HYBRID FORCE/IMPEDANCE CONTROL FOR THE ROBOTIZED REHABILITATION OF THE UPPER LIMBS

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Abstract: This paper presents the architecture and the control hierarchy of a 3 degrees-of-freedom robot destined for the rehabilitation of the upper limbs. A hybrid control law, using a weighted sum of force and impedance, is proposed to implement the *Active-Assisted* rehabilitation mode. Parameters' tuning of the resulting active force feedback controller is based on a compromise between user's safety and dynamic performances. Some results illustrate the performances of the developed controller.

Keywords: Biomedical systems, force/position duality, hybrid control, man/machine interaction.

1. INTRODUCTION

Training and rehabilitation systems are automated devices that are used to: i) perform various types of movements at constant speed (isokinetic), fixed position (isometric), constant load (isotonic); ii) and measure the muscular forces of the target joints (shoulder, knee, wrist…). The use of these systems is becoming popular in the clinical and sport centres. However, the realizable movements are limited to circular motions, because the corresponding machines generally comprise a single rotation axis. Moreover, the fundamental movements of the shoulder are not rigorously circular (Mayer *et al.*, 2001), and they require the 3 degrees of freedom of space. For example, the movements of abduction/adduction in the frontal plane and flexion/extension in the sagittal plane induce a displacement of the shoulder's rotation axis of approximately 8 cm for an articular amplitude of 100° (Walmsey, 2001). Moreover, the physiological movements are not strictly restricted within the plane which defines them. Therefore, during the execution of these movements, the circular trajectories of classical systems force the user to resort to non desired muscular compensations at the elbow, shoulder and trunk. These muscular compensations could distort the muscular evaluation, entailing traumatic rehabilitation and inefficient training.

In order to solve these problems and to increase the range of the realizable motions, we developed a prototype of a robotized arm with 3 degrees of freedom, allowing the execution of physiological movements (fig. 1). The development of this robotized arm benefits from our previous experience related to the specification and the design of rehabilitation and training machines for the lower limbs (Moughamir, *et al.*, 2002; Manamanni, *et al.*, 2005).

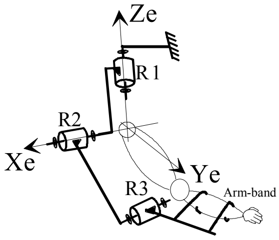

Fig. 1. Robotized arm with 3 degrees of freedom

The use of 3 articulations makes it possible to compensate the misalignment between the rotation axis of the user's shoulder and the rotation axis of the robotized arm. Thus, it becomes possible to carry out the movements of abduction/adduction in the frontal plane and of flexion/extension in the sagittal and scapular planes, while limiting the undesirable muscular compensations. This confers many possibilities to the system in terms of accuracy and efficiency of evaluation and rehabilitation.

The majority of the prototype robots developed for the rehabilitation of the upper limbs (Cozens, 1999; Burgar, et al., 2000 ; Krebs, et al., 2000 ; Reinkensmeyer, et al., 2000; Lum, et al., 2002) use traditional rehabilitation techniques, based on passive or active mobilization. In passive mobilization, the subject is not the actor of the movement and his/her arm is lifted by an external force, while the muscles remain completely relaxed. In active mobilization, the subject carries out a voluntary movement through muscular contraction. The robot must provide the necessary assistance if the subject is not capable of correctly performing the movement. Therefore, the robot should be able to quantify and adjust the degree of assistance necessary to succeed the various stages of rehabilitation. This mode of rehabilitation, known as *Active-Assisted* mode (Moughamir, et al., 2002), requires a rather complex man-machine interaction. To guarantee the safety aspects of this interaction, robots like MIT-MANUS (Krebs, et al., 1998) were designed. This robot, which is highly back-driveable (i.e., it has a low intrinsic endpoint mechanical impedance) and whose low inertia is almost isotropic, is particularly dedicated to the rehabilitation of the cerebral vascular victims. But one of the major disadvantages of these robots, arising from the non-utilization of active force feedback in their control, is the impossibility of varying the apparent inertia felt by the subject to suit the requirements of a gradual rehabilitation. To avoid this inconvenience, the developed prototype (Fig. 1) is not back-driveable and its inertia is not isotropic. Thus, it becomes possible to use an active force feedback to vary the apparent inertia (Clover, 1999; Van der Linde, *et al*., 2003).

In this paper, we present the control architecture of the 3 dof system (Section 2) and the synthesis of the control law for the *Active-Assisted* mode of rehabilitation (Section 3). Simulation results of the flexion/extension movement in the sagittal plane are given in Section 4 to illustrate the performances of this control law.

2. CONTROL STRUCTURE

The architecture of the system (Fig. 2) consists of a mechanical part, a human-machine interface (HMI) and a software development environment. The mechanical part comprises the 3-dof robotized arm with 3 motors-reducers and an effort sensor to measure the forces applied by the user at the robot end-effector. A numerical 3 axes servo-controller is used to control the positions, speeds or torques of the 3 motors, according to training or rehabilitation requirements. The control laws are synthesised using dSPACE and Matlab/Simulink environment, and then compiled and implemented on a DSP board.

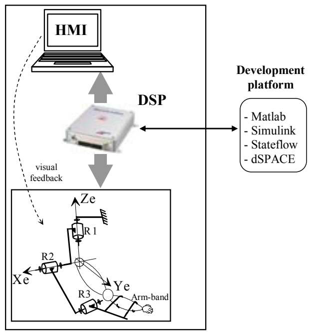

Fig. 2. Architecture of the system.

The human-machine interface allows users to choose a training exercise from a pre-established data base and to select the required parameters.

To facilitate the correct execution of the exercise, a screen is used to provide visual information about the measured forces and the execution of the real movement trajectory compared to the desired one.

The hierarchical control structure (Fig. 3) is inspired from the generic framework, proposed in Moughamir, *et al*. (2002) for the specification and design of any training and rehabilitation machine. This hierarchal structure comprises two parts: a sequential controller divided into 3 levels and a continuous switching control block corresponding to the control laws which are selected to carry out a training session.

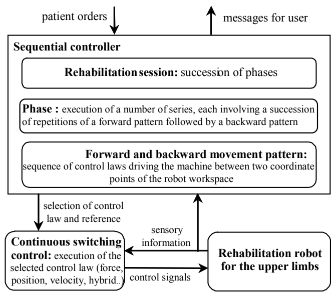

Fig. 3. Hierarchical control structure.

The module corresponding to a Rehabilitation session (fig. 3) is used to co-ordinate the rehabilitation (or training) modes and the consecutive phases forming the session. A phase is given by a succession of training series separated by a period of muscular relaxation. Each series comprises a number of repetitions of a particular *forward* trajectory pattern followed by a particular *backward* trajectory pattern, which are selected as a function of the training type required for the current phase.

This information is to be given by a physiotherapist. The modules of level 3 represent the states and the switching control sequences required to perform a given movement pattern. Many modules, corresponding to movement patterns, were developed to drive the upper limb(s) between two coordinate points of the robot workspace. Each of these modules invokes one or more control laws (position, velocity, force, impedance) as well as a predefined trajectory. The next section presents the control design for the *Active-Assisted* mode, which implements two concurrent control laws: force and force/impedance.

3. CONTROL LAWS

For economic reasons, the majority of force/position control schemes implemented for industrial robots do not compensate the nonlinearities of the dynamic model of the robot. They are generally based on a PID controller with gravity compensation to guarantee good static performances. However, during training and rehabilitation session, both static and dynamic performance requirements must be guaranteed. Therefore, he robot controller must fully compensate the nonlinearities of the dynamic model. A robot model constrained by the environment can be given by:

$$\Gamma = A(q)\,\ddot{q} + n(q,\dot{q}) + J^T F \qquad (1)$$

where q represents the angle positions vector, Γ the torque vector, A the matrix of inertia, n the vectorial sum of the centrifugal and Coriolis torques, gravitation torques and viscous frictions torques, J the jacobian matrix, F the vector of contact forces and torques supported by the robot end-effector. If all the elements of the dynamic system are supposed to be measurable, the control law given by

$$\Gamma = A(q)\,y + n(q,\dot{q}) + J^T F \qquad (2)$$

leads, with a global feedback linearization of the system (1), to the following linear and decoupled system:

$$\ddot{q} = y \qquad (3)$$

Where A is a positive definite matrix. The choice of the new input vector y depends on the required control (position, force, impedance…).

This well-known inverse dynamics control, ensures a good tracking of the trajectory if the parameters of the dynamic model are known with sufficient accuracy. Despite the complexity of this control law is complex, in terms of architecture and calculation, its implementation is feasible thanks to the advent of increasingly fast microprocessors.

The first requirement of the *Active-Assisted* mode is to let the robot end-effector behave like a simple inert mass. The goal of the control law is thus to assign the following dynamics to the end-effector:

$$M_d\,\ddot{x} = F + M_d\,g \qquad (4)$$

where M_d is the matrix of desired inertia, F is the vector of the contact forces and torques supported by the robot end-effector, g is the vector of gravity acceleration and x is the position vector in the operational space. The resulting control scheme is similar to a force controller as (4) can be written as follows:

$$\begin{aligned} &F_r - F = -M_d\,\ddot{x} \\ &\text{where} \quad F_r = -M_d\,g \end{aligned} \qquad (5)$$

The measured force, F, is compared with the reference value, F_r, to determine the dynamics of the movement. The integration of relation (4) into the inverse dynamics control, confers to the robot a similar behaviour to that of an ideal weight machine, with the additional advantage of the ability to execute complex movements, without frictions (Moughamir, et al., 2002).

To avoid a large movement deviation from the desired trajectory, we use the impedance control (Hogan, 1985), usually employed in the man-robot interactions for its advantages in term of safety. In our case, the dynamics of the movement only depends only on the subject and not on desired velocities and accelerations, and the impedance model is given by:

$$F = K\left(x - x_d\right) + B\dot{x} + M\ddot{x} \qquad (6)$$

This relation provides the desired impedance to the robot end-effector. When contact forces arise, the real position x will deviate from the desired position x_d to satisfy (6) and, hence, to establish a compromise between force and position according to the matrices of stiffness K, damping B and inertia M. Relations (4) and (6) can be combined within the framework of the inverse dynamics control as follows:

$$\begin{aligned} \ddot{x}_F &= \alpha\left(M_d^{-1}F + g\right) \\ \ddot{x}_I &= (I-\alpha)M^{-1}\left[K\left(x_d - x\right) - B\dot{x} + F\right] \\ \ddot{x} &= \ddot{x}_F + \ddot{x}_I \\ y = \ddot{q} &= J^{-1}(\ddot{x} - \dot{J}\dot{q}) \end{aligned} \qquad (7)$$

where $x = L(q)$ direct geometric model
and $\dot{x} = J(q)\dot{q}$ direct kinematic model

For the 3 dof robot:

$$\alpha = \begin{bmatrix} \alpha_1 & 0 & 0 \\ 0 & \alpha_2 & 0 \\ 0 & 0 & \alpha_3 \end{bmatrix} \quad \text{and} \quad \alpha_i = \frac{\tilde{x}_{i\max} - \left|\tilde{x}_i\right|}{\tilde{x}_{i\max}} \qquad \tilde{x}_{i\max} > 0$$

where $\tilde{x}_{i\max}$ is the upper limit of the error in the considered direction.

The resulting hybrid-structure controller is based on the weighted sum of force and impedance control. The weighting (α , I - α) is a function of the position

error. α allows fixing the respective weight of the contributions of force and impedance. According to the requirements, when the position error is small, the contribution of force prevails and, reciprocally, when the position error is significant, the impedance control prevails.
The control scheme (Fig. 4) requires compensating of nonlinear terms, inversing the jacobian matrix, measuring the forces at the end effector, and measuring the angular positions and velocities.

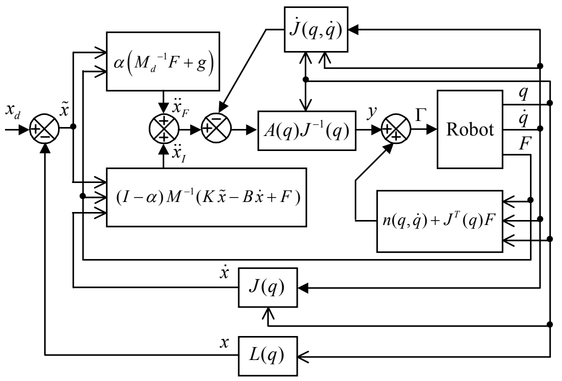

Fig. 4. Control scheme.

Compared with an impedance control alone, the proposed hybrid force/impedance control allows a better satisfaction of requirements imposed by the *Active-Assisted* mode. When the position error tends towards zero, the robot behaves indeed like an ideal weight machine. The parameter $\tilde{x}_{i\max}$ defines the subject's workspace in the considered direction. The larger this parameter is, the lower is the assistance provided by the robot. If this parameter is identical in all directions ($x_{i\max} = x_{\max} \quad \forall i$), the workspace at instant *t* is reduced to a sphere centred on the desired position $x_d(t)$ nearest to the real position $x(t)$ (Fig. 5).

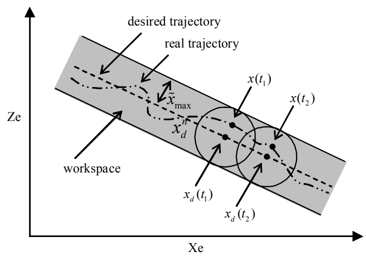

Fig. 5. Workspace in the sagittal plane (Xe, Ze).

The proposed controller forces the subject to remain in this space which, gradually, forms a safety space. To maintain the subject in this workspace, the position error is required to satisfy the following condition:

$$\left|\tilde{x}_i(t)\right| < \tilde{x}_{i\max} \qquad (8)$$

The satisfaction of this inequality depends obviously on the control parameters m_i, k_i and b_i, but also on the capacity of the subject to carry out the desired task.

By considering (7), it is advantageous to choose the matrix M equals to M_d to ensure a good compromise between the contributions of force and impedance and to render the inertia felt by the subject equal to desired inertia irrespective of the position error. The choice of the control parameters k_i and b_i is more delicate, because the control must be sufficiently compliant to guarantee the subject's safety while ensuring a good trajectory correction. According to relation (7), the inequality (8) imposes for any fixed position that:

$$k_i \geq \frac{f_{i\max}}{\tilde{x}_{i\max}} \qquad (f_{i\max} > 0) \qquad (9)$$

where f_{imax} is a higher limit of the maximum force that the subject is likely to exert in the considered direction. This parameter, to be fixed by a physiotherapist, depends on the state of the subject. In addition, k_i must be as small as possible to optimize the comfort of the user. The lower limit of the inequality (9) seems to be a suitable choice:

$$k_i = \frac{f_{i\max}}{\tilde{x}_{i\max}} \qquad (10)$$

To satisfy (8) during the movement, b_i must be tuned so that the force step response, f_{imax}, does not comprise oscillations. This parameter is tuned for the value which marks the appearance of the oscillations. This adjustment ensures the optimization of the dynamic performances of the hybrid control under the constraint of inequalities (8) and (10). In case where M=M_d, this limit corresponds to a fixed damping coefficient of impedance control:

$$\zeta_i = \frac{1}{2}\frac{b_i}{\sqrt{k_i m_i}} = 1,55$$

$$b_i = 2\zeta_i\sqrt{k_i m_i} = 3,1.\sqrt{k_i m_i}$$

It should be noted that for force steps smaller than $f_{i\,max}$ the response comprises small oscillations, but nevertheless the inequality (8) is always satisfied. In fact, these oscillations remain contained because they are filtered by the dynamic play of weightings (α_i , *1* - α_i).

Finally the time required to perform the movement only depends on the subject. The desired trajectory, specified by the physiotherapist, is only defined in term of positions. Define T_d as the set containing the desired trajectory in term of N regularly-spaced points:

$$T_d = \left\{x_d^0, x_d^1, x_d^2, \ldots, x_d^{N-1}\right\}$$

To refresh the desired position according to the real robot's position and to prevent a backward execution of the movement, the following me is used:

initially $n = 0$

$$\begin{cases} \text{While} \quad n+1 \leq N-1 \quad \text{and} \quad \left\|x_d^n - x(t)\right\| > \left\|x_d^{n+1} - x(t)\right\| \\ \text{Do} \quad n = n+1 \end{cases}$$

$$x_d(t) = x_d^n$$

The successive iterations allow refreshing the desired position by determining the minimal distance between the real position $x(t)$ and two successive points x_d^n and x_d^{n+1} of the set T_d.

For obvious safety reasons, the suggested control should be tested exhaustively before experimentation. Indeed, we propose in the following section, simulations which illustrate the potentialities of the controller in terms of dynamics performance and medical interest. In order to validate the control hierarchy and a part of the control architecture, the control laws were compiled and loaded in the DSP board

4. SIMULATION

The aim of this simulation is to illustrate the performances of the proposed controller within the frame of neuromuscular rehabilitation. To simplify the presentation, we will consider the movement of flexion/extension in the sagittal plane, whose desired trajectory brings into play only two of the robot's three degrees of freedom. We will neglect small displacements in the horizontal direction when the movement is carried out in a natural way. The initial position of the robot as well as the trajectory taking into account displacements of the shoulder (glenohumeral) joint (Walmsey, 2001) are depicted in figure 6.

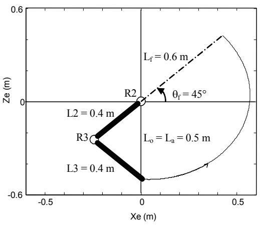

Fig. 6. Initial robot's position and desired trajectory.

The origin of the reference mark, chosen in the sagittal plane (Xe,Ze) to describe this trajectory, coincides with the estimated initial position of the gleno-humeral rotation centre of the subject. This origin is at the intersection of the two axis of rotation of the robot's shoulder (Fig. 1). The axis Ze is directed to the top, in opposite direction to the gravity acceleration. From the initial position, the subject's arm, of length L_a = *50 cm*, is directed downwards, $\theta_0 = -90°$. In the final position, $\theta_f = 45°$, the length L between the origin of the reference mark and the wrist of the robot is: L_f = *60 cm*. Thus, we suppose that the rotation axis misalignment of the subject's shoulder relatively to the rotation axis of the robot's shoulder, R2, entails a *10 cm* increase of the length L for this motion range of *135°*. L2 and L3 are both fixed at *40 cm*. Thus, any proximity with the singular position of the robot (tended arm) is impossible since the sum of these two lengths largely exceed the length of the subject's arm. To form the set T_d, the desired trajectory is discretized in $N = 1000$ points (*1* point per *1,3 mm*). If one neglects the effect of the sampling (Te = *1ms*), the robot's dynamic model is completely compensated. The desired isotropic mass is fixed at *1 kg* and the acceleration of gravity at *9,81 m.s*$^{-2}$. For all directions, we consider that the subject cannot exert forces higher than $f_{i\,max}$ = *100 N*. To simulate his/her behaviour, we suppose that the subject exerts a force of *1 N* in the direction of the anticipated position error $x_d^{n+1} - x(t)$ and entirely compensates the gravity force:

$$F(t) = \frac{1}{\left\| x_d^{n+1} - x(t) \right\|}\left(x_d^{n+1} - x(t) \right) - Mg \qquad (11)$$

In order to study the control reactions with respect to abrupt force variations induced by an abnormal behaviour of the subject (injury, weakness, tiredness…), we add a sinusoidal disturbance (with a frequency of *3 Hz* frequency and amplitude of *20 N* in the vertical direction) to relation (11). The forces simulated in the 2 directions of the sagittal plane are given by:

$$f_{Xe}(t) = f_1(t) = \frac{1}{\left\| x_d^{n+1} - x(t) \right\|}\left(x_{d1}^{n+1} - x_1(t) \right)$$

$$f_{Ze}(t) = f_3(t) = \frac{1}{\left\| x_d^{n+1} - x(t) \right\|}\left(x_{d3}^{n+1} - x_3(t) \right) - m_d g_3 + 20\sin(6\pi t)$$

$$\text{with: } \quad m_d = m = 1\,kg \quad \text{and} \quad g_3 = -9.81\,m.s^{-2}$$

Figure 7 depicts the simulation results for two workspace values ($\tilde{x}_{i\max}$ = *10* and *5 cm* for all directions). The other control parameters are selected according to the method described before and are given by:

for $\tilde{x}_{i\max} = 10$ cm :

$$M = M_d \qquad B\left[Nm^{-1}s \right] = 98.I \qquad K\left[Nm^{-1} \right] = 1000.I$$

for $\tilde{x}_{i\max} = 5$ cm :

$$M = M_d \qquad B\left[Nm^{-1}s \right] = 139.I \qquad K\left[Nm^{-1} \right] = 2000.I$$

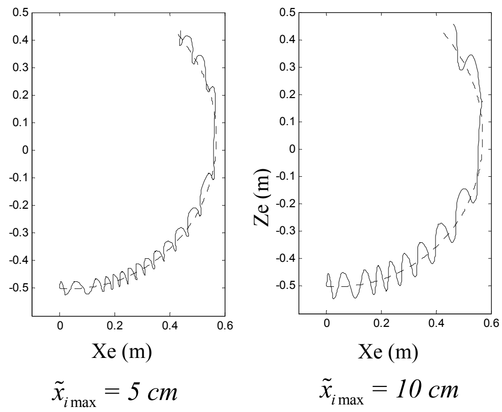

Fig. 7. Simulation of the flexion/extension movement.

For both cases, the movement is completely achieved. The average norms of the vectors of position errors and the matrices α are given, respectively, by:

for $\tilde{x}_{i\max} = 10$ cm : $\|\tilde{x}\|_{moy} = 2{,}88$ cm and $\|\alpha\|_{moy} = 0{,}712$

for $\tilde{x}_{i\max} = 5$ cm : $\|\tilde{x}\|_{moy} = 1{,}61$ cm and $\|\alpha\|_{moy} = 0{,}678$

We notice that the positions errors are more significant if the workspace is large for similar simulated forces. Thus, workspace $\tilde{x}_{i\max}$ will determine the room for manoeuvre of the subject. The index $\|\alpha\|_{moy}$ informs us about the average rate of force control preponderance versus impedance control. This index is of medical interest within the framework of a gradual rehabilitation since it provides a quantification of the quality of execution movement. A higher percentage (about 68 % for $\tilde{x}_{i\max}$ = *5 cm* and 71 % for $\tilde{x}_{i\max}$ = *10 cm*) indicates a better accuracy of movement execution and a lower assistance provided by the robot.

5. CONCLUSION

This paper has presented the architecture and the control hierarchy of a 3 degrees-of-freedom robot destined for the rehabilitation of the upper limbs. This structure is inspired from a generic framework for the specification and design of any training and rehabilitation machine. For robotized neuromuscular rehabilitation in *Active-Assisted* mode, we proposed an original hybrid controller, based on a weighted sum of force and impedance contributions. This control, specially adapted to non back-driveable robots, uses an active force feedback and confers to the robot the behaviour of an ideal weight machine, without frictions, when position error is quasi null. Simulation results have shown that the controller is satisfactory in terms of dynamic performances and safety. However, the type of implementation (torque-based) and the use of an active force feedback raise the issue of stability (Lawrence 1988). To deal with the stability problems of the man-machine coupling, future simulations will integrate a musculo-skeletal dynamic model of the human arm in view to determine the stability conditions of the coupling as a function of the parameters of the proposed controller. Experiments carried out on the 3 dof robotized arm will then make it possible to tune these parameters precisely by taking into account the subject's pathology.

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