An Artificial Gastrocnemius for a Transtibial Prosthesis

Ken Endo, Eric Swart and Hugh Herr

Abstract— **A transtibial amputee does not have a functional gastrocnemius muscle, which affects the knee as well as the ankle joint.   In this investigation, we developed a transtibial prosthesis comprising an artificial gastrocnemius mechanism as well as a powered ankle-foot device.   A pilot study was conducted with a bilateral transtibial amputee walking at a self-selected speed.   The trial compared muscle electromyography and metabolic cost data for the amputee while using the active gastrocnemius prosthesis and a conventional Flex-Foot prosthesis.   The experimental data showed that the compensation for ankle-foot and gastrocnemius function offered by the active device resulted in a reduced metabolic cost for the amputee participant.**

I. INTRODUCTION

Today's commercially available transtibial prostheses are completely passive during stance phase, so their mechanical properties in this period remain fixed over varying walking speeds and terrains. Such transtibial prostheses are typically composed of elastic bumper springs or carbon composite leaf springs that store and release energy during the stance period, e.g. the College Park or Flex-Foot [1].

Unlike conventional ankle-foot prostheses, however, the ankle joint performs an enormous amount of positive mechanical work in pushing off the ground during the terminal stance phase. To replicate the human ankle-foot complex and improve clinical efficacy, a prosthetic ankle-foot mechanism must have active components to control the joint impedance and generate non-conservative motive power during stance. At the same time, the prosthetic device should not exceed the weight of the missing limb. There are several researchers who have developed powered ankle prostheses with active components such as electrical motors or pneumatic actuators [2].

Although some researchers have pursued ankle-foot prostheses with active control, there have been no studies that consider how the gastrocnemius muscles affect the knee joint. A transtibial amputee has lost part of his or her gastrocnemius muscles and his or her ankle foot complex. Gastrocnemius muscles are large biarticular muscles that span the knee and ankle joints and generate torque at both joints. In order to

replicate the normal walking function of a human leg, a transtibial amputee would need both an active ankle-foot prosthesis and a knee flexor that acts like the gastrocnemius muscles. Unfortunately it is generally difficult to estimate how a gastrocnemius muscle-tendon unit behaves due to limitations of in-vivo measurement techniques.

Ishikawa [3] determined the fascicle length change of the soleus and medial gastrocnemius muscles during walking through *in-vivo* measurements using an ultrasonographic apparatus. The fascicle length of the medial gastrocnemius muscle does not change during the early and mid stance phases, indicating that it is the Achilles tendon that is lengthened or shortened instead of the muscle fascicle. Endo and Herr [4] used a quasi passive muscle model of a clutch and spring in series to develop a musculoskeletal model with a minimal number of active actuators. Optimization of the model parameters resulted in a reasonable mechanical economy [5] and muscle-tendon complex behavior that resembled Ishikawa's measurements.

In this paper, we assumed a linear spring and clutch in series could mimic the gastrocnemius function during walking at a self-selected speed. We optimized the engagement time and spring stiffness so that a musculoskeletal model of a transtibial amputee could track the biological knee and hip joint torques with minimum positive mechanical work done by muscle actuators. We then developed an artificial gastrocnemius (AG), combined with a powered ankle-foot prosthesis (PAP), and conducted a pilot trial on a bilateral transtibial amputee.

II. OPTIMIZATION

A previous study showed that the behavior of the gastrocnemius muscle can be mimicked by a spring and clutch in series [4]. This result indicates that a mono knee flexor composed of only a clutch and linear spring should improve the walking behavior and metabolic efficiency of transtibial amputees during walking. In this section we first describe a musculoskeletal model of a transtibial amputee based on Endo's full leg model and then show the optimization strategy to find the best stiffness and engagement time for the knee flexor.

A. Musculoskeletal Model

Fig. 1 shows a two-dimensional musculoskeletal leg model for a transtibial amputee composed of eight series-elastic clutch/actuator mechanisms. In the PAP at the ankle joint, a linear spring and actuator in series as well as a parallel spring are attached [2]. In this study, we assume that the active component of the PAP allows it to mimic the behavior of the

Ken Endo is with the Biomechatronics group, Media lab, Massachusetts Institute of Technology, Cambridge, MA 02139 USA phone: 617-253-2941; fax: 617-253-8542, kene@media.mit.edu

Eric Swart is with the Biomechatronics group, Media lab, Massachusetts Institute of Technology, and Harvard medical school, MA, USA.

Hugh Herr is with the Biomechatronics group, Media lab, Massachusetts Institute of Technology and Harvard/MIT Division of Health Science and Technology, Cambridge, MA 02139 USA.
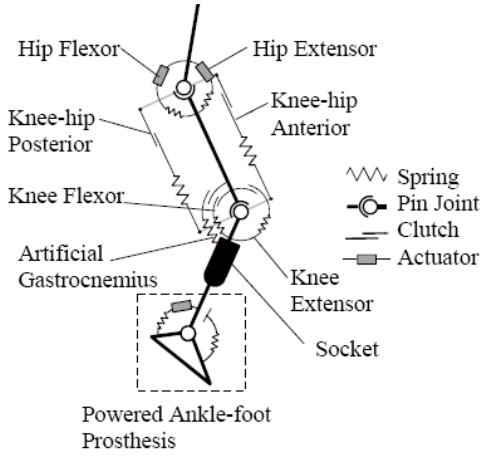
Fig. 1 Musculoskeletal model of a transtibial amputee with a powered ankle-foot prosthesis (PAP) and artificial gastrocnemius (AG).

ankle foot complex and consequently perform an optimization only for the knee and hip joints. The hip extensor and flexor are active components with unidirectional actuators and the other units around the knee and hip joints are composed of clutches and springs. When a clutch is disengaged, the joint rotates without any resistance from spring. When a clutch is engaged, the clutch holds the series elastic spring at its current position and the spring begins to store energy as the joint rotates in a manner comparable to a muscle-tendon unit where the muscle generates force isometrically. The knee and hip joints are composed of five monoarticular and two biarticular tendon-like springs with series clutches or actuators. We assume that all monoarticular units are rotational springs and clutches while all biarticular units act about attached pulleys with fixed moment arm lengths. Moment arms for biarticular units are taken from the literature [6].

B. Optimization Strategy

The model has a total of 12 series-elastic clutch parameters for the knee and hip joints: seven spring constants and five distinct times when clutches are engaged. These parameters and actuator movements define joint torque. Fitting this model to biomechanical data involved determining the spring rates of the model's seven springs and the engagement times of the five associated clutches. The biomechanical data for this procedure was obtained from one participant (male, 81.9kg weight, 1.89m height, 1.27m/s walking speed) using a VICON motion capture system. The data collection process is described in [7].

An optimization procedure was used to fit the parameters of the model to the biomechanical data. We used the same strategy as Endo's optimization [4] except that the optimization was conducted only for the knee and hip joint. The following cost function $f(x)$ was used:

$$f(\mathbf{x}) = a \sum_{j=1}^{100} \sum_{i=1}^{} \left(\frac{\tau_{bio}^{i,j} - \tau_{sim}^{i,j}}{\tau_{bio}^{j,\max}} \right) + W_{act}^{+} \qquad (1)$$

where $\mathbf{x}$ is a vector of the 12 parameters mentioned above, $\tau_{bio}^{i,j}$ and $\tau_{sim}^{i,j}$ are the angular torques applied about joint j at the ith percentage of the gait cycle obtained from biological

data and the model, respectively, and $\tau_{bio}^{j,\max}$ is the maximum biological torque at joint j. Further, W_{act}^{+} is the total positive mechanical work performed by the two muscle actuators during the gait cycle and a is a constant weighting coefficient. The cost function $f(x)$ was minimized subject to the constraint that the net torque at the hip be equal to the target biological torque. The model hip torque was produced by an agonist/antagonist pair of hip actuators while the target biological hip torque was derived from an inverse dynamics calculation.

The hip constraint forces the model to precisely track the biological hip torque throughout the gait cycle. Since the model includes biarticular units capable of transferring energy between joints, the value of the weighting coefficient a in the cost function (1) influences the model's capacity to emulate biological knee torque with quasi-passive elements. Therefore we set the weighting coefficient a equal to a value (50) that was large enough that further increases in a did not result in a larger R^2 value at the knee.

C. Results

Fig. 2 shows the knee torques from the musculoskeletal model simulation and biological data. Since R^2 is 0.96, we see that the gross features of human walking at self-selected speed may be captured using only quasi-passive elements including the AG. The simulated hip torque perfectly tracks the biological data due to the agonist/antagonist pair of active elements.

The contribution of each spring is shown in Fig. 3. The AG is engaged from 17% to 58% of the gait cycle, which is roughly the same as the activation estimated from gastrocnemius electromyography (EMG) measurements [8]. The optimized stiffness of the AG spring was 92.9 Nm/rad. These results were implemented in the design of the actual AG and its controller.

III. CLINICAL STUDY

A. Prosthetic Device for Transtibial Amputees

We developed a prosthetic device for a transtibial amputee as shown in Fig. 4. This device consists of the PAP and the AG. The PAP was designed based on [2].

The AG mechanism is shown in Fig. 5. This device is composed of a clutch, motor and two coil springs. The spring stiffness is set to 50,000 N/m with 0.045 m moment arm (approximately 101.3Nm/rad). The small motor is used not for generating the knee torque but rather for engaging/disengaging the clutch. When the clutch is engaged the spring starts to store energy as the knee extends, resulting in torque being generated about the knee joint. When the clutch is disengaged no torque is applied at the knee joint as the spring stores no energy.

B. Control

Control of the PAP was based on [2] while control of the AG was based on a simple state machine. The knee flexor state machine controller has three states as shown in Fig. 6.
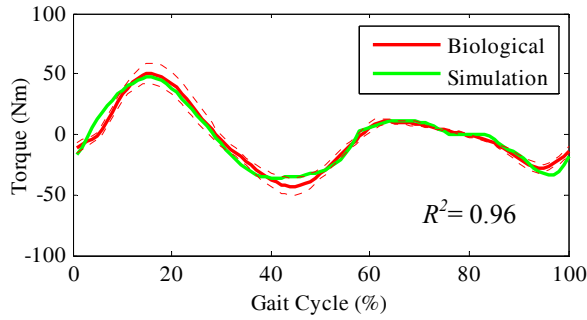
Fig. 2 Knee torque from the optimization result. Green and red curves are the result of the simulation and biological torque, respectively. Dotted curves are one standard deviation from the biological data.
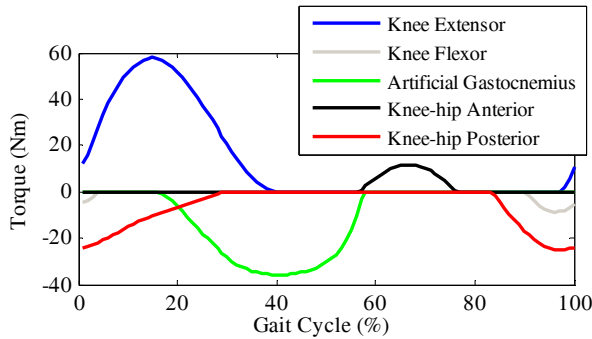

Fig. 3 Contribution from each spring. The Artificial Gastrocnemius (AG) works from the mid stance phase until toe off, similar to the biological gastrocnemius.

The AG is engaged from the time of maximum knee flexion (around 20% of the gait cycle) until toe off (around 58% of gait cycle). The state transition is dependent upon maximum knee flexion; the clutch engages only if the knee reaches maximum flexion in stance phase.

C. Experimental Protocol

The clinical evaluation was conducted at MIT (Cambridge, MA) and was approved by MIT's Committee on the Use of Humans as Experimental Subject (COUHES). A written informed consent was obtained from a participant before data collection began.

One bilateral transtibial amputee participated in the study. The participant was generally in good health and was experienced in prosthesis ambulation. The participant was asked to commit to two experimental sessions.

In the first session, the participant was asked to walk 10 m 10 times at 1.25 m/s with our device and with the Flex-foot. Joint angles were captured with a VICON motion capture system. We also measured EMGs of the rectus femoris (RF), bicep femoris long head (BFLH), and vastus lateralis (VAS) muscles from the skin surface.

In the second session, the participant was asked to walk on a field track for 8 minutes at his self-selected speed with our device, and then, after 5minutes of rest, he was asked to walk at the same speed with the Flex-Foot for 8 minutes. Metabolic cost was measured by indirect calorimetry using an open circuit respirometry system (K4b², COSMED). During resting and walking, we measured the rates of oxygen
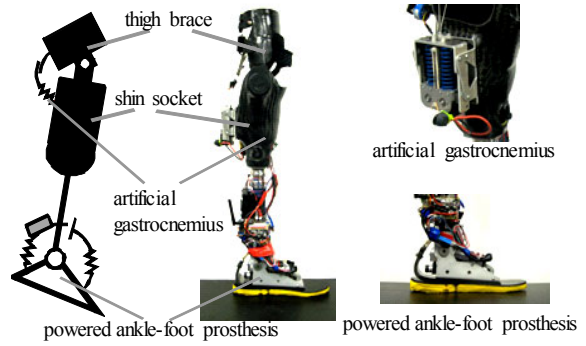

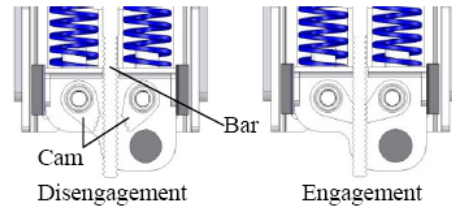Fig. 4 Transtibial prosthesis with the PAP and the AG.

Fig. 5 Clutch mechanism. When cams are disengaged, the bar can move up and down without any resistance. When cams are engaged, the bar moves up under spring force, (but down without the force).
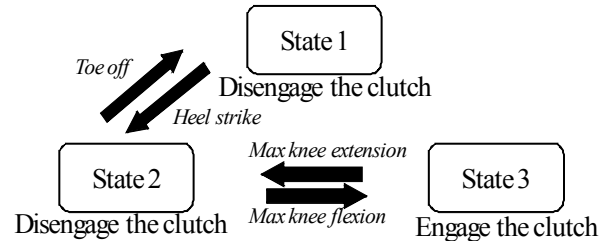

Fig. 6 The AG state machine controller. The clutch is engaged only after maximum knee flexion in the stance phase. The clutch is disengaged automatically when the spring releases all stored energy.

consumption and carbon dioxide production and calculated metabolic rate [9].

D. Result

After 10 walking trials at 1.25m/s in the first clinical study session, the data set was processed for analysis. The observed walking speeds in the 10 trials were all within 5%.

Fig. 7 shows the AG state transitions, knee angle, and torque contribution of the AG in three-step steady state walking. At heel strike the controller transitions from state 0 to state 1 and at the maximum knee flexion in stance phase the controller transitions from state 2 to state 3 with the clutch in the AG being engaged. After toe off, no torque is generated by the AG until the maximum knee flexion in the next stance phase. Fig. 8 shows EMG measurements from VAS, BFLH and RF. The blue and red curves are EMG measurements with Flex-Foot and the PAP/AG, respectively. There is no significant difference in VAS EMG measurements between the two despite the flexion torque generation of the AG at the knee joint from 20% to 62% of the gait cycle. On the other hand, the BFLH EMG shows extra excitation when the participant uses the PAP/AG because the BFLH works as a
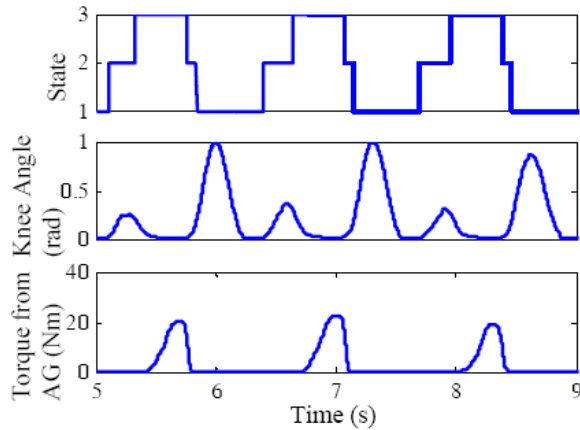
Fig. 7 State transition, knee angle and torque generated by the AG during a walking trial. The data includes three steady steps.
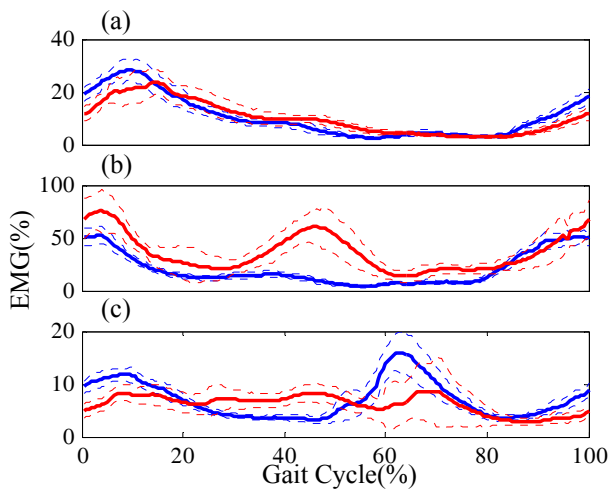

Fig. 8 EMG measurenents of (a) vastus lateraris, (b) bicep femoris long head, and (c) rectus femoris taken from the skin surface. Blue and red curves are EMG measurements from standard prosthesis and the PAP/AG, respectively. 100% of EMG activation indicates the maximum contraction of the muscle.

hip extensor and its hyperextension at the hip joint helps the knee joint extend in order to cancel the AG torque. Finally, the EMG from RF is decreased when the participant uses the PAP/AG. This is because the PAP pushes off the ground and the AG flexes the knee joint at toe off, helping the swing leg move forward.

Fig. 9 shows the ankle and knee angles of the participant with the PAP/AG and Flex-Foot as well as a normal human. There is little difference among the three knee angle curves but the ankle joint curves are distinct. Obviously the Flex-Foot does not have any powered plantar flexion during the terminal stance phase as it is a passive element. The PAP/AG does have powered plantar flexion but its maximum plantar flexion angle is still smaller than in normal human walking. This implies that an intact ankle pushes off the ground harder than the PAP.

After the second session of the clinical study, metabolic cost was calculated while walking at a self-selected speed
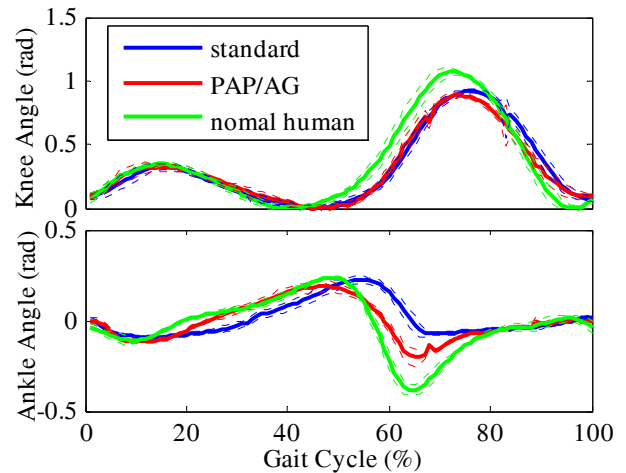

Fig. 9 Knee and ankle angle of trials with standard prosthesis, PAP with AG, and normal human walking.

with the Flex-Foot and the PAP/AG. The metabolic cost was found to decrease by 6.3% when the participant walked with the PAP/AG. A greater metabolic reduction is expected in the future if the powered plantar flexion of the PAP is improved.

IV. CONCLUSION

In this paper, we assumed a linear spring and series clutch can mimic the behavior of the gastrocnemius. We optimized the engagement time of the clutch and spring stiffness for the hardware design implementation. We developed the PAP with the AG and conducted a pilot trial with a bilateral transtibial amputee. In this trial we measured muscle electromyography and the metabolic cost of walking at a self-selected speed with the active device and a passive-elastic, Flex-Foot prosthesis. The measured data showed that our device decreased the metabolic cost of walking at a self-selected walking speed by 6.3%.

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