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HYBRID ORTHOSIS MECHANICS

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ABSTRACT - Mechanical, i.e. mathematical, model of hybrid orthosis is presented. The example of special, simplified hybrid orthosis, applied to spastic paraplegics, was taken for illustration.

INTRODUCTION

When human lower extremities, or more precisely neuromuscular functions are imapired orthoses can be treated as controlling assistive device. The role of external, technical, system depends on handicapped person status as well as on technical possibilities and full understanding of neurophysiological mechanisms. Hybrid orthosis is an assistive system based on maximum use of remained biological functions of handicapped person and functional normalisation by minimum external skeleton application.

Functional electrical stimulation (FES) and active egzoskeletons are two general conceptually different approaches. FES is based on neurophysiological principles, but external skeleton on robotic methods. Hybrid orthosis (HO) is synthetical result of both methods. From the theoretical standpoint HO is dual mechanism consisting of volontary and externally controlled human body and externaly powered and controlled antropomorphic system.

Neurophysiological mechanism of locomotion can be explained as a set of sequencies related to initial body state, biological sensory input and central/peripheral nerve inervation. Everynatural motor act is obtained by coordinate movement of several body segments. This variable sturcture moves under muscle/actuator action. Motion analysis allows just global i.e. "equivalent muscle" action determinitation. The main muscle characteristic is eferent and/of offer stimulation reaction. Muscle reaction contribution is set of internal processes resulting with muscle shorthaving force.

Locomotion by the authors understanding can be explained as an action in which central nervous system influent on primar motor organisation, and afterwards motion control is repeated at subcortical and lower neuron levels. Changes of locomotion activities are volontary by controlled or by reflex arcs (hazard situation).

Neuromuscular functional deficiencies restricts normal locomotion activities. Type and level of impairment are main facts when orthosis prescription is necessary i.e. it defines biomechanical status of handicapped human in rehabilitation sence of word. Various biomechanical requirements (Scwhirtlich, Popović, 1984) have influented hybrid orthosis syntehsis.

BIOMECHANICAL HYBRID ORTHOSIS MODEL

General model of hybrid orthosis for lower extremities is adopted on the basis of simplified neuromuscular locomotion activities (Popović, Schwirtlich, 1984).

Main characteristics of the model are:

- actuator duality - FES and cybernetic actuators (CA)
- soft interface
- full modularity
- bionic sensors.

Control based on state space approach, external skeleton design, CA characteristics determination, interface forces and pressures evaluation are impossible without numerical simulation of the presented model.

If, external skeleton (SFMO) and CA are adopted and if, pressures/ /forces are not of interest simulation is not needed exept for control purposes. When artificial reflex arcs control method (Tomović, 1984) is applied calculations are pure "l´art pour l´art". The author him-self believes that local joint control needs dynamic feed-back. Parameter adaptation can be achieved, but artificial locomotion over all control is compatible to hyman only when AI methodology is applied.

Dual actuators are built in model in order to express remained human functions in paralel action with cybernetic actuator if necessary. CA are ment just for partial external energy/power functioning. Type of FES will not be discussed in this paper. It has been accepted that action exists when FES is applied.

Bionic sensors are for system state estimation, i.e. they are artificial reflex arcs basic elements (Tomović, 1984).

Soft interface and modularity are basic characteristics of SFMO. (Popović, 1981).

For simulation purposes mechanical model is developed. For space analysis general model of rigid body systems with viscoelastic and joint connections is developed (Ito, 1981). Planar analysis is done as well (Popović, 1984).

Motion is described with:

$$(1)\quad \begin{aligned} &(a)\ m_i\ddot{x}_i = X_{i-1} - X_i + S_{ix} \\ &(b)\ m_i\ddot{y}_i = Y_{i-1} - Y_i - m_i g + S_{iy} \qquad i=1,2,\dots,n \\ &(c)\ J_{c_{iz}}\ddot{\phi}_i = \alpha M_i + X_i(y_i - y_{c_i}) + X_{i-1}(y_{c_i} - y_{u-1}) - Y_i(x_i - x_{c_i}) - \\ &\qquad - Y_{i-1}(x_{c_i} - x_{i-1}) + \mathcal{M}_i, \end{aligned}$$

$$(2)\quad \begin{aligned} &(a)\ m_i^*\ddot{x}_i^* = -X_i^* - S_{ix} \\ &(b)\ m_i^*\ddot{y}_i^* = -Y_i^* - m_i^* g - S_{iy} \\ &(c)\ J^*_{c_{iz}}\ddot{\phi}_i^* = (1-\alpha)T_i - \mathcal{M}_i + X_i^*(y_i^* - y^*_{c_i}) - Y_i^*(x_i^* - x^*_{c_i}), \end{aligned}$$

$$(3)\quad \begin{aligned} &(a)\ m^*_{i+1}\ddot{x}^*_{i+1} = X_i^* - S_{i+1,x} \\ &(b)\ m^*_{i+1}\ddot{y}^*_{i+1} = Y_i^* - m^*_{i+1} g - S_{i+1,y} \\ &(c)\ J^*_{c_{i+1,z}}\ddot{\phi}^*_{i+1} = -(1-\alpha)T_i - \mathcal{M}_{i+1} + X_i^*(y^*_{c_{i+1}} - y_i^*) - y_i^*(x^*_{c_{i+1}} - x_i^*), \end{aligned}$$

$$(4)\quad \begin{aligned} &(a)\ S_{ix} = c_i(x_i - x_i^*) + r_i(\dot{x}_i - \dot{x}_i^*) \\ &(b)\ S_{iy} = c_i(y_i - y_i^*) + r_i(\dot{y}_i - \dot{y}_i^*) \\ &(c)\ \mathcal{M}_i = c_i'(\phi_i - \phi_i^*) + r_i'(\dot{\phi}_i - \dot{\phi}_i^*). \end{aligned}$$

X,Y,M,T are joint forces and torques, S, $\mathcal{M}$ interface forcee-viscoelastic connection, m, J inertial properties, g gravity acceleration, x, y position vector, r, c, stifness and elasticity coeficients (Fig. 1) sign is for orthosis.

With geometrical equations (5) $x,y = f(\phi)$, and $x^*,y^* = f^*(\phi^*,x_0^*,y_0^*)$ numerical simulation can be done. Recursion from the last link in open kinematic chain (1) allowds joint torque $M(\phi)$ determination, when $\alpha = 1$. is called HO parameter, and it may vary for 0 to 1. When zero, no muscle action occurs, external torques T can be determined via recursion. When $\alpha\varepsilon(0,1)$ paralel action of muscle and external actuator (CA) in orthosis joint results with motion. Unsuficient muscle contraction may exist if volontary control (muscular impairment, fatique) or external control (neural impairment) is applied. If dual mechanism is involved, because, of variable muscle characteristics it might be impossible so drive control signals for (M,T) realisation.

$$\begin{vmatrix} u_B \\ u_M \end{vmatrix} = C \begin{vmatrix} M \\ T \end{vmatrix}, \qquad \dot{x} = f(X,M,T)$$

Simulation can be done by transformation to first order nonlinear normalysed system of diferential equations:

(6) $f_1(\phi_i^*,\dot{\phi}_i^*,M_i x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (1c)

(7) $f_2(\phi_i^*,\dot{\phi}_i^*,\phi_{i+1}^*,\dot{\phi}_{i+1}^*,x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (2a) + (3a)

(8) $f_3(\phi_i^*,\dot{\phi}_i^*,\phi_{i+1}^*,\dot{\phi}_{i+1}^*,x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (2b) + (3b)

(9) $f_4(\phi_i^*,\dot{\phi}_i^*,\ddot{\phi}_i^*,T_i,x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (2c)

(10) $f_5(\phi_{i+1}^*,\dot{\phi}_{i+1}^*,\ddot{\phi}_{i+1}^*,T_i,x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (3c)

(11) $f_6(\phi_{i+1}^*,\dot{\phi}_{i+1}^*,x_0,y_0,\dot{x}_0,\dot{y}_0) = 0$ (1c)

By introducing $\phi_i^* = x_1$, $\dot{\phi}_i^* = x_2$, $\phi_{i+1}^* = x_3$, $\dot{\phi}_{x+1}^* = x_4$, $x_0 = x_5$, $y_0 = x_6$ system (12)-(14) describes dynamic behaviour of hybrid orthosis

(12) $\dot{x}_\alpha = f(x_\alpha)$ $\quad |x_i| = |x_1 x_2 x_3 x_4 x_5 x_6|^T$

(13) $T_\alpha = g(x_\alpha)$ $\quad \alpha = 1,2,\ldots,n-1$

(14) $M_\alpha = h(x_\alpha)$

If joint torques are known equations (6) and (11) are

(15) $f_1(\phi_i,\phi_i^*,x_0,y_0,\dot{x}_0,\dot{y}_0,\phi_i,\phi_i^*,\phi_i^*) = 0$

(16) $f_6(\phi_i,\phi_i^*,x_0,y_0,\dot{x}_0,\dot{y}_0,\phi_i,\phi_i^*,\phi_i^*) = 0$

and system (12)-(14) becomes

(17) $\dot{x}_\alpha = f(x_\alpha)$, $|x_i| = |x_1,\ldots,x_{10}|^T$, $\alpha = 1,2,\ldots,n-1$

where $x_7 = \phi_i$, $x_8 = \dot{\phi}_i$, $x_9 = \phi_{i+1}$, $x_{10} = \dot{\phi}_{i+1}$.

Runge-Cutta method is very convenient for simulation. Numerical derivation when gait analysis is of interest, i.e. kinematic state $\phi_i(t)$, $i=1,2,\ldots,n$ is known is special problem. Fourier expansion seems to be most appropriate interpolation method when $\phi_i(t)$ are measured, because derivation and Lancosz correction coefficients application makes possible to calculate ground force reaction in range comparable with force plate measurements.

Gait data filtering measurements gives best results, but orthosis controller has to be self contained, real time, portable device.

HYBRID ORTHOSIS FOR SPASTIC PARAPLEGIC ANALYSIS

Numerical simulation of the above presented mathematical model is possible for different orthosis configuration and medical indication. The very simple, realysed, hybrid orthosis is for spastic paraplegics. Biomechanical requirements for such an indication are limitted to maintenance of upright posture and slow walking. Spastic paraplegic results with uncountrolled, lower limbs spastic muscle contraction, usually with hip hyperextension. Hybrid orthosis has to compensate abdominal stability as well as lower extremities supporting role, and to initiate movements of the thigh and shank with sufficient foot control. Precisilly such an process involves about 30 muscles and a number of neural centers controlling 30 degrees of freedom per leg.

Ljubljana REC experience with FES standing and walking proves possibility of just two channel stimulation per leg in order of quasistatic gait synthesis. Muscle fatiques and some other undesirable effects are present, hip control is lacking and under elbow contches use is necessary. SFMO application in RC Belgrade implies just partial external energy, because of soft interface. Active SFMO hips (two dimensional control - - flexion-extension, abduction-adduction) for maintenance of balance, quadriceps stimulation for knee extension, peroneal stimulation for knee and dorsi flexion, knee elastic CA and elastic ankle joint SFMO are adopted for the mentioned indication area. The same configuration can be applied by some other types of impairment.

Gait cycle of such an biomechanical system (Fig. 2) differs from normal gait.

Spastic hip contraction, if existing, may need stimulation control. Multichannel subentaneous or implant stimulation must be used for these purposes. If stimulation control of posture is obtained just above knee (two modules) SFMO has to be applied (Schwirtlich, Popović, 1984). Role of the external skeleton is just safety and standing support. CA is then applied just as elastic-braking device (Popović, 1983).

Control signals are adopted in form

$$u_T = \sum_i U_{T_i} |h(t-\tau_i)-h(t-\tau_{i+1})|, \quad \tau_{i+1}>\tau_i, \quad i = 0,n \quad \tau_o = 0, \quad \tau_n = T_s$$

$$u_M = U_M |h(t)-h(t-T_s)|$$

Two different actuator response are built in the model.

Nonlinear transmission with DC motors and tetanic muscle contractions are actuator characteristics. For adopted HO configuration, neglecting hip stabilisation (multi channel stimulation or external skeleton four actuators control), two channel stimulation (u_{M_1}, u_{M_2}) and two CA channels (u_{T_1}, u_{T_2}) per leg are input signals.

Optimal control synthesis consist of U_i, τ_i, estimation. Minimum energy consumption optimisation criterion is introduced. Minimization of unconstrained functional

$$G = \sum \int_0^T g_i(x_i,u_i)dt$$

where power expenditure is $g_i(x_i,u_i) = e_i-\pi_i-\tau_i$, $e_i = | M_i d\omega_{r_i}| + |T_i d\omega^*_{r_i}|$ "input" power and ${}_i+{}_i$ kinetic-potential power varration $(\pi_i+\tau_i = m_i |u_{c_i} dv_{c_i} + g\dot{y}_{c_i}|)$ is optimisation task (Popović, 1983). Applied deformation method implies minimization of functional

$$G(X(t),k_i) = \min_{u\epsilon U} |k_i \int_0^T F(X,U)dt+(1-k_i) \int_0^T R_i(X,U)dt+G(X(t+t),k_i)|$$

where $R_i(X,U)$ is known family of continual functions and k_i parameters $(k_o=0,\ k_{i+1}>k_i,\ k_N=1)$. It is possible by selected varration of k_i and R_i

to realize uniform convergence in the control space. A simple solution can be determine for different gait aproximations (Fig. 2).

CONCLUSION

Research on and experience with lower limb orthosis open an interesting intersection of neurophysiology and biomechanics and can lead to more functional replacement of imapired quality of locomotion. Hybrid orthoses synthesis is an effort which attempt a synergy between the man and the orthoses, coupling human action to the mechanical response of external, artificial system. The target of this research is identification of the strategy through which external control accomplish an important vital function. Mechanics of hybrid orthoses reflects to all constitutive elements of locomotion i.e. control, interface analysis, energy expenditure.

It is necessary to point out that practical hybrid orthoses design is guarded by our growing knowledge of the neural increasingly control of movement and by wide development of stimulation and robot technology.

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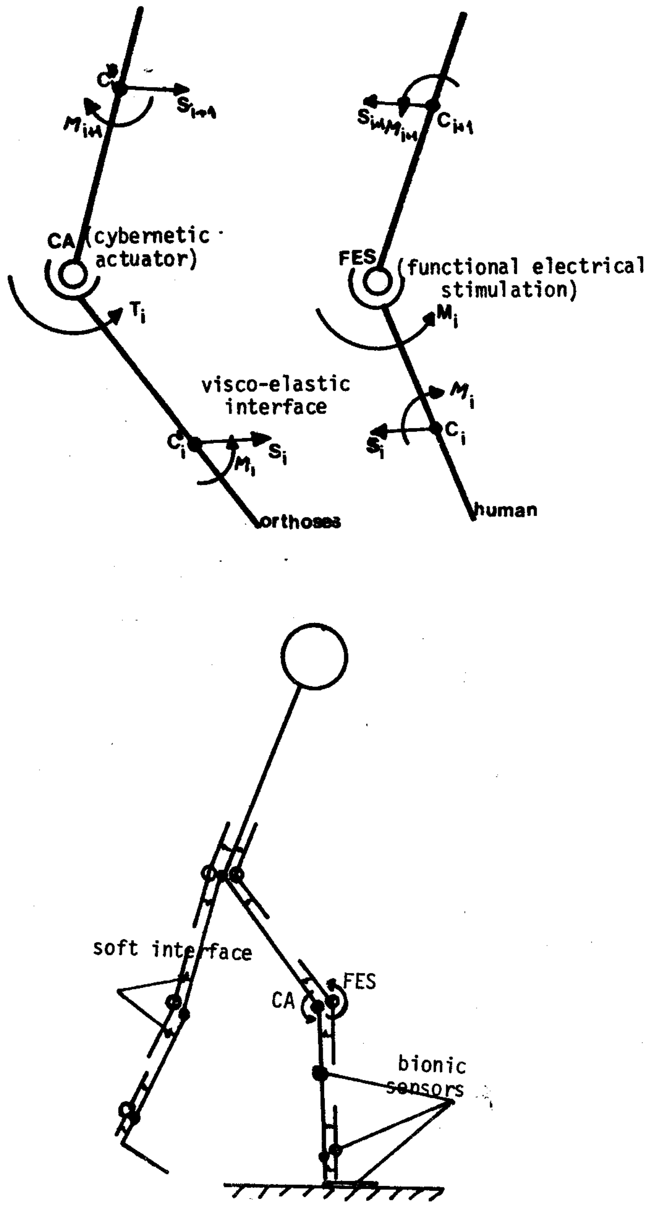

Fig 1. Mechanical model of the hybrid orthoses

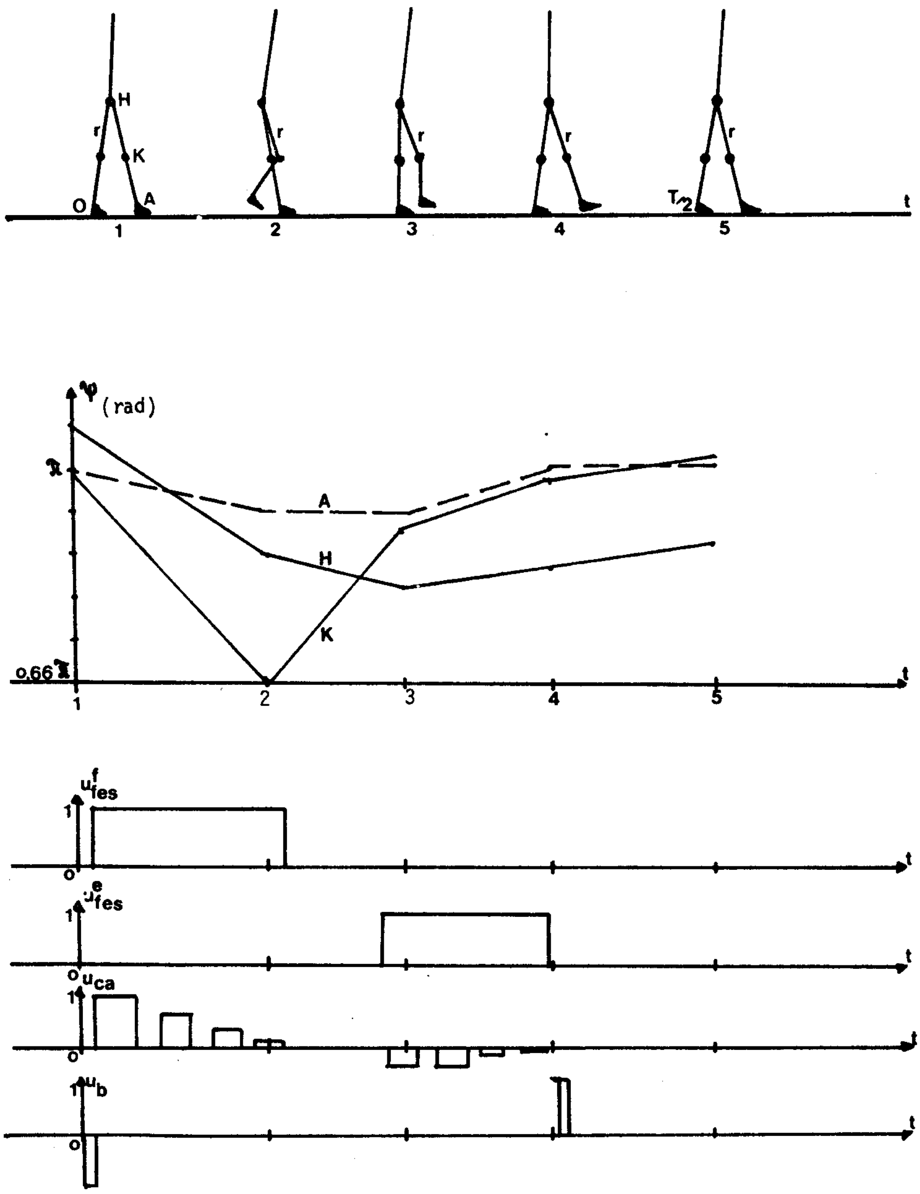

Fig 2. Gait cadence and control signal for HO