Control of a Biomimetic "Soft-actuated" 10DoF Lower Body Exoskeleton

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Abstract: *The successful motor rehabilitation of stroke, traumatic brain/spinal cord/sport injured patients requires a highly intensive and task-specific therapy based approach. Significant budget, time and logistic constraints limits a direct hand-to-hand therapy approach, so that intelligent assistive machines may offer a solution to promote motor recovery and obtain a better understanding of human motor control.*
This paper will address the development of a lower limb exoskeleton legs for force augmentation and active assistive walking training. The twin wearable legs are powered by Pneumatic Muscle Actuators (pMAs), an experimental low mass high power to weight and volume actuation system. In addition, the pMA being pneumatic produces a more natural muscle like contact and as such can be considered a soft and biomimetic actuation system. This capacity to "replicate" the function of natural muscle and inherent safety is extremely important when working in close proximity to humans. The integration of the components sections and testing of the performance will also be considered to show how the structure and actuators can be combined to produce the various systems needed for a highly flexible/low weight clinically viable rehabilitation exoskeleton.

Keywords: *Exoskeleton, Legs, Rehabilitation, Pneumatic muscle actuators, 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 [3]. 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/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 [1].

Locomotor training in particular, following neurological injury has been shown to have many therapeutic benefits [2]. 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. In addition, the treatment can be highly variable from therapists to therapist and throughout the day due to fatigue and the training and rehabilitation activities place extreme physical strain on the patient with the effort required leading to low levels of compliance. All this must be achieved in an area with an international shortage of staff with appropriate training [3].

This paper will present a unique wearable exoskeleton as an intelligent assistive training device showing how the pneumatic Muscle Actuators (pMAs) can emulate much of the action of natural muscle. The design of the mechanical/kinematic structure, making extensive use of this new light & flexible actuator, materials and mechanical design will then be introduced. The mechanism for integrating and controlling the actions of the joints using the pMAs will be studied followed by an analysis of the performance for assistive walking, conclusions and future work.

2. Assistive Walking Devices

In the case of the familiar Reciprocating Gait Orthosis (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 [4-5].

To try to address a variety of modification have been made to the structure of the RGO but in all instances there have been operational issues ranging from limits on the motion that restrict the gait to the weight, size or cosmetic appeal of the device [6-10].

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. 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

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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) [11-17].

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 [3].

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.

3. Design for Safety

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 with the patient is critical, and although there are a variety of safety features built into these designs an overarching paradigm of safety and dependability is critical.

For any robotic systems safety is identified with 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 techniques such as the Head Injury Criteria (HIC) [19]. However, 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.

The systems outlined in this work uses a new pneumatic Muscle Actuator having low inertia and high compliance capacity combined with high power weight ratio (contractile forces up to 7000N in a 100g package). The detailed construction, operation, and mathematical analysis of these actuators can be found in [20-21], however, they key features that make this actuation technique suitable for powered assistive devices include:

- Macroscopic performance similar to natural muscle,
- Muscles can be produced in a range of lengths and diameters and are simple to manufacture,
- Muscles contract by 30-35% of their dilated length, depending upon construction.
- 'Soft' construction and finite maximum contraction make pMA safe for human-machine interaction.
- Muscles can be controlled to a displacement accuracy of 1% and with a bandwidth of 5Hz when operated as an antagonist pair.
- Compared with natural muscle, pMAs provide up to 10 times more force for a similar cross-sectional area.

4. Exoskeleton Legs Mechanical Design

The mechanical structure used to form an exoskeleton to assist those with paralysis or muscle wastage consists of a 10 Degrees of Freedom mechanism corresponding to the fundamental natural motion and range of the human legs from the hip to the ankle but excluding the less significant movements. The hip structure has 3 DoF in total (flexion/extension, abduction-adduction and lateral-medial rotation, but the last one being no co-axial was not fully implemented), 1 DoF at the knee permitting flexion/extension of the lower leg and 1 DoF at ankle (dorsiflexion /plantar flexion), represented in figure 1.

Figure 1 – Author Wearing the Lower Body Exoskeleton

The legs structure is constructed primarily from aluminium, with joint sections fabricated in steel, using precision mechanics. The leg is mounted onto a moulded lower body brace which is light, low cost and comfortable while providing a stable platform. The leg and the brace were constructed to the scale of a typical target human based on data from US Air Force personnel as shown in figure 3, and adjusted to the author without major changes to the set-up, although leg link length changes can easily and quickly be adjusted, if necessary. As with electrical systems it must be recognised that currently this mass does not include the power source (weighting under 5kg). The computational requirements are not severe and operation is possible with Atmel microcontrollers.

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

Each antagonistic scheme includes a high linearity potentiometer for position sensing and an integral strain gauge torque sensor. The muscles used in this project have a diameter of 2cm with an 'at rest' length of 50cm. Hip abduction/adduction have a length of 70cm.

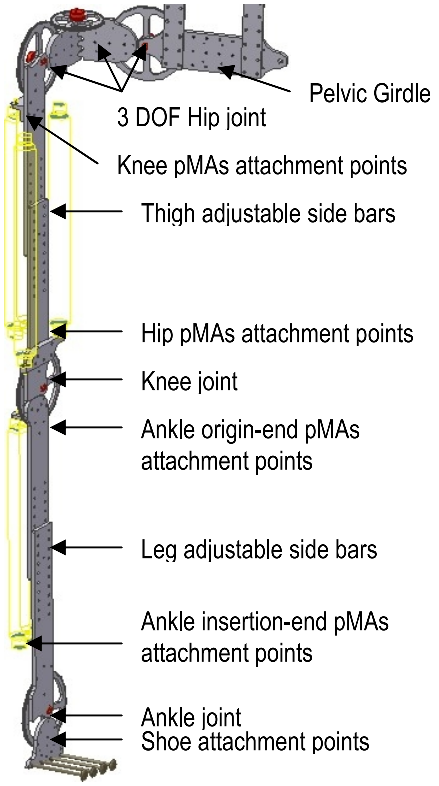

Figure 2 – Exoskeleton 3D CAD graphical representation with assembly and construction plans

The performance specification for the joints of the human lower limbs are shown and compared in table 1, together with the achieved maximum joint torque and range of motion. The total weight of the exoskeleton consisting of the both legs, actuators, electronics and rigid spine is 12kg assembled with an adjustable bar length of 520mm from the hip to the knee and 500mm height from the ankle's base to the top of the knee.

JOINT / SEGMENT MOVEMENT	Human Isometric Strength/Range	Achieved Joint Torque Range
HIP		
Flexion/Extension	110Nm 120°/20°	60Nm 135°/45°
Adduction/abduction	125Nm 45°/30°	65Nm 135°/135°
Internal Rotation	35°-45°	110°
External Rotation	45°-50°	110°
KNEE		
Flexion/Extension	72.5Nm 140°	60Nm 140°
ANKLE		
Plantar Flexion/Dorsiflect	19.8Nm 50°/30°	60Nm 105°/45°

Table 1 – Performance and ranges of motion of human versus pMA powered exoskeleton device

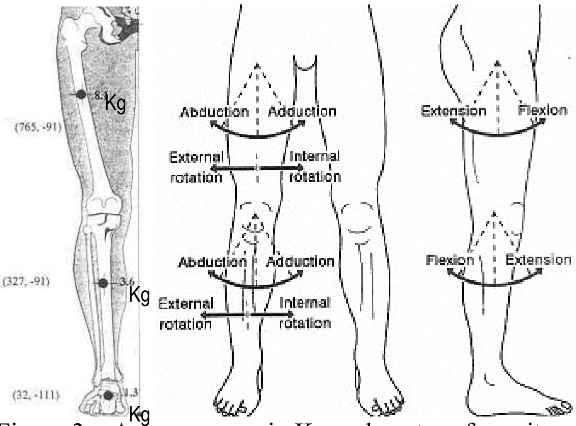

Figure 3 – Average mass in Kg and centre of gravity of limbs and joint coordinates conventions

5. pMAs Design and Control

The original concept of braided pneumatic muscle actuator, the McKibben Muscle was developed for prosthetic applications in the 1950's but it fell into disuse because of the complexity of control, the need for a compressed air supply and the relative ease of use of electrically powered prosthesis controlled by myoelectric signals has lead to their replacement [22].

As is the case with biological muscles in order to drive a joint in two directions an antagonistic pair of muscles is required, since the pMAs is only capable of producing a pulling force when it contracts axially during the expansion constrained by the outer layer. The figure 4 represents the typical control scheme using pMAs.

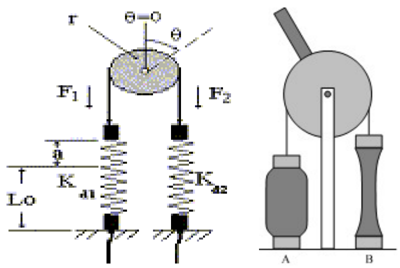

Figure 4 – Torque transmission using antagonistic pair

In the previous figure, the pMAs are modelled as pure springs with variable stiffness K_{d1}, K_{d2}. For an angle θ the forces developed by the actuators are given by [20]:

$$F_1 = K_{d1} \cdot (a + r \cdot \theta) \quad \text{and} \quad F_2 = K_{d2} \cdot (a - r \cdot \theta)$$

To achieve the maximum controllable range of motion, a has been set equal to the half of the maximum displacement.

Where ΔP is computed using a PID control law

$$\Delta P = K_{p1} \cdot e + \frac{1}{T_i} \int e + T_d \cdot \dot{e} \quad \text{and} \quad e = \tau_d - \tau_s$$

is the joint torque error, $P_{\max}$ is the maximum pressure within the pneumatic muscle actuators. Therefore the torque developed by the muscle becomes[20]:

$$T = 2 \cdot r \cdot a \cdot K_p \cdot \Delta P - 2 \cdot r^2 \cdot (K_p \cdot P_{max} + K_e) \cdot \theta$$

K_p, K_e represent the stiffness per unit length and pressure.

6. 3D Mathematical Model - Kinematics

Kinematics is the modeling of the spatial relationships between positions, velocities and accelerations of the structure links of a manipulator, described here in terms the standard Denavit-Hartenberg (DH) parameters. Figure 5 shows the assignment and table 2 the link parameters of coordinate frames to mechanical links in it's zero position.

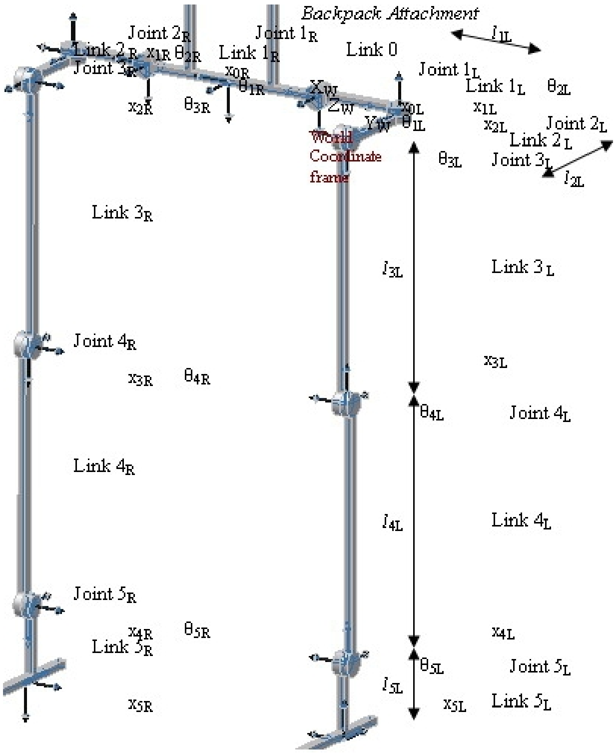

Figure 5 – Exoskeleton model with kinematic parameters

Link	Joint var	Angle θ_i	Twist α_i	Length l_i	Disp d_i	θ_i Maximum range limits
1_R	θ_{1R}	θ_{1R}	+90°	l_{1R}=103	0	-135° to 135°
2_R	θ_{2R}	θ_{2R}	-90°	l_{2R}=103	0	-110° to +110°
3_R	θ_{3R}	θ_{3R}	0	l_{3R}=480	0	-45° to +135°
4_R	θ_{4R}	θ_{4R}	0	l_{4R}=440	8	0° to +140°
5_R	θ_{5R}	θ_{5R}	0	l_{5R}=80	8	-45° to 105°
1_L	θ_{1L}	θ_{1L}	+90°	l_{1L}=-103	0	-135° to 135°
2_L	θ_{2L}	θ_{2L}	-90°	l_{2L}=-103	0	-110° to +110°
3_L	θ_{3L}	θ_{3L}	0	l_{3L}=-480	0	-135° to +45°
4_L	θ_{4L}	θ_{4L}	0	l_{4L}=-440	8	-140° to 0°
5_L	θ_{5L}	θ_{5L}	0	l_{5L}=-80	8	-105° to +45°

Table 2 – Denavit-Hartenberg link parameters for the exoskeleton left and right legs

From the link parameters, Direct kinematic algorithm to multiply the homogeneous transformation matrices of the simplified exoskeleton legs model to get the final exoskeleton homogeneous transform, given by:

Hence, if l_p is the distance of any generic point along the joint axis of the exoskeleton hip, knee or ankle frame, the absolute position of this point $p = \begin{bmatrix} p_x & p_y & p_z \end{bmatrix}^T$ in the upper link of the exoskeleton legs in respect to the defined coordinate base frame, will be given by:

$$p_{hip}(l_p) = \begin{bmatrix} c_1 l_2 c_2 + l_1 c_1 + c_1 c_2 l_3 c_3 - s_1 l_3 s_3 \\ s_1 l_2 c_2 + l_1 s_1 + s_1 c_2 l_3 c_3 + c_1 l_3 s_3 \\ l_2 s_2 + s_2 l_3 c_3 \end{bmatrix}$$

$$p_{knee}(l_p) = \begin{bmatrix} c_1 l_2 c_2 + l_1 c_1 + c_1 c_2 l_3 c_3 - s_1 l_3 s_3 + c_1 c_2 c_3 l_4 c_4 - s_1 s_3 l_4 c_4 - c_1 c_2 s_3 l_4 s_4 - s_1 c_3 l_4 s_4 \\ s_1 l_2 c_2 + l_1 s_1 + s_1 c_2 l_3 c_3 + c_1 l_3 s_3 + s_1 c_2 c_3 l_4 c_4 + c_1 s_3 l_4 c_4 - s_1 c_2 s_3 l_4 s_4 + c_1 c_3 l_4 s_4 \\ l_2 s_2 + s_2 l_3 c_3 + s_2 c_3 l_4 c_4 - s_2 s_3 l_4 s_4 \end{bmatrix}$$

$$p_{foot}(l_p) = \begin{bmatrix} c_1 l_2 c_2 + l_1 c_1 + c_1 c_2 l_3 c_3 - s_1 l_3 s_3 + c_1 c_2 c_3 l_4 c_4 - s_1 s_3 l_4 c_4 - c_1 c_2 s_3 l_4 s_4 - s_1 c_3 l_4 s_4 \\ s_1 l_2 c_2 + l_1 s_1 + s_1 c_2 l_3 c_3 + c_1 l_3 s_3 + s_1 c_2 c_3 l_4 c_4 + c_1 s_3 l_4 c_4 - s_1 c_2 s_3 l_4 s_4 + c_1 c_3 l_4 s_4 \\ l_2 s_2 + s_2 l_3 c_3 + s_2 c_3 l_4 c_4 - s_2 s_3 l_4 s_4 \end{bmatrix}$$

$$+ \begin{bmatrix} (c_1 c_2 c_3 c_4 - s_1 s_3 c_4 - c_1 c_2 s_3 s_4 - s_1 c_3 s_4) l_5 c_5 + (s_1 s_3 s_4 - c_1 c_2 c_3 s_4 - c_1 c_2 s_3 c_4 - s_1 c_3 c_4) l_5 s_5 \\ (s_1 c_2 c_3 c_4 + c_1 s_3 c_4 - s_1 c_2 s_3 s_4 + c_1 c_3 s_4) l_5 c_5 + (c_1 c_3 c_4 - s_1 c_2 c_3 s_4 - s_1 c_2 s_3 c_4 - c_1 s_3 s_4) l_5 s_5 \\ (s_2 c_3 c_4 - s_2 s_3 s_4) l_5 c_5 - (s_2 c_3 s_4 + s_2 s_3 c_4) l_5 s_5 \end{bmatrix}$$

In static consideration the same transformation is used to relate the external force and moments applying at the end-effector to the torques at the joints. These previous statements can be algebraically expressed using the following equations.

$$J \cdot \dot{\theta} = \begin{bmatrix} v \\ \omega \end{bmatrix}, \qquad J^T \begin{bmatrix} F \\ M \end{bmatrix} = \tau$$

Considering a generic exoskeleton legs with n degrees of freedom, as previously described J is the exoskeleton Jacobian, v, ω are the 3x1 vectors of translation and rotational velocity, with F, M being the 3x1 vectors that describe the forces and moments acting at one specified point on the exoskeleton structure, $\dot{\theta}$ is nx1 vector of joint rates and τ is the nx1 vector of joint torques/forces.

Thus, using equation the Jacobian J_{knee}, J_{ankle} and J_{foot} can be formulated.

$${}_0J_{knee} = \begin{bmatrix} -K_{y0} & -c_1 K_{z1} & s_1 s_2 K_{z2} - c_2 K_{y2} \\ K_{x0} & -s_1 K_{z1} & c_2 K_{x2} + c_1 s_2 K_{z2} \\ 0 & s_1 K_{y1} + c_1 K_{x1} & -c_1 s_2 K_{y2} - s_1 s_2 K_{x2} \\ 0 & s_1 & -c_1 s_2 \\ 0 & -c_1 & s_1 s_2 \\ 1 & 0 & c_2 \end{bmatrix}$$

$${}_0J_{ankle} = \begin{bmatrix} -A_{y0} & -c_1 A_{z1} & s_1 s_2 A_{z2} - c_2 A_{y2} & s_1 s_2 A_{z3} - c_2 A_{y3} \\ A_{x0} & -s_1 A_{z1} & c_2 A_{x2} + c_1 s_2 A_{z2} & c_2 A_{x3} + c_1 s_2 A_{z3} \\ 0 & s_1 A_{y1} + c_1 A_{x1} & -c_1 s_2 A_{y2} - s_1 s_2 A_{x2} & -c_1 s_2 A_{y3} - s_1 s_2 A_{x3} \\ 0 & s_1 & -c_1 s_2 & -c_1 s_2 \\ 0 & -c_1 & s_1 s_2 & s_1 s_2 \\ 1 & 0 & c_2 & c_2 \end{bmatrix}$$

The Jacobian formulated here can now be used to calculate the necessary hip, knee and ankle joint torques in order to "reflect " a specified external momentum or load generated at any single point between hip and knee, ankle or foot using the following equation:

$$\begin{bmatrix} \tau_1 \\ \tau_2 \\ \tau_3 \end{bmatrix} = {}_0J^T_{knee,ankleorfoot} \cdot \begin{bmatrix} {}_0F \\ {}_0M \end{bmatrix}$$

As previously noted, special care must be taken in order to ensure that the vector of the external load (both force and moment) has been expressed with respect to the same reference frame as the Jacobian for the validity of computed values. For the exoskeleton legs case, this reference frame is the base frame (frame 0).

Many control schemes require the inverse of the Jacobian. At a kinematic singularity the Jacobian becomes singular, and such simple control techniques will fail. There is a constraint in knee joint, such that it can move only forward, which solves the singularity problem.

6.1. Manipulator rigid-body dynamics

Robot dynamics is concerned with the equations of motion, the way in which the robot moves in response to torques applied by the actuators, or external forces. An impedance control scheme was employed for the overall rehabilitation/training exoskeletal system. The following equations of motion describe the dynamic behaviour of the exoskeleton for an n-axis are given below.

(The impact of the swing leg is assumed to be perfectly inelastic while ensuring that no slippage occurs. Moreover, a physical realisability of motion implies that the foot can't push on the ground. The dynamic equations for the five link structur during a single-support phase are of the form: $M(q)\cdot\ddot{q}+C(q,\dot{q})+G(q)=\tau$

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

Where:

- q — Is the $n\times 1$ vector of generalized joint coordinates describing the pose of the manipulator or the joint variable n-vector
- $\dot{q}$ — Is the vector of joint velocities
- $\ddot{q}$ — Is the vector of joint accelerations
- $\boldsymbol{M}$ — Is the $n\times n$ symmetric joint-space inertia matrix, or robot inertia tensor matrix
- $\boldsymbol{C}$ — describes Coriolis and centripetal torques/effects – Centripetal torques are proportional to $\dot{q}_i^2$, while the Coriolis torques are proportional to $\dot{q}_i\dot{q}_j$
- $\boldsymbol{F}$ — Is the friction vector that describes viscous and Coulomb friction and is not generally considered part of rigid-body dynamics
- $\boldsymbol{G}$ — Is the gravitational loading $n\times 1$ vector torques
- Q — Is the vector of generalized forces associated with the generalized coordinates q
- $\tau_{jo\,\mathrm{int}}$ — is the joint $n\times 1$ vector of the generalized actuated torques
- F_R — is the force that the leg generates at the end-tip
- J^T — is the transpose Jacobian of the manipulator

The above equation can be used to describe the interaction between an user and the exoskeleton.

7. System Controller and 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 200Hz using a PWM signal. This provides rapid, smooth motion. Development and adjustment of an controller and details of the design can be found in [6-8]. By incorporating a pressure sensor into the muscle 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 only used to store the data collected under normal working conditions of the prototype.

Each individual muscle pair or joint is controlled by a local microcontroller (Atmel ATMEGA8 – figure 6) which mates with the valve assembly for compact operation. Every valve microcontroller board consist of two microcontrollers, which allows to board handle with two muscle pairs (two joints).

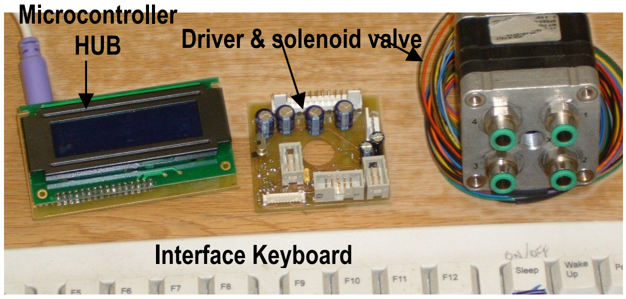

Figure 6 – Hub with interface keyboard & microcontroller board with valve drivers

Each MCU runs at 8 MHz, and can control up to 2 pMAs (2 inlets + 2 outlets). Each antagonistic pair is controlled by three PID controllers (two low-level for pressure and one higher-level for position/torque) on all the joints, figure 5. As the muscles operate in pairs the value provided by the controllers is added to one of the muscles and subtracted from its antagonist pair.

The MCUs are connected through a serial data bus to central controller or HUB. The hub consist one microcontroller Atmel ATMEGA128 with 2functions. Firstly the hub coordinates all valve control units and feed them with self-generated data which are based on the exoskeleton operating mode. Secondly the hub should be used only as the interface between PC and the valve controllers. In this case all inputs are generated by the PC. Communication between the hub and PC is completely wireless making use of new BlueTooth technology. Windows PC based interface and data operation software was developed in Matlab simulink.

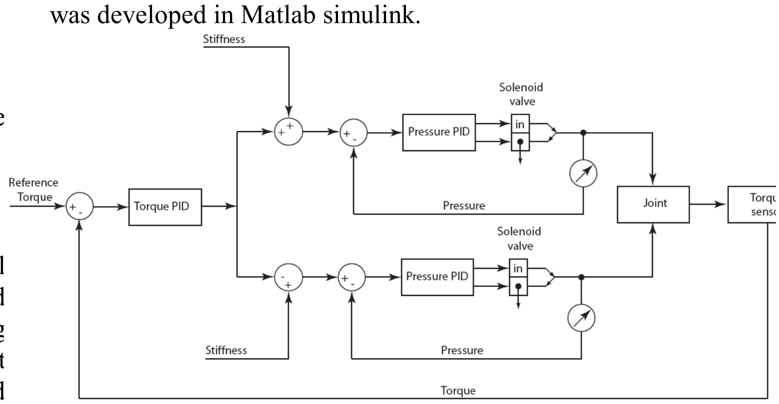

Figure 7 - Joint Torque Control scheme

8. Experimental Method & Results

A chirp signal (figure 8A-C) with different amplitudes and frequencies was used to evaluate the closed loop frequency response of each joint. For the ankle the amplitude was set to ±12º and ±25º with frequency range swiping from 0 to 4 Hz in 0.1Hz steps over a period of 180s as represented in figure 8A. The frequency was reduced to 1.5Hz on the knee/hip and when links were loaded with weight shown in figure3 it was reduced to 1Hz at the hip.

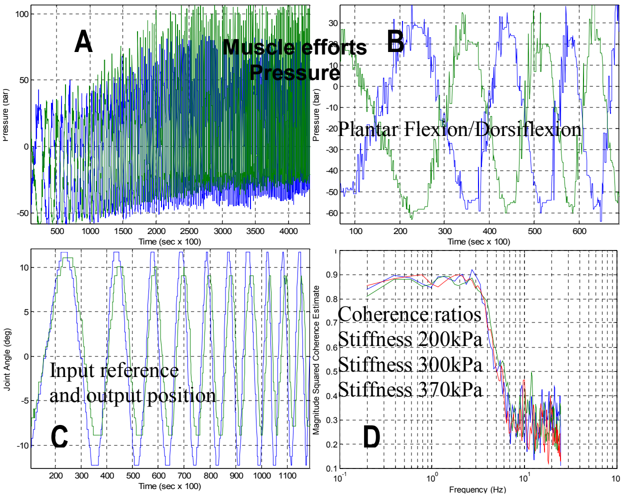

Figure 8 – AB-Joint muscle efforts & C-Position control

Figure 8C illustrates how well the implemented closed loop control scheme is able to track the reference and compensate for the actuator response shortcomings

The Transfer Function Estimate (TFE) was computed by averaging several cross powered spectral densities and using a suitable Kaiser window to reduce the impact of leakage. In parallel, to have an estimate on the precision how good the TFE was the coherence function was used. This coherence is a function of frequency with values between 0 and 1 that indicate how well the input corresponds to the output at each frequency, represented in Figure 8-D, where we can see that inside the excitation frequencies the TFE is very accurate (results over 0.7).

These were both implemented in Matlab, for all joints under different amplitudes, frequency range and stiffness.

Figure 9 AB & CD shows the right ankle transfer function to a 180sec; ±12º & ±25º reference sinusoidal chirp signal with frequencies from 0 to 4Hz, respectively.

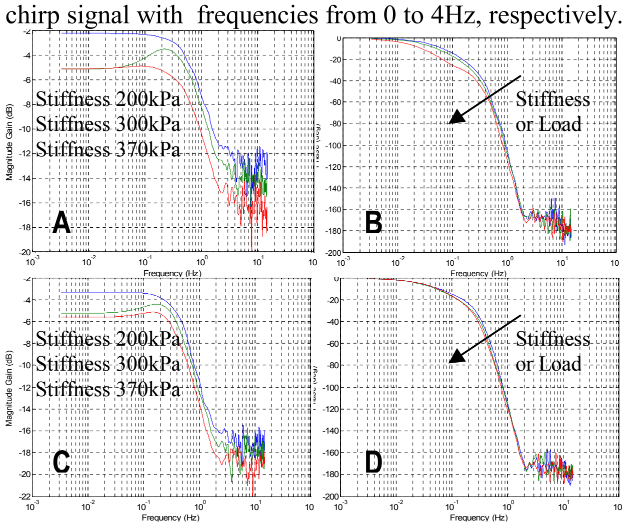

Figure 9 – Unloaded right ankle transfer function

Figure 10 AB & CD shows the same right ankle transfer function running the same test cycle but loaded with the average human weight (as given in figure 3).

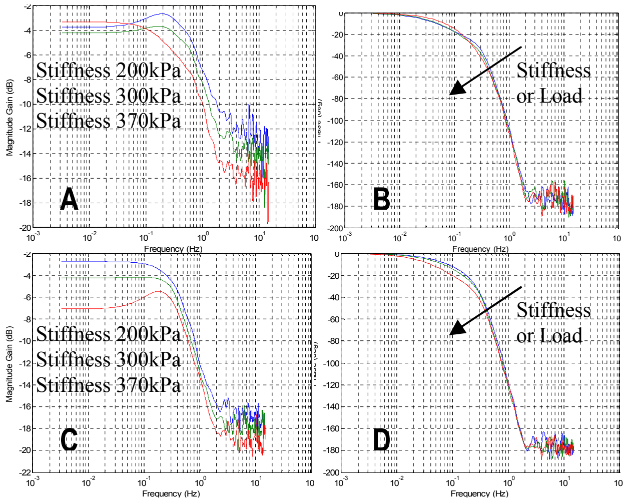

Figure 10 – Loaded right ankle transfer function

These figure where taken using Matlab and considering that the sampling frequency specified is 30Hz and the number of overlapping sample points of overlap from section to section is half of the defined for the Kaiser average window.

Experimental results revealed on previous figures shows a phase lag which follows a characteristic profile typical of a 1st order system. The current performance is less than would be acceptable for an knee prosthetic, however, on-going prototype updates and research suggests that this can be increased by several hundred percent and this will form part of future developments, some of them already being implemented at present for this system. Overall response, muscle efforts, ability to handle load disturbances, human gait system-tracking/guiding capabilities achieved with a healthy individual (to ensure stability and safety) revealed that it may be successfully used for some medical conditions involving degenerative muscle wasting diseases/weak lower limb muscles or reduced coordination of human motor control.

9. Conclusions and Future Work

The work presented here has shown how complex biologically inspired structures can be constructed and powered by a ‘soft’ actuator that macroscopically have many characteristics similar to natural muscle, while still retaining beneficial attributes of conventional mechanical systems. In fact, these characteristics together with the ability to mimic the human muscle principle proved to be well suited for this “biological implementation” and so may provide a valuable insight into the development of new range of powered assistive devices.

The lower body exoskeleton system effectiveness was presented and discussed and currently is being assessed when acting as a power assist device for rehabilitation, prosthetics or training with different task-specific therapy based approach, aimed to guide the human gait being tested. This are achieved/accomplished either by

augmenting, constraining, braking or even stopping the strength/movement of it's user.
We believe that in future it can be successfully used for some medical conditions as rehabilitation of stroke, traumatic brain/spinal cord/sport injured patients that suffering from degenerative muscle wasting diseases/weak lower limb muscles.

Future work will further investigate the use of this structure in different approach active and passive power assistive modes. Key developments will include:

- Enhanced power outputs from the actuators to equal the power of human leg muscles.
- Integration of exoskeleton into a full body support kit based around a treadmill walker.
- Objective evaluation of the use of the exoskeleton system by analyzing the changes in normal walking gait & EMG signals from the lower limb muscles.
- Continued testing and validation with healthy test subjects to develop a library of medical treatment procedures and training modes to obtain a better understanding of human motor control.
- Testing with subjects suffering from muscle wastage, paralysis or other muscular medical restrictive conditions or special physical regimes.

Technical possibilities of this future clinically viable exoskeleton device are one aspect, but correct research approach, time, resources and persistency in wide range of successful medical trials will decide on the appearance and implementation of this promising and challenging field of research to the benefit of patients.

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