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DEVELOPMENT OF A CONTROLLERFOR A 3-DOF ROBOTIC PLATFORM FOR USER INTERACTION IN REHABILITATION THERAPIES

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Abstract

This work deals with the development of a controller for a robotic platform conceived as a rehabilitation device for the human upper limb. The mechanical system is a three-degree-of-freedom parallel mechanism which is coupled to the actuators, three DC-motors and the driversthat provides the desired control signals (position, velocity and current). Regarding the control system, two techniques are utilized: the computed torque control for the motion control and the impedance control for force control. Simulations are performed in MatLabsoftware in order to evaluate the interaction with a user.

Keywords: *Robotic platform. Parallel mechanism. Control techniques. Upper limb rehabilitation.*

1. Introduction

It is estimated that, in a year, 15 million people might suffer a cerebral vascular accident (CVA). Among those, 6 million die and 5 million suffer permanent injuries. The CVA, or stroke, is the second main cause of disabilities, after demency. The injuries can be either vision or speech loss, and paralysis [1]. Among the people that survive a CVA, 10% might recover completely, 25% might suffer light injuries, 40% severe injuries, 10% demand special care and 15% die right after the stroke [2].

In case of motor disabilities, the movement recovery is possible through the physiotherapy. This practice requires that the professional therapeut be always present during all the sessions, in order to guarantee that the pacient performs the correct recovery movements. The utilization of robotic mechanisms during rehabilitation sessions is a growing practice, due to advantages, namely, excelent repetition, more intensity and duration in a comparison with the conventional methods [3]. Basically, the research on rehabilitation robotics relies on the development of end-effector-based platforms (MIT-manus, MIME and GENTLE/s) and exoskeletons. The goal of this work is the development of the control system of a robotic parallel mechanism, in such a way that it provides the desired motions for rehabilitation therapies.

2. Methodology

The methodology we follow here consists of a sequence of activities: (a) description of the mechanical system (b) characterization of the actuators and control hardware; (c) control simulations to predict the user interaction with the real mechanism.

Initially, the platform requirements for rehabilitation applications are identified. Then, the computational models for the simulation of the control system, and perform selected tests to predict, through simulations, the real behavior.

3.The platform requirements

The robotic platform was originally conceived not only for rehabilitation applications but also for pick-and-place operations. Hence, the actuation system specifications should satisfy the requirements for the execution of both tasks.

From the results obtained in previously performed simulations [4], we concluded that the maximum output torque is 3N.m and maximum angular velocity is 16.8 rad/s. By assuming the efficiencies for the motor and the speed reducer are 0.8 and 0.7, respectively, the estimated electrical power is 89.6 W. Then, we selected an actuator system to work in a range from 100 to 150 W.

Additionally, we have decided that the motors' nominal voltage must remain in a range from 24 and 48 V, because motors with bigger nominal voltage usually have bigger torque constant. This choice brings some benefits, namely, the motor will not demand high current values, the use of transmission cables with small diameters and the adequacy of low current power sources.

Then, the following requirements are defined:

- Motor: DC, power between 100 and 150W, nominal voltage between 24 and 48V, maximum weight 15 N, coupled to an encoder and a speed reducer.
- Speed reducer: planetary, maximum weight 10 N. It must support an output torque of 3N.m and maximum angular speed of 160 rpm.
- Encoder: Optical, three channels, with a minimum 1000 pulses/turn.
- Drive: servo-amplifier with the possibility of position, velocity and current controls.

The accurate definition of such parameters is highly important since the control loop is highly dependent of such parameters, not only for the dynamics model computation, but also for the signal processing and its discretization, avoiding either response degradation and system instability.

4. Prototype

Once the mechanical system is already built, we proceeded to the developmentof the dynamic modelof the mechanism. The mechanism analyzed in our research is a three-degree-of-freedom parallel mechanism, and it is shown in Figure 1. Figure 2 represents the kinematic diagram of the mechanism.

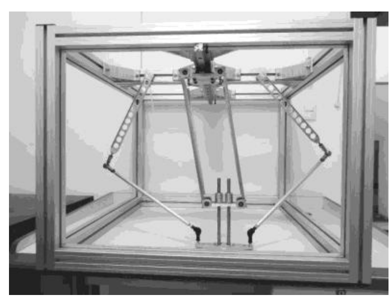

Figure 1- The mechanical system.

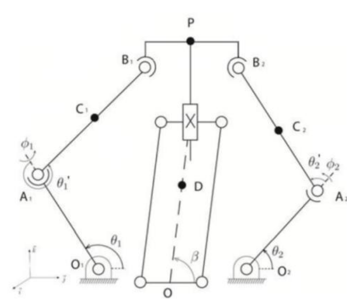

Figure 2– Kinematic diagram of the mechanism

Given the project requirements, the following equipment are chosen: Maxon 148867 motor and Maxon EPOS2 controller.
Maxon 148867 motor has the following specifications,

Nominal voltage	24 V
Rotation (no load)	793.8 rad/s
Nominal rotation	725.7rad/s
Current (no load)	137 mA
Nominal current (maximum current in steady-state)	5.77 A
Nominal torque	170 mNm
Maximum torque with locking shaft	2280 mNm
Maximum efficiency	91%

Table 1 – DC motor specifications

The selected planetary gear has a 43:1 gear ratio and 71% of efficiency, ensuring the required torque. The controller, in addition to meeting the prerequisites mentioned above, also has the characteristics described in Table2. The prototype's architecture is described in Figure 3.

Comutation frequency	50 kHz
Maximum efficiency	94%
Sample rate – current control (PI)	10 kHz
Sample rate – speed control (PI)	1 kHz
Sample rate – position control(PID)	1 kHz
Communication protocols	USB CAN RS232

Table 2 – EPOS2 specifications

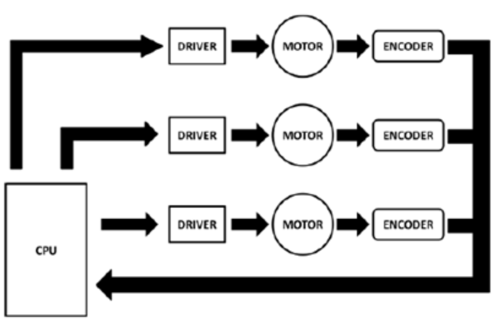

Figure 3- Basic architeture (adapted from [4])

5. The Control system

The control system must ensure the following requirements for therapy to be successful:

- Patient safety (stability of the system);
- Faithfulness to the designated trajectory;
- The patient must exert some force for the movement to occur.

This force is a variable that depends on the level of intensity required for the therapy. To ensure that all therapy requirements are met, we chose an impedance control architecture based on computed torque control [7].

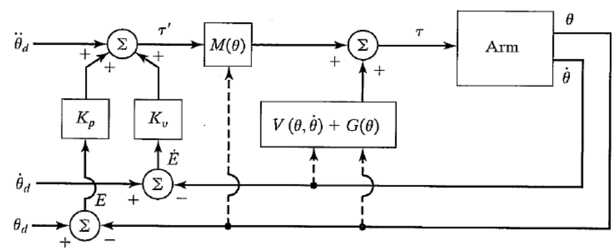

Figure 4– Computed torque control loop (adapted from [7]).

The impedance control structure is focused on the interaction of the mechanism with the environment. The goal is to change the characteristics of the structure (the damping of the system, for instance) to obtain a desired dynamic response for rehabilitation [6]. Although more precise information of the mechanism and a more structured control loop is needed, the possibility of controlling the mechanism as a whole considering the interaction with the environment suggests that it is a good approach.
The impedance control will be implemented using the same control structure shown in the computed torque control by choosing suitable values for the inputs (Cartesian coordinates of de platform and its first and second order derivatives) X_d, $\dot{X}_d$, $\ddot{X}_d$ and the gain matrices $K_p$and K_v. The control loop is shown in Figure 5.

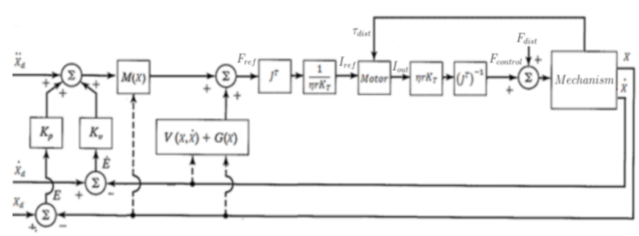

Figure 5– The control loop (adapted from [7]).

The dynamic equation of the mechanism is given by

$$\bar{M}\ddot{X} + \bar{V} + \bar{G} = F$$

where $\bar{M}$ is the mass matrix, $\bar{V}$ is the vector of centrifugal and coriolis terms, $\bar{G}$ is the vector of gravitational terms and F is defined by the following equation

$$F = F_{control} + F_{dist}$$

The $F_{control}$ force is the force applied to the platform dueto the action of actuators and F_{dist} is the disturbance force applied to the platform, in this case, the force applied by the patient imposes a trajectory to the platform. The control law is given by

$$F_{ref} = \bar{M}\left(\ddot{X}_d + K_v(\dot{X}_d - \dot{X}) + K_p(X_d - X)\right) + \bar{V} + \bar{G}$$

Assuming ideal current controller, $F_{control} = F_{ref}$, the equation of the closed loop system is given by

$$\ddot{X} - \ddot{X}_d + K_v(\dot{X} - \dot{X}_d) + K_p(X - X_d) = \bar{M}^{-1}F_{dist}$$

Thus, if K_p and K_v are diagonal with all positive elements and $F_{dist} = 0$, we can see our system in closed loop as 3 uncoupled mass-spring-damper system, with unit mass. Varying the values of K_p, K_v and X_d, we can make some preferred directions.To illustrate, suppose we wish to perform trajectories in the plane $z = C, C \in \mathbb{R}$, parallel to the x or y-axis, with a sinusoidal disturbance (due to motor problems of patients) also in the plane $z = C$, orthogonal to the trajectory.We can assume that the movement in the z axis only aims to make the patient´s be more comfortable while performing the trajectory. For this purpose, the inputs $\dot{X}_d$ and $\ddot{X}_d$ will not help, so they are set to zero.
Thus,by choosing the following values of K_p and K_v

$$K_p = \begin{bmatrix} K_{px} & 0 & 0 \\ 0 & K_{py} & 0 \\ 0 & 0 & 0 \end{bmatrix} K_v = \begin{bmatrix} K_{vx} & 0 & 0 \\ 0 & K_{vy} & 0 \\ 0 & 0 & K_{vz} \end{bmatrix}$$

Additionally, the mass matrix of the mechanism ($\bar{M}$) has the following format:

$$\bar{M} = \begin{bmatrix} M_{xx} & 0 & 0 \\ 0 & M_{yy} & M_{zy} \\ 0 & M_{yz} & M_{zz} \end{bmatrix}$$

with $K_{px}, K_{py}, K_{vx}, K_{vy}, M_{xx}, M_{yy}, M_{zz} \geq 0$.
By neglecting the acceleration and velocity in the z direction, the closed-loop equations of the system are in the following format

$$\begin{cases} M_{xx}\left(\ddot{x} + K_{vx}\dot{x} + K_{px}(x - x_d)\right) = F_x \\ M_{yy}\left(\ddot{y} + K_{vy}\dot{y} + K_{py}(y - y_d)\right) = F_y \\ \dfrac{M_{yz}}{M_{yy}}F_y = F_z \end{cases}$$

Assuming a parallel to the x axis rectilinear motion, we define the following initial and final points of the trajectory

$$X_i(x_i, y_i, z_i) X_f(x_f, y_i, z_i)$$

Choosing the following values to the system inputs,

$$X_d = X_i \quad K_{px} = 0$$

then, the system is as follows

$$\begin{cases} M_{xx}(\ddot{x} + K_{vx}\dot{x}) = F_x \\ M_{yy}\left(\ddot{y} + K_{vy}\dot{y} + K_{py}(y - y_i)\right) = F_y \\ \dfrac{M_{yz}}{M_{yy}}F_y = F_z \end{cases}$$

Thus, the system behaves as a mass-damper system in the x-direction and a mass-spring-damper in the y-direction, with the spring unstressed when $y = y_i$.

This result is only valid assuming ideal current controller. However, we will show through some simulations that the system with the current controller projected behaves very close to the system with ideal current controller, so all the equations shown before are still valid.

6. Results
We will simulate the rectilinear motion of the mechanism on the x-axis from the point $X_i = (0; 0; 0{,}552)$ to the point $X_f = (0{,}300; 0; 0{,}552)$ using a fifth-degree-polynomial interpolation, so that the speeds and accelerations at the beginning and end of the path are null, added to a sinusoidal disturbance in y with $\omega = 3\frac{rad}{s}$ and $A = 0{,}020m$. The following inputs and gain matrices are used in the control law:

$$K_p = \begin{bmatrix} 0 & 0 & 0 \\ 0 & 10 & 0 \\ 0 & 0 & 0 \end{bmatrix}$$

$$K_v = \begin{bmatrix} 10 & 0 & 0 \\ 0 & 10 & 0 \\ 0 & 0 & 10 \end{bmatrix}$$

$$X_d = (0; 0; 0.552)$$

To show that the analysis is still valid using the assumption of ideal current controller, we perform the following procedure:

- The required force (F) and corresponding torques applied to the actuators(τ_{dist}) for the mechanism to accomplish the desired trajectory are calculated (inverse dynamics);
- The force reference (F_{ref}) given by the control law when the mechanism performs this trajectory is calculated;
- From the reference force, the reference current (I_{ref}) that will be used in the motors' loop is calculated;
- The reference current and the torques caused by the movement of the mechanism (torque disturbance) are used to calculate the output current of the motors (I_{out}) through the

difference equations of the current controller;

- Torques applied by the motor and the equivalent force applied to the platform ($F_{control}$) are calculated;
- The force performed by the user to make such a move (F_{dist}) is calculated by the difference between F and $F_{control}$;
- The direct dynamics simulation is performed, assuming ideal current controller, using F_{dist} ;
- The trajectory imposed on the mechanism is compared with that obtained in the direct dynamics simulation.

The simulationswere done using the software MATLAB. The following results were obtained (Figures 6, 7 and 8):

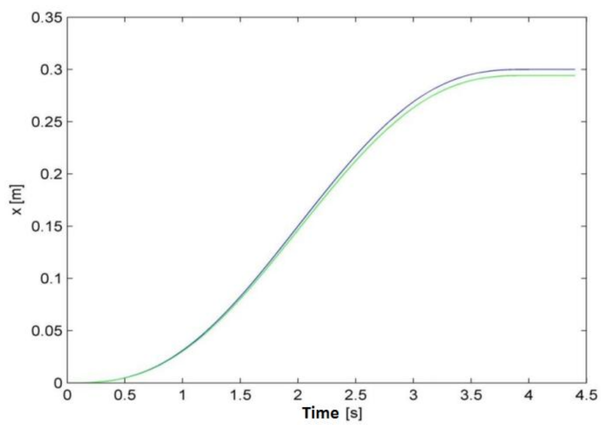

Figure 6- Performed trajectory in the x-axis

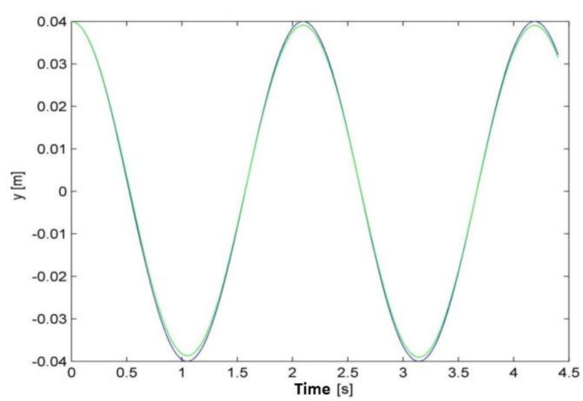

Figure 7 -Performed trajectotry in the y-axis

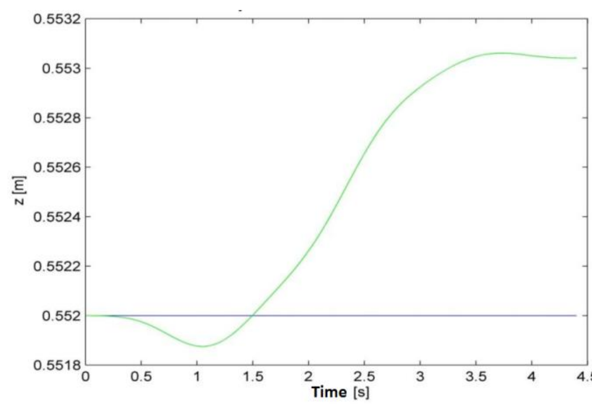

Figure 8 - Performed trajectory in the z-axis

The blue line is the desired trajectory and the green line is the trajectory obtained in the direct dynamics simulation

These results show us that the analysis is still valid using the assumption of ideal current controller. This fact makes it possible to use a simple and explicit expression for the closed loop system and will be very useful for the next analysis.

Another important analysis is the quasi-static displacement. Based on it, it is possible to show that the force applied by the user occurs at low speeds trajectories. Choosing $K_{px} = 0$ and $K_{py} = 10$, we can expect that the system behaves approximately as a mass-spring system in y-direction with no restriction in the x-direction. Thus, setting $z = 0.552m$, we obtain the force maps in Fig. 9-11.

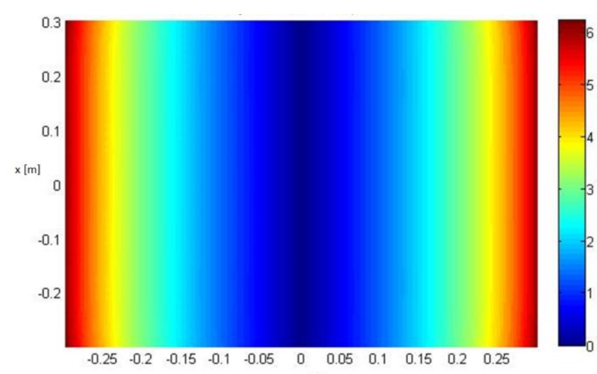

Figure 9–Force [N] applied by the user when $y_d = 0\ m$

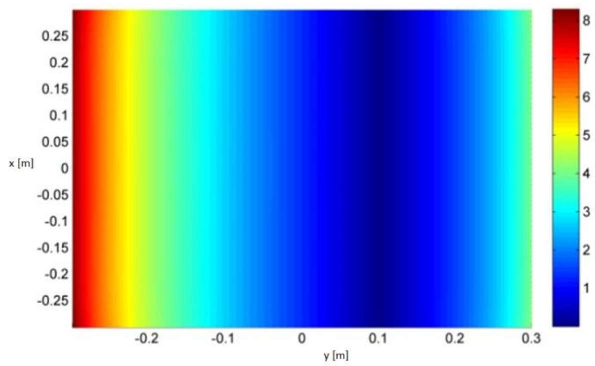

Figure 10 - Force [N] applied by the user when $y_d = 0{,}100m$

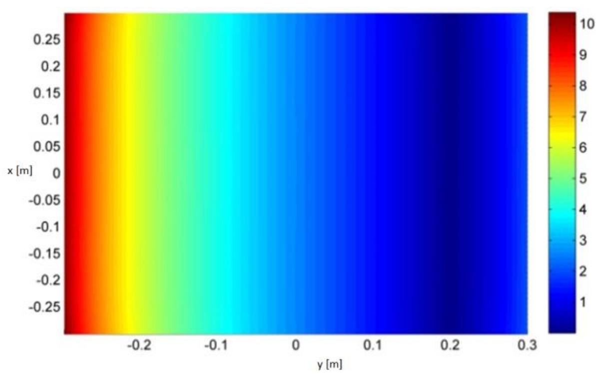

Figure 5 - Force [N] applied by the user when $y_d = 0{,}200m$

It can be concluded that the x-component of the force in this case (quasi-static displacement and $K_{px} = 0$) is null.

7. Discussion

The simulations presented show that the predicted resuts were reached. Either the input changes in each simulation or even the controller parameters allow us to notice different situations, which can occur in experimental tests of the mechanism.

The simulation of the impedance control shows that the errors associated to the end-effector position occur. However, these errors can be tolerated in cases where they do not sacrifice the pacient´ssafety, the presence of such errors can be an advantage because they constitute itself in a compliance effect to accomodate the patient´s movements. Moreover, the simulations also show that the errors associated to the force applied by the patient do not occur.

8. Conclusions

The simulations have shown that the desired results were achieved, providing the theoretical basis for experimental tests in order to mitigate eventual problems to be faced. However, simulations cannot guarantee that the presented strategy will be successful in all the therapy requirements. Uncertainties on the model parameters might be critical for the control system and the unknowledge of certain factors, like friction and approximations on the mechanical parameters, can also be very critical, impossibilitating the therapy. Threfore, it is crutial that those tests must be executed on the real mechanism in order to verify the information presented in this work.

9. Future works

We suggest in future works the parameters optimization, as well as alternative control techniques can be explored in this mechanism, because it can be employed in many applications, not only in the rehabilitation of the upper limb but also in pick-and-place operations.

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