ABSTRACT

We present the design, control, and experimental evaluation of an energy regenerative powered transfemoral prosthesis. Our prosthesis prototype is comprised of a passive ankle and a powered knee joint. The knee joint is actuated by an ultracapacitor based regenerative drive mechanism. A novel varying impedance control approach controls the prosthesis in both the stance and swing phase of the gait cycle, while explicitly considering energy regeneration. This control method varies the impedance of the knee joint based on the amount of force exerted on the shank of the prosthesis. Furthermore, the controller promotes energy regeneration by precisely injecting a designated amount of negative damping into the system. Our control approach leads to a few tuning parameters that cover all of the gait phases for walking and all of the tested walking speeds and eliminates the need for tedious target impedance scheduling. Experimental evaluation is done with an amputee test subject walking at different speeds on a treadmill. The results validate the effectiveness of the control method. In addition, net energy regeneration is achieved while walking with near-natural gait across all speeds.

INTRODUCTION

Energetically passive prostheses enable walking for people with lower limb amputations by using damping and spring-like elements [1]. Accordingly, passive prostheses cannot provide net positive power. Able-bodied gait, on the other hand, requires positive power output at the ankle and knee joints [2]. This results in an increase in energy consumption for amputees during walking [3, 4].

Powered prostheses have been shown to reduce the metabolic cost of transport by providing positive net work [5–7]. However, one main drawback of powered prostheses that hinders their widespread use is their power consumption. For the Vanderbilt leg, which includes both a powered knee and ankle, a walking distance of about 9-12 km and a battery life of approximately 1.8 hours of continual walking have been reported [8, 9]. The Empower ankle (Ottobock, Duderstadt, Germany) can provide a battery life of up to eight hours [10]. The Power Knee
(Ossur, Reykjavik, Iceland) has a battery life of approximately twelve hours depending on usage and a charging time of three and one half hours [11]. Considering power consuming daily activities beyond the average walking pace for which these values are reported, such as fast walking or climbing stairs, the aforementioned powered prostheses would have to be recharged several times daily for an amputee with a moderately active lifestyle.

Energy regeneration technologies have gained much attention due to their potential to reduce the energy cost of systems, allowing them to work for longer periods of time with lower operational costs. Use of energy regeneration technologies in powered prostheses specifically can provide longer battery life and more generous ranges of locomotion, making them more practical for daily use. The concept of energy regeneration is understood here to be the process of recovering energy that would otherwise be dissipated and storing it for later use.

The potential to recover energy in powered lower limb prostheses can be seen when considering power flows occurring in able-bodied walking. During able-bodied gait, the knee mostly acts as a brake, absorbing/dissipating power, while the ankle acts as a motor, providing power [2, 12]. In a powered prosthetic leg, the energy dissipated by the knee joint has the potential to be stored and reused to reduce the overall power consumption.

Many papers discussing energy regeneration in powered prostheses are scarce in the research literature. An MIT group developed a regenerative transfemoral prosthesis in the 1980s [13–15]. Their conclusions suggested the use of larger capacitances, which were not available at the time. Arizona State University has created a powered robotic ankle that uses energy storing elastic elements in series with an electric motor [16–19]. In a powered ankle prosthesis simulation, Everarts et al. [20] also use an elastic element in series with an electric motor to reduce the peak power and regenerate energy. Tucker et al. [21] developed an analytical model of a regenerative powered transfemoral prosthesis. A regeneration manifold is found that limits the actuator damping which can be achieved while regenerating energy.

Many powered lower limb prostheses use impedance control [5, 6, 17, 22, 23]. Impedance control emulates the behavior of physical springs and dampers; any active, variable, and nonlinear behavior can be achieved. For prostheses power is produced by varying the joint impedance during different gait phases. However, the advantages of impedance control over mechanical springs and dampers, as with any powered control method, come at the price of energy consumption.

The general approach for controlling powered lower limb prostheses is to use a finite state impedance controller which divides the gait into discrete states [5–9, 22–27]. Each state has its separate controller and transitions between states are triggered by sensors placed on the prostheses. The control parameters for each state are tuned to individual subjects and different walking speeds. A five state controller with three gains per state across three walking speeds can lead to as many as $5 \times 3 \times 3 = 45$ tuning parameters in addition to gait-phase switching rules [8]. More elaborate methods add variation of the impedance parameters based on joint angles or measured forces during the finite states to reduce the number of tunable parameters [28, 29]. We refer readers to a comprehensive survey [1] of control strategies used for powered lower extremity prostheses.

We present an overview of the design, control, and experimental evaluation of an ultracapacitor based regenerative powered prosthetic knee. A previously introduced framework for analyzing regenerative robotic systems [30–33] is used to design and model the powered prosthesis. Ultracapacitors, also known as supercapacitors, provide an efficient means of storing and reusing energy [34] that is lightweight and durable and has high power densities and the ability to rapidly charge and discharge without damage. This is the first known publication of a human trial with an electro-mechanical energy-regenerative prosthesis. In addition, a novel varying impedance control approach drives the prosthesis in both stance and swing phase, while explicitly dealing with energy regeneration. Our control method varies the impedance of the knee joint based on the amount of force exerted on the shank. This approach provides a natural variation in the impedance of the knee and leads to far fewer tuning parameters compared to some other approaches [6–9]. In addition, the controller allows walking at different speeds without the need for retuning, and with a simple adjustment, the same tuning can be used for different subjects. The prosthesis is evaluated experimentally by having an amputee test subject walk with the device on a treadmill.

**THE REGENERATIVE PROSTHESIS MODEL**

We have previously developed a framework for analyzing robotic systems having regenerative actuators or drive mechanisms [30–33]. Based on this framework, each actuator of a regenerative robotic system is classified as either conventional, external power is used for actuation, or regenerative, energy storing elements store and reuse excess energy from the robot. The conventional actuators are termed fully-active while the regenerative actuators are termed semi-active. This framework is used as a basis for analyzing and modeling our regenerative prosthesis.

Our prosthetic leg prototype, depicted in Fig. 1, is comprised of a powered regenerative knee joint and a passive ankle joint. To model the system, the prosthesis, excluding the regenerative drive mechanism, can be thought of as a four degree of freedom standard open-chain robotic system. The first three degrees of freedom are the horizontal motion, vertical motion, and rotation of the prosthesis socket in the sagittal plane; the fourth degree of freedom corresponds to flexion/extension of the knee joint. The equation of motion for this system can be expressed as

$$D(q)\ddot{q} + C(q, \dot{q})\dot{q} + R(q, \dot{q}) + g(q) + T = \tau$$ (1)
where \( q \) is the 4 × 1 vector of joint coordinates, \( D^o(q) \) is the inertia matrix, \( C(q, \dot{q}) \) is a matrix accounting for Coriolis and centrifugal effects, \( R^o(q, \dot{q}) \) is a general nonlinear damping term, \( T \) is the vector of external forces and moments reflected to the manipulator joints, and \( g(q) \) is the gravity vector.

The motion of the prosthesis socket \((q_1, q_2, \text{and } q_3)\) is controlled by the human subject and can be thought of as fully-active, injecting energy into the system. The prosthetic knee is connected to a regenerative (semi-active) drive mechanism comprised of a transmission with velocity ratio \( n \), inertia \( m \), and viscous damping coefficient \( b \), a DC motor/generator with torque constant \( \alpha \) and resistance \( R \), an ultracapacitor with capacitance \( C \), and a four quadrant motor driver with voltage ratio \( r \). Figure 2 depicts a schematic of the regenerative drive mechanism. The inertial and frictional effects of the motor are assumed to have been reflected to the transmission and are thereby included in \( m \) and \( b \). In the most general case, the transmission ratio can be a function of the knee joint angle. The voltage ratio \( r \) represents the ratio of motor voltage and capacitor voltage. Since the motor driver does not boost the capacitor voltage, \( r \) is constrained to \([-1,1]\). A value \( r < 0 \) corresponds to applying reverse polarity voltage to the DC motor.

Bond graphs [35] are used to facilitate the system representation and equation derivation. From the bond graph model in Fig. 2, the interfacing force \( \tau_4 \) for the knee joint is found as

\[
\tau_4 = -mn^2\ddot{q}_4 - (bn^2 + \frac{a^2}{R})\dot{q}_4 + \frac{ar}{R}V_{cap} \tag{2}
\]

where \( V_{cap} \) is the capacitor voltage, and \( a = \alpha n \). Substituting \( \tau_4 \) from Eqn. (2) into Eqn. (1) and absorbing the terms containing \( \dot{q} \) and \( \ddot{q} \) into the left-hand side, the complete model of the prosthesis with the regenerative drive mechanism is obtained

\[
D(q)\ddot{q} + C(q, \dot{q})\dot{q} + R(q, \dot{q}) + g + T = u \tag{3}
\]

where \( D \) and \( R \) result from augmenting \( D^o \) and \( R^o \),

\[
D_{44} = D^o_{44} + mn^2\ddot{q}_4 \\
R_4 = R^o_4 + (bn^2 + \frac{a^2}{R})\dot{q}_4
\]

The first three joints are actively controlled by the human subject \((u_{1-3})\). For the knee joint

\[
u_4 = \frac{ar}{R}V_{cap} \tag{5}
\]

**Semi-active Virtual Control Strategy**

The semi-active virtual control strategy (SVC) provides a means for controlling semi-active joints [30–32, 36]. To control the semi-active knee joint, a virtual control law \( \tau^d \) is initially designed for \( u_4 \) in Eqn. (3). A virtual matching law is then found that can be solved for \( r \)

\[
u_4 = \frac{ar}{R}V_{cap} = \tau^d \tag{6}
\]

The virtual control \( \langle \tau^d \rangle \) can be of any form to achieve desired control objectives. Moreover, provided exact virtual matching of Eqn. (6), properties of the virtual design such as stability, tracking performance, robustness, etc. will also apply to the actual system [30]. For this system virtual matching is always possible as long as the capacitor voltage is high enough. Note that the semi-active virtual control method decouples the system from the ultracapacitor model by placing \( V_{cap} \) in feedback of the virtual control law [33]. This is a major advantage because it allows modeling, analysis, and control of regenerative systems without involving the complexities of ultracapacitor models.

**Regenerated Energy in the Knee Joint**

Energy provided to the capacitor between any two arbitrary times \( t_1 \) and \( t_2 \) can be derived from the bond graph of Fig. 2,

\[
\Delta E_s = \int_{t_1}^{t_2} V_{cap} i \, dt \tag{7}
\]
where \( i \) is the capacitor current. By deriving \( i \) from the bond graph and using Eqn. (2), Eqn. (7) can be written in terms of the virtual control

\[
\Delta E_s = \int_{t_1}^{t_2} \left( \tau^d \dot{q}_4 - \frac{R}{a^2} (\tau^d)^2 \right) dt
\]

A detailed derivation can be found in [33]. A value of \( \Delta E_s > 0 \) indicates energy regeneration and \( \Delta E_s < 0 \) indicates energy consumption. As a result of SVC, \( \Delta E_s \) is independent of the ultracapacitor model and is only a function of the control law \( \tau^d \), joint velocities \( \dot{q} \), and joint parameters \( R \) and \( a \).

THE VARYING IMPEDANCE CONTROL METHOD

We use a novel varying impedance control method to control the prosthetic knee. Our approach changes the impedance of the knee joint based on the amount of force applied to the prosthesis' shank. This provides a continuous variation of the knee impedance during the gait cycle and enables a soft transition between the stance and swing phases of gait. Moreover, our approach leads to far fewer tuning parameters when compared to finite state impedance control. Five parameters that are independent of walking speed are identified. Furthermore, once the controller is tuned, the same tuning can be used for different subjects with just a simple adjustment.

The equation describing the control structure is

\[
\tau^d = -(B_h + B)\dot{q}_4 - \frac{F}{F_s} K q_4 - K_s (q_4 - \dot{q}_4^e)
\]

where \( B \) and \( B_h \) are virtual damping coefficients, \( F \) is the shank force, \( F_s \) is a normalization factor, \( K \) and \( K_s \) are virtual spring stiffnesses, and \( \dot{q}_4^e \) is the equilibrium point of the virtual spring. We explain the controller's functionality in the stance and swing phases separately.

Swing Phase

In the swing phase the shank force \( F \) is zero and Eqn. (9) reduces to

\[
\tau^d = -(B_h + B)\dot{q}_4 - K_s (q_4 - \dot{q}_4^e)
\]

We mainly rely on the kinetic energy of the prosthesis at the beginning of the swing phase to extend the knee. The virtual stiffness \( K_s \) can be used to further propel the leg if the knee joint does not fully extend before heel strike. During tests with the prototype, we observed that this was not the case and set \( K_s \) to zero.

Virtual damping constant \( B_h \) prevents the mechanical hard stop from making contact at the end of the swing phase and only becomes active when the knee angle approaches full extension,
meaning that the screw displacement becomes less than a certain threshold.

\[ B_h = \begin{cases} 
  b_h & q_4 < q_{\text{threshold}} \\
  0 & q_4 > q_{\text{threshold}} 
\end{cases} \quad (11) \]

This could, however, be achieved mechanically by installing a soft stop insert, avoiding the need to expend extra electrical energy.

The purpose of the virtual damping constant \( B \) is to regenerate energy in the swing phase. The damping constant is set by considering the regenerated energy Eqn. (8) under the case where \( \tau^d = -Bq_4 \)

\[ \Delta E_s = \int_{t_1}^{t_2} \left( B + \frac{R}{a^2} B^2 \right) \dot{q}_4^2 \, dt \quad (12) \]

From Eqn. (12) it can be seen that energy is regenerated (\( \Delta E_s > 0 \)) only if

\[ -\frac{a^2}{R} < B < 0 \quad (13) \]

Equation (13) suggests that power can only be regenerated with a virtual damper if negative damping constants are used. Negative damping constants in the range of Eqn. (13) reduce the damping of the overall system but not to the extent of causing instability. Assuming that \( q_4 \) is mostly governed by the system dynamics and varying \( B \) in the range of Eqn. (13) has negligible effect on \( \dot{q}_4 \), we can differentiate Eqn. (12) with respect to \( B \) and set it to zero to find the optimum damping constant for regeneration

\[ B^* = -\frac{a^2}{2R} \quad (14) \]

and the optimum energy regenerated

\[ \Delta E^*_s = \int_{t_1}^{t_2} \frac{a^2}{4R} \dot{q}_4^2 \, dt \quad (15) \]

**Stance Phase**

Once the foot makes contact with the ground, the shank force \( F \) is non-zero, reinstating the full control law Eqn. (9). The virtual spring \( K \) dominates the control in stance phase due to the smaller knee velocities. Also, the virtual spring constant \( K \) is typically set to much larger values compared to \( K_s \). The stance phase control reduces to

\[ \tau^d \approx -\frac{F}{s} Kq_4 \quad (16) \]

The normalization factor \( F_s \) is the measured shank force \( F \) when the amputee is fully supported by the prosthetic leg.

During stance phase the behavior of the control law dictates that as the user shifts his or her weight to the prosthetic leg, the knee will stiffen, providing the amputee with the necessary support even when the knee remains slightly flexed. This is in contrast to the slow collapse of, for example, a hydraulic knee joint under matching conditions. During late stance when the amputee prepares for swing phase and begins to transfer his or her weight to the opposing leg, the proposed control law causes the prosthesis to soften, initializing the knee flexion required to enter swing phase. Each of these transitions are accomplished without switching between multiple sets of control gains.

**Controller Tuning Procedure**

Tuning the control law therefore requires the selection of five values. These are \( b_h \), \( q_{\text{threshold}} \), \( K \), \( K_s \), and \( q_4^* \). Notably, \( B \) is automatically determined by system parameters; see Eqn. (14). The parameter \( b_h \) is set to be a little larger than the magnitude of \( B \) so that it cancels the negative damping. The smallest sufficient value of \( q_{\text{threshold}} \) to prevent the hard stop is chosen, typically a couple of millimeters. \( K \) is determined by trial and error such that the amputee feels well supported. \( K_s \) and \( q_4^* \) are nonzero only when needed and are increased until the knee fully extends under the user’s volition. Once this initial tuning is completed, it is expected that other amputees could reuse the same tuning parameters. The value for \( F_s \), which is specific to the user’s weight, would only need to be updated. None of the tuning parameters are dependent on the user’s selected speed either. These features of the controller make tuning efforts minimal and very straightforward compared to finite state impedance controllers.

**THE PROSTHESIS PROTOTYPE AND EXPERIMENTAL SETUP**

The prototype used for this work was constructed from off-the-shelf components with an emphasis on creating a low-cost, proof-of-concept system. The overall system can be divided into the following categories: actuation, power storage, control, and sensing. A schematic of the system is given in Fig. 3.

The knee structure was built such that standard pyramid adapters are available at both the thigh and ankle interfaces. Also, it should be noted in relation to stance phase that when the knee is completely straight it can enter a mechanically self-locking region, depending on the location of the user’s center of mass. Under this condition \( q_4 = \dot{q}_4 = 0 \) rad, eliminating active power usage and saving energy. The knee attached to a socket and an Ottobock Triton Vertical Shock foot is shown in Fig. 1.

A 12 V DC motor-driven lead screw (ULTRAMOTION) actuates the knee joint by use of the crank-slider architecture. Power is supplied to the motor from four ultracapacitors linked
in series by balancing circuitry (Maxwell Technologies, BKIT-MCINT). These capacitors are rated for up to 2.7 V and at 650 F each (Maxwell Technologies, BCAP0650 P270 K04), determining a maximum operating voltage of 10.8 V. Regulating the power flow to and from the motor, a 10 A SyRen motor driver was selected (DimensionEngineering). This device is capable of four-quadrant operation. The analog control signal sent to the motor driver are generated by the dSPACE system, specifically the DS1104. The control software run by this system was developed in Simulink with the more complex computations written directly in MATLAB code.

A variety of sensors were installed for both control feedback and performance evaluation. For feedback, motor position, which is kinematically related to knee angle, was measured by an encoder, from which velocity could be computed. Additionally, two strain gages were adhered to the foot and then calibrated to produce shank force. The capacitor voltage was measured for use in the semiactive virtual control method. To be able to evaluate the energy regeneration capacity of the prosthesis, current sensors were installed at both the input and output to the motor driver; see Fig. 3 for the wiring schematic. The voltage applied to the motor as well as the voltage across the capacitors, as previously mentioned, were recorded. Combining these two pairs of measurements provides information regarding the power usage and the efficiency of the motor driver. All data with exception of the knee position were passed through a digital filter with a cutoff frequency of 24 Hz.

Human trials with an amputee subject were completed at the Louis Stokes Cleveland VA Medical Center as approved by its internal institutional review board. A 35-year-old male (81.7 kg, 175.3 cm) with a right transfemoral amputation volunteered to trial the leg; see Fig. 4. The test subject walks with a Freedom Innovations Plie microprocessor-controlled passive knee in combination with an Ottobock Triton Vertical Shock foot on a daily basis. The subject used his personal socket and daily foot for all
Three speeds were selected for the trial protocol, which was executed on a treadmill. These were the amputee’s preferred speed while using his everyday leg and plus and minus 0.15 m/s, giving 0.6 m/s, 0.75 m/s, and 0.9 m/s. All test data were taken on the same day. The amputee was provided two periods of at least 15 minutes on previous, non-consecutive days to familiarize himself with the experimental prosthesis.

### TEST RESULTS

The controller was tuned by trial and error based on the amputee’s feedback and the guidelines previously described. The final parameters are provided in Tab. 1. $K$, the spring constant dominating the stance phase, was tuned before the trial so that the leg could hold the weight of one of the authors. This gain was then fine-tuned with the test subject while he walked on a treadmill. The swing phase spring constant $K_s$ and accordingly $q^s_4$ were zero because the test subject’s gait pattern caused the prosthesis to fully extend without aid. $B$ was computed based on the constants $a$ and $R$ that were identified for the actuator. $b_h$ overrides the negative damping during late swing phase, which is defined as $q_{\text{threshold}} = 2$ mm of screw travel before full extension. Note that the same tuning was used for all tests and walking speeds.

<table>
<thead>
<tr>
<th>$K$ (N/mm)</th>
<th>$K_s$ (N/mm)</th>
<th>$B$ (Ns/mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>200</td>
<td>0</td>
<td>-1.743</td>
</tr>
<tr>
<td>$b_h$ (Ns/mm)</td>
<td>$q^s_4$ (mm)</td>
<td>$q_{\text{threshold}}$ (mm)</td>
</tr>
<tr>
<td>2.5</td>
<td>0</td>
<td>2</td>
</tr>
</tbody>
</table>

Figure 5 also shows the electrical power flows for the three tested walking speeds. Power flows are given for the capacitor and motor side of the motor driver. Positive power indicates power consumption while negative power indicates power regeneration. In the stance phase, very little power is consumed. In able-bodied gait, the knee joint uses positive power while it is being extended during mid-stance [2]. As previously explained, for our tests the knee was fully extended during stance and supported the weight of the amputee without the need for energy expenditure. Hence the controller only needs to provide sufficient power to stabilize the knee. In the swing phase the negative damping term of Eqn. (9) becomes dominant, and power is regenerated and stored in the ultracapacitor. With increased walking speed the peak value of the regenerated power increases. In addition, not all of the regenerated power is stored in the capacitor, and a portion of it (0.7 Watts) is consumed by the motor driver.

As was observed from the power plots, energy regeneration was possible under the test conditions. Integrating the power measurements yields total energy regenerated. This is shown for both the motor and capacitor sides of the motor driver and across all speeds in Fig. 6. As speed increases, it is clear from the motor side values that energy regenerated increases. However, approximately one half or more of the regenerated energy does not reach the capacitor bank to be stored. Efficiency does increase significantly between the slow speed and both higher rates, suggesting that the efficiency of the motor driver is affected by the return voltage and/or current applied. There is a less significant increase between the preferred and fast walking speeds. This is likely partially due to the previously mentioned singularity in the crank-slider. It is also true that the hard stop prevention damping will use more energy at higher swing speeds, causing less energy to be available for storage. Comparing these results with able-bodied data, in [2] the range of available energy is about 15-30 J for slow to fast paced walking, respectively. It therefore seems likely that there is still significant energy to be captured beyond what we have accomplished. Indeed, a regeneration potential is known to exist during the stance phase of able-bodied gait [2]. Our current results do not include energy recovery from this region.

### CONCLUSIONS

Within this work we have developed a powered knee prosthesis and controller, emphasizing control simplicity and energy...
regeneration. An experimental trial conducted with an amputee test subject validated the control method and achieved energy regeneration. Furthermore, basic features of able-bodied gait were replicated in testing, including swing phase knee flexion and transitions between gait phases.

Three traits of the proposed control law differentiate it from alternative methods. First, it has only five parameters, and they are intuitive for the individual adjusting the gains. This makes the tuning process relatively easy; tuning was able to be completed in a matter of minutes while conducting the test. Second, guidelines for tuning for energy regeneration can be developed analytically. Lastly, our approach only provides power to the knee joint when needed, yielding further energy savings.

Considering longer periods of operation in the future, several items must be addressed. As with any system with finite on-board power storage, operation must be stopped once charge (indicated by $V_{cap}$) drops below an acceptable threshold. At this point the system must be recharged. It is important to note that two alternative conditions, self-sustained operation or even charge buildup, can also occur in a system with energy regeneration. Achieving either of these conditions is dependent on system parameters and trajectories. If charging occurs during operation, the regenerated power from the knee can be used for operating a powered ankle, which is a primary long-term goal of this work.

As suggested by the results, the prototype suffers from several sources of energy loss. Additionally, there are some losses that are not even reflected in the measurements taken. In the next hardware iteration, an improved actuator, including the motor,
motor driver, and screw, will aid in eliminating these losses. Because the energy regenerated is directly dependent on the motor parameters, a more optimal motor can be selected with this in mind. A different four-quadrant motor driver should be identified to better transfer the power available for regeneration at low walking speeds. The current screw is a lead screw with a rated efficiency of about 60%. Replacing it with a ball screw can easily raise this value above 90%. In addition, the energy regenerated by the negative damping is inversely proportional to electrical resistance \( R \), Eqn. (15). By embedding the electronics and eliminating the lengthy tether used in the test, further improvements in energy regeneration are possible.

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