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5 **“SOFT” EXOSKELETONS FOR UPPER AND LOWER BODY
 REHABILITATION — DESIGN, CONTROL AND TESTING**

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21 The basic concepts for exoskeletal systems have been suggested for some time with
 23 applications ranging from construction, manufacturing and mining to rescue and emer-
 gency services. In recent years, research has been driven by possible uses in medi-
 cal/rehabilitation and military applications. Yet there are still significant barriers to
 25 the effective use and exploitation of this technology. Among the most pertinent of these
 factors is the power and actuation system and its impact of control, strength, speed and
 27 perhaps most critically, safety. This work describes the design, construction and test-
 ing of an ultra low-mass, full-body exoskeleton system having seven degrees of freedom
 (DOFs) for the upper limbs and five degrees of freedom (DOFs) for each of the lower
 29 limbs. This low mass is primarily due to the use of a new range of pneumatic muscle
 actuators as the power source for the system. The work presented will show how the
 system takes advantage of the inherent controllable compliance to produce a unit that
 33 is powerful, providing a wide range of functionality (motion and forces over an extended
 range) in a manner that has high safety integrity for the user. The general layout of
 35 both the upper and the lower body exoskeleton is presented together with results from
 preliminary experiments to demonstrate the potential of the device in limb retraining,
 37 rehabilitation and power assist (augmentation) operations.

Keywords: Exoskeleton; soft actuation; rehabilitation.

39 **1. Introduction**

41 Biological exoskeletons are common structures in both insects and crustaceans,
 where they form a hard outer casing that provides support or protection for the
 organisms allowing them to perform physical activities that, on a size-to-load basis,

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1 could not be attempted by creatures with endoskeletons, e.g. humans. With robotic
2 exoskeletons, the aim is to provide a structure that permits humans to make use of
3 the concepts and advantages of an external skeleton and actuation to enhance power,
4 endurance, or peak strength while retaining controlling human intelligence. These
5 systems have the capacity to combine decision-making capabilities with machine
6 dexterity and power, to greatly augment a person's physical abilities.

7 Yet, exoskeleton technology represents one of the most challenging areas of
8 robotics research requiring significant advances in materials, mechanisms, electron-
9 ics, sensors, controls, intelligence, communication, power sources and actuation —
10 technology improvements that must be integrated together.^{1,2} Further, the funda-
11 mental criteria for an exoskeleton, i.e. a mechanism that wraps around the operator's
12 limbs allowing the replication or enhancement of forces at body segments, means
13 that the human and the mechanical systems (robots) are inherently coupled and
14 safety is paramount. It is therefore essential to provide drive systems that combine
15 the positive attributes of conventional actuator design with a "softer" safer inter-
16 action capacity. These are features not found in traditional exoskeleton actuation
17 (hydraulic or electric) but which are inherent in braided pneumatic muscle actua-
18 tors (PMAs).³ Using this format, exoskeletons can be developed for augmentation
19 and rehabilitation applications (in particular) that have capabilities beyond those
20 available with conventionally driven designs. This work will present the use of com-
21 pliance regulated and controlled pairs of antagonistic PMAs to drive a combined
22 whole-body exoskeleton incorporating both lower and upper body systems.

23 In this paper, we describe the construction and testing of a seven, degree of
24 freedom (DOF) upper arm exoskeleton system and a ten-DOF lower body system.
25 The total weight of the uncompensated upper arm system is less than 2 kg while
26 the lower body system is less than 12 kg. This low mass is primarily due to the use
27 of a new range of PMAs as the power source for the system. The need for safety,
28 simplicity and lightness is met by this type of actuator, which also has an excellent
29 power/weight ratio. The work presented will show how the system takes advantage
30 of the inherent controllable compliance to produce a unit that is extremely powerful,
31 providing a wide range of functionality (motion and forces over an extended range)
32 in a manner that has high safety integrity for the user (patient). The general layout
33 of the both upper and lower body exoskeleton is presented. This includes the design
34 requirements, and the design description. The control issues of the system are also
35 discussed. Initially, the low level joint control of the system is presented. A training
36 control scheme is introduced which is used to control the exoskeleton. Results from
37 preliminary experiments demonstrate the potential of the device in limb retraining,
38 rehabilitation and power assist (exoskeleton) operations.

39 **2. Historical Development**

40 Some of the first exoskeletons (both fictional and real) were suggested and built in
41 the 1960s and often had a military dominated theme. The hydraulically powered

1 Green-man exoskeleton was designed for the US Army for performance augmen-
2 tion. Designed as a forklift truck replacement, but with greater flexibility, mobility
3 and dexterity, it was beset by many problems with the mass of the power pack and
4 the hydraulics and was never deployed. In 1965, General Electric developed Hardi-
5 man, a 700 Kg, 30-jointed system powered by both hydraulics and electric motors.
6 Again, this system was too large and heavy with severe stability problems.²

7 These early efforts suffered from fundamental technological difficulties, such as:
8 computation speeds which were insufficient to provide the control functions neces-
9 sary to give smooth and effective tracking of the wearer’s input movements; energy
10 supplies that were not compact and light enough to be easily portable; actuators
11 that were too sluggish, heavy, and bulky.⁴

12 There was a significant loss of momentum after these set-backs and research in
13 large-scale exoskeletons was largely forgotten, apart from the work of Kazerooni in
14 force extenders, until the late 1980s and early 1990s when developments in virtual
15 reality and telepresence prompted renewed developments looking at both portable
16 and fixed base exoskeletons. Military requirements were again to the fore but in this
17 instance the focus was on upper limb systems as opposed to legged or whole-body
18 units.^{5,6} The EXOS ArmMaster had five active DC powered DOFs and additional
19 passive DOFs to increase comfort and permit size adjustment.² Output torques
20 ranged from 0.38 Nm to 6.3 Nm with a total mass of approximately 10 kg.

21 The Force REFlecting EXoskeleton (FREFLEX) is a seven-DOF DC motor
22 powered exoskeleton with a workspace of approximately 125 cm × 100 cm × 75 cm
23 and an output force of at least 25 N. The FREEFLEX is a non-portable system.⁷
24 The hydraulically powered, seven-DOF Sarcos Arm Master developed for subsea
25 teleoperation is one of the most powerful of all the exoskeletons. In addition to
26 the seven-DOF in the arm, there are three additional DOFs to allow control of
27 the thumb and one finger. The actuators provide output torques that range from
28 97.7 Nm for the shoulder, to 5.5 Nm for the hand joints. Due to the power of this
29 system, safety mechanisms were incorporated, including mechanical joint limits,
30 dead man switches and “watch dog” routines.⁸ The four-DOF SMU exoskele-
31 ton has three DOFs at the shoulder and one DOF at the elbow. Force feed-
32 back is generated by means of pneumatic cylinders. To support its weight, active
33 gravity compensation is used, which is controlled from a neural network based
34 controller.^{2,7}

35 Within Europe, the GLAD-IN-ART exoskeleton is a seven-DOF DC motor pow-
36 ered exoskeleton having a configuration broadly similar to the EXOS with one extra
37 DOF for forearm pronation/supination and two DOFs at the wrist. The torque out-
38 put ranges from 20 Nm for the shoulder to 10 Nm for the upper arm rotation and
39 elbow flexion extension joint and 2 Nm for the forearm joints. The unit again weighs
40 10 kg but has active gravity compensation to offset this loading.⁹ Other notable
41 systems include the KIST exoskeleton with nine DOFs (seven rotational and two
42 prismatic joints) powered by pneumatic cylinders¹⁰ and the seven-DOF DC motor
43 powered Sensor Arm II exoskeleton developed at the University of Tokyo.¹¹

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1 In recent years, there has once again been renewed military interest in exoskele-
2 tons with DARPA funding a wearable robotic system to help soldiers carry heavier
3 loads. Two programs in particular have addressed this problem. Bleex 1 and 2 have
4 two electromechanical legs that strap to the outside of the wearer's legs. These
5 systems have good user maneuverability, permitting users to walk, squat, twist,
6 kneel and run at speeds exceeding 2 m/s while carrying a payload of 45 kg. Bleex
7 2 is a compact and lightweight (14 kg) hydraulically powered design.¹² The Sarcos
8 robotic system is reported to be able to carry 84 kg with no user perception of load.
9 The system permits walking and running and can react to disturbance inputs, e.g.
10 stumbling. The exoskeleton uses a special internal-combustion engine that can use a
11 variety of fuels and deliver enough hydraulic power to meet the strength and speed
12 requirements of the robotic limbs.¹³

13 In non-military applications, one of the most important systems is the HAL
14 (short for Hybrid Assistive Limb) full-body suit designed to aid people who have
15 degenerated muscles or those paralyzed by brain or spinal injuries. HAL-5 is con-
16 structed from a nickel molybdenum and extra-super-duralumin frame, strengthened
17 by plastic casing. The metal frame is strapped to the body and supports the wearer
18 externally, its several electric motors acting as the suit's muscles to provide powered
19 assistance to the wearer's limbs. The exoskeleton is powered by both nickel-metal
20 hydride and lithium battery packs giving almost three hours continuous operation.
21 HAL-5 weighs about 21 kg.¹³

22 From a medical rehabilitation perspective there have been a number of devel-
23 opments both for lower and upper body. Much of the early pioneering work was
24 developed by Vukobratovic.⁴ About the same time, the Balanced Forearm Orthosis
25 (BFO) was developed as a wheelchair mounted passive device to enable a person
26 with weak musculature to move their arms in the horizontal planes.¹⁴ In 1975, the
27 Burke rehabilitation centre developed a five-DOF version of the BFO powered by
28 means of electric motors, but this never gained significant acceptance.¹⁵ The Hybrid
29 Arm Orthosis (HAO), developed by the authors of Ref. 16, aimed to provide upper
30 arm motion assistance. This system offered shoulder abduction and elbow flexion,
31 wrist supination and a three joint jaw chunk pinch. The shoulder and the elbow
32 joints were body powered while the wrist supination and the three-point jaw chuck
33 pinch power were generated by two separate DC motors. More recently, a system
34 developed at MEL¹⁷ used a parallel mechanism to suspend the upper arm at the
35 elbow and wrist level. Each point was suspended by an overhanging plate using
36 three strings arranged in parallel.

37 Among the most interesting of the powered orthosis systems is a motorized
38 upper limb orthosis system (MULOS) developed at the University of Newcastle in
39 mid-90s.^{18,19} The system has five DOFs and is designed to work in three different
40 modalities: Assistive, Continuous Passive Motion (CPM) and Exercise. This device
41 appeared to have good potential but development was discontinued in 1997.

42 MIT MANUS is another robotic system that was developed for the physical
43 therapy of stroke victims. The system supports up to five DOF through two separate

1 modules. Although it is claimed to be portable, its weight is around 45 kg.^{20–21} The
2 system developed by the authors of Ref. 22 was also designed to provide upper limb
3 motion assistance. This system provides eight DOFs supporting upper limb and
4 hand. Special care was taken in the design of the system to ensure user safety.

5 The ARM Guide^{23,24} consists of an instrumented linear constraint that can be
6 oriented in different directions across the subject's workspace using a three-spline
7 steel shaft. This device was primarily used as a measurement tool for force and
8 motion during mechanically guided movement. The most recent system of all is the
9 GENTLE project, which uses a haptic master device to provide exercise for an arm
10 using an interactive virtual environment. The system has six DOFs, but only the
11 three translational motions are active.²⁵

3. Design Requirements

13 Based on the knowledge of the state of the art as well as user needs for safe,
14 compliant powered actuation for both rehabilitation and human performance aug-
15 mentation applications, fundamental technical specifications could be developed.²⁶
16 These requirements focused on:

17 (i) *A structure having low mass/inertia.* The mass of the exoskeleton must be
18 minimized, to enhance user comfort, portability and to reduce energy wastage in
19 simply moving the structure. Low mass and inertia also have secondary benefits
20 when considering the safety aspects of what is in essence a robot physically attached
21 to the human.

22 (ii) *Safety.* As the system is in direct contact with the human operator the safety
23 requirement is paramount. Safety is enhanced through low mass/inertia structures
24 that combine high strength with actuator and/or structural compliance. The device
25 must be safe technically and perceived as safe by potential users.

26 (iii) *Comfort of wearing.* As the extended use of the device is certainly possible
27 and probably necessary, the device must be comfortable, causing no fatigue to the
28 operator even after long periods, e.g. 1–2 hours of operation. This requirement
29 should include ease of fitting adjustment and removal.

30 (iv) *Extensive range of motion.* A generic specification for the display range of
31 motion can be defined as the "typical" human workspace.

32 (v) *Accurate force feedback.* Accurate representation of the forces means that the
33 device must have sufficient force resolution capabilities. The human arm force res-
34 olution capability for the different arm joints has been studied. Experiments from
35 Ref. 27 revealed that the average force resolution is around 0.36 N and tended to
36 increase as the target force increases. Using this as a specification value will ensure
37 that the operator feels no force discontinuities during motion.

38 (vi) *Good motion sensing resolution.* The motion sensing requirements of the device
39 obviously depend on the position resolution capabilities of the human. Since the

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1 human position sensing resolution varies for different joints in the human body, the
2 motion sensing resolution will be determined by the sensing capabilities of the part
3 of the body that the system is attached to. These values are typically in the range
4 0.8° – 2.0° .^{27,28}

5 (vii) *Accurate automatic compensation for gravity forces.* Since medical applica-
6 tions will involve individuals with at best weakened arm structures, active, easily
7 updated gravity compensation forms a keystone of the design.

8 (viii) *Reliability.* As with all systems, user acceptance is dependent to a large extent
9 on the reliability and utility of the mechanism. It is therefore vital that appropriate
10 design concerns are given to reliability in all operations and in environments where
11 materials like water, dust or grease are presented.

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

15 **4. Mechanical Design**

16 The mechanical structure is formed in two major portions, an upper body exoskele-
17 ton to provide power to the arm and a lower body exoskeleton to provide augmen-
18 tation to the legs during walking activities. In both instances, the primary design
19 goal has been for aspects of rehabilitation. For the upper body, the scenario is
20 rehabilitation after strokes while for the lower body paralysis, muscle wastage and
21 stroke are the main targets. At the same time, the structure is capable of perform-
22 ing the general aspects of human performance augmentation. A hand exoskeleton
23 has also been developed and is shown in later figures but is not described in this
24 paper.²⁹

25 **4.1. Upper body**

26 The mechanical arm structure has seven DOFs corresponding to the natural motion
27 of the human arm from the shoulder to the wrist but excluding the hand. The
28 structure which is constructed primarily from aluminium and composite materials,
29 with high stress joint sections fabricated in steel has three DOFs in the shoulder
30 (flexion/extension, abduction-adduction and lateral-medial rotation), two DOFs at
31 the elbow permitting flexion/extension, pronation/supination of the forearm, and
32 two DOFs at the wrist (flexion-extension and abduction-adduction); Fig. 1.

33 The arm is constructed for use by a “typical adult” with only minor changes to
34 the set-up. Arm link length changes can easily and quickly be effected, if necessary
35 making it easy to accommodate a range of users, which is an important aspect of
36 the design. The arm is attached to the user arm at the elbow and wrist level using
37 two wide velcro bands. This was found to be adequate during operation, providing
38 a secure attachment together with easiness of system donning and removal.³⁰ High

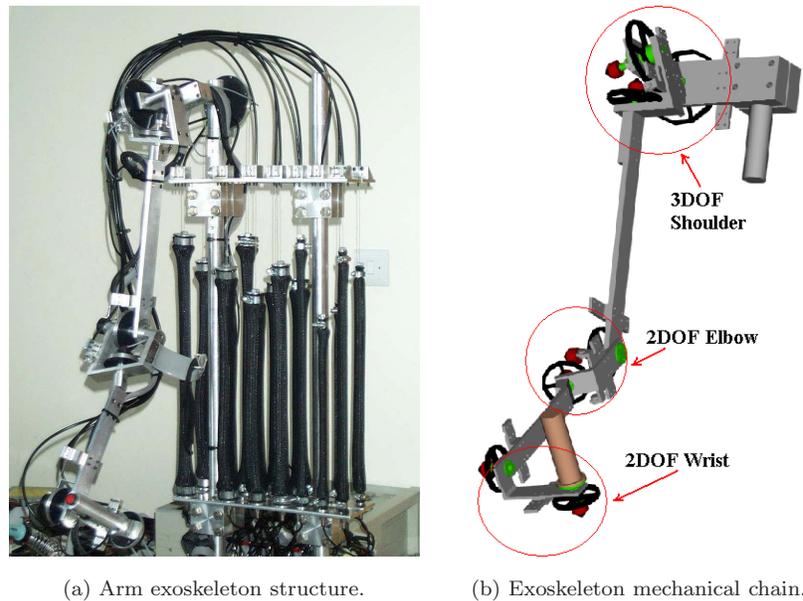


Fig. 1. Upper limb mechanical structures.

1 linearity sensors provide position sensing on each joint, with joint torque sensing
 2 achieved by integrating two strain gauges (mounted on internal spokes) inside each
 3 joint pulley. Ranges of motions of the arm exoskeleton and forces generated are
 shown in Table 1.³¹

5 4.2. Lower body

7 The design concepts and principles of the upper body that have been adopted were
 reasonable in the lower body design. The mechanical structure, which is once again
 9 constructed primarily from aluminium with joint sections fabricated in steel, con-
 sists of a ten-DOF (five DOFs per leg) mechanism corresponding to the fundamental
 11 natural motion and range of the human legs from the hip to the ankle, but excluding
 lower leg rotation and ankle flexion/extension.

13 The hip structure has three DOFs as in the shoulder (flexion/extension,
 abduction-adduction and lateral-medial rotation), one DOF at the knee permit-
 15 ting flexion/extension, and one DOF at ankle (dorsi/plantar flexion); Fig. 2. The
 leg is mounted in a moulded lower body brace which is light, low cost and comfort-
 17 able, designed for ease of change of leg link length. The overall mass of the system
 is less than 12 kg.³² As with electrical systems, it must be recognized that this mass
 19 does not include the power source. Ranges of motions of the lower body exoskele-
 ton and forces generated are shown in Table 1. The total weight of the exoskeleton
 consisting of the both legs and a rigid spine is 12 kg assembled with a length of

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

Motion	Human Arm (°)	Exo + Human Arm (°)	Human Isometric Strength	Achieved Torque Peak (Min)	Leg Motion	Range (°)	Force
Wrist flexion	90	70	19.8 Nm	28 Nm (5 Nm)	Plantar Flexion	50	60 Nm
Wrist extension	99	70			Dorsi flexion	30	
Wrist adduction	27	30	20.8 Nm	24 Nm			
Wrist abduction	47	45		(7 Nm)			
Forearm supination	113	45	9.1 Nm	25 Nm			
Forearm pronation	77	40		(9 Nm)			
Elbow flexion	142	100	72.5 Nm	76 Nm (19 Nm)	Knee	140	60 Nm
Shoulder flexion	188	110	110 Nm	95 Nm (30 Nm)	Hip flexion	120	60 Nm
Shoulder extension	61	25			Hip extension	20	
Shoulder adduction	48	20	125 Nm	128 Nm	Hip adduction	30	65 Nm
Shoulder abduction	134	100		(27 Nm)	Hip abduction	45	
Shoulder medial rotation	97	48		46 Nm (5 Nm)	Hip int. rot.	45	
Shoulder lateral rotation	34	46			Hip ext. rot.	50	

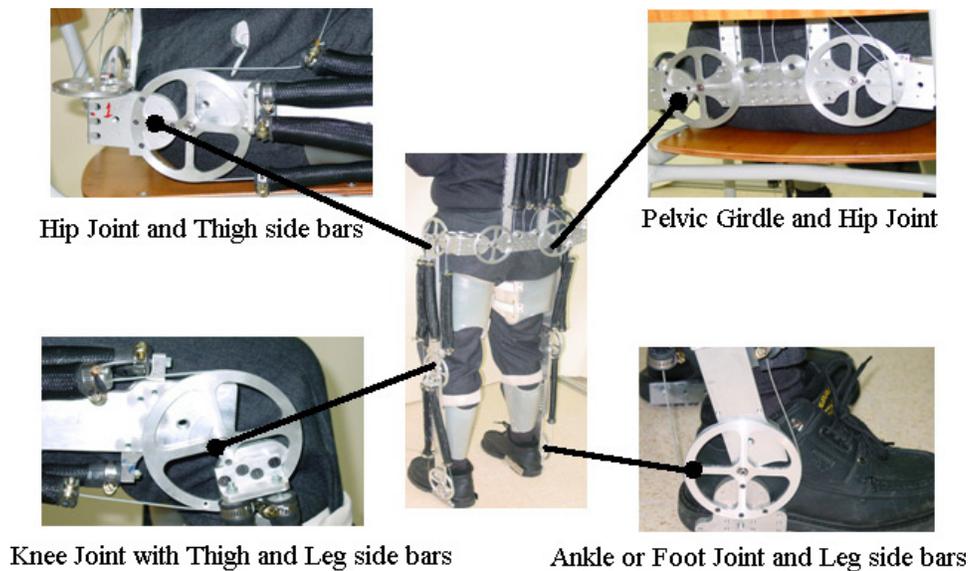


Fig. 2. Lower limb exoskeleton mechanical structure.

1 520 mm from the hip to the knee while the length from the ankle’s base to the top
2 of the knee is 500 mm.

3 **4.3. Actuator design**

4 As identified in the failings of the early Hardiman and Greenman systems, actua-
5 tors and actuation systems are essential, perhaps defining, parts of all exoskeleton
6 structures, providing the forces, torques and mechanical motions needed to move
7 the joints, limbs or body. Their performance is usually characterized by parame-
8 ters such as power (particularly power/weight and power/volume ratios), strength,
9 response rate, physical size, speed of motion, reliability, controllability, compliance
10 and cost. The nature of the drive source proposed in this work forms a key sub-
11 system making use of “soft” compliant actuator technology that permits a change in
12 the operational paradigm for the design and use of the exoskeletons. This is funda-
13 mentally different from methods previously developed and is a key to the success of
14 this technique. This system uses braided pneumatic muscle actuators (PMAs) that
15 provide a clean, low cost actuation source with a high power-to-weight ratio and
16 safety due to the inherent compliance. These PMAs are constructed as a two-layered
17 cylinder; Fig. 3.

18 This design has an inner rubber liner, an outer containment layer of braided
19 nylon and endcaps that seal the open ends of the muscle. Within the actuator,
20 pressure sensors, have been incorporated to monitor the internal state of the muscle,
21 while miniature strain gauge based load cells can be used to directly measure the
22 force in any actuation system. The complete unit can safely withstand pressures
23 up to 700 kPa (7 bar), although 600 kPa (6 bar) is the operating pressure for this
24 system. The detailed construction, operation, and mathematical analysis of these
25 actuators can be found in Refs. 33 and 34. The structure of the muscles gives the
26 actuator a number of desirable characteristics:

- 27 (i) Muscles can be produced in a range of lengths and diameters with larger sizes
28 producing larger contractile force.
- 29 (ii) The actuators have exceptionally high power and force to weight/volume
30 ratios > 1 kW/kg.
- 31 (iii) The actual achievable displacement (contraction) is dependent on the con-
32 struction and loading but is typical 30%–35% of the dilated length — com-
33 parable with the contraction achievable with natural muscle.

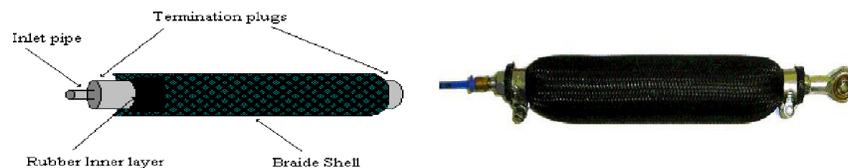


Fig. 3. Pneumatic muscle actuator design.

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- 1 (iv) “Soft” construction and finite maximum contraction make PMAs safe for
human–machine interaction.
- 3 (v) Controllers developed for the muscle systems have shown them to be control-
lable to an accuracy of better than 1% of displacement.
- 5 (vi) Bandwidth for antagonistic pairs of muscles of up to 5 Hz can be achieved.
- 7 (vii) Force control using antagonistic pairs for compliance regulation is possible as
for natural muscle.
- 9 (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 $20\text{--}40\text{ N/cm}^2$ for natural muscle.
- 11 (ix) The actuators can operate safely in aquatic, dusty or other liquid environ-
ments.
- 13 (x) The actuators are highly tolerant of mechanical (rotational and translational)
misalignment, reducing the engineering complexity and cost.

15 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,
17 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
19 tailored to this application.

4.4. Actuator attachments

21 Joint motion/torque is achieved by producing appropriate antagonistic torques
through cables and pulleys driven by the pneumatic actuators. Since the PMA
23 is a single direction-acting element (contraction only), this means that for bidi-
rectional motion/force two elements are needed. These two acting elements work
25 together in an antagonistic scheme simulating a biceps-triceps system to provide
the bidirectional motion/force; Fig. 4.

27 In the above set-up, L_{\min} denotes the length of the muscle when it is fully
contracted; $L_0 = L_{\min} + \frac{L_{\max} - L_{\min}}{2}$ is the initial dilated length of muscle, which
29 is equal to half of maximum muscle displacement needed to maximize the range
of motion of the joint; L_{\max} is the maximum dilated length; r is the radius of the
31 pulley; and P_1, P_2 are the gauge pressures inside the two muscles.

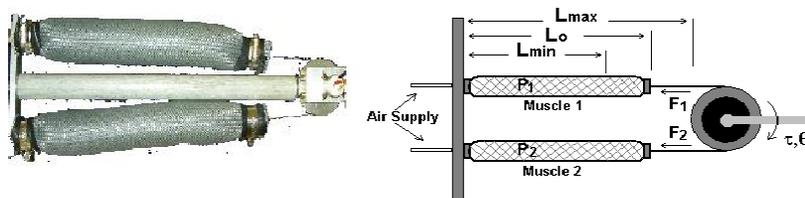


Fig. 4. Antagonistic pairs of muscles.

1 Flexible steel cables are used for the coupling between the muscles and the pulley.
2 Since most of the joints require a range of rotation in excess of 90° , double groove
3 pulleys have been employed. The pulleys have been made from solid aluminium
4 pieces internally machined to form a four spoke structure. On each of these internal
5 spokes, strain gauges are mounted to form a joint torque sensor. The compact
6 actuator structure allows for integration close to their respective powered joints.
7 This makes the overall design compact, in line with the design requirements. The
8 muscles used in the upper limb have a diameter of 2 cm–4 cm with an “at rest”
9 length of 15–45 cm.

10 For the lower limb systems, the ankle actuators (two actuators) and the lower
11 leg flexion/extension actuator (two actuators) are mounted on the side. The knee
12 actuators (two actuators) and the leg hip rotation actuators (two actuators) are
13 mounted on the thigh side while the hip actuators (four actuators) are mounted on
14 the body brace behind the operator’s back. The muscles used in this project have
15 a diameter of 2 cm, with an “at rest” length of 50–70 cm.

4.5. *System controller and user interface*

17 The activation of the PMA is reliant on the effective control of the airflow into
18 and from the muscles. This is controlled by MATRIX valves that incorporate four
19 3/3 controllable ports in a package having dimensions of 45 mm \times 55 mm \times 55 mm
20 and weighing less than 320 g. The valves can be driven and controlled at up to
21 400 Hz PWM signal providing rapid, smooth motion. Development of an adaptive
22 controller and details of the design can be found in Ref. 33. By incorporating a
23 pressure sensor into the valve inlet, closed loop pressure control is also possible.
24 Pulsing of the valves along with data collection from the position, pressure and
25 torque sensors is controlled from local dedicated microcontrollers with I/O, ADC
26 and communication port facilities. The external PC is used to supervise the working
27 conditions of the prototype.

28 Specifically, each individual muscle pairs (joint) are controlled by a local DSP
29 which is again designed to mate with the valve assembly for compact operation.
30 Each MCU controls up to eight PMAs (eight inlets + eight outlets). The DSPs
31 are connected through a data bus to the interface PC, running Windows based
32 monitoring software.

33 5. Modeling and Control

5.1. *Modeling of the pneumatic muscle actuated joint*

To make effective use of the “soft” properties of the PMAs, accurate modeling
and control is essential and extensive work has sought to achieve this. Two main
approaches to the force generation problem for the PMAs have been developed by
researchers, with the first based on energy modeling,³⁵ and the second using force
profiles of the surface pressure.^{33,34} Both methods provide the same base model

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described by the following equation:

$$F = \frac{\pi D_0^2 P}{4} (3 \cos^2(\theta) - 1), \quad (1)$$

where θ is the braid interweave angle, D_0 is the maximum theoretical muscle diameter at a braid interweave angle of $\theta = 90^\circ$ and P is the operating pressure. Modeling of the joint dynamics based on a simplified muscle model derived from Eq. (1) considers the PMAs as a nonlinear spring with an elastic constant that is related to the muscle properties. Equation (1) can be written as a function of the muscle length L and the length of one thread of the braided structure b as

$$F = \frac{\pi D_0^2 P (3L^2 - b^2)}{4b^2}. \quad (2)$$

Simulated results using the above equation and experimental data (Fig. 5) for different sizes of muscles suggest that the pneumatic muscle behaves like a pressure dependent variable compliance spring. Based on this observation and considering constant pressure due to the small length change dL , the muscle spring stiffness per unit length can be computed from Eq. (2):

$$K = \frac{dF}{dL} = \frac{6\pi D_0^2 PL}{4b^2}. \quad (3)$$

As can be seen from Eq. (3) the stiffness of the muscle is a function of the operating pressure P and the length of the muscle L . We can now calculate the stiffness per unit pressure K_{pr} :

$$K_{pr} = \frac{dK}{dP} = \frac{6\pi D_0^2 L}{4b^2} = K_{gas} L. \quad (4)$$

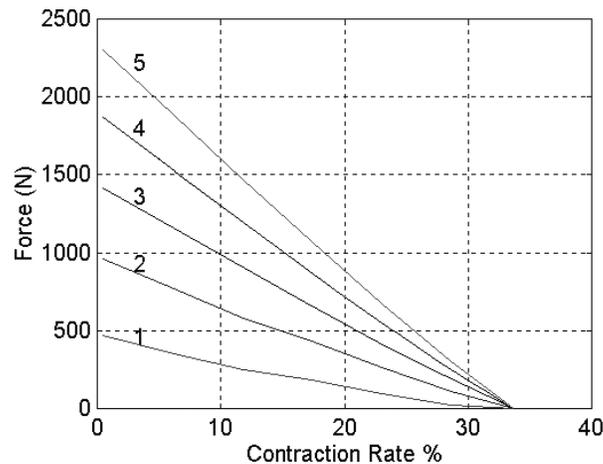


Fig. 5. Experimental force/displacement profiles.

The muscle force equation can now be rewritten to replicate a spring force equation:

$$F = K_{\text{pr}}P(L - L_{\text{min}}), \quad (5)$$

where L_{min} is the length of the muscle when it is fully contracted. Equation (5) represents the muscle as a nonlinear quadratic spring. The muscle force equation in the form presented above is used for the modeling of the pneumatic muscle actuated joints of the compliant arm haptic system. Based on the static force equation of the PMA, the state space model of the joint powered by two antagonistic muscles can be computed. The equation that describes the motion of the joint is given by

$$J\ddot{\theta} + D\dot{\theta} = \tau, \quad (6)$$

where J , D and τ denote the joint inertia, damping and torque, respectively. Considering now the antagonistic pair of muscles shown in Fig. 5 and using the muscle force equation (5), the forces developed are given by

$$\begin{aligned} F_1 &= K_{\text{pr}}P_1(L_0 - L_{\text{min}} + r\theta), \\ F_2 &= K_{\text{pr}}P_2(L_0 - L_{\text{min}} - r\theta). \end{aligned} \quad (7)$$

Equations (7) can be rewritten as

$$\begin{aligned} F_1 &= K_{\text{pr}}P_1(\Delta L_0 + r\theta) = K_{\text{gas}}P_1L_1\Delta L_1, \\ F_2 &= K_{\text{pr}}P_2(\Delta L_0 - r\theta) = K_{\text{gas}}P_2L_2\Delta L_2, \end{aligned} \quad (8)$$

where $\Delta L_0 = L_0 - L_{\text{min}} = \frac{L_{\text{max}} - L_{\text{min}}}{2}$ is the initial dilation for the muscles and $L_1 = L_{\text{min}} + \Delta L_1$, $L_2 = L_{\text{min}} + \Delta L_2$ are the lengths of the muscles at any joint positions. The dilations for the two muscles at any joint position are $\Delta L_1 = \Delta L_0 + r\theta$ and $\Delta L_2 = \Delta L_0 - r\theta$. Using (8), the torque developed at the joint can now be computed:

$$T = (F_2 - F_1)r = (K_{\text{gas}}P_2L_2\Delta L_2 - K_{\text{gas}}P_1L_1\Delta L_1)r. \quad (9)$$

The equation of motion of the single joint in Fig. 5 can be written as

$$\begin{aligned} \ddot{\theta} &= \frac{-D}{J}\dot{\theta} + \frac{K_{\text{gas}}r^3(P_2 - P_1)}{J}\theta^2 - \frac{K_{\text{gas}}r^2(P_2 + P_1)(2\Delta L_0 + L_{\text{min}})}{J}\theta \\ &+ \frac{K_{\text{gas}}r(P_2 - P_1)\Delta L_0(L_{\text{min}} + \Delta L_0)}{J}. \end{aligned} \quad (10)$$

By selecting the states of the system to be $x_1 = \theta$ and $x_2 = \dot{\theta}$, the state equations of the system can be formulated from (10) as

$$\begin{aligned} \begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \end{bmatrix} &= \begin{bmatrix} 0 & 1 \\ \frac{K_{\text{gas}}r^3(P_2 - P_1)}{J}x_1 - \frac{K_{\text{gas}}r^2(P_1 + P_2)L_{\text{max}}}{J} & -\frac{D}{J} \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \end{bmatrix} \\ &+ \begin{bmatrix} 0 \\ \frac{K_{\text{gas}}r(L_{\text{max}}^2 - L_{\text{min}}^2)}{4J} \end{bmatrix} [P_2 - P_1]. \end{aligned} \quad (11)$$

- 1 The above joint model describes the joint dynamics but since it is based on the static force equation of a PMA it does not incorporate the dynamics of the actuator air

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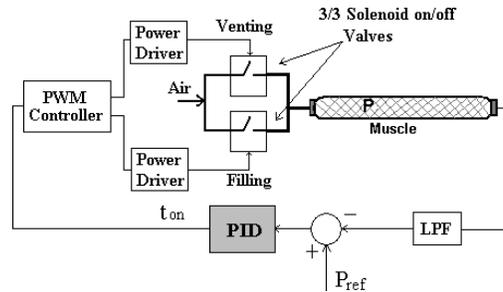


Fig. 6. Control scheme of the muscle pressure using on/off solenoid valves.

1 pressures P_1 and P_2 . In this system, the air pressure inside each of the two muscles
is regulated using on/off solenoid 3/3 valves; Fig. 6.

3 As can be seen from Fig. 8, a PID controller is used to calculate the on/off times
of the venting/filling valves and effectively regulates the pressure. The dynamics of
5 each of the two pressure regulators for P_1 and P_2 can be identified in MATLAB
using experimental input/output pressure data from each muscle and can be finally
7 incorporated into the control scheme.

5.2. Joint model verification

To verify the joint model developed in the previous section, the open loop step
response of one of the system's joints was recorded and the outcome of the model
(11) was compared with the experimental data. The air pressure was controlled
using pulse width modulation with a switching frequency up to 200 Hz and a fast
response time of less than 2 ms. The open loop step response of the system to
the input pressure variation $\Delta P = P_{\text{ref}_2} - P_{\text{ref}_1}$ was recorded. The two pressure
references ($P_{\text{ref}_1}, P_{\text{ref}_2}$) were set as

$$(P_{\text{ref}_1}, P_{\text{ref}_2}) = \left(\frac{P_{\text{max}} - \Delta P}{2}, \frac{P_{\text{max}} + \Delta P}{2} \right), \quad (12)$$

9 with $P_{\text{max}} = 400,000$ (Pa). This gives a balanced input pressure variation and
the system can be considered as a SISO system. The same generated $\Delta P \in$
11 $[-100000 \ 100000]$ (Pa) profile was used as input for the theoretical system model,
and the theoretical step response generated by MATLAB was plotted against the
13 experimental one in Fig. 7.

Figure 8 shows a good transient equivalence between the theoretical and the
15 experimental responses. A steady state error between the two responses can be
observed which is believed to be due to the non-modeled system parameters (e.g.
17 pressure dynamics, delays, friction, saturation, etc.).

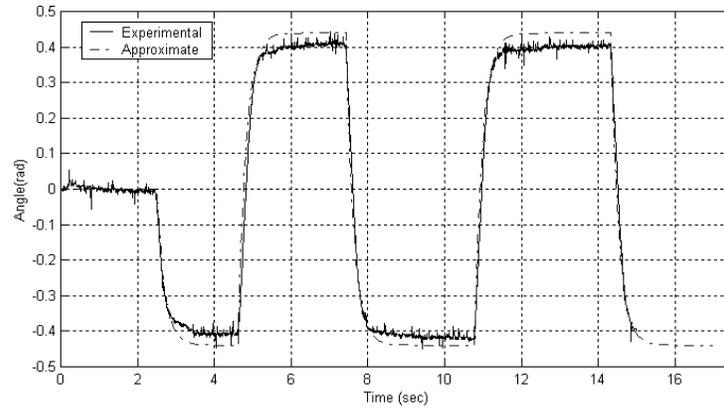


Fig. 7. Experimental and theoretical step response.

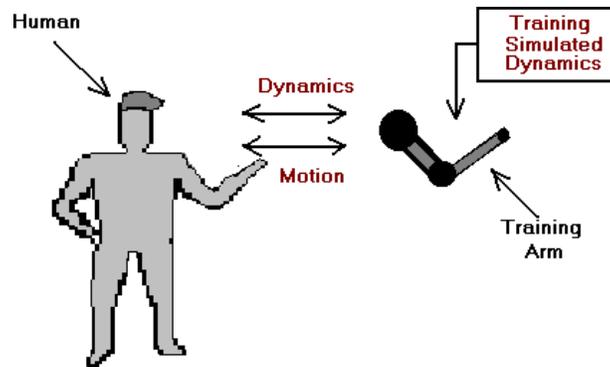


Fig. 8. Schematic diagram of the human and exoskeleton.

1 5.3. Upper limb exoskeleton impedance control scheme

When operating in an assistive/augmentation (or resistive for exercise) mode, an impedance control scheme is employed for the overall exoskeleton system to enable execution of complex assistive/resistive exercises. The following equation describes the dynamic behavior of the exoskeleton:

$$M(q) \cdot \ddot{q} + V(q, \dot{q}) + F(\dot{q}) + G(q) + J^T \cdot F_R = \tau_{\text{joint}}, \quad (13)$$

3 where q is the joint variable n-vector, τ_{joint} is the joint torque vector, $M(q)$ 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, and J^T is the transpose Jacobian of the exoskeleton.

5 The above equation can be used to describe the interaction between an operator and the exoskeleton. Considering the scenario described in Fig. 8, where the user's
7 limb is attached to the exoskeleton.

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Let F_H denote the force that the human exerts on the exoskeleton arm endtip (which is actually the force felt by the user), F_R is the force that the exoskeleton applies to the operator, and $Z_E(s)$ is the system's simulated mechanical impedance. To make the operator feel the simulated augmentation/resistive dynamics, the following equation must be applied:

$$Z_E(s) \cdot (x - x_E) = M_E \cdot \ddot{x} + B_E \cdot \dot{x} + K_E \cdot (x - x_E) = F_H, \quad (14)$$

where M_E , B_E , and K_E are the inertia, damping, and stiffness coefficients. The above equation defines the desired characteristics of the motion of the pair (**Operator, Exoskeleton**). Having specified the desired behavior of the system, the control law can now be derived by eliminating \ddot{x} and \ddot{q} from Eqs. (13) and (14). 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:

$$x = J \cdot \dot{q}, \quad (15)$$

$$\ddot{x} = J \cdot \ddot{q} + \dot{J} \cdot \dot{q}. \quad (16)$$

Solving Eqs. (14) and (16) for \ddot{x} and \ddot{q} , respectively, gives

$$\ddot{x} = M_E^{-1} \cdot (F_H - B_E \cdot \dot{x} - K_E \cdot (x - x_E)), \quad (17)$$

$$\ddot{q} = J^{-1} \cdot (\ddot{x} - \dot{J} \cdot \dot{q}). \quad (18)$$

Combining Eqs. (13), (17) and (18), \ddot{q} can be eliminated to give

$$\begin{aligned} M(q) \cdot J^{-1} \cdot (M_E^{-1} \cdot (F_H - B_E \cdot \dot{x} - K_E \cdot (x - x_E)) - \dot{J} \cdot \dot{q}) \\ + G(q) = \tau_{\text{joint}} - J^T \cdot F_R. \end{aligned} \quad (19)$$

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

$$M_E = J^{-1} \cdot M \cdot J^{-T}. \quad (20)$$

Considering slow motions typical in rehabilitation applications and that $F_R = -F_H$, Eq. (19) gives

$$\tau_{\text{joint}} = -J^T \cdot (B_E \cdot \dot{x} + K_E \cdot (x - x_E)) + G(q). \quad (21)$$

The above equation describes the impedance control law for the overall upper limb exoskeleton. The damping and the stiffness matrixes B_E and K_E are 6×6 diagonal matrices and depend on the dynamics to be modeled. To enable effects such as static force to be simulated, the control Eq. (21) can be modified by including a bias force matrix F_{bias} as follows:

$$\tau_{\text{joint}} = -J^T \cdot (B_E \cdot \dot{x} + K_E \cdot (x - x_E) + F_{\text{bias}}) + G(q), \quad (22)$$

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

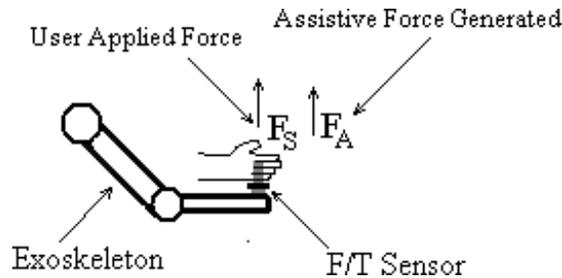


Fig. 9. Assistive mode setup for the exoskeleton arm.

1 5.4. Upper limb assistive impedance control

2 Under the assistive control mode, the exoskeletal system applies assistive force sig-
 3 nals dependent on the operator's desired motion. To enable detection of the user's
 4 desired motion, an F/T sensor is mounted at the distal section (endtip) of the
 5 exoskeleton. This sensor monitors the force signals applied by the user; Fig. 9. Based
 6 on these force signals, the desired position of the system x_E is updated in Eq. (21).

We use the following formula to derive the new desired position using the sensed
 force signal

$$x_E^i = x_{E-1}^i + \int x_f^i dt, i = 1, \dots, 6, \quad (23)$$

$$x_f^i = \left\{ \begin{array}{ll} k_a(F_s^i - a) & F_a^i > a \\ 0 & -a < F_a^i < a \\ k_a(F_s^i + a) & F_a^i < -a \end{array} \right\}, \quad (24)$$

7 where $x_E^i = [x_E^1 \dots x_E^6]^T$ is the desired position vector, $F_s^i = [F_s^1 \dots F_s^6]^T$ is
 8 the force vector from the F/T sensor, k_a is a sensitivity coefficient that can be
 9 adjusted according to the user's physical state, and a is the deadband parameter.
 10 By injecting the desired position vector derived from (23), (24) into (21) assistive
 11 forces towards augment the user desired actions/motions can be generated.

6. Experimental Results

13 A number of preliminary experiments were carried out to evaluate the performance
 14 of the system as a physiotherapy and augmentation device initially for the upper
 15 limb. The results are presented in the following sections.

6.1. Shoulder strengthening with weight training

17 The shoulder is considered to be one of the most complex joints of the human body,
 18 but it is also one of the most vulnerable to injury. This complexity of movement
 19 makes the shoulder joint distinctive from a training and rehabilitation perspective.

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1 One group of rehabilitation exercises often used for shoulder training or treatment
2 after injury is based on consistent repetitive motions using small weights. In this
3 experiment, the training exoskeleton was configured to simulate the forces generated
4 by a virtual constant load located at the elbow joint. The arm exoskeleton was
5 securely attached to the operator's arm and the control matrices were set up as
6 follows to simulate a 2 kg load; Fig. 10.

7 Since the load is located at the elbow frame, the Jacobian up to the elbow was
8 used to resolve the load into shoulder joint torques. During the experiment, the
9 operators repeated a shoulder abduction/adduction exercise as shown in Fig. 10.
10 During these motions, the position and the output torque of the shoulder abduc-
11 tion/adduction joint torque were recorded.

12 The results introduced in Fig. 11(a) show the abduction/adduction motions for
13 a typical test subject. Graph 11(b) introduces the desired and the output torque of
14 the shoulder abduction/adduction joint as recorded during the experiment.

15 Graphs 11(c) and 11(d) illustrate the output load as a function of the time and
16 the joint position. The maximum external load error is less than 2.5% for the whole
17 motion range. In terms of the actual sensation, the test subjects (ten male subjects
18 aged 22–35) reported that the sensation of the external load is very close to the
19 natural sensation, giving very encouraging feedback about the possibility of using
20 the system as a training/rehabilitation device and applying this to a more extensive
21 range of physiotherapy training regimes in addition to applications as a power assist
22 system.

23 **6.2. Assistive/augmentation mode experiment**

24 In this experiment, the assistive capacity was assessed by evaluating the perfor-
25 mance of the upper limb exoskeleton in providing assistive force signals in healthy
26 subjects. The device was securely attached to the operator's arm and it was con-
27 figured to work in assistive mode. A load of 2 Kg was attached at the exoskeleton

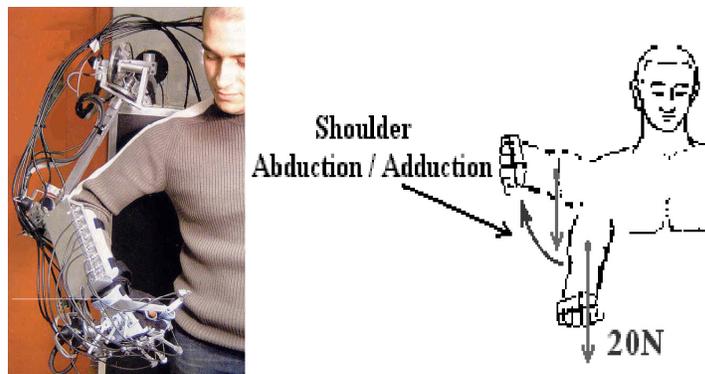


Fig. 10. Shoulder training experiment.

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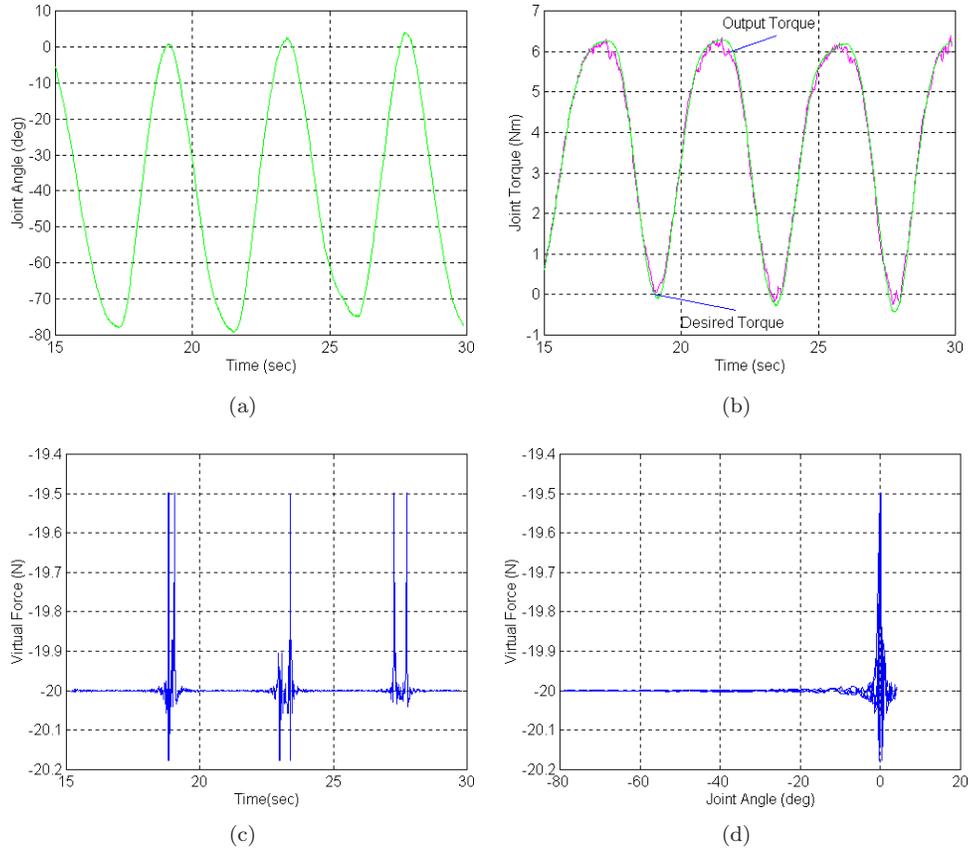


Fig. 11. Shoulder training experimental results.

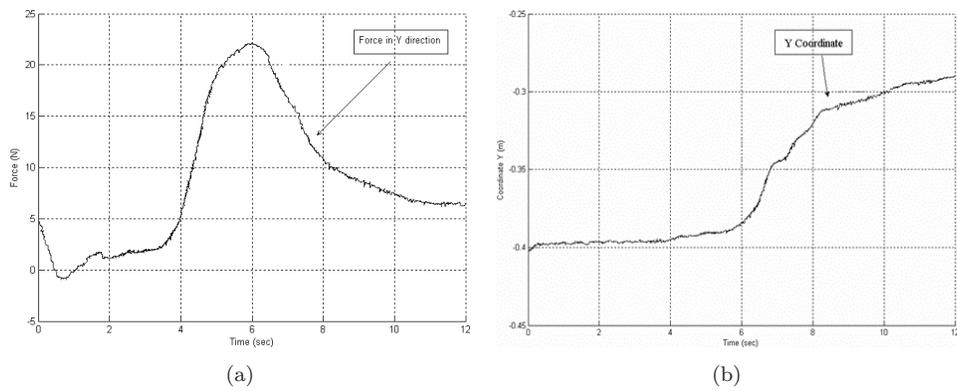


Fig. 12. (a) Force felt by the user, and (b) assistive motion response.

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1 endtip and the users were prompted to manipulate the exoskeletal device and try
2 to lift the attached load vertically. During this time, the force sensed at the F/T
3 sensor was recorded. Figure 12(a) shows the force profile sensed by the sensor,
4 which is actually the force felt by the user. The force initially increases as the user
5 starts trying to move the exoskeletal system in order to lift the load. The increase
6 force levels cause changes in the desired vertical y -coordinate [Fig. 12(b)] which
7 subsequently causes assistive forces governed by (22) to be generated at the tip of
8 the system. As a result, the force applied by the user [Fig. 12(a)] decreases. This
9 indicates the capability of the system and the proposed assistive control scheme to
10 provide assistive forces under user's motion control.

11 7. Conclusions and Further Work

12 The concepts for exoskeletal systems have been around for some time and the
13 potential for these devices in applications ranging from construction, manufacturing
14 and mining to rescue and emergency services is immense. In recent years, research
15 has been driven by possible uses in medical/rehabilitation and military applications.

16 Yet despite the obvious benefits from the use of exoskeletons, there still remain
17 barriers to the effective use and exploitation of this technology. One of the most
18 pertinent of these factors is the power and actuation system. In particular, since
19 an exoskeleton is in essence a robot coupled to a human at several contact points,
20 there are issues of safety and control.

21 This paper address these issues through the development of lightweight upper
22 and lower body systems that are powered by braided pneumatic muscle actuators
23 (PMAs) which provide systems with high power, good control, accurate motion
24 and compliant behavior that permits a soft, and therefore safer, interaction with
25 the user. The paper has shown the background to the development of the systems,
26 showing a brief history of exoskeletal systems. Based on this, a user requirement
27 has been defined. The paper has shown the design, construction and testing of ultra
28 low mass arm and leg based exoskeletons. These systems are shown to be able to
29 augment the power of users (either able-bodied or unwell) providing up to 100%
30 assistance for seven DOFs in the arm and five DOFs in the legs. Along with the
31 above functionality, the system was designed to meet the specifications of lightness,
32 gravity compensation, ease of fitting and adjustment and relatively low mechanical
33 complexity, which are essential for any system that is in direct contact with the
34 human operator. It has been shown that the system can provide a stable, compact
35 and viable augmentation system for rehabilitation activities involving both arm
36 and leg actions. Results have been introduced to demonstrate the capability of the
37 system as a physiotherapy training facility.

38 Ongoing research will seek to address several issues related to the individual
39 lower and upper body systems and the combined unit. For the upper limb, future
40 work is addressing rehabilitation post stroke through task based therapies, which
41 is a new technique that shows excellent potential for improved long-term stroke

1 recovery. The use of daily exoskeleton based therapy has shown excellent potential.
 2 Studies will also look at the use of the exoskeletons for daily use by those with
 3 muscle wastage, permitting increased independence. For the lower limb, the goal
 4 is assistive activity post paralysis and stroke. The exoskeleton has been integrated
 5 with a tread mill and active body support systems provide graded limb loading
 6 during training.

7 In the non-medical direction, work is ongoing to develop the system to augment
 8 power assistance during the execution of simple tasks, such as load lifting or tasks
 9 that require repetitive stress motions. This will include the understanding of the
 10 power requirements of these tasks and the development of effective force amplifica-
 11 tion control strategies that will enable the execution of the physical activities.

12 Further studies will also consider the development of fully independent power
 13 sources and increased actuator power for operations involving user power outputs
 14 exceeding current physical levels.

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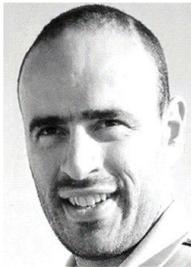
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