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DYNAMICALLY CONTROLLED ANKLE-FOOT ORTHOSIS (DCO) WITH REGENERATIVE KINETICS: INCREMENTALLY ATTAINING USER PORTABILITY

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ABSTRACT – A portable wearable robotic device that can actively supplement locomotion of partially limited ambulators in their normal environment (variable terrain, weather, man made structures, etc.) seems highly desirable but currently short of attainment due to several key technology gaps. Low energy and power density in current actuation technology, inadequate control schemes and safety of use are leading challenges towards a portable, complementary device. This paper presents the *dynamically controlled ankle-foot orthosis (DCO) with regenerative kinetics* which seek to incrementally attain portability by solving the energy/power density issue in powered elements by harnessing elastic energy of uniquely tuned mechanical elements and reducing the control problem and increasing safety by introducing compliant elements between the human-machine-environment interfaces.

Index Terms – Ankle-Foot Orthosis, Power/Energy Density, Robotic Tendon, Power Amplification, Dynamic Control

I. INTRODUCTION

A. BACKGROUND

Using powered assistive devices and orthotics to aid patients with locomotor deficits is not new. In the 1970's Vukobratovic built a pneumatic exoskeleton to supplement human walking, and Seireg et al. designed a hydraulic multitasking exoskeletal walking device [1, 2, 3]. Because studies have recently shown that neural networks in the brain and spinal cord possess the ability to reorganize, interest in robotic therapy has significantly increased. It was proven that through repetitive task training neural networks can be re-mapped [4-8]. With renewed energy, the rehabilitation community has pushed for assistive devices.

Most of the currently available systems are large clinic based devices such as The Lokomat System by Hocoma which includes a table top controller, orthoses, treadmill and a body support system [9-10]. Other well known assistive devices include the BLEEX (Berkley Lower Extremity Exoskeleton) robot and HAL-3 (Hybrid Assistive Leg) robot [11-13]. Both of these devices are rigidly attached to the wearer and are directly driven so that the human-machine-environment interfaces are not inherently compliant. The BLEEX robot uses hybrid hydraulic actuators to drive the system, whereas the HAL-3 robot uses DC motors and gearboxes to provide power for movement of the user.

In other work, a robotic powered knee, RoboKnee [14] and an active ankle foot orthosis, AAFO [15], have been developed to assist with gait. Each of these devices feature the linear Series Elastic Actuator [16] that uses a helical spring in series with a ball screw mechanism. The spring aids in force and impedance control task stability but their system compliance is derived mostly from its controller.

All of these devices have varying degree of potential, especially within a clinic setting that can accommodate the large form factor of these devices. However, for a large segment of patients with only partial deficits, a portable wearable device would allow for use in specific gait tasks such as walking in unstructured environments, modulating speed, climbing stairs, etc. These complementary devices could be both therapeutic and assistive.

B. ANKLE COMPLEX DURING WALKING GAIT

Gait is a cyclical pattern of leg and foot movement that creates locomotion. Gait is commonly discussed in terms of a percentage of a single gait cycle. A gait cycle is defined for a single leg and begins with the initial contact of the foot with the ground or 'heel strike'; the conclusion of a cycle occurs as the same foot makes a second 'heel strike'. To illustrate a typical pattern of gait, consider the illustration of the ankle complex during stance phase of a single cycle of gait, figure 1 and the kinematics and kinetics of a normal ankle, figure 2. Notice that in figure 2, peak ankle moment occurs at roughly 45% of the gait cycle and at a normalized value of -1.25 Nm/kg. The negative sign represents the physiological direction of the plantarflexing ankle complex. The foot rotates downwards to push off from the ground. At the point at which the peak moment occurs, the ankle angle begins a rapid decent to its lowest overall value of -24 degrees at 60% of the gait cycle. The region of gait approximately between 45% and 60% of the gait cycle is known as 'push off'. At the conclusion of 'push off', now considered ‘toe off’, the leg initiates ‘swing’ and the foot is then positioned for the next ‘heel strike’.

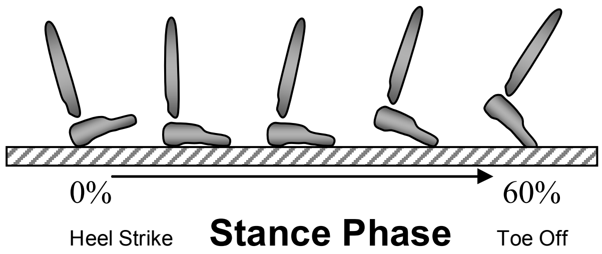

Figure 1. Stance phase of a single gait cycle. 60-100% of gait is the swing phase, not shown.

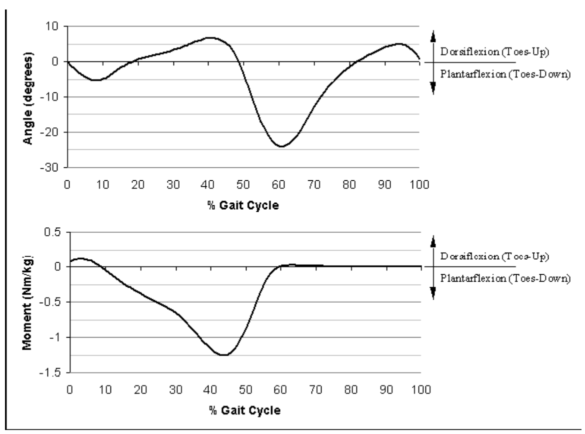

Figure 2. Normal Ankle Gait: Kinematics and Kinetics [25].

C. ENERGY AND POWER DENSITY

An ankle joint requires 250W of peak power and in the direct drive scenario 36 Joules of energy per step (80kg subject at 0.8Hz walking) [17]. This is significantly more than what is required at the knee or hip joints. An actuator and transmission system that can provide the necessary peak power would most likely be an electric motor and gearbox system that under a traditional approach weights 6-7 kg. Providing 36 Joules of energy per step would require a significantly larger battery for a modest 8 hours of operation. Other actuator technologies that are used in gait therapy devices such as pneumatic and hydraulic actuators like the “McKibben Muscle” can provide the required power with a small device but are impractical as portable devices because they require separate pumps or other air supply[18]. Herr and Kornbluh describe a “new horizon” in artificial muscle in [19]. Electroactive polymers have been used to demonstrate stationary bicycle pedaling and bicep movement in a human size skeleton but even the most promising of these materials, the dielectric elastomer, would require more than 5000V of electricity for operation of a much smaller 100W actuator [19].

D. COMPLIANCE

Traditional robots are inherently stiff and interfacing with humans can be hazardous. Understanding force control is essential in developing compliant actuators needed for human-robotic interaction. As a manipulator interacts with its environment, the forces must be controlled so that it is compliant and safe. Early work in force control focused on controlling stiff robots. Fine or precise robotic force control has been difficult to achieve because of the stability problems mentioned in literature [20]. In addition, the mechanical stiffness of these systems is typically very large and it is necessary to rely on high-performance actuators (expensive, heavy motors) and high bandwidth control to produce compliance. Such a scheme will have inherent limitations during interaction with a stiff and unstructured environment such as repeated impacts with an uneven walking surface.

In contrast, a system with mechanical compliance built into the device will have intrinsic compliance regardless of the stiffness or unstructured nature of the environment. As a result, the requirement on the actuator and the system control are much more modest [21-23].

II. DCO DESIGN

We have designed and built a light weight, compliant, energy efficient, active ankle-foot orthosis [17, 24], figure 3. A small DC motor, a highly efficient lead screw and spring assembly actuate the orthosis. The orthosis is a single axis device that provides sagittal plane movement at the ankle joint. The DCO is controlled in real time using Real Time Workshop and Simulink from Mathworks. The Simulink model is compiled on to the embedded target PC with the xPC Target Operating System. An encoder, linear potentiometer and one force sensitive resister embedded at the heel provides the necessary sensor feedback.

Advantech’s 650MHZ PC-104 with 512MB on board memory is selected to run the system. Multifunctional I/O board from Sensoray Co., Model 526, which is connected to the PC104 via an ISA bus, controls a RE-30 Maxon DC motor by using its encoder. An MC-7 motor controller from Diverse Electronic Service is used with a 24V voltage supply to control the motor with necessary current while an ATX power supply is used for the computer.

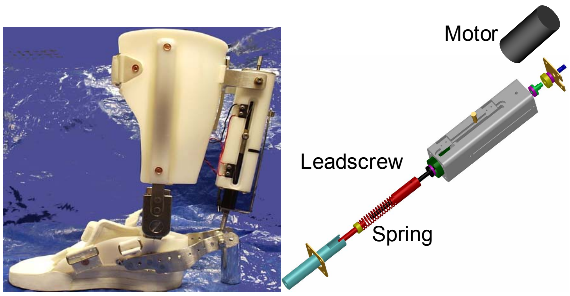

Figure 3. DCO with the Robotic Tendon Actuator.

III. DATA AND ANALYSIS

A. THE ROBOTIC TENDON

The Robotic Tendon [17] (figure 3) is a small and lightweight actuator that features a low energy motor that is used to adjust the position of the helical spring using a very simple position controller. Figure 4 illustrates how the desired spring deflection and consequently via Hookes Law the desired force is achieved using a spring. As the ankle rotates over the foot during stance phase, a lever position profile as` shown in figure 4 is obtained. By correctly positioning the motor, a desired spring deflection as shown in the shaded area of figure 4 is obtained. A heavy, powerful, impedance controlled motor is not needed because the Robotic Tendon stores a portion of the stance phase kinetic energy and additional motor energy within the spring. The spring releases its stored energy to provide most of the peak power required during push off, figure 5. Therefore, the power requirement on the motor is significantly reduced. As described in [17], peak motor power required is 77W compared to 250W for a direct drive system in the 80kg subject at 0.8hz example, figure 6. And consequently, the weight of the Robotic Tendon, at just 0.95kg, is 7 times less than an equivalent direct drive motor and gearbox system that is required to provide the necessary peak power. In addition, ideal energy requirements, as determined by the integration of the power curves, were reduced from nearly 36 Joules to 21 Joules per step (80kg subject walking at 0.8hz) significantly reducing battery requirements so that a commercially available battery pack worn in a fanny pack could potentially power the DCO for 8 hours of continuous operation.

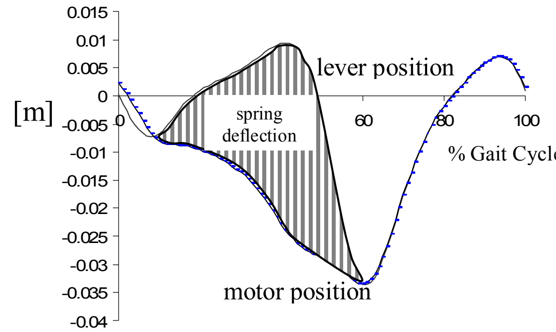

Figure 4. Desired spring deflection, shaded area, is achieved by controlling the motor position and capitalizing on the cyclical nature of gait.

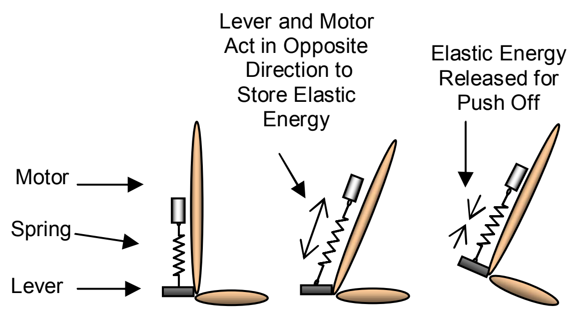

Figure 5. Motor and lever act in opposite directions to store the necessary elastic energy in the spring required for push off. The spring provides the majority of the peak power required for push off.

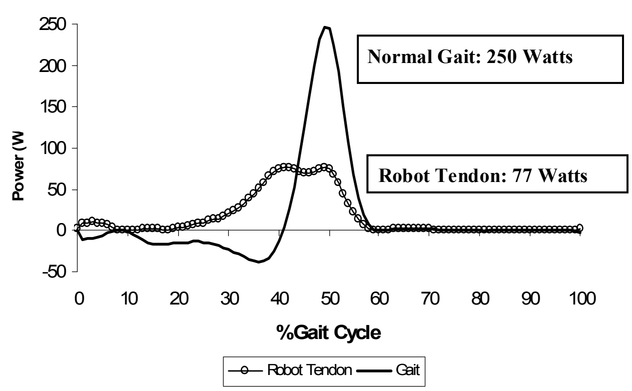

Figure 6. 250 Watts of peak power is required for normal gait. The motor provides 77 Watts of peak power and the spring provides the remaining power required for gait.

B. POWER AMPLIFICATION

In our most recent experiment, we tested two able-bodied subjects (male, 60kg and female, 70kg both at 0.8hz) outfitted with the DCO on a treadmill. The DCO was configured to provide 50% assistance, which was done by scaling the body weight by half to determine the power requirements. The purpose was to confirm our new dynamic control scheme and to confirm that by harnessing the stored elastic energy in the mechanical elements, motor requirements were significantly reduced – a critical requirement towards portability.

Equation 1, developed in [17], provides a relationship between spring stiffness K and peak motor power P_m. From this relationship an optimum spring stiffness is selected that minimizes the motor power requirement. Note that F is the force exerted on the spring and x_g is the lever position.

$$(P_m)_{peak} = \max\left|F \cdot \dot{x}_g + \frac{F \cdot \dot{F}}{K}\right| \qquad (1)$$

For a 70kg subject walking with 50% assistance with a typical gait would require a peak power of approximately 108W. Consequently, a direct drive system would be forced to provide 100% of the 108W. By selecting an optimal K value of 15850 N/m for the Robotic Tendon using equation 1, minimum peak power required by the motor is determined to be 43W. This is a significant reduction in motor requirement and therefore, size and weight. Figure 7 illustrates the ideal power sharing between the motor and the spring to provide the required output power. Notice that around 40% of gait, the motor and spring work in opposite direction to store elastic energy in the spring so that at 50% of gait, the spring can provide almost 100 Watts of peak power while the motor provides the remainder.

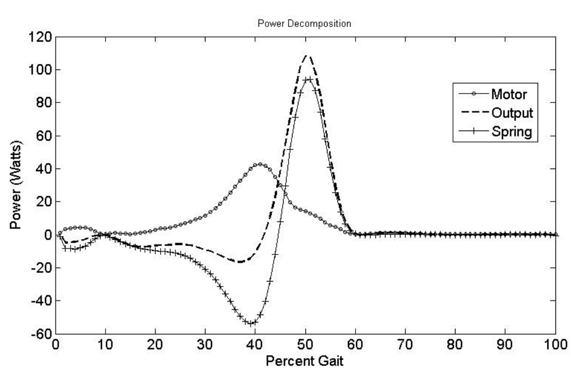

Figure 7. The spring and motor power add to provide the desired output power required for gait. Notice that at 40% of gait, the spring and motor work in opposite direction to store elastic energy and at 50% gait, the spring provides majority of the output power.

Results of our initial evaluation can be seen in figure 8. Input power was determined by the product of the linear velocity and the force at the motor. Output power was determined by the product of the linear velocity of the ankle displacement and the force acting on the spring. The measured output peak power was 131W. The corresponding input power generated by the motor was 55W. This means that the spring provided 76 W of the 131W of peak power. The output was 238% higher than the power generated by the motor – a 2.38 power amplification. This is a very positive result in terms of kinetic efficiency. This demonstrates the power of harnessing spring energy in gait assistance. The challenge now is to further increase power amplification without compromising gait kinematics and to achieve higher level consistency in output power.

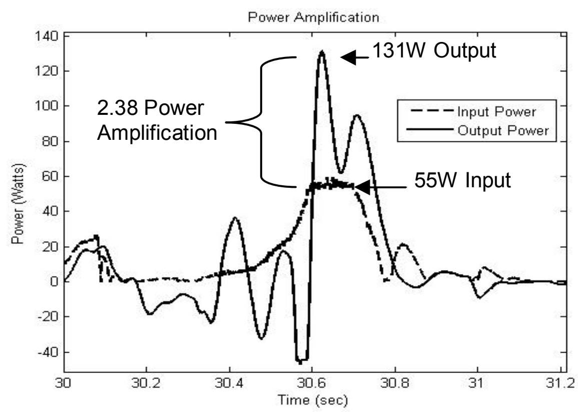

Figure 8. The input power is the power generated by the dc motor. The output power is the combination of motor and spring power. Note: output power profile includes noise due to the use of numerical differentiation method.

C. DYNAMIC CONTROL

Together with power and energy density, computer control of orthoses and prostheses remain a significant challenge. Work by Au et al in EMG position control [26] and by Pappas et al in state based control [27] seems promising because of its simplicity. Sugar's effort to reduce the control problem using compliant simple force control [23] is a key finding towards simplifying control methodology and served as our starting point with the Robotic Tendon.

Unlike a clinic based device, a controller for a portable orthosis must have the capability to support the user in his complete environment (home, work, outdoors, etc.) and conditions (weather, terrain, etc.). In this paper, we present our initial DCO controller that supports walking gait initiation, cessation and speed modulation based on user intent on a treadmill. Additional details on the control methodology can be found in [28]. A more robust state based controller is currently under development that identifies and controls 4 separate states of a user's gait cycle.

The DCO controller has a predetermined gait pattern expressed as a time-based function embedded in the controller, which drives the motor controller and thus the system. Gait is initiated at heel strike with activation of the force sensitive resister embedded in the heel. As the user initiates gait, the motor drives the lead screw nut through a pattern predetermined for each subject with closed loop feedback. The ankle, however, is not forced to follow the specific pattern because the compliant spring is between the motor and user, safely absorbing environmental irregularities such as a rock under foot or user errors. This inherent compliance not only provides for a safer interface but allows for a much simpler control scheme because we no longer require high bandwidth high precision force control.

Through experimentation, a Proportional Derivative control with a P gain of 1 and a small derivative value of 0.02 was selected. As seen in Figure 10 the encoder output (smooth line) closely follows the checkered curve which is the embedded gait pattern with an expected 0.03 sec lag.

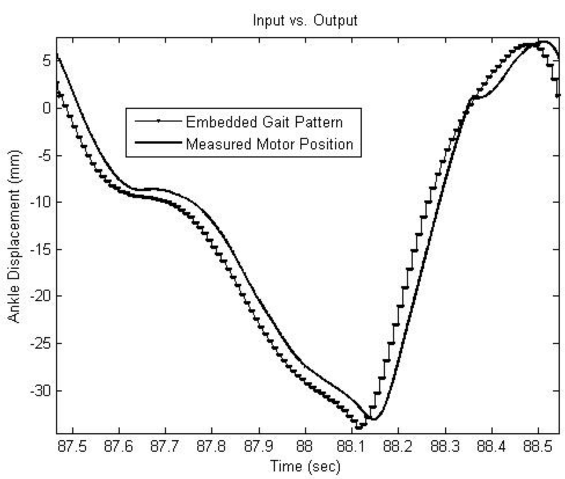

Figure 10. The checkered line is the embedded gait pattern and the smooth line is the measured motor position. It is a very close match.

Figure 11 illustrates the dynamically changing gait cycle. The top graph illustrates the changing pace of gait. The subject during this period varied her walking speed from 1.0 sec/step to 1.8 sec/step as indicated by the plot. Pace of gait is determined by determining the duration between a heel strike to a heel strike of the previous gait cycle. The bottom plot in Figure 11 is the embedded gait pattern. By scaling the coefficients of the time-based gait pattern function, we can modulate the frequency of the embedded gait pattern [28]. This simple controller is very effective in modulating speed, gait initiation and cessation under linear treadmill walking conditions as tested on three able-bodied and two stroke survivors.

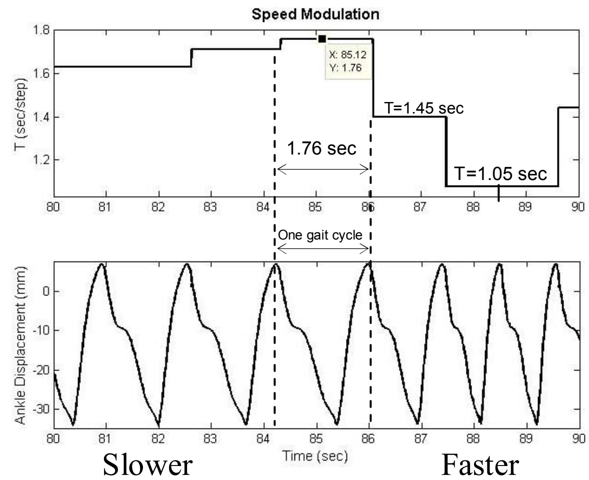

Figure 11. The top graph indicates the time measured between consecutive heel touches of the previous gait cycle, T. The bottom graph shows the frequency modulation based on the duration time, T. A smaller T correlates to a higher frequency of the embedded gait pattern. Higher frequency of the input gait pattern is in response to higher speed of the user.

IV. CONCLUSION

Significant advances have been achieved towards portable wearable robotic devices that can actively complement the partially limited ambulators in their normal environment and conditions. The Proprio Ankle [29] by Ossur or the MIT's Ankle-Foot Emulator [30] is a good example of the most recent achievements. However, low power and energy density and inadequate control methodology remain as key challenges towards realizing biomimetic wearable devices. We presented in this paper the Dynamically Controlled Orthosis with Regenerative Kinetics as one solution to the power and energy density problem. We showed that our approach gains kinetic advantages by leveraging elastic energy potential in uniquely tuned helical springs. As the tibia rotates over the stance foot ankle during walking gait, we position the spring to maximize elastic energy storage. We presented one example where we achieved a power amplification of 2.38 with the motor providing 55W and the spring providing the remaining 76W. In testing with two able-bodied subjects we achieved power amplification of up to 2.7 and our latest work suggests power amplification of twice our current amount may be possible as we better understand the detailed movement of the human ankle complex and the muscle-tendon structures. This is a very significant finding because at this level of amplification, we can downsize the mechanical system to a wearable level and energy consumption is also within wearable levels. Also, as significant is that this level of power amplification brings powered running devices within sight. In addition, we presented our new controller with the capability to start, stop and modulate gait speed on a treadmill. Several months of testing on three able-bodied and two subjects with gait deficits show that the control methodology is sufficient for in-clinic use. We believe state logic is the best near-term control approach as we expand the environment and conditions of our DCO and as we develop our new generation of Spring Ankle with Regenerative Kinetics (SPARKy) prostheses.

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