33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

A Pneumatically-Actuated Lower-Limb Orthosis

Sai-Kit Wu, Matthew Jordan, and Xiangrong Shen

Abstract— **Powered lower-limb orthosis is a type of wearable mechanical devices that can serve a wide variety of important biomedical purposes. Due to the constraints associated with the actuation technology, the majority of current lower-limb orthoses are either passive or tethered to external power sources, limiting the functionality of such devices. In this paper, the authors present their preliminary research results towards a fully mobile (i.e. untethered) powered lower-limb orthosis, leveraging the high power density of pneumatic actuators for the joint power generation. The design of the orthosis is presented, with the objectives of providing full assistance in the locomotion of various common locomotive modes, and generating minimum level of restriction to the wearer's daily activities. To regulate the power delivery on the joints for a natural gait assistance, a finite-state impedance controller is developed, which simulates an artificial impedance to enable an effective interaction with the wearer. Preliminary testing demonstrated that the orthosis was able to provide a natural gait and comfortable user experience in the treadmill walking experiments.**

I. Introduction

Developed as wearable mechanical devices working in parallel with the human structure, orthoses can serve a wide variety of important biomedical purposes. Specifically for the lower-limb orthoses, the applications include the gait assistance of elderly persons, and the rehabilitation of persons with lower limb pathologies. Currently, the majority of clinically prescribed lower-extremity orthoses, including ankle-foot orthoses (AFO) and knee-ankle-foot orthosis (KAFO), are energetically passive devices. In spite of the successes with these devices, the energetically passive nature poses significant limitation to their functionality. Specifically, the gait achieved through such devices is largely different from the normal gait of healthy subjects, due to the constraint applied to the joint motion. Also, the healthy joints, such as the hip joint, are required to provide a higher torque to compensate for the lack of power generation of the locked joints, increasing the metabolic energy consumption of the wearer. Furthermore, fixed kinetic characteristics provide little flexibility in adapting to different locomotive modes and environments.

Manuscript received April 15, 2011.
S-K. Wu is with the University of Alabama, Tuscaloosa, AL 35487, USA. (e-mail: swu10@ua.edu).
M. Jordan is with the University of Alabama, Tuscaloosa, AL 35487, USA (e-mail: jorda002@crimson.ua.edu).
X. Shen is with the University of Alabama, Tuscaloosa, AL 35487, USA. (corresponding author, phone: 205-348-6743; fax: 205-348-6419; e-mail: xshen@eng.ua.edu).

To address such issues, orthotic devices with active joint power delivery have been investigated by the researchers in this area. Due to the significant difficulty in providing energy supply on the go, the majority of such active devices are stationary or tethered to an energy source. These stationary or tethered devices are usually combined with a treadmill or overhead harness to provide safety protection. Note that this type of system is usually bulky and complex, and thus the application is limited to the rehabilitation training for major clinics. Furthermore, without the energy source, such system is not able to provide mobile assistance in daily activities.

Aware of the constraints with the tethered or stationary systems described above, researchers have also explored the self-powered mobile orthotic devices to serve the purposes such as the rehabilitative training of spinal cord injury patients and assistance of elderly persons. To obtain a self-powered orthosis, a key challenge is the development of a compact and powerful actuation system. In the existing works, a major approach is the series-elastic-actuator (SEA), which consists of a DC motor that drives the orthotic joint through a ball screw mechanism in series with a coil spring [1]. Though the electric motor and the corresponding energy source (battery) are easy to use, the lower power density of the DC motor results in a heavy and bulky system, which is exacerbated by the complex spring-based driving system. In addition to the SEA, hydraulic power system has also been explored [2]. However, the complex hydraulic supply increases the complexity and weight/volume of the system.

In this paper, we present the design and control of a novel pneumatically-actuated lower-limb orthosis, with the objective of exploring the use of high-power-density pneumatic power in lower-limb orthotics to reduce the weight and volumetric profile of the system. In addition, an impedance-based control approach is presented, with the purpose of achieving natural interaction with the wearer. Pneumatic actuators, in general, have a significantly higher power density than DC motors. Also, compared with hydraulic actuators, pneumatic actuators have an inherent, physically existing, and controllable compliance, which is very beneficial with respect to the interaction with the humans. More importantly, for the mobile applications, a liquid propellant-powered pneumatic supply approach was proposed by Goldfarb and his collaborators [3]. This approach provides significantly higher energetic

978-1-4244-4122-8/11/$26.00 ©2011 IEEE

characteristics by utilizing the gaseous product of the catalytic decomposition of liquid propellant to drive the pneumatic actuators. In the long term, such system will be able to provide a highly compact and powerful actuation system for self-powered orthotic devices, and enable their practical application in biomedical practices. Rather than directly construct a fully self-power version, the authors have developed a power-tethered device first, and started the experimental testing and controller development. The subsequent section describes the design of the power-tethered prototype of the orthosis.

II. Orthosis Design

The design of the proposed self-powered orthosis aims to provide actuated knee and ankle joints to assist the wearer's locomotion. Leveraging the basic structure of a standard KAFO, the orthosis is easy to don and doff, with little restriction to the wearer's daily activities. Also, with the pneumatically actuated knee and ankle joints, the orthosis is able to provide full assistance in the most common locomotive modes, including the slow and normal walking, and stair ascending and descending.

A. *Standard KAFO-based structure*

The prototype developed in this work utilizes a standard metal-frame KAFO as the basic structure. As a mature clinical product, KAFO has been demonstrated to provide effective load transfer to the human limb when restricting the joint motion. Furthermore, adapting from this mature product, the resulting powered orthosis has a better potential for the further clinical application in the long term. Among the common types of KAFO, a metal-frame device, with two upright bands extending proximally to the thigh, is chosen as the basis of the prototype (Fig. 1). In this structure, two thigh bands are used to connect the uprights, maintaining the structure integrity and allowing the joint-extensive force to be distributed over a large area on the posterior side of the thigh. Furthermore, this structure is also able to provide sufficient strength and stiffness, preventing excessive deformation under the forces applied by the actuators.

Fig. 1. The lower-limb orthosis fitted to a healthy subject.

Based on the standard KAFO, pneumatic cylinder-type actuators were attached to generate torque and power output to assist the wearer's locomotion. Note the cylinder-type actuator is chosen primarily because of its greater torque capacity in comparison with vane-type actuator. Furthermore, unlike the pneumatic muscle actuator, the cylinder-type actuator does not expand radially, and thus is free of the interference with the limb segment in the actuation. In the design of the actuation structure, a fundamental requirement is the easiness to don and doff, with minimal restriction to the wearer's daily activities, e.g. allow the wearer to sit when wearing the orthosis. To meet this requirement, the knee actuator is mounted on the outer flank, while the ankle actuator is mounted on the posterior side of the orthosis. With this actuation structure, the wearer can don and doff the orthosis in the same manner as the standard KAFO. Furthermore, this arrangement results in a sufficient range of motion for both the knee and ankle joint in the gait, and allows the wearer to sit with the orthosis on.

B. *Meeting the torque requirements of common locomotive functions*

Unlike the activities with upper limbs, the lower limbs are involved in the locomotion, and thus require significant torque output on the joints. To assist the elderly persons or persons with lower-limb pathologies, the proposed powered orthosis is expected to provide sufficient torque output for the common locomotive functions. For the design of the current prototype, the objective is to provide 100% joint torque for a 75 kg wearing during the level walking at slow and fast cadence, as well as the stair ascent and descent.

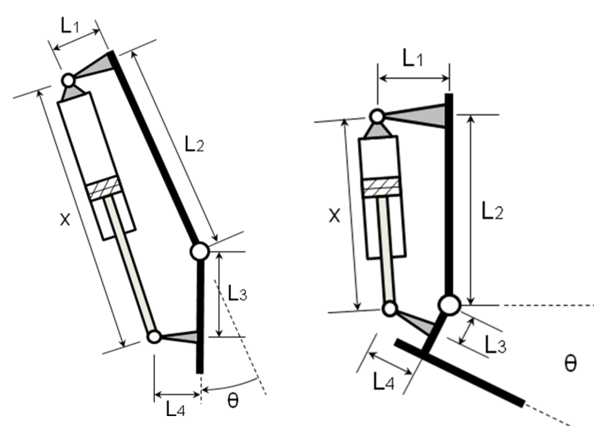

Fig. 2. Schematics of the actuation structures for the knee joint (left) and the ankle joint (right).

Note that the actuation torque capacity is proportional to the maximum actuation force of the cylinder, and also changes as a function of the joint angle, as determined by the kinetic characteristics of the slider-crank mechanism (Fig. 2). The equation for the maximum actuation force F_{MAX} can be derived as the product of the maximum air pressure in the cylinder chambers (usually the supply pressure P_s) and the corresponding piston area A_p:

$$F_{MAX} = P_s \cdot A_p \tag{1}$$

where $$A_p = \begin{cases} \pi D^2 & rodless\ side \\ \pi\left(D^2 - d^2\right) & the\,rod\ side \end{cases} \tag{2}$$

In the above equation, D is the piston diameter, and d is the piston rod diameter. Note that the piston areas in the two directions are slightly different due to the existence of the piston rod. To obtain the corresponding maximum torque, the method of virtual work can be applied, which results in the following equation:

$$\tau = -F \cdot \frac{dx}{d\theta} \tag{3}$$

where τ is the torque corresponding to a certain actuation force F. According to the geometric relationships shown in Fig. 2, actuator length x can be expressed as a function of the joint angle θ according to the following equations:

$$x = \sqrt{L_H^2 + L_L^2 - 2L_H L_L \mathbf{cos}\left(180° - \alpha_H - \alpha_L - \theta\right)} \tag{4}$$

where $$L_H = \sqrt{L_1^2 + L_2^2} \tag{5}$$

$$L_L = \sqrt{L_3^2 + L_4^2} \tag{6}$$

$$\alpha_H = \mathbf{tan}^{-1}\left(L_1 / L_2\right) \tag{7}$$

$$\alpha_L = \mathbf{tan}^{-1}\left(L_4 / L_3\right) \tag{8}$$

Substitute (4) into (3), the following eqn. can be obtained:

$$\tau = \frac{L_H L_L \mathbf{sin}\left(180° - \alpha_H - \alpha_L - \theta\right)}{\sqrt{L_H^2 + L_L^2 - 2L_H L_L \mathbf{cos}\left(180° - \alpha_H - \alpha_L - \theta\right)}} F \tag{9}$$

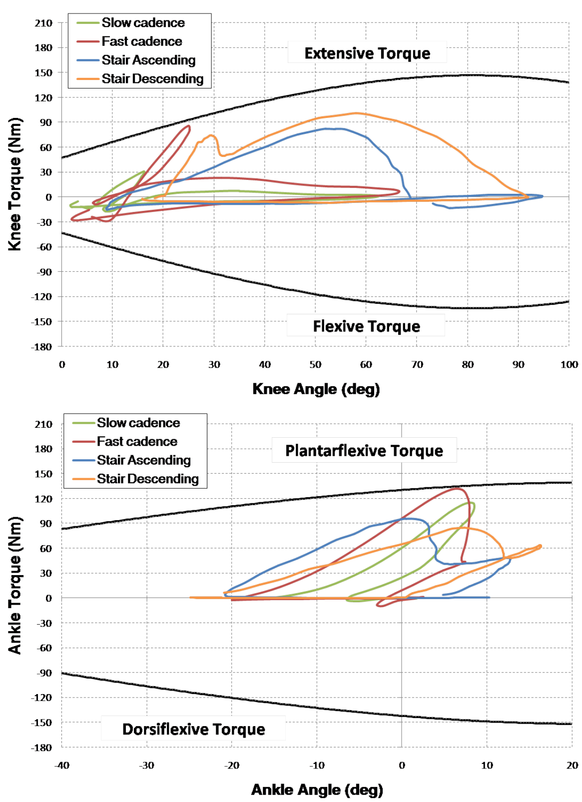

Fig. 3. Comparison of torque capability of the powered joints to the torque requirement during various gait modes for a 75 kg user.

From a design perspective, there are a large number of design variables, including the mounting position related variables L_1, L_2, L_3, L_4, and the cylinder-related parameters, including diameters D, d, and the stroke L. The values of the design variables are determined according to the practical considerations, with the objective of meeting the torque requirements of common locomotive functions and providing sufficient ranges of motion. Note that the torque requirements are derived from the standard torque trajectories obtained from the related biomechanical studies, including the level walking data from Winter [4] and stair ascent/descent data from Riener et al. [5]. The comparison of the design torque versus the torque requirements during various gait modes are shown in Fig. 3.

III. Orthosis Control

With the capability of supplying a significant amount of torque on the joints, the robotic lower-limb orthosis requires a reliable and effective control approach to enable a natural and coordinated assistance to the biological limb and joint motion. In the existing literature, the majority of orthosis control approaches regulate the joint position to follow predetermined joint trajectories, which are usually obtained from standard biomechanical studies. Such an approach poses multiple problems. First, enforcing predetermined joint trajectories largely precludes the interaction between the orthosis and the wearer's biological limb, and thus tends to generate an unnatural gait and affect the comfort in use. Moreover, with the significant variation in the human gait, the standard joint trajectory may not provide a close match to the user's natural walking gait. Finally, the standard gait trajectory involves a large amount of data, making it nearly impossible to conduct subject-specific tuning, posing a significant issue in the clinical application.

Aware of the weaknesses of the position-based control approaches, the authors propose the use of impedance-based approach for the control of the orthosis presented in this paper. The general impedance control theory was proposed by Hogan in 1980s for the control of robotic manipulators in interactive tasks [6]. Applying the impedance control to the lower-limb orthosis described above, it is expected to enable the interaction with the biological lower limb, and provide a more natural and comfortable user experience. From an energetic perspective, the impedance control is able to generate active joint power with a piecewise-passive behavior. Specifically, the controlled joint behavior in the orthosis joint is strictly passive (i.e. always converge to a local minimum) within a certain mode, while the joint power is generated when the equilibrium point of the artificial spring shifts with the mode transition. This provides a high level of safety in the operation, and enables the user to control the power delivery through the mode switching. In addition, the impedance control only involves a limited number of control parameters with clearly defined physical meanings, significantly simplifying the subject-specific tuning in the clinical use.

To obtain a fully functional control system for the orthosis, the impedance control is combined with a finite-state machine to provide assistance synchronized with the cadence of the gait. Specifically, a standard gait cycle consists of two distinct states, stance and swing, both of which can be further divided into a number of sub-states according to the dynamic characteristics. Sup et al. demonstrated that, by dividing the gait cycle into four states, the impedance representation is able to provide a highly accurate reproduction of the standard torque trajectory with the curve fitting technique [7]. Furthermore, the transition between the states is triggered by a certain set of conditions, often associated with highly definitive events (such as heel strike) that can be easily detected with sensor signals. Leveraging these unique advantages, the finite-state impedance control was applied to the lower-limb prosthesis presented in this paper to achieve a natural interaction in the locomotive assistance.

The finite-state machine in the orthosis control system consists of four distinct states, with the details described as follows, including: (1) Weight Acceptance (WA). Triggered by the heel strike, WA marks the start of the gait cycle. During this state, the joints maintain a high impedance, and absorb the energy of ground contact. (2) Push Off (PO). PO is the second phase of the general stance mode. This phase starts when the knee joint reaches the maximum flexion after the initial ground contact. During this phase, energy is generated to obtain the impetus for the forward motion. (3) Early Swing (ES). ES is the first phase of the general swing mode. Since the lower-limb is not in contact with the ground, the dynamic behavior of the joints can be modeled with a weak damper to regulate the free motion of the limb segments. (4) Late Swing (LS). LS is the second phase of the general swing mode, and also the last phase in the entire gait cycle. Also, LS is usually associated with the extension of the knee joint. In this mode, both knee and ankle joints are expected to go back to the initial state to prepare for the new gait cycle and the heel strike.

IV. EXPERIMENTS

To demonstrate the effectiveness of the orthosis in locomotive assistance, the preliminary testing was conducted on a healthy subject. The subject was fitted with the orthosis and conducted level walking experiments on a commercial treadmill modified with enhanced safety features. Also, the testing was conducted under a reduced supply pressure (approximately 100 psi instead of 200 psi), and thus the orthosis was expected to provide approximately 50% of the required torque for fast-walking and stair ascending /descending. In the experiments, the control parameters were tuning according to the objective measures (such as the torque and angle trajectories) as well as the feedback reported by the subject. Figure 4 shows the knee angle trajectory in the preliminary testing, which is very similar to that in the subject's natural (i.e. unassisted) gait, demonstrating the effectiveness of the proposed control approach. Also, the calculated joint power trajectory shows that the orthosis was able to provide approximately 30% of the locomotive power, demonstrating the significant assistance provided by the orthosis.

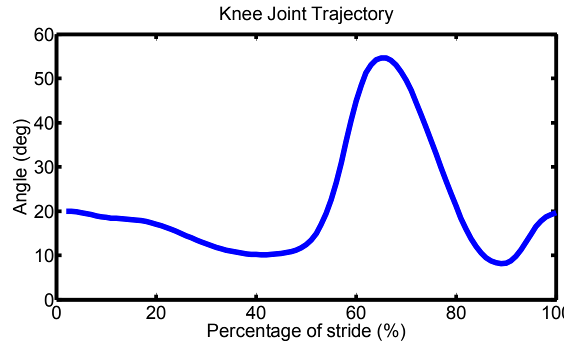

Fig. 4. Average knee joint trajectory in the level walking when the healthy subject was assisted with the powered orthosis.

V. CONCLUSIONS

In this paper, a pneumatically actuated lower-limb orthosis was presented. Developed based on the standard metal frame knee-angle-foot orthosis, this device incorporates two pneumatic actuators to provide locomotive assistance on the knee and ankle joints. The kinetic structure was specially design to provide full assistance during locomotion in a variety of common modes, and generate a minimum level of restriction to the daily activities. To control the assistance in the locomotion, a finite-state impedance controller was developed, which simulates artificial impedances on the joints to enable a natural interaction with the wearer. Preliminary testing results show that this orthosis was able to provide effective assistance during treadmill walking experiments.

REFERENCES

[1] J.A.Blaya and H. Herr, "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 12, no. 1, pp. 24-31, 2004.

[2] K.A. Strausser, T.A. Swift, A.B. Zoss, and H. Kazerooni, "Prototype medical exoskeleton for paraplegic mobility: first experimental results," in *Proceedings of ASME Dynamic Systems and Control Conference*, DSCC2010-4261, 2001.

[3] M. Goldfarb, E.J. Barth, M.A. Gogola, and J.A. Wehrmeyer, "Design and energetic characterization of a liquid-propellant-powered actuator for self-powered robots," *IEEE/ASME Transactions on Mechatronics*, vol. 8, no. 2, pp. 254-262, 2003.

[4] D.A. Winter, *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, Waterloo, ON, University of Waterloo Press, 1991.

[5] R. Riener, M. Rabuffetti, and C. Frigo, "Stair ascent and descent at different inclinations," *Gait and Posture*, no. 15, pp. 32-44, 2002.

[6] N. Hogan, "Impedance control: an approach to manipulation," *ASME Journal of Dynamic Systems, Measurement, and Control*, no. 107, pp. 1-24, 1985.

[7] F. Sup, A. Bohara, and M. Goldfarb, "Design and control of a powered transfemoral prosthesis," *International Journal of Robotics Research*, vol. 27, no. 2, pp. 263-273, Feb. 2008.