INVESTIGATIONS ON THE CONTROL OF LIMBS BY MEANS OF THE FUNCTIONAL ELECTRICAL STIMULATION OF MUSCLES

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Abstract

Problems occuring during the development of control systems within orthoses for paralyzed patients are analysed. Experimental results obtained from a simple closed-loop position control system showed that more sophisticated control strategies are necassary to remove the deficiencies of actual control systems. With regard to the implication of appropriate control strategies a model for the "electrically stimulated muscles-limb-load" system is derived. A hierarchical control strategy involving adaptive features for the Functional Electrical Stimulation is then described and, in view of the implementation on a digital computer, a realization of this strategy is presented. The effects of different modulation techniques (pulse amplitude modulation and pulse frequency modulation) on the feasibility of control systems are also investigated. Pulse frequency modulation systems are found more robust than pulse amplitude modulation systems.

1. Introduction

The first successful application of the Functional Electrical Stimulation (FES), however without control terms, was executed by Liberson and others (1). Since then several investigations have been carried out on this field concerning fundamental problems (2), (3) as well as the aspects of application (4), (5), (6).
So far the applications are introduced mainly in two areas: the provision of functional movements in the hand of quadriplegic patients C4-C6 (7) and the improvement of the gait of hemiplegic patients (8). For all these applications accurate, regulated control of electrically elicited muscle contractions will improve the execution of the movements to a great extend . Therefore the control problem plays an important rolle within the FES. The investigation on this problem was however not consistently pursued. In the studies carried out hitherto either a closed loop control of position (9) or a closed loop control of force (10) was developed based on simple control models. So far the control loop of position was mainly closed by the feedback pathway of the eyes. The applications were consequently time expensive and troublesome for the patients. The control was itself inaccurate, its efficacy instable.
The control of but one of the two essential variables: the position and the force is not sophisticated for many applications and problems as:

- the time variance of the controlled system, i.e. the muscles-limb-load system (MLL), due to the changes in efficacy of stimulation arising from small shifts of stimulating electrodes during movements and to the fatique of muscles
- the nonlinear imput-output relationship of the electrically stimulated muscles

are further reasons of the insufficiencies of the control (9),(10). These effects also severely impede the development of appropriate control systems.
In order to solve these problems one should consider them fundamentaly. Recently some progress was made mainly in circumventing the fatique of muscle (7), (11). The variations of muscle responses due to electrical stimulation could also be diminished by using intramuscular electrodes (12). Crago et.al. (10) have presented in 1980 a model for the control of force that involves different modulation techniques improving the linearity of the control. This paper discusses the control aspects of the problem more fundamentaly which results in a working model to control the execution of the muscle contraction in relation to the given input.

2. A simple closed-loop control of the limb position

In order to get an insight on the methods of FES and to check how far a control system based on simplifying assumptions can be employed in some simple applications, a closed-loop control system for limb position was investigated (13). Fig. 1 shows the principal procedures of this system.

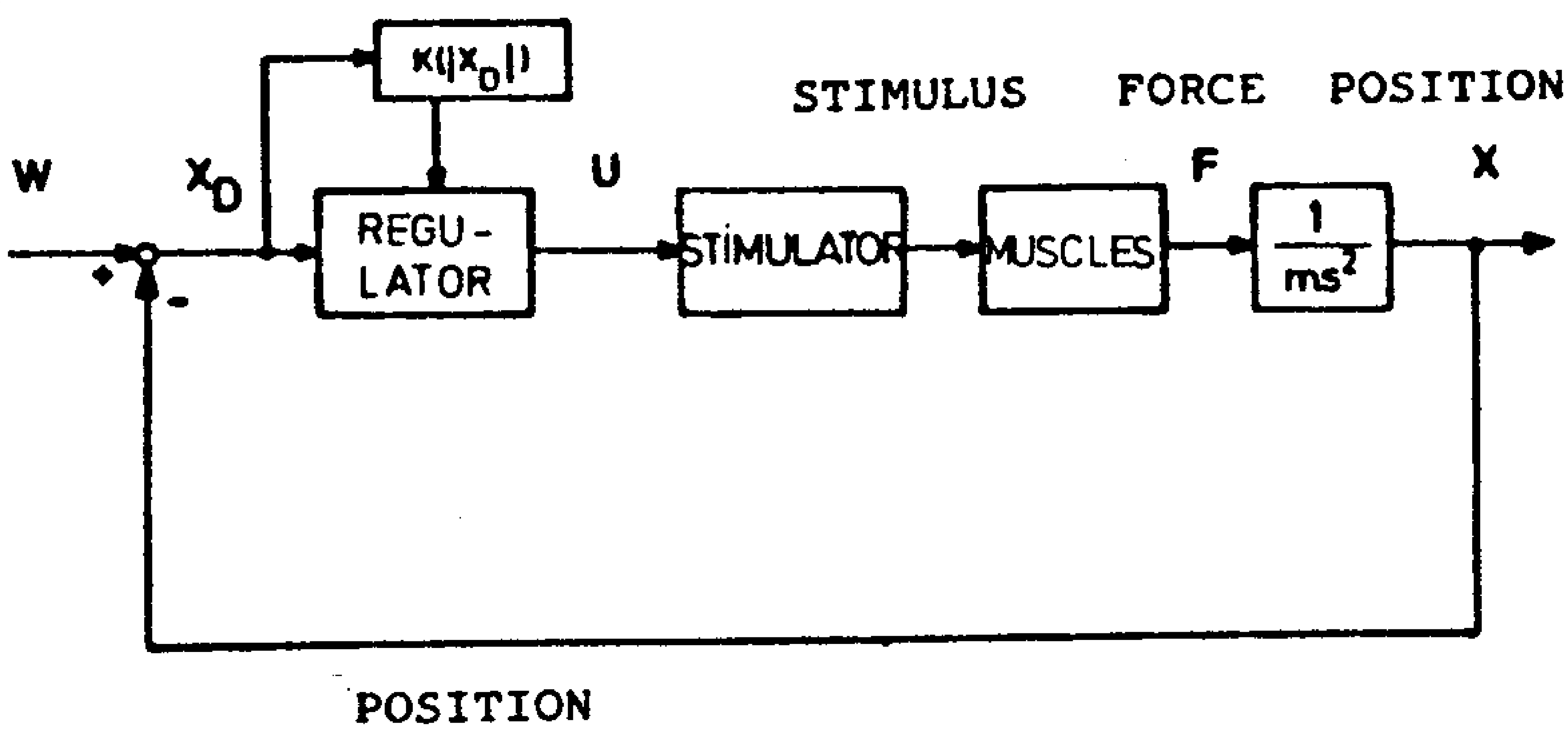

Fig. 1: The block diagram of the simple closed-loop control system

Assumptions

The following assumptions were made:

- The antagonistic muscle pair has a linearized input-output relationship. The stimulus is the input, the elicited force is the output. Its transfer function is given by:

$$G_p(s) = \frac{V}{(1 + s.t_k)^2} \qquad (1)$$

  where s is the complex frequency, V is the gain factor ant t_k the contraction time of the muscle.

- The system "limb-external load" is assumed to be an undamped mass m:

$$G_L(s) = 1/(ms^2) \qquad (2)$$

Methods

Experiments were carried out on the gastrocnemius and tibialis anterior muscles in the hind leg of anaesthetized adult cats. The hind leg of the cat was fixed by a mechanical frame so that only the foot remained movable in one direction. The angle of the ankle joint was measured by an electrogoniometer The output of the controller (a digital computer) was multiplexed, digital-analog converted and drove the multichannel stimulator. The pulse amplitude modulation (PAM) and pulse frequency modulation (PFM) were simultaneously performed in all experiments. The amplitude of the rectangular pulse ranged from 0 to 10 mA, the frequency from 10 to 35 Hz. Intramuscular stimulating electrodes were used.

Closed loop systems with proportional plus integral controllers were investigated:

$$u = K.x_D + \frac{1}{T_I}\int x_D.dt \qquad (3)$$

$$K = c.\left| x_D \right|$$

u : output of the controller
x_D : difference between the command w and the position feedback x
K : proportional gain factor
T_I : time constant of the integrator
c : constant

Results

Sinusoidal command input frequencies up to 1.2 Hz were applied to assess the performance of the control systems. The results can be summarized as follows (Fig. 2):

- The most cases of the control were stable. Some data exhibited rather large deviations between x and w.
- The position reversals were sometimes unsatisfactory.
- Too high a proportional gain caused oszillations in the systems.

The results, similar to those of other investigations (9), (10) are in fact not suprising because of the reasons mentioned in the introductory section. Moreover multielectrode stimulation in a muscle is necessary for a larger control range.

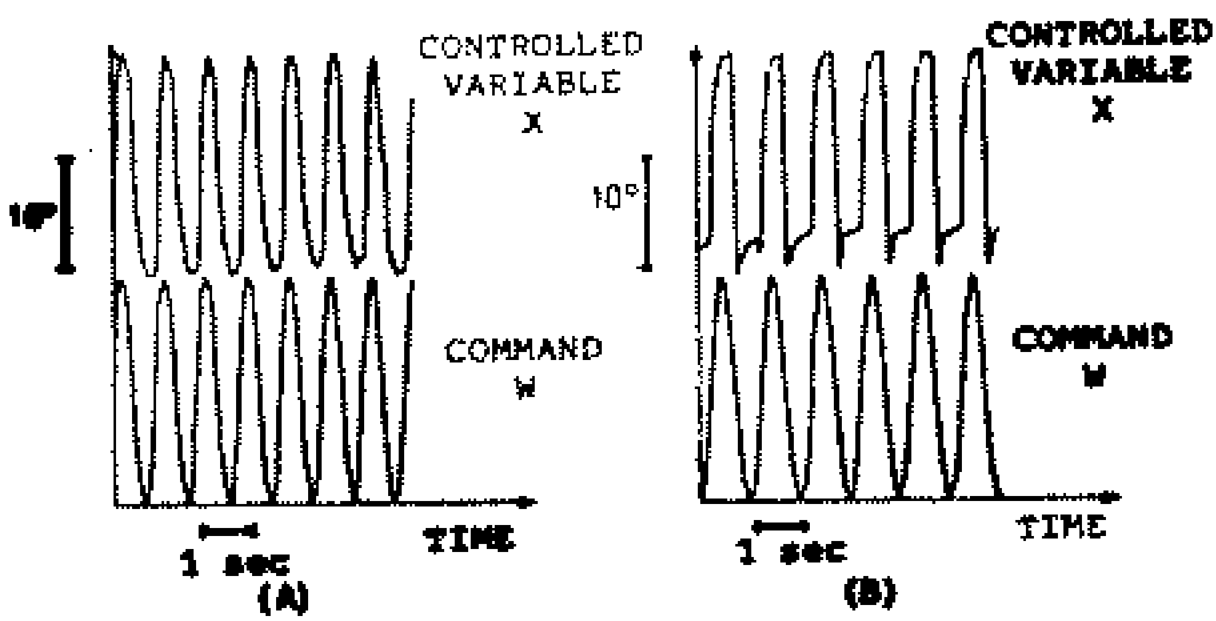

Fig. 2: Results from the simple control system.
(A) satisfactory results
(B) unsatisfactory position reversals.

Consequently more complex control strategies, which take into account the above mentioned effects, should be develop. Intuitively adaptive control is best suited for diminishing the influences of the time variance and nonlinearities in the system. Moreover for clinical applications it is necessary to automatically adjust and readjust the controller parameters during operation.

3. A model for the MLL system

The model is assumed to have a nonlinear static part and a linear dynamic part with a dead time. A model must be precise enough to yield sufficient control effects, but need not be too complex with regard to evaluation during operation.
Our own experiments have shown, that the nonlinear part consists of a dead zone, a nearly proportional zone and a saturation zone for both modulation techniques.
The linear part can be described by a mechanical system as shown in Fig. 3.

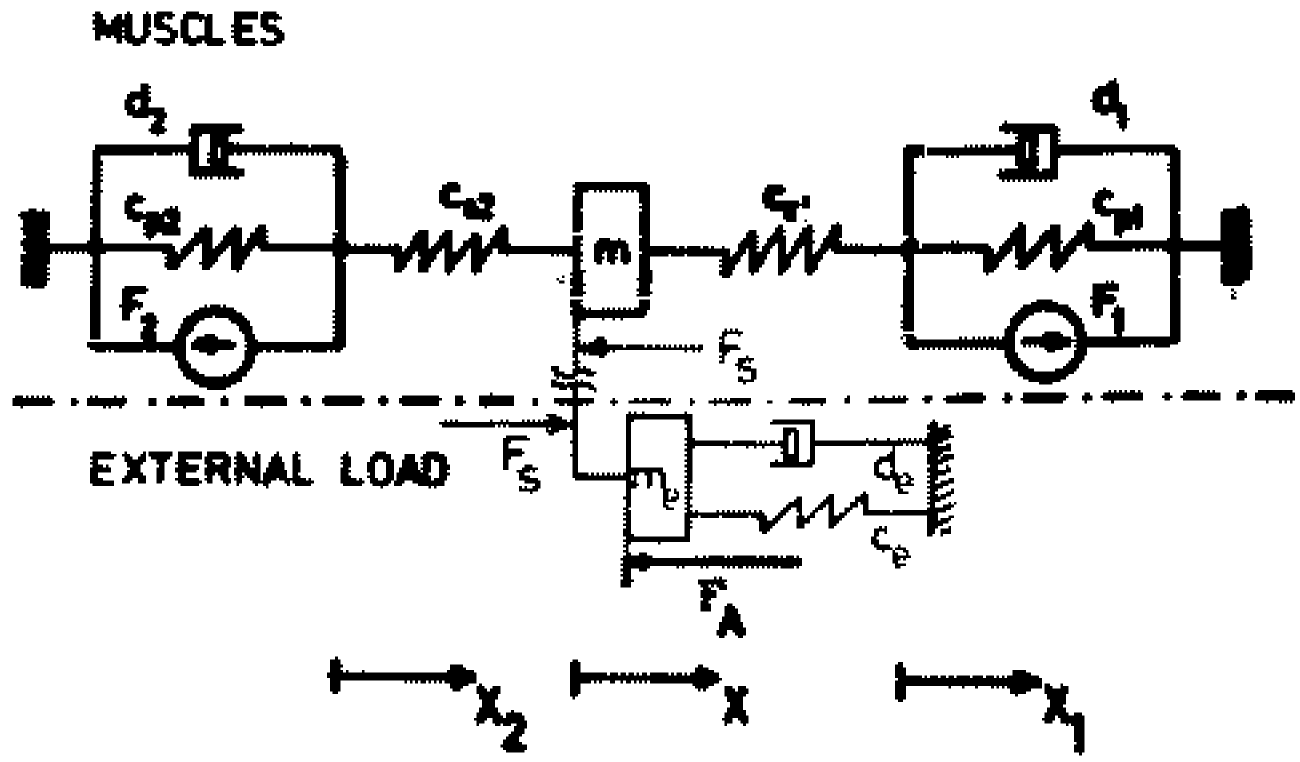

Fig. 3: Mechanical model for the MLL system.

The antagonistic pair and the limb are characterized by their mass m, their serial and parallel elastic components c_{si}, c_{pi} $(i = 1, 2)$ and their damping components d_i $(i = 1, 2)$. Elasticity and damping are assumed to depend on the actual neuronal muscle innervation (14). The input F_i $(i = 1, 2)$ is the internal force generated by the contractil elements. The external load is characterized by its mass m_e and its elastic and damping components c_e and d_e. F_A is an additional external force. By assuming a rigid connection between limb and load the same displacement x is valid for both:

$$d_1\dot{x}_1+(c_{p1}+c_{s1})x_1-c_{s1}x = F_1 \tag{4}$$

$$d_2\dot{x}_2+(c_{p2}+c_{s2})x_2-c_{s2}x = F_2 \tag{5}$$

$$c_{s1}x_1+c_{s2}x_2-(c_{s1}+c_{s2}+c_e)x - d_e\dot{x}-(m+m_e)\ddot{x}=F_A \tag{6}$$

Equ. 4-6 yield a 4th order differential equation in x, where the generator force F_i and the external F_A are the excitation input functions. By assumption of the symmetry conditions:

$$c_{s1}/d_1 = c_{s2}/d_2 \tag{7}$$

$$c_{p1}/d_1 = c_{p2}/d_2 \tag{8}$$

and further $c_e = 0$; $d_e = 0$ it becomes a 3rd order differential equation:

$$\dddot{x}+\tilde{b}\ddot{x}+\tilde{c}\dot{x}+\tilde{d}x= \tilde{V}_P F_1-\tilde{V}_P F_2-\tilde{V}_F\dot{F}_A-b\tilde{V}_F F_A \tag{9}$$

The assumtions 7 and 8 lead to the fact that muscles of the same type have the same contraction time, though their gains can be different from each other.

The influence of the external load on the limb is involved in the interacting force F_S which effects contrary to the limb movement. Replacing the external load and the force F_A, F_S acts exactly in the same way on the limb as the load and F_A do together. By using of F_S instead of F_A Equ. 9 retains its validity, however with another set of coefficients which are determined only by the parameters of muscles and limb:

$$\dddot{x}+b\ddot{x}+c\dot{x}+dx= V_P F_1-V_P F_2-V_F\dot{F}_S-bV_F F_S \tag{10}$$

where $F_S = F_A + F_L$

$G_L(s) = \frac{x(s)}{F_L(s)}$; $G_L(s)$: transfer function of the load

It follows from Equ. 10 that the position of the limb (i.e. that of the load by a rigid connection between the limb and the load) may be characterized by two relationships: that between electrical stimulation and limb position (G_P) and that between external force and position (G_F). This consideration is important for further implications.

The model according to Equ. 10 is shown in Fig. 4. It is an extension of the linear model proposed by Magdaleno (14) for small perturbances. Under certain circumstances the representation according to Equ. 10 has some advantages. In this way the model parameters depend only on the internal state of muscles and limb. Even by load variations the identification need not be repeated. It allows also the separation of the MLL system into two relationships.

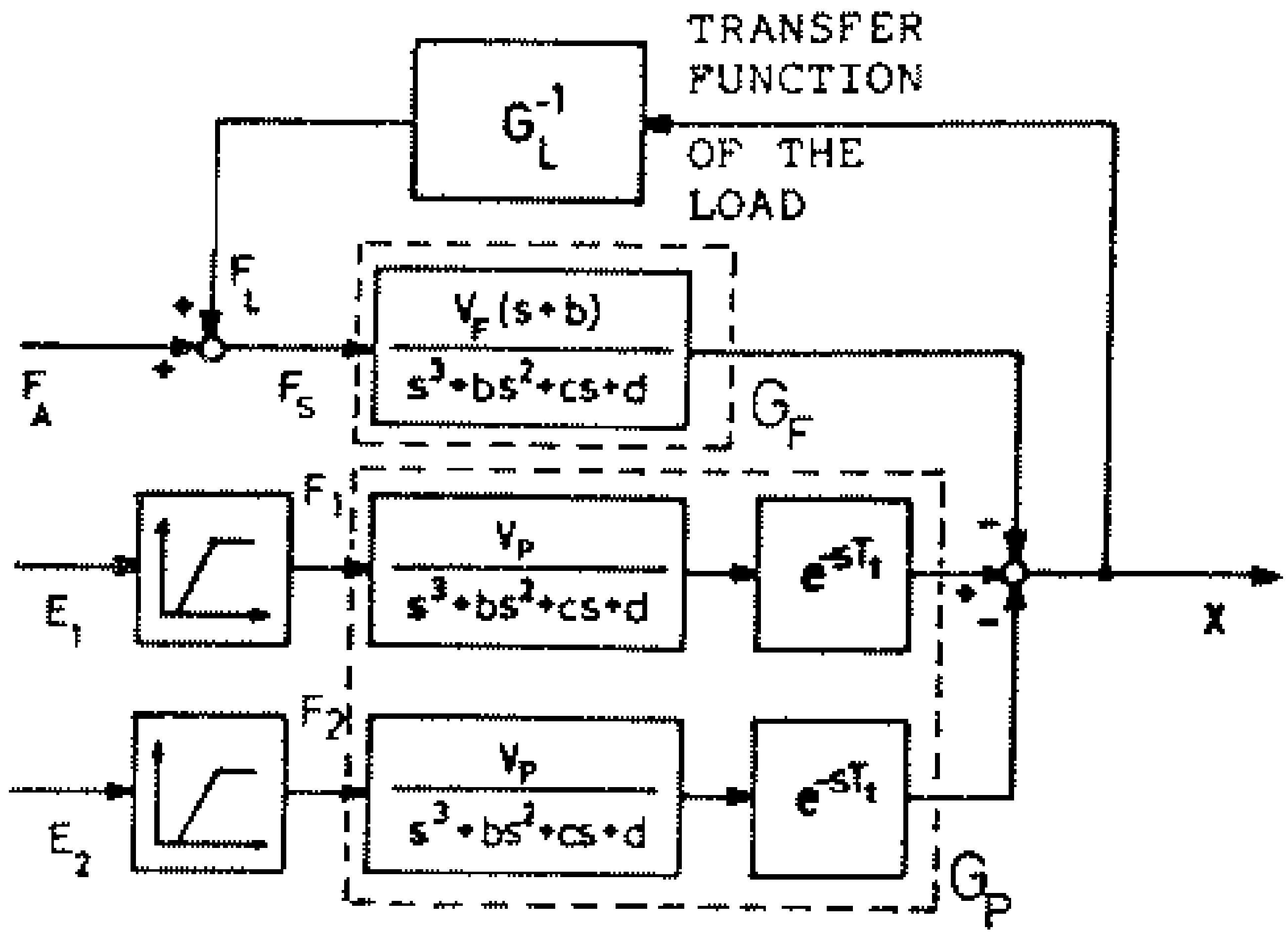

Fig. 4: The MLL model

4. Adaptive control of limb movements

Based on the above mentioned model the principal features of an adaptive control strategy which may compensate the influences of time variance in the system can be outlined. Fig. 5 shows the block scheme of the low and the intermediate levels of this hierarchically organized control system.

- The lowest level of the hierarchy is a closed control loop where the limb position and the force applied to the limb are inputs in the controller. This level should guarantee the stability of the limb movement and compensate small disturbances. The control algorithms at this level must be able to actuate multichannel stimulators, further more they should be easy to modify.
- The next level deals with the adaptive features of the system:system identification, synthesis and modification of the control algorithms of the lowest level.
- The highest level deals with planning and initiation of intended movements and additionally it is involved with the supervision of the entire system. This level is not investigated in this study.

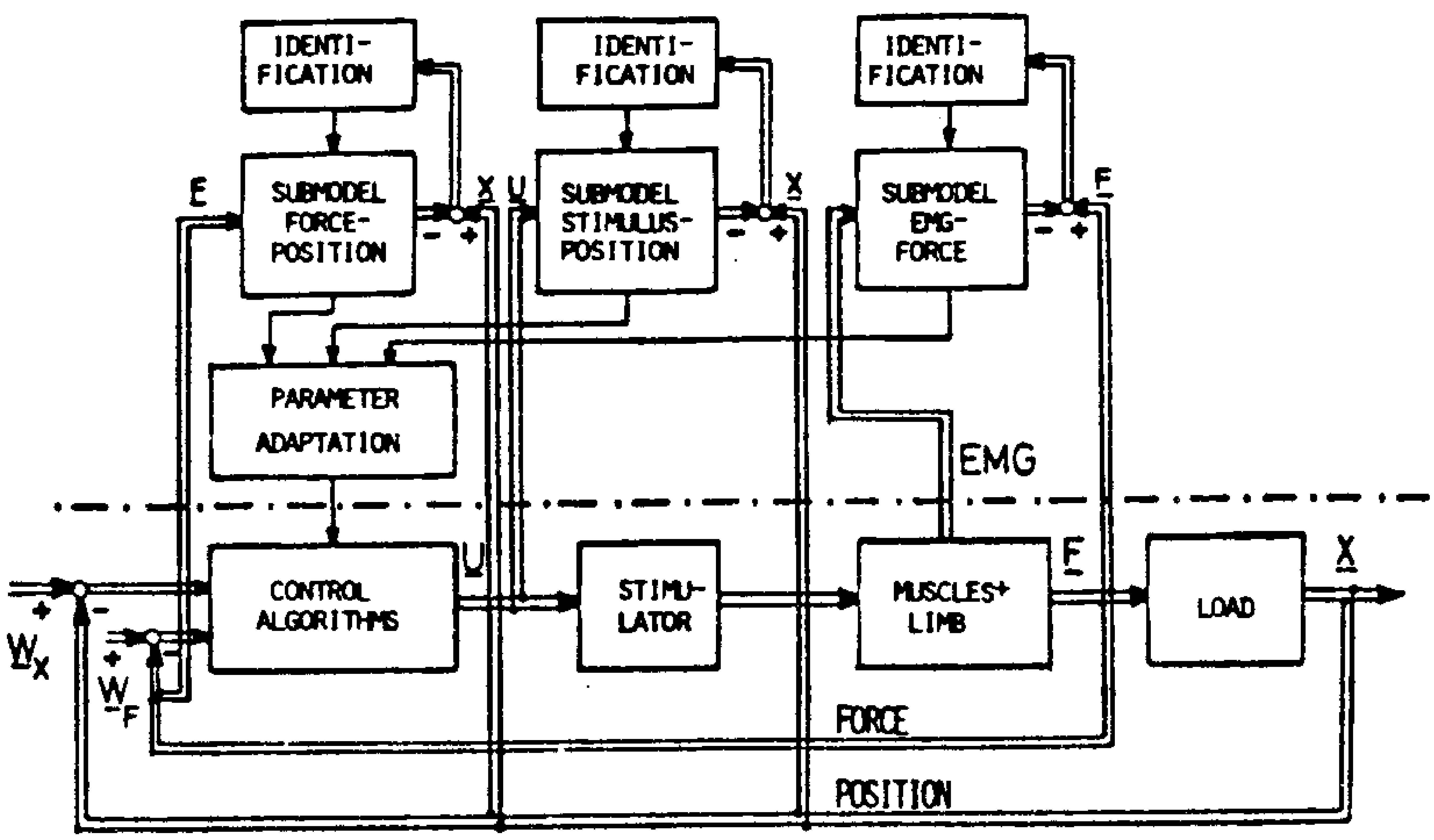

Fig. 5: The block scheme of the low and intermediate levels of the hierarchical control strategy.

The linear part of the submodel "stimulus-position" and the submodel "force-position" correspond respectively with the relationships Gp and G_F in the previous section. The first one represents the formal relation between stimulus (stimulus parameters: pulse amplitude, pulse frequency) and resulting displacement of the unloaded limb. It is nonlinear. The second one characterizes the interaction between the limb and the external load.
The submodel "EMG-muscle force" delivers a prediction of the force F_M produced by the muscles (compared to the muscle force obtained indirectly via Equ. 11 with a certain delay). The relation between F_M and other parameters of the MLL model is given by:

$$F_M = c_{s1}x_1+c_{s2}x_2 - (c_{s1}+c_{s2})x \qquad (11)$$

$$\text{or} \quad F_M = m\ddot{x}+F_S \qquad (12)$$

Thus, the effect of the dead time involved in the system may be reduced by using the faster response time of this submodel. Moreover it gives us the possibility to analyze the produced force and to estimate the recruted parts of the muscles due to certain stimulation patterns. By now this submodel is not yet investigated.

5.A Realization of the control model at the low and the intermediate level

As discussed in the 3th section Equ. 10 allows the physical separation of the MLL sytem into two submodels, one for the stimulus-position relationship,one for the force-position relationship. Thereby we obtain the possibility to interpret the force or the position as a disturbance in according to the application requirements. By this procedure two separate control pathways can be installed. Subsequently a realization of a system controlling the position of the loaded limb is presented, in which the external force due to the load should be interpreted as a disturbance. The control of the force proceeds analogously.

System indentification

For the identification one of the methods of parameter estimation may be used. Amongst these the least square method (LS) has been often applied because of its reliability in convergence, its relatively small computing time and its ease of mathematical manipulation. The nonlinearities in the system can also be determined, if they are described by polynoms. For digital computation a recursive LS algorithm should be chosen (15).

System: $$N(u) = n_1 u + n_2 u^2 + \ldots + n_q u^q \qquad (13)$$

$$G_p(z) = \frac{b_1 z^{-1} + \ldots + b_n z^{-n}}{1 + a_1 z^{-1} + \ldots + a_m z^{-m}} z^{-\tau} = \frac{B(z^{-1})}{A(z^{-1})} \qquad (14)$$

$$G_F(z) = \frac{c_1 z^{-1} + \ldots + c_p z^{-p}}{1 + a_1 z^{-1} + \ldots + a_m z^{-m}} = \frac{C(z^{-1})}{A(z^{-1})} \qquad (15)$$

where N is the polynom of the nonlinearity, z is the discrete complex variable and τ the dead time. a_i, b_i and c_i are the coefficients of the z-transfer functions corresponding to the relationships G_P and G_F.

Estimating model:

$$\hat{N}(u) = \hat{n}_1 u + \hat{n}_2 u^2 + \ldots + \hat{n}_q u^q \qquad (16)$$

$$\hat{G}_p(z) = \frac{\hat{b}_1 z^{-1} + \ldots + \hat{b}_n z^{-n}}{1 + \hat{a}_1 z^{-1} + \ldots + \hat{a}_m z^{-m}} z^{-\hat{\tau}} = \frac{\hat{B}(z^{-1})}{\hat{A}(z^{-1})} \qquad (17)$$

$$\hat{G}_F(z) = \frac{\hat{c}_1 z^{-1} + \ldots + \hat{c}_p z^{-p}}{1 + \hat{a}_1 z^{-1} + \ldots + \hat{a}_m z^{-m}} = \frac{\hat{C}(z^{-1})}{\hat{A}(z^{-1})} \quad (18)$$

The elements of the vector:

$$\underline{\hat{\theta}} = \left[\hat{a}_1 \ldots \hat{a}_m, \hat{b}_{11} \ldots \hat{b}_{n1}, \ldots, \hat{b}_{1q} \ldots \hat{b}_{nq}, \hat{c}_1 \ldots \hat{c}_p\right] \quad (19)$$

where $\hat{b}_{ij} = \hat{b}_i \hat{n}_j$

are unknown and should be estimated.

Controller

The controller consists of two separate pathways R_P and R_F (Fig. 6). Since the design and the modification of a dead beat controller (DB) is particularly simple, it will be used for controlling the displacement due to the stimulus (relationship G_P). The parameters of this controller pathway (R_P) can be directly calculated from the estimated parameters in $\hat{G}_P$, $\hat{G}_F$ and $\hat{N}$ (15).
The other pathway R_F is for compensation of the disturbance due to F_S. The transfer function of this compensating part is given by:

$$R_F = \frac{\hat{G}_F}{\hat{N} \cdot \hat{G}_P} \quad (20)$$

Fig. 6 shows the block diagram of the lowest level.
Two major problems are of interest here: the convergence of the parameter estimation and the stability of the closed-loop system. The mathematical proofs to these are for from trivial due to the nonlinearities and time variance in the system. Earlier mathematical proofs to some of these problems and extensive simulations on similar adaptive control systems using LS parameter estimation and DB controller (16) have however indicated that the estimated parameters converge as well as such systems are stable in many circumstances.

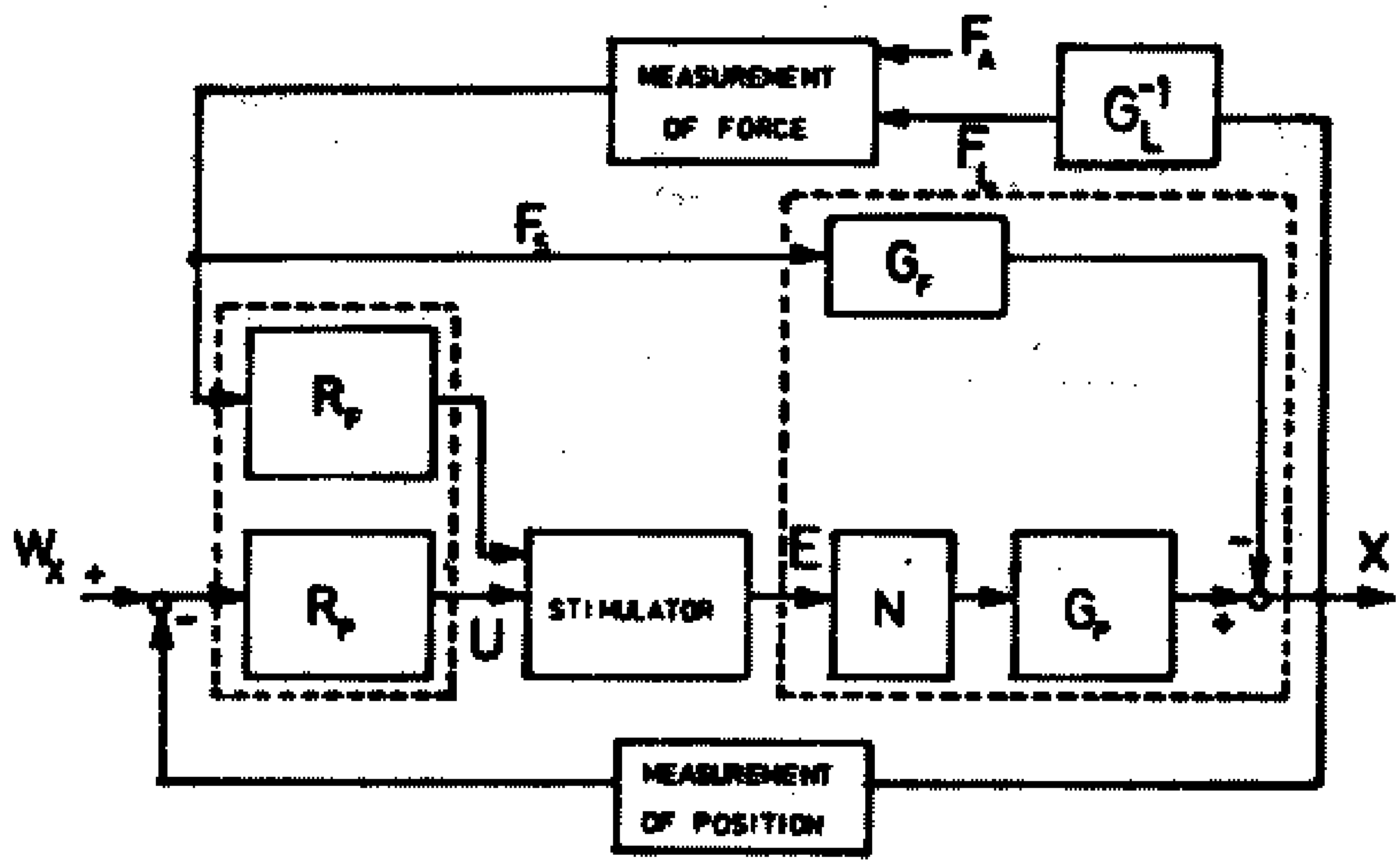

Fig. 6: The block diagram of the lowest level.

6. Effects of the modulation techniques

In our applications two modulation techniques: the PAM and the PFM techniques were used to increase the operation range of the muscles. The effects of the PAM and PFM techniques on the properties of the control systems, investigated by means of computer simulations, are interesting with respect to the improvement of the control efficiency. In order to obtain generally available results of these, the nonlinearities in the model (a 4th order model was used) were neglected in the simulations.
Closed loop control systems with proportional plus integral plus derivative controllers acting in PAM technique, and those with integral controllers acting in connection with integral pulse frequency modulators (17) and additional velocity and acceleration feedback were simulated. Fig. 7 shows two step-forced responses of the simulated systems. The simulations show that PAM systems are superior to PFM sytems with respect to the settling time and the stationary accuracy. But PAM systems are significantly more sensitive to parameter variations than PFM systems.

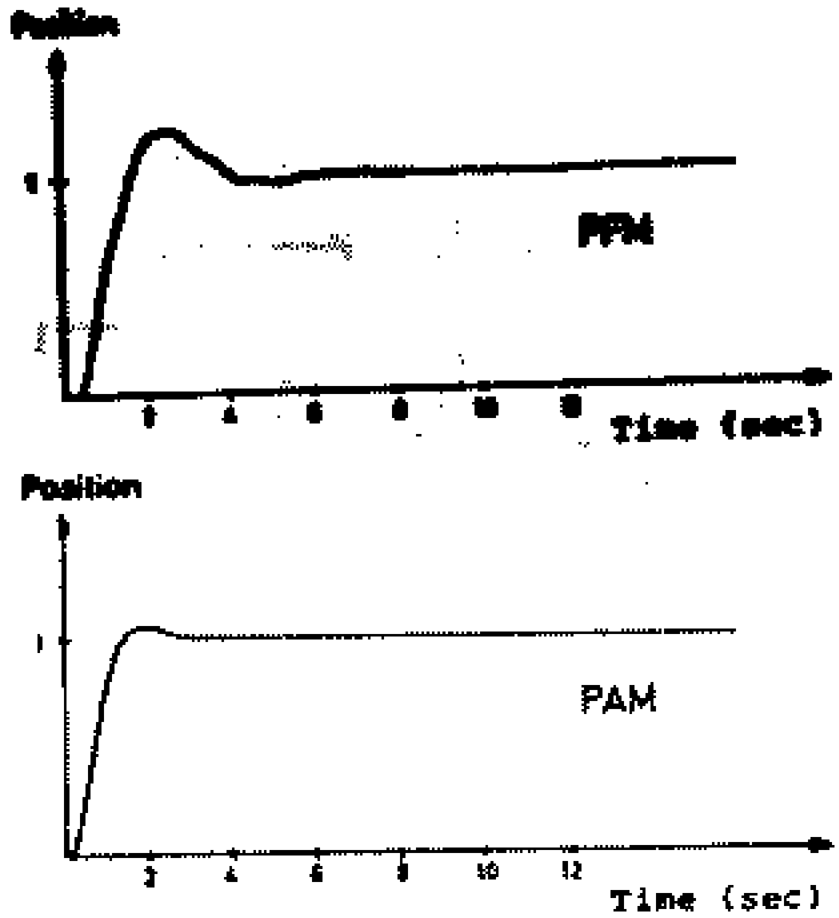

Fig. 7: Simulated step-forced responses.

7. Conclusion

Theoretical and experimental aspects resulting from the development of control systems for FES were discussed. Such control systems that involve both the position and the force feedback are of interest and consequently proposed in this study. It is further necessary to apply adaptive control strategies in order to compensate the influence of the time variance and the non-linearities inherent in the muscle and limb.

Based on a model of the muscles-limb-load system an adaptive control system was presented which takes into consideration the position and the force of the limb. This allows to install two separate controller parts (one for the position and one for the force).

A realization of the presented control strategy involves parameter estimation, a self tuning regulator and a compensation of disturbance.

In addition, the combination of two modulation techniques, PAM and PFM, may be used to increase the range of control. Our investigations show that PFM technique augments the robustness of control, whereas PAM technique improves the settling time and the accuracy of the steady state.

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