

Autonomous Exoskeleton Reduces Metabolic Cost of Walking

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Abstract—We developed an autonomous powered leg exoskeleton capable of providing large amounts of positive mechanical power to the wearer during powered plantar-flexion phase of walking. The autonomous exoskeleton consisted of a winch actuator fasted to the shin which pulled on fiberglass struts attached to a boot. The fiberglass struts formed a rigid extension of the foot when the proximal end of the strut was pulled in forward by the winch actuator. This lightweight, geometric transmission allowed the electric winch actuator to efficiently produce biological levels of power at the ankle joint. The exoskeleton was powered and controlled by lithium polymer batteries and motor controller worn around the waist. Preliminary testing on two subjects walking at 1.4 m/s resulted in the exoskeleton reducing the metabolic cost of walking by 6–11% as compared to not wearing the device. The exoskeleton provided a peak mechanical power of over 180 W at each ankle (mean standard \pm deviation) and an average positive mechanical power of 27 ± 1 W total to both ankles, while electrically using 75–89 W of electricity. The batteries (800 g) used in this experiment are estimated to be capable of providing this level of assistance for up to 7 km of walking.

I. INTRODUCTION

Using a wearable device in parallel with the body to reduce the metabolic cost of walking has proven to be a challenge [1], [2]. The earliest record of exoskeleton development is a late 19th century U.S. patent [3]. Since that time, interest in such technologies has increased substantially, driven by the accelerating pace of innovation in several mechanical and computer-related disciplines [4]–[6]. The overarching goal of such technologies is the reduction of the metabolic energy (i.e. calories) required during locomotion. [1], [2]. To our knowledge, no autonomous exoskeleton has provided a reduction in the metabolic energy required for unloaded walking.

Many factors have hindered the development of these performance-enhancing exoskeletons including substantial added mass, limited mechanical power and tethered energy supplies. Autonomous devices capable of providing biologically equivalent levels of joint mechanical power

necessary for locomotion have been investigated, but these devices have been bulky with substantial mass [1], [7], [8]. Adding mass to the lower limbs requires additional metabolic power, and the effects are amplified as the mass is moved distally or further away from the hip [9]. In an effort to reduce the weight of the worn exoskeleton, researchers have investigated passive and quasi-passive exoskeletons. Without an active actuator, these devices are not able to provide biological levels of positive power [10]–[12]. Researchers have also reduced weight and provided substantial positive power by tethering them to an energy supply not worn by the human wearer [13], [14]. These devices were able to provide a metabolic improvement during walking using pneumatic artificial muscles, but required a tether to an air supply and extensive valving control network, thus distancing them from an autonomous solution.

Despite challenges in the development of autonomous exoskeletons, researchers have reduced the metabolic energy consumed in other activities, such as hopping and walking with load carriage. By adding a spring in parallel with the legs, the metabolic cost of hopping was reduced by 24% [15]. This exoskeleton exploited the energetically conservative nature of hopping to reduce the mechanical work done by the muscles of the legs. Additionally, the metabolic energy consumed during walking with load carriage was demonstrated by the device presented in the current study [16]. The metabolic cost of loaded walking was reduced by 36 W, 8% of the metabolic cost of walking without the device. Such work is encouraging, however autonomously reducing the energy consumed during unloaded walking has remained difficult.

In this study, we present the design and testing of an autonomous leg exoskeleton. The intent of this research is to develop a technology that can reduce the metabolic cost of level ground walking. Our hypothesis is that a leg exoskeleton capable of providing biological levels of positive mechanical power with minimal added distal mass can provide such a metabolic benefit. In the evaluation of this hypothesis, we chose to augment the ankle joint because it is responsible for over 40% of the average positive mechanical power during walking [17]. We tested the metabolic effect of the ankle exoskeleton while walking at 1.4 m/s.

This work was supported by the National Science Foundation Graduate Research Fellowship award number 1122374.

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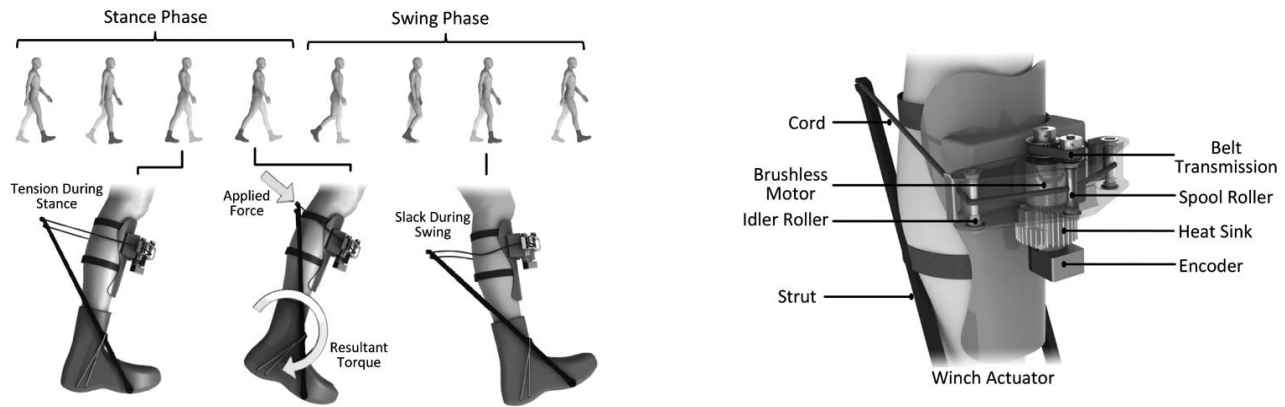


Fig. 1: The mechanical design of the autonomous powered leg exoskeleton. The winch actuator applies a force on the proximal end of fiberglass struts attached to the boot. The force of the winch actuator is reacted with the shin of the subject. The applied forces results in a large plantar flexion torque during push-off, but the actuator can release excess cord during swing to become transparent to the subject.

II. EXOSKELETON DESIGN

The exoskeleton was designed and controlled to provide assistance to the ankle during the powered plantar-flexion phase of walking.

A. Mechanical Design

The mechanical hardware was comprised three main assemblies: a pair of fiberglass struts attached to each boot, a unidirectional actuator mounted on the anterior shank segment, and a battery and control package worn on the waist (Fig. 1). Each boot had two medial and lateral fiberglass struts, pinned to the medial and lateral aspects of the metatarsophalangeal joints. A lightweight inextensible cord coupled each strut to the heel of the boot. The fiberglass struts were an extension of the ankle-foot complex; when an anterior force was applied to the proximal end of the strut it was converted into a torque about the human ankle joint. The struts acted as a moment arm (≈ 300 mm from the ankle joint) for the winch actuator to apply the plantar-flexion assistive torque about the ankle joint. The exoskeleton used a custom winch actuator powered by a brushless DC (BLDC) motor. The 200 W BLDC motor (model: 305015, Maxon Motor, Sachseln, CH) actuated an 8 mm diameter spool through a belt transmission with a 13:8 speed reduction. The spool wrapped a 1 mm diameter ultra-high molecular weight polyethylene cord (Dyneema, Stanley, NC) attached to the proximal end of the fiberglass struts. The effective transmission ratio between the BLDC motor and ankle joint was approximately 120:1. The geometric transmission, comprised of a spool, idler roller and strut, eliminated the need for a traditional mechanical transmission, reducing weight and complexity of the device.

The exoskeleton motors were powered and controlled by batteries and motor controllers worn around the waist. Sensory information was provided to the controller. A gyroscope was integrated into each actuator to measure the angular shank velocity in the sagittal plane (model: LPY550ALTR, STMicroelectronics, Geneva, CH). The angular position of the BLDC motors were measured with 500 count quadrature incremental optical encoders (model:

HEDL 5540, Maxon Motor, Sachseln, CH). Each winch actuator had a corresponding BLDC motor controller (model: SBL1360, Roboteq, Scottsdale, AZ). The motor control loop iterated at 1000 Hz. The motors, sensors and controllers were all powered by two 24 V lithium-polymer batteries, with a capacity of 2.5 Ah each. The total mass of the system was 3.8 kg, with 1.7 kg worn on the waist and 2.1 kg worn on the legs.

B. Walking Control

A biomechanically-inspired control strategy was implemented to detect gait phases and assist the user during the push-off region of the gait cycle. To apply the appropriate assistance, the location within the gait cycle was detected and the desired mechanical power characteristics were determined. To detect gait phase location, the gyroscopes on each actuator were used to detect heel strike. Terminal swing phase of walking was estimated by a period of positive shank angular velocity with duration of greater than 300 ms. Subsequently, heel strike was estimated as occurring 50 ms before the shank velocity reached a speed of zero. The 50 ms accounted for the lag introduced by a 2nd order, 6 Hz low pass Butterworth filter applied to the gyroscope signal. During controlled dorsiflexion, the winch maintained tension in cord by applying a low open loop torque (10-20 Nm). Powered assistance was then achieved by applying a parabolic power profile over 150 ms to assist with plantar-flexion. The appropriate power characteristics for the following cycle were determined at the end of each gait cycle. The motor's angular velocity and the voltage were used to obtain the peak magnitude and timing of the applied power. These values were compared to Winter's reference ankle power profiles [18]. The controller incrementally scaled and shifted the parabolic peak magnitude and initiation timing of the power assistance so that the peak power would occur at 54% of the gait cycle [18], and the magnitude would be 2 W/kg. Powered assistance did not impede additional plantar flexion by the biological ankle due to the nature of the unidirectional actuator. After 150 ms of powered plantar-flexion, the controller entered swing phase. During swing phase the exoskeleton used position control to

quickly release the cord over 100 ms and provide slack in the drive cord to allow the user to freely dorsiflex the biological ankle.

During the calculation of exoskeleton applied mechanical power, the motor's voltage and angular velocity were used with a linear model of motor efficiency. The linear motor model included motor parameters, applied voltage and motor velocity to estimate the mechanical motor power. Previously, an external force sensor was used to empirically determine the transfer efficiency between the model predicted mechanical power and the measured mechanical power [16]. The average transfer efficiency for positive mechanical power was 0.68 ± 0.03 which resulted in a root mean square error of 11 W, tested over a range of output amplitudes and frequencies.

III. METHODS

The metabolic effect of the exoskeleton was tested on two male subjects (82 & 86 kg weights; 175 & 191 cm heights) walking on a treadmill at 1.4 m/s. The subjects were healthy and exhibited no gait abnormalities. This study was approved by the MIT Committee on the Use of Humans as Experimental Subjects. Consent was obtained from experimental participants after the nature and possible consequences of the exoskeletal studies were explained. The experimental protocol involved four walking trials and two standing trials, all performed while wearing a portable pulmonary gas exchange measurement instrument (model: K4b², COSMED, Rome, IT). To account for natural variation in metabolism, the control conditions of no exoskeleton were tested before and after the exoskeleton conditions. First, the subjects stood for 5 minutes to measure an initial resting metabolism. The subjects then walked for 10 minutes without the device. Afterwards, the subjects walked with the powered exoskeleton providing assistance for 20 minutes in order to allow for human-machine adaptation [19]. Next, the subjects walked for 20 minutes with the unpowered exoskeleton to measure the effects of the added mass. The subjects then walked for another 10 minutes without the device. Finally, after the last no device trial, the subjects stood for 5 minutes in order to obtain the final resting metabolic rate.

Metabolic rate was calculated from oxygen and carbon dioxide exchange rates measured by the portable pulmonary gas exchange measurement unit. The average flow rates of the last two minutes of each trial were converted into metabolic power using the equation developed by Brockway et al. [20]. The metabolic rate of standing was subtracted from the gross metabolic rates of walking in order to obtain

the net metabolic cost of walking. The net metabolic rates measured from the two control trials were averaged and compared to the net metabolic rates of the exoskeleton trials.

The mechanical and electrical power of the exoskeleton were wirelessly recoded via Bluetooth at a sampling rate of 83 Hz. The mechanical power was estimated through the linear motor model and experimentally measured transfer efficiency discussed in a previous manuscript [16].

IV. RESULTS

The metabolic results supported the hypothesis that an ankle exoskeleton can reduce the energetic cost of walking by supplying biological levels of mechanical power and minimizing the added distal mass. (Table 1.).

The mechanical power provided by the exoskeleton was measured over 40 strides. The profile of the mechanical power for subject 1 is compared to Winter's reference ankle power profile in Fig. 2 [18]. The exoskeleton provided an average positive mechanical power of 27 ± 1 W (mean \pm standard deviation) to subject 1, and 26 ± 1 W to subject 2. The exoskeleton required 75 ± 3 W of electrical power for subject 1, and 89 ± 4 W of electrical power for subject 2. The peak mechanical power, 182 ± 6 W for subject 1, and 183 ± 14 W, occurred at 54% of the gait cycle.

V. DISCUSSION

The magnitude of the positive power applied by the exoskeleton appeared to be a critical factor. The pneumatic powered ankle exoskeleton developed by Sawicki and Ferris, which has a similar mass distribution as the presented device, generated an average positive mechanical power of 16 W and reduced the metabolic cost by 10 W [14]. The pneumatic exoskeleton also applied a burst of power during powered plantar-flexion. The autonomous exoskeleton increased the average mechanical power by approximately 70%, but increased the metabolic reduction by 80-200%. This may suggest that once the effects of added mass are overcome, additional power assistance results in a metabolic reduction greater than the added mechanical power, as supported by previous studies [13], [14], [16], [21]–[23].

In order to be autonomous, the exoskeleton must efficiently convert electrical energy into mechanical energy. Lithium polymer batteries have a high energy density, but the efficiency of transforming the chemical energy to mechanical energy is limited by the characteristics of the brushless motors. Brushless motors are most efficient when operating at high speeds and low torques, contrasting the conditions seen at the ankle joint, which are considerable lower speeds and higher torques. Traditional transmissions such as ball screws or gears can be quite efficient, but it is often at the cost of added mass and complexity. The autonomous exoskeleton achieves an efficient, lightweight transmission by simply extending the moment arm attached to the foot. The large moment arm produced by the fiberglass struts

Table 1: Net Metabolic Results

Subject	Without (W)	Powered (W)	Unpowered (W)
1	295	278	319
2	263	233	295

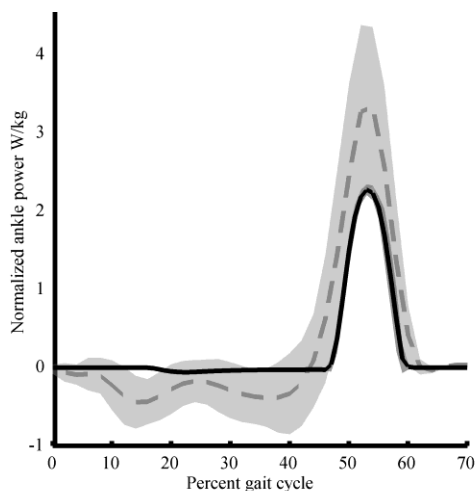


Fig. 2: The estimated mechanical power applied by the exoskeleton to subject 1 and a reference ankle power profile. A reference ankle power profile is shown with a dashed grey line, and one standard deviation is shaded in light grey. The estimated mechanical power applied by the exoskeleton, averaged over 40 strides, is shown in black. One standard deviation is shaded in a dark gray, but can only be seen near the peak power.

reduces the required forces needed to actuate the ankle. Furthermore, the actuator is able to react against the shin, which is able to comfortably withstand the winch loads. The moment arm and winch transmission do not constrain the sagittal motion of the ankle. The current estimated range of the system, 8 km, scales with the batteries carried at the waist. Studies of weight carrying energetics suggest that the metabolic cost of extending the exoskeleton range would be approximately 3.7 W per 10 km [9]. The range of the exoskeleton could be extended to 40 km and still provide a metabolic benefit.

Future work will involve the testing of more subjects and measuring kinetics and kinematics. It is necessary to test more subjects in order to determine the metabolic effects of the exoskeleton across many individuals. It will also be interesting to measure the biomechanics of subjects using the exoskeleton and determining how the exoskeleton affects the mechanical work done at each joint.

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