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Development and Control of a 'Soft-Actuated' Exoskeleton for Use in Physiotherapy and Training

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Abstract. Full or partial loss of function in the upper limb is an increasingly common due to sports injuries, occupational injuries, spinal cord injuries, and strokes. Typically treatment for these conditions relies on manipulative physiotherapy procedures which are extremely labour intensive. Although mechanical assistive device exist for limbs this is rare for the upper body.

In this paper we describe the construction and testing of a seven degree of motion prototype upper arm training/rehabilitation (exoskeleton) system. The total weight of the uncompensated orthosis is less than 2 kg. 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 shows 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. A training control scheme is introduced which is used to control the orthosis when used as exercise facility. Results demonstrate the potential of the device as an upper limb training, rehabilitation and power assist (exoskeleton) system.

Keywords: pneumatic muscles, exoskeleton, rehabilitation, upper limb

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 injuries, occupational injuries, 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.

Reducing the task load for these professionals through the use of assistive orthotics could have major benefits in terms of the overall healthcare provided and the cost of this provision, while at the same time providing greater assess to effective rehabilitation regimes could be of significance to the healing process of the patient. Therapeutic results, from Reinkensmeyer et al. (2000), shows that medical benefits can gained in the chronic hemiparetic arm using active assist therapy. Indications up to date suggest that robot based rehabilitation regimes have a positive effect on the reduction of impairment of the human brain. Robots allow the control of the amount of exercise delivered to the subject and provide a patient's performance measuring tool (Krebs et al., 2000).

For lower limb rehabilitation there are 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 rehabilitation processes, although there have been a small number of significant devices ranging from passive mechanical arm support devices to electrically powered arm orthotic systems. Both approaches having advantages and disadvantages.

One of the earliest upper limb orthotics was the Balanced Fore arm Orthosis (BFO) (Alexander et al., 1992). 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 (Alexander et al., 1992). In 1975 the Burke rehabilitation centre developed a 5dof version of the BFO powered by means of current motors but this never gained significant acceptance (Stern and Lauko, 1975).

The Hybrid Arm Orthosis (HAO), developed by Benjuya and Kenney (1990), aimed to provide upper arm motion assistance. This system offered shoulder abduction and elbow flexion, wrist supination and a three joint jaw chunk pinch. Two different power sources were used to achieve this. The shoulder and the elbow joints were interconnected and simultaneously abduct and flex respectively by contra lateral shoulder elevation (body powered). Two separate DC motors generated the wrist supination and the three-point jaw chuck pinch power. Control of the orthosis motion was achieved with contra lateral shoulder movements and air switches operated by the head.

MIT/Manus is another rehabilitation robot developed for the physical therapy of stroke victims (Hogan et al., 1992; Krebs et al., 1998). A person sitting at a table puts the lower arm and wrist into a brace attached to the arm of the robot. MIT-MANUS can assist or guide the movement of the subject's upper limb and can record quantities such as position, velocity and force. Impedance control is used to regulate the compliance of the robot making the robot interact safely and gently with humans. The total weight of the system is 45 kgr.

Another system developed at MEL (Homma and Arai, 1995) used a parallel mechanism to suspend the upper arm at the elbow and wrist level. An overhanging plate using three strings arranged in parallel suspended each point. 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 orthoses is a motorised upper limb orthosis system (MULOS) developed at the University of Newcastle in mid 90's (Johnson and Buckley, 1997; Marchese et al., 1997; Yardley et al., 1997). This project aimed at empowering the disabled and elderly by the improvement, restoration or substitution of motor function in the upper limb. The developed system has 5 degrees of freedom (3 d.o.f at the shoulder, 1d.o.f at the elbow, 1 d.o.f for pronation/supination of the forearm) and is designed to work in 3 different modalities: Assistive, Continues Passive Motion (CPM) and Exercise. In assistive mode, the system acts as an amplifier of arm muscles and can lead the hand to a desired fixed position while being controlled by a joystick or other suitable input. In Continuous Passive Motion-CPM the system can apply pre-programmed cyclical movements to selected joints according to a pattern selected by the user or therapist. In the Exercise each joint of the machine can be moved only by the exertion of force. The force required to move a particular joint may be modified and the user has the option to evaluate the maximum force he can exert. Safety and control issues still need to be addressed in this project. This device appeared to have good potential but development was discontinued in 1997.

The ARM Guide developed by Reinkensmeyer et al. (1999) 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 measurement tool for force and motion during mechanically guided movement.

Researchers at the Department of Veteran Affairs along with researchers at Stanford University have developed and clinically tested three mechatronic systems (using PUMA robots) for post-stroke therapy (Burgar et al., 2000). The third generation was able to handle multiple functional movement patterns, to fully support the limb during 3D movements and to provide passive, active-assisted, resistive, and self-guided modes of therapy.

The GENTLE/S project (Johnson et al., 2002) is a three-year project in the European Community Framework 5 (starting in February 2000). Its objectives are to explore and identify "best therapies" for machine mediated stroke rehabilitation. The prototype is based upon a haptic master of 6 degrees of freedom (dof). Only the three translational are active. The system operates under three modes of control: passive, active assistive and resistive.

It is clear from the structures considered above that there are still significant problems to be solved in the development of upper limb rehabilitation orthotsis 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 testing of a seven degree of motion prototype upper arm training/rehabilitation (exoskeleton) system. The system presented here accommodates the issues mentioned above and has the following advantages when compared to other robotic rehabilitation facilities.

(i) Low mass
(ii) Excellent power/weight ratio with
(iii) Inherent safety
(iv) Natural compliance
(v) Ease of fabrication
(vi) Low Cost

The above advantages are due to the use of a new range of pneumatic Muscle Actuators (pMA) as power source for the system. The nature of this drive source in this system forms a key sub-system within this rehabilitation unit. This is very different from methods previously developed. The total weight of the uncompensated orthosis is less than 2 kg. Safety is met by this type of actuator due to limit displacement of this actuator, which has also an excellent power/weight ratio. This device compared to most of the other systems which use software control methods to regulate their compliance, takes advantage of the inherent controllable compliance of this type of actuator 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 is has high safety integrity for the patient. Finally, the high tolerance of this actuator to mechanical (rotational and translational) misalignments makes easy the construction of the system and keeps the engineering complexity and cost low.

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 also discussed. Initially the low level joint control of the system is presented. A training control scheme is introduced which is used to control the orthosis when used as exercise facility. Results from preliminary experiments demonstrate the potential of the device as an upper limb training, rehabilitation and power assist (exoskeleton) system.

2. Design Requirements

Based on the knowledge, and particularly the limitations of previous systems and the requirements outlined above the fundamental technical specifications for the upper limp rehabilitation training system are:

(i) *Light with low mass/inertia.* The mass of the training 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.
(ii) *Safety.* As the system is in direct contact with the human operator the safety requirement is paramount for a rehabilitation device. The device must be designed not only to be safe but also to seem safe from the operator's viewpoint.
(iii) *Comfort of wearing.* As the extended use of such a device is certainly possible and probably necessary, the device must be comfortable, causing no fatigue to the operator even after long periods e.g. 1–2 hrs of operation. This 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) *Accurate force feedback.* Accurate representation of the simulated training forces means that the device must have sufficient force resolution capabilities. The human arm force resolution capability for the different arm joints has been studied. Experiments from Tan et al. (1994) revealed that the average force resolution is around 0.36 N and tended to increase as the target force increases. Using this as a specification value will ensure that the operator feels no force discontinuities during the training exercises.

(vi) *Good motion sensing resolution.* The motion sensing requirements of the device obviously depends on the position resolution capabilities of the human. Since the human position sensing resolution varies for different joints in the human body (Tan et al., 1994; Kalawsky, 1993), the motion sensing resolution will be determined by the sensing capabilities of the part of the body that the system is attached to. The device should have position sensing capabilities at least equal to the position sensing of the part of the body. Considering an arm attached device the joint position resolution needs to match the joint position resolution of the human arm, which varies from 0.8° at the shoulder to 2.0° at the wrist (Tan et al., 1994).

(vii) *Accurate automatic compensation for gravity forces.* Since the application will by its nature involve individuals with at best weakened arm structures the mass of the rehabilitation aid and possibly the patient 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.

(viii) *Reliability*. As with all systems user acceptance is dependent to a large extent on the reliability and utility of the mechanism. It is therefore vital that appropriate design concerns are given to reliability in all operations and in environments where materials like water, dust or grease are presented. Issues in the area of maintenance will of course be of related importance.

(ix) *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 d.o.f. corresponding to the natural motion of the human arm from the shoulder to the wrist but excluding the hand. The structure, which can be seen to function as a powered exoskeleton, has 3 d.o.f. in the shoulder (flexion/extension, abduction-adduction and lateral-medial rotation), 2 d.o.f. at the elbow permitting flexion/extension, pronation-supination of the forearm and 2 d.o.f. at wrist (flexion-extension and abduction-adduction), Fig. 1.

The arm structure is constructed primarily from aluminium and composite materials, with high stress joint sections fabricated in steel. This resulted in a light, low cost and comfortable structure providing a stable platform. 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 adjusted, if necessary making it easy to accommodate a range of users, which is an important aspect of the design. Ease of wearing the device was considered of particular importance to enable the use of the system from persons with neurological disorders. To accommodate this two Velcro attachments located at the elbow and wrist level facilitate easy mounting and detachment of the device requiring minimum effort.

High linearity sensors are employed to perform the position sensing on the joints. Apart from the position sensor, each joint is equipped with a torque sensor. The torque sensor was implemented by integrating two strain gauges inside the pulley of each joint. The strains gauges were mounted on a internal spokes of each joint pulley as described later.

3.2. Actuation System

The nature of the drive source in this system forms a key sub-system within this rehabilitation/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 (pMA) that provide a clean, low cost actuation source with a high power/weight ratio and safety due to the inherent compliance. These pneumatic Muscle Actuators (pMA) are constructed as a two-layered cylinder, Fig. 2.

This design has an inner rubber liner, an outer containment layer of braided nylon and endcaps that seal the open ends of the muscle. Within the actuator pressure sensor have been incorporated to monitor the internal state of the muscle, while miniature strain gauge based load cells can be used to directly measure the force in any actuation system.The complete unit can safely withstand pressures up to 700 KPA (7 bar), although 600 kpa (6 bar) is the operating pressure for this system. The detailed construction, operation, and mathematical analysis of these actuators can be found

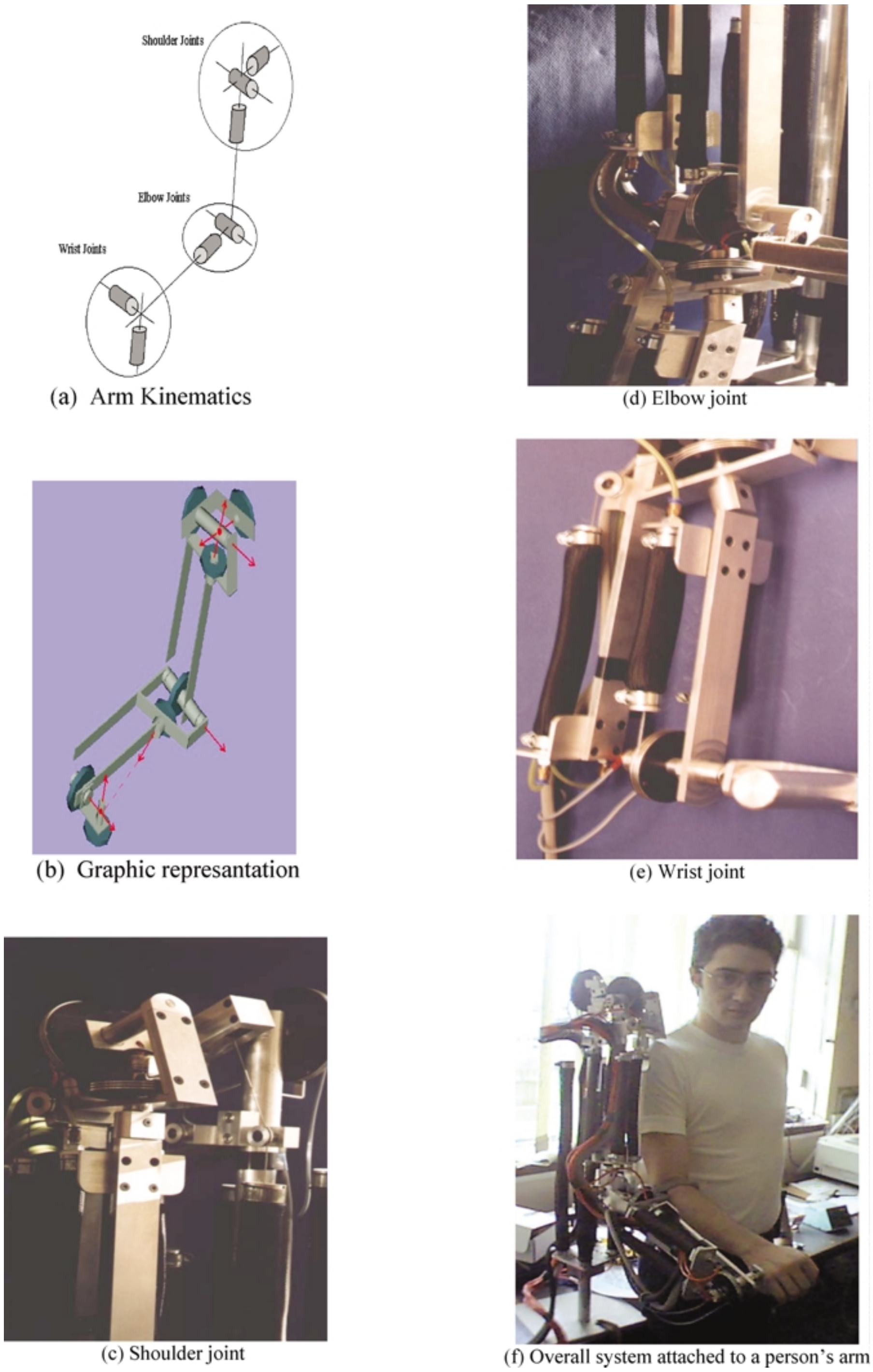

Figure 1. Exoskeleton mechanical structure.

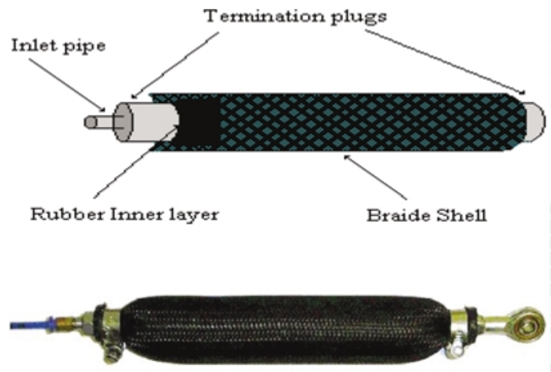

Figure 2. Pneumatic muscle actuator design.

in Caldwell et al. (1995), Chou and Hannaford (1996), and Tsagarakis and Caldwell (2000). The structure of the muscles gives the actuator a number of desirable characteristics (Caldwell et al., 1994):

(i) This muscle can be made in a range of lengths and diameters with increases in sizes producing increased contractile force.
(ii) Actuators have exceptionally high power and force to weight/volume ratios >1 kW/kg.
(iii) 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.
(iv) 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.
(v) Controllers developed for the muscle systems have shown them to be controllable to an accuracy of better than 1% of displacement.
(vi) Bandwidths for antagonistic pairs of muscles of up to 5 Hz can be achieved.
(vii) Force control using antagonistic pairs for compliance regulation is possible again comparable with natural muscle action.
(viii) When compared directly with human muscle the contractile force for a given cross-sectional area of actuator can be over 300 N/cm^2 for the pMA compared to 20–40 N/cm^2 for natural muscle.
(ix) The actuators can operate safely in aquatic, dusty or other liquid environments.
(x) The actuators are highly tolerant of mechanical (rotational and translational) misalignment reducing the engineering complexity and cost.

The activation of the pMA is reliant on the effective control of the airflow into and from the muscles. This is controlled by MATRIX valves that incorporate 4 3/3 controllable ports in a package having dimensions of 45 mm $\times$ 55 mm $\times$ 55 mm and weighing less than 320 g. The valves are controlled using a pulse width modulation (PWM) regime with a pulsing frequency of 100 Hz, providing rapid yet smooth motion. The duty cycle of the pulsed signal forms a controlled variable as will be described later. Development of an adaptive controller and details of the design of the system can be found in Caldwell et al. (1994, 1995).

It is worth noting that a commercial form of the pMA 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.

3.3. *Actuator Attachments*

Joint motion/torque on the rehabilitation/training arm is achieved by producing appropriate antagonistic torques through cables and pulleys driven by the pneumatic actuators. Since the pneumatic Muscle Actuator is a single direction-acting element (contraction only) this means that for bidirectional motion/force two opposal elements are needed. These two acting elements work together in an antagonistic scheme simulating a biceps-triceps system to provide the bidirectional motion/force, Fig. 3. Flexible steel cables are used for the coupling between the muscles and the pulley. Since

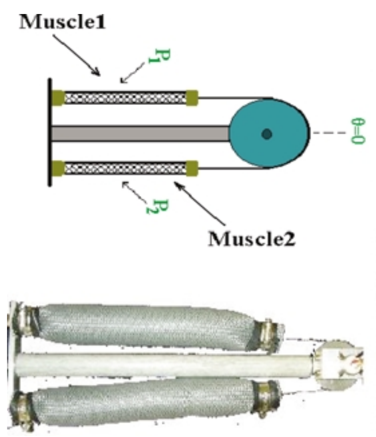

Figure 3. Antagonistic pairs of muscles.

most of the joints required 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 theses 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 force difference between the agonist and the antagonist muscle generates a positive or negative torque/motion at the joint. The compact actuator structure allows for integration close to their respective powered joints, which makes the overall design compact in line with the design requirements.

The wrist actuators (four actuators) are mounted on the forearm. The forearm pronation/supination actuators (two actuators) are mounted on a support structure, which lies parallel to the forearm link. Two idler pulleys at the elbow level are used to direct the coupling cables to the pronation/supination joint pulley. The elbow flexion extension actuators (two actuators) are mounted on the upper arm. The shoulder medial/lateral rotation actuators (two actuators) are fixed to a support structure similar to the one for the forearm rotation. The forces are transmitted in a similar manner through cables the direction of which is controlled by small idler pulleys on the shoulder level. The shoulder actuators (four actuators) are mounted on the body brace behind the operator's back. The shoulder adduction/abduction actuators are directly coupled with the adduction/abduction pulley, while the flexion extension joint is activated through cables, which are routed through the adduction/abduction joint to the flexion/extension pulley. Due to this routing, motion of the adduction/abduction joint affects the movement of the flexion/extension joint. The coupling between the two joints is minimised by reducing the diameter of the routing pulleys mounted on the adduction/abduction shaft.

The muscles used in this project have a diameter of 2 cm to 4 cm with an 'at rest' length varying from 15 cm to 45 cm. Two factors determine the length and the diameter of the actuator. The first is the required range of motion for the particular joint and the second is the torque required at that joint.

4. Joint Control

Joint torque control has been implemented on each joint, Fig. 4. Using the torque feedback provided by the torque sensor on each joint, a high bandwidth torque control loop can be formed around each individual joint.

The torque control loop uses the torque 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

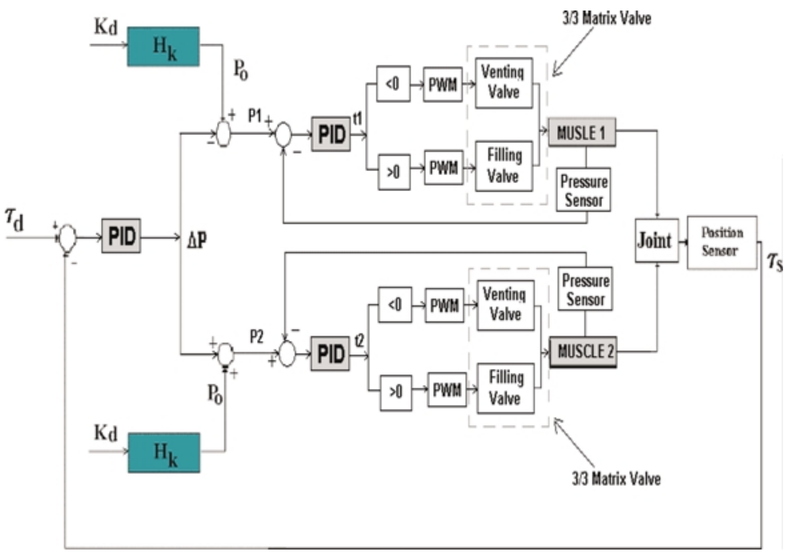

Figure 4. Block diagram of a single joint torque control scheme.

by:

$$P_1 = P_o - \Delta P \tag{1}$$

$$P_2 = P_o + \Delta P \tag{2}$$

Where ΔP is computed using a PID control law

$$\Delta P = K_{pr} \cdot e + \frac{1}{T_i} \int e + T_d \cdot \dot{e} \tag{3}$$

and

$$e = \tau_d - \tau_s \text{ is the joint torque error} \tag{4}$$

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. The pressure feedback signal is provided by the means of the 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. The consequence of the above control loop is that the actuator/joint system behaves as a pure torque source and provides improved torque response.

5. Trainer Control Scheme

An impedance control scheme was employed for the overall rehabilitation/training exoskeletal system. The following equation describes the dynamic behaviour of rehabilitation exoskeleton.

$$M(q) \cdot \ddot{q} + V(q, \dot{q}) + F(\dot{q}) + G(q) + J^T \cdot F_R = \tau_{\text{joint}} \tag{5}$$

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

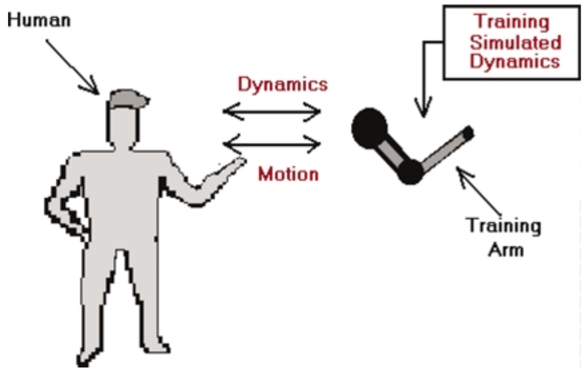

Figure 5. Schematic diagram of the human, and the rehabilitation training arm.

The above equation can be used to describe the interaction between an operator and the training exoskeleton. Considering the scenario described in Fig. 5 where the patient's arm is attached to the rehabilitation trainer. Patient and rehabilitation trainer interact with each other through motion and dynamic coupling. During the motion of the subject's arm dynamic or static forces are generated from the rehabilitation trainer according to the dynamic/static forces of the regime to be simulated.

Let F_H denote the force that the human exerts on the training arm endtip (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 trainer's simulated mechanical impedance. To make the operator feel the simulated training dynamics the following equation must be applied.

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

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 Eqs. (5) and (6). To do this the following equations, which relate the velocities and accelerations of the exoskeletal trainer end-point with the velocities, accelerations in the joint space are introduced.

$$x = J \cdot \dot{q} \tag{7}$$

$$\ddot{x} = J \cdot \ddot{q} + \dot{J} \cdot \dot{q} \tag{8}$$

Solving Eqs. (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, Eqs. (5), (9) and (10) $\ddot{q}$ can be eliminated to give:

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

To keep the cartesian inertia of the human arm/ exoskelton unchanged:

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

Considering slow motions typical in rehabilitation applications and that $F_R = -F_H$, Eq. (11) gives

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

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 6×6 diagonal matrices and depend on the training dynamics to be simulated.

$$B_E = \begin{bmatrix} {}^tB_E^x & 0 & 0 & 0 & 0 & 0 \\ 0 & {}^tB_E^y & 0 & 0 & 0 & 0 \\ 0 & 0 & {}^tB_E^z & 0 & 0 & 0 \\ 0 & 0 & 0 & {}^rB_E^x & 0 & 0 \\ 0 & 0 & 0 & 0 & {}^rB_E^x & 0 \\ 0 & 0 & 0 & 0 & 0 & {}^rB_E^x \end{bmatrix},$$

$$K_E = \begin{bmatrix} {}^tK_E^x & 0 & 0 & 0 & 0 & 0 \\ 0 & {}^tK_E^y & 0 & 0 & 0 & 0 \\ 0 & 0 & {}^tK_E^z & 0 & 0 & 0 \\ 0 & 0 & 0 & {}^rK_E^x & 0 & 0 \\ 0 & 0 & 0 & 0 & {}^rK_E^x & 0 \\ 0 & 0 & 0 & 0 & 0 & {}^rK_E^x \end{bmatrix}$$

The superscripts on the left of the coefficients refer to the translational and rotational motions while the superscripts on the right of the coefficient denote the direction of motion. To enable effects such as static force to be simulated the control Eq. (13) can be modified by including a bias force matrix F_{bias} as follows:

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

where F_{bias} is a 6×1 bias force/torque matrix, which can be used to for simulation of special effects like virtual weight lifting.

6. Preliminary Experiments/Results

6.1. Joint Range of Motion

The first series of test conducted on the rehabilitation exoskeleton involved consideration and measurement of the work volume of the system. The motion range of each of the joints of the exoskeleton is shown in Table 1. The first column gives the typical range of motion of the human arm joints (Harold et al., 1972). The second column presents the limits of motion of each joint of the mechanical structure itself. Finally, the third column presents the range of motion of each joint when the user's arm is attached to the rehabilitation trainer.

As can be seen in Table 1 the ranges of joint motions of the arm exoskeleton largely corresponds to the range achieved by the human arm in normal motion. Areas were the correlation is less than perfect include shoulder abduction/adduction and flexion extension where the range of motion is narrower than for the corresponding human arm motions.

Table 1. Motion ranges for the training/rehabilitation arm.

Arm motion	Human arm	Training arm	Training + Human arm
Wrist flexion	90	70	70
Wrist extension	99	70	70
Wrist adduction	27	45	30
Wrist abduction	47	45	45
Forearm supination	113	45	45
Forearm pronation	77	40	40
Elbow flexion	142	100	100
Shoulder flexion	188	110	110
Shoulder extension	61	25	25
Shoulder adduction	48	20	20
Shoulder abduction	134	100	100
Shoulder medial rotation	97	48	48
Shoulder lateral rotation	34	46	46

In addition, 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 training shoulder. Finally, shoulder medial rotation and the forearm rotations have also a limited range of motion compared to the human arm motions. 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. Upgrades to improve these features are already in development.

6.2. *Joint Output Torque*

The second area of performance evaluation was the output torque that the system could simulate or restrain. The torque output performance for each joint of the rehabilitation trainer/exoskeleton is shown in Table 2. The first column gives the typical human isometric strength for each of the arm joints (An et al., 1986).

The above torque values are the minimum (worst case) output torques. The 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 (pMA) which exhibit spring characteristics exerting higher forces for longer dilated lengths.

6.3. *Shoulder Strengthening with Weight Training*

To show the utility of the rehabilitation trainer/exoskeleton a series of test scenarios have been developed based on standard physiotherapy treatment

Table 2. Joint output torque for the training/rehabilitation arm.

Joint	Human isometric strength	Achieved torque
Shoulder		
Flexion/extension	110 Nm	30 Nm
Adduction/abduction	125 Nm	27 Nm
Rotation	–	6 Nm
Elbow		
Flexion/extension	72.5 Nm	6 Nm
Supination/pronation	9.1 Nm	5 Nm
Wrist		
Flexion/extension	19.8 Nm	4 Nm
Adduction/abduction	20.8 Nm	4 Nm

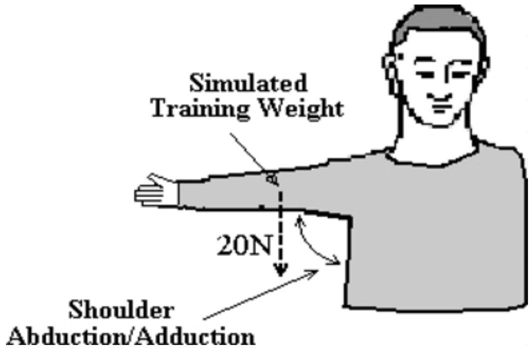

Figure 6. Shoulder training experiment.

regimes. The shoulder is considered to be one of the most complex joints of the human body, but it is also one of the most vulnerable to injury. This complexity of movement makes the shoulder joint distinctive from a training and rehabilitation perspective. One group of rehabilitation exercises often used for shoulder training or treatment after injury is based on consistent repetitive motions using small weights.

In this experiment the training exoskeleton was configured to simulate the forces generated by a virtual constant load located at the elbow joint. The arm exoskeleton was securely attached to the operator's arm and the control matrices were set up as follows to simulate a 2 kg load, Fig. 6.

$$B_E = \begin{bmatrix} 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \end{bmatrix}$$

$$K_E = \begin{bmatrix} 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 \end{bmatrix} \quad F_{\text{bias}} = \begin{bmatrix} -20 \\ 0 \\ 0 \\ 0 \\ 0 \\ 0 \end{bmatrix}$$

Since the load is located at the elbow frame the Jacobian up to the elbow was used to resolve the load into shoulder joint torques. During the experiment the operators repeated a shoulder abduction/adduction exercise as shown, Fig. 6. During these motions the position and

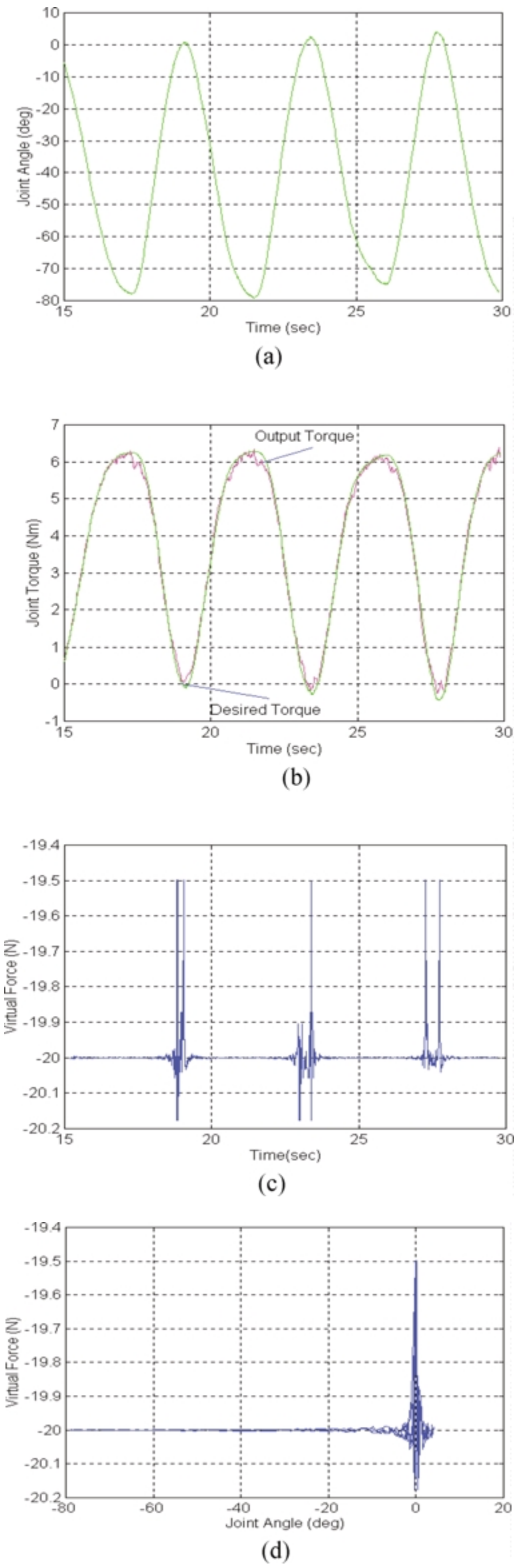

Figure 7. Shoulder training experimental results.

the output torque of the shoulder abduction/adduction joint torque were recorded.

The results are introduced in Fig. 7(a) show the abduction/adduction motions for a typical test subject. Graph (b) introduces the desired and the output torque of the shoulder abduction/adduction joint as recorded during the experiment. To give a more clear idea of the system performance the next graphs (c), and (d) illustrate the output load as a function of the time and the joint position. The output load was calculated using the output torque of the joint. The maximum external load error is less than 2.5% for the whole motion range. In terms of the actual sensation the test subjects (10 male subjects aged 22–35) reported that the sensation of the external load is very close to the natural sensation, giving very encouraging feedback about the possibility of using the system as a training/rehabilitation device and applying this to a more extensive range of physiotherapy training regimes in addition to applications as a power assist system.

7. Conclusions/Further Work

An upper limb multipurpose device was proposed. The mechanical and actuation system was described. The device has 7 DOF corresponding to the natural motion of the human arm from the shoulder to the wrist and is constructed primarily from aluminium and composites for low mass <2 kg. The achievable work volume and joint torque outputs have good correlation with the values for human subjects. Particular attention is applied to the use of 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 power device is in close proximity to a user. This is particularly apt in medical applications were the users, due to their conditions, may be more a risk than a fully healthy operator in an industrial type environment. It has been shown that this design concept can lead to an inherently safer system due to defined limits set on actuator contraction (35% maximum).

In addition, antagonistic action permits compliance control. This has advantages in terms of safety and more human ‘soft’ interaction providing a facility that is reminiscent of the compliance controlled feel of human manipulation. It has been shown that the device can be used as:

(i) an exercise facility for the joints of the upper limb,

(ii) a rehabilitation/power assist orthosis or as a joint power for those with loss/reduced power in the limb and
(iii) a motion analysis system.

Along with its above functionality the system was designed to meet the specifications of lightness, gravity compensation, ease of fitting and adjustment and relatively low mechanical complexity, which are essential for any system that is in direct contact with the human operator. Finally a training control scheme was presented and results were introduced to successfully demonstrate the capability of the system as a physiotherapy training facility.

Further studies will include the following:

(i) Objective evaluation of the effects of the use of the exoskeleton system by analysing the EMG signals of the muscles during different experimental conditions.
(ii) The assessment of the system effectiveness as a power assist device, for people with weak upper limb muscles.
(iii) The use of the system to provide power assistance during the execution of simple tasks such as load lifting or tasks that require repetitive stress motions. This will include the understanding of the power requirements of these tasks and the development of effective force amplification control strategies that will enable the execution of the physical activities.
(iv) The development of software to demonstrate the use of the system as a motion and power analysis system.
(v) Further enhancements and develops of the physiotherapy training regimes to form a library of treatment procedures.

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