Design of an Agonist-Antagonist Active Knee Prosthesis

Ernesto C. Martinez- Villalpando, Jeff Weber, Grant Elliott and Hugh Herr

Abstract - The majority of commercial prosthetic knees are passive in nature and therefore cannot replicate the positive mechanical work exhibited by the natural human knee in early and late stance. In contrast to traditional purely dissipative prosthetic knees, we propose a biomimetic active agonistantagonist structure designed to reproduce both positive and negative work phases of the natural joint while using series elasticity to minimize net energy consumption. We present the design and physical implementation of the active knee prosthesis prototype.

I. INTRODUCTION

A. State of the Art

Modern transfemoral prostheses can be classified in three major groups: passive, dynamically damped, and powered, each of which exhibits fundamental deficiencies. Passive and dynamically damped knees are by their nature capable only of negative mechanical power and therefore cannot replicate the generative phases of the natural knee [1]. As a result, the adequate locomotion of above-knee amputees is impaired; leading to asymmetric gait and elevated metabolic energy consumption relative to healthy subjects [2-4].

Powered prostheses can significantly improve these problems; nevertheless they still face challenges in weight, energy efficiency, and range of motion. Current approaches to the design of powered knee prostheses have focused mainly on the use of single actuator-transmission systems directly coupled to the knee joint [5, 6]. Ossur's "Power-Knee" is one such commercially available device. Such a direct drive system, however, utilizes power inefficiently as it cannot leverage the natural dynamics of the system. In this paper we describe the design of a biomimetic active knee prosthesis which mimics knee biomechanics in level ground walking and utilizes series elasticity to minimize net energy consumption.



Figure 1: Typical angle and mechanical power curves of the knee joint during level ground walking at self selected speed [7][8].

B. Human Knee Biomechanics in Level Ground Walking

The natural level-ground walking cycle consists of five distinct stages (Figure 1) [8].

1. Beginning at *heel strike*, the stance knee begins to flex slightly. This *Stance Flexion* phase allows for shock absorption upon impact while maintaining the body's center of mass at a more constant elevation throughout the stance period. During this phase, the knee acts as a spring, storing energy in preparation for the *Stance Extension* phase.

2. After maximum flexion is reached in the stance knee, the joint begins to extend, until maximum extension is reached. This knee extension period is called the *Stance Extension* phase. During the first $\sim 60\%$ of *Stance Extension*, the knee acts as a spring, releasing the stored

E.C. Martinez-Villalpando is with the Massachusetts Institute of Technology, Biomechatronics Group, Media Laboratory, Cambridge, MA, 02139 USA (email: ernesto@media.mit.edu).

J. Weber is with the Massachusetts Institute of Technology, Biomechatronics Group, Media Lab (email: jaweber@media.mit.edu).

G. Elliott is with the Massachusetts Institute of Technology, Biomechatronics Group, Media Laboratory (email: gelliott@mit.edu).

H. Herr is with Massachusetts Institute of Technology and Harvard-MIT Health Sciences and Technology Program, Biomechatronics Group, Media Laboratory (email: hherr@media.mit.edu).

energy from the Stance Flexion phase of gait. During the last \sim 30% of *Stance Extension*, the knee absorbs energy in a second spring to be released during *Pre-Swing*.

3. During late stance or *Pre-Swing*, the knee of the supporting leg begins its rapid flexion period in preparation for the *Swing* phase. During early *Pre-Swing*, as the knee begins to flex in preparation for toe-off, the stored elastic energy from *Stance Extension* is released.

4. As the hip is flexed, and the knee has reached a threshold angle in *Pre-Swing*, the leg leaves the ground and the knee continues to flex. At toe-off, the *Swing Flexion* phase of gait begins. Throughout this period, knee power is generally negative as the knee's torque impedes knee rotational velocity. During terminal *Swing Flexion*, the knee can be modeled as a extension spring in series with a variable damper, storing a small amount of energy in preparation for early *Swing Extension*.

5. After reaching maximum flexion angle during *Swing*, the knee begins to extend forward. During the early *Swing Extension* period, the spring energy stored during late *Swing Flexion* is then released. During the remainder of *Swing Extension*, the knee outputs negative power (absorbing energy) to decelerate the swinging leg in preparation for the next stance period. During terminal *Swing Extension*, the knee can be modeled as a flexion spring in series with a variable damper, storing a small amount of energy in preparation for early stance. After the knee has reached full extension, the foot once again is placed on the ground, and the next walking cycle begins.



Figure 2: Knee Torque Vs. Angle with phases of gait[9]. HS-heel strike, MSF –maximum stance flexion, MSEmaximum stance extension, TO- toe off, MWFmaximum swing flexion.

II. PROSTHESIS DESIGN

A. Prosthetic Knee Design Criteria

The ideal knee prosthetic is capable of replicating the full range of motion of the human knee as well as the ranges of torque and stiffness which are observed in normal human walking. Just as importantly, the device should not exceed the size or weight of the amputated limb. Moreover, the prosthetic should exhibit dynamics similar to those of the natural limb [9, 25]. Reasonable design criteria are given in the following table.

| Length | 330 mm |
|---------------------|-----------|
| Width | 70 mm |
| Weight | 3 Kg |
| Flexion Angle range | 0-120 deg |
| Output torque | 100 Nm |

Table 1: Design criteria for the prosthetic knee.

B. Agonist Antagonist Architecture

To overcome the inadequacies of existing knee prosthetic devices, we propose an agonist-antagonist active knee prosthesis (AAAKP) with an architecture inspired by the muscle anatomy of the natural human knee joint. In particular, this architecture (see Figure 3) allows for independent engagement of flexion and extension series springs (k_F and k_E) so that joint position and stiffness are independently controllable. This allows the controller to utilize the passive dynamics of the system and store absorbed energy in the springs for later use, thereby increasing efficiency.

The actuation of the AAAKP is comprised of a pair of series elastic actuators (SEA) [10, 11], each of which consists of a torque source (active element) and a series spring, connected via a transmission. Two opposing SEAs are used to emulate the elasticity and damping characteristics of antagonistic muscle actuation.

In level-ground walking, this design permits the use of a quasi-passive control scheme. Active elements are used primarily to independently control the engagement of each tendon-like spring, thereby controlling the transformation of potential energy into kinetic energy. Since the ideal behavior of the knee on level ground is net dissipative, such a quasi-passive approach can in theory be extremely efficient.

We hypothesize that this device can produce an adequate and appropriate positive power output at the knee joint, thereby reducing the net work produced at the hip and, consequently, the metabolic cost of level ground walking.



Figure 3: Simplified mechanical architecture of the agonist-antagonist active knee prosthesis.

C. Gait Control Strategy

Control of the AAAKP is accomplished using a finite state machine inspired by the phases of the natural gait cycle described previously.

- 1. *Stance Flexion* begins at *heel strike*. The extension spring is engaged storing energy in preparation for the *Stance Extension*.
- 2. Stance Extension begins as maximum flexion is reached in stance. Initially, energy is released from the extension spring and subsequently stored in the flexion spring. This spring is engaged at *Flexor Angle 1* (θ_{F1}) .
- 3. *Pre-Swing* begins as the knee flexes past *Flexor Angle I* in preparation for the swing phase. Energy stored in the flexion spring is released prior to toe off.
- 4. Swing Flexion begins at toe off, as the knee reaches *Extensor angle 1* (θ_{E1}). Energy is stored in this extension spring in preparation for swing extension.
- 5. Swing Extension begins after maximum flexion occurs. Energy is initially released from the extension spring and subsequently absorbed in the flexion spring, which is engaged at *Flexor Angle 2* (θ_{F2}), to decelerate the swinging leg in preparation for the next stance period. During terminal *Swing Extension*, the extension spring is servoed in preparation for engagment and energy storage at heel strike.

D. Component Selection

a) Torque Sources and Transmission

As shown in Figure 4, the knee joint is driven by a set of cables connected to a linear carriage which is free to move along the length of the device. This carriage can be engaged on either side by the extension and flexion springs, each of which is positioned by a ballscrew driven by an electric motor. Both series elastic actuators feature transmissions comprised of a 2:1 belt drive coupled to a m ballscrew (10x3 mm). All actuation mechanisms are fully supported by an aluminum structure resembling the shape of the knee and shin. Each of the series elastic actuators in this prototype uses a brushed DC motor capable of providing sufficient power for both level ground walking and more energetically expensive tasks, such as climbing stairs.





b) Sensors and Electronics

Feedback to the controller is provided by the sensors listed in Table 2. In particular, locations and compression of both springs are monitored, as is the knee angle itself. An inertial measurement unit (IMU) is also used to detect motion of the limb and determine gait phase.

| Measurement | Sensor |
|--------------------|---------------------------|
| Ankle angle | Digital encoder |
| Motor displacement | Digital encoder |
| Heel strike | Hall effect |
| Spring compression | Hall effect |
| Limb acceleration | Inertial measurement unit |

Table 2: Sensors utilized by the AAAKP controller.

For the purposes of this prototype, all electronics are implemented in a single board which runs along the lateral side of the knee. Motors are driven by H-bridge controllers with speed governed by 20KHz pulse width modulation (PWM) and powered by a six cell Lithium polymer battery (22.2V nominal). Analog sensors are read through a 10-bit analog to digital converter (ADC). The system is controlled by an AVR microcontroller and may be monitored by either USB or Bluetooth.

Because all processing is done on-board and power can be supplied by a relatively small battery, the prototype is completely self-contained and does not require tethering.



Figure 5: Active agonist-antagonist knee prosthesis.

c) Series elastic components

In order to size components, an optimization is performed over the extension and flexion series spring constants as well as the three threshold angles which parameterize the control strategy. We minimize the mean magnitude deviation between joint power extracted from experimental gait data and the joint power predicted by simulation of the gait control strategy. That is, we minimize over a metric of the energetic cost of achieving perfect power tracking:

$$E_{cost}(k_F, k_E, \theta_{F1}, \theta_{F2}, \theta_{E1}) = \int \left\| P_{bio}(t) - P_{sim}(t) \right\| dt$$
$$= \int \left\| P_{bio}(t) - \tau_{sim} \dot{\theta}_{bio}(t) \right\| dt$$

where $P_{bio}(t)$ and $P_{sim}(t)$ are biological and simulated knee power respectively; $\dot{\theta}_{bio}(t)$ is knee velocity; and τ_{sim} is the net joint torque predicted by a simulation utilizing the gait control strategy when given the biological position data. The reference biological knee power $P_{bio}(t)$ and velocity $\dot{\theta}_{bio}(t)$ were obtained from fifteen trials with a healthy 81.9 kg, 1.87m subject walking at 1.31 m/sec. A Vicon motion capture system was used to extract sagittal plane joint angles, moments, and power. The determination of the desired global minimum for the objective function was implemented by using an unconstrained gradient optimizer implemented in MATLAB.

Utilizing this optimization, it is possible to select the gait control strategy threshold angles, θ_{F1} , θ_{F2} , and θ_{E1} , which minimize the energy cost for a given choice of k_F and k_E . This optimized energy cost as a function of spring constants is presented in Figure 6. In particular, notice that this system is significantly more sensitive to choice of extension spring than to flexion spring.

Optimizing over all parameters yields the optimal choice of springs and engagement angles:

Flexor Spring: $k_F = 104.5$ N.m/rad Extensor Spring: $k_E = 146.0$ N.m/rad Flexor Angle 1 (Start of stance extension): $\theta_{F1} = 0.42$ rad Flexor Angle 2 (Swing extension): $\theta_{F2} = 0.36$ rad Extensor Angle 1 (Swing flexion): $\theta_{E1} = 0.18$ rad



Figure 6: Energy cost (J) during a gait cycle as a function of flexion and extension spring constants.

III. SIMULATION RESULTS

Using these optimized parameters, the energy and knee torques produced by the gait control strategy were simulated for one gait cycle. Figure 7 compares these simulated results to measured data for a healthy subject. With the exception of an abnormally large flexion torque immediately prior to heel strike, an acceptable agreement with the natural joint is achieved. Figure 8 decomposes the knee torque into components provided by each of the series springs. Notice in particular the presence of cocontraction during stance, serving to stiffen the joint.

The simulation predicts a minimal energy consumption of 5.66J per step. Assuming an average gait cycle of 1.22sec duration, this corresponds to approximately 4.7 Watts. This value will be affected by the efficiency of the actuator-transmission system in the built prototype.



Figure 7: Comparison between biological and simulated prosthetic knee torque and power during a complete level-ground walking cycle.



Figure 8: Knee torque contributions from the flexion and extension springs of the series elastic actuators.

IV. CONCLUDING REMARKS

A biomimetic knee prosthesis prototype with agonistantagonist series elastic actuation was designed and built using elastic components optimized to minimize the energetic cost of level ground walking in simulation. This architecture permits prosthetics which are lightweight, efficient, and extremely versatile. Gait experiments and analysis on above knee amputees will be conducted shortly to validate simulation results. By closely tracking the natural knee behavior, the AAAKP is expected to reduce the metabolic cost of level ground walking for transfemoral amputees. Moreover, the powered nature of the device is expected to facilitate more energetically complex tasks, such as stair and ramp traversal.

REFERENCES

[1] J. L. Johansson, D. M. Sherrill, P. O. Riley, P. Bonato and H. Herr, "A clinical comparison of variable-damping and mechanically passive prosthetic knee devices," *Am. J. Phys. Med. Rehabil.*, vol. 84, pp. 563-575, Aug. 2005.

[2] J. G. Buckley, W. D. Spence and S. E. Solomonidis, "Energy cost of walking: comparison of "intelligent prosthesis" with conventional mechanism," *Arch. Phys. Med. Rehabil.*, vol. 78, pp. 330-333, Mar. 1997.

[3] G. A. Carmona, A. Lacraz and M. Assal, "Walking activity in prosthesis-bearing lower-limb amputees," *Rev. Chir. Orthop. Reparatrice Appar. Mot.*, vol. 93, pp. 109-115, Apr. 2007.

[4] S. V. Fisher and G. Gullickson Jr, "Energy cost of ambulation in health and disability: a literature review," *Arch. Phys. Med. Rehabil.*, vol. 59, pp. 124-133, Mar. 1978.

[5] A. O. Kapti and M. S. Yucenur, "Design and control of an active artificial knee joint," *Mechanism and Machine Theory*, vol. 41, pp. 1477-1485, 12. 2006.

[6] F. Sup, "Design and Control of a Powered Knee and Ankle Prosthesis," *Robotics and Automation, 2007 IEEE International Conference on,* pp. 4134-4139, 2007.

[7] D. A. Winter, "Biomechanical motor patterns in normal walking," *J. Mot. Behav.*, vol. 15, pp. 302-330, Dec. 1983.

[8] V. T. Inman, H. J. Ralston, F. Todd and J. C. Lieberman, *Human Walking*. Baltimore: Williams & Wilkins, 1981, pp. 154. [9] D. A. Winter and D. G. Robertson, "Joint torque and energy patterns in normal gait," *Biol. Cybern.*, vol. 29, pp. 137-142, May 31. 1978.

[10] G. A. Pratt, "Series elastic actuators," *Intelligent Robots and Systems 95. 'Human Robot Interaction and Cooperative Robots', Proceedings. 1995 IEEE/RSJ International Conference on*, vol. 1, pp. 399-406 vol.1, 1995.

[11] D. W. Robinson, "Series elastic actuator development for a biomimetic walking robot," *Advanced Intelligent Mechatronics, 1999. Proceedings. 1999 IEEE/ASME International Conference on,* pp. 561-568, 1999.

[12] Farber, Boris, S. and Jacobson, Jacob S., "An aboveknee prosthesis with a system of energy recovery:" *Journal of Rehabilitation Research and Development*, vol. 32, pp. 337, Nov. 1995.

[13] K. Fite, "Design and Control of an Electrically Powered Knee Prosthesis," *Rehabilitation Robotics, 2007. ICORR 2007. IEEE 10th International Conference on*, pp. 902-905, 2007.

[14] A. O. Kapti and M. S. Yucenur, "Design and control of an active artificial knee joint," *Mechanism and Machine Theory*, vol. 41, pp. 1477-1485, 12. 2006.

[15] Popovic, D. and Schwirtlich, L. Belgrade active A/K prosthesis. Presented at Electrophysiological Kinesiology (International Congress Series, no. 804).

[16] D. Popovic, M. N. Oguztöreli and R. B. Stein, "Optimal control for an above-knee prosthesis with two degrees of freedom," *Journal of Biomechanics*, vol. 28, pp. 89-98, 1. 1995.

[17] D. B. Popovic and V. D. Kalanovic, "Output space tracking control for above-knee prosthesis," *IEEE Trans. Biomed. Eng.*, vol. 40, pp. 549-557, Jun. 1993.

[18] M. L. van der Linden, S. E. Solomonidis, W. D. Spence, N. Li and J. P. Paul, "A methodology for studying the effects of various types of prosthetic feet on the biomechanics of trans-femoral amputee gait," *Journal of Biomechanics*, vol. 32, pp. 877-889, 9. 1999.

[19] H. A. Varol, "Decomposition-Based Control for a Powered Knee and Ankle Transfemoral Prosthesis," *Rehabilitation Robotics, 2007. ICORR 2007. IEEE 10th International Conference on*, pp. 783-789, 2007. [20] H. A. Varol, "Real-time Intent Recognition for a Powered Knee and Ankle Transfemoral Prosthesis," *Rehabilitation Robotics, 2007. ICORR 2007. IEEE 10th International Conference on,* pp. 16-23, 2007.

[21] D. A. Winter and S. E. Sienko, "Biomechanics of below-knee amputee gait," *Journal of Biomechanics*, vol. 21, pp. 361-367, 1988.

[22] Flowers, W.C. (1973). A man-interactive simulator system for above-knee prosthetics studies. PhD Thesis, Cambridge, MA, MIT, Department of Mechanical Engineering.

[23] Flowers, W.C., and Mann, R.W. (1977).Electrohydraulic knee-torque controller for a prosthesis simulator. ASME Journal of Biomechanical Engineering, 99(4): 3—8.

[24] S. K. Au, Weber, J, Herr, H. Martinez-Villalpando, E.C. "Powered Ankle-Foot Prosthesis for the Improvement of Amputee Ambulation," Engineering in Medicine and Biology Society, 2007. EMBS 2007. 29th Annual International Conference of the IEEE, pp. 3020-3026, 2007.

[25] W. Braune, O. Fischer. "Determination of the Moments of Inertia of the Human Body and Its Limbs". Springer-Verlag Berlin Heidelberg 1988.

[26] R. Riener, M. Rabuffetti and C. Frigo. "Stair ascent and descent at different Inclinations". *Gait & Posture*, Vol. 15, Issue 1, Pages 32 – 44. 2002.