# BIONIC RUNNING FOR UNILATERAL TRANSTIBIAL MILITARY AMPUTEES

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## ABSTRACT

A team from the United States Military Academy and Walter Reed Medical Center, in collaboration with private industry partner, SpringActive, Inc., have designed, built, and demonstrated a first of its kind motor powered, single board computer controlled, running prosthesis for military transtibial amputees. This paper presents the design and initial results of the new prototype, which includes successful testing with one unilateral transtibial Military amputee running at 3.6 m/s (8 mph) on a treadmill. The 2011 prototype described in this paper is intended to support a unilateral transtibial Military amputee on an Army Physical Fitness Test which includes a 2 mile timed run on a level ground.

## 1. INTRODUCTION

A team of faculty, staff, and cadets from the United States Military Academy, with support from the clinicians at Walter Reed Medical Center, in collaboration with private industry partner, SpringActive, Inc., have designed, built, and demonstrated a first of its kind motor powered, single board computer controlled, running prosthesis for military transtibial amputees.

Despite several versions of passive and active transtibial prostheses in various stages of technological readiness levels, none currently provide the military transtibial amputees a single component solution for returning to full duty. The desired outcome is to develop a robust walk-run, all terrain, all-weather, quiet device that requires one battery charge per day and with a total weight less than the replaced limb. The device must be metabolically efficient and kinematically beneficial. The device will integrate with individual soldier equipment. The design will allow simple removal of active components, such as the computer and the motor, for occasions when a passive device provides sufficient performance or when Military operations such as airborne and waterborne operations dictate.

### 1.1 Passive Prostheses

The limitation of passive prostheses ultimately stems from their constant stiffness characteristics and inability to supplement the potential energy stored during the stance phase of the gait cycle.

The prevalence of activity-specific foot-ankle prosthetics highlights the consequences of the constant stiffness characteristics for passive devices. These purpose-built passive prosthetics are optimized for a desired level of energy efficiency and stability for a given activity, in a given environment, and under the constraint of the mechanics properties for the components that store and release potential energy during the stance phase of the gait cycle. Varying the design constraints is critical to transitioning between various gait speeds and conditions because they affect the overall compliance and energy regeneration capabilities of the prosthetic limb. The amount of compliance or effective stiffness of the limb is effectively a compromise between metabolic efficiency and stability for a given condition of user biometrics, gait speed, and running surface (Daley and Usherwood, 2010). The transtibial amputee is often thus confronted with the decision to interchange between several prosthetic feet or use one, likely suboptimal for several environments. These decisions become more challenging for Military amputees who may be in austere environments under unpredictable operational conditions.

There exist unique solutions using active controls to vary the kinematics of the lower limb prosthesis. One such example is the Propio ankle by Ossur, which uses a computer based motor to control ankle angle based on position in gait cycle, gait, and other external conditions (Proprio Technical Manual, 2009). Unfortunately, it does not provide any additional power to the gait cycle, which is a limitation of all passive prostheses. The human gait cycle has an energy deficit for a 70kg person ranging from 36J/step while walking, to 100J/step while running, which is provided in non-amputees by the lower limb muscle network (Hitt et al., 2010). This energy deficit attributes to greater energy consumption while walking

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14. ABSTRACT
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and running for transtibial amputees, with tests indicating that they can expend up to 35% more energy while walking (Rao et al., 1998).

### 1.2 Active Prostheses

Active prostheses serve to replicate the lower limb muscle-tendon system in the sagittal plane with a combination of a linear springs and actuators. The linear springs regenerate energy during the stance phase, while the actuator effectively decreases the equilibrium length of the spring after dorsi-flexion of the stance phase to supplement the amount of potential energy it stores and releases. Decreasing the effective equilibrium length of the spring increases the displacement of the spring from equilibrium during dorsi-flexion, which by Hooke's Law, stores additional energy in the spring equal to the product of the spring rate and difference in the squares of the effective displacements. Brushless (Blaya and Herr, 2004) and brushed motors (Hitt et al., 2010), in addition to pneumatic muscles (Kao et al., 2010; Versluys et al., 2008; Hitt et al., 2010) provide the energy required to increase the spring potential in the stance phase of both prostheses and orthotics. The benefit of storing the potential energy during dorsi-flexion is it effectively provides a motor power multiplication factor of up to 3. That is, the muscle-tendon system is able to maintain a 450W power output while using a 150W motor while walking (Hitt et al., 2010).

### 1.3 Transitioning from Walking to Running

As previously discussed, there is nearly a threefold increase in energy required during the gait cycle to transition from walking to running. In addition, the stride frequency nearly doubles, necessitating close to six times the power output from the muscle-tendon complex while running when compared to walking. What further exacerbates this power requirement is the increased effort required to overcome actuator inertial characteristics, which were found to account for a significant 18% (80W) of the muscle-tendon output simply during walking (Hitt et al., 2010). Not only does the gait cycle frequency increase in the transition to running, but the acceleration of actuator components within each gait cycle increases as a result of the changes in kinematics between walking and running. While walking, the stance phase accounts for 62% of the gait cycle, which decreases to 36% as the gait transitions from walking to running (Novacheck, 1998). The effect of an increased gait cycle frequency and reduced time for storing and releasing energy per cycle is a dramatic elevation in the power required to overcome the inertial effects of the actuators and supply the needed quick power spike at push-off.

Another important consideration in the transition to running is the biped's natural neuromuscular adaptation while changing gait from walk to run. Comparisons between gait speed and leg stiffness indicated a positive correlation between the two. However, clinical trials determined that increases in gait speed caused solely by increases in stride length had little effect on the leg stiffness (Farley and Gonzales, 1996). Additional trials determined a strong correlation between stride frequency and leg stiffness (Farley and Ferris, 1998).

When comparing leg stiffness to surface compliance at constant speeds in human subjects (Ferris et al., 1998) and guinea fowl (Daley and Biewener., 2006; Daley and Usherwood, 2010; Biewener, 2007), results demonstrated that biped neuromuscular control maintains a constant effective leg stiffness, which accounts for the leg stiffness in series with the surface stiffness. While some experiments determined that the majority of the neuromuscular stiffness adjustments occurred at the knee joint (Burdett, 1982), there exists convincing evidence that neuromuscular adaption exists at the ankle as well. Active transtibial prosthesis testing of subjects loaded with different weights at a specified gait demonstrated that tendon stiffness adjustments affected actuator efficiency (Hitt and Sugar, 2010). Clinical trials conducted to determine the reduction in muscle-tendon loading using an active orthosis demonstrated that human subjects were also likely to reduce leg stiffness (Ferris and Farley, 1998) when supplemented by an actuator. However, most interesting was the change in ankle kinematics when using an active orthosis. Research suggests that the neuromuscular response at the ankle seeks to maintain a specified moment within the joint, to the point that the subject will change his or her gait to maintain that ankle moment (Kao et al., 2009). Thus, in order to prevent disruption to the transtibial amputee's running gait, the actuation of the active ankle prosthesis must accurately mimic the kinematics and kinetics of the ankle. This requires a robust control system that allows the active prosthesis to properly mimic the ankle moment profile for given user-specific gait kinematics.

### 1.4 Tendon-Muscle Control Schemes

Finite state machine control systems exist for active transtibial prosthesis in various levels of robustness. Single state control schemes estimate the start and frequency of a specified gait cycle, and control the effective length of the spring based on an assumed gate profile (Oymagil et al. 2007). However, due to the variability in gait profile under realistic conditions, the limitations of this simple state machine overcome the inherent stability of the compliance offered by series actuator spring muscle tendon system. This limitation is particularly critical when transitioning from walking to running as the magnitude and timing of dorsi and plantar flexion differ significantly between the two gaits.

Robust finite state control systems adapt to changes in the gait by controlling invariant parameters within different portions of the gait. One implementation (Bohler et al., 2008) divides the stance phase into five zones based on the ankle angular position, and controls either ankle torsional stiffness or angular velocity within each zone. This method seeks to control angular velocity when the ankle is plantarflexing, and stiffness when the ankle is dorsiflexing during the stance phase. The optimal angular velocity and stiffness parameters vary based on the position within the gait cycle, which the control method addresses by defining specific parameters for three separate plantarflexion zones and two separate dorsiflexion zones, accounting for five unique parameters for a given gait cycle. As the optimal angular ankle angular velocity and stiffness profiles also vary under realistic conditions, this control system uses state logic to determine the amputee activity given the conditions, and maps the five optimal parameters to that specific activity.

Impedance Control methods (Shaeffer and Hirzinger, 2002) provide actuator inputs to change the effective inertia, damping, and stiffness characteristics of the robotic limb. Though this method requires knowledge of the reaction forces and moments on the ankle, it offers flexibility within the system to adapt to changing environmental conditions. Applications in active ankle foot orthotics use state logic to determine different zones within the stance and swing phases to vary the effective impedance of the limb (Blaya and Herr, 2004). Such applications have been effective in stabilizing plantarflexion during the beginning of the stance phase to prevent toe slap, while still allowing sufficient dorsiflexion during the beginning of the swing phase to prevent toe drag, thereby reducing hip circumduction.

The critical function within finite state control methods is properly estimating the dynamic state. Unfortunately, robust state estimation algorithms are increasingly complex and computationally expensive, potentially limiting their effectiveness during real time control. One simplification to state estimation algorithms is to measure the amputee's electromyographic (EMG) signals directly. Studies suggest there is a correlation between the amputee's EMG signals and the desired actuator function (Ferris, 2006; Novacheck, 1998). However, the dominant factor in the effectiveness of myoelectric-based control is the system's ability to reliably acquire and process an accurate EMG signal, which still remains a significant challenge (Holgate et al., 2009).

A novel approach to overcome the limitations of finite state machine controls is to define gait parameters invariant of amputee activity in a continuous control system. One implementation couples tibia angular velocity and angular position in the sagittal plane to identify the desired ankle angle and moment, adaptive to any gait kinematic (Holgate et al., 2009). Using a single rate gyro, this method captures tibia velocity in the inertial earth fixed reference frame, determines an angular position, and identifies the corresponding stride length and position within the gait cycle. The challenge in the tibia based control system is integrating the rate gyro's angular velocity signal, which is prone to offset and drift. Rather than augment the rate gyro with an accelerometer to perform complementary or Kalman filtering, the tibia based control method implements a novel analog signal processing technique that tenuates the drift. This method is more robust than the aforementioned dual signal processing techniques, as it provide a variable neutral reference based on gait kinematics rather than constant neutral reference defined in the direction of acceleration due to gravity. Thus, the neutral tibia position is referenced in the center of the stance phase which directs more in line with the normal to the walking surface. This allows the system to adapt to uneven terrain and varying amputee activities without state logic.

## 2. IMPLEMENTATION

### 2.1 Mechanical Design

The running prosthesis was essentially a redesign of the SPARKy walking active ankle prosthesis (Hitt et al., 2010). Refining the walking prosthesis for running required component modification to resist the increased dynamic loads on the structural components and provide the increased actuator power necessary for running. Refer to Fig. 1 for an illustration of the embodiment design and Fig. 5 for a comparison of the walking and running feet designs.

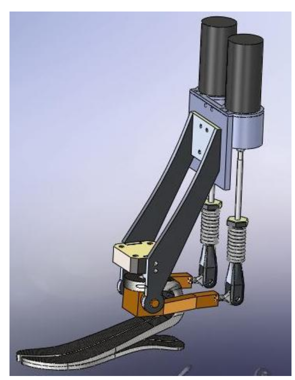

Fig. 1. West Point bionic running foot design incorporates a dual actuator, dual spring muscle-tendon system

Simulation and testing revealed 4200N as the objective peak load on the robotic muscle-tendon system for a specified foot-ankle configuration.

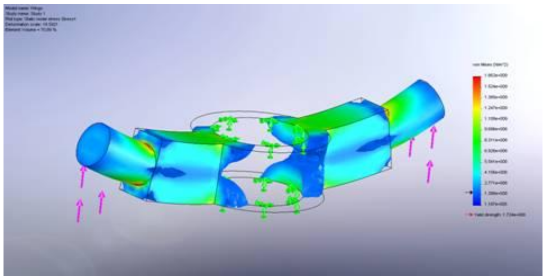

Fig 2. Finite element analysis of the muscle-tendon interface component revealed excessive internal stresses, necessitating redesign for the running foot.

Finite element analysis of the components indicated three critical areas for redesign: the robotic muscle-tendon interface; the tibia-ankle interface; and the tendon-ankle interface. The robotic muscle-tendon interface originally transmitted inputs from the single linear actuator to both springs, resulting in significant internal moments and shear loads in the interface component. Refer to Fig. 2 for a depiction of internal stresses of the muscle-tendon interface component. To mitigate the excessive normal and shear stresses at the interface surfaces, the team included a second actuator such that each actuator interfaced with a single spring.

Increased external moments exerted on the revolute joint at the ankle caused excessive shear loads on the bolted tibia-ankle interface adjacent to the ankle joint. To mitigate the risk of failure at the revolute joint, the team replaced the bolted connection with a unitary, solid component. Refer to Fig. 3 for a comparison of the tibia-ankle interface component for the walking and running feet.

With increased internal moments in the ankle-tendon interface component, the team removed the stress concentrations in the component, with emphasis near the revolute ankle joint. The redesign also incorporates an increased second area of moment of the interface component in the direction of the applied moment.

Fig. 3. Comparison of critical system interfaces between the running foot (foreground) and the walking foot.

The 300-400W actuator power requirement mandated replacement of the Maxon RE40 based actuator system rated up to 150W. The team evaluated three alternatives with similar power characteristics: a single Maxon RE75 brushed motor actuator; a dual Maxon RE40 actuator system; and a single EC40 brushless motor actuator system. Evaluation parameters included the complexity of the motor control required, the overall weight and inertial characteristics of the components, and the clearance provided for the lower limb socket. The RE75 allowed for the simplest motor control algorithm, but provided less than optimal socket clearance due to the large diameter of the motor casing. In addition, the RE75's rotating inertia was double to quadruple of the inertia of the other two alternatives, which would have limited its efficiency. Furthermore, the single actuator system would have generated excessive internal moments in the muscle-tendon interface component. The dual actuator system required additional control requirements to synchronize the two motors, but mitigated risk of muscle-tendon component failure. Additionally, though inertial power requirements were significantly less than the RE75 system, they were double the EC40 brushless motor actuator power requirements. However, the EC40 brushless motor system control had inherent instabilities during actuator direction reversal, and tested methods proved unreliable at that particular transition in the gait cycle. Therefore, the RE40 dual system proved the best compromise between motor control requirements, efficiency, and compatibility with the amputee.

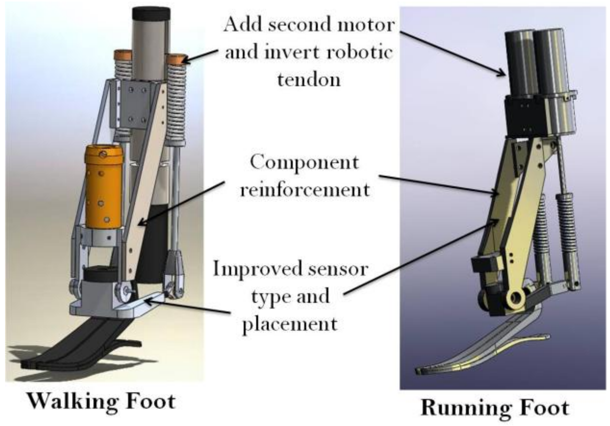


Fig 5. Additional power and load requirements mandated component redesign at the muscle-tendon, tendon-ankle, and tibia-ankle interfaces.

**2.2 Control System Design**

The controller hardware is composed of a PC104 microprocessor with a Sensoray DAQ unit attached. Utilizing control code downloaded from MalLab, the PC104 drives two custom brushed motor position controllers, see Fig. 6.

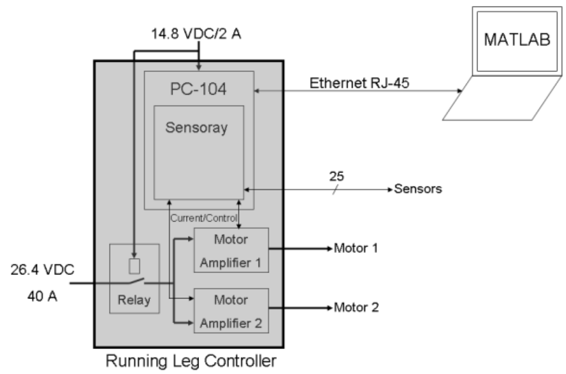


Fig. 6. Control Hardware Diagram.

The electronics are portable and are powered by a 26.4V battery pack. An Ethernet connection to the Matlab PC is used to download the control code as well as collect the data measured from the robot.

The controller software logic for running is based strongly on the team's prior work with walking gait, and the tibia controller (Holgate et al., 2009). The advantages of this control approach for both walking and running are that its modeling is not based upon time, and is inherently continuous, i.e. no state base decision logic is used. In addition, the method detects user movement intent 1000 times every second.

Although based upon the same methods used in development of the original tibia based control model, the development of the running controller required significant modification for the task of running. Due to these distinctive differences in coordinated motion, a modified motor reference pattern distinctive to running gait was required. Refer to Fig. 7 and Fig. 8 for a sample ankle-gait profile and ankle moment-gait profile determined by stride length, respectively. Statistical analysis of running gait kinematics determined an appropriate ankle angle and ankle moment for a given position in the gait cycle, where the position is defined as a percent within the gait cycle.

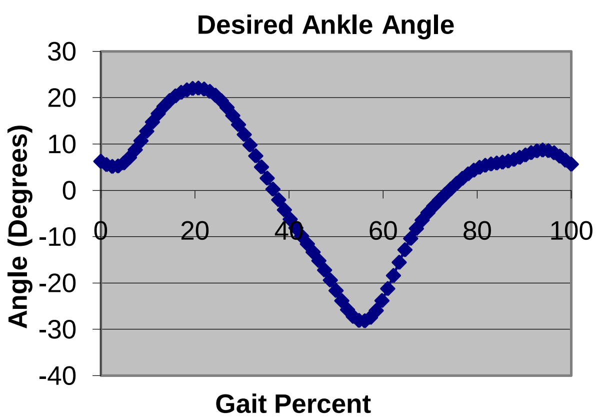


Fig. 7. Desired ankle kinematics as a function of the position within the gait cycle for a specified stride length.

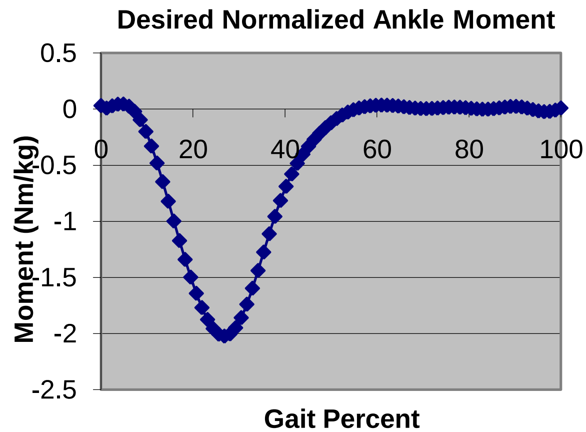


Fig. 8. Desired ankle kinetics as a function of the position within the gait cycle for a specified stride length.

Using a rate gyro to measure the motion of the tibia, percent gait and stride length can be determined. Corresponding gait percentage and stride lengths determine the current desired moment at the ankle, which is reflected in the displaced length of the spring. The position of the actuator nut determines the effective equilibrium length of the spring, which is controlled to maintain the appropriate spring displacement. The lever position is a measure of the ankle angle, and indicates the vertical displacement of the tendon-ankle interface from neutral, i.e., an ankle angle of 90 degrees. See Fig. 9 for a sample motor position-gait profile.

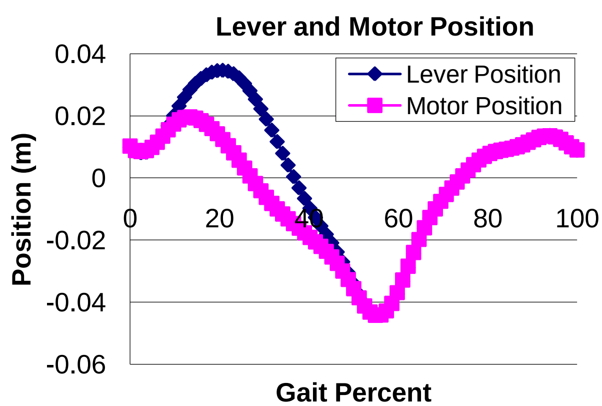


Fig 9. Desired motor position and tendon-ankle interface (lever) position in the gait cycle for a specified stride length.

The team then developed a two dimensional mapping of the motor position as a function of stride length and position within the gait cycle. See Fig. 10 for a sample look-up map. This two dimensional array is stored within the control system memory, providing a lookup table for the desired motor position indirectly as a function of the measured tibia angular rate and calculated tibia angular position.

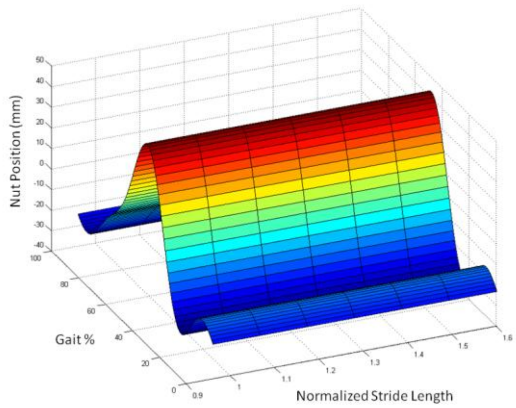


Fig. 10: Displacement of the spring attachment nut (mm) as a function of stride length and gait percent.

### 3. RESULTS

The team conducted clinical tests with an 80kg unilateral transtibial amputee at the United States Military Academy 13-25 April 2010. See Fig. 11 for the testing apparatus. IRB approval was obtained from the United States Military Academy.

Fig. 11. Single subject treadmill running tests 13-25 April.

During tests, the amputee, a United States Army Special Operations noncommissioned officer, sustained a moderate running gait speed up to 3.6 m/s (8 mph), see Fig. 12. In one test, after reaching a speed of 3.6 m/s, the subject jumped off the moving treadmill coming to an abrupt stop. The continuous tibia-based control effectively adapted to the significant and rapid change in gait kinematics and kinetics within one sampling cycle.

Fig. 12. Subject running on a treadmill.

The clinical trials validated the tibia-based control algorithm for running and allowed the team to refine simulation models for accurate analysis of required actuator power input in future designs. Though the running foot proved effective for gaits speeds up to 3.6 m/s, testing also highlight key areas of emphasis for the next iterative design process with the objective of sustained overground running. Visual inspection of the subject's gait indicated hip circumduction likely caused by an underpowered actuator system. Model refinement based on the test results indicated that the actuators delivered up to 550W of the 590W peak power required for the subject's weight and gait speed. See Fig. 13 for the refined power requirement analysis based on gait position. See Fig. 14 for the measured (current and voltage) motor input power. Thus, future redesign efforts will focus on reducing the power required to overcome inertial characteristics of the actuator system and structural components, as well as increase the power capacity of the actuator system. The solution emphasis for the next design is incorporating a brushless motor actuator system that halves the actuator inertial properties and more than doubles the actuator's nominal power capacity to 800W.

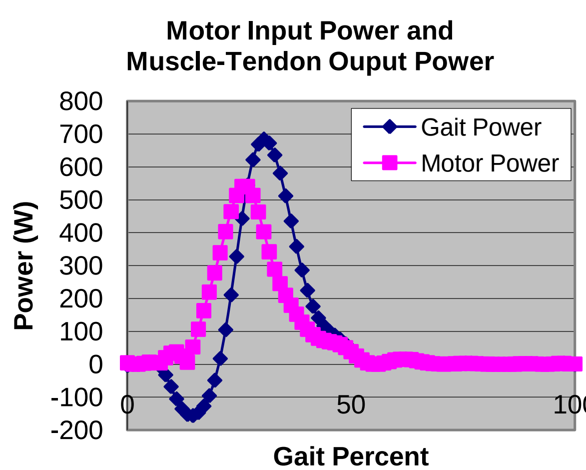


Fig. 13. Motor and muscle-tendon power capacities for 80kg subject running at 3.6 m/s.

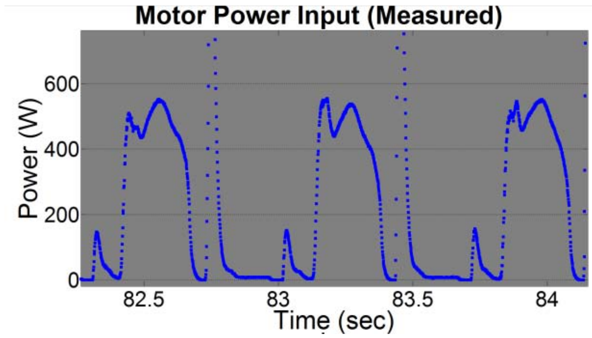


Fig. 14. Motor power input determined as the product of the measured voltage and current for 80kg subject running at 3.6 m/s. Note that the spike in power is due to reversal of the motor between stance and swing.

Testing also indicated service life challenges with the lead screw assembly in the actuator. The lead screw assembly is required to convert the rotational motion of the electric motors to translation with a high enough mechanical advantage to limit the size of the electric motors used. Therefore, as second emphasis of the next redesign is incorporation of a roller screw assembly in place of the lead screw, and elimination of any external moments not exerted axially on the roller screw. See Fig. 15 for a rendering of the future design concept.

Fig. 15**.** Conceptual design of next running foot iteration includes a single brushless motor actuator, reduced overall weight, and isolated roller screw assembly.

## 4. CONCLUSION

Preliminary testing indicates that it is possible to use a series actuator-spring system to mimic the tendon-muscle power amplification strategy for running. Overground tests continue with full system portability, see Fig. 16. However, running gait kinematics, particularly the reduced stance phase within a gait cycle, increases the importance of component inertial characteristics, thereby generating a requirement to use brushless motor systems in future designs and complex manufacturing techniques that facilitate significant weight reduction. As ancillary efforts to develop battery systems with increasing energy density continue, the development of a feasible robust foot capable of replacing limb functions under any activity and environmental condition becomes a more realizable goal.

Fig 16. Unilateral transtibial amputee conducting overground walking tests with fully portable tibia-based controller.

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