Design of Human-Friendly Powered Lower Limb Rehabilitation Orthosis

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Abstract. Many patients with spinal injures are confined to wheelchairs, leading to a sedentary lifestyle with secondary pathologies and increased dependence on a carer. Increasing evidence has shown that training using devices such as Reciprocating Gait Orthoses (RGO) reduces the incidence of these secondary pathologies, but the physical effort involved in this training is such that there is poor compliance.

This paper reports on the design of a new "human friendly" orthosis powered by high power pneumatic Muscle Actuators. The combination of a highly compliant actuation system with an intelligent embedded control mechanism which senses hip, knee and ankle position, velocity, acceleration and force produces a powerful yet inherently safe operation for paraplegic patients. The application of this technology will greatly improve the rehabilitative protocols for paraplegic patients.

Keywords: *Pneumatic muscle actuators, Exoskeleton, Design, Legs, Wearable, Bio-mimetic, soft actuator*

1. INTRODUCTION

Although reciprocal walking for paraplegic patients with complete thoracic lesions has been routinely available since the early 1980s, many patients with spinal damage are almost permanently confined to a wheelchair [1]. However, research has shown that the ability to stand and walk:

i). decreases the instances and severity of secondary problems including; the formation of contractures in the lower limbs, pressure sores, bowl infections, lower limb spasticity, osteoporosis, and kidney and urinary tract infections,

ii). reduces patient dependence on a carer,

iii). improves cardiopulmonary functions,

iv). has a positive psychological effect which impacts on the rehabilitation process, attempts to gain employment, and family and social life [2].

Locomotor training in particular, following neurological injury has been shown to have many therapeutic benefits [3]. Current treatment which often involves treadmill stepping with manual assistance and partial body weight support facilitates increased health care. However, manual assistance relies on physiotherapy procedures which are extremely labour intensive requiring high levels of one to one attention from highly skilled medical personnel. In addition, the treatment can be highly variable from therapists to therapist and throughout the day due to fatigue. At the same time the training and rehabilitation activities place extreme physical strain on the patient and the tremendous effort required leads to low levels of compliance. All this must be achieved in an environment in which there is an international shortage of staff with appropriate training [4].

This paper reports on the design of a new "human friendly" orthosis powered by high power pneumatic Muscle Actuators. Initial sections will introduce the area of passive and powered orthosis and show the benefits that can be gained through such devices. Section IV onwards considers safety aspects of the systems and them proposes the use of soft actuators as a possible solution. The design of a lower limb orthosis provided. The paper concludes with a discussion oif the current use and future developments needed.

II. PASSIVE ORTHOSES

In the case of the familiar Reciprocating Gait Orthoses (RGOs) used by patients with traumatic paraplegia (paralysis), ambulation is always achieved with knees locked since such patients are at a high risk of falling due to the lack of muscle power in the knee extensors. Hence RGO wearers currently only have the option of walking with straight leg(s) which is inefficient, un-cosmetic and makes walking more difficult [5-6].

To try to address this Simcox et al. designed a solenoid-activated orthotic knee joint but confirmed the findings of Harrison et al. [7] who had problems with unlatching the knee joint when the knee was not fully extended. An experimental electro-pneumatic active gait orthosis based around a modified ARGO (Advanced Reciprical Gait Orthosis, RSL/Steeper) was designed by Belforte et al. [8] but this was heavy and bulky and cosmetically undesirable. Suga et al developed a knee-ankle-foot orthosis with a lateral orthotic knee joint unit that controlled sagittal plane knee movements using a microcomputer and external sensors, however, patients again commented that the orthosis was heavy, and that the knee was extremely bulky and cosmetically unacceptable [9]. Saitoh et al, described a new hip-knee-ankle-foot orthotic system using a medial single hip joint for paraplegic standing and walking [10]. Unfortunately, this did not supply the total range of biomechanical functions required for a normal gait. Genda et al designed a new walking orthosis for paraplegics using a hip and ankle linkage system, but the hybrid orthosis did not facilitate walking with flexed knees during swing i.e the knees were still locked in extension [11].

III. POWERED TRAINERS

One alternative to traditional passive orthosis for locomotor training is the use of powered limbs. Within this approach there are two generic trends: one based on stimulating the muscles of the patient to provide the power for motion and support, and a second using powered orthoses to provide assistive inputs to drive the joints and support the body mass.

i). Function Electrical Stimulation (F.E.S.)

In the first of these approaches based on Function Electrical Stimulation (FES), the muscles of the patient's leg are externally stimulated to generate the support and motions. FES or as it is on occasions termed Function Neuromuscular Stimulation (FNS) is the use of low-voltage electrical signals to elicit a skeletal muscle response. The basic premise of functional neuromuscular stimulation is that a viable muscle, even though atrophied, can still be activated and controlled by means of electrical stimulation applied below the level of injury. Research into FES has been on-going for over 3 decades and considerable progress has been made during that period, with FES based bikes used to increase muscle strength and even a first commercial FDA approved walker (Parastep) [12].

There have also been attempts to provide hybrid orthoses linking F.E.S to traditional orthosis for more efficient orthotic intervention [13]. However, a nine-fold increase in energy consumption for patients to stand and ambulate has been reported [14-16] due to over stimulation of anatomical muscle groups and subsequent unsuitably high torque production [17]. As well as being tiring and inefficient, this walking is also ungainly and anecdotal evidence suggests that patients often reject their orthosis when walking with the leg in extension [18]. The high energy/effort required by the patients in all the current rehabilitation options means that compliance is low.

ii). Externally Powered Orthosis.

With the second approach based on the externally powered orthosis, support and power is provided by actuators attached to the external mechanical structure. The best known of these powered designs are Lokomat a 4 dof per leg treadmill based system and the Fraunhofer walker which provides 3 dof per leg in a crank slider motion with rotation for the ankle [4].

The Lokomat, in particular, is a motor driven exoskeleton device that employs a body weight support suspension system and treadmill. Attached to the lower limbs, the Lokomat moves the patient's legs through position controlled trajectories that aim to mimic normal human gait patterns. This is accomplished by utilizing motors that are precisely synchronized with the speed of the treadmill. The hip and knee joint angles are controlled in real time by software to achieve kinematically correct stepping behaviours. Each of the four motor-driven joints is individually controlled to correspond precisely to the desired joint angle trajectories. Sensors in the motors provide an indirect indication of the amount of effort the patient is generating to achieve walking in an upright posture. Since this process relieves therapists from the rigours of manually assisted treadmill gait therapy, the training sessions can be longer and more repeatable.

IV. SAFETY IN REHABILATION SYSTEMS

Fundamental to the operation and development of a powered locomotor trainer with operational parameters suited to use by paraplegics is the safety. Externally powered devices such as the Lokomat have achieved some success but the nature of the interaction between the patient and the user is critical, and although there are a variety of safety features built into these design a overarching paradigm of safety and dependability is critical.

In recent years, there has been great interest generated in the emerging field of human-centered robotics or Human Adaptive Mechatronics [19] which studies close interactions between robotic systems and humans, including direct human-robot contact. In such applications, traditional figures of merit such as bandwidth, peak force or torque, and work envelop, do not fully define the requirements and specifically, do not address the safety requirements.

For any robot or mechanical systems safety is dependent on a mechanism's mechanical, electrical, and software design characteristics. However, the biggest danger when working in close proximity with robots/mechanisms is the potential for impacts resulting from the large effective inertia (effective impedance).

This risk can be evaluated using the Head Injury Criteria (HIC) developed by the car industry and used to predict accelerations likely to cause serious injury during an impact between a mechanism and a human. This research has shown that simply adding a compliant (rubber) covering would be impractical requiring more than 15cm of skin to reduce values for a small robot to an acceptable level. Neither does the use of a covering address the root cause– namely the large effective inertia of most mechanisms [20].

If inherent safety is to be achieved, it is necessary to design mechanisms that have naturally low impedance and then couple this with robust electrical and software safety systems.

V. ACTUATORS DESIGN

Actuators and actuation systems are essential parts of all mechatronic structures providing the forces, torques and mechanical motions needed to move the joints, limbs or body. To provide safety in human centre robotics the work at Salford has focused on the use of pneumatic Muscle Actuator (pMA) and has led to the development of a range of actuators with enhanced performance characteristics [21] based on an adaptation of a braided pneumatic actuation design dating back to at least the 1920s.

The actuator used in these designs consists of an inner liner with a tubular braided mesh outer. When the air within the actuator is pressurised it contracts by 25-35% and depending on the muscle diameter can produce a contractile force of up to 7000N, from a muscle with a 70mm nominal diameter and a weight of less than 100g. The detailed construction, operation, and mathematical analysis of these actuators can be found in [21-22], however, they key features that make this actuation technique suitable for powered assistive devices include:

- Muscles can be produced in a range of lengths and diameters and are simple to manufacture.
- An extremely high power to weight ratio.
- Muscles contract by 30-35% of their dilated length, comparable with natural muscle.
- ‘Soft’ construction and finite maximum contraction make pMA safe for human-machine interaction.
- Muscles can be controlled to a displacement accuracy of 1% and can have a bandwidth of 5Hz when operating with an antagonist.
- Compared with natural muscle pMAs provide up to 10 times more force for a similar cross-sectional area.

VI. MECHANICAL DESIGN CONCEPTS

The mechanical structure proposed to form an orthosis to assist those with paralysis or muscle wastage consists of a 10 Degrees of Freedom mechanism (5 DoF for each leg). This corresponds to the motion of the human legs from the hip to the ankle excluding lower leg rotation.

Conceptually the basic structure is based on the well proven reciprocating gait orthosis (RGO) common in many aspects of spinal injury rehabilitation, which has been augmented to provided enhanced mobility as well as powered operation.

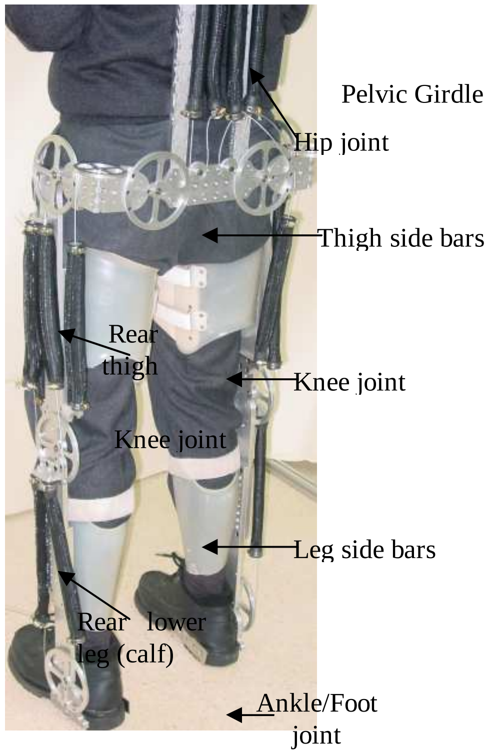


Figure 1 – The Exoskeleton Legs

The hip structure has 3 DoF (flexion/extension, abduction-adduction and lateral-medial rotation) as shown in outline in figure 1. This provides equivalence with the normal human motion at the hip but exceeds the single degree of freedom of the RGO where only hip flexion-extension is possible. The range of motions at the 3 joints of the hip for “normal” motion and the exoskeleton are shown in table 1 and it can be seen that there is good correlation between the needs for leg motion and the capacity of the powered orthosis.

The structure has 1 DoF at the knee permitting flexion/extension of the lower leg. This is in line with designs for current RGOs. The range of motion at the knee is shown in table 1 and this is once again comparable with achievable human motion ranges.

The final DoF is at the ankle permits Dorsi/plantar flexion of the foot, figure 1. Eversion/inversion is not currently possible but this still exceeds the RGO design although it is less than normal ankle motion.

The leg structure is constructed primarily from aluminium, with joint sections fabricated in steel, using precision mechanics. Support for the thighs and the calves are provided by moulded structures tailored to the anatomy of the patient. These moulded structures are mounted on the aluminium structures.

The total weight of the exoskeleton, consisting of the both legs and a rigid spine (on to which the muscles for hip flexion/extension and abduction/adduction mounted) is 12kg. The length from the hip to the knee is 520mm with 500mm from the ankle's base to the top of the knee.

As with electrical systems it must be recognised that this mass does not include the power source, but does include all valves and electrical control systems. This power source is compressed air which is readily and safely available in most hospitals and clinics.

Joint/ Segment	**Movement**	**Human Range of Motion**	**Exoskeleton Range of motion**
Hip	Flexion	100°-125°	135°
	Extension	10°-30°	0°
	Abduction	40°-45°	80°
	Adduction	0°-25°	45°
Extended Hip	Internal Rotation	35°-45°	70°
	External Rotation	45°-50°	70°
Knee	Flexion	120°-150°	135°
Ankle	Plantar Flexion	20°-50°	30°
	Dorsi Flexion	15°-30°	30°

Table 1 Ranges of motion for the "normal" leg and exoskeleton [23].

The compact actuator structure allows for integration as close as possible to their respective powered joints. The ankle actuators (two actuators) are mounted at the side of the calf while actuators for the knee and the leg hip rotation actuators (two actuators) are mounted on the lateral surface of the thigh. The remaining hip actuators (four actuators) are mounted on the body brace behind the operator's back.

Each antagonistic muscle pair attaches around an instrumented pulley which maintains a constant defined moment arm at differing joint rotation angles. The attachment over the pulley is cable driven which in combination with the muscles permits tolerance of mechanical misalignments. This ability to cope with misalignments is a key feature of the pMA that permits lower cost in the design and construction of the exoskeletal frame. The position of each joint is sensed using high linearity potentiometer with torque feedback on the muscle tension through strain gauges integrated into the spokes of the pulleys. Depending on the activated joint, muscles of differing sizes are used to produce the propellant action. In general, the muscles were of 20 mm diameter, with an 'at rest' length varying from 500mm – 700mm.

The performance specification for the joints of the human leg are shown in table 2 [23], together with the projected joint torque. This shows that the forces are less than required for typical healthy walking. Nonetheless, this should permit assistive action for those with limb wastage and paralysis.

Joint	**Human Isometric Strength**	**Exoskeleton torque @ 5 bar**
Hip		
Flexion/ Extension	110Nm	40Nm
Adduction/ Abduction	125Nm	45Nm
Rotation	-	6Nm
Knee		
Flexion/ Extension	72.5Nm	40Nm
Ankle		
Flexion/ Extension	19.8Nm	40Nm

Table 2 – Performance of pMA in Powered Assist Device

VII. SYSTEM CONTROL / USER INTERFACE

As a pneumatically powered structure, air flow control valves are needed. Eight port 2/2 valves in an integrated package 45mm x 55mm x 55mm weighing less than 300g (MATRIX) are used within this design and mounted at the base of the spine. These valves can be driven and controlled at up to 400Hz using a PWM signal. This provides rapid, smooth motion. By incorporating a pressure sensor into the valve inlet, closed loop pressure control is also possible. Pulsing of the valves along with data collection from the position, pressure and torque sensors is controlled from local dedicated microcontrollers with I/O, ADC and communication port facilities. The external PC is used to supervise the working conditions of the prototype.

Each individual muscle pair (joint) is controlled by a local micro-controller (Motorola MC68HC811) which mates with the valve assembly for compact operation.

Each MCU runs at 2 MHz, and can control up to 8 pMAs (8 inlets + 8 outlets). Each antagonistic pair is controlled by simple PID feedback controller on all the torso joints. As the muscles operate in pairs the value provided by the controller is added to one of the muscles and subtracted from its antagonist pair.

The MCUs are connected through a data bus to the interface PC, running Windows based software developed in Borland Delphi.

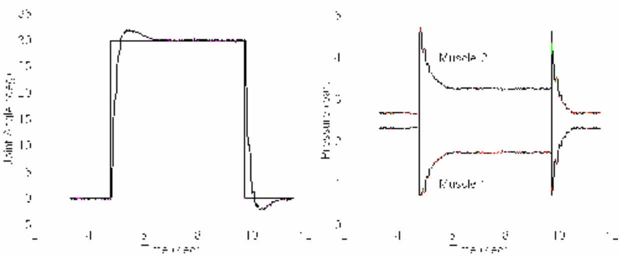
VIII. EXPERIMENTAL RESULTS

A series of joint operational tests were conducted to determine the "stability" of the system actuators.

Figure 2a illustrates a fast step response with a rise time of 0.3sec and an overshoot of 6%. Figure 2b displays the gauge pressure inside the two pneumatic actuators during the step response.

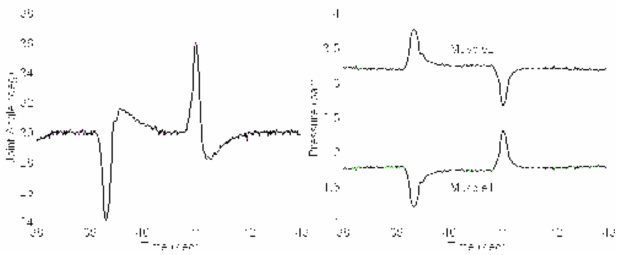
In the second experiment the sensitivity and response of the system to load variations were explored. Figure 3a shows the position of the joint settled initially around the desired position $\theta_d = 30\,\deg$. At $time \approx 39\sec$ a negative disturbance load is applied to the joint (~1Nm). As can be seen in the figure, the joint initially moved away from the desired position due to the disturbance load. The control scheme responds by increasing the pressure in muscle2 and reducing the pressure of muscle1, figure 3b and as can be seen, the joint rapidly settles again to the desire position. At $time \approx 40.8\sec$ the disturbance load was removed. The joint position again initially moved away from the settled point but quickly moved back and settles again to the desired position. These experimental results revealed that the control scheme had a good ability to cope with load variations.

In the next experiment, the bandwidth of the system was measured by applying a series of sine-wave inputs of different frequency. The results in figure 4a show a system bandwidth of approximately 1.4Hz. Additionally, figure 8b illustrates the position of the joint while the system is tracking a sine-wave input of 0.5Hz. The experimental results revealed that the above closed loop joint control scheme compensates for the actuator shortcomings and enhances the quality of the response, in terms of accuracy and bandwidth, of the antagonistic scheme to position commands. Figure 4b also reveals a phase lag which follows a characteristic profile typical of a 1st order system. The current performance is less than would be acceptable for a knee prosthetic, however, on-going research suggests that this can be increased by several hundred percent and this will form part of future developments of this system [8].



(a) (b)

Figure 2 – (a) System step response, (b) Muscle Efforts.



(a) (b)

Figure 3 – (a) System response to load disturbances, (b) Muscle Efforts.

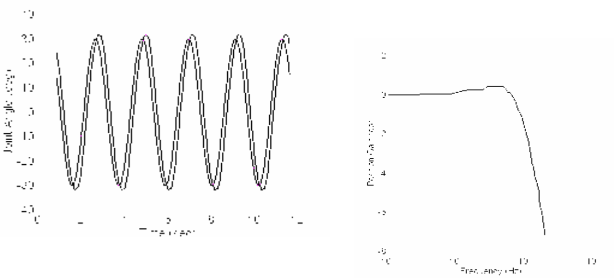


Figure 4 – (a) System frequency response (b) System-tracking response of sine-wave input.

Having achieved a level of joint control and co-ordination the exoskeleton was tested with a healthy individual to ensure stability and safety. These initial tests showed that motion was possible in a regulated and safe manner with the structure providing support and power motions at the joints.

VIII. CONCLUSIONS AND FUTURE WORK

This work has shown how complex biologically inspired structures can be constructed and powered by a 'soft' actuator that macroscopically has many characteristics similar to natural muscle, while still retaining beneficial attributes of conventional mechanical systems. These structures and the ability to provide controlled compliance regulated power could be of significant benefit in the construction of a power assist device that can be used to augment the strength of those suffering from degenerative muscle wasting diseases.

Future work will further investigate the use of this structure in power assist modes. Key developments will included:

i) Enhanced power outputs from the actuators to equal the power of human leg muscles.
ii) Integration of exoskeleton into a full body support kit based around a treadmill walker.
iii) Continued testing and validation with healthy test subjects.
iv) Testing with subjects suffering from muscle wastage or paralysis.

The use of the design philosophy outlined in this paper and in particular the utilisation of a soft actuation system provides a system with a inherent safety and dependability profile that cannot be easily achieved with conventional designs and may provide a valuable insight into development of powered assistive devices.

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