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Development of a lightweight, underactuated exoskeleton for load-carrying augmentation

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Abstract **- Metabolic studies have shown that there is a metabolic cost associated with carrying load [1]. Several leg exoskeletons have been developed by various groups in an attempt to augment the load carrying capability of the human. Previous research efforts have not fully exploited the passive dynamics of walking and have largely focused on fully actuated exoskeletons that are heavy with large energy requirements. In this paper, a lightweight, underactuated exoskeleton design is presented that runs in parallel to the human and supports the weight of a payload. Two exoskeleton architectures are pursued based on examining human walking data. A first architecture consists of springs at the hip, a variable impedance device at the knee, and springs at the ankle. A second architecture replaces the springs at the hip with a non-conservative actuator to examine the effect of adding power at desired instances throughout the gait cycle. Preliminary studies show that an efficient, underactuated leg exoskeleton can effectively transmit payload forces to the ground during the walking cycle.**

Index Terms – exoskeleton, spring, underactuated, walking

I. INTRODUCTION

When compared to wheeled vehicles, exoskeleton-based assistive devices have several advantages, such as allowing the user to traverse irregular terrain surfaces. Individuals employed in specific recreational, occupational and military pursuits often carry heavy loads using a variety of backpack systems. Recreational hikers and backpackers commonly carry subsistence and comfort items in backpacks [2]. Fire fighters and other emergency personnel carry oxygen tanks for breathing and other equipment using backpack systems [3]. Still further, foot soldiers often carry extremely heavy backpack loads and walk long distances across rough terrain [2].

A leg exoskeleton could benefit people who engage in load carrying by increasing load capacity, lessening the likelihood of injury, improving efficiency and reducing the perceived level of difficulty. Exoskeletons have been developed that amplify the strength of the wearer, apply assistive torques to the wearer's joints and support a payload for the wearer. Several exoskeleton design approaches have employed hydraulic actuators to power hip, knee and ankle joints in the sagittal plane [4]. Such an exoskeleton design demands a great deal of power, requiring a heavy power supply to achieve system autonomy. For example, the exoskeleton in [4] consumes approximately 2.27kW of hydraulic power, 220 Watts of electrical power, and has a total system weight of 100 lbs. This approach leads to a noisy device that has a very low payload to system weight ratio. Further, this type of exoskeleton is heavy, and if failure were to occur, could significantly harm the wearer.

Evidence from biology [5] and passive walkers [6] suggests that legged locomotion can be very energy efficient. The exchange between potential and kinetic energy suggests that walking may be approximated as a passive mechanical process. Passive walkers reinforce this fact. In such a device, a human-like pair of legs settles into a natural gait pattern generated by the interaction of gravity and inertia. Although a purely passive walker requires a modest incline to power its movements, researchers have enabled robots to walk on level ground by adding just a small amount of energy solely at the hip or the ankle joint [7]. Recent evidence suggests that elastic energy storage is also critical for efficient bipedal ambulation. Palmer [8] showed that by characterizing the human ankle during the stance phase of walking in terms of simple mechanical spring elements, sagittal plane dynamics of a normal ankle can be reproduced at least at slow to moderate walking speeds. Further, in [9] it was shown in numerical simulation that an exoskeleton using passive elastic devices can substantially reduce muscle force and metabolic energy in walking.

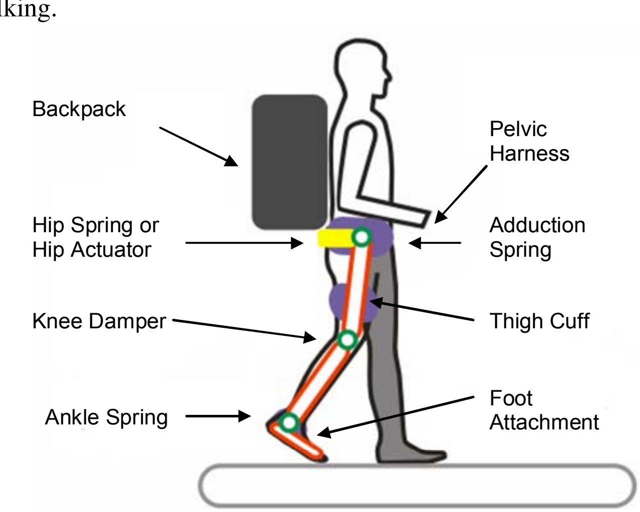

Fig.1 Concept sketch of the main components of the exoskeleton.

This paper examines biomechanical data from human walking and outlines the design of an alternative, more efficient exoskeleton that uses both passive and active elements. Two exoskeleton architectures are outlined. A first architecture consists of springs at the hip, a variable impedance device at the knee, and springs at the ankle. A second architecture replaces the springs at the hip with a non-conservative actuator to examine the effect of adding power at desired instances throughout the gait cycle.

II. Exoskeleton Design

A. *Methodology*

The exoskeleton was designed to provide a parallel load path to transfer the weight of the backpack directly to the ground. The exoskeleton had sufficient degrees of freedom to minimize kinematic constraints experienced by the wearer. The system was designed so that the distal mass of the exoskeleton was minimized. The requirement for hip actuation in the sagittal plane was to assist both the exoskeleton and human in walking.

B. *Degrees of Freedom*

The exoskeleton was implemented with three degrees of freedom at the hip, one at the knee, two at the ankle and one at the foot. The joint ranges of motion accommodated normal human walking. A cam mechanism was implemented at the hip joint to enable hip abduction/adduction. During abduction in the coronal plane, there was a length difference between the biological leg and the exoskeleton leg. This length difference resulted from dissimilar centers of rotation between the biological leg and the exoskeleton leg. This effect impeded normal walking motion and caused discomfort. A mechanism was designed to automatically adjust the exoskeleton leg length and project the center of rotation of the exoskeleton leg onto the biological hip center of rotation.

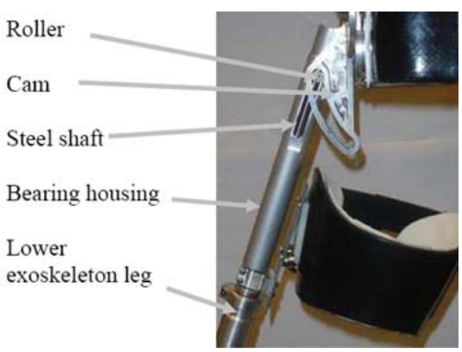

Fig. 2 The roller is grounded to a hollow ½" steel shaft. When the leg abducts, the roller rides up the cam and the steel shaft slides up the bearing housing. The steel shaft is connected to the lower part of the exoskeleton leg so as the leg is abducted, the exoskeleton leg shortens.

C. *Interface to the Human*

The exoskeleton interfaced to the human via shoulder straps, a waist belt, thigh cuffs, and a shoe connection. A compliant belt interfaced the lower torso to the backpack frame and the backpack's shoulder straps interfaced the upper torso. The physical connection between the exoskeleton and the human enabled the exoskeleton to passively track the human's leg motion. A standard military issued backpack, Alice Pack, was selected to carry the load. The exoskeleton was attached to the standard military backpack through a harness. The hip joints of the exoskeleton legs were mounted to the harness. There was sufficient clearance between the pelvic harness and the wearer to minimize disturbance to the wearer's gait.

D. *Structure*

A parallel orthotic structure was the basic framework to transfer the load from the backpack to the ground. The main structural elements consisted of standard prosthetic aluminum tubing. This tubing was chosen since it is lightweight, rated for human use, and interfaces with standard prosthetic connectors and components.

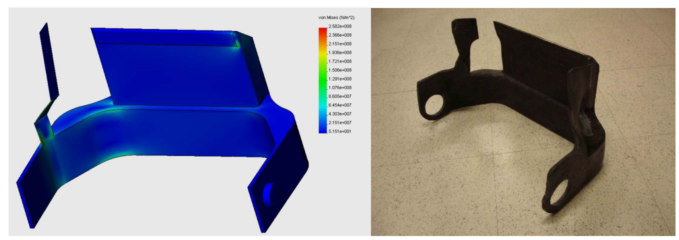

Fig. 3 Finite element modeling results from optimization of harness design along with final molded part.

The harness connected rigidly to the backpack frame to transfer the load from the backpack to the exoskeleton. The pelvic harness was made from carbon fiber and the stiffness to weight ratio was optimized using finite element analysis. The structure consisted of a hollow core with 1/16th inch thickness of carbon fiber layer over it. A box was also incorporated into the harness for electronic part storage while at the same time offering an improved structural integrity.

III. Actuation Design Specifications

A. *Human Walking Biomechanics*

Human walking data were used in order to specify the design requirements for actuation at the exoskeleton joints. The power profiles for the hip, knee and ankle in the sagittal plane are plotted for a number of sets of gait data ([9], [10], [11], [12]). A conservative estimate of the weight of the exoskeleton and payload was chosen to be 60kg and the normative data were scaled to a 60kg person in order to estimate the torques and powers required at the joints of the exoskeleton. In estimating the torque and power requirements at the hip joint of the exoskeleton, the normative data were scaled to a 135kg person. This was due to the fact that the design goal was to have the actuator at the hip assist the exoskeleton (60kg) as well as the human (75kg). The human was assisted by means of a thigh cuff attachment between the human and exoskeleton thigh. The torque vs. angle plots use the data set from [11] which is for a walking speed of 0.8m/s.

A number of assumptions were made in the application of the human biomechanical data to the design of the exoskeleton. The first is that the exoskeleton carries its own weight, power supply and payload. The second assumption is that joint torques and joint powers scale linearly with mass. This second assumptions seems reasonable given that increases in vertical ground reaction force have been found to be proportional to increases in the load being carried [13]. The third assumption is that the exoskeleton will not greatly affect the gait of the wearer. Changes in gait have been shown to increase the physiological energy expended during locomotion [14]. Fig. 4 illustrates the significant regions of positive and negative power during the gait cycle. Specifications for actuation components as well as control strategies are extracted from angle, torque and power data at the human hip, knee and ankle joints in the sagittal plane.

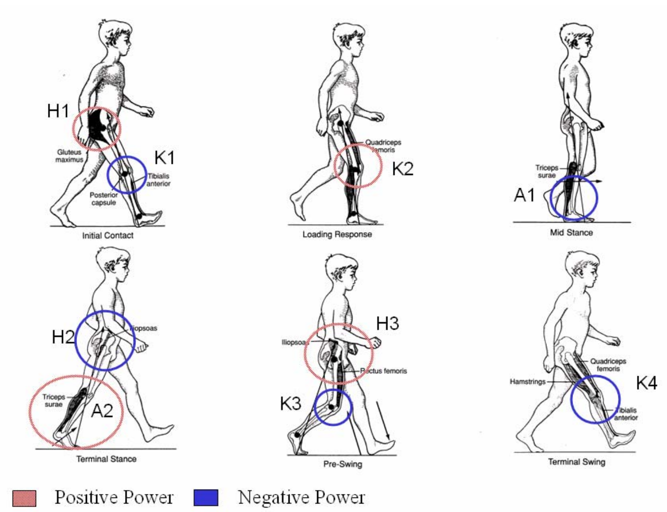

Fig. 4 Summary of significant regions of positive and negative work in walking. The instances labeled here are referred to later in this paper when examining human walking data.

1) *Hip*

During normal walking the human hip joint follows an approximate sinusoidal pattern with the thigh flexed forward on heel strike and then the hip moves through extension during stance as the body is pivoted over the stance leg in a pendulum-like motion. Positive power is required on heel-strike to raise the center of mass of the human over the stance leg. A peak negative hip torque of approximately 130Nm is experienced as the leg accepts load and the body's center of mass is raised. A maximum positive torque of about 100Nm occurs during the swing phase as the hip muscles provide energy to swing the leg forward. This action is sometimes referred to as "pull off," and is the muscular system's second largest contribution of propulsive power during the gait cycle. The power profile at the hip as a function of gait cycle is shown in Fig 5. The hip joint is the preferred location for a non-conservative actuator as proximal mass is less expensive metabolically in walking than distal mass.

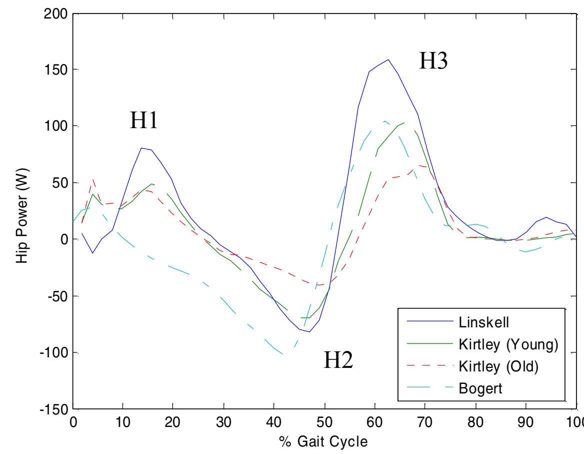

Fig. 5 Hip joint power profile scaled for a 135kg person as a function of the gait cycle. H1 is a small region of positive power, not always present, which corresponds to concentric hip extensor activity during loading response, H2 is a region of negative power, corresponding to eccentric hip flexor activity during mid-stance and H3 is a region of positive power, corresponding to concentric activity in the hip flexors during pre-swing and initial swing.

An actuator could assist in adding power in the H1 and H3 regions. From Fig. 5 it can also be seen that a spring placed at the hip joint could absorb the negative energy in H2 and release it during H3 to assist in swinging the leg forward. In Fig. 6 an approximate linear relationship can be seen between the hip torque and angle during the stance phase for slow walking (0.8m/s). The spring constant for such an "extension spring" was estimated as 115Nm/rad. As well as adding power throughout the gait cycle, a force-controllable actuator at the hip could be programmed to experiment with various impedance values.

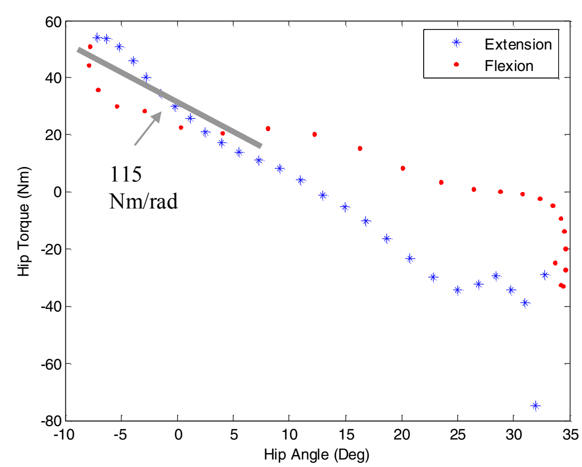

Fig. 6 Hip angle plotted versus hip torque for a walking speed of 0.8m/s.

2) *Knee*

In walking, the knee joint acts primarily as a variable-damper. Fig. 7 outlines the power of the knee as a function of gait cycle. It can be seen that the power is largely negative indicating that the knee absorbs power for the majority of the gait cycle. At heel strike there is a region of negative power followed by a period of positive power as the knee goes through stance flexion-extension. This is followed by a period of negligible joint power as the knee is passively extended. For a large part of the swing phase the leg has a pendulum like motion with the knee varying the damping to control the swing leg duration.

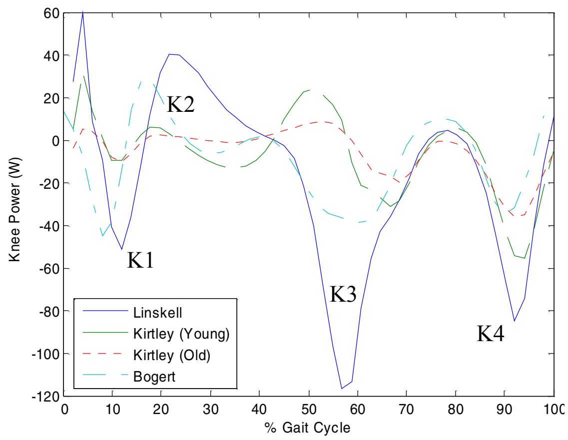

Fig. 7 Knee joint power profile scaled for a 60kg person as a function of gait cycle. K1 is a region of negative power, corresponding to eccentric knee extensor activity during the loading response, and K2 is a region of positive power, corresponding to concentric knee extensor activity during mid-stance. K3 is a region of negative power, corresponding to eccentric activity in the rectus femoris during pre-swing, and K4 is a region of negative power, corresponding to eccentric activity in the hamstrings during terminal swing.

It can be seen in Fig. 7 that during flexion-extension during early stance, the knee behaves like a spring as there is a region of negative energy followed by a region of positive energy of similar size. Fig. 8 shows a plot of knee angle vs. torque and a linear relationship can be seem during the stance phase. For the remainder of the gait cycle, the knee acts like a variable-damper to control leg during the swing phase.

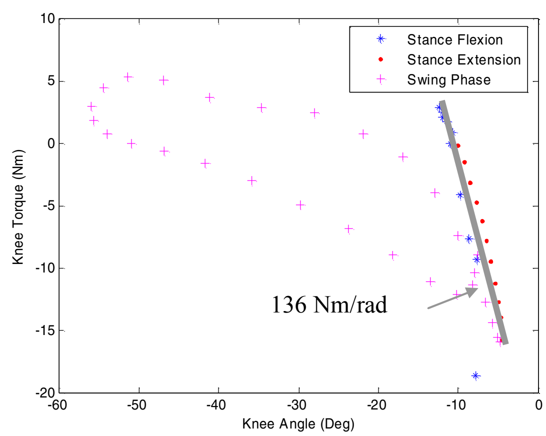

Fig. 8 Plot of knee angle versus knee torque for the walking cycle. It can be seen that the knee behaves primarily as a variable damping device throughout the gait cycle.

From the gait data it appears that the ideal actuator for the knee of the exoskeleton is a spring with a variable-damper. The spring would provide a resistive torque at the knee on heel strike as energy is absorbed and this energy is then released to aid in knee extension during stance. During the swing phase, the variable-damper would be engaged to control the swinging of the leg. It should be noted, for walking on a decline or down stairs, the variable-damper would be required during the stance phase to dissipate energy. For the initial implementation a variable-damper mechanism was used without the spring. The damper was able to provide the necessary resistive torque during stance but the negative energy was dissipated as heat.

3) *Ankle*

During the mid and late stance phases of walking the ankle joint torque is negative for approximately 40% of the gait cycle as the ankle controls the forward movement of the center of mass. The maximum positive power input during the gait cycle occurs at toe off. At that time, the ankle torque is at its largest, approximately 90Nm.

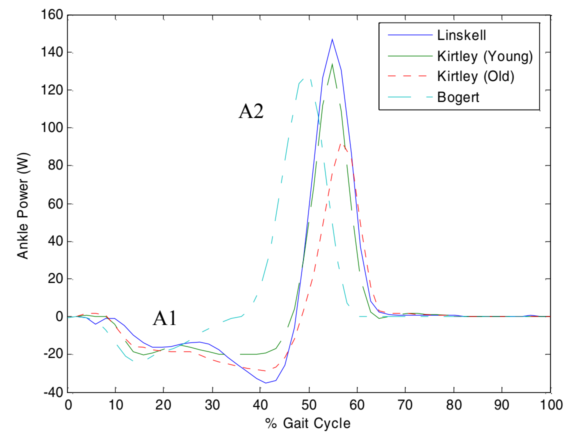

Fig. 9 Ankle joint power profile scaled for a 60kg person as a function of gait cycle. A1 is a region of negative power, corresponding to eccentric plantar flexor activity at the ankle during mid-stance and terminal stance, and A2 is a region of positive power, corresponding to the concentric burst of propulsive plantar flexor activity during pre-swing.

Fig. 10 shows the ankle torque plotted vs. angle for walking at 0.8m/s. A linear fit yields a spring constant for the ankle of 301Nm/rad for this walking speed.

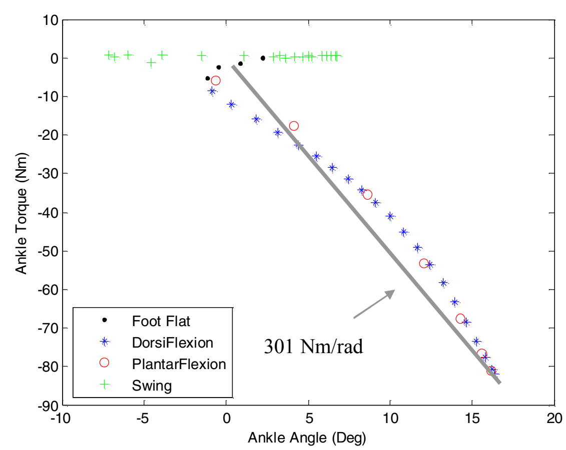

Fig. 10 Plot of ankle angle versus ankle torque for the walking cycle. It can be seen that the ankle behaves like a spring at a walking speed of 0.8m/s.

For slow walking speeds a spring is the ideal choice for actuation at the ankle as the energy absorbed during dorsiflexion is almost equal to the positive energy generated during controlled plantarflexion. Based on the ankle data in Fig. 10 the total energy that is absorbed and then later released is approximately 9J. At faster walking speeds the positive power becomes increasing large and in this case a hybrid actuation approach may be beneficial where a small motor is used in conjunction with the spring.

IV. Actuation Implementation

A. Hip

Two different actuation strategies were advanced at the hip. The first strategy comprises uni-directional springs placed at the hip. One spring is compressed during late stance hip extension as the backpack pendulums over the exoskeleton leg. In addition, a second uni-directional spring stores energy during hip adduction in walking and releases that energy during hip abduction. The second strategy comprises a force-controllable actuator placed at the hip joint to replace the spring for that is compressed during hip extension. The actuator was programmed to provide positive and negative power at the appropriate points during the gait cycle to augment hip extension and flexion movements.

1) Linear Series Elastic Actuator

Series elastic actuators (SEA) [15] were chosen as they provide a means for implementing lightweight and inexpensive force control in a bandwidth similar to that of natural muscle. The SEA has a spring in series with the output of the motor. The spring acts as a sensor, filter and impedance limiter. The actuator has a brushed DC motor driving a 3mm lead ball screw via a 2:1 reduction belt drive. The ball screw nut is coupled to the output through four die compression springs and the spring compression is measured with a linear potentiometer.

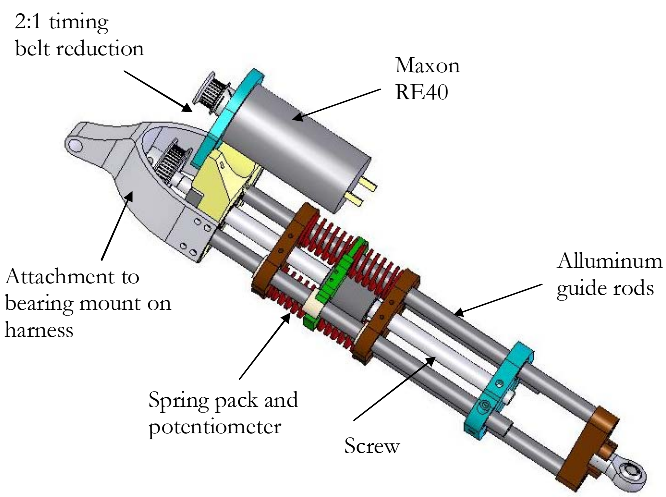

Fig. 11 Linear series elastic actuator.

The specifications for actuation at the exoskeleton hip joint outlined in the previous section were used to design the series elastic actuator. A 150W Maxon RE40 Brushed DC motor was chosen for its power to weight ratio. When the actuator is used at the hip of the exoskeleton it experiences two boundary conditions. The actuator may be either directly in contact with the environment or it may be connected to a freely moving inertial load. These boundary conditions represent the stance and swing phase of the walking cycle, respectively. During the stance phase, the load position can be considered a fixed position source, and in the swing phase, the load position is defined as a function of the force in the spring and the load mass. These boundary conditions were characterized separately in order to determine the performance of the actuator for each case.

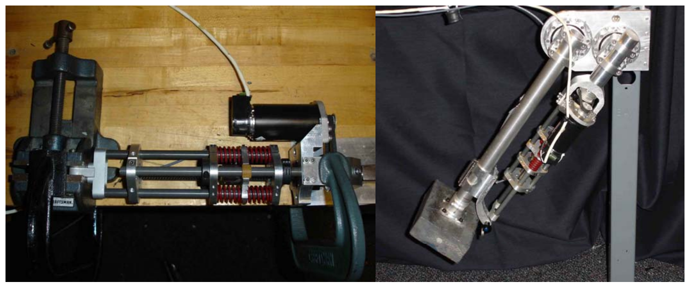

Fig. 12 Testing the hip actuator for two different boundary conditions. For the first case the actuator output is fixed. In the second case the actuator is moving a load mass.

The actuator was modeled as per Robinson [16] and based on the open loop Bode plot a lead compensator for the system was implemented. Shown below is a comparison of the theoretical and experimental Bode plot of the closed loop system for the fixed boundary condition. The bandwidth of the system for the fixed end condition was 35Hz. The closed loop Bode plot for the free end condition with a mass was also plotted and the bandwidth was found to be 40Hz.

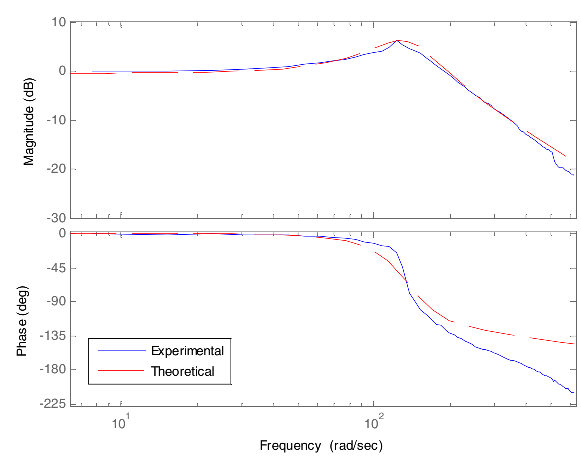

Fig. 13 Closed loop Bode plot for the series elastic actuator with the end condition fixed.

Shown below is the actuator tracking a sine wave with amplitude of 1600N at 5Hz for the fixed end condition.

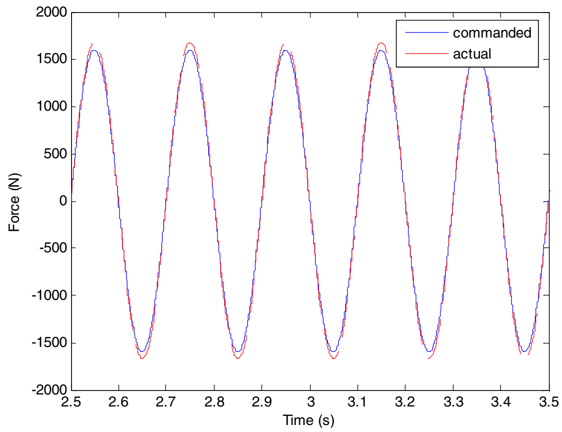

Fig. 14 Tracking a 5Hz sine wave with an amplitude of 1600N.

2) *Passive Hip Extension Spring*

The hip extension spring stores energy during hip extension and releases that energy during the thrusting, hip flexion phase. The hip spring was 1.5" in diameter, 4" in length, and its stiffness was 110 lb/in. A plunger was grounded to the cam. During hip extension, the plunger compressed the spring. A thin delrin plate covered the spring and acted as a low friction surface for contact with the plunger. A clear finger guard was placed on top of the spring holder for safety.

Fig. 15 Extension spring assembly mounted on the exoskeleton.

B. *Variable Damper Mechanism at the Knee*

During slow human walking, the knee behaves largely as a variable damper where minimal positive power is exerted. The knee of the exoskeleton was implemented with a magnetorheological damper with the fluid in the shear mode similar to [17]. The damper at the knee could exert a maximum braking torque of 60Nm and consumed on average approximately 1W of electrical power during walking.

C. *Uni-directional Spring at the ankle*

For slow walking, it has been shown that the ankle behaves like a spring. For the exoskeleton, a linear spring located at the ankle joint was designed to capture the negative energy during controlled dorsiflexion. This energy is subsequently released to assist the exoskeleton foot in plantar flexion as the foot comes off the ground. This rotary ankle spring was implemented by having a lever compress a linear urethane spring.

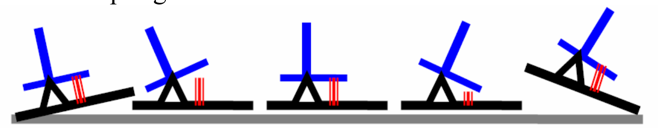

Fig. 16 Energy is stored in the red uni-directional ankle spring as it is engaged in mid-stance and the shin rocks forward. The energy in the spring is released to assist power plantar flexion.

Urethane was used for the spring material as it is lightweight. The linear spring was seen at the ankle joint as an effective rotary stiffness and the linear spring constant was transformed through an effective transmission due to the lever. The spring constant was 247kN/m and the lever arm was 0.0381m (1.5 in) which gave a joint rotary stiffness of 356Nm/rad.

D. *Adduction Spring at the Hip*

When the exoskeleton wearer stands on one leg, a moment was created by the backpack load since it was off center from the biological hip joint. This moment was undesirable and caused discomfort. To solve this problem a linear compression spring was placed at the hip joint of the exoskeleton so that it would compress to counter this moment during hip adduction. It is unidirectional so it allowed the hip to freely abduct. The spring was kept in place by means of a keeper that was attached to the exoskeleton hip joint assembly. The keeper was attached to the sagittal plane bearing so that the keeper rotated with the exoskeleton leg. A brass plunger fitted through a hollow spring and was kept from falling out by means of a nut that threaded on to the end of the plunger.

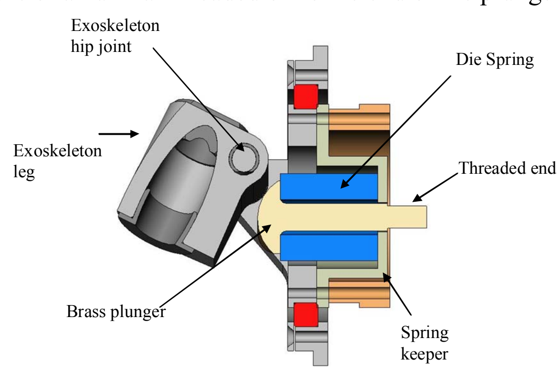

Fig. 17 Section view of the adduction spring assembly at the hip joint.

The rotary spring constant for adduction was obtained by using a free body diagram of the exoskeleton leg during single support to calculate the moment felt at the hip joint. This was calculated to be 8Nm. Then by examining human walking data it was assumed that the human leg undergoes approximately five degrees of adduction during normal walking. From these two values an effective rotary stiffness of 96Nm/rad was calculated. This was achieved by means of a linear spring of value 425kN/m compressed by a lever arm of 15mm.

VI. Preliminary Results

Initial walking experiments have been conducted with the exoskeleton loaded with a 75lb payload. The configurations with the passive extension springs and the non-conservative actuators at the hip were studied. It was discovered that the hip and knee angles followed similar trajectories to that of normal human gait kinematics, indicating that the exoskeleton does not negatively affect gait patterns in the sagittal plane. Further, it was determined that at least 90% of the weight of the payload and exoskeleton mass was transferred through the exoskeleton leg structure. Fig. 18 shows the load in the exoskeleton leg as a function of the gait cycle during a walking experiment. It can be seen that during the stance phase approximately 500N was transferred through the leg. At approximately 62% gait cycle, the load in the exoskeleton leg dropped to near zero as the leg entered the swing phase.

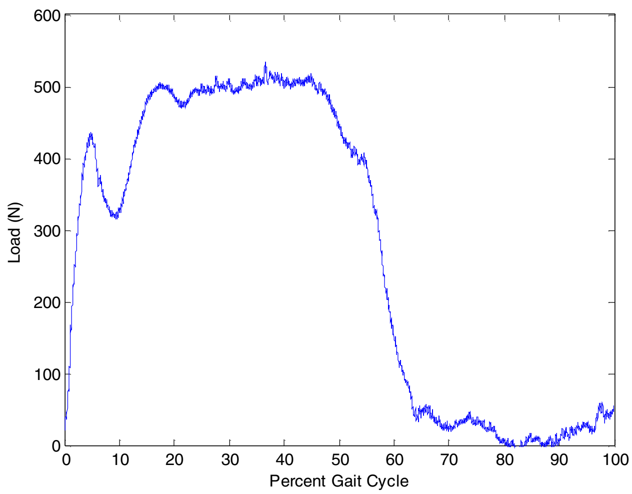

Fig. 18 Load in the exoskeleton leg as a function of gait cycle. Heel strike is evident by a large increase in measured force level. During the swing phase it can be seen that there is minimal load in the exoskeleton leg.

VII. Conclusions and Future Work

In this paper, a lightweight, underactuated exoskeleton is presented that runs in parallel to the human leg and transmits payload forces to the ground. Although primarily passive in design, the leg exoskeleton mechanism is shown to effectively transmit payload forces to the ground during walking. Through the analysis of fast walking and running biomechanics, we hope to design exoskeletons in the future that will allow for higher speed locomotory function. In addition to augmenting human strength and endurance, the advancement of leg exoskeletons can contribute to the science of bipedal walking and lead to a better understanding of locomotory biomechanics, energetics and control.

VIII. Acknowledgements

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