A MECHATRONIC DEVICE FOR THE REHABILITATION OF ANKLE MOTION

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Abstract: The paper presents the main results from a research aiming at the design of an electro-mechanical actuator to assist walking movements of the ankle articulation, for use in the rehabilitation of lower limb motion in patients suffering neurological disease. Motivations for the research project are discussed within the framework of the application of mechatronic concepts within rehabilitation practice. The entire design process is then described, from the definition of project target through the mechanical concept and control design steps until design validation by means of numerical simulations and tests on a prototype.

1 INTRODUCTION

In recent years, mechatronics has emerged as a powerful approach to provide innovative solutions in many technical fields related with mechanical and electronic engineering, and also in the field of bioengineering. This paper presents an application of mechatronic concepts to improve the effectiveness of therapies addressing the rehabilitation of lower limb motion in patients suffering neurological disease.

The project is being developed as a joint cooperation of the Politecnico di Milano (Technical University of Milan) and the Villa Beretta rehabilitation centre, part of Valduce Hospital, and is part of HINT@Lecco, a large research program aiming at the promotion of research fostering new applications in the fields of medical diagnostics and therapy.

Aim of the project is to design an electro-mechanical actuator to assist walking movements of the ankle articulation, referred to as "ankle actuator". The system has been designed for use either as integrated with the Lokomat rehabilitation device (Colombo et Al, 2000) presently in use at Villa Beretta, or as a standalone device to assist physiotherapy.

In the paper, the need and possible applications of the ankle actuator are described under the point of view of rehabilitation practice (Section 2), then the results of measurements and numerical simulations based on biomechanic multi-body models are reported, with the aim of setting the targets for the project (Section 3). The mechatronic design of the system and the identification of appropriate control strategies to meet the targets of the project are described in Sections 4 and 5. Finally, Section 6 reports about the final assessment of system performances, that was pursued by a combination of testing on a prototype demonstrator and multi-physics simulation of the actuator fitted on a person, involving the modelling of the lower limb and of the actuator together with the control unit.

2 USE OF THE DEVICE FOR REHABILITATION PURPOSES

The equinovarus foot is the most common pathological lower limb posture after lesions that result in an upper motor neuron syndrome (UMNS).

The lack of normal motor control and/or the presence of static foot deformity alters the cyclical kinematic pattern of lower limb and trunk during gait. There may be impairment of advancement of the body weight over the supporting limb and to

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swing the unloaded limb forward in preparation for the next step. Foot pain, skin breakdown (lateral border, fifth metatarsal) and knee hyperextension (and/or varus) are frequently associated to this gait deviation and the compensation needed for the lack of adequate base of support, limitation of ankle dorsiflexion, dysrhytmic and restrained forward translation of body mass, asymmetrical weight transfer and interference with weight bearing on the involved limb. Gait deviations and compensations in the involved limb induce compensations for the non-involved limb, pain and fatigue.

The equinovarus foot impairment is the result of different combinations of the following dysfunctions:

a) decrease of dorsiflexor muscles motor control during swing phase;
b) increase activity of plantarflexor muscles;
c) reduction of the elastic properties of the calf muscles.

The main goal of rehabilitation procedures is to maintain ankle passive range of motion , to reduce “learning non use” due to weakness of dorsiflexor muscles, to maintain the elastic properties of dorsiflexor/plantarflexor muscles.

3 TARGETS FOR THE PROJECT

A first step of the research consisted in defining quantitative targets for the mechatronic ankle actuator, in view of allowing the correct choice of the actuation system and the proper mechanical design of the system. This was done through a combination of experiments to measure relevant gait parameters and of gait mechanics modelling and simulation by a biomechanical multi-body model.

3.1 Measure of Gait Parameters

An experimental campaign was performed on healthy subjects, with the aim of obtaining the references corresponding to correct motion, to be reproduced under the assistance of the mechatronic ankle actuator. Lower limb movements were measured by means of an ELITE opto-electronic system for motion analysis, whereas contact forces under the footprint were measured by means of dynamometric platforms.

These measurements allowed to quantify the required maximum speed, maximum force and power targets for the ankle actuator. Furthermore, a rather large set of measured data was made available, allowing the validation of the bio-mechanical multi-body model in view of its use in a later stage of the project (see Section 6.1). Some of the experimental results are shown in the next paragraph, where they are compared with the results of a simulation model. Despite it is known that in case of patients with neurological diseases, muscle configurations may present a different situation compared to the healthy population, the aim of this device is to reproduce a healthy subject gait on patients with moderate neurological diseases. Applications on subjects with more serious diseases have to be verified by means experimental tests.

3.2 Numerical Simulations

A multi-body human model was defined in ADAMS/LifeMOD environment, with the aim of complementing measurements to define the targets of the project. Moreover, the same model was used in a later stage of the project, being interfaced with a model of the ankle actuator, for performance assessment purposes (see Section 6.1).

The model includes the pelvis and the two legs, and was used to perform inverse and forward dynamic simulations. In the target definition phase, numerical simulations were used to evaluate gait parameters that could not be directly measured, as the ankle torque, and to derive gait parameters under conditions that could not be tested.

The multi-body model was validated based on comparison with the measurements described in Section 3.1. As an example, Figure 1 shows the time history of the measured and simulated vertical contact force component for a healthy male person weighting of 68kg and being 1,72m tall. Figure 2 shows the experimental vs. numerical comparison of right foot marker position during gait on a treadmill, for the same subject as in Fig. 1.

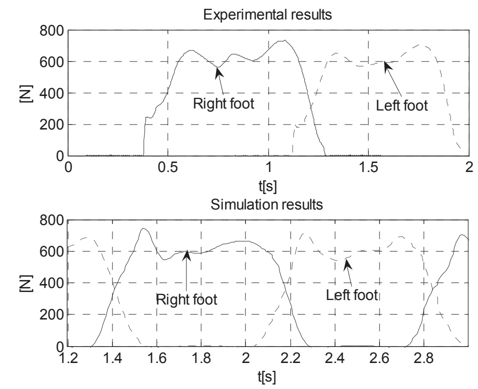


Figure 1: Contact forces for a male person. Up: experimental results; down: results of numerical simulation.

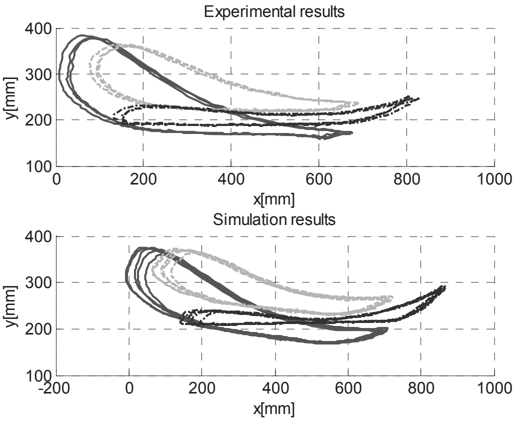


Figure 2: Right foot marker position for a male person: a) experimental results; b) results of numerical simulation.

The comparison of measured vs. simulated results allows to conclude that the mathematical model is able to capture correctly the main issues of gait mechanics and can be hence used to evaluate gait parameters under conditions that cannot be physically tested.

After validation, the model was used to derive quantitative targets for the ankle actuator, taking into consideration the effect of the Lokomat body weight support (b.w.s) system. The kinematic targets, not dependent upon the b.w.s. level, are listed in Table 1, while torque and power requirements for different levels of b.w.s. are compared in Table 2.

Table 1: Kinematic targets for the ankle actuator.

	MIN	MAX
Ankle rotation	-10°	20°
Ankle velocity	-200°/s	150°/s

Table 2: Torque and power targets corresponding to different body weight support (b.w.s) levels.

		MAX	RMS
b.w.s 0%.	Torque	112 Nm	54.4 Nm
	Power	152.6 W	48.6 W
b.w.s 25%	Torque	81.6 Nm	43.6 Nm
	Power	125.9 W	41.5 W
b.w.s 50%	Torque	65.1 Nm	31.2 Nm
	Power	90 W	29.3 W
b.w.s 75%	Torque	31.8 Nm	15.6 Nm
	Power	53.8 W	14.5 W
b.w.s 90%	Torque	24 Nm	8.37 Nm
	Power	34 W	7.36 W

Measurements described in a companion paper (Bocciolone et Al., 2008) showed that the Lokomat body weight support system provides a relief of at least 85% of patient's weight. Accordingly, torque and power targets for the project were assumed in a precautionary way to correspond to 75% body weight support.

The target motion of the ankle articulation was derived based on ELITE measurements (averaged to remove the intrinsic variability of each step) and on multi-body simulations. This was made considering a healthy subject, since the target of the system is to have the ankle performing a physiologically correct motion. As an example, the reference time history of ankle rotation for the same male subject considered above and for a walking speed of 2km/h is presented in Figure 3.

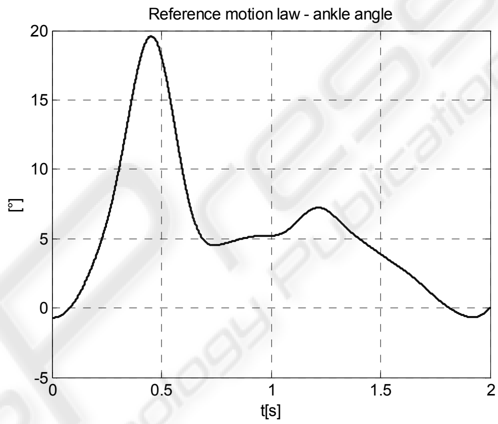


Figure 3: Reference motion law (talo-crural joint) for a male person.

4 THE CONCEPT

Besides the quantitative performance requirements resumed in Section 3, a number of qualitative requirements were defined for the ankle actuator, as detailed below:

- high intrinsic safety;
- fast and simple installation on the patient's leg (wearability);
- low weight, small size and low visual impact;
- low emission of noise and heat;

Furthermore, in case the device is used on a Lokomat machine, the mounting/dismounting of the ankle actuator over the Lokomat exoskeleton should not require any modification of the original structure.

The concept phase was then divided in two stages: selection of the type of actuator and design of the interface with the patient.

As far as the choice of the actuator is concerned, four alternatives were initially identified:

a) controlled electrical drive with rotating motor;
b) controlled electrical drive with linear motor;
c) controlled pneumatic actuator;
d) controlled Shape Memory Alloy (SMA) actuator.

These solutions were thoroughly analysed and compared evidencing for each advantages and weak points in view of the specific application. The main results of this comparison are summarized in Table 3.

Table 3: Comparison among main possible solution for the actuators of device.

Requirement	Solution			
	Rotating motor	Linear motor	Pneumatic actuator	SMA actuator
Max. Torque	Good	Very good	Very good	Poor
Dynamical response	Very good	Very good	Good	Poor
Wearability	Good	Good	Short	Fair
Cost	Good	Fair	Fair	Fair
Weight and dimensions	Very good	Very good	Poor	Good

The electrical drive with a rotating motor (DC or AC brushless) satisfies all main requirements: it has low weight and appropriate dimensions, affordable cost and simple structure which allows to build a device with a good wearability. Finally, a quite wide variety of solutions (in terms of size and specific features) is available on the market, allowing to tailor the choice in view of the requirements of the specific application. The drawback of this solution is the need of using an epicycloidal gear reducer, because rotating motors present high speed and low torque, while for ankle actuation, mechanical power is needed in the form of a relatively high torque acting at relatively low speed. The use of a gear reducer implies an increase in the weight of the device, which can be however kept within reasonable limits.

The linear motor mainly has the same advantages of the rotating motor and does not bear the drawback of using a gear reducer, because linear motors provide relatively high force at low speed, but is much more expensive than rotating motors, requires a more complicate mechanical design of the system and less alternatives are offered on the market.

Pneumatic actuation was considered also referring to previous applications to ankle actuation: (Sawicki, 2005) and (Ferris, 2005) showed that this kind of actuation is able to reproduce with good accuracy the features of actual ankle motion. In view of the aims of this project however, this solution compared to electromechanical solution provides lower actuation speed, lower maximum control force and requires auxiliary systems of relatively large size and potentially disturbing, like the air compressor.

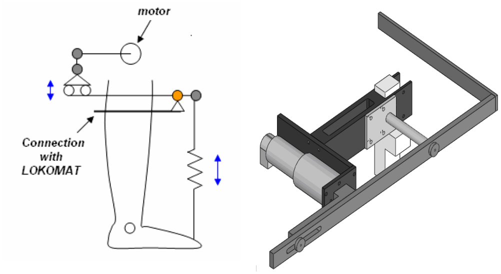
The SMA actuator solution was also considered based on some tentative applications in the bio-medical field, including a parallel project within the HINT research project. However, besides some practical disadvantages like the need for a rather complicated cooling system, present SMA actuation technology appears to suffer from an insufficient dynamical response which is not able to ensure the required actuation speed to implement ankle motion control.

Based on the analysis summarized above, the solution based on rotating electric motor was identified as the most appropriate in terms of technical requirement and related costs and was chosen for development. In particular, a 150 W DC motor (MAXON RE40 150W), driven by a 4-Q-DC servoamplifier (MAXON ADS 50/10) and coupled with an epicycloidal gear reducer with a gear ratio of 1:113 (MAXON GP 52C) was chosen.

The second stage of the concept development consisted in the design of the interface between the actuator and the leg. The basic idea underlying this activity was to implement a configuration similar to the one used in the Lokomat machine. On that device, the feet are restrained by a sling held by an arm rigidly connected to the exoskeleton (Figure 4). The concept for interfacing active ankle actuation with the patient’s leg was then based on the idea of replacing the passive arm with an articulated mechanical system, properly connected with the motor, so that ankle actuation may be obtained by an appropriate movement of the final link in the articulated system. The functional scheme of the solution identified is reported in Figure 5a: a support is fixed over the Lokomat exoskeleton, and connected by a hinge with a link which is actuated by a crank fitted on the gear output shaft. The rotation of the link produces a vertical motion of the sling holding patient’s foot, which can be suitably controlled by the electric motor. Figure 5b shows a 3D rendering of the ankle actuator, including motor+gear, crank, link and connection to the Lokomat exoskeleton.

As shown in Figure 6, the mechanism has a gear ratio between the rotation of the gear output shaft and the rotation of the link which is almost constant and equal to 1:2.5. Figure 7 shows a picture of the developed device.

Figure 4: Passive sling of LOKOMAT.



(a) (b)

Figure 5: a) Concept of active ankle actuation; b) Mechanical design of the ankle actuator.

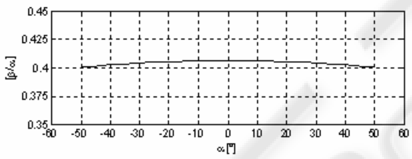


Figure 6: Gear ratio of kinematic mechanism versus angle of crank α.

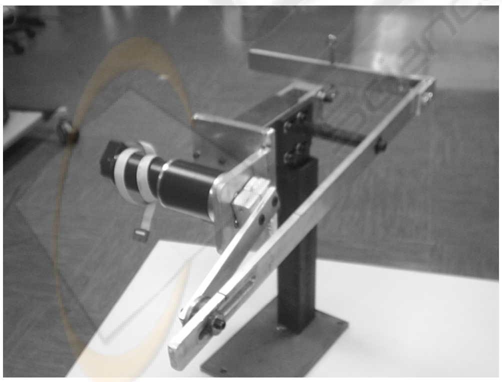


Figure 7: The developed device.

5 CONTROL DESIGN

A relevant part of the research consisted in the design of the control system for the electro-mechanical ankle actuator. A typical control architecture for electrical motors is composed of an internal feedback loop (called current loop) and an external feedback loop which in this case was defined to control the angular position of the motor (position loop), as shown in Figure 8. This configuration enables to increase the performances of the control system and to control at any time the electrical motor current in order to avoid working condition potentially dangerous for the electrical actuator.

The current feedback loop is implemented by the servoamplifier, which has a very high dynamic response. Control parameters for the current loop are optimised by the manufacturer, so it was decided to use it without any modifications. On the other hand, the regulator for the position loop was implemented using a programmable control board (DSP board). The inputs for the DSP board are the reference position signal and rotor position measured by an encoder, whereas the output of the position regulator is the reference current signal, which is fed into the current regulator.

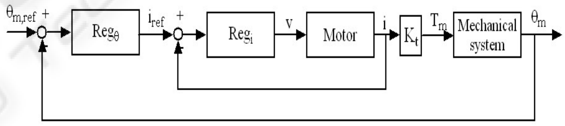


Figure 8: Block diagram of control system.

In order to define the reference signal for the position regulator, first of all a typical wave shape for ankle angular motion was derived taking into account the patient's anthropometry, by means of numerical simulations supported by measurements, as described in section 3. Figure 3 shows an example of reference motion law for ankle rotation.

Then, under the simplifying assumption of having a constant gear ratio between the output shaft of the gear and the ankle, the reference for ankle rotation was converted into a reference for the angular position of the motor.

In this first stage of study, a proportional-derivative (PD) regulator has been used for the design of the position loop. The gains of regulator were chosen to provide a good performance of control system and to assure a good dynamic response. For the optimal design of the regulator, a simple linear model of the ankle actuator was defined. The model equations are:

$$\begin{aligned} v &= R \cdot i + L\frac{di}{dt} + e \\ e &= K_e \cdot \dot{\theta}_m \\ T_m &= K_t \cdot i \end{aligned} \tag{1}$$

$$J^* \cdot \ddot{\theta}_m = \eta_{tot} T_m - \frac{T_c(t)}{n} \tag{2}$$

where v and i are respectively the voltage and the current of electrical motor characterized by a resistance R and an inductance L, e is the electromotive force, K_e and K_t are respectively the speed and torque constants, T_m is the torque of motor, J^* is the equivalent moment of inertia of all parts of device, θ_m is the angular rotation of motor, η_{tot} is the coefficient of efficiency of the reducer, T_c the resistant torque and n the total gear ratio.

For the control design, the resistant torque T_c is considered as a known noise input signal and for this reason it's not considered during the design of regulator.

Finally, numerical tests were performed to verify the correct tuning of the gain parameter for the position regulator, using the more detailed non-linear model described in section 6.1.

6 SIMULATIONS AND TESTS FOR FINAL ASSESSMENT

6.1 Multi-physics Simulations

A multi-physics model of the ankle actuator was established, to verify the correct design of the system (including the regulator) and to assess its overall performances. The model is composed of:

- an electro-mechanical component, representing the electrical motor and the gear (see Fig. 9a);
- an electronic component, reproducing the control loops implemented in the motor drive (Fig. 9a);
- a multi-body biomechanical component, representing the lower limb and the mechanical linkages connecting the motor with the foot (Fig. 9b).

Contrarily to the simplified model used to perform the control design stage of the project, the multi-physics model used for verification and assessment is fully non-linear and accounts for the interaction effects between the different components of the system.

In particular, the linear equation (2), representing in a simplified way the mechanical component of the system, is replaced by a non-linear equation for the gear and by a fully non-linear model of the leg and of the mechanical linkages.

As far as the gear is concerned, a non-linear formulation is used, accounting for the different expression of power dissipation in the gear depending on the power flow being directed from the motor to the output shaft or vice-versa (Hannah, Hiller, 1995), so that having defined the mechanical power flowing in the gear from the motor to the output shaft W_g according to the following expression:

$$W_g = \left(T_m - J_m \ddot{\theta}_m\right)\dot{\theta}_m \tag{3}$$

the mechanical equation of the motor takes the following non-linear expression:

$$\begin{aligned} \eta_d T_m + \frac{T_c}{n} &= \left(\eta_d J_m + \frac{J_c}{n^2}\right)\ddot{\theta}_m \quad \text{for } W_g > 0 \\ T_m + \frac{\eta_r T_c}{n} &= \left(J_m + \frac{\eta_r J_c}{n^2}\right)\ddot{\theta}_m \quad \text{for } W_g < 0 \end{aligned} \tag{4}$$

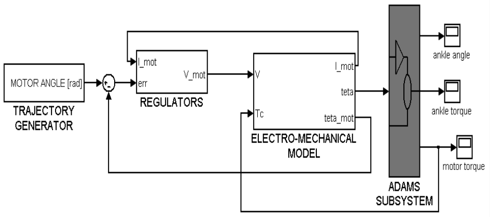
where J_m is the moment of inertia of the rotor and gear input shaft, J_c is the moment of inertia of the gear output shaft, η_d and η_r are the efficiency coefficients of the gear (for other symbols, see Section 5).

The leg and the linkages in the ankle actuator are represented by a multi-body model with an excess of 30 states defined in ADAMS/LifeMOD environment. The model takes as input the rotation of the gear output shaft θ_c defined as:

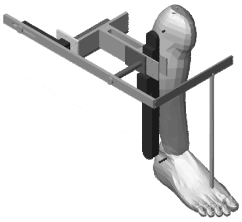
$$\theta_c = n\theta_m \tag{5}$$

and by solving the direct dynamic problem for the linkages and leg returns the motion of the modelled bodies and the value of the resistant torque T_c, which is fed back into equation (4). The use of the non-linear multi-body model allows to represent in full detail the actual non-constant gear ratio of the linkage (see Figure 6), and to take into account the effect of actual loads (e.g. contact forces exchanged in the footprint) in the calculation of the resisting torque T_c, avoiding the simplifying assumptions that were introduced for control design purposes.

The model was implemented in Matlab/Simulink environment, using co-simulation with ADAMS to account for the coupling between the electro-mechanical and electronic components of the model (implemented in Simulink) and the bio-mechanical model component (implemented in ADAMS). Figure 9a shows a high-level representation of the complete Simulink model, where co-simulation with the ADAMS subsystem is represented.



(a)



(b)

Figure 9: a) Co-simulation model; b) detail of ADAMS subsystem.

Figure 10 shows the results of a simulation reproducing the behaviour of the ankle actuator under the reference motion law shown in Figure 3 and considering 75% of body weight support. In the upper subfigure, the deviation of motor rotation with respect to the reference value is plotted: it is observed that the maximum deviation is in the range of 0.3%, indicating that the design of the regulators was performed correctly, so that the actuator is able to track position reference within accuracy levels that are fully satisfactory for the considered application.

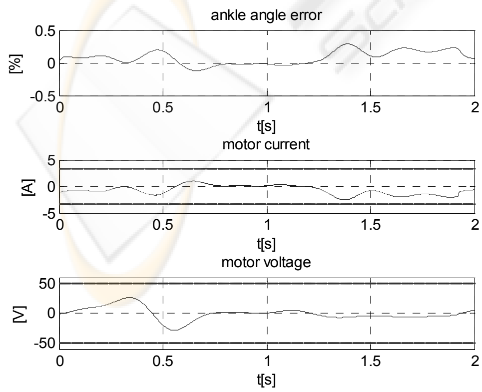


Figure 10: Co-simulation results.

In the central and lower subfigures of Figure 10, motor voltage and current are plotted, together with their respective limit values that cannot be exceeded to prevent damage and ensure durability of the motor. It is observed that for both voltage and current the maximum values achieved during the simulation are well below the limit values, demonstrating the correct dimensioning of the motor and actuator for the considered application.

In order to quantify the further ability of the device to withstand higher working loads (produced e.g. by unforeseen additional disturbances or by uncertainties in some design parameters), safety factors can be defined for voltage and current as the ratio between the limit value and the actual maximum absolute value of each quantity. For motor voltage, the safety factor is 1.70 whereas for motor current the safety factor is 1.32. These values are high enough to ensure the robustness of the system, but not so high to suggest the over-dimensioning of the motor.

6.2 Tests on a Demonstrator

Preliminary tests were performed on the physical demonstrator shown in Figure 7, to assess experimentally the performances of the device. At the present stage of the research, the demonstrator is not yet connected to the patient's leg, so passive masses are used to produce a resisting torque and inertial effect reproducing in first approximation the connection with the ankle.

In this section, three working conditions are considered:

a) An unloaded condition, which means that no external or inertial load is applied on the ankle actuator;
b) A loaded condition obtained by connecting a 5kg mass to the extremity of the sling, thus introducing a resisting force of about 50N and an additional inertial effect;
c) A loaded condition realised through a 10kg mass, producing a resisting force of about 100N and an inertial effect higher than in case b).

Tests were performed on the ankle actuator by using the reference ankle rotation represented in Figure 3 and measuring the actual motor position and the motor current.

Figure 11 shows the results obtained for working condition c) in the list above, corresponding to the most loaded condition. It is observed that the positioning error is always below 2%, which is a quite low value but still higher that the one obtained from the numerical simulation. This result could however be affected by some electrical noise disturbance in the measure of the motor angular position, so that the actual positioning error could be in the same range of values as the one obtained by numerical simulation. More accurate tests are planned in order to clarify this point.

The motor current is quite low, with maximum value below 1A, to be compared with a limit value of 3.33A. This means that there is a large margin for further loading of the actuator.

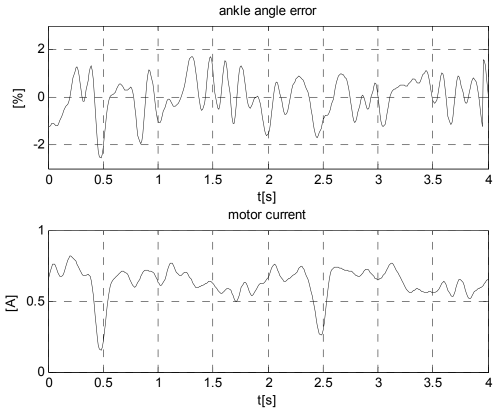


Figure 11: Experimental result in 100N load condition. Up: position error; down: motor current.

Table 4 compares the results of tests performed in working conditions a) to c), in terms of maximum angular position error of the motor and of maximum and r.m.s. value of the motor current required by the actuator. The maximum position error in the unloaded condition is much higher than in the two loading conditions. This is produced by the effect of plays that are present in the mechanical parts connecting the final link with the gear output shaft. The effect of plays is otherwise eliminated by the presence of the gravitational load applied by the masses in working conditions b) and c).

Table 4: Values of maximum ankle angle error, maximum motor current and RMS motor current in different loading condtions.

	$Error_{max}$	I_{max}	I_{RMS}
No load	5.12%	0.42A	0.1A
50N load	1.95%	0.71A	0.59A
100N load	2.55%	0.82A	0.64A

Considering a different working condition where the ankle actuator is connected with the patient's leg, the effect of plays can seriously affect the performances of the system, and therefore a new stage of the research was started to find an alternative design for the connection between the gear output shaft and the final link moving the patient's foot. A solution based on the use of toothbelt was developed and is presently under realisation. This new solution is expected to improve substantially the performances of the device in actual working conditions.

7 CONCLUSIONS

The paper has reported some results from an ongoing research having as objective the development of an electro-mechanical ankle actuator for use in the rehabilitation of lower limb motion in patients suffering neurological disease.

Preliminary results obtained by numerical simulation and tests show the ability of the developed system to actuate ankle motion within accuracy levels that are appropriate for rehabilitation purposes. Research is on going on introducing this device in physiotherapy. Preliminary test on healthy and diseased subjects are foreseen in a short time.

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