Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Functional Electrical Stimulation: A MatLab Based Tool for Designing Stimulation Patterns

Strahinja Došen, *Student Member, IEEE*, Dejan B. Popović, *Member, IEEE*

***Abstract*—We developed user-friendly software that generates stimulation profiles by using user-customized model-based control of walking. The model is a multi-segment structure with pin and ball joints. A pair of an agonist and an antagonistic muscles acts at each joint. Each muscle is modeled by a three-compartment multiplicative model. The control is based on optimization that uses a cost function that minimizes the tracking error of the joint angles and levels of muscles activations. The inputs to the simulation are trajectories and user characteristic model parameters. The outputs of the simulation are levels of muscle activations *vs.* time. The software allows for interactive testing of various walking trajectories and model parameters since the simulation is integrated into a database of individuals and reference trajectories. The simulation was realized in the MatLab environment with multiple windows graphical user interface. Here we present an example: stimulation patterns for the shank-foot system that is applicable for walking control in hemiplegic individuals.**

I. Introduction

LACK of adequate control is a limitation for practical application of Functional Electrical Stimulation (FES) for daily use in rehabilitation of paralyzed and paretic individuals. The control for FES that we adopted as effective mimics biological control and has a hierarchical structure. The top, coordination level is based on a rule-based discrete control. The rules are If-Then relations that are connecting joint trajectories with muscle activations. The said rules can be determined by machine mapping of the sensory information (kinematics, dynamics) and actuation (muscle activity in EMG space). Typically, the output (muscle activity space) was created from the EMG recordings in healthy individuals [1]. This procedure does not consider substantial differences in properties of sensory-motor systems between the healthy and disabled individuals. We developed a tool in the MatLab environment that uses dynamic programming and parameters that are user characteristic for generating the appropriate stimulation patterns for the disabled individuals.

Manuscript received April 3, 2006. This work was partly supported by the Danish Research Council, DK, and partly by Ministry of Science and Environmental Protection of Serbia, Belgrade, Serbia.

S Došen is with SMI, Department for Health Science and Technology, Aalborg University, DK (phone: +45 96358828; fax: +45-98154008; e-mail: sdosen@hst.aau.dk), and with Faculty of Technical Sciences, University of Novi Sad, Serbia.

DB Popović, is with SMI, Department for Health Science and Technology, Aalborg University, DK (phone: +45 96358726; fax: 45-98154008; e-mail: dbp@smi.auc.dk), and with Faculty of Electrical Engineering, University of Belgrade, Serbia.

We adopted the multi-segment chain model where major body segments are represented as rigid bodies (foot, shank, thigh segments and head/arm/trunk segment). The segments are connected with pin and ball joints as appropriate. We consider that the ankle and knee joints can be modeled as pin joints when analyzing walking; yet, the hip joint comprises at least two degrees of freedom. The actuation of the body model was assumed in the form of a pair of muscles: agonist and antagonist modeled by a three-component multiplicative schema [1, 3, 4]. The novelty in this software is that the parameters that characterize muscle model are user dependent. The method for determining parameters is described in [1]. We have not considered biarticular muscles, and we have not modeled reflex responses that could occur.

II. Interactive Simulation of FES Assisted Walking

The novelty in this research is the interactive software that runs at most computer platforms in MatLab environment. The software is available on request for research purposes from the authors (sdosen@hst.aau.dk).

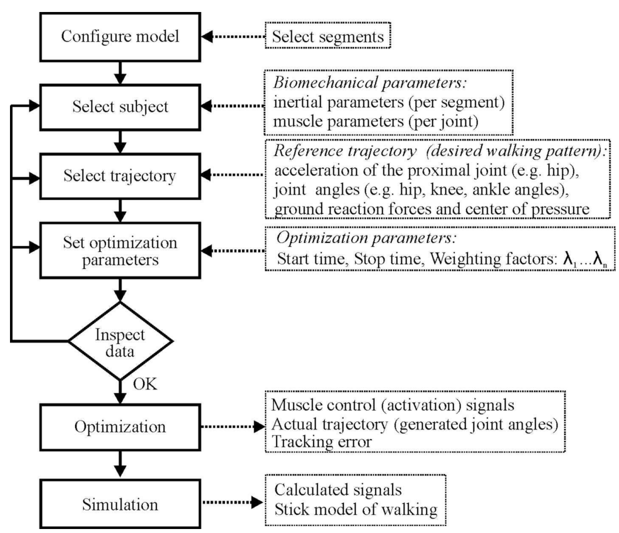


Fig. 1. The flow chart of the interactive MatLab based simulation tool. Rectangles show user activities when using the new software. The arrows show the input and/or output for each user activity.

The flow-chart diagram of the algorithm is presented in Fig. 1. The model is configured by selecting segments that will be used in simulation. User loads customized set of parameters which characterize the individual that is assisted with FES. He/she than selects desired walking pattern that will be generated by applying FES. The pattern can be chosen from the library of trajectories recorded in able-bodied or disabled subjects (level walking with different speeds, slope walking, etc.) After setting optimization parameters: time interval and weighting λ_i, i = 1, n of each of the n joints that are actuated by means of FES (the sum of values λ_1 to λ_n has to be 1), optimization algorithm can be started. The software implements dynamic programming in order to minimize two components: 1) the tracking errors from the desired trajectories, and 2) total muscle effort (sum of weighted muscle efforts in each joint). The time step used in simulation can be set at the beginning of operation. Since the electrical stimulation is applied with frequencies between 10 and 50 pulses per second, the time step, used in this simulation, of 10 ms is sufficient. The outputs from the simulation are discrete values of muscle activation. This information will provide initial stimulation patterns for the selected paralyzed subject.

III. EXAMPLE

A. Model

We present here an example that is of interest for therapy of hemiplegic individuals: 4-channel stimulation for control of the shank and foot. For this case, the user selects the two-segment model (Fig. 2). This mechanical model comprises two rigid bodies connected by two pin joints. The system interacts with the body via force and torque at the thigh segment. The model is reduced to a planar model and can be analyzed as double pendulum with the knee joint being a moving hanging point.

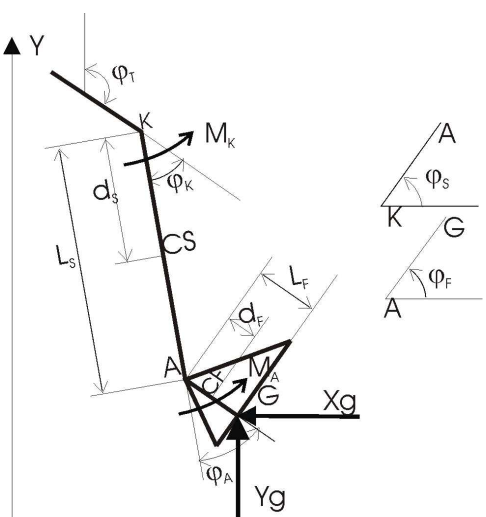


Fig. 2. Biomechanical model of the lower leg. The notations are: CS and CF are the centers of masses of the shank and the foot; J_{CF} and J_{CS} are moments of inertia of the foot and the shank; φ_A, φ_K are relative angles of the ankle and the knee; φ_S and φ_F are the absolute angles of the shank and the foot from the horizontal; φ_T is absolute angle of the thigh from the vertical; x_G is point of application of the ground reaction forces X_G and Y_G; M_K and M_A are the net torques on the knee and ankle joints.

Application of Lagrangian approach gives the following equations of motion:

$$A_1\ddot{\varphi}_F + A_2\ddot{\varphi}_S\cos(\varphi_S - \varphi_F) + A_3\dot{\varphi}_S^2\sin(\varphi_F - \varphi_S) - A_4\ddot{x}_K\sin\varphi_F - A_5(\ddot{y}_K + g)\cos\varphi_F - X_G L_F\sin\varphi_F - Y_G L_F\cos\varphi_F = M_A \quad (1)$$

$$B_1\ddot{\varphi}_S + B_2\ddot{\varphi}_F\cos(\varphi_S - \varphi_F) + B_3\dot{\varphi}_F^2\sin(\varphi_S - \varphi_F) - B_4\ddot{x}_K\sin\varphi_S - B_5(\ddot{y}_K + g)\cos\varphi_S - X_G L_S\sin\varphi_S - Y_G L_S\cos\varphi_S = M_K - M_A \quad (2)$$

$$A_1 = J_{CF} + m_F d_F^2; B_1 = J_{CS} + m_F L_S^2 + m_S d_S^2; A_2 = m_F d_F L_S; B_2 = A_2$$
$$A_3 = A_2; B_3 = B_2; A_4 = m_F d_F; B_4 = m_F L_S + m_S d_S; A_5 = -A_4; B_5 = -B_4$$

Biological actuators are represented as equivalent flexor and extensor muscles acting on the joints. The multiplicative three-component muscle model that takes into account neural activation, torque-angle and normalized torque-velocity characteristics is used [3], [4]. Passive-elastic properties of the joints are modeled as nonlinear resistive torques [5]. Equations have the following form (ankle joint):

$$M_A^f = (c_{12}\varphi_A^2 + c_{11}\varphi_A + c_{10})g_A^f(\dot{\varphi}_A)u_1 \quad (3)$$

$$M_A^e = (c_{22}\varphi_A^2 + c_{21}\varphi_A + c_{20})g_A^e(\dot{\varphi}_A)u_2 \quad (4)$$

$$M_A^r = d_{11}(\varphi_A - \varphi_{A0}) + d_{12}\dot{\varphi}_A + d_{13}e^{d_{14}\varphi_A} - d_{15}e^{d_{16}\varphi_A} \quad (5)$$

$$g_A^f(\dot{\varphi}_A) = \begin{cases} c_{14}, & \dot{\varphi}_A < (1 - c_{14})/c_{13} \\ 1 - c_{13}\dot{\varphi}_A, & (1 - c_{14})/c_{13} \le \dot{\varphi}_A < 1/c_{13} \\ 0, & 1/c_{13} \le \dot{\varphi}_A \end{cases} \quad (6)$$

$$g_A^e(\dot{\varphi}_A) = \begin{cases} 0, & \dot{\varphi}_A < -1/c_{23} \\ 1 + c_{23}\dot{\varphi}_A, & -1/c_{23} \le \dot{\varphi}_A < (c_{24} - 1)/c_{23} \\ c_{24}, & (c_{24} - 1)/c_{23} \le \dot{\varphi}_A \end{cases} \quad (7)$$

Letters e, f and r denote extensor, flexor and resistive moments respectively. Positive directions for moments in the joints are given in the Fig. 2. The equations for the knee joint have the same form, only the parameters have different indexes (e.g. c_{3j}, c_{4j}, d_{2j}).

If we adopt $x_1 = \varphi_F$, $x_2 = \dot{\varphi}_F$, $x_3 = \varphi_S$, $x_4 = \dot{\varphi}_S$ as state variables, then by solving the system (1) and (2) with respect to $\dot{x}_2$ and $\dot{x}_4$ we have:

$$\dot{x}_1 = x_2; \quad \dot{x}_2 = P_2 + \sum_{j=1}^{4} G_{2j}u_j$$
$$\dot{x}_3 = x_4; \quad \dot{x}_4 = P_4 + \sum_{j=1}^{4} G_{4j}u_j \quad (8)$$

The terms P_2, P_4, G_{2j}, G_{4j}, j = 1,2,3,4 are nonlinear combinations of state variables and values that are given as inputs into the simulation (i.e. $X_G, Y_G, \ddot{x}_K, \ddot{y}_K$). The values u_j, j = 1, 2, 3, 4 are normalized muscle activations (i.e. 1 - maximal activation, 0 – relaxed muscle).

The cost function was assumed as follows:

$$R(\mathbf{u}) = \int_{t_o}^{T+t_o} \{[(x_1(t) - z_1(t))/z_{1\max}]^2 + [(x_3(t) - z_3(t))/z_{3\max}]^2 + \lambda_1[u_1^2(t) + u_2^2(t)] + \lambda_2[u_3^2(t) + u_4^2(t)]\}dt \quad (9)$$

This cost function comprises a member that is the quadratic error normalized to the maximum value, and two members that are related to the activation levels. The maximum of each of the members in cost function is 1.

The details of transforming the system and the cost function into a discrete form, as well as principles of dynamic programming are described elsewhere [6].

B. Procedure

Here, we present simulation results of two simulations for the same subject; yet, along two different trajectories.

Biomechanical parameters for the simulation are set in the dialog window shown in Fig. 3. The biomechanical parameters describe the geometry inertia and properties of the muscles for a given subject. All of the parameters are described in the section A. Once the set of parameters are entered into the window, this data can be saved in a file for later use. The files are stored in a database of subjects. One of possible, not perfected procedure of determining these parameters is given in [7]. Here we present the case for a subject with moderate spasticity, and muscles that are not atrophied; thus, almost normal force generators (Fig. 3).

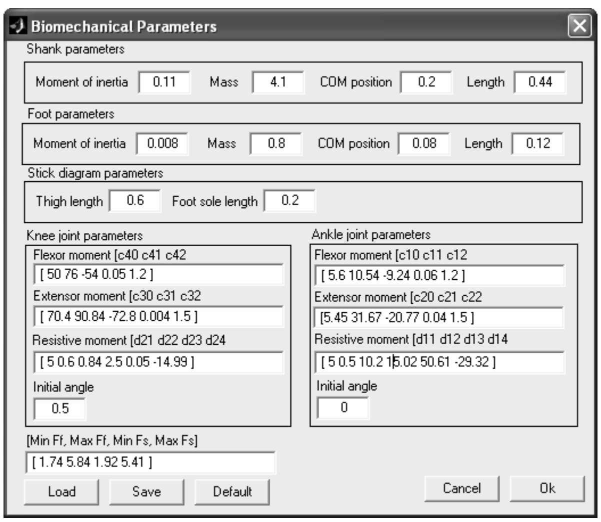


Fig. 3. The dialog window for setting the biomechanical parameters of the model. All parameters have dimensions in international systems.

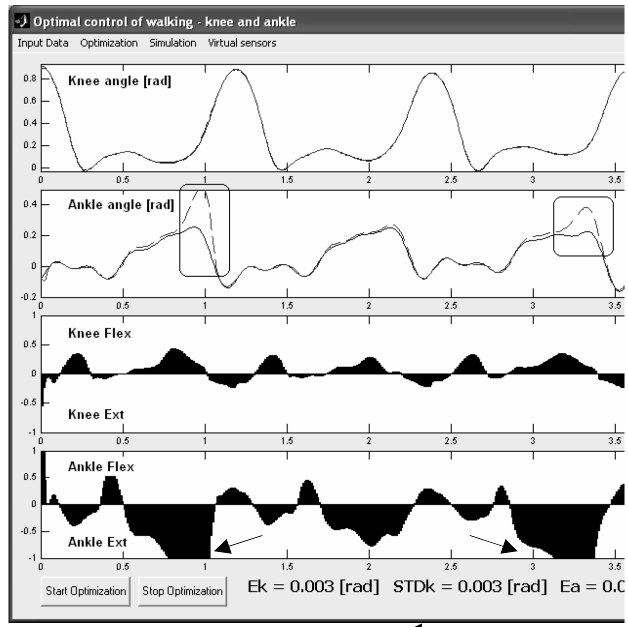
The trajectory data consist of the following: joint angles, ground reaction forces (GRF), knee joint accelerations, and motion of the center of pressure (COP) along the foot in the direction of progression (Table I and Fig. 2). The software is organized in a manner that it comprises a database with several trajectories; thereby, the user can select desired trajectory based on characteristics that he/she prefers.

TABLE I
INPUT DATA

Knee accelerations	$\ddot{x}_K, \ddot{y}_K$
Angles	φ_K, φ_A (joint angles), φ_T (segment angle)
GRF, COP	X_G, Y_G, x_G

The first case presented here uses the trajectory with following parameters: speed of walking v = 1 m/s, $\varphi_{Amax} \sim 0.3$ rad, $\varphi_{Kmax} \sim 0.9$ rad.

The program calculates muscle activations for the equivalent knee and ankle flexor and extensor muscles as well as actual trajectories (joints angles) that are generated by applying these activations as inputs into the model (Fig. 4). The main application window after optimization is in Fig. 4. The actual trajectories are superimposed onto reference trajectories so that tracking error can be assessed visually. The mean absolute errors (e_A, e_K), their standard deviations (σ_A, σ_K), and maximum absolute errors (e_{Amax}, e_{Kmax}) are given as numerical measures of the quality of tracking. Outputs from the simulation are summarized in the Table II.



Desired trajectory - - - Actual trajectory Saturation

Fig. 4. Main application window for the walking with v = 1 m/s. Saturation in muscle activations can be seen on the bottom plot. The saturation is the consequence of the not strong enough muscles for the desired trajectory. Deviations from the reference joint angle for the ankle are encircled on the second panel from the top.

TABLE II
OUTPUT DATA

Generated joint angles	φ_K^*, φ_A^*
Tracking error	e_K, e_A, e_{Kmax}, e_{Amax}, σ_A, σ_K
Muscle activations	u_1, u_2 - ankle flexor and extensor u_3, u_4 - knee extensor and flexor

It can be seen in Fig. 4 that muscle activations for the ankle extensor muscles went into saturation. This result means that the muscles are not strong enough to generate torques needed to ensure tracking of the reference trajectory. One can see significant deviations from the reference trajectory (rounded boxes in Fig. 4).

After optimization, in the simulation window one can visually inspect the walking pattern in the form of a stick diagram (Fig. 6A). The tracking error caused excessive dorsi flexion and abnormal push off in the late stance phase. This behavior is certainly not desirable pattern, and it is likely to lead to instability and possibly fall.

In the second simulation, we selected the walking with the following characteristics: v = 0.8 m/s, $\varphi_{Amax} \sim 0.2$ rad, $\varphi_{Kmax} \sim 0.8$ rad. Simulation results for this case are shown in Fig. 5. There are no saturations of the activations, no significant deviations from the reference, walking is stable and when visualized, pattern looks normal (Fig. 6B).

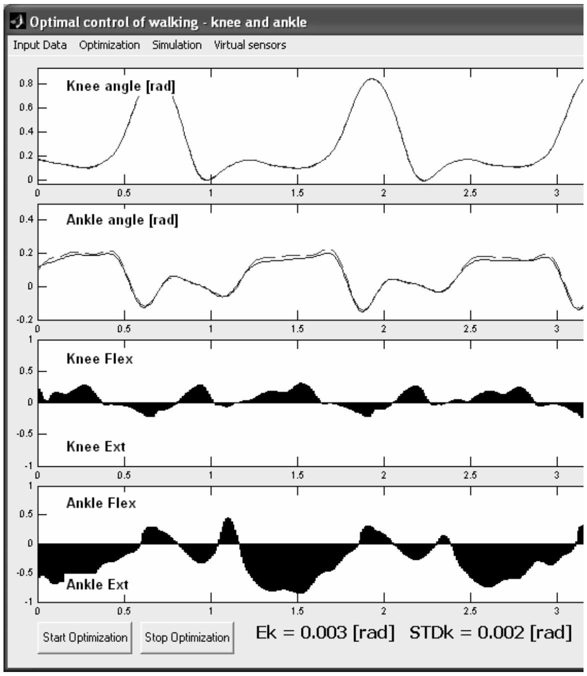


Fig. 5. Main application window. There is no saturation of muscle activations for the velocity of walking v = 0.8 m/s. Tracking error is small.

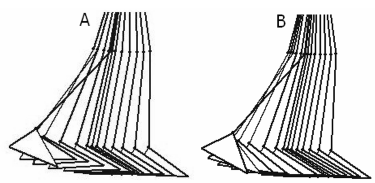


Fig. 6. A) Walking pattern with excessive dorsi flexion B) Normal walking pattern

IV. Discussion

The software tool described in this paper presents user-friendly windows based environment for automated generation of activation profiles. By simulating different trajectories for a given subject, it is possible to find achievable walking patterns with respect to the current state of his/her sensory-motor systems. By virtually experimenting with achievable trajectories of different speeds and different joint angle ranges, the level of effort (i.e. activation) needed for the realization of trajectory can be assessed. The minimal activation is likely to lead to less fatigue, although we are aware that a minimum coactivation of agonist and antagonist can greatly contribute to stability (stiffness control). On the basis of these criteria (i.e. achievability, effort, and quality of tracking) suitable trajectory for the walking of the given subject can be selected.

The overall aim of the research is to develop control for a neural prosthesis (NP) that allows life-like walking of post-stroke hemiplegic individuals. The control that we suggest is applicable to both surface and implantable NP. The work presented here is an important phase in the design of a controller; it allows identification and analysis of plausible trajectories for a given disabled individual. This work follows directly our work from late nineties [6], [8]; yet, includes spatial (3D) dynamic modeling of walking and it is expanded in the dimensions (number of joints). We developed the software package that incorporates dynamic and kinematic properties of the selected subject, that is, individual features of the sensory-motor pathways that are modified due to the injury or disease.

V. Conclusion

In summary, the software presented in the paper is a tool allowing evaluation of a variety of minimal muscle sets that neurologists and physiotherapists believe are necessary to generate functional walking in hemiplegic individuals. The outcome of this analysis is the “muscle set” that can potentially generate functional walking. Once the muscle set is identified, the software package can be used to identify the output (forces) of these muscles. In the case that the simulation demonstrates that it is sufficient to use only FES; then the output of simulation gives the stimulation timings and profiles required for real-time rule base control. The rules will be determined by using machine learning where the inputs are trajectories, and the outputs are the timing and stimulation profiles as described in Jonic *et al*. [2].

If it is revealed that the output requirement for a single muscle in a set is higher than the nominal muscle output, such muscle set will be discarded. The number of potential muscle sets that can generate functional walking will be further evaluated. If the selected muscles, when externally stimulated can not produce the walking, then an external skeleton can be envisioned. This result will suggest the use of hybrid orthosis [1].

References

[1] Popović, D. B. and Sinkjær, T., *“External Control of Movement” in Control of movement for the physically Disabled*, Springer, London, 2000

[2] Jonić S., Janković, T., Gajić, V., and Popović, D. B., "Three machine learning techniques for automatic determination of rules to control locomotion," *IEEE Transactions on Biomedical Engineering* , vol. 46, no. 3, pp. 300-310, 1999.

[3] Shue, G., Crago, P. E., and Chizeck, H. J., "Muscle-joint models incorporating activation dynamics, moment-angle, and moment-velocity properties," *IEEE Transactions on Biomedical Engineering*, vol. 42, no. 2, pp. 212-223, 1995.

[4] Veltink, P. H., Chizeck, H. J., Crago, P. E., And Elbialy, A., "Nonlinear joint angle control for artificially stimulated muscle," *IEEE Transactions on Biomedical Engineering*, vol. 39, no. 4, pp. 368-380, 1992.

[5] Riener, R. and Edrich, T. Passive elastic joint moments in the lower extremity, *Proc. 19th Annu. Int. Conf. IEEE/EMB Society*, Chicago, IL, 4, 1717-1720. 1997.

[6] Oguztoreli, M. N., Popović, D. B, and Stein, R. B., "Optimal control for musculo-skeletal systems," *Journal of Automatic Control*, vol. 4 pp. 1-16, 1994.

[7] Stein, R. B., Zehr, E. P., Lebiedowska, M. K., Popović, D. B., Scheiner, A., and Chizeck, H. J., "Estimating mechanical parameters of leg segments in individuals with and without physical disabilities," *IEEE Transactions on Rehabilitation Engineering,* vol. 4, no. 3, pp. 201-211, 1996.

[8] Popović, D., Oguztoreli, M. N., and Stein, R. B., "Optimal-control for the active above-knee prosthesis," *Annals of Biomedical Engineering*, vol. 19, no. 2, pp. 131-150, 1991.