Abstract – This paper describes the simulation in the MatLab environment and the data acquisition system used for collecting the input to determine the muscle activity profiles needed for electrical stimulation of paralyzed legs. The simulation operates off-line and generates stimulation profiles by minimizing the total muscle activity and tracking errors of desired joint angles (hip and knee). An optimal solution determined by dynamic programming for determining electrical stimulation patterns was used. This software allowed for feasibility testing in walking patterns of individuals with paralyzed or paretic legs (e.g., hemiplegia, paraplegia, etc.).

1. INTRODUCTION

Effective rehabilitation of walking that is assisted with electrical stimulation poses specific optimization problems in the use of paralyzed muscles due to the nature of human-computer interaction. It is the problem of muscle use optimization, and consequently gait optimization, with which the following is concerned.

For patients with a walking disability, normal gait can be difficult to reproduce because of a variety of issues such as modified reflex responses, muscle fatigue, atrophy, joint contractures, etc. [1]. Finding optimal solutions to muscle activations for those patients with a walking disability [2] is essential for gait control. The proposed solution considers that the walking pattern of a disabled individual should be similar to that of a healthy individual. However, the motor systems of individuals with paralysis are modified and inadequate; hence, normal gait patterns cannot be achieved due to the physiological limitations imposed by disability [2]. Therefore, it is of interest to have user friendly software in order to test feasible trajectories and design the first iteration of the stimulation profiles for restoring walking of individuals with paralysis.

The developed simulation, described in this document, can provide information on gait pattern achievability and what stimulation profiles should be applied in order to generate the selected gait pattern. The specific advantage of this software is that the model can be customized to the features of the individual to which the stimulation will be applied. This simulation is not appropriate for real time control; yet, it is adequate for the synthesis of rule-based control that operates in real time [2].

2. OPTIMIZATION METHODOLOGY

The presentation here is limited to a model for solving the problem of optimization of a planar (2D) model consisting of two joints (hip and knee). The ankle joint was considered fixed with the anticipation of the use of an ankle-foot orthosis (Figure 1). The dynamic equations of the system are to be primarily solved – these equations will be omitted in lieu of their general knowledge in the field [3]. The state variables for the system, derived from the dynamic equations, were the angular velocities of the two joints which have been used to solve for muscle torques at their respective joints – it was assumed that each joint can be modeled by one equivalent flexor muscle and one equivalent extensor muscle. The cost function was assumed to be the sum of the activations of all muscles acting at the two joints, and the sum of the normalized quadratic tracking errors. The optimization was resolved by minimizing the cost function.

The discrete-time cost function (Eq.1) [3] can be minimized by differentiation, then by evaluating muscle activations and joint angle trajectories with respect to each other.

\[
R(\mathbf{u}) = \frac{h}{2} \sum_{n=0}^{N-2} r_n (X_n, u_n) + r_{N-1} (X_{N-1}, u_{N-1})
\]  

(Eq.1)

Fig. 1: Model diagram for the simulation [1]

X is the vector of state variables (joint trajectories). Solving the discrete-time system for \( \mathbf{U} = (u_1, u_2, u_3, u_4) \), where \( \mathbf{U} \) is the set of muscle activation levels between zero (no activation) and one (fully activated) at the hip and knee joints, yields the optimal solution. Also, \( r_n \) is dependent on the biomechanical properties of the system. Variable “h” in Eq. 1 is the step in dynamic programming. When the values
for $\hat{U}$ are calculated, some values tend to exceed the threshold of which $\hat{U}$ should stay between zero and one. Values exceeding the upper and lower threshold are set to be either zero or one, depending on whether the value is negative or greater than one.

An important realization within the solution of minimizing the activation levels is called Dynamic Programming. Dynamic programming defines time dependant optimality as: “The best action minimizes the sum of the cost incurred at the current stage and the least total cost that can be incurred from all subsequent stages, consequent on this decision.” [4]. As such, at each interval, subsequent desired trajectories must be calculated to ensure that the system retains optimality. This can be accomplished via a simple discrete-time transformation approximation. Data from the current interval is taken by the approximation and then is used to calculate the necessary coefficients for the following calculation interval.

![Flow chart for the simulation program](image)

**Fig. 2: Flow chart for the simulation program**

We developed a MatLab program that optimizes this solution. The inputs to the simulation were the trajectories describing gait and the body parameters of the test subject. The outputs were the actual trajectories, driving torques, and most importantly the muscle activation profiles. In accordance with this, we developed a new acquisition system that was practical for recording gait data and even more so for rule-based control [1].

Figure 2 shows the flow chart of the simulation program. The simulation program in MatLab was also created with dialog windows and many user changeable features to encourage a user-friendly environment.

3. ACQUISITION SYSTEM

The acquisition system was used for obtaining kinematic and dynamic information during gait that is to be used as input for the aforementioned simulation. The hardware components of the acquisition system are shown in Figure 3.

We used a 6062-PCM CIA A/D card (National Instruments, Austin, Texas), and a laptop computer for data acquisition. We also developed a portable ensemble of an amplifier for the FSR sensors, goniometers, and accelerometers. The FSR (piezoresitive) sensors are fastened to the insole of a shoe and they converted the pressure exerted on the section of feet into a voltage, which was then amplified and sent to the A/D card. Two channels of the A/D card received information from the goniometers (strain gauge full bridge configuration) which were attached to the knee and hip joints and measured relative flexion and extension of neighboring segments during gait. The next three channels of the A/D card received data from the accelerometers – ADXL311 (analog MEMS device), which measured the trunk angle, and the hip accelerations in the horizontal and vertical direction. The entire hardware system was developed at the Laboratory for Biomedical Instrumentation at the Faculty of Electrical Engineering, University of Belgrade. The hardware development was partly contributed by the company “UNA Consulting”, Belgrade.

The data obtained by the accelerometer for the trunk angle where low-pass filtered with a cutoff frequency at 2.5 Hz.

Software for the acquisition system was developed in LabVIEW 6.1. The sampling rate for all channels was set at 500 samples per second. The incoming buffer size is 50000 and we used a first in – first out method [6]. The acquisition
program cyclically reads the data stored in the buffer and determines how many samples to read at a time using the parameter “number of scans to read at a time”. The sample cluster size in this case was chosen to be 200. In addition, the delay in the input buffer was set at 5 µs. It is also important to prevent incorrect data to be written to the buffer, which is accomplished by comparing whether or not the function “number of scans to be read at the time” is greater than “scan backlog”. “Scan backlog” is responsible for indicating at which point the buffer must initiate reading information, where zero indicates the absolute beginning.

Fig. 3: Hardware Components of the Acquisition system

After acquiring the forces acting on the foot during gait, the signals were processed and a resultant vertical force was assessed. The horizontal and vertical forces were estimated by using known relations between them and the total force from literature.

The input data for simulation was recorded by using the following protocol: We placed a two-axis accelerometer at the hip to measure the hip acceleration, and one accelerometer at the pelvis in order to measure the trunk angle. Two goniometers were placed at the hip and the knee and the insole based FSR sensors in a low-cut sneaker. During the recording session the subject was followed by the examiner carrying a laptop used for data acquisition. Gait testing consisted of five level walking trials, of which each had ten strides at natural walking speed. The parameter data was determined by using the technique that was developed by Stein et al. [4].

4. RESULTS

The input for the simulation was three-fold: joint angles, forces acting on the foot and the biomechanical properties of the individual. The data used for the simulation was from a healthy individual. As expected the results of the simulation yielded joint angle trajectories which were very close to the actual trajectories (Figure 4 and Figure 5).

Fig. 4: The desired and estimated knee angle

Fig. 5: The desired and estimated hip angle

Furthermore, the results for the knee and hip muscle activation levels were not saturated and are depicted in Figures 6 and 7, respectively. Note that the flexor and extensor muscle activations are never ‘on’ at the same time in order to eliminate co-contractions [3]. It is, however, possible to include co-contractions in order to provide an increased stiffness in the joints, which may be necessary in some cases [3].

5. DISCUSSION

As indicated, the calculated optimal results of the simulation are approximate to the patterns acquired from the able bodied individual. However, some errors are noticeable in the superimposed traces.

Recall, that the muscles were modeled as being single equivalent flexor and extensor muscles, where in reality there are several layers of muscle groups which fire at different times providing a different response.

The mean magnitude of errors are $\hat{\epsilon}_K = 0.041$ radians and $\hat{\epsilon}_H = 0.016$ radians for the knee and hip joints, respectively. Also, their maximum absolute errors were $\hat{\epsilon}_{K,\text{max}} = 0.113$ radians and $\hat{\epsilon}_{H,\text{max}} = 0.049$ radians. The errors are reasonably small which shows that the approximation was acceptable.

Moreover, the muscle activation diagrams (Figures 6 and 7) show some signs of overlapping, but for the most part are consistent and no co-contractions are present.

The main aim of this presentation is to demonstrate that a tool has been developed for analyzing walking abilities. The
intention is to have this software in public domain, so that others can use it for their functional electrical stimulation studies via the Internet. The software and the hardware will allow for the analysis of specific walking patterns in individuals with decreased physiological resources and to show whether or not they are feasible. The software will also serve to determine the initial set of stimulation profiles that can then, through an iterative procedure, be improved and used in rehabilitation procedures. This simulation package can also serve as a template for discussion with therapists and physiatrists on the interfaces that are most suitable in clinical applications for improved stimulation.

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LITERATURE


This software is a step in the development of a spatial (3D) simulation of the whole leg that is to include abduction/adduction in the hip and dorsi/plantar flexion at the ankle joint. The simulation is of great interest for therapeutic programs in post-stroke hemiplegic patients.