A Compliant exoskeleton for multi-planar upper limb physiotherapy and training

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Abstract - In general, treatment of full or partial loss of function in the upper limb due to injury relies on extremely labour intensive physiotherapy procedures. Although mechanical assistive device exist for limbs this is rare for the upper body. In this paper we present a seven degree of motion prototype upper arm training/physiotherapy (exoskeleton) system. The total weight of the uncompensated orthosis is less than 2kg. This low mass is primarily due to the use of a new range of pneumatic Muscle Actuators (pMA) as power source for the system. This type of actuator, which has also an excellent power/weight ratio, meets the need for safety, simplicity and lightness. The work presented focuses on the physiotherapy and training control architecture used to make the system a multi-functionality facility which permits the execution of physiotherapy and training regimes under isokinetic, isotonic and training mode operation.

Keywords: Exoskeleton; upper limb; physiotherapy; training; pneumatic actuator

1. INTRODUCTION

Full or partial loss of function in the shoulder, elbow or wrist is an increasingly common ailment associated with a wide range of injuries, disease processes, and other conditions including sports, occupational, spinal cord injuries, and strokes. Typically treatment for these conditions relies to some extent on manipulative physiotherapy procedures which by their very nature are extremely labour intensive requiring high levels of one to one attention from highly skilled medical personnel. Major benefits can be gained in terms of the overall healthcare provided through the use of physiotherapy orthotics. For lower limb rehabilitation and physiotherapy there is an increasingly large and well regarded range of mechanical assistive products that aim to improve the quality of the rehabilitation process. Unfortunately this is not generally true for upper limb physiotherapy processes, although there have been a small number of significant devices ranging from passive mechanical arm support devices to electrically powered arm orthotic systems.

One of the earliest upper limb orthosis was the Balanced Fore arm Orthosis (BFO). This was a wheelchair mounted passive device developed in 1965 to enable a person with weak musculature to move their arms in the horizontal planes. A later version of the same device incorporated additional joints at the base to allow additional vertical movements. In this case the weight of the orthosis was compensated by means of rubber bands attached to the joints, but this approach gave poor gravity compensation and the device was rarely used [1]. In 1975 the Burke rehabilitation centre developed a 5 dof version of the BFO powered by means of electric motors but this never gained significant acceptance [2]. The Hybrid Arm Orthosis (HAO), developed by [3], aimed to provide upper arm motion assistance. This system offered shoulder abduction and elbow flexion, wrist supination and a three joint jaw chunk pinch. The shoulder and the elbow joints were body powered while, the wrist supination and the three-point jaw chuck pinch power were generated by two separate DC motors.

More recently a system developed at MEL [4] used a parallel mechanism to suspend the upper arm at the elbow and wrist level. Each point was suspended by an overhanging plate using three strings arranged in parallel. Motion of the upper limb was generated by changing the length of each string according to the command given by the user using voice or head motion. Only simulated results have been presented for this system.

Among the most interesting of the powered orthosis is a motorised upper limb orthosis system (MULOS) developed at the University of Newcastle in mid 90's [5-7]. The developed system has 5 degrees of freedom (dof) and is designed to work in 3 different modalities: Assistive, Continues Passive Motion (CPM) and Exercise. This device appeared to have good potential but development was discontinued in 1997.

MIT MANUS is another robotic system that has been developed [8,9] for the physical therapy of stroke victims. In the robot-aided therapy, a video screen prompts the person to perform an arm exercise. If movement does not occur, MIT-Manus moves the person's arm. If the person starts to move on his/her own, the robot provides adjustable levels of guidance and assistance to facilitate the person's arm movement. The system supports up to 5dof through two separate modules. Although it is claimed to be portable its weight is around 45kg.

The ARM Guide [10,11] consists of an instrumented linear constraint that can be oriented in different directions across the subject's workspace using a three-splined steel shaft. The subject is attached to the ARM Guide using a custom splint that rides along the linear constraint. This device was primary used as a measurement tool for force and motion during mechanically guided movement.The most recent system of all is the GENTLE project [12], which uses a haptic master device to provide exercise for an arm using an interactive virtual environment. The system has 6 dof but only the three translational motions are active. It is clear from the structures considered above that there are still significant problems to be solved in the development of upper limb rehabilitation orthosis that will fulfil the aspirations of the patient and the medical and technical communities. Issues that are of particular concern include:

i) Orthosis mass.
ii) Accurate automatic compensation for gravity forces.
iii) Provision of a multipurpose facility for upper limb training and joint motion assisting/analysis in up to 7 dof.
iv) Safe operation and importantly safe perception from the patient's viewpoint.
v) Reliability in all operations and in environments where materials like water, dust or grease are presented.
vi) Relatively low complexity and low engineering and construction cost.

vii) Simple fitting and removal.
viii) Low/no maintenance.

In this paper we describe the construction and control of a seven degree of motion prototype upper arm training/rehabilitation (exoskeleton) system. The total weight of the uncompensated orthosis is less than 2kg. This low mass is primarily due to the use of a new range of pneumatic Muscle Actuators (pMA). The need for safety, simplicity and lightness is met by this type of actuator, which has also an excellent power/weight ratio.

The work presented will show how the system takes advantage of the inherent controllable compliance to produce a unit that is extremely powerful, providing a wide range of functionality (motion and forces over an extended range) in a manner that has high safety integrity for the patient. The general layout of the arm orthosis is presented. This includes the design requirements, and the design description. The control issues of the system are discussed. The physiotherapy and training control schemes which are used to control the orthosis during isokinetic, isotonic or training mode are introduced. Initially, the low level joint control of the system is presented. Finally a training control scheme is introduced which is used to control the orthosis when used as an exercise facility.

2. DESIGN REQUIREMENTS

Based on the knowledge of user needs and the system requirements outlined above, the fundamental technical specifications for the upper limp rehabilitation training system are:

i) A structure having low mass /inertia.
The mass of the trainer must be kept to a minimum, as it interferes with the forces transmitted from the actuators to the human. A device with large mass requires excessive use of the actuator power to counterbalance gravity effects. Larger actuators add to the mass and cost and increase safety concerns.

ii) Safety.
As the system is in direct contact with the human operator the safety requirement is paramount for a rehabilitation device.

iii) Comfort of wearing
As the extended use of the device is certainly possible and probably necessary, the device must be comfortable, causing no fatigue to the operator even after long periods eg. 1-2hrs, of operation. This requirement should include ease of fitting adjustment and removal.

iv) Extensive range of motion.

A generic specification for the display range of motion can be defined as the workspace of the human arm motion.

v) Support of different physiotherapy and training exercises.
The system must be able to execute different exercises including isokinetic and isotonic exercises. During these exercises accurate simulated forces (isotonic) and velocities (isokinetic) means that the device must have sufficient force and position resolution capabilities. The motion sensing requirements of the device obviously depend on the position resolution capabilities of the human. Since the human position sensing resolution varies for different joints in the human body, the motion sensing resolution will be determined by the sensing capabilities of the part of the body to which the system is attached. Joint position resolution of the human arm varies from 0.8° at the shoulder to 2.0° at the wrist [13,14].

vi) Accurate, automatic compensation for gravity forces.
Since the application will by its nature involve individuals with at best weakened arm musculature, the mass of the rehabilitation aid and possibly the patient's arm will impinge on the exercises and motions that can be attained. To enhance this it is important that active easily updated gravity compensation forms a keystone of the design.

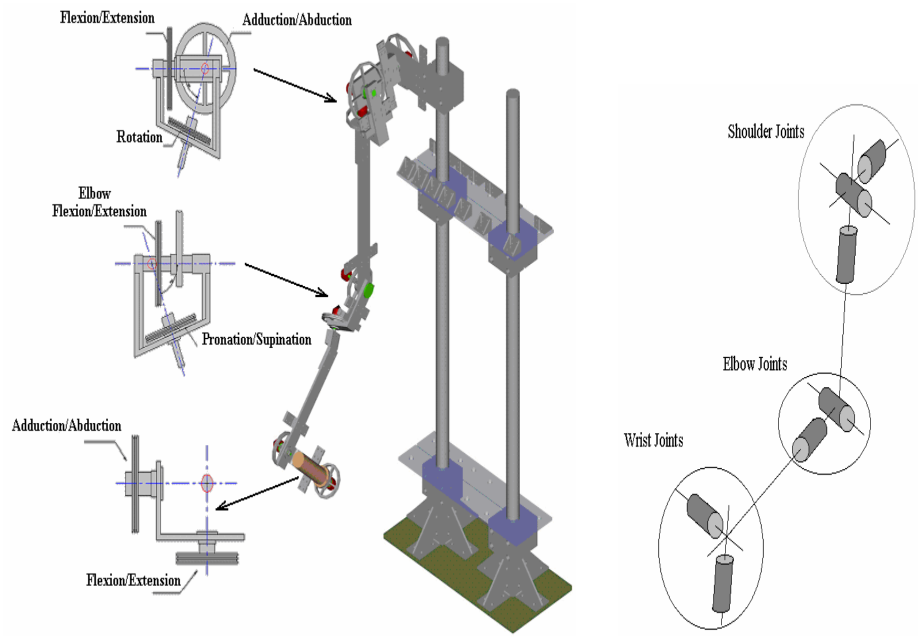
vii) Complexity
As with most designs, options that keep complexity to a minimum will tend to improve reliability, and reduce cost and these should always be under consideration during the design process.

3. DESIGN DESCRIPTION

3.1. Mechanical Structure

The mechanical arm structure to be used as the basis for this system has 7 dof corresponding to the natural motion of the human arm from the shoulder to the wrist but excluding the hand. The structure, figure 1, which can be seen to function as a powered exoskeleton, has 3 dof in the shoulder (flexion/extension, abduction/adduction and lateral-medial rotation), 2 dof at the elbow permitting flexion/extension, pronation/supination of the forearm and 2 dof. at the wrist (flexion/extension and abduction/adduction).

The arm structure is constructed primarily from aluminium and composite materials, with high stress joint sections fabricated in steel. The arm is constructed for use by a 'typical adult' with only minor changes to the set-up. Arm link length changes can easily and quickly be effected, if necessary making it easy to accommodate a range of users which is an important aspect of the design. High linearity sensors are employed to perform the position sensing on the joints.



(a) Arm Structure (b) Arm Kinematics

Figure 1. Exoskeleton mechanical structure.

3.2. Actuation System

The nature of the drive source in this system forms a key sub-system within this physiotherapy/training unit making use of the "soft" nature of the actuator operation. This is very different from methods previously developed and is a key to the success of this technique. This system uses braided pneumatic Muscle Actuators (pMAs) that provide a clean, low cost actuation source with a high power/weight ratio and safety due to the inherent compliance. The type of pneumatic actuator was firstly used for applications in rehabilitation robotics by McKibben [15]. At the time the power/weight performance of the system and the inherent compliance were seen as positive features but control was still a problem and the mechanism development was discontinued. In the 1980's the principle of the pMA was resurrected by Bridgestone in their Rubbertuator and this was incorporated into their "Soft arm" robot which was produced for a number of years [16]. By 1990's work on the Rubbertuator had stopped but several other groups had noted the potential of this form of actuation [17-21]. New applications were also identified particularly in the area of bio-robotics [22-24], and rehabilitation [25,26], however, applications were also found in unrelated domains such as the nuclear industry [27].

The pneumatic Muscle Actuators are constructed as a two-layered cylinder, figure 2. Within each actuator a pressure sensor has been incorporated to monitor the internal state of the muscle. The complete unit can safely withstand pressures up to 700KPA (7 bar), although 400kpa (4 bar) is the operating pressure for this system.

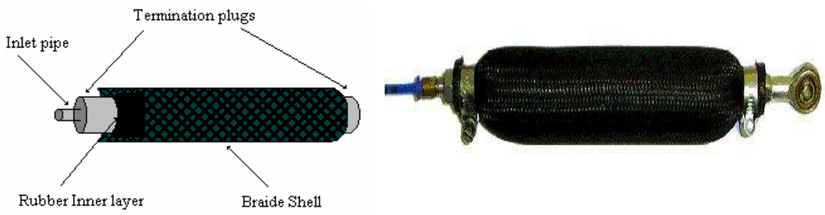


Figure 2. Pneumatic Muscle Actuator Design.

The detailed construction, operation, and mathematical analysis of these actuators can be found in [16-19,28]. The structure of the muscles gives the actuator a number of desirable characteristics [29].

i). Actuators have exceptionally high power and force to weight/volume ratios >1kW/kg.

ii). The actual achievable displacement (contraction) is dependent on the construction and loading but is typical 30%-35% of the dilated length - this is comparable with the contraction achievable with natural muscle.

iii). Being pneumatic in nature the muscles are highly flexible, soft in contact and have excellent safety potential. This gives a soft actuator option, which is again comparable with natural muscle.

iv). Force and position control using antagonistic pairs for compliance regulation is possible. This is once more comparable with natural muscle action.

v). The actuators are highly tolerant of mechanical (rotational and translational) misalignment reducing the engineering complexity and cost.

It is worth noting that a commercial form of the braided actuator with characteristics similar to the pMA is available from Festo. While it is possible to use these actuators they were not selected since in-house manufacture permits greater control over the dimensions, forces and general performance of the drives allowing them to be tailored for this application. Joint motion/torque on the rehabilitation/training arm is achieved by producing appropriate antagonistic torques through cables and pulleys driven by the pneumatic actuators. Two elements work together in an antagonistic scheme simulating a biceps-triceps system to provide the bi-directional motion/force, figure 3.

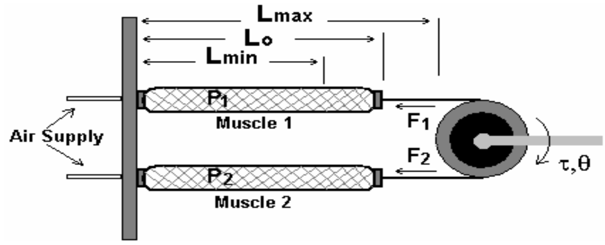


Figure 3. Antagonistic pairs of muscles.

In the above set-up $L_{\min}$ denotes the length of the muscle when it is fully contracted, $L_o = L_{min} + \frac{L_{max} - L_{min}}{2}$ is the initial dilated length of muscle which is equal to the half of maximum muscle displacement in order to maximize the range of motion of the joint, $L_{\max}$ is the maximum dilated length, r is the radius of the pulley and P_1, P_2 are the gauge pressures inside the two muscles. Modelling of the joint dynamics is based on a simplified model that considers the pMAs as a nonlinear spring with an elastic constant that is related to the muscle properties. Details on the development of the joint model can be found in [17,30]. Figure 4 shows the completed system with the pneumatic actuators mounted at the support structure of the exoskeleton system.


Figure 4. The system developed with the pneumatic muscle actuators.

Flexible steel cables are used for the coupling between the muscles and the pulley. Since most of the joints require a range of rotation in excess of 90°, double groove pulleys have been employed. The pulleys have been made from solid aluminium pieces internally machined to form a 4 spoke structure. On each of these internal spokes strain gauges are mounted to form a joint torque sensor. The pulleys are fastened on the joint shafts and rotate on bearings to minimise friction. The compact actuator structure allows for integration close to their respective powered joints. This makes the overall design compact, in line with the design requirements. The muscles used in this project have a diameter of 2cm to 4cm with an 'at rest' length varying from 15cm to 45cm. The factors that determine the length and the diameter of the actuator are range of motion and the required torque at the joint.

3.3. Control and Interfacing Hardware

The overall schematic of the system hardware is shown in figure 5. The activation of the pMA is reliant on the effective control of the pressure inside the muscles. This is controlled by MATRIX valves that incorporate 4 3/3 controllable ports in a package having dimensions of 45mm x 55mm x 55mm and weighing less than 320g.

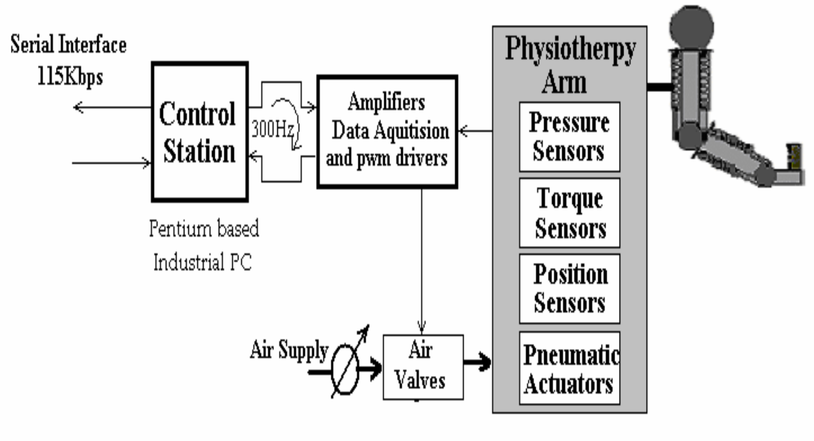


Figure 5. Control and interfacing hardware

The valves are controlled using a pulse width modulation (PWM) regime with a pulsing frequency of 100Hz, figure 6. The duty cycle of the pulsed signal forms a controlled variable as will be described in the next section. Development of an adaptive controller and details of the design of the system can be found in [18,28].

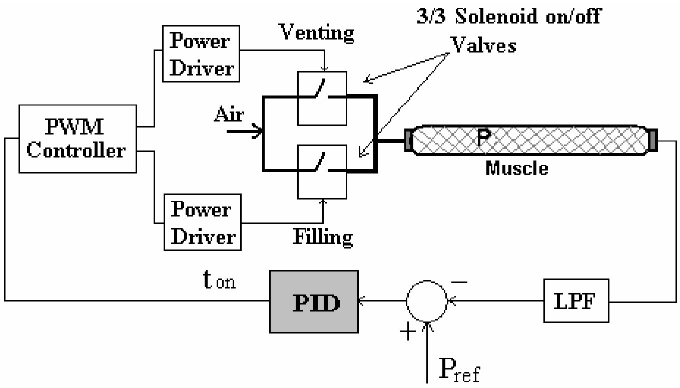


Figure 6. Control scheme of the muscle pressure using on/off solenoid valves

4. PHYSIOTHERAPY AND TRAINING CONTROL MODES

The control architecture of the device is designed to suit the requirements of the physiotherapy and training tasks. Three mode of operation are supported: Isokinetic Mode, Isotonic Mode and Training Mode. Figure 7 shows the relationship of these modes with the control layer of the device. Isokinetic mode (constant velocity) is used for execution of passive isokinetic exercises. In this mode the device joint velocity is explicitly controlled by an independent joint position control scheme where individual and multiple joint isokinetic exercises can be executed. Execution of coordinate motion exercises is also possible under this mode.

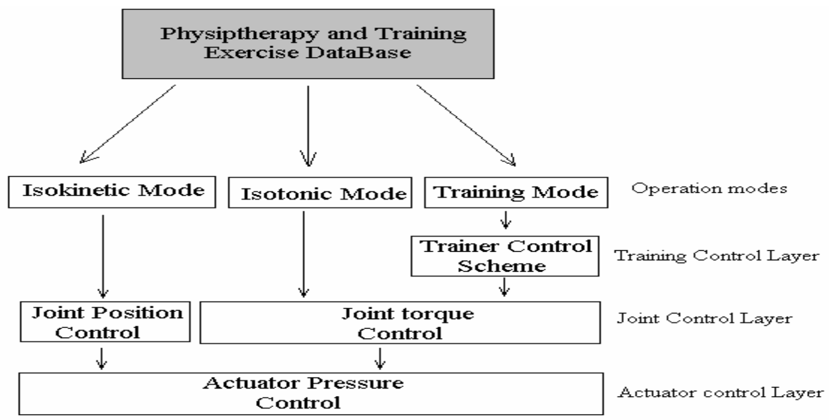


Figure 7. Layered control scheme and physiotherapy operation modes of the device.

The isotonic mode (constant force) enables the execution of individual or multiple independent joint isotonic exercises using joint torque control. Under training mode the device is controlled by a trainer control scheme based on impedance control, which enables the execution of muscle strengthening exercises. The different software control layers were implemented using an object-oriented approach. The interaction of the different software modules is shown in figure 8.

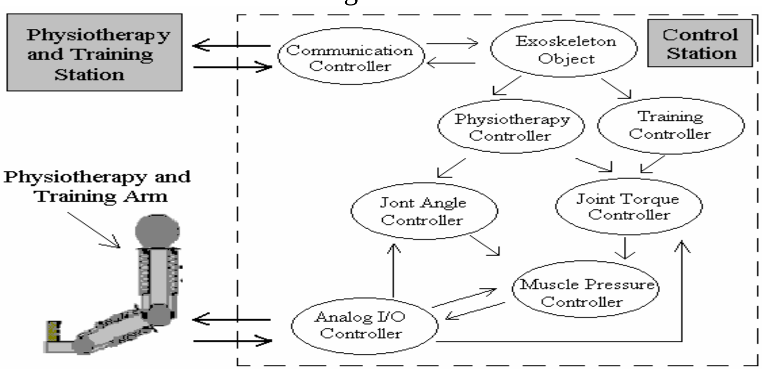


Figure 8. Object-oriented implementation of the device control.

Information exchange between the two stations is achieved through the communication controller. The joint control layer, (torque or position) depending on the operation mode, has been implemented on each joint using the scheme in figure 9. Using the feedback provided by the torque or position sensor on each joint, a torque or position control loop can be formed around each individual joint. This control loop uses the torque/position error to calculate the required amount of pressure change in the two muscles of the antagonistic pair. The command pressures for the muscles at each cycle are given similarly by:

$$P_1 = P_o - \frac{u}{2} \quad (1) \quad P_2 = P_o + \frac{u}{2} \quad (2)$$

Where u is the pressure control signal computed using a PID control law:

$u = K_{pr} \cdot e + \frac{1}{T_i}\int e + T_d \cdot \dot{e}$ (3) and $e = \tau_d - \tau_s$ (4) is the joint torque or position error depending on the operational mode. With this control method, open loop joint stiffness control is also possible by varying the amount of pressure, P_o, that is added to the output of the PID loop.

The coefficients of the PID law were experimentally estimated. These two command pressures form the input for the two inner pressure control loops, which form the actuator control layer. The pressure feedback signal is provided by means of pressure sensors contained within each muscle of the antagonistic pair. The output of these inner pressure control loops are the times t_1 and t_2 which corresponds to the duty cycle of the PWM signal that drives the solenoid valves. Positive values for t_1 and t_2 activate the filling valves while negative values switch on the venting sequences.

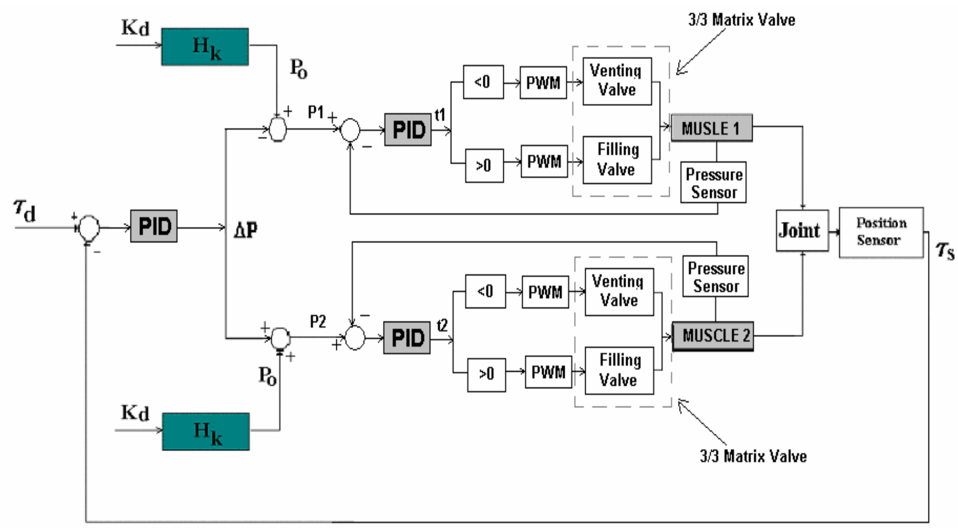


Figure 9. Block diagram of a single joint position or torque control scheme.

5. TRAINER CONTROL SCHEME

When the exoskeleton switches to the training mode of operation an impedance control scheme is employed for the overall exoskeletal training system to enable the execution of complex training exercises. The following equation describes the dynamic behaviour of exoskeleton.

$$M(q)\cdot\ddot{q}+V(q,\dot{q})+F(\dot{q})+G(q)+J^{T}\cdot F_{R}=\tau_{joint} \quad (5)$$

Where

q	is the joint variable n-vector
τ_{joint}	is the joint n-vector of the generalized torques
$M(q)$	is the inertia matrix
$V(q,\dot{q})$	is the coriolis/centripetal vector
$F(\dot{q})$	is the friction vector
$G(q)$	is the gravity vector

F_R is the force that the arm generates at the end-tip J^T is the transpose Jacobian of the manipulator

The above equation can be used to describe the interaction between an operator and the training exoskeleton. Considering the scenario described in figure 10 where the patient's arm is attached to the trainer.

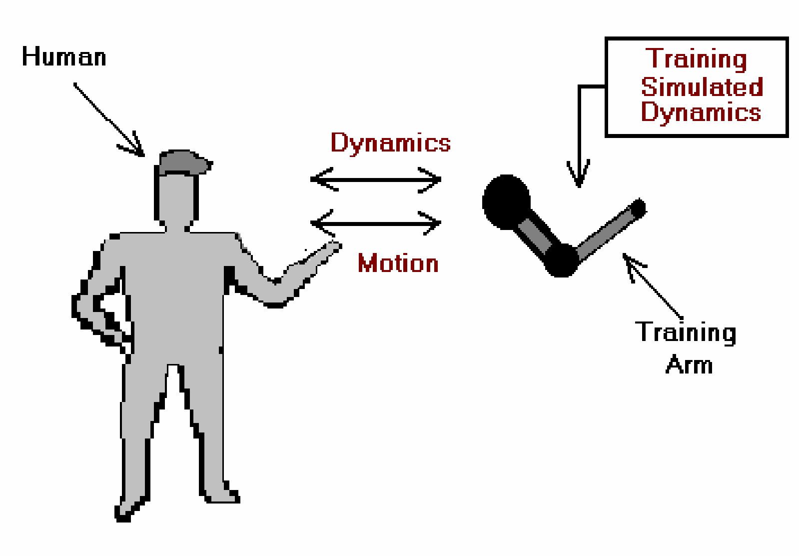


Figure 10. Diagram of the Human, and the Arm.

Let F_H denote the force that the human exerts on the training arm end-tip (which is actually the force felt by the user), F_R is the force that the arm applies to the operator and $Z_E(s)$ is the exoskeleton trainer simulated mechanical impedance. To make the operator feel the simulated training dynamics the following equation must be applied.

$$Z_E(s)\cdot(x-x_E)=M_E\cdot\ddot{x}+B_E\cdot\dot{x}+K_E\cdot(x-x_E)=F_H \quad (6)$$

Where M_E, B_E, K_E are the inertia, damping and stiffness coefficients. The above equation defines the desired characteristics of the motion of the pair (**Operator, Training Exoskeleton**). Having specified the desired behaviour of the system the control law now can be derived by eliminating $\ddot{x}$ and $\ddot{q}$ from equations (5) and (6). To do this the following equations, which relate the velocities and accelerations of the exoskeletal trainer end-point with the velocities and accelerations in joint space are introduced.

$$x = J\cdot\dot{q} \quad (7) \qquad \ddot{x} = J\cdot\ddot{q}+\dot{J}\cdot\dot{q} \quad (8)$$

Solving equations (6) and (8) for $\ddot{x}$ and $\ddot{q}$ respectively gives:

$$\ddot{x} = M_E^{-1}\cdot(F_H - B_E\cdot\dot{x} - K_E\cdot(x-x_E)) \quad (9)$$

$$\ddot{q} = J^{-1}\cdot(\ddot{x}-\dot{J}\cdot\dot{q}) \quad (10)$$

Combining, equations (5), (9) and (10) $\ddot{q}$ can be eliminated to give:

$$M(q)\cdot J^{-1}\cdot(M_E^{-1}\cdot(F_H - B_E\cdot\dot{x}-K_E\cdot(x-x_E))-\dot{J}\cdot\dot{q})+G(q)$$
$$=\tau_{joint} - J^T\cdot F_R \quad (11)$$

To keep the Cartesian inertia of the human arm/exoskeleton unchanged:

$$M_E = J^{-1}\cdot M\cdot J^{-T} \quad (12)$$

Considering that the motions are slow which is typical in physiotherapy based training applications and that $F_R = -F_H$, equation (11) gives

$$\tau_{joint} = -J^T\cdot(B_E\cdot\dot{x}+K_E\cdot(x-x_E))+G(q) \quad (13)$$

The above equation describes the impedance control law for the overall rehabilitation/trainer exoskeleton. The damping and the stiffness matrixes B_E and

K_E are 6x6 diagonal damping and stiffness matrices and depend on the training dynamics to be simulated. To enable effects such as static force to be simulated the control equation (13) can be modified by including a bias force matrix F_{bias} as follows:

$$\tau_{joint} = -J^T \cdot (B_E \cdot \dot{x} + K_E \cdot (x - x_E) + F_{bias}) + G(q) \quad (14)$$

Where F_{bias} is a 6x1 bias force/torque matrix, which can be used for simulation of special effects like virtual weight lifting.

6. SYSTEM PRELIMENARY PERFORMANCE MEASURES

A number of preliminary results are presented in the following sections to demonstrate the performance of the system.

6.1 Joint Motion Range

The first fundamental result to be presented is the arm exoskeleton joint motion ranges and the work-volume provided for the operator's arm. The motion range of each of the joints of the exoskeleton is shown in table 1. The first column gives the unrestricted range of motion of the human arm joints while the second column presents the limits of motion of each joint of the mechanical structure itself.

ARM MOTION	HUMAN ARM RANGE (°)	EXOSKELETON & ARM (°)
Wrist Flexion	90	92
Wrist Extension	99	92
Wrist Adduction	27	35
Wrist Abduction	47	46
Forearm Supination	113	82
Forarm Pronation	77	81
Elbow Flexion	142	103
Elbow Extension	-	16
Shoulder Flexion	188	105
Shoulder Extension	61	20
Shoulder Adduction	48	25
Shoulder Abduction	134	96
Shoulder Medial Rotation	97	52
Shoulder Lateral Rotation	34	52

Table 1. Motion ranges for the exoskeleton arm.

As can be seen in table 6.1 the ranges of joint motions of the arm exoskeleton do not entirely correspond to those of the human arm. The shoulder abduction/adduction and flexion/extension ranges of motion are narrower than the corresponding human arm motions. Motions above the operator's head and behind the operator's body are primarily restricted due to the mechanical structure and actuator displacement limit on the exoskeleton's shoulder. The shoulder medial rotation and the forearm supination have also a limited range of motion compared to the unrestricted human arm. The restriction of these motions is due to the size of the actuators (length) powering these particular joints. The range of these motions can be increased with the use of longer muscles, however, this may also require changes in the mechanical structure of the arm. The joint position resolution for the entire joint motions is 0.007deg.

6.2 Joint Torque Capability

The maximum output torque of each joint of the exoskeleton was also measured. The results for the seven joints of the exoskeleton are introduced in table 2.

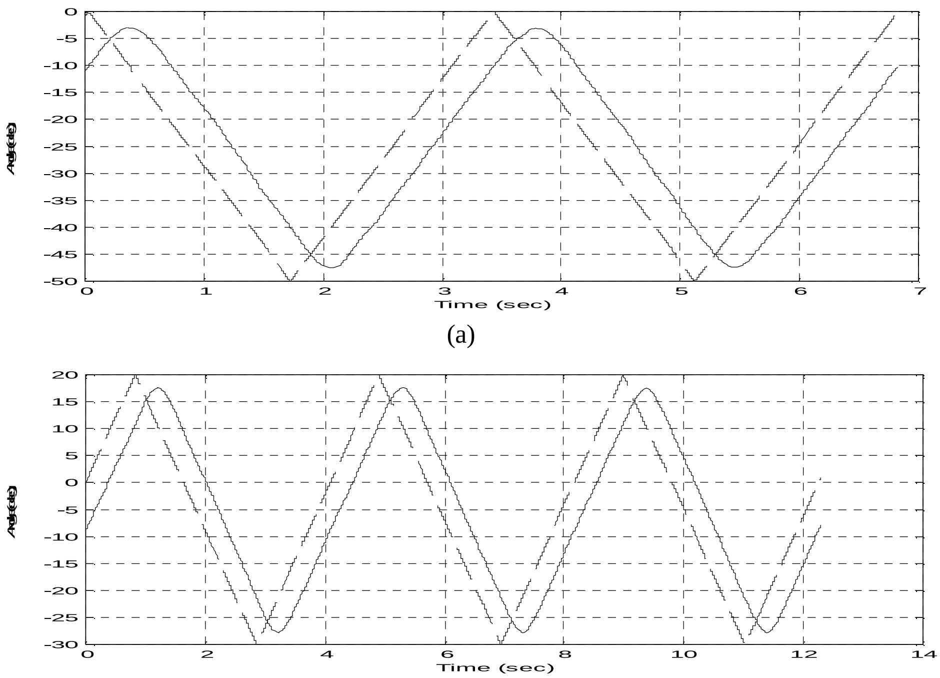
ARM JOINT	HUMAN ISOMETRIC STRENGTH [12]	MIN TORQUE	MAX TORQUE
Wrist Flexion/ Extension	19.8Nm	5Nm	28Nm
Wrist Adduction/Abduction	20.8Nm	7Nm	24Nm
Forearm Supination/ Pronation	9.1Nm	9Nm	25Nm
Elbow Flexion/Extension	72.5Nm	19Nm	76Nm
Shoulder Medial/Lateral Rotation	-	5Nm	46Nm
Shoulder Flexion/ Extension	110Nm	7Nm	95Nm
Shoulder Adduction/Abduction	125Nm	11Nm	128Nm

Table 2. Joint torque output for the exoskeleton arm.

The maximum output torque is not constant for the whole range of motion but depends on the joint position. This is because of the nature of the actuators used. The pneumatic muscles actuators exhibit spring characteristics exerting higher forces for longer dilated lengths. As a result of this the maximum output torque at the exoskeleton joints is also a function of the joint position. The minimum torque values in the above table appear at the borders of the joint motion space. Inside the area of the joint space that is of particular important during physiotherapy regimes the output torques are more than 30% of the human isometric strength which is probably adequate for the execution of the physiotherapy exercises. The torque resolution depends on the particular joint varying from 0.01Nm for the shoulder abduction/abduction joint to 0.003Nm for the joints of the wrist.

6.3 Isokinetic Mode Experiment

The performance of the control scheme was evaluated during isokinetic regimes for the elbow joint. In this isokinetic mode experiment the user's arm was securely attached to the exoskeleton which guided the human arm to follow a pre-programmed trajectory. Based on initial conditions such as speed and start and stop angles, a linear function was generated, which produced angle values for a series of temporal instances. These angle values feed the position controller, which in turn undertook to guide the exoskeleton in order to perform the exercise. The results are shown in figure 11. The dashed line denotes the desired input and the solid line the actual exoskeleton position as recorded by the position sensors. As it can be seen in the figure although there is a small delay between the isokinetic regime input and output this is not important as the shape of the output trajectory replicates well the regime input. These results indicate the initial feasibility of the exoskeleton as a potential physiotherapy training system.



(a)

Figure 11. Experimental results during isokinetic regimes for the elbow flexion/extension and supination/ pronation joint.

7. CONCLUSIONS / FURTHER WORK

An upper limb multipurpose rehabilitation, physiotherapy and training device was proposed. The mechanical structure, the actuation system and particularly the control architecture and the operational modes of the device were described. Attention is applied to the use of braided pneumatic Muscle Actuators which have extremely good power/weight ratios and due to their highly flexible and soft nature have beneficial attributes in applications were a powered device is in close proximity to a user. In addition, antagonistic action permits direct compliance control, which again has advantages in terms of safety and more human friendly ‘soft’ interaction providing a facility that is reminiscent of the compliance controlled feel of conventional human manipulation. The operational modes provided by the device enable the execution of isokinetic, isotonic and training regimes. To achieve this the device control architecture employs joint position/torque control for the execution of the isokinetic/isotonic exercises while training regimes are performed using an impedance trainer control scheme. In particular it has been shown that the device can be used as:

i) a physiotherapy exercise facility for the joints of the upper limb supporting both isokinetic and isotonic exercises.
ii) a training and strengthening exercise machine using an impedance based training control scheme.
iii) a motion analysis system.

Further studies will include the following:

i) Development of the physiotherapy training regimes to form a library of treatment procedures.
ii) The assessment of the system use as a power assist device, for people with weak arm muscles.
iii) Further development of the control architecture incorporating force amplification control strategies that will enable the execution of heavy physical activities or provide power assist for people with weak arm muscles.
iv) The development of software to demonstrate the use of the system as a motion and power analysis system.

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