Development of an Active Exoskeleton for Assisting Back Movements in Lifting Weights

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***Abstract*—In the present work, we show the development of an active exoskeletal device to aid the spine. This device is targeted at able-bodied people, who for various reasons may need help with lifting weights from the ground or more simply in daily movements. The characteristics of the device must respond to the characteristics of ergonomics, lightness, functionality, resistance, durability, and, last but not least, inexpensiveness, so as to be able to meet the needs of a large public. In order to equip the exoskeletal device with the listed characteristics, it was decided to create it using a combination of materials: polylactic acid for the back and the terminal shell, steel for the pins and spacers, and aluminum for the structure. This way, all the basic requirements for the success of the device have been met. Moreover, as for actuators, McKibben-type pneumatic muscles have been used, as they have particular characteristics of compliance and high power-to-weight ratio and are well suited to bioengineering systems.**

***Index Terms*—exoskeleton, pneumatic muscle, service robot, biomechanics, control**

I. INTRODUCTION

As reported by the European Union Labor Force Survey in 2007 [1], musculoskeletal disorders are the leading cause of occupational diseases in all work sectors.

It was found that, among the main musculoskeletal disorders, the diseases of the intervertebral discs are the most relevant and the pathologies of the rachis are widespread in the population, whether they carry out purely manual work (handling loads, construction workers, etc.) or they perform more sedentary jobs. The global characteristic of the spread of these diseases in the human body has led researchers to take care of the development of devices that aid the skeletal muscle system.

There are currently no exoskeletal amplifiers dedicated for the rachis, except for passive assistance devices of the bust. One of these is called Springzback and has been patented [2] under the name "Personal Upper Body Support Device for Lower Back Muscle Support." It is worn at the front of the body, and an adjustable spring opposes resistance to the user's chest when bending: this resistance partially reduces muscle strength while lifting, providing benefits during static postures with forward torsos. Another nonmarketed device, for exclusive use for the Marines, is known under the name HULC, which stands for Human Universal Load Carrier. This is an exoskeleton with electrohydraulic motorization born and developed for assistance in lifting and transporting heavy loads. The motor part of the exoskeleton is mounted behind it and serves to operate two cords connected to two grips gripped by the person: these knobs are the part that comes into contact with the object to be lifted and grasps it by friction. Very heavy containers have particular geometries on the sides that allow the insertion of the knobs, increasing the safety of lifting them. The anthropomorphic-type exoskeleton discharges the forces generated for lifting directly to the ground through the feet, which are anchored to the person's shoes.

There are several devices to aid the spine under study, such as the PLAD, developed by PeakWorks together with Queen's University and Ryerson University. The PLAD, which stands for Personal Lift Assist Device, is a passive aid device for personal assistance when lifting loads, which was patented in 2009 [3]. It offers a passive type of aid by exploiting the action of springs that accumulate elastic energy when the trunk lowers and release it when the it rises. It is formed by a rigid part that is worn and rested on the back and by bands secured to the legs below the knees. Elastic elements are connected to these parts to provide assistance.

Another passive assistance system is the Lift Assist Device (LAD) that was developed by the University of Utah [4]. There are two versions of the LAD: the first, Torsion Spring LAD, uses torsion springs as accumulators of elastic energy. The aid is equipped with a rigid extension, which is positioned behind the back and reaches the head. By modifying the preloading of the springs and the extensions, it can be adapted to various users. The second version uses elastic bands placed along the back (Bending LAD), which has been designed to provide the user with a wide range of movements.

Active devices are aids that provide energy and strength to the human body through an external source. One of these was developed in 2004 at Hokkaido University by a group of researchers [5], which was in the form of an exoskeleton, wearable by the user, on which an electric motor and a transmission are mounted with a cable wound on a drum.

The exoskeleton is worn and fixed to the rib cage, pelvis, and knees. The bearing structure of the

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exoskeleton is made of aluminum, and it allows the cable to apply a force perpendicular to the torso. The engine used to move the drum has a power of 40 W.

Another active device was studied at Okayama University [6] and developed as a servo-assisted garment for lumbar muscle support. This device consists of a prototype with pneumatic actuators mounted inside the garment that can be worn as a jacket. The actuators are positioned behind the back. There are two types of pneumatic muscle actuators: pneumatic rubber muscles of the elongation type and pneumatic muscles positioned inside the garment, cushion type, to create, when pressurized, a greater thickness between the back and the elongation actuators. Therefore, they have the task of increasing the arm between the useful force, generated by the actuators, and the sacral lobe joint L5-S1 so that it increases the useful torque provided by the device accordingly. The advantages of this aid are certainly the lightness and the increase of the arm between the useful force and the back, but there are also some aspects that can be improved. The force generated by the actuators is discharged on the vertebral column, adding to the compressive stresses. In addition, the cushion actuators, when pressurized, press perpendicular to the vertebral column, increasing the shear stress on it. The device presented in this work is an active, wearable exoskeleton of a simple design that is light, reliable, and easy to use, and that can overcome the problems presented by the devices described above.

II. Biomechanics Of Lifting

A. Low Back Biomechanics of Lifting

In order to correctly design the device, it is necessary to know and study the movement of the human body, as well as the distribution of the forces inside it during the lifting of loads. Therefore, it is necessary to identify a biomechanical model that represents its kinematics.

TABLE I. Segment Masses and Center-of-gravity Locations [9].

Segment	Mass (percentage body mass)	Location of centre of gravity
1. Head and neck	8.4	57% of distance from C7 to vertex
1a. Head	6.2	20 mm above tragion
2. Head and neck and trunk	58.4	40% of distance from hip to vertex
2a. Trunk	50.0	46% of distance from hip to C7
2b. Trunk above lumbo-sacral joint	36.6	63% of distance from hip to C7
2c. Trunk below lumbo-sacral joint	13.4	Approximately at the hip joint
3. Upper arm	2.8	48% of distance from shoulder to elbow joints
4. Forearm	1.7	41% of distance from elbow to wrist joints
5. Hand	0.6	40% of hand length from wrist joint (at centre of an object gripped)
6. Thigh	10.0	41% of distance from hip to knee joints
7. Lower leg	4.3	44% of distance from knee to ankle joints
8. Foot	1.4	47% foot length forward from the heel (half height of ankle joint above the ground) — mid-way between ankle and ball of foot at the head of metatarsal [1]

The model of Drillis and Contini is used, where they have considered different segments of the body by associating the dimensions according to a criterion of proportionality between the parameters of each segment with the height of the body [7]. As for the positions of the centers of gravity, a reference is made to Pheasant's studies [8], in which the trunk was further subdivided with respect to what was done by Drillis and Contini (Table I). Moreover, as for the posture-dependent forces, the Chaffin model is considered, which was specifically proposed for the movement for lifting loads [9, 10].

The Chaffin model is of a static type. Starting from the dimensions and mass of the body, from the posture and the load to be lifted, it provides forces on the lumbosacral joint, L5-S1, namely, F_{COMP} (the compression force) and F_{SHEAR} (the shear force) (Fig. 1). It provides the torque given by the loads with respect to the joint. The L5-S1 joint is the most critical in the entire spine, as it is considered the connection between the spine and the hip bone and the place where the load shows its effects (low back pain) during the activity of lifting weights. The model is based on the free body diagram of the human body with the loads using the L5-S1 plane as a cutting plane, which is oriented by the α angle with respect to the horizontal plane (Fig. 1).

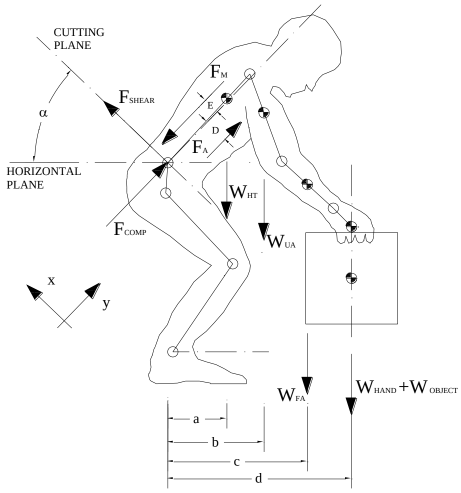

Figure 1. Chaffin model.

The model shows the F_{M} and F_{A} forces applied by the back muscles and the abdominal muscles, respectively, and the corresponding arms E and D with respect to the joint. The model quantifies the arm on which the F_{M} force acts relative to the L5-S1 joint in 50 mm and provides an expression for D, and it also allows an evaluation for the F_{A} force on the basis of the pressure of the abdominal muscles, P_{A} [10]:

$$D\ (\mathrm{mm}) = 6.7\ \mathrm{mm} + [(14.9 - 6.7)\ \mathrm{mm} \times \sin T], \quad (1)$$

$$P_A\ (\mathrm{mmHg}) = 10^{-4}\ (0.6516 - 0.005447 * \theta_H) * M_{L5S1}^{\ 1.8} \quad (2)$$

where T is the torso angle, θ_{H} is the included hip angle (knee-hip-shoulder), and M_{L5S1} is the force momentum with respect to the L5-S1 joint. In Fig. 2, the other characteristic angles of the model are the shinbone angle F, the included knee angle K, the sacrum angle β, the L5-S1 shoulder angle γ, the shoulder angle θ_{SE}, and the elbow angle θ_{EW}.

In the model, there are the arms that the forces have with respect to the L5-S1 joint (Fig. 1). They depend on the characteristic angles defined above. The one shown in Fig. 2 is the Chaffin model that has been extended by the authors, differentiating, with respect to the original model, the loads acting on the arms and on the load. The Chaffin model also allows the α angle to be determined according to the movement phase. The α angle is a function of the β angle, which in turn depends on other characteristic angles between the various segments of the body (Fig. 2). The following are the expressions of α, β [9, 10], and γ determined by the authors:

$$\alpha = 40 + \beta \tag{3}$$

$$\beta = -17.519 - 0.11863T + 0.22687K + 0.11904\ 10^{-2}\ TK + 0.499\ 10^{-2}\ T^2 - 0.753\ 10^{-3}\ K^2 \tag{4}$$

$$\gamma = T + \sin^{-1}\left[\frac{0.066}{0.262} \times \sin(T-\beta)\right]. \tag{5}$$

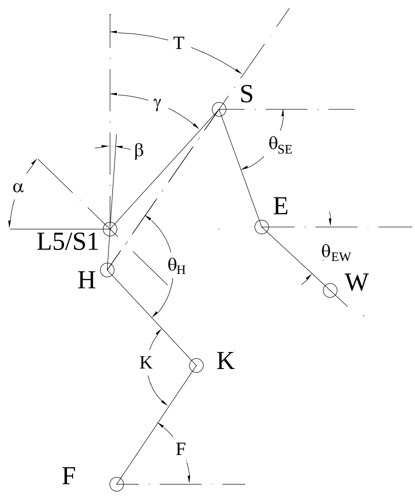

Figure 2. Characteristic angles.

Through the described relationships, once the equilibrium equations for the free body diagram of the Chaffin model are written, it is possible to derive the trends of the forces acting on the L5-S1 joint, F_{SHEAR}, F_{COMP}, and the force F_M on the back muscles:

$$F_M = \frac{F_A D - W_{HT}a - W_{UA}b - W_{FA}c - (W_{HAND} + W_{LOAD})d}{E}, \tag{6}$$

$$F_{SHEAR} = (W_{HT} + W_{UA} + W_{FA} + W_{HAND} + W_{LOAD})\sin\alpha, \tag{7}$$

$$F_{COMP} = F_M - F_A + (W_{HT} + W_{UA} + W_{FA} + W_{HAND} + W_{LOAD})\cos\alpha. \tag{8}$$

B. *Multibody Model*

To obtain reliable results, it is necessary to have a set of laws of motion of the characteristic angles that are coherent with each other and congruent with a natural movement. This was obtained through a multibody model constructed according to the methods presented above. By means of the multibody models, it is possible to take into account the dynamics of mechanical systems [11]. With the laws of motion of the characteristic angles, it was possible to construct the trends of the forces of interest in order to have a reference to evaluate the performance of the device to be developed. The multibody model was also validated by comparing the results that it provided for the forces of interest with those of the analytical model above. The user was considered to be 176 cm tall and to weigh 75 kg.

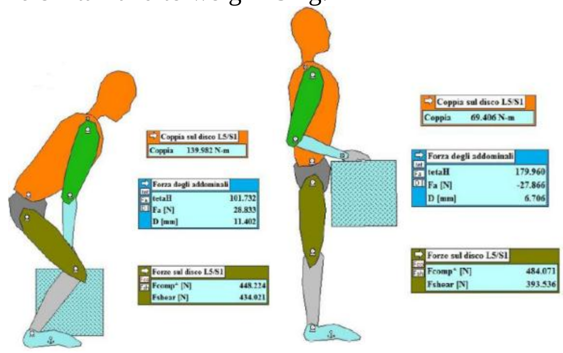

Figure 3. Multibody model.

In Fig. 3, the multibody model in the initial and final positions of the lifting movement is presented. The adopted laws of motion were −as follows:

$F(t) = 56 + 3.4\ t$; $K(t) = 102.7 + 7.73\ t$; $\beta(t) = 4.078 - 0.5078\ t$; $\gamma(t) = 42.438 - 4.2135\ t$; $\theta_{SE}(t) = 105$; $\theta_{EW}(t) = 92.4 - 7.25\ t$,

with t varying from 0 to 10 s.

Figs. 4–6 show comparisons between the results of the analytical model and those of the simulations by the multibody model. It can be seen that the results are quite similar, although the multibody model presents some oscillations due to the dynamics.

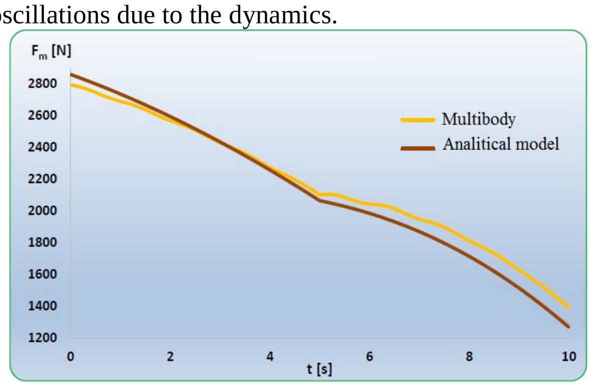

Figure 4. Comparison of the back muscles' force F_M.

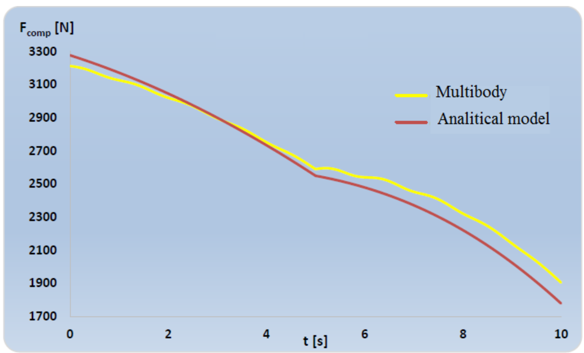

Figure 5. Comparison of the L5-S1, F_{COMP}.

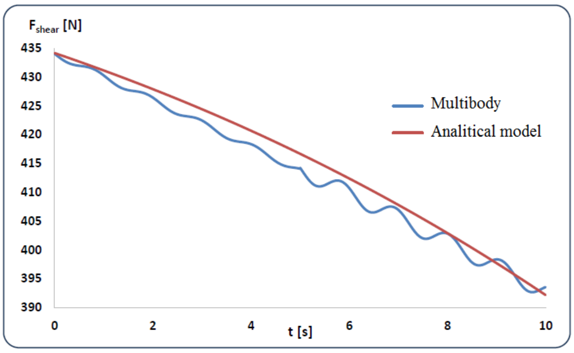

Figure 6. Comparison of the L5-S1, F_{SHEAR}.

III. Device for Lower Back Support

By means of a survey conducted on 180 people, it was possible to understand that a device that is able to prevent and alleviate back pain is desirable and to understand what its technical requirements should be. The device whose main function is to assist in lifting weights must be ergonomic, easy to use, and possibly aesthetically pleasing and have small dimensions.

A. *Technical Requirements*

The technical requirements are summarized in Table II, where the general requirements are translatedinto more technical ones.

TABLE II. Design Specification.

Assistance in lifting weights
Lifting load: 25 kg with 50% effort reduction
Management of assistance level
Ease of use and ergonomics
Easily wearable, comfortable
Intuitive operation
Respect for natural movements
Low weight
Not cumbersome
Low energy consumption

B. *Conceptual Design*

The idea is to develop a wearable device that is positioned parallel to the back and that does not apply on the body loads outside the strictly functional ones. The ideal would be to apply pairs on the trunk and the pelvis that tend to help the extension of the back by values of smaller forces possible and with directions always coinciding with those of the movements of the points on which they are applied in order to always have the maximum useful work without unnecessary and annoying additional components of strength.

The objective can be achieved by a device equipped with two structural elements—one corresponding to the back and one corresponding to the pelvis—connected to each other by a hinge below the plane containing the joint L5-S1 to avoid increasing the cutting force. The two structural parts will be connected in the upper part of the torso and on the pelvis and can control the extension of the back through a linear actuator according to the diagram in Fig. 7.

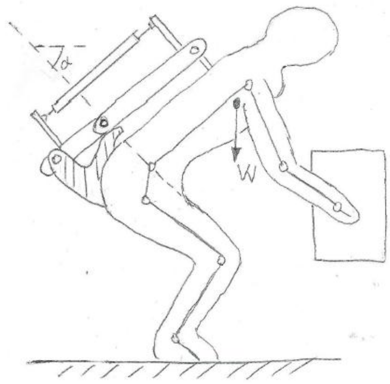

Figure 7. Conceptual design.

This way, it will be possible to apply three forces: one in the upper part of the shoulders, one in the lower part of the pelvis directed backward, and one in the upper part of the pelvis directed forward, which can apply the torques necessary for the extension of the back. The actuator is of a pneumatic muscle type. The device will be directly supplied by a pneumatic line since it is addressed to work predominantly indoors.

Based on this idea, two free body diagrams were considered for the evaluation of the effects of the device described on the forces of interest. The first derived from the Chaffin model in which the user's body is represented with, in addition to the forces already present, the force applied by the device. The cutting plane is always the transverse plane through the L5-S1 joint, and another plane vertically cuts the connection between the human body and the device behind the back. The other one concerns the upper structure of the device delimited by the same cutting planes (Fig. 8). By writing the equilibrium equations for the two free body diagrams, it is possible to obtain the forces of interest, F'_{COMP} and F'_{SHEAR}:

$$F'_M = \frac{F_A D - W_{HT} a - W_{UA} b - W_{FA} c - (W_{HAND} + W_{LOAD}) d - R_s h}{E}, \quad (9)$$

$$F'_{COMP} = F'_M - F_A + (W_{HT} + W_{UA} + W_{FA} + W_{HAND} + W_{LOAD}) \cos\alpha, \quad (10)$$

$$F'_{SHEAR} = (W_{HT} + W_{UA} + W_{FA} + W_{HAND} + W_{LOAD}) \sin\alpha - R_S. \quad (11)$$

We can immediately see that there is a reduction in all forces in relation to the presence of the *Rs* force applied by the device in the upper part of the trunk.

In addition, in this case, to better quantify the performance along the entire movement and to better define the dimensions of the device, the multibody model already considered was used, to which the model of the aid device conformed according to the indications that emerged when the conceptual design phase was added.

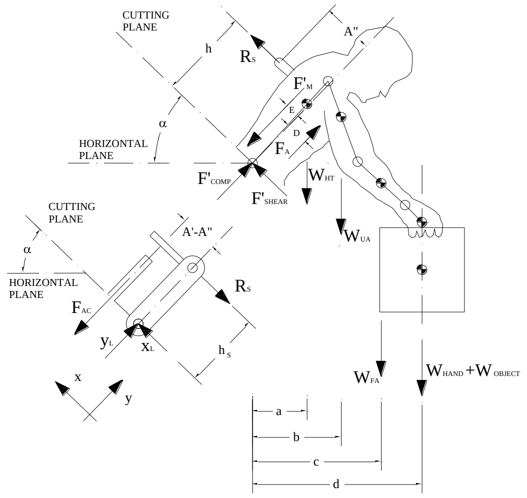

Figure 8. Free body diagrams of the human body and the device.

C. Functional Design

Using the multibody model, it was possible to identify the definitive kinematic architecture of the device. Based on the conceptual idea exposed, it was possible to define the dimensions of the segment device to obtain a correct movement for lifting a load. The position of the hinges and the positioning of the device on the user's body have been defined. Again, the user was considered to be 176 cm tall and to weigh 75 kg. The movement of the device during lifting was studied, and it was possible to define the positions of its attachments. The technical specifications of the actuator have therefore been defined. This is linear and takes a length ranging from the maximum extension of 290 mm to the minimum extension of 230 mm. From this model, it was possible to derive all the structural loads on the different segments of the device. The considered load is 25 kg.

The designed device is able to reduce the forces acting on the L5-S1, F_{COMP}, F_{SHEAR}, and the force by the back muscles F_M.

Figs. 10–12 show the reduction of the forces of interest.

A decrease of the forces by 35% on the L5-S1 joint and by 43% on the back muscles can be noted at the beginning of the lift.

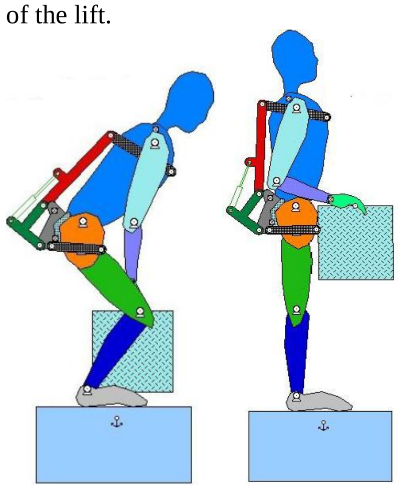

Figure 9. Multibody model for the functional design of the ausilium.

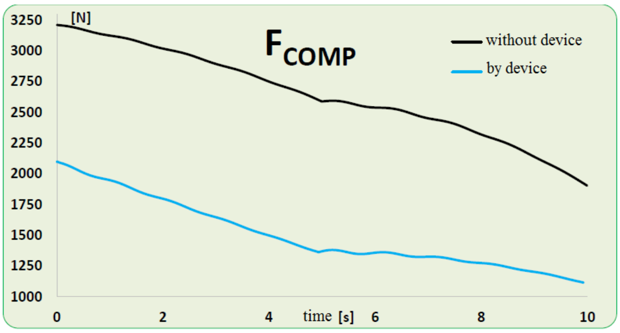

Figure 10. Reduction of the compression force on L5-S1.

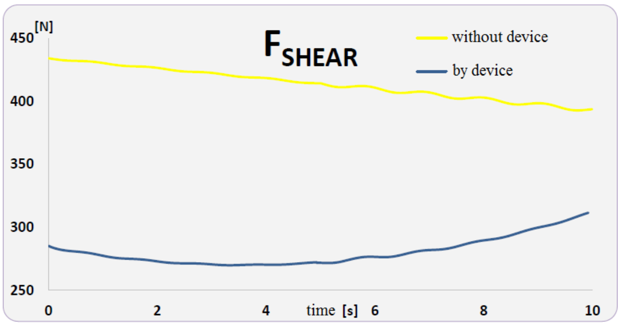

Figure 11. Reduction of the shear force on L5-S1.

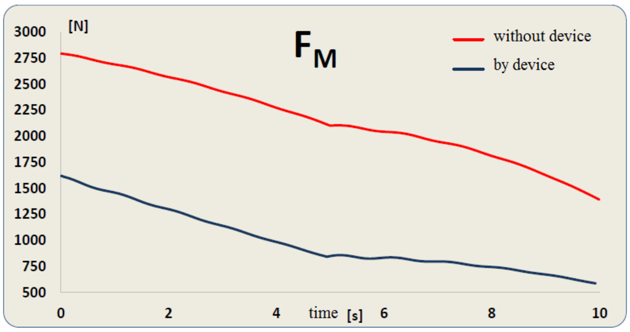

Figure 12. Reduction of the force of the back muscles.

D. Actuators

With regard to the actuators, it was decided to use pneumatic muscles. This is because they are easy-to-assemble actuators, with a good ratio of developed force to weight, and they are economic and compliant [12]. In particular, McKibben muscles are used [13]. Pneumatic muscles of this type have a force versus shortening characteristic, which linearly decreases with shortening for a given level of pressure. In the simulations, a feature of this type was considered, where preliminary investigations revealed that it can be satisfied by two parallel McKibben muscles with dimensions of about 290 mm length and 30 mm diameter at rest. In particular, through the multibody model of the exoskeleton-user system, the characteristic required for the actuator system was determined and this was compared with the characteristic of the selected muscles.

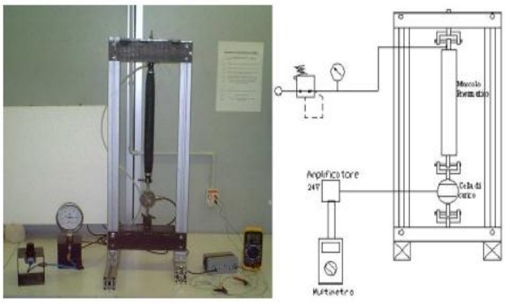

Figure 13. Experimental setup for the characterization of the pneumatic muscle.

The chosen muscle has been characterized experimentally in order to derive its performance. By means of an experimental test bench instrumented with a load cell, it was possible to detect the static characteristics. For a given percentage shortening, the strength that can be developed at the different pressures was detected. Figure 13 shows test bench and Fig. 14 shows the characteristic of the considered pneumatic muscle.

In Fig. 15, a comparison between the characteristic required by the device in the simulations and the characteristic of the chosen muscle is presented. It can be seen how the request of the device can be satisfied by two muscles in parallel with a 2.5-bar pressure.

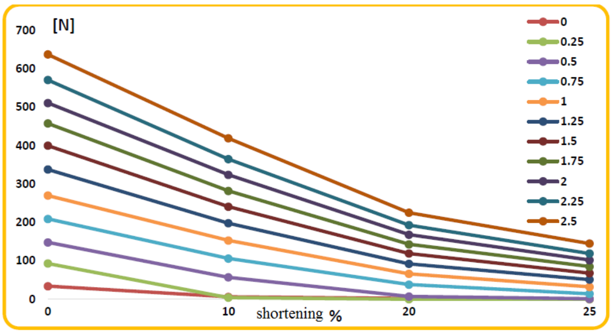

Figure 14. Characteristics of the pneumatic muscle.

E. Detailing and Structural Design

Once the functional project is completed and the sizing of the actuators is carried out, all the details necessary for the realization are defined. All structural bearing components or accessories, such as the vertebrae, the pins, and the plates, have been sized.

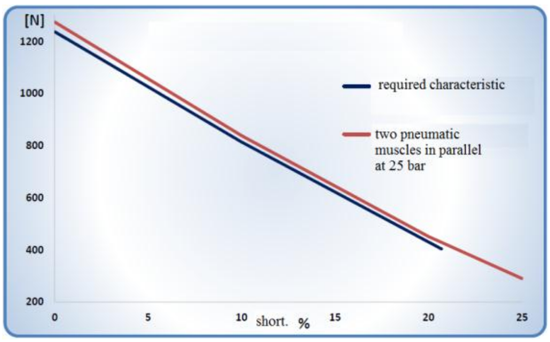

Figure 15. The required characteristic and the characteristic of two pneumatic muscles at 2.5 bar working in a parallel configuration.

Sizing was done by hand and using FEM models. Some parts of the human body interface have been designed and sized to be made of plastic by rapid prototyping. Fixation straps were provided in the upper part of the trunk and in the pelvis. Figure 16 shows an image of the 3D CAD model of the whole device.

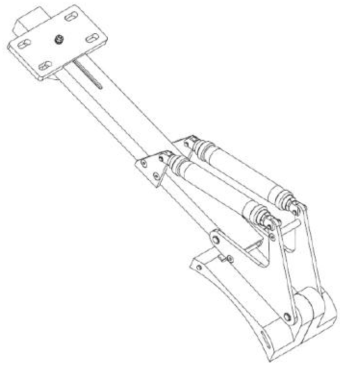

Figure 16. CAD model of the developed device.

F. Control

Regarding the control, it was decided to use a system already experimented on by the authors, which gave good results on a lower limb orthosis to assist in the movements to get up and sit down [14]. This is a control system that involves the use of a sensor to detect the user's intention and a simple algorithm able to command the device with a force proportional to the user's intention. This way, the user becomes a central element of the control system, defines the speed references, and checks the evolution of the system, while the device applies force for movements. The man–machine system thus applies the control of the mechanical impedance on itself, which is what normally happens in the movements of the animals. The result is an intuitive control system that can be applied immediately without any training on the user's part. In other projects, the authors used airbag sensors or myoelectric sensors for this purpose. In this case, myoelectric sensors (EMG) seem more suitable if the muscles of the back are to be used for the command. Figure 17 shows a signal conditioning unit developed by the authors and one MMG electrode for the user's intention detection. The control system makes use of a controller Arduino. Other systems for controlling the device are currently being studied, in particular, one that makes use of instrumented gloves for gripping the load to be lifted and others that use accelerometers for the detection of muscular activity [15] or also low-frequency or piezoelectric accelerometers for detecting the attitude of the trunk of the user [16].

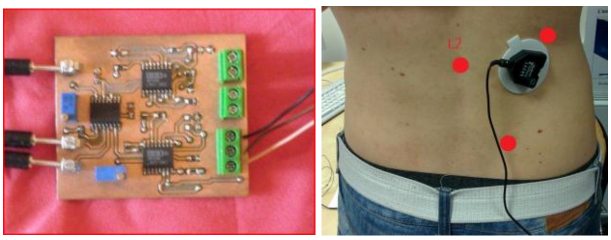

Figure 17. EMG signal conditioning board and electrode.

IV. First Experimental Test

Figure 18 shows the built device. The evaluations concerned, first of all, the wearability and the mobility that the worn device leaves to the user. Others have concerned the ability to help. The latter was evaluated considering the device through the use of elastic elements mounted instead of the actuators for time reasons. Work is progressing and the results of the active device operation will be documented in the future.

First, rubber elements were chosen and experimentally characterized. The rigidity of the elastic elements was detected by a simple experimental bench. Subsequently, estimates have been made of the capacity of the device to assist back movements. Two rubber elements were considered to work in parallel with a stiffness of 0.6 N/mm, each for an equivalent stiffness of 1.2 N/mm.

Figure 18. The exoskeleton for the back.

The two rubber spring elements have a rest length of 145 mm. The springs mounted on the device assume a length with the maximum extension of the back equal to 240 mm. So, the springs apply a total force of 114 N. When the utilizer has his/her back totally inflected, the length of the springs becomes 290 mm, thus providing a total force of 174 N. By simple calculation, it is possible to estimate the torque provided by the exoskeleton to be 18.2 Nm. During the lifting of a 25 kg load, the user has to provide a torque of 143.3 Nm at the L5-S1 joint level. It derives the device in the tested conditions provides the 13% of the total requested torque. The exoskeleton was tested in this configuration (Fig. 19).

Figure 19. The test of the device in the passive configuration.

V. Discussion

During the tests, the user could detect the following aspects:

- The device is quite light.
- It is possible to wear it in an easy way.
- The device provides the ability to freely move and also to sit down without any annoyance.
- Even if in a limited way because of the passive behavior, in the testing conditions, the device gives tangible help into the extension of the back.

Some problems may arise due to the imperfect anchoring of the lower structural part of the device to the user's body at the hip bone level. In fact, this part has a limited extension, and acting on a soft part of the body could be subject to rotations that could compromise the efficiency of the device, but this will be experimentally tested in future activities.

VI. Conclusions

In this paper, the development of an active exoskeleton was presented to provide back assistance in lifting movements. The device has a structure formed by two segments articulated by means of a hinge and is positioned behind the back, parallel to it, and is fixed to the user's body by means of straps. It is equipped with a muscular-pneumatic actuator and a control system that makes use of myoelectrical sensors able to detect the intention of the user. Preliminary tests have been carried out, which have shown that the device is easily wearable, allows good mobility for the user, and allows sitting. Tests were made with the passive device (i.e., with springs instead of actuators). Tangible help in the movements for the extension of the back has been achieved. In the future, there will be a complete experimental characterization of the device in order to quantify the performance.

VII. Acknowledgments

Our thanks are due to Mr. Davide Cimini and to Mr. Paolo Cetra for the support in the experimental activity given to this project.

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