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Antagonistic Active Knee Prosthesis. A Metabolic Cost of Walking Comparison With a Variable-Damping Prosthetic Knee

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Abstract—**This paper examines the impact of a biomimetic active knee prosthesis on the metabolic costs associated with a unilateral transfemoral amputee walking at self selected speed. In this study we compare the antagonistic active knee prosthesis developed at MIT to an electronically controlled, variable-damping commercial knee prosthesis, the Otto Bock C-leg. Use of the active knee prosthesis resulted in both, a 17% increase in an amputee's average self selected walking speed from 1.12 m/s to 1.31 m/s, and a 6.8% reduction in metabolic cost. The results of this study suggest that an agonist-antagonist active knee prosthesis design with variable impedance control can offer walking energetic advantages over commercially available systems.**

I. Introduction

Walking fatigue is synonymous with higher metabolic expenditure and is a common affliction of lower limb amputees. The rate of oxygen consumption in unilateral below the knee amputees is 20-30% higher than in intact subjects [1][2], with an additional 25% increase seen in above the knee amputees [3][4]. Conventional knee prostheses, despite their damping and compliance features have not provided a metabolic advantage for amputees [5][6]. In addition to higher metabolic consumption, transfemoral amputees show a reduction in self-selected speed and present overall diminished endurance.

Variable-damping knees are among the most advanced prostheses available. These knees require a power source to modulate damping levels and adapt to different modes of gait, whereas powered prosthetic knees are capable of performing non-conservative positive work. Variable-damping knees offer several advantages over mechanically passive designs, including enhanced knee stability and adaptation to different ambulatory speeds [6]-[11].Examples of commercially available variable-damping knees include the Blatchford Endolite Intelligent Prosthesis, the Otto Bock C-leg, and the Össur Rheo.

While microcontroller based, variable-damping knees offer some advantages over purely passive knee mechanisms, they are nonetheless unable to produce positive mechanical power, and therefore cannot replicate the positive work phases of the human knee joint for activities such as sit-to-stand maneuvers, level-ground walking, and stair/slope ascent ambulation. Not surprisingly, transfemoral amputees experience gait pathologies such as asymmetric gait patterns, slower gait speeds, and elevated metabolic cost requirements as compared to non-amputees [6].

Current approaches to the design of powered prostheses have focused mainly on the use of single motor-transmission systems directly coupled to the knee joint [12][13]. Such direct-drive designs, however, require high electrical power consumption to fully emulate the mechanical behavior of the human knee joint even during level-ground ambulation. One reason for this lack of energetic economy perhaps is that such designs do not adequately leverage the passive dynamics of the leg, and elastic energy storage structures, in a manner comparable to highly economical walking machine designs [14]-[16] or simpler mechanical knee designs with extension assist compliant elements [17].

In the following sections, we begin by briefly describing the active knee prosthesis design, its model, and control strategy for level ground walking. We then describe the experimental methods for speed adaptation on treadmill walking and the metabolic evaluation of a unilateral transfemoral amputee walking at self-selected speed comparing the active knee with a conventional C-leg. We finally discuss the results and implications of the agonist-antagonist design architecture of the active knee along with a variable impedance controller and its impact on step to step variability and metabolic cost during level ground walking.

II. Active Knee Prosthesis

A. Design and Hardware Description

The biomimetic active knee prosthesis developed at MIT's Biomechatronics research group [19],[20] incorporates an agonist-antagonist arrangement of two unidirectional series-elastic actuators (SEAs) [18]. This design, shown in Figure 1, is motivated by a variable-impedance prosthetic knee model, comprising two series-elastic clutch mechanisms and a variable-damper.

The two SEAs allow the prosthesis to behave elastically during stance phase, and provide both active assistance and variable dampening during swing phase, resulting in an energetically economical knee prosthesis for level-ground walking. Moreover, the fully motorized design with series-elastic force sensing allows for joint torque to be directly controlled for more energetically expensive tasks, such as stair and ramp ascent gaits and standing from a seated

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posture. The knee architecture is designed to accommodate non-conservative, high mechanical power movements, while still providing for a highly economical level-ground walking mode.

In each of the two series elastic actuators (one acting purely in flexion and one in extension), a brushed motor acts through a belt drive and ballscrew on one side of a linear spring, the stiffness of which is derived from models of human locomotion. The other side of each spring acts on a linear carriage whose translation is directly coupled to knee joint rotation. If an active force in a given direction is undesirable, the appropriate motor may disengage its spring from the linear carriage entirely. Motors may therefore act actively through the series elements or quasipassively by repositioning the springs, holding the springs stationary during energy storage, or dissipating energy by regulating spring movement.

The knee prosthesis is completely self-contained and does not require tethering; all electronics are implemented in a single onboard printed circuit board. The electronics system is based on an AVR microcontroller and custom motor controllers with speed governed by 20 KHz pulse width modulation (PWM). The knee is powered by a six cell Lithium polymer battery (22.2V nominal).

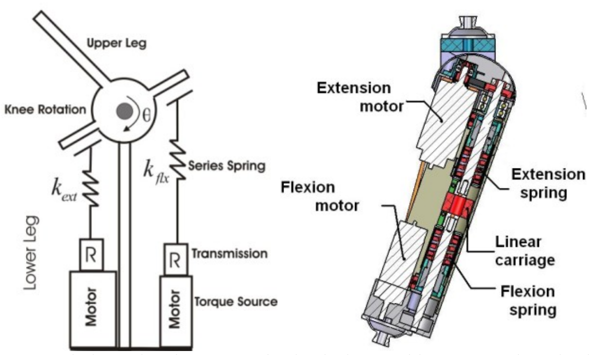

Fig.1. Antagonistic active knee prosthesis design architecture and sagittal cut view of the knee's electromechanical design

In order to develop the low-level actuator controllers, each motor (flexion or extension) in the antagonistic architecture is modeled as a torque source T_{flx} or T_{ext} with rotary internal inertia I_{flx} or I_{ext}, applying force to a series spring of stiffness k_{ext} or k_{flx} through transmission R_{flx} or R_{ext}. Additional damping terms b_{flx} and b_{ext} represent unavoidable brush and bearing friction acting on the motors (Figure 2). Each transmission converts rotary motion of its motor into linear compression of its series springs, which in turn applies a force at a moment arm of r from the knee joint . Finally, T_k and θ represent external knee joint torque and angular displacement, respectively. This model considers the lower leg inertial properties I_k. Moreover, it ignores nonlinearities due to stick-slip friction and transmission. Transmission ratios are assumed constant and do not vary with knee angle. Internal resonances are not considered.

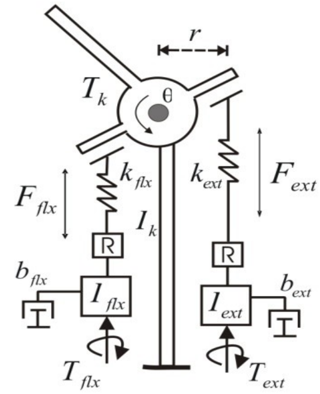

Fig.2. Active knee prosthesis system model

B. Control Strategy. Level Ground Walking

A finite-state controller for level-ground walking was implemented to replicate the intact knee behavior. The decoupled SEAs allow the controls to take advantage of the passive dynamics of the artificial limb in order to replicate the biomechanics of an intact knee joint during the gait cycle with an energy efficient and speed adaptable control strategy. While energy regeneration could be used to restore some of the energy lost to the virtual dampening, it was not implemented in this prosthesis. The three states implemented in the controller were Stance (ST), Swing Flexion (SF) and Swing Extension (SE).

1) Stance (ST) begins at heel strike. The knee's extension actuator acts as a virtual clutch and engages its series elastic component by maintaing the spring equilibrium position as the knee flexes slightly (~0.2 rad); energy is thus stored in the extension spring. This period of flexion allows for shock absorption. After reaching maximum stance flexion, the knee joint begins to extend (~15% gait cycle), until maximum stance extension is reached (~42% gait cycle). During this extension, the series elastic element of the flexion actuator is compressed. The energy stored in the extension spring during stance flexion is transferred into the flexion spring during stance extension. This energy transfer between the extension and flexion SEAs modulates the joint stiffness during stance, producing adequate torque behavior around the knee joint.

2) Swing Flexion (SF) is initiated once the knee begins to flex and before toe off. The first phase of SF involves accelerating the knee into full swing (~42% to ~62% gait cycle). All of the energy stored in the flexion spring during stance extension is rapidly released to propel the knee into SF. Once the energy in the spring is depleted additional energy is provided to assist in swing by using the flexion actuator. If the knee angle increases above 30 degrees, the extension actuator begins to behave as a viscous damper using low gain position control of the series spring equilibrium angle to regulate force. Using the extension actuator as a virtual damper instead of a hard-stop allows the knee to adapt to walking speed.

3) Swing Extension (SE) occurs after reaching a maximum swing flexion angle and the knee begins to extend forward. During swing extension (from ~73% to ~100% gait cycle) the knee joint experiences a deceleration in preparation for the next gait cycle. The flexion actuator provides active variable dampening by following a specified impedance trajectory. The beginning of SE (~73% to ~90%) is characterized by a constant viscous dampening. The flexion SEA effectively behaves as a linear viscous damper, maintaining a constant impedance. Once the knee angle is within ~10 degrees of full extension the flexion SEA begins to linearly increase the impedance as a function of angle. This results in a rapid but smooth deceleration of the lower limb at the end of swing. Once the knee comes to a complete stop at full extension the state controller re-enters ST.

III. EXPERIMENTS

A. Able-body subject testing.

Prior to evaluating the active knee prosthesis with a transfemoral amputee, we tested the artificial knee prosthesis' ability to accommodate different walking speeds without modifying the impedance parameters used in the finite state machine control for level ground walking. For this study we used a kneeling socket adaptor which enables an able-bodied subject to walk on the prosthesis to emulate amputee walking, facilitating development of the control strategy. The device is a modified hands-free crutch (i-Walk Free). For the study, we tested the prosthesis on an 81 Kg, 1.76 m tall able-body subject walking on a treadmill at three different speeds (09, 1.1 and 1.3 m/s) for a duration of 5 minutes at each speed. Using the onboard sensors and electronics system we recorded joint angle and torque.

Fig. 3.Able body subject using kneeling socket adaptor to test active knee prosthesis on a treadmill.

B. Amputee Testing.

We evaluated the impact of the active knee prosthesis on the metabolic cost of ambulation of a transfemoral amputee walking at self-selected speed. For the study, one healthy male participant with unilateral above-knee amputation (with no other musculoskeletal problems or any known cardiovascular, pulmonary or neurological disorders) was recruited. The subject weighs 97 Kg and has a height of 1.97m. The subject has a capacity of ambulation at a K3 level (i.e. having the ability or potential for ambulation with variable cadence).

In the first part of the assessment, the subject was asked to walk along a 10-m level walkway at a comfortable self-selected walking speed with his conventional prosthesis (Otto Bock's C-leg) and then with the active knee prosthesis. 10 walking trials were performed with each prosthesis. Before the trials with the powered knee prosthesis, the subject acclimated to the new device for approximately ten minutes. Parallel bars were utilized along the walkway for added safety. Both prostheses were fitted by the same certified prosthetist. Moreover, the patient used the same prosthetic socket, prosthetic foot (Flex-Foot LP-VariFlex® from Össur, Inc.) and shoe while testing each knee device. The self-selected speed when walking with each prosthesis was compared. The faster of the two average self-selected speeds was chosen for metabolic cost evaluation.

In the second part of the study we assessed the metabolic cost of walking by measuring rates of oxygen consumption and carbon dioxide production with a breath by breath portable telemetric system (Cosmed K4b2, IT). These measurements may be used to estimate the metabolic power P for each walking trial using the well documented [21] linear expression:

$$P = 16.48*VO_2 + 4.48*VCO_2 \qquad (1)$$

The participant was asked to first walk in an indoor athletic track for 8 minutes with his conventional knee prosthesis (C-Leg) at the walking speed determined from the first session in order to establish a control metabolic rate when walking.

Fig. 4 Unilateral above-knee amputee wearing portable K4b2 telemetric system equipment for metabolic cost assessment.

After resting for 10 minutes, the patient changed to the use of the active prosthesis and acclimated to the device by walking for 5 minutes. He then walked on the track for 8 minutes with the active knee while gas exchange rates were measured. To maintain a constant walking speed during the trials, the patient was instructed to follow an electric vehicle programmed to move at the chosen speed. Resting gas exchange rates were also measured with the participant seated for 5 minutes before and after each walking trial. The net metabolic power associated with each condition may be calculated using equation (1) and subtracting the calculated average resting power.

IV. Results and Discussion

The active knee prosthesis demonstrated the ability to provide qualitative agreement with intact knee biomechanics during level-ground ambulation at different walking speeds in the controlled treadmill study. Figure 5, depicts the average angle vs. torque curves at the knee joint for each walking speed. These results suggest that the antagonistic architecture in coordination with the variable-impedance control can facilitate step to step adaptation to speed variation during amputee locomotion. This architecture and controller take advantage of the passive dynamics of the artificial limb in order to replicate the biomechanics of an intact knee joint.

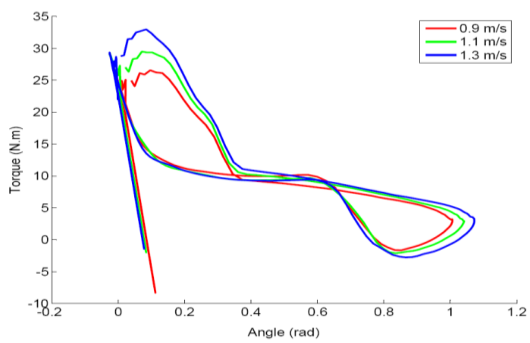

Fig 5. Average angle vs torque curves of the active knee prosthesis during treadmill walking of able-body subject using kneeling-socket adaptor.

The average self-selected speed in level ground walking of the amputee subject with the conventional C-leg was 1.12 m/s, while the average self-selected speed with the active knee was 1.31m/s. This represents an improvement of approximately 17%. During the metabolic assessment session, the walking speed was set at 1.3 m/s. The average resting metabolic cost was 1.06 W/kg. The metabolic power associated with the C-leg was 6.50 W/kg (5.44 W/kg above resting) while that associated with the active knee was 6.13 W/kg (5.07W/kg above resting). This corresponds to a 6.8% reduction in average metabolic power above resting. To the best understanding of the authors, this is the first time that the use of an active knee prosthesis attached to a passive foot-ankle system has shown a metabolic reduction when compared to the subject's conventional prosthesis. For this study the overall weight of the artificial limb (knee prosthesis with subject's own foot and shoe) was 2.8 kg when using the c-leg and 3.6 kg when using the active prosthesis. The metabolic cost reduction despite the increased weight of the powered prosthetic system suggests the possibility of further decreasing metabolic cost by optimizing the artificial lower limb weight.

Future work will expand this research to include a larger number of transfemoral amputees in order to further determine the impact on the metabolic cost of walking, as well as other clinical advantages of the antagonistic powered knee.

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