Mechanized Knee Flexion in Lower Limb Orthoses: A Prototype and Tests

Marko Ackermann
University of Sao Paulo - PMR
ackermann@mechb-uni.stuttgart.de

Fabio Gagliardi Cozman
University of Sao Paulo - PMR
Av. Prof. Mello Moraes, 2231
05508-900
fgcozman@usp.br

Abstract. *Orthoses for lower members are often plagued by a high rejection rate, due both to the lack of aesthetics of the resulting gait and to the excessive effort demanded from the user of the orthosis. These factors are caused to, a great extent, by the fact that orthoses force the whole gait to be performed with the knee articulation in a fixed position. This work presents a light and compact device, with a low energy consumption, that can improve the performance and the gait aesthetics for knee-ankle-foot orthoses (known as KAFO). To do so, we explore the natural dynamics of the lower members, adding a spring to the standard orthosis system. We also add control circuitry, a small electric motor, sensors and microprocessors to a standard orthosis, so that we can drive the whole gait in a relatively natural manner. We have designed the system through a large set of simulations and tests with a patient at the AACD (Associacao de Assistencia a Crianca Deficiente --- Association for the Assistance to the Handicapped Children). We describe the design of the device and the tests we have conducted. The work described in the paper is the natural sequence of previous publications in COBEM 2001 and CONEM 2002, which described partial steps in the project; the present paper offers a complete implementation of our ideas.*

Keywords*.* *Lower limb orthoses, knee flexion, biomechanics.*

1. Introduction

KAFOs (knee-ankle-foot orthoses) are prescribed to paraplegic patients with low level spinal cord injury (level T10 to T12) with good control of the trunk muscles. They are usually designed in such a way as to maintain the knee joint locked in its extended position. In Brazil, according to approximate data obtained from AACD (Associacao de Assistencia a Crianca Deficiente --- Association for the Assistance to the Handicapped Children), 8000 people suffer spinal cord injury every year. From this population, about 1500 could use lower limb orthoses; however, only about 5 % do use them. The most important reasons for such a rejection rate are the lack of aesthetics of the resulting gait and the excessive effort demanded from the user of the orthoses to walk. These factors are caused to a great extent by the fact that orthoses force the whole gait to be performed with the knee articulation in a fixed position. The literature presented in Section 2.1 suggests the importance of improving the orthotical gait by imp lementing automatic knee flexion.

This article describes the development of a new device capable of providing knee flexion for orthoses during the swing phase of the gait. In order to achieve a light and energy saving system we explore the natural dynamic of the gait, which is briefly explained in Section 2.2. We propose a system, described in Section 3, whose main actuator is a spring that flexes the knee joint at the beginning of the swing phase. After the knee flexion, the spring is decoupled and the knee reaches its extended position in an entirely passive way. As far as we know, this is an original strategy that leads to an energy saving gait similar to a normal one.

A simple model of the lower limb, presented in Section 4.1, was used in order to design the device and to study the sensitivity of the system behavior (discussed in Section 4.2). A prototype, described in Section 5, was built and tests with a real patient were performed (presented in Section 6). The analysis performed through the simulations and the tests with the patient showed the potential of our strategy in improving traditional KAFOs, and showed the necessity of further improvements of the device (as discussed in Section 7).

This work is the natural sequence of previous publications in COBEM 2001 (Ackermann; Cozman; Dias, 2001) and CONEM 2002 (Ackermann; Cozman, 2002), which described partial steps in the project. The present paper offers a complete implementation of our ideas.

2. Bibliographic review

2.1. Importance of knee flexion during the gait

Many authors have studied knee flexion as an improvement to gait quality. Indeed, Abdulhadi; Kerrigan and LaRaia (1996) showed an increase of 20% in oxygen consumption when the knee joint of a person with normal gait is immobilized. This fact suggests that orthoses with locked knee induce energy losses during the gait of a paraplegic. Greene and Granat (2000) showed that a knee flexion associated to an ankle flexion during the swing phase of the gait of a paraplegic person can lead to a reduction of the effort required from the user. Kaufman et al. (1996) showed through experiments that the use of a KAFO that allows knee flexion during the swing phase of the gait can reduce the energy consumption of the gait compared with a KAFO that keeps the knee joint locked in its extended position. Allard et al. (1981) showed that the vertical oscillation amplitude of the center of mass of a person who uses a conventional KAFO increases 65% compared to the situation in which no orthoses are used. Such results show the inefficiency of the gait when a conventional orthosis with locked knee is used.

Ackermann; Cozman and Dias (2001) showed through a very simple anthropometric study that a knee flexion superior to about 40 degrees can potentially reduce the hip joint raise during the swing phase in relation to the case in which the knee joint is maintained extended. This reduction leads to a decrease in user effort. The knee flexion is not only functionally profitable; it also provides a more normal-looking gait that surely increases the aesthetics of the gait and, as a result, can substantially reduce the rejection rate of the orthoses (Ackermann, 2002).

2.2. Systems for automated gait

In order to promote knee flexion in a simple and energy-saving way, this work employs the natural dynamics of the human gait. In the following paragraphs we summarize previous work that used this concept and that inspired the present work.

The work of Gharooni; Heller and Tokhi (2000) is the most similar to ours. They developed a system based on the use of a spring to initiate the swing phase. To promote the knee extension after the knee flexion, they use FES (Functional Electrical Stimulation) to stimulate the knee extensor muscles against the action of the spring.

The classical work of McGeer (1990) explored and developed the so called *Passive Dynamic Walking*. He demonstrated mathematically that it is possible to build a two-legged machine capable of walking on a shallow slope in a passive way, with energy provided solely by gravity forces, without any active control. Camp (1997) demonstrated that the implementation of a simple open-loop controlled actuation is enough to promote a stable gait on an even surface.

According to Pratt (2000), very simple or even open-loop control strategies, that explore the natural dynamic of the human gait, applied to prosthetical, orthotical or robotic systems, can lead to energy-saving gait patterns that are very similar to normal ones.

3. A new system to promote knee flexion in Knee-Ankle-Foot Orthoses

Here we propose a new solution for promoting knee flexion on KAFOs. The solution employs the natural dynamics of the gait by implementing the following strategies:

– the required energy to flex the knee at the beginning of the gait is given by a spring (a low impedance element), and

– the necessary knee extension at the end of the gait is reached in an entirely passive way by simply decoupling the spring from the joint after a specific knee flexion.

As far as we know, this is an original strategy that leads to a very simple solution, because it does not require any closed-loop control nor any type of actuator or locking device at the hip joint. The use of a motor directly coupled to the joint to flex the knee would substantially increase the energy consumption by avoiding the exploration of the natural dynamics, and would require more power from the batteries. The use of FES (Functional Electrical Stimulation) to extend the knee at the end of the swing phase, as done by Gharooni; Heller and Tokhi (2000), would considerably complicate the system.

To achieve an effective swing phase, a knee flexion must be accompanied by a hip flexion. This is achieved as a reflex of the knee flexion (Gharooni; Heller; Tokhi, 2000), in such a way that an actuator coupled to the hip joint is unnecessary. Statically, the flexion of the knee leads to the flexion of the hip so that the potential energy of the entire lower limb is minimized. Dynamically, the effect of the acceleration of the lower segment of the lower limb also contributes to the flexion of the hip joint. According to simulations and to the tests with a patient, the generated hip flexion is enough to provide a satisfactory swing phase, although its value at the moment of extension of the knee is substantially smaller than that achieved during a normal swing phase.

The support phase happens with the knee joint locked in its extended position. This strategy provides gait stability during the support phase of the gait and avoids the need for an actuator to maintain the lower limb extended. The user unlocks the knee joint to initiate the swing phase; the knee is then automatically locked at the end of the swing phase as soon as it reaches its extended position. The storage of energy in the spring is made by a small motor during the whole period in which the knee joint is locked.

A drawing that explains the proposed solution is shown in Fig. (1) followed by an explanation of each one of the events of the gait cycle. The figure shows the realization of a swing phase whose kinematics is similar to that of the normal gait.

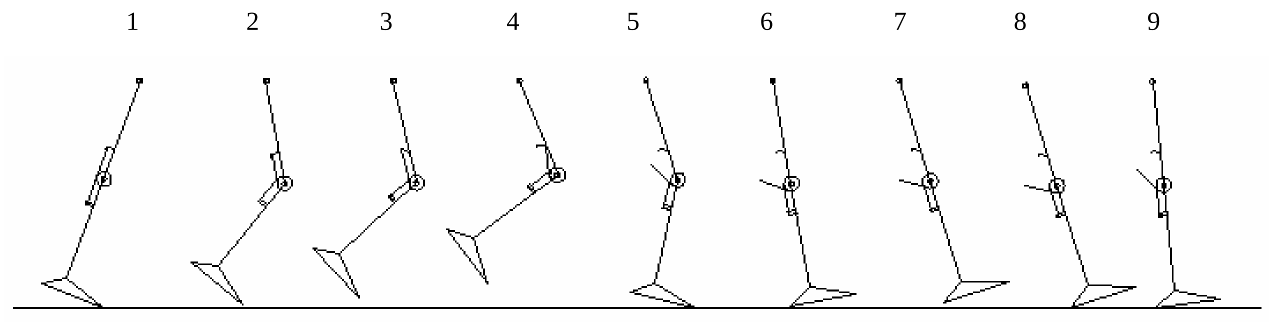

Figure 1 - Drawing of the gait cycle generated by the system developed.

1- The knee joint is unlocked;

2- The knee flexes under the effect of the spring. The hip flexes as a reflex of the knee flexion;

3- As soon as the knee reaches a specific knee flexion angle, the spring is decoupled and stops to exert moment on the knee joint;

4- As an effect of the kinetic energy acquired by the lower limb before the decoupling of the spring, the knee and the hip joints keep flexing until them, under the decelerating effect of the gravity, reach their maximum flexion level;

5- The knee and hip joints extend under the effect of gravity;

6- The knee joint reaches its totally extended position, while the hip joint is still flexed. The knee joint is automatically locked, as soon as it reaches its totally extended position. A small motor begins to store energy on the spring. This storage takes place during the whole period, in which the knee joint is locked in its extended position.

7- As an effect of the momentum of the lower segment of the lower limb, the entire limb advances after the knee locking (the hip joint flexes).

8- Beginning of the support phase (heel strike).

9- Support phase.

4. Dynamics of the lower limb system

In order to study the sensitivity of the lower limb behavior to several factors, such as friction at the joints and weight and height of the user, as well as to determine the spring features, we used a simple model of the lower limb valid only for the swing phase of the gait. The differential equations of the movement were calculated and we simulated the lower limb under several scenarios.

The model employed consists of a double pendulum on the sagittal plane with two degrees of freedom (rotation of the knee and of the hip joints) as showed schematically on Fig. (2). The hip joint is fixed. The foot was considered fixed in relation to the lower segment of the lower limb, which simulates the situation in traditional lower limb orthoses. Average mass distribution features and dimensions of the lower limb segments in relation to the height and the weight of an individual were found in (Winter, 1990). Average friction at the joints were found in (Stein et al., 1996). Average stiffness at the joints, which depends on the angle of flexion, were found in (Riener; Edrich, 1999) and in (Audu; Davy, 1985). *q1* and *q2* are respectively the angle between the upper segment of the lower limb and the vertical and the angle between the lower and the upper segments of the lower limb. A detailed explanation of the simplifications adopted as well as an explanation of the models of stiffness and friction used can be found in (Ackermann, 2002).

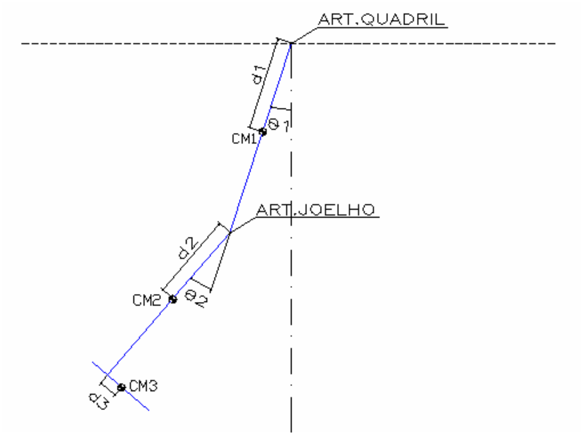

Figure 2 - Double pendulum model of the lower limb during the swing phase of the gait with two degrees of freedom.

The differential equations were obtained trough the Lagrange equations, which are shown in (Ackermann, 2002). The equations were solved by the numerical integration method Runge-Kutta implemented on the software Matlab^{TM}.

We performed several simulations to design the spring so as to achieve a relatively normal gait and to minimize the necessary hip rise; an extensive discussion of results can be found in (Ackermann, 2002). The most important conclusions are:

– There is a spring that leads to lower limb behavior that is similar to the normal gait;
– the needed hip rise during the swing phase is reduced in comparison to the situation where the gait is performed with the knee locked in its extended position;
– the sensitivity of the behavior of the lower limb to variations on weight and height is relatively small, given the average anthropometric relations presented by Winter (1990) and
– the sensitivity of the system to friction and to passive stiffness of the joints is substantial.

Using simulations as a guide, we designed a linear spring with a stiffness (between 14 and 17 Nm/rd) that leads to a lower limb behavior similar to the normal one. Moreover, the designed spring maintains the hip rise as small as possible. We also observed that the spring should be decoupled when the knee flexion angle reaches about 60 degrees.

5. Prototype

We have built a prototype that is depicted in Fig. (3) and is composed of four systems:

1-) The spring system (including a steel spring), and the mechanical structure of aluminum, which permits the coupling of the prototype to traditional orthoses produced by AACD.

2-) The locking system, which keeps the knee joint locked and extended during the support phase of the gait and free during the swing phase of the gait. We developed a new locking system, which is very light, energy saving and quite compact (Ackermann; Cozman, 2002). It contains a moving part, which actually locks the knee joint, and a small solenoid commanded by the control system capable of unlocking the joint by moving the moving part. The locking of the joint occurs automatically, under the effect of gravity, as soon as the joint reaches its extended position. A small burst of energy into the solenoid is enough to unlock the joint. Its total weight is about 200 g.

3-) The spring storage system, whose function is to store energy on the spring during the period in which the knee joint is locked. This system is composed of a small 12 V DC motor, a 43:1 speed reducer, a pulley and a steel cable connected to the free stick of the spring and to the pulley. The system is capable of storing energy in the spring in about 1.3 s. The motor and reductor set weights about 180 g.

4-) The control system, which controls the spring storage system and the locking system by means of the information obtained from the sensors. It contains a microprocessor (PIC16F84). The microprocessor receives information from two sensors (locking knee and coupling of the spring) and from one button commanded by the user to initiate the swing phase of the gait. The microprocessor controls the solenoid, the motor and a knee locking signal to inform the user that the knee has been safely locked at the end of the swing phase.

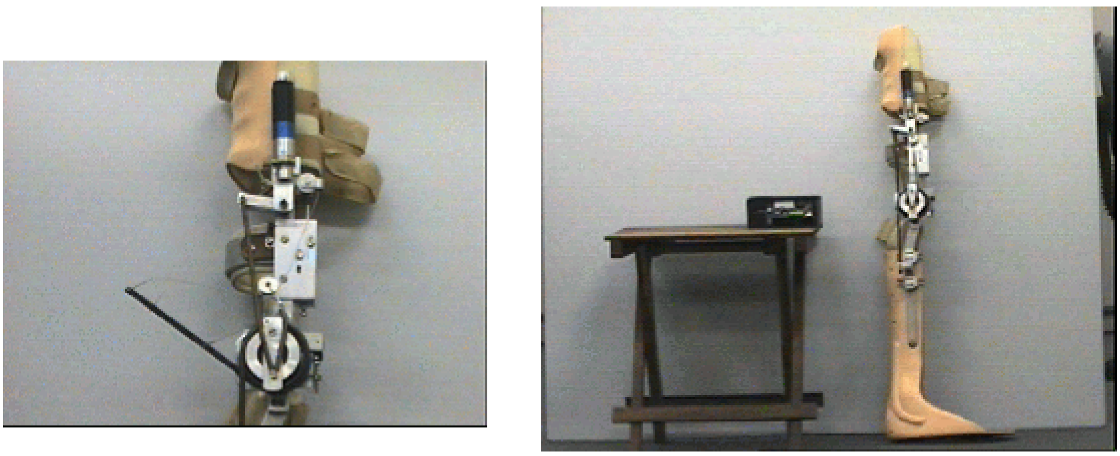

Figure 3 - Pictures of the Prototype built.

A drawing indicating the system operation is presented in Fig. (4). The numbers in the figure are explained below.

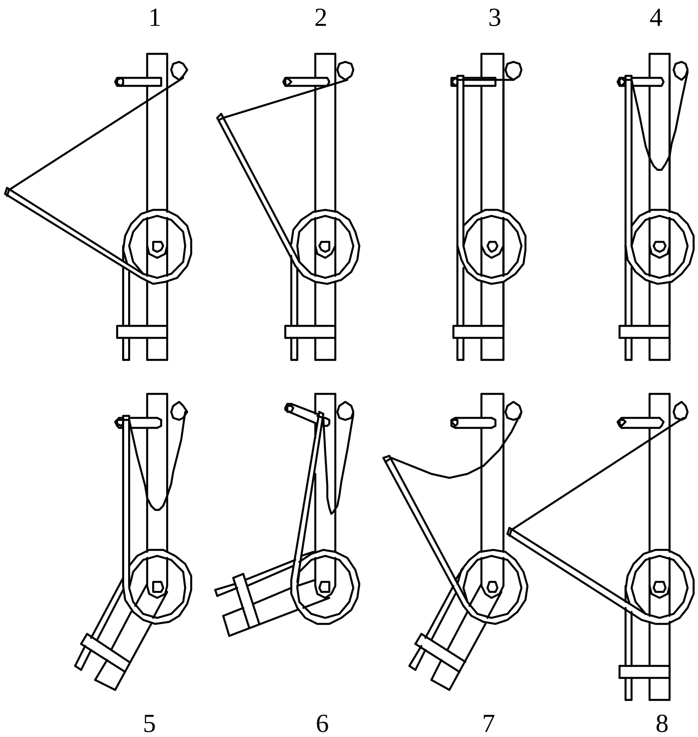

Figure 4 - Drawing indicating the system operation.

1-) As soon as the knee extends at the end of the swing phase, the joint is automatically locked by the locking system, under the effect of the gravity. The control system detects the locking event by means of an optical sensor and commands the begin of the energy storage on the spring by the small DC motor of the spring storage system. The motor is turned on by the control system and begins to pull the free stick of the helical spring by means of a steel cable.

2-) Elastic potential energy is stored on the spring by means of a small DC motor. This occurs during almost the whole period in which the knee joint is locked and extended.

3-) The free stick of the spring is mechanically coupled to the "coupler". An optic sensor detects the coupling and the control system turns off the DC motor.

4-) The control system turns on the motor in the opposite direction, so as to relax the tension in the cable. After a pre-defined period, the control system permits that the user commands the begin of the swing phase by pressing a button.

5-) As the patient commands the unlocking of the knee joint, the spring begins to transfer energy to the lower limb, flexing the knee joint.

6-) As soon as the flexion angle of the knee reaches the neutral angle of the spring, the free stick of the spring is mechanically decoupled, so as to allow the totally passive motion of the lower limb.

7-) Under the decelerating effect of the gravity, the knee begins to extend.

8-) The knee joint reaches its totally extended position and is automatically locked by the locking system.

6. Tests with a paraplegic patient

To verify the central ideas of this work, we proceeded two tests with a paraplegic patient of AACD, user of a traditional KAFO produced by AACD. The objective of the first test was to verify the effect of the spring on the behavior of the lower limb during the swing phase of the gait. This test demonstrated the viability of the new strategy proposed, and led us to identify weak points. After designing and building the complete prototype, the second test was performed. This test showed that the prototype works well, but it showed also the need for improvements.

Three pictures of the first test proceeded can be observed on Fig. (5). The first picture shows the initial position, the second shows an intermediary position of the lower limb and the third one shows the final position. Tab. (1) presents information about some events: the time and the angles reached by the hip joint (*q1*) and the knee joint (*q2*).

The tests were performed at AACD. The tested patient was 25 years old, 1.71 m, 53.7 kg, paraplegic level T12, with low spasticity and good control of the trunk muscles. He was positioned between parallel bars in such a way to simulate the use of crutches. The left lower limb was positioned on a stage, whose high was about 0.15 m, what lets the patient's right lower limb free. The right lower limb, to whose orthosis the prototype was coupled, was hold backwards (hip angle about 20 degrees) simulating the initial position of the lower limb immediately before the begin of the swing phase.

Figure 5 - Lower limb movement during the test performed with the patient.

Table 1 - Time and angles reached by the hip joint (*q1*) and the knee joint (*q2*).

Event	**Time(s)**	q**1**	q**2**
Begin	0	23^o	0^o
Spring decouple	0.2-0.233	-	60^o
$\theta 2$ maximum	0.267	-	60^o – 65^o
Knee extension	0.467	-	0^o
Max. hip flexion	0.733	-25^o	0^o

Conclusions from the tests :

First Test:
– As observed in Fig. (5), the kinematics of the swing phase promoted is similar to that of the normal gait, which corroborates the potential of this new system to improve the gait with orthoses. To illustrated this fact, we indicate the period of time comprehended between the beginning of the swing phase and the extension of the knee: this period was 0.47 s during the test and is about 0.4 s in the normal gait.
– This behavior fits well the behavior predicted by simulations;
– The knee flexion speed was substantially higher than that observed in the normal gait; this is caused by the type of spring used, which applies a high torque on the joint at the very beginning of the swing phase.
– The user is able to control the gait speed by controlling the moment in time in which the user gives commands.
– The measured length of the gait was longer than that observed when the patient walks with locked knees.

Second Test:
– The second test showed that the system designed mostly functions accordingly to specifications.
– In spite of the fact that the test conditions were very similar to that of the first test, the behavior of the lower limb observed was substantially different. The total extension of the knee in a passive way at the end of the swing phase was not observed. After analysis, we concluded that this difference occurred due to a relevant increase of the passive stiffness of the user joints. Relevant variations in the lower limb behavior due to variations in the knee joint stiffness had been previewed by the simulation performed. This increase was probably caused because the patient spent a long period (about 1 year) without using the orthosis and without attending a physiotherapy program.
– The fact described showed the importance of considering carefully the passive moments at the joints. Besides, it indicates the need for maintaining the patient enrolled in a physiotherapy program to avoid stiffness increase at the joints.

7. Conclusion

This work presented a new, compact and energy saving system that provides knee flexion for lower limb orthoses of type KAFO. The system explores the natural dynamic of the lower limb. We proposed the utilization of a spring at the knee joint, which allowed us to implement the following strategy: store energy on the spring during the whole period in which the knee joint is locked at its extended position (with low power consumption), and release energy at high rates during the beginning of the swing phase (to provide knee and hip flexion). After the knee flexion, the spring is decoupled leading to an entirely passive dynamic behavior of the lower limb, causing the total extension of the knee at the end of the swing phase.

The solution proposed is capable of providing a normal-looking kinematics to the lower limb during the swing phase of the gait, as shown through simulations and tests. The provided knee flexion during the swing phase increases the aesthetics and reduces the energy required for the user to walk. These two facts can considerably reduce the high rejection rate of lower limb orthoses.

The prototype only requires the user to indicate the moment when the swing phase must begin. The prototype is light (1.26 kg) and energy saving (a 270 g NiMH battery is capable of providing an autonomy of about 2300 steps). The battery and the control system, enclosed in a 149x97x63 mm box attached to a waistband, weigh 470 g.

In spite of the good perspectives of the new system proposed, some improvements are needed:

–The simulations and the tests with the patient indicated the high sensitivity of the kinematics of the system to joint stiffness. This fact associated with the substantial variability of this feature in the population can lead to bad performance of the system. Strategies to reduce the sensitivity to this feature are required; an idea is to apply a personalized stiffness mechanism in order to neutralize the passive stiffness of the knee joint.

–Another high sensitivity detected is in relation to friction, particularly at the knee joint. This fact indicates to the importance of minimizing the friction at the joints when the orthosis is designed and built.

–The reduction of the knee flexion speed is desirable in order to further approximate the kinematics of the lower limb to the normal one. One idea here is to use a non-linear spring with appropriate dynamics.

–Last but not least, the simultaneous flexion of the ankle and of the knee joints during the swing phase would provide a substantially further reduction of the hip movement needed, which would lead to a better performance of the device.

8. References

Abdulhadi, H.M., Kerrigan, D.C., Laraia, P.J., 1996, "Contralateral Shoe-Lift: Effect on Oxygen Cost of Walking with an Immobilized Knee", Archieves of Physical Medicine and Rehabilitation, Vol. 77, pp. 670-672.

Ackermann, M., 2002, "Sistema de acionamento e travamento para flexao mecanizada de joelho em orteses de membros inferiores", Sao Paulo, Escola Politecnica - University of Sao Paulo, Aug., 110 p. (Masters dissertation).

Ackermann, M.; Cozman, F.G.,2002, "Um sistema de travamento para a articulação do joelho de órteses de membros inferiores", In: Proceedings of the II Cogresso Nacional de Engenharia Mecanica, João Pessoa - PB, pp. 116-124.

Ackermann, M., Cozman, F.G., Dias, A.S., 2001, "Sistema para Flexão de Joelho Mecanizada em Órteses de Membros Inferiores", Proceedings of the 16th Brazilian Congress of Mechanical Engineering, Vol. 1, Uberlândia, Brazil, pp. 254-263.

Allard, P.L.; Duhaime, M.; Thiry, P.S.; Drown, G., 1981, "Use of gait stimulation in the evaluation of a spring-loaded knee joint orthosis for Duchenne muscular dystrophy patients", Medical and Biological Engineering and Computing, Vol.19, pp.165-70.

Audu, M.L., Davy, D.T., 1985, "The influence of muscle model complexity in muscleskeletal motion modeling", Journal of Biomechanical Engineering, Vol.107, pp.147-57.

Camp, J., 1997, "Powered "Passive" Dynamic Walking", Ithaca NY: The Sibley School of Mechanical and Aerospace Engineering, Cornell University, Aug. (Masters of Engineering Project Report).

Gharooni, S., Heller, B., Tokhi, M.O., 2000, "Hybrid Spring Brake Orthosis for controling hip and knee flexion in the swing phase", Research Report No. 764, Department of Automatic Control and Systems Engineering, University of Sheffield, Sheffield, UK.

Greene, P.J., Granat, M.H., 2000, "The effects of knee and ankle flexion on ground clearance in paraplegics gait", Clinical Biomechanics, Vol. 15, n. 7, pp. 536-540.

Kaufman, R.K., Irby, S.E., Mathewson, J.W., Wirta, R.W., Sutherland, D.H., 1996, "Energy-efficient Knee-Ankle Foot Orthosis: A Case Study", Journal of Prosthetics and Orthotics, Vol. 8, n. 3, pp. 79-85.

McGeer, T., 1990, "Passive Dynamic Walking", The Int. Journal of Robotics Research, 9(2):504-524.

Pratt, J.E., 2000, "Exploiting inherent robustness and natural dynamics in the control of bipedal walking robots", June, 157p, PhD Thesis – Department of Electrical Engineering and Computer Science, Massachusetts Institute of Technology.

Riener, R.; Edrich, T., 1999, "Identification of passive elastic moments in the lower extremities", Journal of Biomechanics, Vol. 32, pp. 539-44.

Stein, R.B.; Zehr, E.P.; Libiedowska, M.K.; Popovics, D.B.; Scheiner, A.; Chizeck, H.J., 1996, "Estimating mechanical parameters of leg segments in individuals with and without physical disabilities", IEEE Transactions on Rehabilitation Engineering, Vol.4, n.3, pp.201-11.

Winter, D. A., 1990, "Biomechanics and Motor Control of Human Movement", Ed. Wiley-Interscience, 2nd ed.