CONTROL OF ACTIVE ORTHOSIS BY EMG SIGNALS

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Abstract: *The article discusses control of active prosthesis and orthosis by using of EMG (electromyography) signals. It is designed an active orthosis for human lower limb and analyzed kinematic parameters of healthy and injured knee joints. There are considered different types of electrodes and found more optimal design. Also are analyzed electrodes location on human body to obtain more useful results. Problems of active orthosis controlling are also discussed.*

Keywords: EMG SIGNALS, ACTIVE PROSTHESIS AND ORTHOSIS, KNEE JOINT, SURFACE ELECTRODES.

1. Introduction

A long time it is known different types of prosthesis that can be used instead of amputated limbs of human. During rehabilitation period of injured limbs usually are used orthosis.

Presently more useful are not so much cosmetic prosthesis [1, 2] how many controlled bioelectrical active prosthesis and orthosis [3, 4]. At designing of this devices kind take place a lot of technical problems: energy consumption, adaptation to the parameters of the human body, noiselessness, cost etc. But the most difficult task is controlling of prosthesis system provide direct connection between human and prosthesis and also feedback from prosthesis to the human body.

Usually there are three different types of prosthesis controlling [5]:

- use of end EMG signals,
- use of central EEG signals,
- use of intermediate signals (registration from nerve fibers).

The most promising seems to be use of central signals but in this research we consider control of active prosthesis and orthosis by help of EMG signals.

2. Active orthosis

Active orthosis to be considered (Fig. 1) consist of two main parts (lower and upper) and their connecting component (6) simulates a knee joint [6]. The main purpose of this device is to allow movement of an injured knee joint according to trajectories of healthy knee joint. By help of the orthosis it is possible to unload the injured knee joint, partially or fully compensate muscular effort required to bend the lower extremities, and also restoration of the joint moving functions during the rehabilitation period.

The base components of the device are shown in the Figure 1. Rotation of orthosis lower part around axis (6) is made by help of servomotor (4), gears (3), ball screw (2) and stock (1). The resulting device will remove the load provided by the human body on the knee joint during the movement. The linear actuator will fully or partially refund the flexion of the knee joint function. The device autonomy will easily let to use it in daily life, and thanks to the flexibility of settings, it would be easy to adjust it to each individual patient [7].

For correct movement of orthosis parts it is necessary to control carefully velocities and accelerations of system components. For this purpose a set of experiments were made by help of St. Petersburg Institute of Cinema and Television. Each person from experimental group performed several tasks, such as walking in a circle and walk in a straight line. During the experiments a Vicon system with ten cameras Bonita was used. Locations of the markers for optical system are shown in Fig. 2 a, (point 4 is on the back of the leg). Used inertial system consist of three accelerometers gy-61 and the board Arduino Mega [8]. The location of the sensors on the leg is shown in Fig. 2 b.

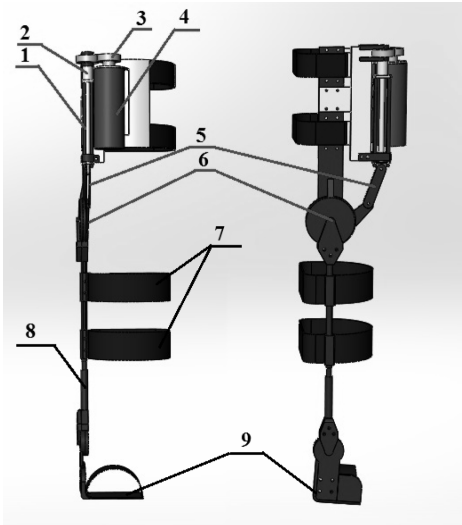


Fig. 1. *Design of the orthosis prototype*
1 – stock, 2 – ball-screw, 3 – gears, 4 – servomotor, 5 – "patella", 6 – flat "knee" joint, 7 – fixing belts with electrodes, 8 – flat "ankle" joint, 9 – "heel-support".

The obtained results were filtered for noise exclusion. Filtration by a Butterworth filter should be implemented first in front direction, then in reverse direction to avoid the phase shift. Results of filtration of data received by sensor 7 (Fig.2) are shown in Fig. 3.

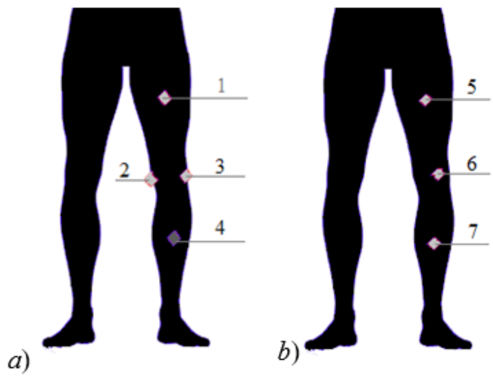


Fig. 2. *Location of the markers and accelerometers*

a - location of the optical system's markers,
b - location of the accelerometers.

Data obtained by inertial system are shown in Fig. 4. Yellow line shows the acceleration obtained during the motion capture, the red and blue lines represent velocity and displacement obtained by integration.

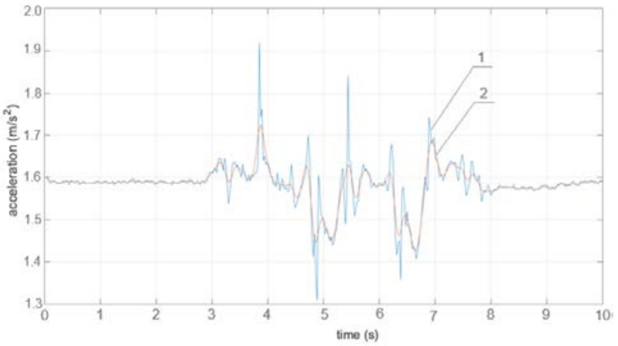


Fig. 3. *Filtering of the data*

1 - data received from the sensor 7,
2 - filtered data.

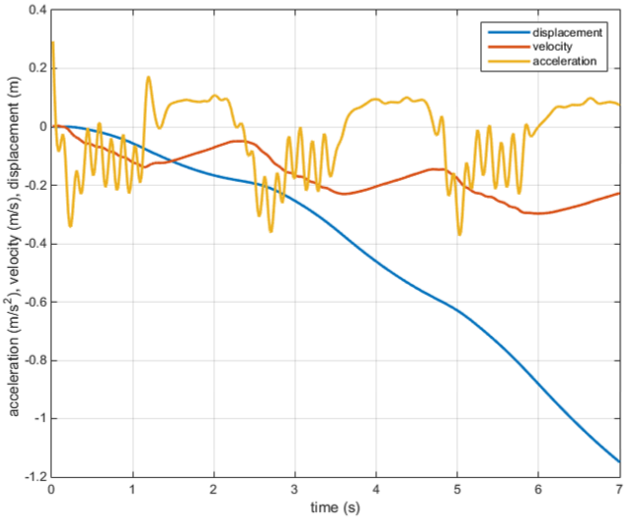


Fig. 4. *Acceleration, velocity and displacement of the point 7 (Fig. 2)*
yellow line – acceleration,
red line – velocity,
blue line – displacement.

On base of obtained results it was developed a model of knee joint and analyzed motion of healthy and injured knees [6].

3. Measurement and filtration

The EMG signals have very small amplitude (20 µV … 2 mV) and the noise amplitude can be much more [9]. Amplitudes of EMG signals are different for different people and signal amplifier should be configurable over wide range.

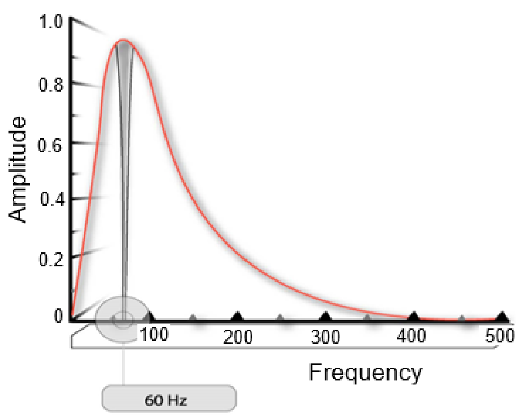


Fig. 5. *Schematic representation of a typical EMG power spectrum [10]*

Frequency of EMG signals are between 10 … 500 Hz. In this case one enough big problem can be power-line (A/C) noise that frequency is between 50 … 60 Hz (Fig. 5). For removing of this noise usually are used a sharp notch filter but the result of this kind filtering is the loss of important EMG signal information [10] and notch filtering should be avoided as a general rule.

The second problem during measurement of EMG signals is influence of other muscles groups. The decreasing of this influence can be obtained by optimal electrode design, distance between electrodes, measurement points etc. Conditions between skin and electrode are also very important for obtaining of signal information. They depend on electrodes polarization effect, electrodes joining, skin conditions, subcutaneous fat etc.

EMG signal amplitude dependence on electrode location on the muscle is shown in the Fig. 6. Correct positioning of the electrode increases important signal information several times.

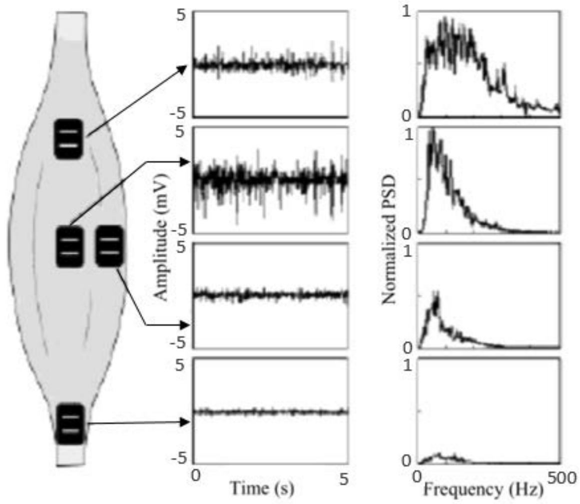
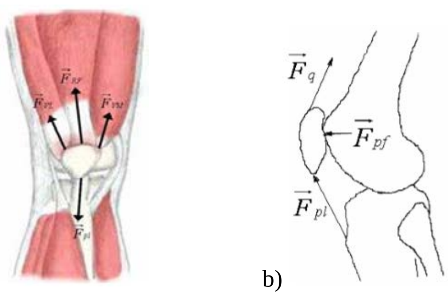


Fig. 6. *Signal amplitude dependence on electrode location [9]*

Signal amplitude is also depend on muscle strength and correct selection of electrode joining points can also increase the signal values significantly. Skeletal muscle cells are formed by the muscle fibers, which constitute the structural contractile units. Each fiber, if excited, has the ability to stretch or contract. The activation of the muscle fiber by nerve endings induces two waves of depolarization that travel at a velocity of 3 to 6 m/s [11]. The internal tissue is electrically conductive. Thus, electrical signals related to depolarization of the fiber can be recorded by electrodes on the skin or muscle. Fig. 7 shows how the different portions of quadriceps acting on the knee extension. F_{pl} is the patellar tendon force, F_q is the resultant quadriceps force, F_{pf} is the tibiofemoral contact force, VM – vastus medialis, VL – vastus lateralis, RF – rectus femoris.



a) b)

Fig. 7. *Quadriceps forces during contraction [11]*

a) schematic representation of the forces involved in quadriceps action,
b) forces acting on the patella.

4. Surface electrodes

One very important part in correct EMG signals measurement is electrode type. There are two categories of surface electrode: passive and active. Passive electrode consists of conductive detection surface that senses the current on the skin through electrode interface. Active electrodes contain a high input impedance electronics amplifier in the same housing as the detection surfaces [12].

One problem at electrode joining on the skin can be usage of paste or gels. These components are widely use in practice for obtaining of stabile contact between electrode and skin. But in this case electrode can slide on the contact surfaces and measurement conditions will change in the time. The solution can be found by special design of electrode [9] where instead of flat surface (Fig. 6 a, upper design) will be used more complicated surface (Fig. 6 a, lower design).

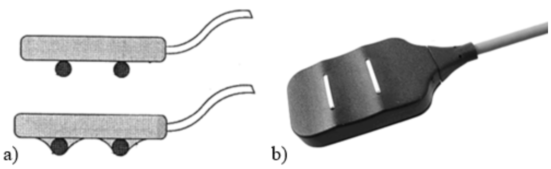


Fig. 8. *Examples of active surface electrodes*

Figure 8 b shows an industrial active electrode in bipolar configurations from Delsys Inc [12]. The spacing between the bars is 10 mm, the length of the bars is 10 mm and the thickness is 1 mm. These electrodes do not require any skin preparation or conductive paste or gels.

5. Control system

Control of active orthosis can be provided by help of dynamic human body system (DHBS) or by direct force control (DFC). In both systems, a hierarchy of two control loops exists. The high-level control loop evaluates EMG signals and the current state of the human body and the orthosis. The output of this loop is the desired motion expressed, as either the desired knee angle or torque. The low-level loop controls the actuator with a PID controller [13].

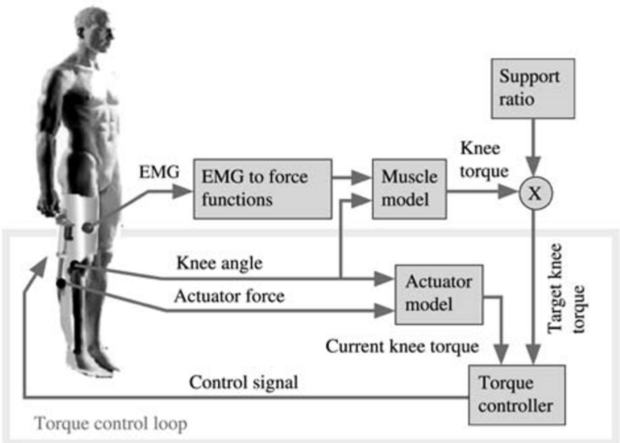


Fig. 9. *Orthosis with the direct force control system [13]*

In the first case (DHBS) the recorded EMG signals are converted into muscle forces and used as an input to the body model together with the joint angles and floor contact information. The model outputs the desired knee angle of the operator, which is passed to the low level control loop as the target value.

In the second case (DFC) the resulting knee torque from the muscle activation is calculated in the high-level control loop from the EMG signals (Fig. 9). The supporting torque is calculated by multiplying torque and by the amplification factor, which can be chosen according to the support required.

Conclusions

Necessary parameters of healthy knee joint were obtained by help of Vicon system. On base of this date a model of the knee joint were developed and movement of healthy and injured knee joints were analyzed. Design and components of an active orthosis are presented. EMG signals measurement system was considered and parameters influencing on measurement results are analyzed.

Acknowledgments

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