0.97	0.86	**0.98**	0.64	0.6	**0.94**	0.97	0.96	**0.98**
14	0.68	0.54	**0.91**	0.86	0.63	**0.88**	0.9	0.85	**0.91**	0.96	0.76	0.87
15	0.87	0.8	**0.91**	0.85	0.84	**0.95**	0.97	0.96	**0.98**	0.95	0.94	**0.97**
16	0.16	0.42	**0.46**	-0.6	-0.6	**0.8**	0.04	0.05	**0.33**	-0.8	-0.9	**-0.8**
17	0.48	0.40	**0.9**	0.3	0.25	**0.84**	0.83	0.74	**0.86**	0.9	0.89	**0.94**
18	0.47	0.4	**0.91**	0.52	0.33	**0.77**	0.85	0.81	**0.94**	0.94	0.9	**0.96**
19	0.76	0.71	**0.98**	0.81	0.79	**0.85**	**0.91**	0.79	0.85	0.9	**0.98**	0.94
20	0.81	0.79	**0.98**	0.67	0.61	**0.77**	0.83	0.81	**0.88**	0.78	0.7	**0.8**
Mean	0.61	0.58	**0.87**	0.70	0.67	**0.85**	0.70	0.69	**0.85**	0.81	0.75	**0.83**
SD	0.29	0.24	0.10	0.38	0.35	0.23	0.25	0.26	0.13	0.39	0.40	0.39

Note. Generally, the strongest ($0.77 \leq r \leq 0.99$) correlation coefficients were found between O1 and O2; weaker but strong ($0.35 \leq r \leq 0.98$) correlation appeared for CD and O1; the weakest coefficients ($0.12 \leq r \leq 0.96$) occured for CD and O2.. Correlation coefficients for individuals 9 and 16 are negative in the comparison of CD-O1 and CD-O2. Mean correlation coefficients are higher than 0.58 for all conditions.

Thesis II. The effect of objects with distinct masses on the variances of endpoint (EP), joint configuration (JC), sEMG and simulated muscle forces (FORCE) as levels of the bilogical motor apparatus during holding the object in the hand.

Related publications: [Tibold et al., 2011, Tibold et al., 2009, Laczkó and Tibold, 2009]

In order to reproduce complex arm movements artificially first the issue of muscle synergies, namely how the CNS chooses its strategy to select proper muscles for optimal solution of a given motor task and what is controlled by the nervous system under changing object conditions, must be understood.
The following group of theses summerizes how motor variances on different levels of the motor control were effected by a heavier object held in the hand during the time interval of holding. Variances of the performed movements were computed during holding as functions of normalized time and were averaged across time for both object conditions (CD, O2) within all subjects in lifting and putting-down. These values were computed for the endpoint, joint configuration, sEMG of 4 arm muscles (BI, TR, DA, DP) and for computed muscle forces of the same muscles separately. The mean variance across holding with load was divided by the mean variance across holding without load for 1) all individuals separately and for

2) the mean variances (Figure 4) across all individuals (Table 2, *RATIO*).

II.1. I have proven that motor stability highly depends on altering load conditions. Movements executed with load (O2) varied in a higher manner than movements executed without load (CD) during holding time interval.

This has been proven by analysing variances (Figure 4) in external workspace, in internal joint space and in the space of muscle activation patterns when an object was in the hand of the actual person (HOLDING time interval).
In endpoint variances results did not show significant difference between the two object conditions either in lifting ($F(1,19)=1.62$, $p=0.21$) or in putting-down ($F(1,19)=1.99$, $p=0.17$) at $p<0.05$. In joint configuration there was no significant difference between the object conditions during lifting ($F(1,19)=0.73$, $p=0.4$). However, in putting-down the arm configuration variance was significantly ($F(1,19)=8.11$, $p=0.01$) greater if movements were executed with load proving a less enhanced effect of the gravity on movement execution while acting against the gravity. High significant differences occurred in both surface EMG and virtually computed muscle force levels depicting minimal p values in the ranges of $0.0002 \leq p_{EMG} \leq 0.024$ and $0.00002 \leq p_{FORCE} \leq 0.0005$ respectively.

However, in computed muscle forces DP showed no difference between the object conditions either in lifting or in putting-down.

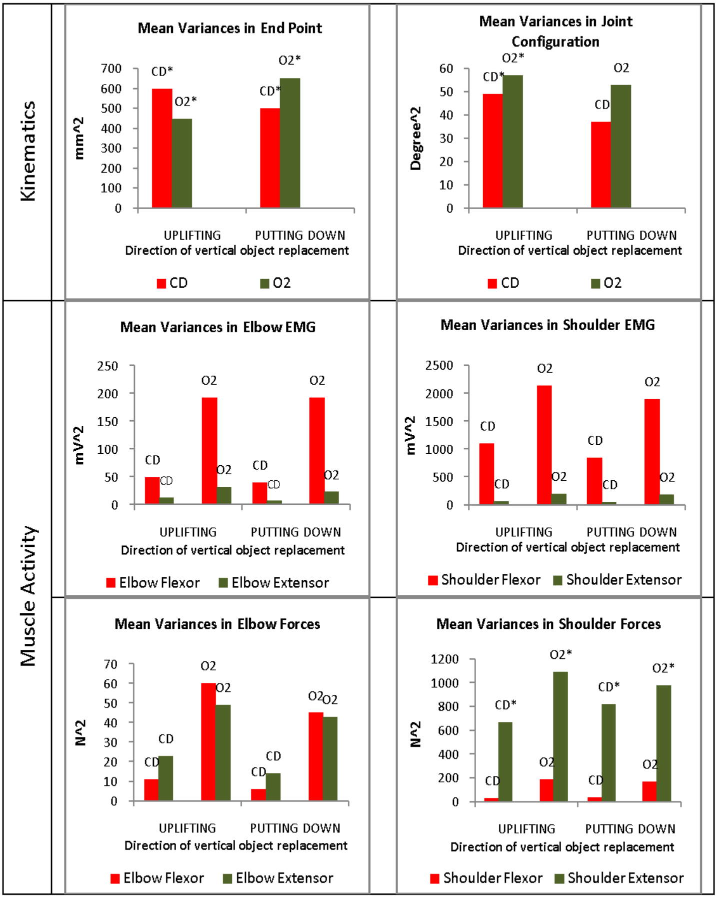

Figure 4. *Mean values of variances across 20 individuals for all motor control levels in both directions (UPLIFTING, PUTTING DOWN) under both conditions (CD, O2). In the cases of coherent CD-O2 pairs * means that even though the variance of*

movements executed with O2 is higher than movements executed with CD but this difference was not significant at $p<0.05$.

II.2. I have shown that RATIO of movement variances with load (O2) to movement variances without load (CD) was smaller for endpoint and joint configuration variances than for sEMG and muscle FORCE variances respectively representing that enhanced muscle synergies stabilize the movement at kinematic level by controlling primarely through the hand position and less by the combined joint rotations and not by individual muscle activities.

To compare the effect of load on different control levels I computed the ratios of mean variances across all individuals by dividing the mean variance of movements with load by the mean variance of movements without load. (*RATIO*) $RATIO_{JC}>RATIO_{EP}$ for both uplifting and putting-down. $RATIO_{EMG}>>1$ and $RATIO_{FORCE}>>1$ concerning all investigated muscles except DP.

Thus, it is obvious that the load effected the muscle variances at a higher rate than kinematic variances (Table 2) proving that the effect of the heavier object on variances was the highest at muscle activity level, smaller at joint configuration level and it was the smallest at endpoint level.

Table 2.

Ratios of mean variances (RATIO)

	UP		DOWN	
EP	0.7		1.3	
JC	1.2		1.4	
EMG	**BI**	**TR**	**BI**	**TR**
	3.9	2.5	4.8	3.3
	DA	**DP**	**DA**	**DP**
	1.9	2.83	2.2	3.42
FORCE	**BI**	**TR**	**BI**	**TR**
	9.6	2.3	5.75	2.83
	DA	**DP**	**DA**	**DP**
	5.3	1.1	4,83	1,12

Note. In the table the RATIO of movement variances with load (O2) to movement variances without load (CD) is given for the studied motor control levels. The RATIO was smaller for endpoint and joint configuration variances than for sEMG and muscle FORCE variances respectively representing that enhanced muscle synergies stabilize the movement at kinematic level by controlling primarely through the hand position and less by the combined joint rotations and not by individual muscle activities.

The shoulder extensor (DP) had a slightly different behavior than other muscles because the variances of DP were higher than the variances of DA in contrast to EMG variances (Figure 4). A possible reason of this is that the main action lines of muscle forces may alter and influence muscle force variances caused by the structure of the deltoid muscle [43], [44].

II.3. I have shown that using virtually predicted muscle forces, generalized muscle activity patterns for rehabilitation of the upper extremity containing the shoulder complex cannot be generated in the cases of the main arm muscles. Instead of using generalized action patterns for all patients, personalized movement patterns rather lead to adequate rehabilitation processes of individuals.

Table 3.

Between subjects mean and SD of predicted muscle forces

		Mean ***(N)***		**SD** ***(N)***	
		CD	**O2**	**CD**	**O2**
UP	**BI**	23	220	16	168
	TR	24	55	15	38
	DA	12	64	8	42
	DP	679	742	839	604
DOWN	**BI**	32	184	28	143
	TR	18	51	14	36
	DA	17	82	14	70
	DP	785	886	928	758

Note. *Between subjects mean and SD of predicted muscle forces (BI, TR, DA, DP).. Standard deviations of predicted muscle forces were remarkably high compared to the mean of predicted muscle forces across participants indicating different execution patterns between subjects.*

Predicted muscle force standard deviations (SD) across all individuals were relatively high compared to the mean across participants for both object conditions (Table 3) indicating distinctive force patterns between persons exerted by individual muscles under the execution of the same point-to-point movement. Therefore, the execution of movements under altered object conditions varies subject by subject. To demonstrate this, standard deviations were computed within subjects (*for 10 trials*) for both directions (UP, DOWN) under each load condition (CD, O2) to all individuals separately and then averaged across all participants (Table 4).

Table 4.

Averaged within subjects predicted muscle force SD-s

[N]	UP		DOWN	
CD	BI	TR	BI	TR
	2.4542	2.7512	3.255	2.5224
	DA	DP	DA	DP
	2.0863	16.166	2.3067	19.475
O2	BI	TR	BI	TR
	5.5247	2.8967	4.7365	3.0204
	DA	DP	DA	DP
	5.0782	30.423	3.6851	16.128

Note. Averaged within subjects standard deviations are significantly smaller than between subjects SD-s (Table 3) at $p<0.05$ significance level. This suggests that personalized movement patterns lead to efficient rehabilitation processes.

Within subjects SDs (Table 4) were significantly smaller ($p<0.05$) than between subjects SDs for both directions under each object condition.

As a consequence of this fact, generalized movement patterns using predicted muscle forces for rehabilitation processes cannot be generated because it changes between individuals. However, using the presented 3D biomechanical model (*Theses I.*) personalized movement patterns might be better applicable in FES rehabilitation procedures of tetraplegic patients.

Future applications

In the first theses group I summarized the most important novel results of a personalized biomechanical model.

These results will be useful to generate personalized muscle activity patterns for patients having rehabilitation of their upper extremity. Personalized movement patterns will be created based on anthropometric data to define typical muscle geometries characterizing the individuals anatomical muscle geometries as accurately as possible. Furthermore, in the next step of the research I plan to reveal the relationship between virtually computed muscle forces and stimulation patterns. Such stimulation patterns should be applied by using electrical stimulators in FES applications to activate the main muscles of the upper limb and generate reaching-grasping arm movements artificially in the 3D space. In the second theses group I summarized how the heavier object affected different levels of the motor apparatus thus supporting that enhanced muscle synergy stabilizes the movement at kinematic levels.

This issue is going to be very useful in medical rehabilitation and in occupational therapy by assisting medical doctors in selecting which muscles or body parts should be trained for more efficient motor performance.

My theses contribute to human motor control research by better understanding how the central nervous system (CNS) reacts for changing external conditions.

Acknowledgement

First and foremost I would like to thank my supervisor and scientific adviser *József Laczkó* for his consistent support, for sharing his concerns and suggestions related to my studies and scientific work. I would also like to express my gratitude to the head of the doctoral school *Prof. Péter Szolgay* and *Prof. Tamás Roska* for the support and the chance to be the member of the doctoral school. The interdisciplinary environment they provided proved to be truly important in the successful conclusion of my PhD studies. I am also thankful to my collaborators *Gábor Fazekas*, *Györgyi Stefanik* without whom the biomechanical measurements would have never been done in the National Institute for Medical Rehabilitation. I am also very grateful to the members of the team *Tamás Pilissy, Bence Borbély, Péter Katona* working under the leadership of József Laczkó in the Pázmány and Semmelweis University for their useful advice concerning my work and for the valuable meetings.

I am in the same way grateful to my fellow Ph.D students *Éva Bankó, Ádám Fekete, Richard Fiat, László Füredi, Barnabás Hegyi,* *Zoltán Kárász, András Kiss, László Laki, Csaba Nemes, Tamás Pilissy, Ákos Tar, Dávid Tisza, Gergely Treplán, József Veres, Béla Weiss* for all their assistance and the time we have spent together. I acknowledge the help of *Lívia Adorján, Judit Tihanyi, Vida Tivadarné Katinka* and the staff of the *Registrar's Office, Dean's Office and the Financial Department.* I would like to thank *Márton Péri* for his work to improve the English style of the dissertation.

Last but not least I am tremendously grateful to my *Parents* and *Grandparents* for all their love and support. Finally I would like to thank my darling *Bogi* for all her love, support and patience during my busy days. Without them I could have never managed to succesfully conclude my studies.

Publications

The author's journal publications related to the theses book

R. **Tibold**, J. Laczkó, „The effect of load on torques in point to point arm movements, a 3D model", *Journal of Motor Behavior* (accepted), 2012

R. **Tibold**, G. Fazekas, J. Laczkó, „Three-dimensional model to predict muscle forces and their relation to motor variances in reaching arm movements", *Journal of Applied Biomechanics* vol. 27, no. 4, pp. 362-374, 2011

The author's conference publications related to the theses book

R. Tibold, J. Laczkó, „The effect of load on variances of object replacing arm movements", *4th International Symposium On Applied Sciences In Biomedical And Communication Technologies*, ISBN 978-1-4503-0913-4/11/10, 2011

J. Laczkó, **R**. **Tibold** „Variances of joint torques and muscle activities during arm movements with loads". *Neuroscience Meeting*, Program No. 810.19, Washington, DC: Society for Neuroscience, Online, 2011

J. Laczkó, **R**. **Tibold**, „3D analysis to reveal muscle activity timing in object replacing arm movements", *11th International Symposium on the 3D Analysis of Human Movement*, pp. 90-93., San Francisco, 2010

R. **Tibold**, A. Poka, B. Borbely, J. Laczkó, „The effect of load on joint- and muscle synergies in reaching arm movements". *Accepted at VII. Conference on Progress in Motor Control*, Marseille, France 2009. July, 2009

J. Laczkó, **R**. **Tibold**, G. Fazekas, „Neuromuscular synergy ensures kinematic stability during 3D reaching arm movements with load". *Neuroscience Meeting 2009*, Program No. 272.2, Chicago, IL: Soc. for Neuroscience, 2009. Online, 2009

J. Laczkó, T. Pilissy, **R. Tibold**, „Neuro-mechanical Modeling and Controlling of Human Limb Movements of Spinal Cord Injured Patients”. *Proc. of the 2nd International Symposium on Applied Sciences in Biomedical and Communication Technologies*. ISBN 978-80-227-3216-1, On CD Isabel 2009, File number: 285283, 2009

Cited publications in the theses book

[1] J. Sanyal, D. P. Chakraborty, and V. R. Rao, “Environmental and familial risk factors of Parkinsons disease: case-control study.,” *The Canadian journal of neurological sciences. Le journal canadien des sciences neurologiques*, vol. 37, no. 5, pp. 637-642, 2010.

[2] S. Judd, *Genetic Disorder Sourcebook*, Fourth Edi. Omnigraphics.

[3] P. Czobor, J. Vitrai, S. Marosfi, and I. Toth, “Steady-state visual evoked-potential tests in Sclerosis Multiplex.,” *Electroencephalography and Clinical Neurophysiology*, vol. 50, no. 3-4, p. 117, 1980.

[4] J. Czopf, K. Hegedus, M. Kissantal, and G. Karmos, “Statistical-analysis of EEG and clinical data in multiple-sclerosis - significance of visual evoked-response in diagnosis of multiple-sclerosis.,” *Electroencephalography and Clinical Neurophysiology*, vol. 41, no. 2, p. 210, 1976.

[5] T. Pilissy, K. Pad, G. Fazekas, M. Horvath, G. Stefanik, and J. Laczkó, “The role of ankle-joint during cycling movement task.,” *International Journal of Rehabilitation Research*, vol. 30, pp. 58-59, 2007.

[6] J. Szecsi, C. Krewer, F. Müller, and A. Straube, “Functional electrical stimulation assisted cycling of patients with subacute stroke: kinetic and kinematic analysis.,” *Clinical Biomechanics*, vol. 23, no. 8, pp. 1086-1094, 2008.

[7] G. Sándor, "The Identification of the Model of the Stimulated Leg," UniversityOfTwente, 1991.

[8] J. Szecsi, S. Krafczyk, J. Quintern, M. Fiegel, A. Straube, and T. Brandt, "Paraplegic cycling using functional electrical stimulation. Experimental and model-based study of power output," *Der Nervenarzt*, vol. 75, no. 12, pp. 1209-1216, 2004.

[9] J. Szecsi, C. Schlick, M. Schiller, W. Pöllmann, N. Koenig, and A. Straube, "Functional electrical stimulation-assisted cycling of patients with multiple sclerosis: biomechanical and functional outcome--a pilot study.," *Journal of rehabilitation medicine official journal of the UEMS European Board of Physical and Rehabilitation Medicine*, vol. 41, no. 8, pp. 674-680, 2009.

[10] P. N. Taylor, J. Esnouf, and J. Hobby, "The functional impact of the Freehand system on tetraplegic hand function - Clinical results.," *Spinal cord the official journal of the International Medical Society of Paraplegia*, vol. 40, no. 11, pp. 560-566, 2002.

[11] A. Naito, M. Yajima, H. Fukamachi, and Y. Shimizu, "Functional electrical stimulation (FES) to the biceps brachii for controlling forearm supination in the paralyzed upper extremity.," *The Tohoku journal of experimental medicine*, vol. 173, no. 2, pp. 269-273, 1994.

[12] P. N. Taylor, J. Esnouf, and J. Hobby, "Pattern of use and user satisfaction of Neuro Control Freehand system.," *Spinal cord the official journal of the International Medical Society of Paraplegia*, vol. 39, no. 3, pp. 156-160, 2001.

[13] R. Raikova, "A general approach for modelling and mathematical investigation of the human upper limb.," *Journal of Biomechanics*, vol. 25, no. 8, pp. 857-867, 1992.

[14] J. Langenderfer, S. A. Jerabek, V. B. Thangamani, J. E. Kuhn, and R. E. Hughes, "Musculoskeletal parameters of muscles crossing the shoulder and elbow and the effect of sarcomere length sample size on estimation of optimal muscle length.," *Clinical Biomechanics*, vol. 19, no. 7, pp. 664-670, 2004.

[15] S. L. Delp, J. P. Loan, M. G. Hoy, F. E. Zajac, E. L. Topp, and J. M. Rosen, “An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures.,” *IEEE Transactions on Biomedical Engineering*, vol. 37, no. 8, pp. 757-767, 1990.

[16] G. L. Gottlieb, “A computational model of the simplest motor program.,” *Journal of Motor Behavior*, vol. 25, pp. 153-161, 1993.

[17] M. A. Lemay and P. E. Crago, “A dynamic model for simulating movements of the elbow, forearm, and wrist.,” *Journal of Biomechanics*, vol. 29, no. 10, pp. 1319-1330, 1996.

[18] A. A. Amis, D. Dowson, V. Wright, and J. H. Miller, “The derivation of elbow joint forces, and their relation to prosthesis design.,” *Journal of medical engineering technology*, vol. 3, no. 5, pp. 229-234, 1979.

[19] A. A. Amis, D. Dowson, and V. Wright, “Analysis of elbow forces due to high-speed forearm movements.,” *Journal of Biomechanics*, vol. 13, no. 10, pp. 825-831, 1980.

[20] A. A. Amis, J. H. Miller, D. Dowson, and V. Wright, “Elbow joint forces – Basic data for prosthesis designers.,” *Journal of Bone and Joint Surgery-British*, vol. 62, no. 2, pp. 251-252, 1980.

[21] K. N. An, M. Jacobsen, L. Berglund, and E. Chao, “Application of a magnetic tracking device to kinesiologic studies,” *Journal of Biomechanics*, vol. 21, no. 7, pp. 613–615, 1988.

[22] K. N. An, K. R. Kaufman, and E. Y. Chao, “Physiological considerations of muscle force through the elbow joint.,” *Journal of Biomechanics*, vol. 22, no. 11-12, pp. 1249-1256, 1989.

[23] J. T. London, “Kinematics of the elbow.,” *The Journal of Bone and Joint Surgery*, vol. 63, no. 4, pp. 529-535, 1981.

[24] J. C. Otis, R. F. Warren, S. I. Backus, T. J. Santner, and J. D. Mabrey, “Torque production in the shoulder of the normal young adult male. The interaction of function, dominance, joint angle, and

angular velocity.," *The American Journal of Sports Medicine*, vol. 18, no. 2, pp. 119-123, 1990.

[25] S. I. Backus, J. D. Mabry, M. A. Kroll, R. F. Warren, and J. C. Otis, "Torque production in the shoulder of the normal young-adult male.," *Physical Therapy*, vol. 65, no. 5, p. 715, 1985.

[26] "Stroke Center," 2011. [Online]. Available: http://www.strokecenter.org/patients/stats.htm.

[27] J. W. McDonald and C. Sadowsky, "Spinal-cord injury," *The Lancet*, vol. 359, no. 9304, pp. 417-425, 2002.

[28] J. Laczkó, A. J. Pellionisz, B. W. Peterson, and T. S. Buchanan, "Multidimensional sensorimotor 'patterns' arising from a graphics-based tensorial model of the neck-motor system.," *Society for Neuroscience Abstracts*, vol. 13, no. 1, p. 372, 1987.

[29] J. Laczkó, A. Pellionisz, H. Jongen, and S. C. A. M. Gielen, "Computer modeling of human forelimb muscle activation in multidimensional Intrinsic coordinate frames.," *Society fot Neuroscience. Abstract*, vol. 14, no. 2, p. 955, 1988.

[30] J. Laczkó, "Modeling of limb movements as a function of motoneuron activities.," *Kalokagathia*, vol. 43, no. 3, pp. 24-34, 2005.

[31] J. Laczkó, K. Walton, and R. Llinas, "A neuro-mechanical transducer model for controlling joint rotations and limb movements.," *Ideggyogyaszati Szemle*, vol. 59, no. 1-2, pp. 32-43, 2006.

[32] J. Laczkó, T. Pilissy, and A. Klauber, "Modeling of limb movements for controlling functional electrical stimulation of paraplegics.," *Proc. of the Third Hungarian Conference on Biomechanics*, pp. 151-157, 2008.

[33] J. Laczkó, "Modeling of multi-joint movements.," *Kalokagathia 2001, Spec. Issue*, pp. 91-96, 2001.

[34] J. Laczkó, R. Tibold, and G. Fazekas, “Neuromuscular synergy ensures kinematic stability during 3D reaching arm movements with load.,” *Neuroscience Meeting*, 2009.

[35] M. L. Latash, J. K. Shim, and V. M. Zatsiorsky, “Is there a timing synergy during multi-finger production of quick force pulses?,” *Psychopharmacology*, vol. 177, no. 1-2, pp. 217-223, 2004.

[36] M. L. Latash, V. Krishnamoorthy, J. P. Scholz, and V. M. Zatsiorsky, “Postural synergies and their development.,” *Neural Plasticity*, vol. 12, no. 2-3, pp. 119-30; discussion 263-72, 2005.

[37] G. Torres-Oviedo, J. M. Macpherson, and L. H. Ting, “Muscle synergy organization is robust across a variety of postural perturbations.,” *Journal of Neurophysiology*, vol. 96, no. 3, pp. 1530-1546, 2006.

[38] E. Kellis and A. Katis, “Hamstring antagonist moment estimation using clinically applicable models: Muscle dependency and synergy effects.,” *Journal of Electromyography and Kinesiology*, vol. 18, no. 1, pp. 144-153, 2008.

[39] D. Domkin, J. Laczkó, S. Jaric, H. Johansson, and M. L. Latash, “Structure of joint variability in bimanual pointing tasks.,” *Experimental Brain Research*, vol. 143, no. 1, pp. 11-23, 2002.

[40] D. Domkin, J. Laczkó, M. Djupsjöbacka, S. Jaric, and M. L. Latash, “Joint angle variability in 3D bimanual pointing: uncontrolled manifold analysis.,” *Experimental Brain Research*, vol. 163, no. 1, pp. 44-57, 2005.

[41] H. E. J. Veeger, B. Yu, K. N. An, and R. H. Rozendal, “Parameters for modeling the upper extremity.,” *Journal of Biomechanics*, vol. 30, no. 6, pp. 647-652, 1997.

[42] J. Laczkó, J. Quintern, and S. Krafczyk, “Modeling of joint-rotations during line tracking arm movements.,” *In: “Neuroprosthetics. From basic research to clinical application”,*, pp. 305-314, 1996.

[43] J. B. Wickham and J. M. Brown, “Muscles within muscles: the neuromotor control of intra-muscular segments.,” *European Journal Of Applied Physiology And Occupational Physiology*, vol. 78, no. 3, pp. 219-225, 1998.

[44] J. M. Brown, J. B. Wickham, D. J. McAndrew, and X. F. Huang, “Muscles within muscles: Coordination of 19 muscle segments within three shoulder muscles during isometric motor tasks.,” *Journal of Electromyography and Kinesiology*, vol. 17, no. 1, pp. 57-73, 2007.