Decomposition-Based Control for a Powered Knee and Ankle Transfemoral Prosthesis

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Abstract— This paper describes an active passive torque decomposition procedure for use in controlling a fully powered transfemoral prosthesis. The active and passive parts of the joint torques are extracted by solving a constrained least squares optimization problem. Rather than utilize "echo control" as proposed by others, the proposed approach generates the torque reference of joints by combining the active part, which is a function of the force and moment vector of the interaction between user and prosthesis and the passive part, which has a nonlinear spring-dashpot behavior. The ability of the approach to reconstruct the required joint torques is demonstrated in simulation on measured biomechanics data.

I. INTRODUCTION

Despite significant technological advances over the past decade, commercial transfemoral prostheses remain essentially limited to energetically passive devices. The knee and ankle joints of the native limb, however, generate significant net power output over a gait cycle during many locomotive functions, including walking upstairs and up slopes, running, and jumping [1-8].

A major reason for the absence of powered joints in transfemoral (and transtibial) prostheses is the lack of energy source and actuator with suitable power density to provide the power and energy required for gait in a low-weight device. Recent advances in power supply and actuation for self-powered robots, such as the liquid-fueled approach developed by Goldfarb et al. [9-12], offer a power density that makes a powered lower limb prosthesis feasible. Based on this liquid-fueled approach, the authors have developed a prototype of a powered knee and ankle transfemoral prosthesis, as described in [13] and shown in Fig. 1. The development of a powered prosthesis changes significantly the nature of the user-prosthesis interface and control problem. Unlike a passive device (e.g., a modulated damper knee joint) that can fundamentally only react to a user’s input, a powered device can both act as well as react. As such, the prosthesis necessitates a reliable control framework for generating required joint torques while ensuring stable and coordinated interaction with the user and the environment.

II. PRIOR WORK IN POWERED TRANSFEMORAL PROSTHESES

The authors are not aware of any prior work in the development of a powered knee and ankle transfemoral prosthesis, although prior work has been published in the development of powered knee transfemoral prostheses. Specifically, Flowers et al. [14-20] developed a tethered electrohydraulic transfemoral prosthesis that consisted of a hydraulically actuated knee joint tethered to a hydraulic power source and off-board electronics and computation. They subsequently developed an “echo control” scheme for gait control, as described by Grimes et al. [16], in which a modified knee trajectory from the sound leg is played back on the contralateral side. Popovic and Schwirtlich [21] report the development of a battery-powered active knee joint actuated by DC motors, together with a finite state knee controller that utilizes robust position tracking control algorithm for gait control. Finally, Ossur, a major prosthetics company based in Iceland, has announced the limited launch of a powered knee, called the “Power Knee.”

III. INTERFACE APPROACHES

Unlike existing passive prostheses (including microprocessor-controlled devices), the introduction of power into a prosthesis provides the ability for the device to act, rather than simply react. As such, the development of a suitable controller that provides for stable and reliable interaction between the user and prosthesis is paramount. The user interface and control issue can be addressed with widely varying approaches and at widely varying degrees of invasiveness. The major categories of interface, in order of increasing invasiveness, are (1) mechanical sensory...
interface, (2) surface electromyography (EMG) interface, (3) implantable peripheral nervous systems (PNS) interface, and (4) implantable central nervous system (CNS) interface. Mechanical sensory interface (MSI) approaches use only sensors pertaining to the biomechanics of gait (i.e., as opposed to the physiology of gait), such as measurement of forces, torques, joint angles, and vertical orientation (i.e., inclination). Surface EMG, which is the approach used by actively powered myoelectric upper extremity prostheses, incorporates surface electrodes (often in the prosthesis socket) to extract command signals from the muscles in the residual limb. Some researchers have investigated the use of surface EMG control approaches for knee control in a lower limb prosthesis, including [22-26]. In the case of an upper extremity above-elbow myoelectric prosthesis, the combination of the biceps and triceps EMG provides a single bipolar signal, which is switched between the control of the terminal device and control of the elbow (i.e., does not enable simultaneous control of both joints). This approach would not be appropriate for an active knee and ankle joint leg, however, since locomotion requires simultaneous control of the knee and ankle. As such, an EMG approach would require at least two control channels (i.e., measurement from two sets of antagonist muscles). Implantable PNS approaches include the use of percutaneous electrodes implanted in the nerves, and/or the use of implantable capsules for extraction of the EMG signals, from which neural or EMG commands can be extracted. Finally, implantable CNS approaches utilize electrode arrays implanted in the cortex of the brain, from which motor commands can also be extracted. Presumably, the extent of control over the prosthesis would vary roughly inversely with the extent of invasiveness.

As with any medical device or procedure, one would optimally wish to incorporate the least invasive approach that achieves the desired specific aims, and as such, the proposed controller utilizes the non-invasive mechanical sensory interface approach. The lower limb, in particular, lends itself much more readily to non-invasive interface approaches than does the upper limb, since (1) the lower limb fundamentally interacts mechanically with the environment (i.e., the ground and the user) and (2) the tasks in which the lower limb engages are typically periodic in nature. Both of these qualities are leveraged in the proposed design.

The previously cited works on powered knees all incorporate a variation of echo control, in which the sound-side (i.e., unaffected) leg is instrumented to provide position commands (delayed by one half cycle) for the powered prosthesis. The obvious drawback to such an approach is that the sound-side (or unaffected) leg must be instrumented, which requires the user to don and doff additional equipment and associated wiring. The echo control approach also restricts the use of the prosthesis to unilateral amputees and also presents a problem for “odd” numbers of steps, in which an echoed step is undesirable. A more subtle, although perhaps more significant shortcoming of the echo-type approach is that suitable motion tracking requires a high output impedance of the prosthesis, which forces the amputee to react to the limb rather than interact with it. Specifically, in order for the prosthesis to dictate the joint trajectory, it must assume a high output impedance (i.e., must be stiff), thus precluding any dynamic interaction with the user and the environment, which is in turn contrary to the way in which humans interact with their native limbs.

IV. DECOMPOSITION-BASED CONTROL

This work offers an alternative control which employs the non-invasive MSI approach and obviates the need for sound side instrumentation. Figure 2 shows the structure of the proposed control system for the powered prosthesis. In the figure, the feedback loop for each joint incorporates the open loop joint dynamics and the decomposed passive impedance behavior, specified by the function $\tau_p = f_p(\theta, \dot{\theta})$. The input to the feedback loop is an active torque component, which is a function of the user input measured by the socket load cell and specified by the function $\tau_a = f_a(F_a)$.

$$\text{Fig. 2. Structure of the transfemoral prosthesis control system.}$$

One of the most significant aspects of the proposed controller is the inherent passivity of its structure. Specifically, because the behavior of the leg is decomposed into an active portion that is superimposed upon an underlying passive impedance, and because the active component is a function solely of the user’s input, the overall device behavior is guaranteed to be passive. As shown in Fig. 2, since the function $f_p(\theta, \dot{\theta})$ represents a passive mapping from $\theta$ and $\dot{\theta}$ to $\tau_p$ (by definition of the mapping), and since the natural leg dynamics from $\tau$ to $\theta$ and $\dot{\theta}$ are also passive, passivity theory guarantees that the closed-loop formed by these two mappings is also passive [27]. As such, the closed-loop portion of the device simply reacts to the user’s input, much the way a state-of-the-art passive prosthesis does. The only non-passive (i.e., active) portion of the controller is the component of torque directly resulting from the user’s input (i.e., from the socket interaction forces), over which the user has instantaneous and direct control. The proposed controller therefore creates a highly stable device over which the human has complete and reliable control. Alternatively stated, because of the structure of the proposed controller, the prosthesis simply reacts to what the user does to it, and as such will form a natural extension of the user.

In the remaining part of the paper, the active passive decomposition method will be explained and the joint torque
reference generation procedure is presented. Finally, simulation results based on measured biomechanical data validate the effectiveness of the proposed approach.

V. ACTIVE PASSIVE TORQUE DECOMPOSITION

Passive joint torque, $\tau_p$, is defined as the part of the joint torque, $\tau$, which can be represented using spring and dashpot constitutional relationships (passive impedance behavior). The system can only store or dissipate energy due to this component. The active part can be interpreted as the part which supplies energy to the system and the active joint torque is defined as $\tau_a = \tau - \tau_p$. This active part will be represented as an algebraic function of the user input via the MSI (i.e., socket interface forces and torques).

Gait can be considered a mainly periodic phenomena with the periods corresponding to the strides. Hence, the decomposition of a stride will give the required active and passive torque mappings for a specific gait mode. In general, the joint behavior exhibits varying active and passive behavior in each stride. Therefore, segmenting of the stride in several parts is necessary. In this case, decomposition of the torque over the entire stride period requires the decomposition of the different segments and piecewise reconstruction of the entire segment period. In order to maintain passive behavior, however, the segments cannot be divided arbitrarily, but rather can only be segmented when the stored energy in the passive elastic element is zero. This requires that the phase space can only be segmented when the joint angle begins and ends at the same value. Figure 3 shows the phase portrait of normal speed walking and the four different stride segments, $S_1, S_2, S_3$, and $S_4$. Thus, the entire decomposition process consists of first appropriate segmentation of the joint behavior, followed by the decomposition of each segment into its fundamental passive and active components.

The decomposition of each segment shown in Fig. 3 is converted to an optimization problem. In each segment of the stride, $2n$ data points are selected by sampling the angular position in equal intervals between its minimum and maximum and selecting the corresponding positive and negative angular velocities. In this work, the number of angular position samples for each segment, $n$, is set to be 100. The constrained least squares optimization problem given in (1) is constructed and solved.

$$\min_x \frac{1}{2} \|Cx - d\|^2 \quad \text{s.t.} \quad 0 \leq x$$

(1)

where $C$, $x$ and $d$ are defined in (2) and (3) respectively. The indexing of the joint angular position, angular velocity and moment samples are explained via the sketch in Fig. 4.
\[
\begin{align*}
\mathbf{x}_{3xl} &= \begin{bmatrix}
  k_1 \\
  k_2 \\
  \vdots \\
  k_{n-1} \\
  b_1 \\
  b_2 \\
  \vdots \\
  b_{2n-1} \\
  b_{2n}
\end{bmatrix}, \\
\mathbf{d}_{4xl} &= \begin{bmatrix}
  \tau_1 \\
  \tau_2 \\
  \vdots \\
  \tau_{2n-1} \\
  \tau_{2n} \\
  \tau_1 - \tau_2 \\
  \tau_2 - \tau_3 \\
  \vdots \\
  \tau_{2n-1} - \tau_{2n}
\end{bmatrix}
\end{align*}
\]

(3)

The matrix \( C \) consists of three sub-matrices, \( C_1, C_2 \) and \( C_3 \). \( C_1 \) is the main part responsible for the fitting of the spring and dashpot constants, \( k \) and \( b \). \( C_2 \) bounds the rate of change of the passive joint torque and ensures smoothness in the resulting passive joint torque, and \( C_3 \) is basically a row of penalty constants, \( \beta \), which penalizes large values of the spring and dashpot constants and thus limits the magnitudes of both. In this work, \( \beta \) is set to 0.1.

The origin of each virtual spring is also added to the optimization problem formulation as a parameter in order to obtain a tighter passive torque fit. Therefore, the optimization problem given by (2) will be solved iteratively for a range of values of spring origin constant, \( \alpha \). The solution with the least error norm is selected as the optimal solution.

The result of the above stated constrained optimization problem for segment 1 is shown in Fig. 5. As can be seen from the figure, the decomposed passive part is very similar to the joint torque, and thus it can be stated that the behavior of the joint is mainly passive. The result of the decomposition for the segment \( S_i \) is stored in \( R_i \) of the form given in (4).

\[
R_i = \begin{bmatrix}
  \theta & \dot{\theta} & \tau_{\text{pas}} & F_{S1} & F_{S2} & \tau_{\text{act}} \\
\end{bmatrix}_{20x6}
\]

(4)

\[
\tau_{\text{pas}} = C_1 \mathbf{x}
\]

(5)

VI. JOINT TORQUE REFERENCE GENERATION

The procedure presented in the previous section decomposes the joint torques into active and passive parts. The joint torque references for the control of the prosthesis are generated by combining this active and passive torques. There are two major challenges to be solved. Firstly, the correct motion segment must be selected. Secondly, after the motion segment is selected at each sampling instant a new joint torque reference should be generated using the discrete mappings for the active and passive torque parts.

A switching system modeling approach incorporating both discrete and continuous states is used for the reconstruction of the torque reference signal. The state chart shown in Fig. 6 will govern the discrete dynamics of the controller. Since the sequence of the segments is ordered (i.e., the direction of the motion for a specific gait mode does not change), each segment can transition only to the next one, where the transition guard function can be written as an inequality in terms of \( \theta \) and \( \dot{\theta} \). The transitions between segments take no time and the dynamics of the controller are governed by the \( \{ f_{pi} (\theta, \dot{\theta}); f_{ai} (F_S) \} \) pair at each sampling instant. The joint reference torque is

\[
\tau_{\text{ref}} = \tau_a + \tau_p = f_{pi} (\theta, \dot{\theta}) + f_{ai} (F_S)
\]

(6)
The decomposition algorithm presented in this work gives the result matrix, $R$, for each segment. The discrete data in $R$ is used to construct the joint torque reference for the continuous measurements of another trial in the same gait mode. At each sampling instant of the algorithm, the continuous measurements of another trial in the same gait is used to construct the joint torque reference for the result matrix, $R$, for each segment. The discrete data in $R$ is calculated as shown in (7) and stored in the vector $e_i$.

$$e_i = \sqrt{(\theta_{m} - \theta_{i})^2 + (\dot{\theta}_{m} - \dot{\theta}_{i})^2} \quad (7)$$

Then two elements of this vector with the least error norm are found and the passive knee torque reference is found as a weighted linear combination of the passive knee torques corresponding to these points. The reconstruction of the active knee torque part is similar where only $\{\theta, \dot{\theta}, \tau_{pas}\}$ is changed with $\{F_{S1}, F_{S2}, \tau_{pas}\}$.

VII. IMPLEMENTATION AND EXPERIMENTAL VALIDATION

In order to generate appropriate kinematic and kinetic data for the design of the controller and for testing of the approach, gait experiments were conducted in the Biomechanics and Sports Medicine Laboratory of the University of Tennessee at Knoxville. The experimental setup consisted of a 7-camera 240 Hz VICON MX system and 2 AMTI multiaxis force platforms. As is standard in such a setup, hip, knee and ankle angles were measured via the cameras, the ground reaction force was measured using the multiaxis force plates, and the joint torques and forces were computed in post-processing via inverse dynamics, based on inertial parameters estimated with lookup tables. The sampling rate for the cameras and the force plates were 100 Hz and 1000 Hz, respectively. Data collection consisted of four trials each slow walking (WS), normal walking (WN) and fast walking (WF). For the slow, normal, and fast walking, the subject was asked to walk at those respective rates, while an optical interrupt timing system measured the walking speed between the subject’s entrance and exit of the camera viewing volume. The minimum, maximum and average speeds for each class of walking speeds are shown in Table I. The experimental setup and the human subject are shown during a measurement trial in Fig. 7. Figure 8 shows the angular position, angular velocity and torque of the knee joint for a walking experiment with normal speed.

![Fig. 6. Structure of the switched control system.](image1)

![Fig. 7. Data collection for the demonstration of the proposed approach.](image2)

![Fig. 8. The angular position, angular velocity and torque of the knee joint for a walking experiment with normal speed.](image3)

<table>
<thead>
<tr>
<th>Walking Mode</th>
<th>Minimum Speed (m/s)</th>
<th>Average Speed (m/s)</th>
<th>Maximum Speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow</td>
<td>1.01</td>
<td>1.04</td>
<td>1.07</td>
</tr>
<tr>
<td>Normal</td>
<td>1.19</td>
<td>1.22</td>
<td>1.26</td>
</tr>
<tr>
<td>Fast</td>
<td>1.45</td>
<td>1.51</td>
<td>1.56</td>
</tr>
</tbody>
</table>
VIII. RESULTS AND DISCUSSION

The code for the decomposition algorithm and reconstruction is implemented in Matlab. One stride of the normal walking experiment data is decomposed and the result matrix \( R \) is acquired. Then using \( R \), the knee joint torque reference for another trial of the gait data is represented by the structure shown in Fig. 2. Instead of the socket interface forces, the ground reaction force and the hip moment are used.

The original knee torque and the reconstructed knee torque reference are shown in Fig. 9(a). As can be observed from the figure, the reconstructed knee torque trajectory is a good approximation of the original knee torque. Figure 9(b) shows the angular velocity versus the knee torque. The loops in the original knee torque versus angular position plot are also present in the reconstructed knee torque reference versus angular position plot. This shows that the active passive decomposition preserves the energetic behavior of the system.

![Knee torque and reconstructed knee torque](image)

**Fig. 9.** Original knee torque and the reconstructed knee torque reference versus time (a) and the angular position (b).

IX. CONCLUSION

This paper proposes a method for the control of a powered transfemoral prosthesis. The approach is based on the decomposition of knee torque into 1) a fundamentally passive component, which is an algebraic function of joint angle and angular velocity, and 2) an active component, which is an algebraic function of the measured forces between the user socket and the prosthesis. The paper presents a method for decomposing measured data into these passive and active algebraic functions, and describes a method for generating real-time torque trajectories based on these functions and the measured joint angles and angular velocities and interaction forces. The effectiveness of the approach is demonstrated on measured biomechanical data. Future work will utilize the proposed approach for the real-time control of the powered transfemoral prosthesis.

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