Active Orthosis for Rehabilitation and Passive Exercise
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Abstract

Starting from an overview of nowadays adopted orthoses and from a healthy man gait analysis, the design characteristics of a new active orthosis had been defined. A particular attention had been paid to limit the device complexity and to its cosmetics. To supply the necessary power for the knee flexion and extension pneumatic actuators had been introduced. In this paper two different solutions of the cylinders forces transmission systems are proposed.

1 Introduction

An effective project for a system which helps paraplegics to walk requires a study of their disability characteristics and an analysis of both physiological and psychological problems. From a pathological standpoint there is a close link between the lesion level and the compromised functions [1], whereas the psychological problems vary with each subject. They are nevertheless very important and can determine or not the success of the system [2].

The selection of the appropriate ambulatory aid is linked to the stage of rehabilitation and to the type of pathological damage; in this paper a particular attention to the orthoses is paid. Over time passive orthoses have evolved considerably to the extent of having easily wearable structures with reduced weights and dimensions. However patients, due to the energy costs and the gait obtainable, except in rare cases, abandon them. The aim of the research is an orthosis which, even satisfying wearability requirements, enables a gait similar to the physiological one and reduces energy consumption. To this aim, an active orthosis capable of supplying mechanical energy is presented.

2 Ortheses

The numerous models of orthoses, although very different, base themselves on a common operating principle: to provide a support structure which, stiffening the lower limbs, prevents the collapse of the patient.
All types of orthosis are highly personalised and are constructed or adapted to the patient both in order to account for the different anthropometric measures, and to adapt to the pathologies congenital to paralysis (clubfoot, flexion paralysis, extension paralysis etc.). The orthoses which are here described are divided into passive, semi-active and active.

2.1 Passive Ortheses

Passive orthoses were the first to be developed and technological progress in this field has led to devices which can be used by patients with high lesion levels. The gait pattern, although varying with the orthosis type, involves the use of the muscles which can still be controlled to advance the swinging leg. Walking aids, such as crutches and deambulators, are necessary to maintain balance during walking, and in some cases, also for orthostatism. The following is a brief orthoses overview, beginning with the simpler models up to the more recent ones.

The Vannini-Rizzoli stabilisers were studied with the aim of giving walking independence to paraplegics using an easy-to-wear and sufficiently aesthetically pleasing device. They are leg-foot orthoses that allow to regain the upright position free of supports, and to permit walking using crutches or walking frame.

The orthoses A.F.O. (Ankle-Foot-Orthosis), K.A.F.O. (Knee-Ankle-Foot-Orthosis) and H.K.A.F.O. (Hip-Knee-Ankle-Foot-Orthosis) have a support capacity which, depending on the typology, goes from the foot up to the thigh of the limb in question, causing the articulations stiffening. In Fig. 1 a scheme of the orthosis known as Craig-Scott is shown. The structure of the support is very similar to the other K.A.F.O., the two pairs of aluminium hoops are fixed to the paralysed limb using Velcro strips or belts.

The HGO (Hip Guidance Orthosis) [3], known as Parawalker and shown in Fig. 2, represents a considerable step forward with respects to the thigh-leg-foot models examined previously, permitting ambulation to patients with lesions at a high level, up to T1.

The rigid metal structure is obtained by assembling steel components and it can only be worn on the clothes.

With the Parawalker the upright position is maintained without manual supports, walking however requires stabilising aids.

The ambulatory technique initially requires the patient to lean laterally, pushing on a crutches, while the other acts as a simple support. As a result of this action, one leg is lifted from the ground and, owing to the force of gravity, goes forward with a pendulum movement.

The contemporaneous extension of the trunk and the consequent forward movement of the
pelvis, immediately followed by the repositioning of the crutch, concludes the step.

Recently a new kind of Parapodium orthosis had been presented in [4]. It consists of two parallel linkages and feet rest on a small platform attached to the links. During gait the feet platforms are all the time, even during the swinging phase, parallel to the floor.

The RGO (Recipocating Gait Orthosis) represents a further step and, thanks to the innovative introduction of reciprocating cables, it offers a more natural gait, limiting greatly the lateral movements which have to be performed to achieve displacement in the previous models. Its structure (Fig. 3), similar to that of the Parawalker’s, is characterised by valves made in thermoplastic material and linked to the side metal segments. The knee hinge can be unlocked by a rear bridge. The hip articulations are connected by a pair of reciprocating cables. When the subject pushes with the upper limbs on the stabilising aid, he carries out a propulsion of the pelvis and hence a propulsion of the support hip. As he does this he transmits to the opposite side, by mean of the two rear cables, a force which causes the flexion of the opposite hip and, therefore, the forward movement of the entire lower limb. Ambulation requires the use of stabilisers, while the upright position can be maintained without any supports.

The ARGO (Advanced Reciprocating Gait Orthosis) [5], is very similar in structure to the RGO, so it does not substitute the preceding one, but represents a possible alternative. The two rear cables are replaced by just one and a couple of gas springs are introduced to slow down the patient while sitting down, and push him while standing up, reducing fatigue during these phases.

### 2.2 Semi-active orthoses or hybrid systems

Semi-active orthoses consist of an exhoscheletron, usually a traditional passive orthoses, and a functional electrostimulator which forces the contraction of the paraplegic’s muscles using electrical impulses [6]. Functional electrostimulation (FES), which can in theory be applied to all muscles, permits to reduce the energetic cost in the use of traditionally passive orthoses.

### 2.3 Active Ortheses

Active orthoses are a natural progression from passive ones and concentrate on improving the paraplegic’s ambulation, making it more similar to physiological
gait. This result can be obtained only by increasing the degrees of freedom of the structure, in particular correspondingly to the knees and ankles joints. Active orthoses consist of an exoskeleton, which supports the paraplegic and provides enough rigidity to the lower limbs, and some actuators, which have the double function of moving the articulations and stiffening them to guarantee the stability of the structure.

All active orthoses up until now have been developed as prototypes, and generally they are complex enough both from a realisation and a control standpoint. A problem for all the solutions is definitely linked to the energy source and the consequent working autonomy. Different versions of the active exoskeleton, gradually more evolved, were made by Vukobratovic [7]. The first used a rigid structure with one degree of freedom for each leg. In the second model the degrees of freedom for each leg were three, and finally the third prototype controls the lower limbs, and also the upper part of the trunk to guarantee the patient’s equilibrium.

Fig. 4 shows the orthosis in the third and final version. This is made of three distinct parts: the corset and the two legs. The corset wraps the patient’s body from the pelvis to the armpits, and is attached above the shoulders with a leather belt. Fourteen three-way electrovalves, which control the six pneumatic cylinders and the two diaphragm actuators for the movement of the frontal plane, are housed in a leather belt attached to the corset. The leg is made of three parts: the thigh, the calf and the foot.

Movement is controlled by an electronic system in order to perform a predetermined cycle. The limitations of this orthosis are the excessive weight, cosmetics which have been greatly compromised by the complexity of the structure, and donning and doffing time.

The Rabishong system is a master-slave type orthosis [8]. The master is worn by the rehabilitator, four encoders at the level of the hip and knee joints provide the necessary instructions to move the second orthosis, the slave one. This is worn by the patient, who is forced to copy the rehabilitator’s movements. Realisation is both pneumatic or hydraulic and electric.

Miyamoto’s orthosis [9] is made of an orthosis worn by the patient and a trolley with the control electronics and the energy supply for the actuators. The orthosis, made in plastic reinforced with carbon fibre, is articulated at the hips, the knees and the ankles, with one degree of freedom for each hinge. A polyethylene corset, fixed to the orthosis at the level of the hips, limits the movements of the trunk in the sagittal plane. The knees are also operated by electro-hydraulic actuators while the ankles only have a mechanism to prevent planter hyperflexion. For this
orthosis, a fully computer-controlled ambulation cycle was chosen, meaning that the patient had only to keep his balance. Also in this case the limitations for daily use are the weight of the orthosis and of the trolley, which has, however, always to be moved by the patient.

3 Design of the active orthosis

3.1 Orthoses requirements

From the previous description of commercial orthoses and of those existing only as prototypes, some important considerations emerge. The success of an orthosis depends on a large number of factors, not least psychological ones. In fact not only functional, but also cosmetics requirements must be satisfy. During the design is important to consider that ambulation must be attained completely independently. This means that the patient must be able to don and doff the system, sit and stand up on his own and to overcome light inclinations and small steps. Cosmetics are valuation parameters concerned with the emotive area of the patient, and are a compromise of various factors: the gait, which should be as close as possible to a physiological one; the possibility of disguising the presence of the prosthesis under the clothes; the hindrance caused by devices which have to ensure stability; the aesthetical design, the propulsion systems and all the other apparatus used by the patient. However the problems relating to aesthetics can be a little overlooked when the intended use of the orthosis is strictly physiotherapeutic and as such will not be worn by the patient outside a hospital or rehabilitation centres.

Finally, both the economic and the energetic costs have to be considered. From the standpoint of the last it is important that the ambulation does not require excessive efforts and that the speed must be acceptable. When designing a high performance active orthesis a compromise between the system’s complexity, resulting from the number of degrees of freedom controlled, and its simplicity, which can reduce the problem of the patient’s psychological refusal, has to be found.

3.2 Technical Specifications

The choice of the number of degrees of freedom to be controlled was done by analysing kinematics angles, forces and torque trends developed during physiological walk. To read and visualise gait analysis data Gaitlab software [10] was used; this other than plotting the various kinematics and dynamic quantity trends, animates also the human gait biomechanics in the anthropometric planes.

Different authors have analysed human gait parameters. Saunders [11] describes the main walking characteristics, which are an optimisation of the gait cycle, as having the aim of reducing the gravity centre excursion and hence the physical effort. The hip flexion enables an alternate forward motion. Even if this motion can be substituted by a pelvis movement, the hip flexion and extension enables a more fluid and efficient walk. When traditional orthoses are used, both the
trunk movement and the hip extension are obtained by pushing on the crutches and then using the back muscles to extend the trunk; with reciprocating orthoses, the extension of the trunk implies a flexion of the opposite limb. In case of paralysed hip abductor muscles, an external moment to prevent the collapse of the pelvis during the supporting phase has to be applied. This can be obtained using an actuator or a mechanical block of the pelvis. The second solution enables to create a simpler system. As can be seen in Fig. 6, obtained using Gaitlab, the flexion of the knee is relevant both during the support phase, and during the swinging phase in which a shortening of the lower limb avoids foot contact with the ground.

In absence of the knee joints the contact with the ground can be avoided by lifting the limb onto the support foot, or lifting the pelvis to the level of the swinging limb followed by a circular trajectory of the foot to the outside. These methods are, however, less efficient from an energetic standpoint and they involve unnatural movements. Finally, the knee flexion during the swinging phase enables a pelvis lowering, minimising the vertical excursion of the gravity centre during the step.

With regards to the ankle articulation, in the physiological gait during the swinging phase, a light back flexion of the ankle prevents the foot from touching the ground. This movement can be recuperated when this joint is absent, via a greater flexion of the knee. The ankle articulation can be omitted, while the flexo-extension of the hip and, above all, of the knee must be present. The trends of the hip and the knee joints torque, obtained once again with Gaitlab, are shown respectively in Figs. 6a and 6b.

With regards to the hip the greatest moment during the flexion is about 75 Nm immediately after the heel strike of the opposite limb on the ground, while the maximum moment in extension is at the end of the swinging phase and is equal to -50 Nm. Physiological angles for the hip vary between ±20°.

The knee flexo-extension moments are of less absolute value, in extension the maximum torque applied is 15 Nm in the final part of the swinging phase, and about -40 Nm in flexion in the first part of the support phase. The values shown above are the definitive technical specifications relative to the movements of the hip and the knee joints.
3.3 Design

Two possible solutions were examined. The first consists of a porpoise built exoskeleton; the second uses a reciprocating gait orthosis, on which the movement of the knee had to be introduced. Whereas in the first case there is the advantage of having a structure specific to the aim, in the second case the orthosis is already engineered. Furthermore, the presence of the reciprocating cable enables not to actively move the hip, leaving up to the patient to provide the propulsive energy for the forward motion of the limb. With regards to psychological acceptance, this is a structure which most paraplegics already use and know. One final advantage is linked to the possibility of a modular realisation of the knee, in fact being able to adapt the knee actuation to a commercial orthosis a reduction of the economical cost may be achieved. Among the models with a reciprocating gait, RGO and ARGO, the latter was chosen as it had some improvements, particularly with regards to the reciprocating cable. The hip articulation of the ARGO has a range of ±20°; and it guarantees the seating of the patient with hip and the knee joints angles of 90°. In the prototype version to activate knee joints a pneumatic solution had been chosen,
excluding both the electric one for weight reasons, and the hydraulic one for supply problems. Two different solutions were realised, with a ‘strut’ and a ‘muscular’ cylinder. The first, which can be seen in Fig. 7, is made of a double-acting pneumatic cylinder, the ends of which (the rear plate and the rod) are connected with hinges to the calf and thigh segments; in this way the piston force creates the desired torque.

The second solution comes from the need to reduce to a minimum the system’s dimensions. With the previous solution, the cylinder occupies the third side of a triangle made by the thigh and calf segments. This triangle increases its area as the knee flexion increases, making it difficult to wear and having to keep large masses moving with consequent problems.

To develop the ‘muscular’ cylinder system, the human body was used for inspiration, trying to simulate mechanically the function of the tendons. In the human locomotory system, the moment required to the leg flexo-extension leg comes from the tendons tension caused by the muscles contraction. In the exoskeleton, however, the piston tends a chain which, on its turn, transmits a moment to the knee joint. The function which in the human body is obtained from the rotula, is simulated using a toothed wheel similar to that on a bicycle. Two other toothed wheels complete the system carrying out respectively the jobs of chain tightening and alignment, as can easily be seen in Fig. 8. The double-acting actuator, with a passing rod, is fixed to the calf segment.

4 Experimental Tests

A first series of tests were carried out with the orthesis not worn by a patient, to make calibrations and to define the optimum step cycle. With this aim the device was connected to a support structure and it was equipped with an actuator able to generate artificially the hip joint torque necessary to walk. In the photograph of Fig. 9 the test set-up, with the ‘muscular’ cylinder solution, can be seen.

The hip was activated, during this test phase, by a rotary pneumatic actuator. Only the articulation of the left hip was connected to the motor, as the reciprocating cable ensures the movement of the opposite articulation. The torque generated by the rotary motor, whose shaft was in axis with the hinge to be moved, was transmitted to the femoral section using a flanged plate and a lever with a prismatic guide.

Electrovalves for the command of the actuators, and the ambulation control system made with a PLC, were mounted on the support structure. The action exchanged with the ground was ensured by the friction in the soles of the rubber boots worn by the orthesis and fixed to the calf upright and the sole.
The rigidity relative to the tarsic articulation was simulated with air chambers inserted inside the boots. The tests made, both with the ‘strut’ and the ‘muscular’ cylinder solutions, demonstrated the viability of the device and permitted to define angles and optimum pressure values for the gait. Furthermore, the tests highlighted the notable obstruction of the ‘strut’ cylinder solution, which was then abandoned.

The photograph of Fig. 10 shows the active orthosis from side and front views in the actual execution of the prototype. The command electrovalves of the ‘muscular’ actuators are connected to the rear tube of the ARGO, and receive both a pneumatic supply and the electrical command signals from a fixed control pulpit.

Experimental tests are currently carried out on ‘wearers’, both healthy and paraplegic.
5 Conclusion

In this paper, research carried out until now on an active, pneumatically operated, system has been presented. The realisation of a device and the preliminary experimental tests made have shown the project’s feasibility. Further research is directed to sensorise the position of the hips and the knees joints and towards to optimise the gait pattern. Furthermore, a command interface of the cycle directly mounted on the crutches will be realised, with the aim of allowing the step cycle to be completely controlled by the patient. Finally a wide series of tests, using paraplegic wearers with different lesions will be carried out to completely defining the orthesis’s performance.

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References