Robotic Augmentation

Of Human Locomotion

For High Speed Running

by

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ABSTRACT

Human running requires extensive training and conditioning for an individual to maintain high speeds (greater than 10mph) for an extended duration of time. Studies have shown that running at peak speeds generates a high metabolic cost due to the use of large muscle groups in the legs associated with the human gait cycle. Applying supplemental external and internal forces to the human body during the gait cycle has been shown to decrease the metabolic cost for walking, allowing individuals to carry additional weight and walk further distances. Significant research has been conducted to reduce the metabolic cost of walking, however, there are few if any documented studies that focus specifically on reducing the metabolic cost associated with high speed running. Three mechanical systems were designed to work in concert with the human user to decrease metabolic cost and increase the range and speeds at which a human can run.

The methods of design require a focus on mathematical modeling, simulations, and metabolic cost. Mathematical modeling and simulations are used to aid in the design process of robotic systems and metabolic testing is regarded as the final analysis process to determine the true effectiveness of robotic prototypes. Metabolic data, (VO2) is the volumetric consumption of oxygen, per minute, per unit mass (ml/min/kg). Metabolic testing consists of analyzing the oxygen consumption of a test subject while performing a task naturally and then comparing that data with analyzed oxygen consumption of the same task while using an assistive device.

Three devices were designed and tested to augment high speed running. The first device, AirLegs V1, is a mostly aluminum exoskeleton with two pneumatic linear actuators connecting from the lower back directly to the user's thighs, allowing the device

to induce a torque on the leg by pushing and pulling on the user's thigh during running. The device also makes use of two smaller pneumatic linear actuators which drive cables connecting to small lever arms at the back of the heel, inducing a torque at the ankles. Device two, AirLegs V2, is also pneumatically powered but is considered to be a soft suit version of the first device. It uses cables to interface the forces created by actuators located vertically on the user's back. These cables then connect to the back of the user's knees resulting in greater flexibility and range of motion of the legs. Device three, a Jet Pack, produces an external force against the user's torso to propel a user forward and upward making it easier to run. Third party testing, pilot demonstrations and timed trials have demonstrated that all three of the devices effectively reduce the metabolic cost of running below that of natural running with no device.

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CHAPTER 1
INTRODUCTION

There are many methods to increase the speed and duration for which humans can travel. One common method for faster travel is through the use of wheeled vehicles, however, there are times when it would be beneficial for a person's natural ability to run to be augmented. When compared to common apparatuses such as wheeled vehicles, the human gait cycle allows for increased mobility where wheeled vehicles cannot travel due to incompatible terrain such as narrow passages, rocky terrain, or stairs. Although walking or running can allow for transportation of individuals through areas where wheeled vehicles cannot go, the duration of transportation and speed is heavily constrained by the physical capabilities of the human in question. Some individuals that are required to run for long distances at elevated speeds include infantry soldiers, police officers, fire fighters, and athletes.

Individuals that would benefit from exceeding their natural physical capabilities are consistent with those that perform strenuous activities where physical performance directly relates to their overall success. In cases regarding military infantry, soldiers are often required to traverse long distances over rough terrain which can greatly impact their success in the battlefield (Fletcher, 1974). A robotic exoskeleton which augments human locomotion of the legs could increase a soldier's ability to accomplish strenuous missions, preserve cardiovascular performance leading to improved decision making, reduce the potential for injury due to over exertion, and enhance peak running speeds making a soldier faster than their enemies. The improved characteristics of soldiers using mechanical devices for augmentation could potentially result in the preservation of

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human life when combined with strenuous, dangerous missions. For athletes, over-speed running and training has become popular using pulley systems and springs connecting two runners together (Behrens, 2011). Athletes could utilize mechanical devices to improve their performance in a variety of sports or could use devices for over-speed training so their capabilities are improved after using the device.   Previous research has led to a variety of exoskeletons being developed that increase user strength; however, often times these devices are designed using hydraulics with a focus on generating extreme forces up to 5 times of what a human can naturally produce (Chu, 2005). This may create a very powerful system, but the systems are also heavy and slow, often increasing the metabolic cost of natural movement. Furthermore, if a system malfunction occurs, this could result in significant damage to the user due to the extreme forces the exoskeleton creates.

The research contained within this thesis explores the application of light weight robotic systems which augment human locomotion for running at high speeds through a variety of actuators and application of external forces to a user. Furthermore, it explores the requirements of control systems to sense the user's movement for autonomous control and how properly timed control systems can inject energy into the human gait cycle. The systems designed to augment human running described in this thesis consist of two pneumatic devices (AirLegs V1 and V2) that apply torque at the hips and or ankles, and a Jet Pack that applies an external forward and upward force at the user's torso.
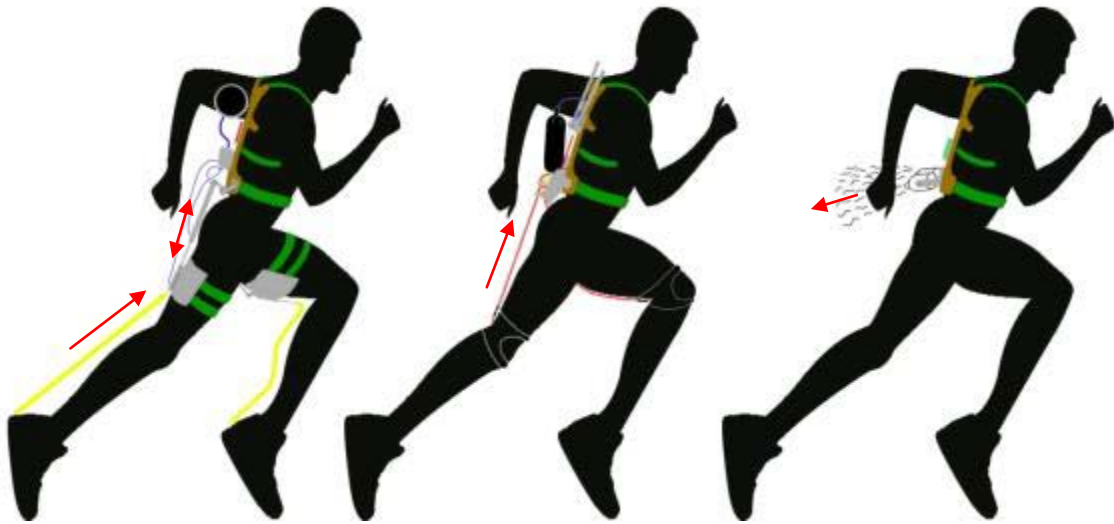
Figure 1.1. Three Devices with Forces Shown in Red Arrows, From Left to Right; AirLegs V1, AirLegs V2, Jet Pack

1.1   Force Effects on Humans Running Over Ground

During the running gait cycle, the main external forces acting on the runner's

motion are air drag and rolling friction.

Propulsive Force

$F_{propulsive} = 0.1mg$

Drag Force

$F_{drag} = 0.5\rho ACV^2$

Rolling Resistance

$F_{friction} = 3mgA_r$
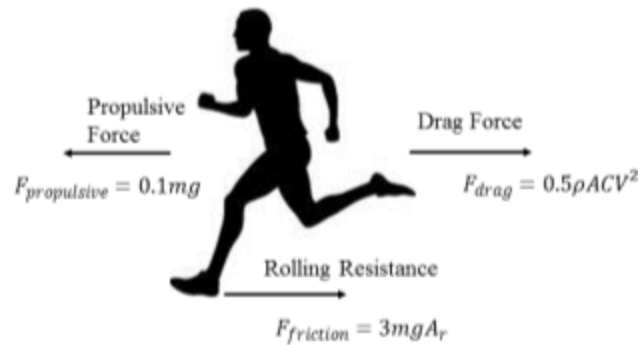
Figure 1.2. Forces Acting on the Runner

The external forces acting on a runner, coupled with supporting the mass of the runner,

are attributed to metabolic expenditure because the user's musculoskeletal system must

overcome these forces to run at high speeds requiring significant amounts of oxygen.

Furthermore, the metabolic cost associated with running has been shown to increase as

relative running speeds increase (Gottschall, 2003). Three power curves based on the

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runner's speed are demonstrated in figure 1.3. The rolling friction creates a large braking power proportional to weight. The power loss due to air drag is proportional to the cube of speed but is still smaller than frictional drag at high speeds (6.7 m/s or 15 mph). The total propulsive power must be equal to the total braking power plus drag power at high speeds which is reasonable for steady state motion.
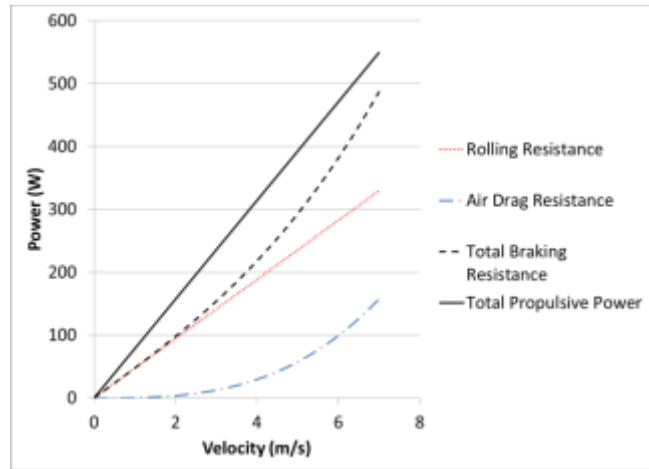
Figure 1.3. The Propulsive Power Needed to Run Versus the Braking Power

Research also suggests that external forces significantly greater than the required propulsive force may not be optimal for reduction of metabolic cost in regards to the human gait cycle. One example of this is demonstrated in a study where different levels of an external force were applied to a user on a treadmill (Gottschall, 2003). An optimal metabolic reduction of 47% was determined when 10% of the user's body weight was applied as a forward force. In similar tests, this metabolic reduction was diminished as the forward force was increased above 10%, demonstrating that too much force was not beneficial. Thus, an exoskeleton or mechanical device designed to augment running must be comprised of lightweight components and offer the correct amount of assistance. If too much assistance is applied, the user's metabolic savings may diminish.
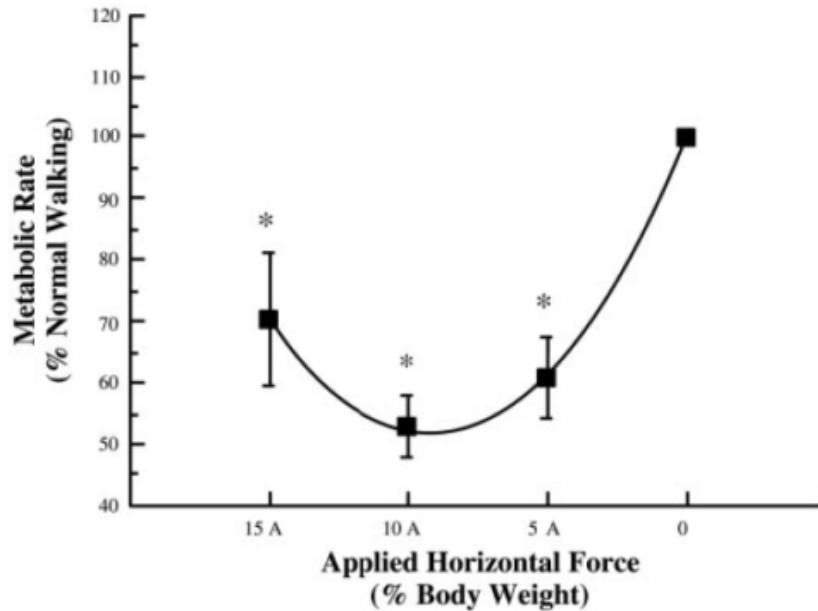
Figure 1.4. Metabolic Cost Versus Applied Horizontal Force (Gottshall, 2003)

1.2　Requirements of control systems

To achieve autonomous control of a robotic exoskeleton, a control system must be designed in a way that enables the device to operate in concert with its user. This can be done by utilizing various sensors, switches, or user input (Ollinger, 2007). One problem that is often encountered when developing a control system in which a robot is to augment human locomotion is that if sensors are used to monitor limb movement, then applying forces to that limb adds impedance. This can cause the control system to lose stability because it is now manipulating its own input. Groups have attempted to address this issue by means of adding inertia compensation for more precise control; however this then leads to a delayed control system, especially if using force sensors that are not mounted closely to the physical interface of the user (Zanotto, 2013). Another approach to exoskeleton control is the use of a feed forward control system. Feed forward control systems have been used to compensate for the effects of gravity on a user's limbs but

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these can quickly become unstable as well when coupled with the added inertia of an exoskeleton component combined with the user's limb. Furthermore, control systems designed to function based on negative damping of the user's limb, coupled with feed forward control and no inertia compensation, are likely to become unstable as well (Zanotto, 2013). Although there is no known research regarding high speed running and disturbances to the human body such as an exoskeleton becoming unstable or moving out of sync with the user, it is easy to imagine that a poor control system would cause the user to feel uncomfortable, especially with the risk of falling. It can also be assumed that if an exoskeleton is not perfectly in sync with the human i.e. significantly delayed or advanced actuators, the exoskeleton would not induce running augmentation because the user would be working against the robot for a portion of the gait cycle (Ollinger, 2007).

1.3   History of Exoskeletons

The first recordings of exoskeleton type devices date back to the late 1800's when Nicholas Yagn began building what he called an "Apparatus for facilitating walking, running, and jumping" (Yagn, 1890). At the time, microprocessors had yet to be developed to control the device which resulted in a mostly passive apparatus; however, there were some pneumatic hand controls to activate artificial muscles.
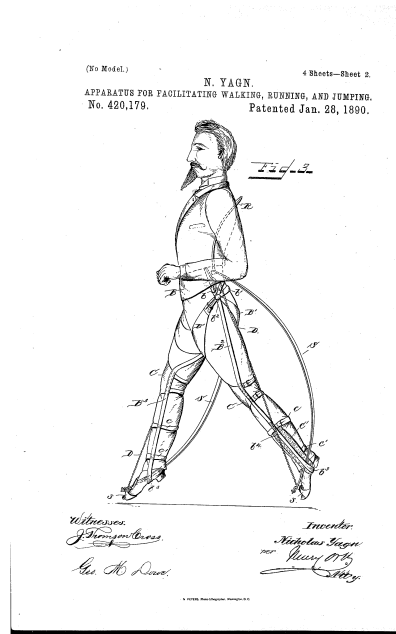
Figure 1.5. Apparatus for Facilitating Walking, Running, and Jumping (Yagn, 1890)

From information contained in a patent filled on February 11th 1890, Yagn developed an air powered device which was intended to utilize passive springs and cords to transfer momentum of the torso into energy for moving the feet at a higher rate of speed while carrying a load. Due to lack of documentation besides the patent which it was filed under, the success of this device cannot be determined. It was not until the mid 1960's that documentation of another exoskeleton device was published when General Electric developed and tested an exoskeleton named Hardiman (General Electric, 1968).

Figure 1.6. Hardiman Designed by General Electric (General Electric, 1968)

The Hardiman device, a hydraulic full body exoskeleton, was designed to allow a user to manipulate extremely heavy objects up to 250lbs and consisted of a master-slave type approach to have the exoskeleton follow the movements of its pilot. Although they created a full scale prototype, the Hardiman project was eventually abandoned because the device was never capable of functioning properly due to weighing 1500lbs and limitations in sensing and actuator technology. When tested, the Hardiman would become violently out of control. Since the 1960's, advancements in technology have lead to many improvements in not only sensors, but hardware as well. With the development of the microprocessor in 1972 and continuous development thereafter, devices to assist human locomotion became much more practical (Ceruzzi, 2003). Within the last 10 years, exoskeleton type devices have become much more prevalent with many researchers developing their devices in conjunction with robotics. An example of an

exoskeleton using microprocessors and integrated sensors is the Berkeley Lower

Extremity Exoskeleton (BLEEX). BLEEX was developed in 2001 and connects to a

user's back and legs.

Figure 1.7. BLEEX Exoskeleton Designed by Berkeley Researchers (Chu, 2005)

BLEEX applied torques to a user's hip, knee and ankle joints with hydraulic

actuators which were powered through an onboard combustion engine (Chu, 2005). The

device was designed to follow the user's legs and support additional payload on the user's

back. Lack of success of the device was attributed to the use of hydraulic systems and

actuators that could not respond fast enough and the device was not effective for running.

1.4   Objective

The objective of this thesis is to present alternative methods for augmentation of

the human running gait through a variety of different robotic devices. Three devices have

been designed, built, and tested to determine the most effective method for adding energy

into the human gait cycle and thereby reducing the metabolic cost associated with

running at high speed.

Figure 1.8. Devices Designed to Augment Human Running. From Left to Right, AirLegs V1, AirLegs V2, and Jet Pack

Furthermore, the devices described in this thesis were designed to be as simple as

possible to minimize the number of sensors and actuators required for operation while

still attempting to augment human locomotion.

1.5   Hypothesis

The general hypothesis for this research project is that by utilizing robotic

mechanisms that exert an appropriate magnitude of force at the right time, human

locomotion for running at high speed can be augmented, effectively reducing the

metabolic cost of running. Three methods for achieving human locomotion augmentation

were designed, built and tested. The specific hypotheses for this research is as follows:

- An exoskeleton device consisting of pneumatic actuators connected to the legs and ankles will allow for running at high speeds while reducing the metabolic cost of running.

- A soft-suit exoskeleton using cables connected to pneumatic actuators at the user's back and knees will assist human locomotion and reduce metabolic cost for running.

- A propulsive device at the user's back will assist human locomotion and reduce the metabolic cost for running

1.6   Thesis Outline

The remaining sections of this thesis are devoted to the exploration of human running gait, development of a phase controller, and development of devices which augment human locomotion for high speed running. Chapter 2 presents a general background on the human gait cycle and explores methods for determining the required forces needed to enhance running. Chapter 3 describes the development of a phase oscillator to control exoskeleton movement. Chapter 4 describes the design of the three devices developed to augment human running. Chapter 5 presents methods for analytical testing and determining the metabolic efficiency of the devices. Chapter 5 focuses on reviewing the devices and conclusions.

CHAPTER 2

GAIT EXPLORATION

Chapter two explains the human gait cycle and background methodology for analyzing necessary forces for running enhancement and the significant aspects of the gait cycle are determined.

2.1   Human Gait Cycle

Humans, like most other bipedal animals, have the capability of walking by alternating their legs in a particular fashion. Walking or running, although a complicated procedure of movements, requires almost no conscious thought about individual muscle actuation. For a lower limb robotic exoskeleton, each actuator must be independently controlled in perfect time with its human pilot to keep the device from obstructing the path of the user's legs. The human gait cycle is commonly broken down into two main phases; the stance phase, and the swing phase (Vaughn, 1992). The two phases combined make up a common walking procedure starting at heel strike of one foot, and continues until that same foot has left the ground and returns to its original orientation. The stance phase begins when one foot first touches the ground known as a heel strike and continues until push off. The swing phase then contains motions where that same foot is in the air, until just before heel strike. The swing and stance phases contain multiple orientations throughout their duration. For most individuals, general walking and running gaits are largely uniform across a broad spectrum of people.
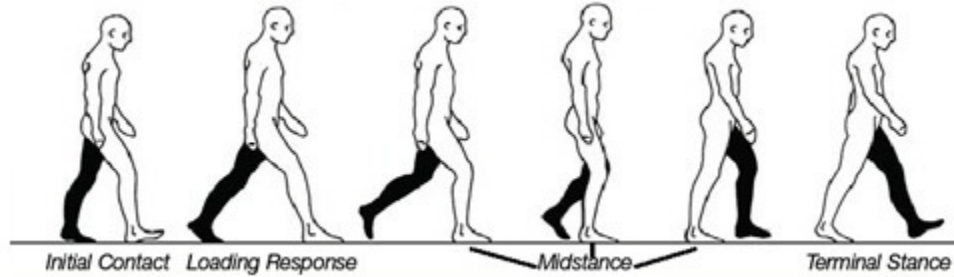
Figure 2.1. The Walking Gait Cycle (Donatelli, 1996). unshaded leg in stance phase and shaded leg in swing phase.

According to the Gait Analysis Laboratory (Vaughn, 1992), the stance phase is responsible for "first double support" which is when both feet are on the ground. It is also responsible for "Single limb stance" where only one foot is making contact with the ground and "second double support" where both feet are back on the ground. Although similar, the average running gait for humans contains slight variations within the stance and swing phase because at high speed, both feet never make contact with the ground at the same time.
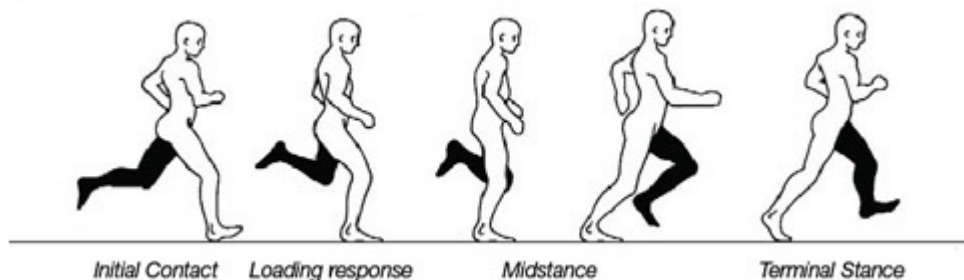
Figure 2.2. The Running Gait Cycle (Donatelli, 1996) ; unshaded leg in stance phase and shaded leg in swing phase.

With regards to the running gait cycle, focusing only on the right (unshaded) leg in figure 2.2, it can be seen that propulsive forces originate during the stance phase when a runner is pushing off the ground. During this time, the runner's leg is rotating clockwise,

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utilizing a clockwise torque at the hip. The opposite (shaded) leg is in swing phase at the same time and a counterclockwise torque is required to rotate the leg forward.

Because walking and running gaits are different which can be seen when comparing figure 2.1 and figure 2.2, one critical component of a controller must be its capability to transition from a walking gait to a running gait if it is to keep from inhibiting its user. This poses a difficult problem for robotic systems that use a state based approach because the robot must detect and recognize different gait cycle patterns. This concept is explored further in Chapter 3.

During the running gait cycle, the main external forces acting on the runner's motion are air drag and rolling friction.

Propulsive Force
$F_{propulsive} = 0.1mg$

Drag Force
$F_{drag} = 0.5\rho ACV^2$

Rolling Resistance
$F_{friction} = 3mgA_r$
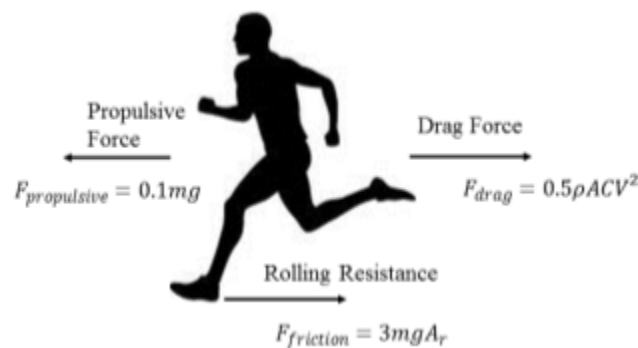
Figure 2.3. Forces Acting on the Runner

These forces, coupled with supporting the mass of the runner, are attributed to metabolic expenditure because the user's musculoskeletal system must overcome these forces to run at high speeds. It can be determined by the common observer that propulsive forces for natural running are only generated during the stance phase, while the feet are on the ground. Thus, for a moment in the running gait cycle, there is no propulsive force being generated while both feet are off the ground. Also, rolling resistance creates a large braking power proportional to runner's weight which also only occurs during the stance

phase as well. (Gottschall, 2003). The power due to air drag is proportional to the cube

of speed but is still smaller than frictional drag at high speeds (6.7 m/s or 15 mph) and

occurs at all times. For steady state motion, the total propulsive power must be equal to

the total braking power at plus drag power. All three power curves based on the runner's

speed are demonstrated in figure 2.4.
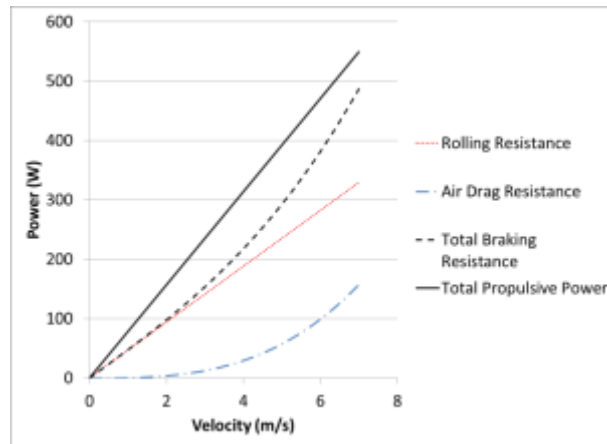
Figure 2.4. The Propulsive Power Needed to Run Versus the Braking Power

2.2   Pulling Forces Required for Augmentation

Studies have shown that researchers have previously analyzed human running

gate using treadmills in combination with pulling forces and discovered that a savings in

metabolic cost of up to 47% is possible with the correct towing force (Kram, 2008). Over

ground testing was selected, unlike Kram's research, avoiding the use of a treadmill

because there are reduced inertial effects and reduced wind resistance. Over ground

testing would allow for testing with more realistic rolling friction and wind resistance as

well as inertial effects. A testing rig was developed to explore the human gait cycle for

running over ground to encompass all forces experienced for natural running. The testing

rig was designed for over-ground experimentation and was outfitted with a load cell in

series to a towing cable, which was then connected to a climbing harness located on the

pelvic area of a test subject. The testing rig was then attached to an electric golf cart to impose towing forces on a runner. Two test variables were determined for this experiment, towing force, and towing angle. Tests at different speeds were performed with different towing angles to determine the most effective combination of towing force and angle to produce the greatest metabolic reduction.

Figure 2.5. Testing Rig Attached to an Electric Golf Cart

Similar to Kram's research, where testing was conducted by adding towing forces on a treadmill, experiments were created using the testing rig to determine what forces would be necessary to reduce metabolic cost when running outdoors on an athletic track. During the experiments, the runner wore a respirator like device to record metabolic data. The load cell, in series with the towing cable, was used to determine the magnitude of pulling force required to increase running speeds and reduce the metabolic cost a runner experiences while running at high speeds.

Figure 2.6. Pull Force Testing with Golf Cart Rigging

Figure 2.7. Pulling Test Subject While Recording Metabolic Pilot Data

Various trials of assisted and unassisted runs were completed with tow angles

ranging from 5 degrees below horizontal to 35 degrees above horizontal. With each trial,

the golf cart was kept at a constant speed, allowing the test subject to determine the

appropriate amount of towing force based on what felt the best and how much force they

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exerted against the ground. The test subjects demonstrated that it felt most comfortable

to utilize around 10% of their body weight as a towing force. Towing forces above 10%

made the runner feel unstable causing their feet to slap into the ground increasing their

rolling friction and effectively increasing metabolic cost. Towing forces below 10%

forced the runner to exert higher push off forces increasing metabolic cost as well. The

following figures demonstrate how different runners, running at different speeds, all

utilized nearly 10% of their body weight as a towing force to produce the most metabolic

savings.
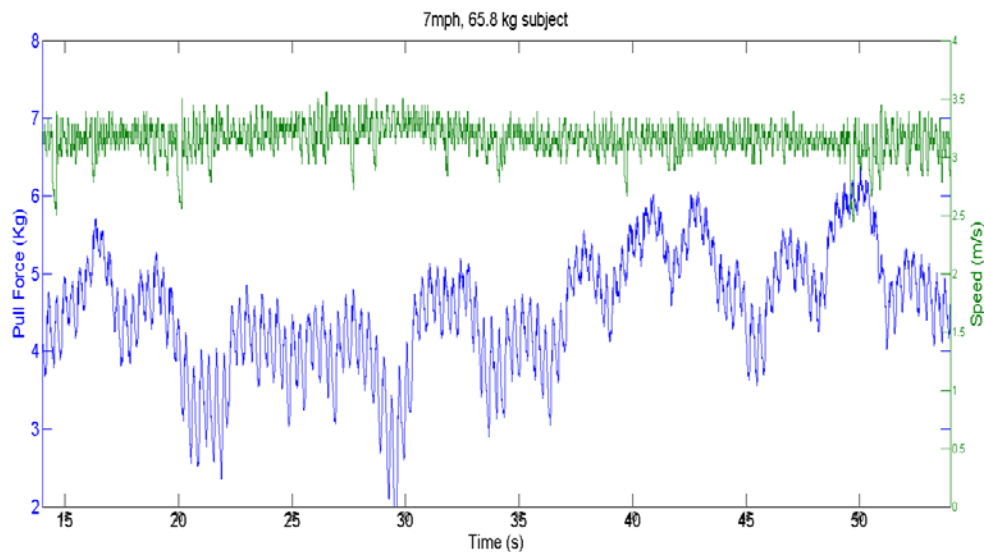
Figure 2.8. Speed Versus Pulling Force at 7Mph; The speed is shown in green and is measured using the vertical axis on the right. The pulling force for a 65.8 kg subject varies between 2 - 6 kg, and averages 5 kg or approximately ten percent of body weight. The pulling force is measured using the vertical axis on the left.
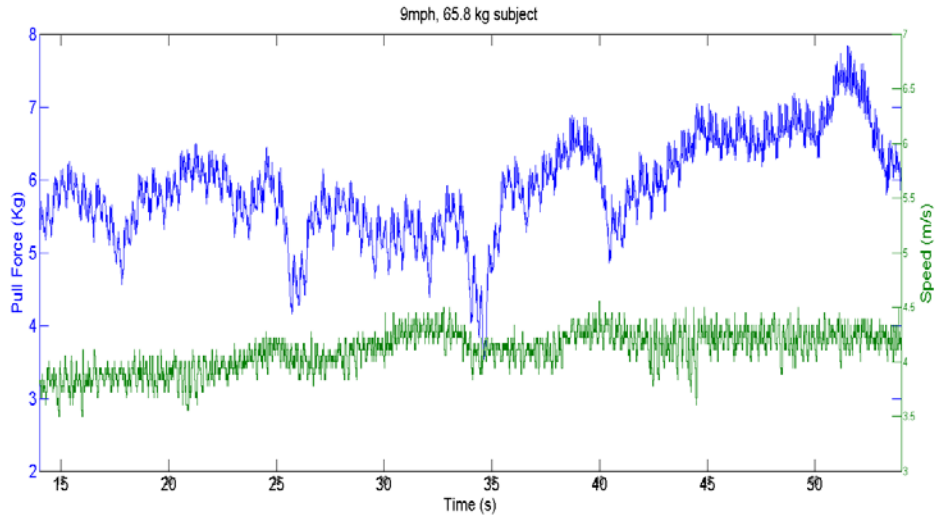
Figure 2.9. Speed Versus Pulling Force at 9Mph; The speed is shown in green and is measured using the vertical axis on the right. The pulling force for a 65.8 kg subject varies between 4 - 8 kg, and averages 6 kg or approximately ten percent of body weight. The pulling force is measured using the vertical axis on the left.
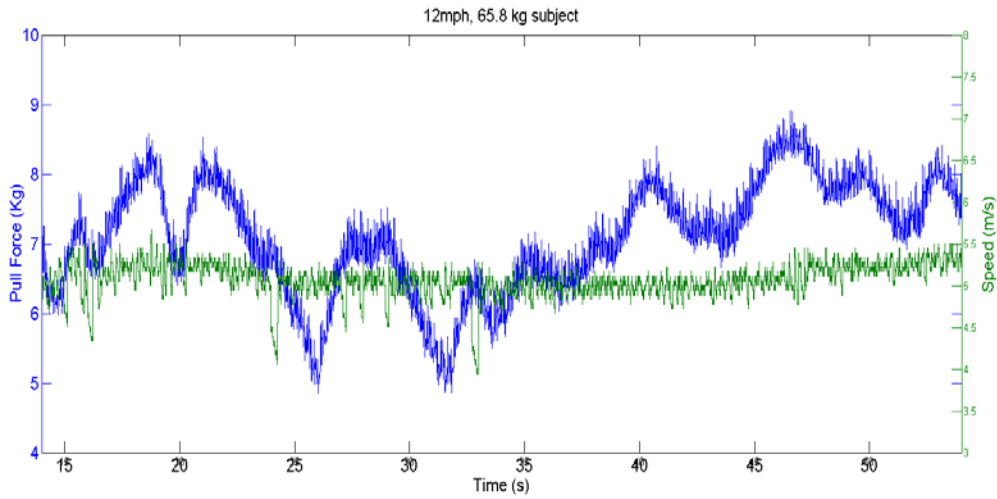
Figure 2.10. Speed Versus Pulling Force at 12Mph; The speed is shown in green and is measured using the vertical axis on the right. The pulling force for a 65.8 kg subject varies between 5 - 9 kg, and averages 7 kg or approximately ten percent of body weight. The pulling force is measured using the vertical axis on the left.

The towing angle was also controlled while towing runners at different speeds and

monitoring the amount of force imposed on the runner.

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Testing showed that using around 10% of body weight as a towing force resulted in peak metabolic savings and was a significant finding because the data was very similar to the data recorded by the experimental tests performed by Kram (Kram, 2008). However, unlike Kram's research, the metabolic savings for running over ground with a towing force were closer to 30% savings. The metabolic cost is believed to be slightly off from Kram's work because running over ground induces drag force. With experimental data that validated Kram's research, it was clear that not only would a towing force help runners on a treadmill, but over-ground as well. Peak metabolic reductions were obtained when the towing angle was close to 25 degrees above horizontal resulting in a reduction in metabolic cost of 29.8%.

Figure 2.11: Greatest Reduction in Metabolic Cost from 25 Degree Angle

It was determined that the force should be angled at 25 degrees so that an upward force as well as a propulsive force is generated. If the force was angled downward, the feet would slap the ground adding discomfort and increasing the potential breaking force. Similar to a runner carrying an additional payload or increasing their body weight, increased breaking force is the result of increased rolling friction. A slight upward force acts against the natural gravitational force, reducing ground reaction forces upon landing with

each foot. Kram's research on treadmills also concluded that both vertical (reducing the runner's potential mass) and horizontal (forward forces) made running easier (Kram, 2008). When both the optimal tow force of 10% of the user's body weight and a 25 degree angle towing force were used, significant metabolic reductions were exhibited. In multiple trials over a 400 meter course, using data from 60 seconds of the test to 120 seconds, an average metabolic savings of 29.8% was recorded. Data points in the middle of the test, 60 seconds - 120 seconds, were used because during this time span, the runner and golf cart were traveling at a nearly constant pace. For times outside of this span, the runner and golf cart were either speeding up or slowing down and acceleration of the subjects and cart were not monitored.

Metabolic Cost 12mph Tow VS Free Run
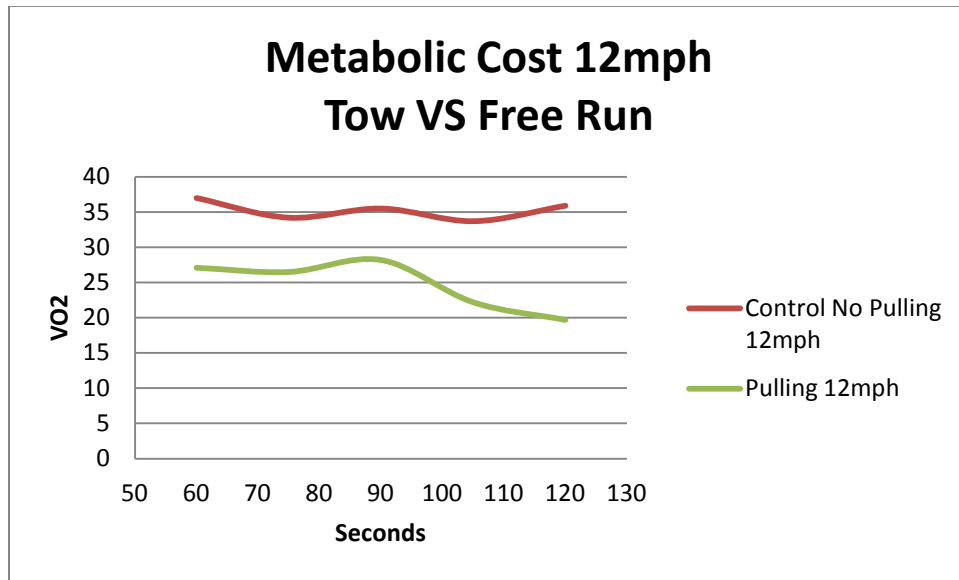
Figure 2.12: Metabolic Cost Free Running Versus Towed Running

The heart rate of the runners was also recorded during testing and was reduced as well with an external horizontal/vertical force.

2.3   OpenSim Modeling of External Propulsive Force

A three dimensional model was created using Stanford's OpenSim software to identify the specific muscle groups assisted with a towing force at 10% of a user's body mass applied at an angle above horizontal.
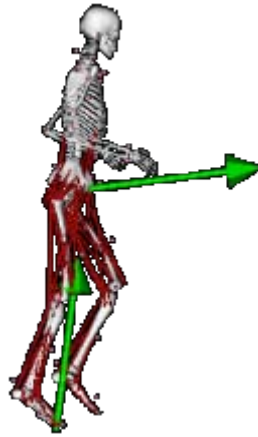
Figure 2.13. OpenSim 3D Model of Muscular and Skeletal System

The model was imported from a running model created by Samuel Hamner and utilizes kinematics from a subject running at similar speeds (Hamner, 2010).   Additional forces consistent with that of a towing force were simulated by manipulating the forces data file manually prior to calculating inverse kinematics. Based on Hamner's research, the soleus and gastrocnemius muscles are two of the muscles considered to be the "greatest contributors to propulsion and support," (Hamner, 2010). The soleus and gastrocnemius are located on the lower shank of the leg, commonly referred to as the calf muscle. It can then be determined that because these muscles largely contribute to the propulsive force, assisting them would yield the greatest metabolic reduction. The simulation validates that a forward and upward force applied at the pelvis decreases the muscle activation forces of the soleus and gastrocnemius muscles as seen in figure 2.14
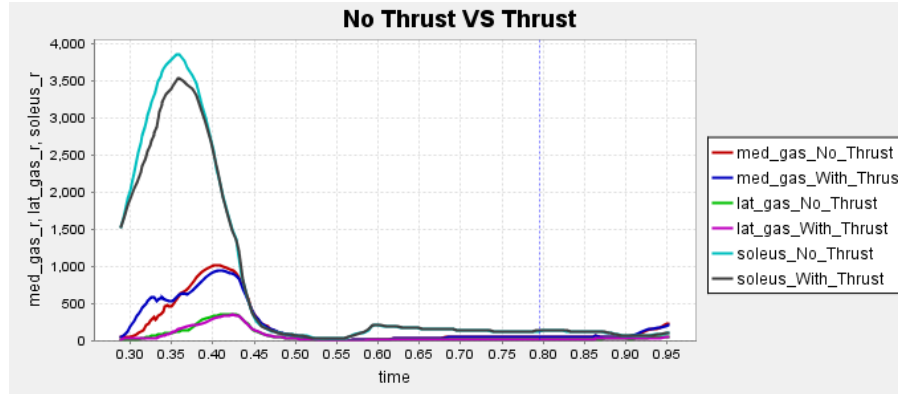
Figure 2.14. Muscle Activation Energy With and Without Towing/Thrust Force.

However, unlike initial testing via tow rope and load cell, the simulation does not reflect

an optimal towing force to assist locomotion and implies that more towing force is better

without a limit to how much force is too much. The Stanford OpenSim software depicts

that adding an exponentially large towing force helps the leg muscles much more than

adding 10% of the runners mass because the software relies on input data from real test

subjects and cannot modify the trajectory of the model. The input data for OpenSim

includes marker data from a motion capture system and ground reaction forces acquired

from force plates during real experiments; thus, the output data produced by modifying

input files may have a small margin of error. Experiments performed with the golf cart

and towing rig showed that adding too great of a forward force made the runner feel

unstable with the sensation they were falling forward. The simulation does not model

this stability issue as the marker trajectory and force plate data cannot be modified

without performing new gait trails. Instead, the software places additional energy into

what is known as residuals, limiting the decrease of simulated muscle activation force.

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CHAPTER 3

CONTROLLER METHODOLOGY

Control systems are needed to add energy and assist body motion in areas such as powered orthotics, prosthetics, and exoskeletons (Hitt, 2007 and Wang, 2007). Systems can add energy based on negative damping (Ollinger, 2007), patterns based on phase angles (Holgate, 2008) and impedance control (Hogan, 1984). Using a phase angle approach, a method was developed to add energy to assist human locomotion.

- The system adds a bounded amount of energy to create an oscillatory type of motion.

- The control method has been shown to work on linear and rotary mechanical systems

- A powered exoskeleton has been built and demonstrated using this new control method.

3.1   Phase Oscillator

A phase-based approach is beneficial because the phase controller produces a continuous forcing function that acts on a hybrid dynamic system. The phase approach generates a continuous signal even though the dynamics of gait continuously change because the forcing function does not change regardless of what a runner's legs are doing.

Phase plane analysis can be studied to determine if the system moves in a repeating or limit cycle (Graham 1971). Use of a non-linear forcing function based on the sine of the phase angle is beneficial because the sine function can never be greater or smaller than one or negative one; the function is bounded. A standard, second order,

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mechanical system equation is given: m represents the mass, b represents the damping, k

represents the stiffness.

$$m\ddot{x} + b\dot{x} + kx = 0 \tag{1}$$

Groups have added negative damping to the system to force the system to move, but the

negative damping can become unstable because $c\dot{x}$ grows as the velocity gets larger

(Ollinger, 2007).

$$m\ddot{x} + b\dot{x} + kx = c\dot{x} \tag{2}$$

It was decided to add energy to the mechanical system by using a "phase oscillator" by

using the sine of the phase angle, ϕ. In figure 3.1, the phase angle, ϕ, is defined.

$$m\ddot{x} + b\dot{x} + kx = c\sin(\emptyset) = \frac{c\dot{x}}{\sqrt{\dot{x}^2 + x^2}} \tag{3}$$

If c is positive, the system oscillates back and forth. The energy is always bounded

because the forcing function must be between plus or minus c, see figures 3.2 and 3.3.

If c is negative, the energy is damped, and the system state goes to zero, see figures 3.4

and 3.5.
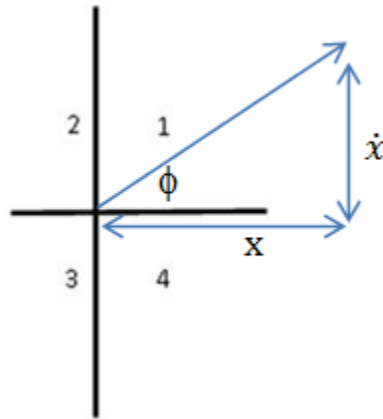
Figure 3.1. Phase Plot Definition; horizontal axis is the position, x, and the vertical axis
is the velocity, $\dot{x}$. The sine of the phase angle phi is given by the opposite over the
hypotenuse. The hypotenuse is given by $\sqrt{\dot{x}^2 + x^2}$
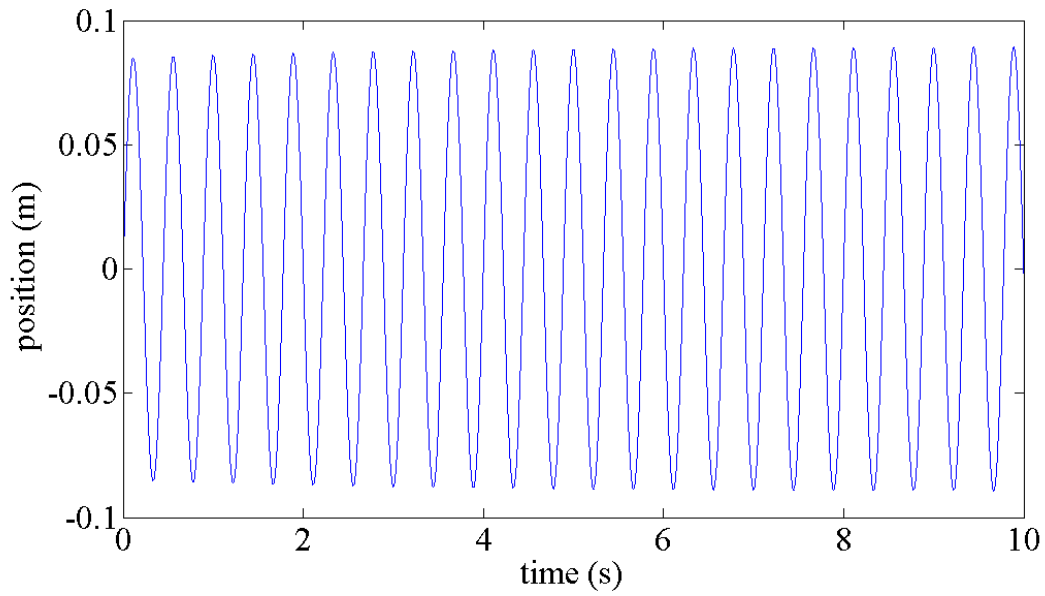
Figure 3.2. System Position Oscillations; position back and forth. m = 1 kg, k = 200 N/m, b = 0.6 Ns/m, c = 0.6. Initial velocity = 1.2 m/s, initial position = 0 m.
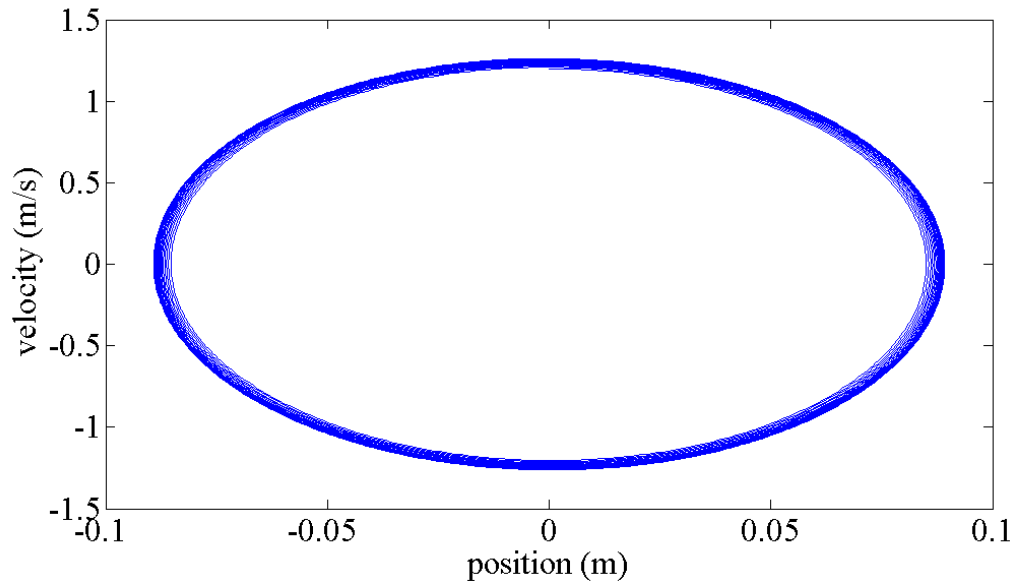
Figure 3.3. Phase Plot Limit Cycle; the phase plot demonstrates a limit cycle as the system position oscillates back and forth. m = 1 kg, k = 200 N/m, b = 0.6 Ns/m, c = 0.6; Initial velocity = 1.2 m/s, initial position = 0 m.
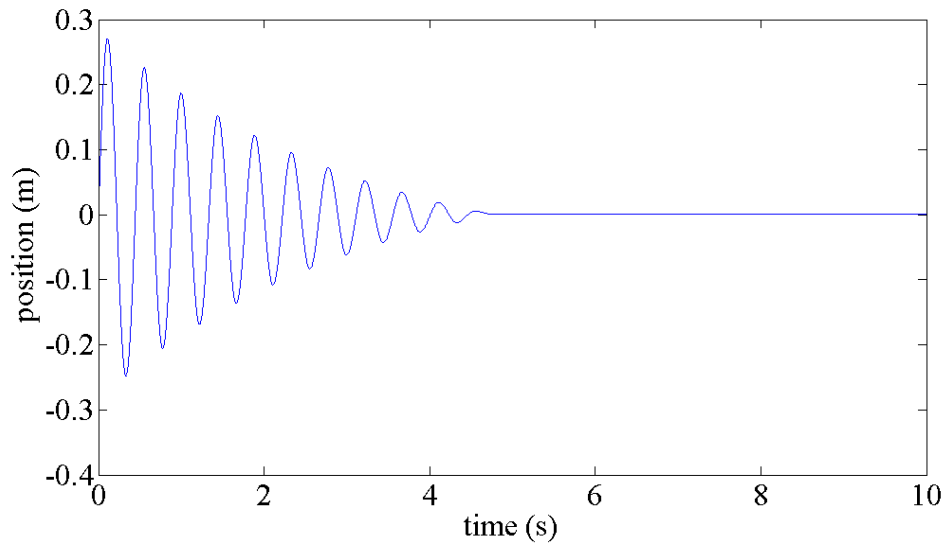
Figure 3.4. System Damps to Zero. m = 1 kg, k = 200 N/m; b = 0.6 Ns/m, c = -0.6. Initial velocity = 4 m/s, initial position = 0 m.
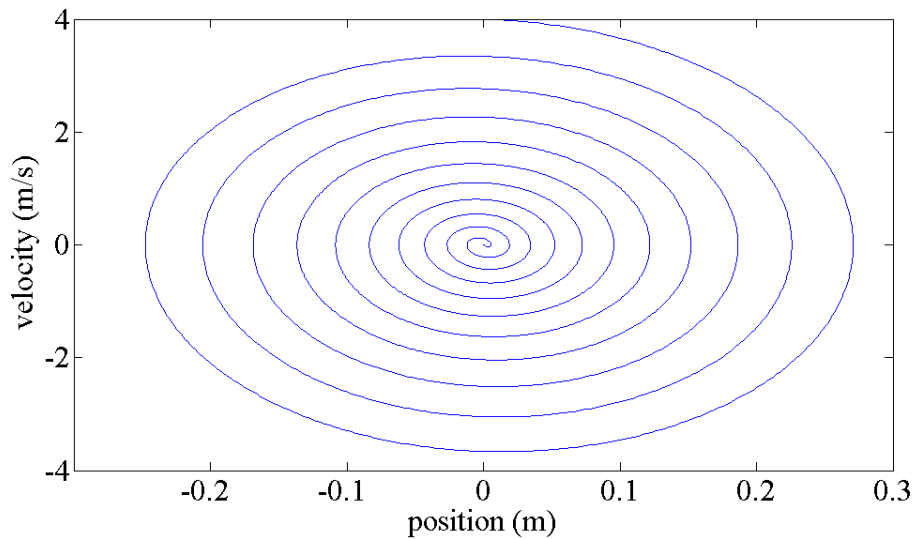
Figure 3.5. Phase Plot Position and Velocity Go to Zero; m = 1 kg, k = 200 N/m, b = 0.6 Ns/m, c = -0.6. Initial velocity = 4 m/s, initial position = 0 m.

The analysis can be repeated for a rotational model: I represents the inertia, b represents

the damping, k represents the rotational stiffness.

$$I\ddot{\theta} + b\dot{\theta} + k\theta = 0 \tag{4}$$

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Again, energy was added to the mechanical system by using a phase oscillator by using

the sine of the phase angle, phi. Other trigonometric functions were studied as well to

determine their effectiveness, but sine of the phase angle achieved the best results.

$$I\ddot{\theta} + b\dot{\theta} + k\theta = c\sin(\emptyset) = \frac{c\dot{\theta}}{\sqrt{\dot{\theta}^2+\theta^2}} \tag{5}$$

If c is positive, the system oscillates back and forth. The energy is always bounded

because as $\dot{\theta}$ gets bigger, in the limit the numerator and denominator cancel and just

equal c. If c is negative, the energy is damped out and the system state goes to zero.

3.2   Phase Controller

A testing device was designed where a small flywheel accelerates and decelerates,

creating inertial torque, causing a 1 meter pendulum with 3.4 kg at the end to oscillate

easily via parametric excitation similar to a small child pumping a swing, see figure 3.6.

The testing device demonstrates that a small torque can add energy to the relatively large

angular velocity of a pendulum. In an inverted pendulum, when c is set to negative, a

small inertia can accelerate and decelerate creating toques that dampen movement of the

pendulum. If the pendulum is swinging when the motor is turned on and c is negative, its

movements are quickly stopped. If the pendulum is placed in an inverted position, the

controller can keep the pendulum vertically upward like a well balanced inverted

pendulum.
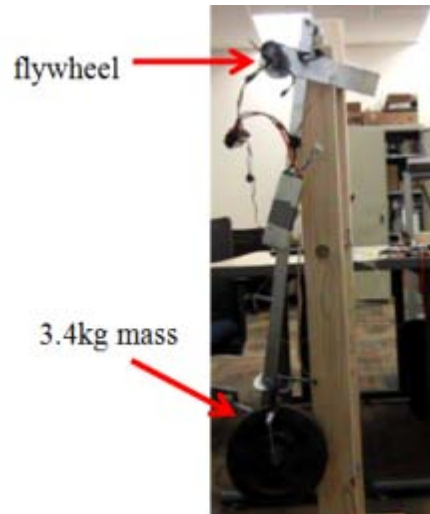
Figure 3.6. Inverted Pendulum Oscillates Similar to a Human Leg.

In a simulated example, a pendulum system with length 0.5 m, lumped mass of 1 kg, damping 0.007 Nm/(rad/s), and spring stiffness of 230 Nm/rad can be oscillated back and forth with a small excitation torque of 0.05 Nm. The limit cycle is robust and a wide range of initial conditions converge to one limit cycle defined by the constant c. The system oscillates at its natural frequency of 4.83 Hz, see figure 3.7.
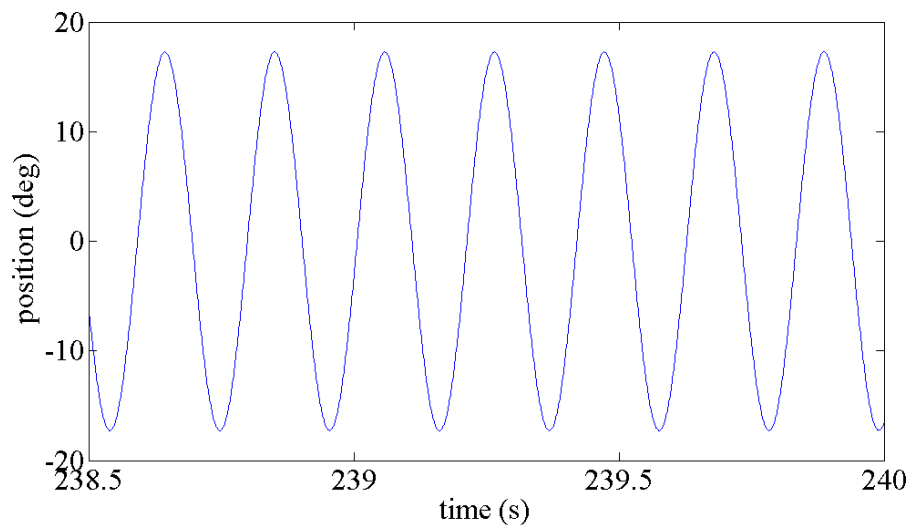
Figure 3.7. System Natural Frequency; I = 0.25 kgm^2, b = 0.007 Nm/(rad/s), k = 230 Nm/rad, c = 0.05, Initial velocity = (2/3)*pi rad/s, Initial position = (1/9)*pi rad.

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The damping of the system creates a torque to slow the system down while the torque created by the sine of the phase angle assists the movement of the system. Both signals are shown in figure 3.8, but the damping torque is shown with a positive sign instead of a negative sign to allow the signals to be plotted on top of each other.
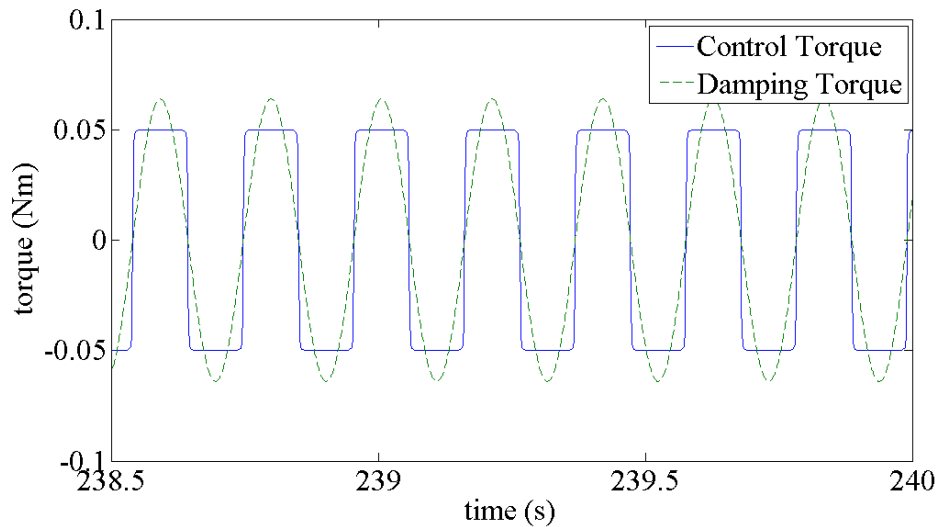
Figure 3.8. Damping Torque Oscillations; I = 0.25 kgm^2, b = 0.007 Nm/(rad/s), k = 230 Nm/rad, c = 0.05, Initial velocity = (2/3)*pi rad/s, Initial position = (1/9)*pi rad.

The power into the system created by the control torque and the power out of the system created by the damping torque are similar but do not exactly match, see figure 3.9. However, if the power curves are integrated over a cycle, the energy into and out of the system are equal and match. Even though the control power is quite low with peaks that are approximately 0.5 W, the power in the oscillating spring is quite high, 300W, see figure 3.10. Note, the sign of the damping torque was changed to positive instead of negative so that the power curves could be plotted on top of each other. The spring power is quite high because the system is oscillating quickly and the spring stiffness is large. The spring stiffness was modeled based on the stiffness at the hip joint.

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This control method is beneficial because a low power oscillator can pump energy into

the system and create large oscillations with high kinetic energy and high potential

energy in the spring. The oscillations are bounded and create a limit cycle. This method

is beneficial in developing wearable robots because small actuators (motors, pneumatic

cylinders, hydraulic cylinders etc.) can pump energy into the gait cycle or limb

movement.
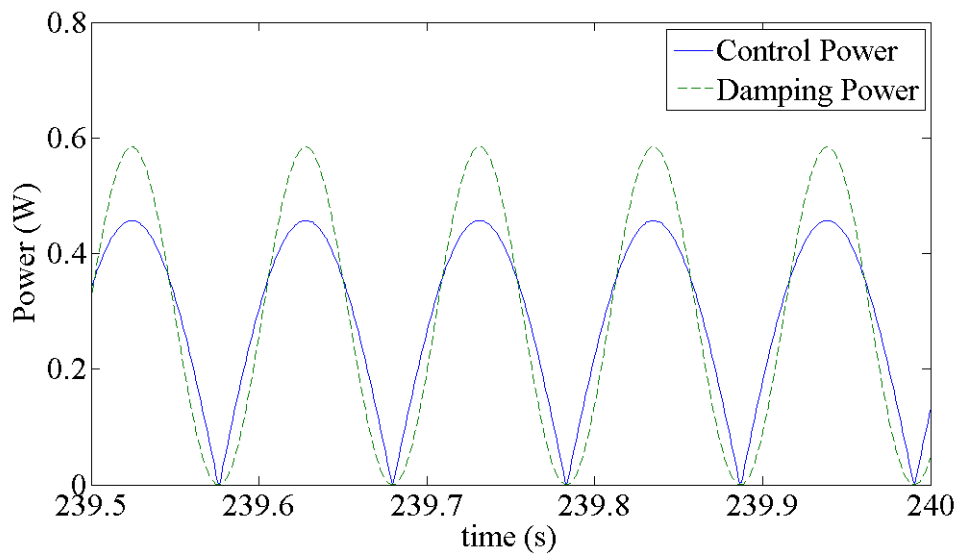
Figure 3.9. Inverted Damping Torque; the damping is shown positive instead of negative
so it can be compared easily with the control power. The powers are similar but not
equal. The energy for each curve over a cycle does match.
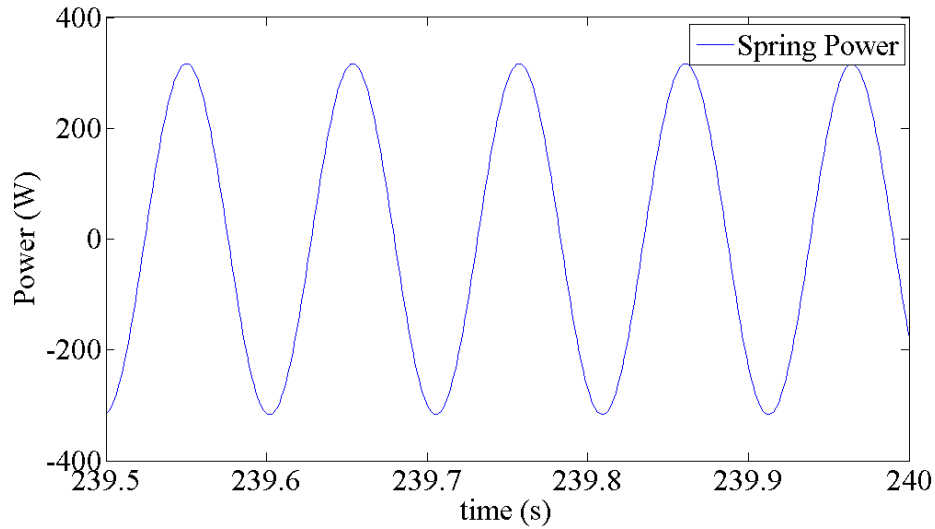
Figure 3.10. Spring torque Oscillations; the system is oscillating back and forth quickly and the spring torque is high resulting in large power oscillations in the spring.

Following the example of a torque applied to a pendulum structure based on the sine of the phase angle, torque is applied to help the legs oscillate while walking and running by applying a torque at the hips based on a phase oscillator. The leg of the human body can be assumed to be a pendulum-like structure with inertia, damping, and a spring stiffness (Blickhan, 1989). To enhance the pendulum motion, a parametric excitation torque can be added (Seyfarth, 2002). The direction of the torque must be switched at the correct timing and frequency and should be tuned with the frequency of gait. A phase oscillating term based on the phase portrait determines the desired torque to add positive power to the system.

AirLegs was designed and built to enhance walking and running using the phase angle approach. The control torque signal is used to trigger pneumatic valves to create torque at each hip joint which assists flexion and extension of the thigh. The pneumatic cylinder pulls and pushes on the thigh plate attached to the leg. The linear force of the cylinder creates an assistive torque at the hip joint.

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One rate gyro is placed on each thigh above the knee joint to determine the angular

velocity of each thigh.
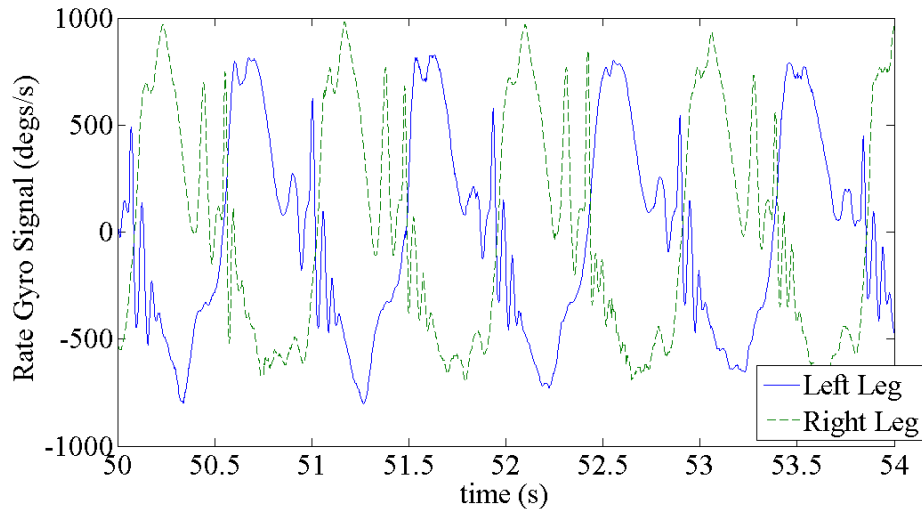
Figure 3.11. Rate Gyro Thigh Signal; a rate gyro is mounted on each thigh. The signals
are not exactly the same. The time duration and the pattern for each leg differ.

The signals for the left and right leg are not inversed and do not match exactly because

the stance phase is slightly longer than the swing phase. This results in variations in

duration for each phase of the gait cycle and the gait motion differs which requires a

sensor mounted on each leg instead of only one sensor driving both legs of the

exoskeleton. The signal from the rate gyros is then pseudo integrated to determine the

thigh angular position and the sine of the phase angle is determined using the angular

position and rate of the thigh.
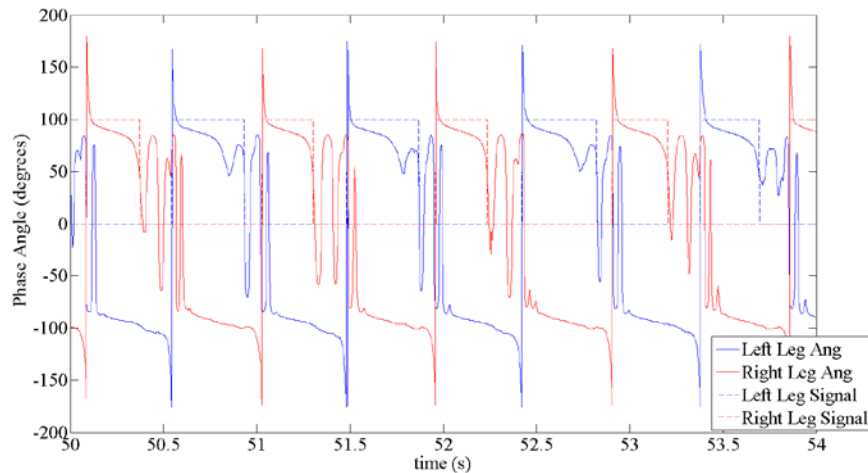
Figure 3.12. Phase Angle for Left and Right Leg; the signal is calculated 500 times per second in the microprocessor. The left and right control signals are shown in dashed lines. As the leg swings forward, the control signal turns high, 100. As the leg swings backward, the control signal turns low, 0.

Demonstration data with someone jogging on the treadmill is shown. The stance and swing phase angles are periodic but are not exactly the same. The time that the right leg swings forward is slightly larger than the time the same leg to swing backward. As the leg waits for heel contact, the phase angle oscillates up and down for a brief 200-300 ms.

The control signal in Figure 3.12 is used to trigger a two-position, four-way valve. When the control signal is high, the valve is energized and the thigh is pushed forward. When the control signal is low, the spring inside the valve moves it into the second position to pull the leg back. Even on a treadmill, the gait motion is not perfectly the same. The time duration between the high signals for the left and right leg slightly change. Each leg's movement is slightly different.

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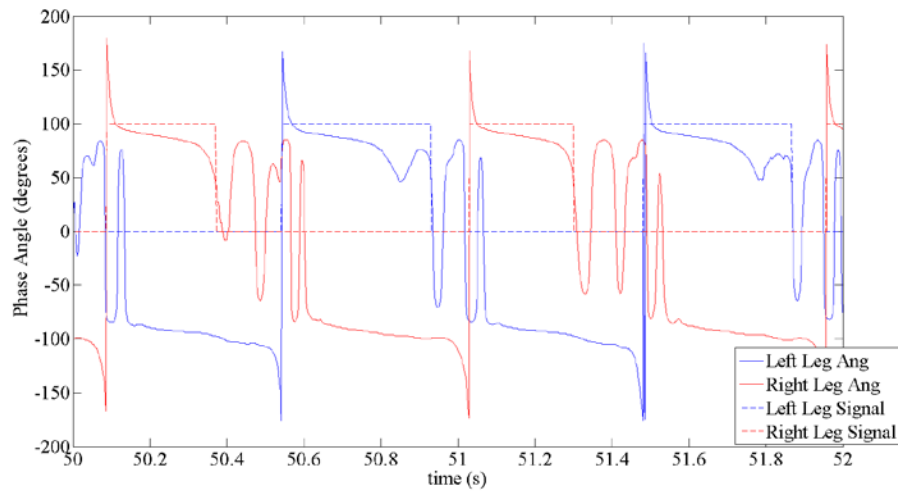
Figure 3.13. Phase Angle and Control Signal for Left and Right Leg; The left and right control signals are shown in dashed lines.

Using the phase controller, the exoskeleton can seamlessly assist walking, running, and can transition from walking to running and back to walking without any user input other than the user changing between walking and running. The seamless transitions are possible because a continuous control signal is generated and used as a triggering mechanism at 500Hz.

Implementation of the phase controller was accomplished using Matlab's Simulink environment and a DSPic programming add-on to program a microprocessor.

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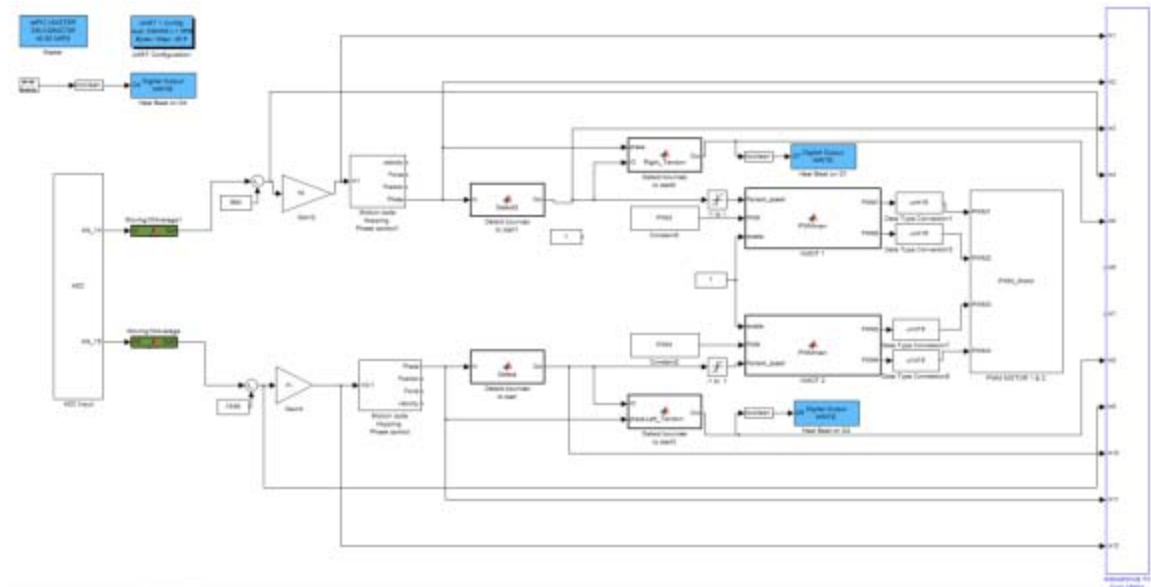
Figure 3.14. Phase Oscillator Control Software Written in Matlab

Figure 3.14 shows the basic logic of the control software with two analog inputs on the left hand side which are the sources of the rate gyros mounted on each leg. The analog values from the rate gyros are then manipulated to control the exoskeleton.

The microcontroller chosen to run the control software is Microchip's DSPic33FJ128MC710 which was available with Digilent's Cerebot MC7 breakout board. The Cerebot MC7 was selected because it comes standard with two H-bridge motor drivers which are used to control the pneumatic valves of the device.

Figure 3.15. A Digilent Cerebot MC7

CHAPTER 4

DESIGN OF RUNNING AUGMENTATION DEVICES

The design of a running augmentation device plays a critical role in the effectiveness of the device. Robotic systems that are overly complicated with many sensors and actuators can become cumbersome for the user, complicated for the developer, and more susceptible to failure because there are more components that can experience issues. The devices discussed in this thesis were designed focusing on four main objectives;

- The device should be safe for the user to wear, even in the event of a malfunction

- The device should utilize as few sensors and actuators as possible and the design should be minimalistic in nature to reduce weight

- The device should be autonomous, requiring no continuous input from the user except for their natural movement

- The device should be compatible with the average user and be capable of performing on different users with little to no modification

Many other researchers have taken a different approach to designing exoskeletons and assistive devices by designing very complex systems. Commonly, exoskeletons contain a plethora of sensors with the goal of having the robot interpret every possible scenario similar to Berkely's BLEEX (Chu, 2005). Many of these devices are also designed with a great deal of actuators to assist many unnecessary motions and further still, devices are often designed with focus on precise motion control. The primary problem with these designs is that they have components and system architecture that is

not beneficial for their purpose. Extra sensors make the software and control algorithms extremely complicated and designing a robot to interpret different scenarios becomes very difficult because there are countless new scenarios that the developer cannot foresee. Similarly, devices with a complex mechanical design consisting of many actuators are more likely to restrict natural movement and weigh more because they require more components. Devices that utilize precise motion control also increase in complexity and require additional sensors with feedback actuators.

4.1   AirLegs V1
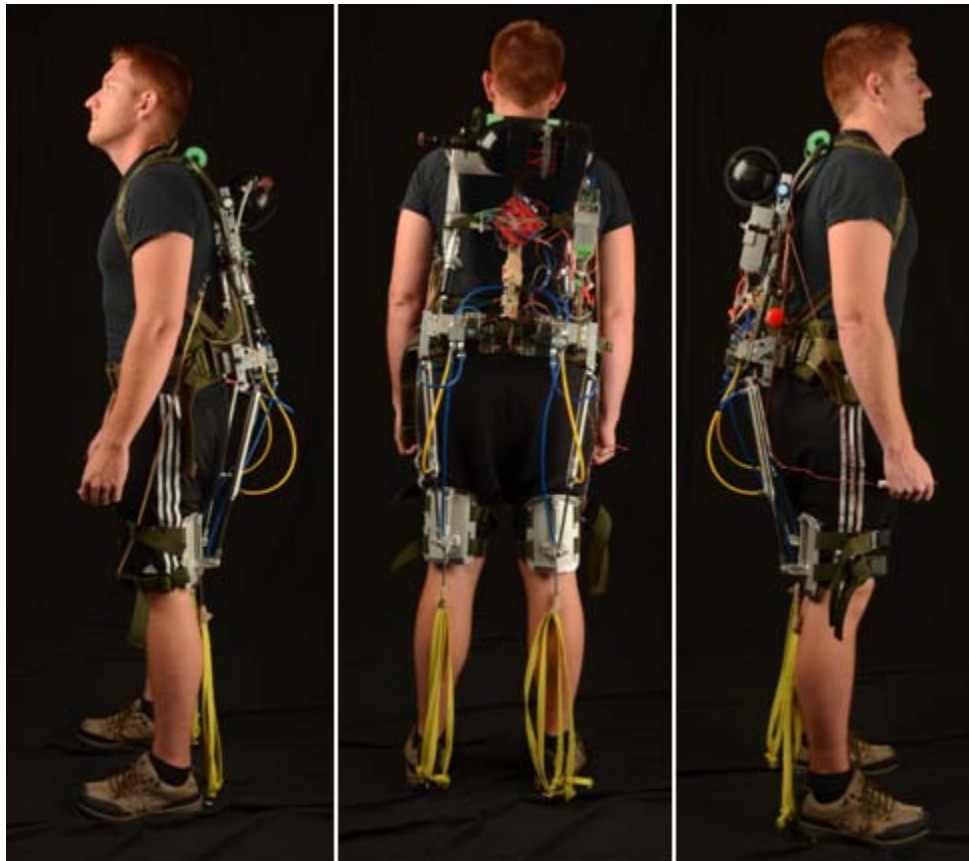
Figure 4.1. AirLegs V1 Left, Back, and Right Views; A pneumatic exoskeleton with actuators connecting directly to the user's thighs with artificial tendons connecting at the ankles.

In an effort to keep the running augmentation devices as simple as possible, the results of the experiments with the golf cart tow testing and OpenSim simulations were studied and AirLegs V1 was developed to only assist the muscles responsible for maintaining high speed running. This is accomplished by adding torque to the hip and ankle joints at precisely the right time.

The design of the AirLegs V1 exoskeleton was derived from studying the motion of the gait cycle and the trajectories of the legs while running. A novel hip joint was designed to allow flexibility in torso orientation. Often, when switching from walking to running, users change the angle of their torso relative to the ground. The novel hip joint allows for an extra partial degree of freedom by allowing approximately 15 degrees of rotation.

Figure 4.2. Hip Joint Assembly

The hip joint allows for a user to either walk upright or run slightly bent over which is more comfortable based on user feedback and video analysis of subjects transitioning from walking to running without the device.

Ankle support was also incorporated into the design of AirLegs V1 by creating an additional actuator mounted on the thigh plates. The purpose of assisting the rotational torque of the ankle was incorporated because the ankle is rotated naturally for push off by the soleus and gastrocnemius muscles which were discovered to generate most of the forces for push off. The additional small actuators mounted on the thigh plates are connected to a lever arm on the back of the user's shoes resulting in a bi-articular spring that is triggered similarly to the actuators at the hips. Figure 4.3 shows the small actuator of one leg connected to elastic cords which are then attached to a lever arm on the back of the heel.

Figure 4.3. Actuator Assembly Connected to Heel to Provide Torque at the Ankle

AirLegs V1 is powered by both battery and compressed air, using batteries to control the pneumatic valves while compressed air powers the cylinders. A pneumatic system was designed because preliminary analysis of motors available showed that a motor capable of producing the necessary torques was not only heavy, but had difficulty

in changing direction fast enough to keep up with the oscillations of the human leg. While there are motors capable of such speeds, they are heavy and draw a significant amount of power to overcome high inertias associated with oscillations.

A carbon fiber storage tank is mounted on the device and stores compressed air at 4,500 psi. The compressed air is then regulated down to 800psi from a regulator built into the tank. The air supply is then reduced again by an inline regulator to 90 psi which is fed to the cylinders actuating the artificial tendons adding torque to the ankles. Further reduction of pressure is accomplished by another inline actuator which reduces pressure to 40 psi and is fed to the hip actuators. By utilizing light weight aluminum pneumatic air cylinders, the total system weight is only 13.9 lbs.

FIGURE4.4. AirLegs V1 Gait Cycle; from left to right, top to bottom, the device pulls the left leg back during the stance phase and applies torque at the ankle during push off. The device then pushes the leg forward during the swing phase.

A significant benefit of mechanics incorporated in AirLegs V1 is that it pushes the user's leg forward during the swing phase. This helps the user to accelerate their leg forward in a faster manner which can be beneficial when running at high speeds to overcome the inertia of the leg after push off.

4.2   AirLegs V2
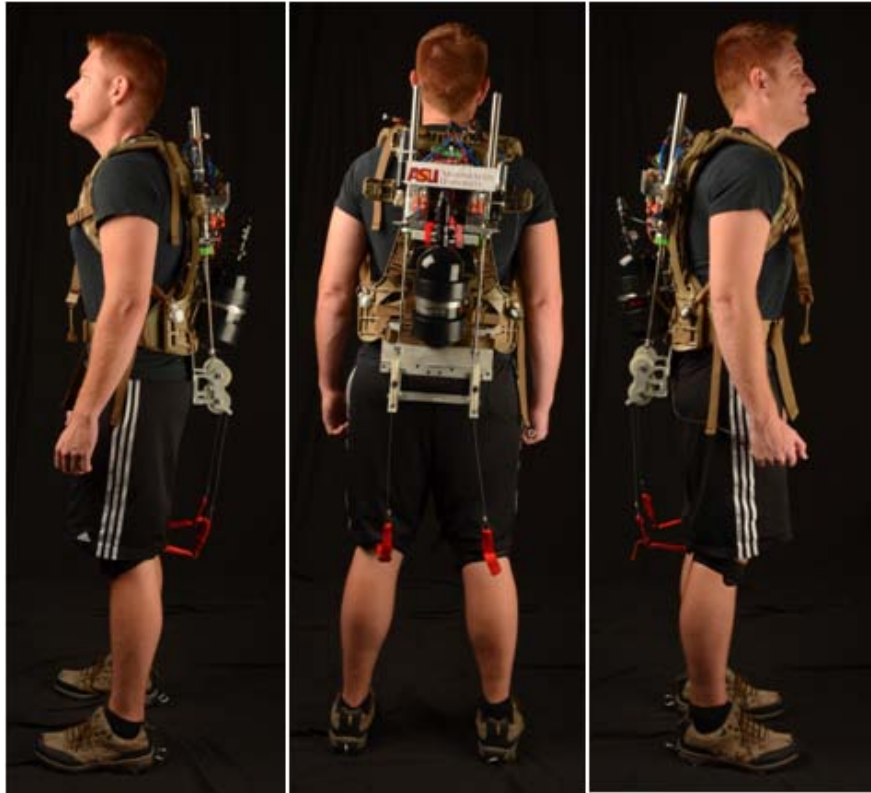
Figure 4.5. AirLegs V2 Left, Back, and Right Views.

AirLegs V2 is a soft suit version of AirLegs V1 that has been designed to greatly simplify the requirements of putting on or taking off the device. Although AirLegs V1 was successful in reducing metabolic cost, it was difficult to put on and could not be donned without the assistance of another individual. AirLegs V2, on the other hand, is simple for a user to don and doff without additional assistance from others. AirLegs V2 is made of a pneumatic system which controls two vertically mounted linear pneumatic

cylinders attached behind the shoulder blades which are attached to a cable and pulley system mounted on the user's back. The cables are then routed through a pulley system to create a torque at the hips similar to that of AirLegs V1 but attach to two easily removable Velcro knee braces. Limitations of the cable design result in forces only pulling at the back of the thighs because a pushing force cannot be made from flexible cable. The torque is applied to the hips during the stance phase increasing the ground reaction forces created during push off.

A carbon fiber storage tank is mounted on the device and stores compressed air at 4,500 psi. The compressed air is then regulated down to 800psi from a regulator built into the tank. The air supply is then reduced again by an inline regulator to 40 psi which is fed to the cylinders actuating the cables that pull on the knee braces. Compared to AirLegs V1, AirLegs V2 offers a greater range of motion, reduced overall weight, and it easy to don and doff. AirLegs V2 is also much simpler to adjust for other users and adjustment from one user to the next only requires lengthening or shortening of the connecting straps that connect to the knee braces. The straps connected to the pulley system are then attached to the knee braces by means of a plastic connector that can be clipped or unclipped using one hand.

The soft suit design uses the same phase controller with only slight modifications because it does not include ankle support. The phase controller allows for seamless transition from walking to running and back to walking without any user input. AirLegs V2 is also lighter than AirLegs V1 by 2.3lbs weighing in at 11.6lbs

4.3   Jet Pack
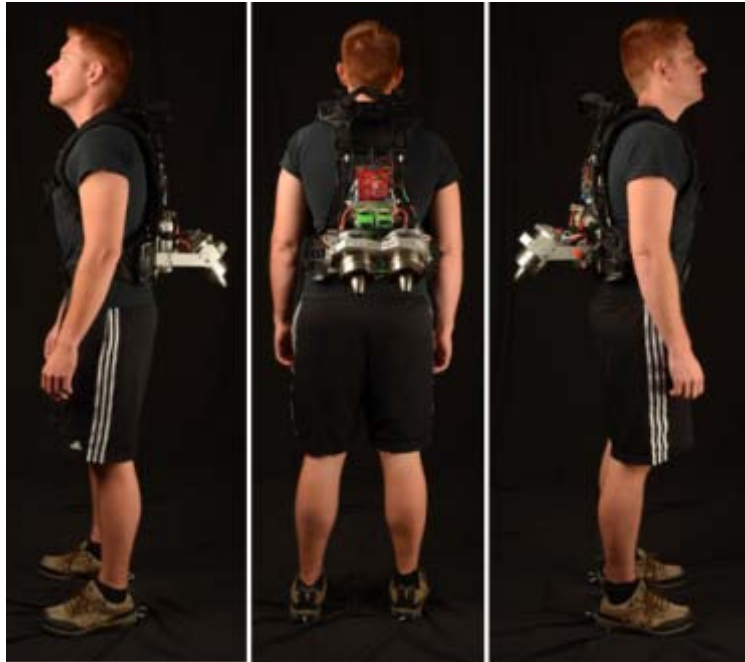
Figure 4.6. Jet Pack Left, Back, and Right Views; the prototype is worn like a common backpack and exhibits a forward and upward external force on the user's torso.

The Jet Pack system was designed as an alternative method to enhance running gait and utilizes two statically mounted high speed ducted fans. Benefits of the Jet Pack when compared to the other devices includes; no adjustment necessary when switching between users and no limitations on size of the potential user. Because there are no components that attach to a users joints or limbs, no adjustments are necessary.

The Jet Pack's thrust is created by two ducted fans made of high speed brushless motors (67,500 RPM) with 12 blades each. The ducted fans operate on 25 volt, 6 cell lithium polymer batteries. Each brushless motor has its own battery and draws approximately 100 amps from a brushless speed controller at peak thrust levels, resulting in roughly 4 minutes of continuous use before depleting the battery to 22 volts. The amount of thrust can be controlled by the user and the total run time increases if less than

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100% thrust is used. The fans are mounted onto a military pack frame at an angle so that the forces they produce are near 25 degrees below and behind the runner. By exerting force which pushes the runner from below and behind the runner, forces similar to those induced by the golf cart testing rig are recreated in a small light weight package with no mechanical interaction between the device and user's legs.

Analysis of user height determined that while different users may be taller or shorter, the most effective positioning of the fans created a force axially in line through the runner's center of mass. In other words, the thrust vector needed to be in line with the user's center of mass, on average just below their belly button. If the thrust force was too high on the user's back, they felt as if they were going to tip forward. After lowering the fans to a position which was closer to the center of mass of one user it was found that they no longer had this feeling. To explore the positioning of the fans further, multiple users each put on the device and found even though they all were of different height, the relative offset of the fans to their center of mass was negligible between users. Because of this relationship, the need for having the user control the height of the fans on their back while running was not significant enough to add that feature into the controller and it could be left as a static adjustment.

Analysis of the thrust angle was also conducted to determine if a controller should include the ability to adjust the angle of the fans while running. It was found that the thrust angle relative to horizontal was not only dependent on the angle between the fans and the pack frame, but also the angle of the user's torso while running. For example, a user that runs with their body more tipped forward would yield a higher forward thrust

than vertical thrust. Inversely, a user that ran with their torso angle closer to vertical would produce a greater vertical thrust than horizontal.

Figure 4.7. Jet Pack Back Pack Prototype

The Jet Pack system also incorporates a hand held controller, see figure 3.7, allowing the runner to dynamically change the magnitude of thrust produced. This allows the runner to "tune" the Jet Pack so the force produced is closer to their 10% body weight and feels good to the runner. The hand held controller allows the runner the ability to not only control the amount of thrust produced, but also control on/off operation of the ducted fans.

Figure 4.8. Jet Pack Hand Held Controller

In order for the hand controller to be capable of trust modulation, an onboard microcontroller was programmed to handle user input. The microcontroller was programed in C code generated by Matlab Simulink software that utilizes a visual interface for programming similar to the methods for programming AirLegs. The microcontroller operates by creating a pulse width digital modulated signal that can then be interpreted by the onboard motor controllers. When the processor is initially turned on, the thrust setting is at 50%. When the user squeezes the trigger, the fans turn on to the desired speed. By pressing up or down on the controller hat, the thrust setting is increased or decreased, producing more or less thrust.
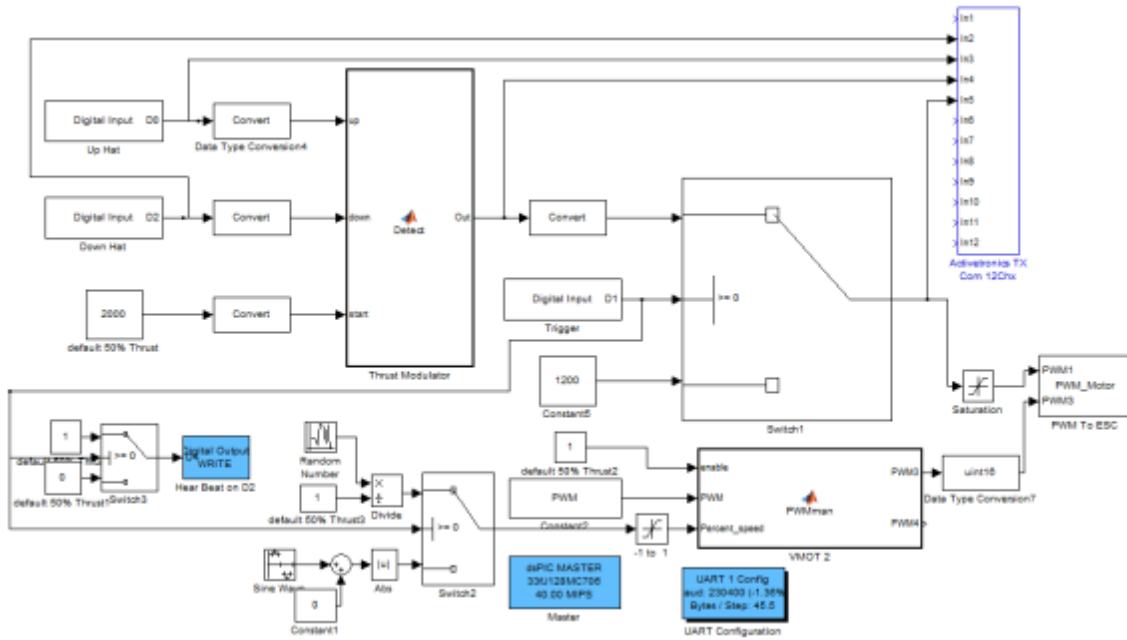
Figure 4.9. Matlab Simulink Control Software for Jet Pack

The software runs on a Digilent Cerebot MC7 board with a DSPic33FJ128MC710

microcontroller. The brushless motor controllers are then sent a PWM signal at 50Hz to

control the speed of the motors.

Chapter 5

DEVICE EXPERIMENTATION AND ANALYSIS

In chapter 5, the experimental process and data analysis are discussed for each device. Due to the complexity of IRB protocol, not all devices have been approved for human subject testing at the time of this document and have only been demoed at this time. Testing has been concluded for the Jet Pack and AirLegs V1 has been tested by the Army Research Laboratories.

5.1   Army Research Laboratory Third Party Testing of AirLegs V1

At the Army Research Laboratory, AirLegs V1 was tested on two subjects with hip actuators only. The users were strapped into the device and wore a chest pack metabolic unit after placing reflective markers on reference points on their body. A motion tracking system was used to record marker data and an Opti-gate analysis system tracked their gait. A treadmill with force plates inside was used to record ground reaction forces. The first test subject to wear AirLegs V1, a 5'1" female weighing 105lbs, did not experience optimal results because she was much smaller than anticipated and the pneumatic cylinders were much too large. At slower speeds, the metabolic cost was neutral, but at higher speeds, the metabolic cost increased. The second subject, a tall male, experienced runner, demonstrated metabolic savings shown at both 6mph and 8mph, see Table 5.1. The user had reduced metabolic cost comparing device powered on versus no device at all. Testing was performed at ARL on September 26-27, 2013. All testing was performed under a protocol approved by an institutional review board supplied by the Army Research Laboratories.

Table 5.1

Metabolic Cost When Running With Powered Hips on a Treadmill at ARL

Test	Metabolic Cost (ml/min/kg)
Subject 1, Small Female (mass = 59.1 kg, ht = 162 cm)	
1. Running at 5.5 mph on a treadmill, No Device	26.52 (3.7)
2. Running at 5.5 mph on a treadmill with Powered hip system ON	26.48 (4.0) (-0.2%, no difference)
3. Running at 6.5 mph on a treadmill, No Device	29.4 (2.7)
4. Running at 6.5 mph on a treadmill with Powered hip system ON	31.3 (3.5) (6.7% increase)
Subject 1, Tall Male (mass = 66.1 kg, ht = 181.8 cm)	
5. Running at 6 mph on a treadmill, No Device	31.6 (2.8)
6. Running at 6 mph on a treadmill with Powered hip system ON	**29.1 (3.4) (8.0% savings)**
7. Running at 8 mph on a treadmill, No Device	40.8 (2.4)
8. Running at 8 mph on a treadmill with Powered hip system ON	**36.6 (1.7) (10.2% savings)**

AirLegs V1 has also been demonstrated on other occasions over-ground and the device can successfully transition between walking and running gaits as well as irregular stepping. The device has also successfully demonstrated assistance with stair climbing and uneven terrain. Currently, the device can assist a user to travel half a mile on one tank of compressed air.

5.2   AirLegs V2 Testing

Initial pilot demonstrations of AirLegs V2 have shown similar metabolic reductions to V1, however, a detailed study has not yet been completed as the device is

pending IRB approval. Demonstrations of the device have shown that AirLegs V2 can successfully transition between walking and running gaits, irregular stepping, and assist while climbing stairs. The device also exhibits significant improvement with regards to range of motion for the user when compared to AirLegs V1. The user can perform lateral movements, hip abductions and a variety of different stances such as kneeling, bending, and turning thanks to its soft suit cable design. In a demonstration in which a user attempted to climb stairs repeatedly, the device was capable of assisting the user to climb over 50 flights of stairs on one tank of compressed air. Although no data was recorded, the user was capable of demonstrating the device without any signs of fatigue, stress, or discomfort which they could not do naturally.
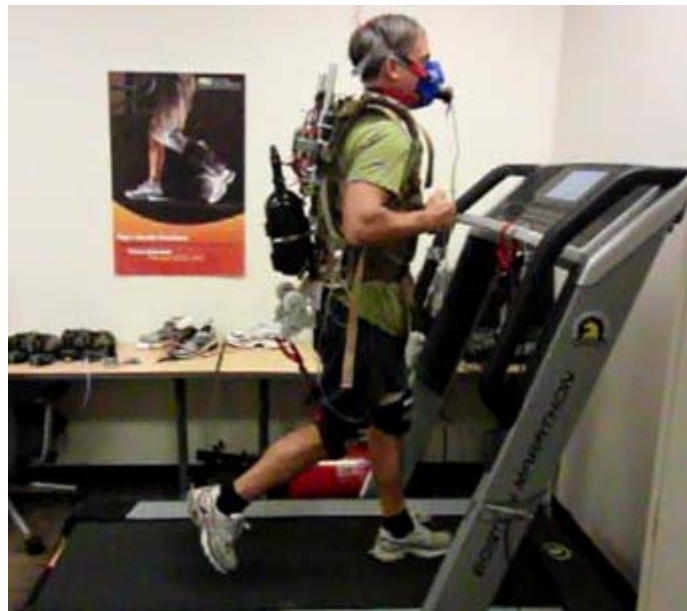
Figure 5.1. AirLegs V2 Metabolic Demonstration

AirLegs V2 has been demonstrated on a treadmill as well as over ground. Treadmill demonstrations indicate that the device significantly helps with muscle fatigue and has received feedback as feeling "great"

5.3   Jet Pack Testing

Figure 5.2. Subject Testing Jet Pack on 1 Mile Course.

Upon receiving IRB approval for human testing (Appendix A), tests were performed on one subject. Preliminary measurements of the user's heart rate, speed, distance, and time were recorded and observed to show significant improvements in speed and reduced heart rate.

Testing included 4 timed trials, two trials of the subject sprinting a 200 meter course, and two with the subject running for 1 mile. The goal of the testing was to determine if the subject could complete the specified courses in a faster time using the device and if their heart rate would change based on using the device or not.

Testing for the 200 meter course proceeded with the subject attempting to run 200 meters without any assistive device first, then after resting for a period of 30 minutes, the

subject ran the same 200 meters with the device. The primary test demonstrated a

decrease in time of 3 seconds required to run the 200 meter course. The Subject

completed the first 200 meter trial in 28 seconds with no device and completed the same

course in 25 seconds with the device.

Secondary testing, when the subject ran one mile, demonstrated consistent

findings with decreases in time and heart rate as well, see figures 4.3 and 4.4.
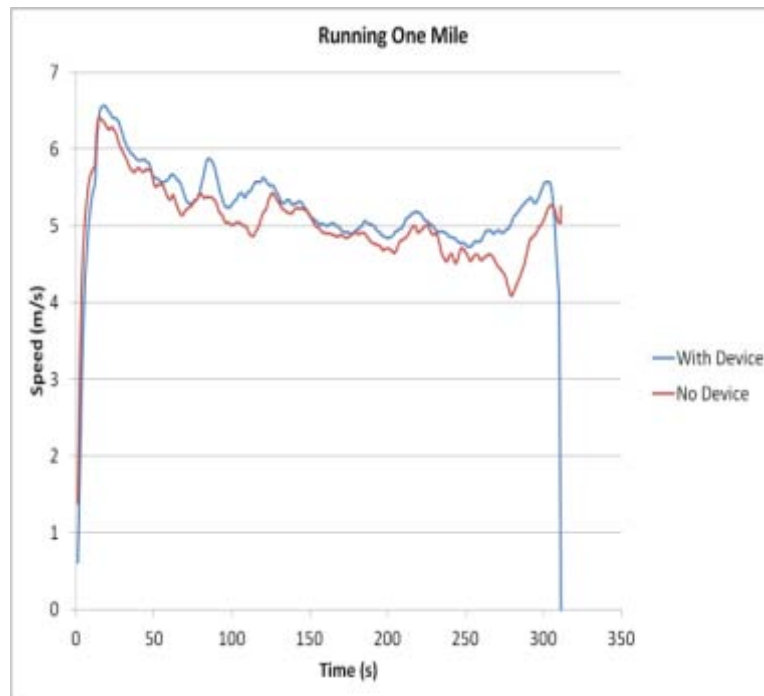
Figure5.3. Speed on 1 Mile Test With and Without Jet Pack

Testing with the device not only decreased the subject's time to complete one mile

by 18 seconds, but lowered their heart rate as well. The time to run one mile without the

device was 5 minutes and 20 seconds. The user became fatigued towards the end of the

first mile and demonstrated relatively high heart rate. After resting one hour, the same

subject attempted to run the one mile course again only this time with the Jet Pack. Even

though the user was carrying additional weight (Jet Pack and helmet weigh 12lbs or

5.4kg), they were capable of completing the same course and distance with a time of 5

minutes and 2 seconds. These findings show that even when carrying additional weight, a

runner can sustain higher speeds for longer periods of time when utilizing the Jet Pack as

shown by an increase in average and peak speeds. A reduction in metabolic cost can be

assumed based on a decrease in average and peak heart rates even with subject carrying

an additional 12lbs and running faster.
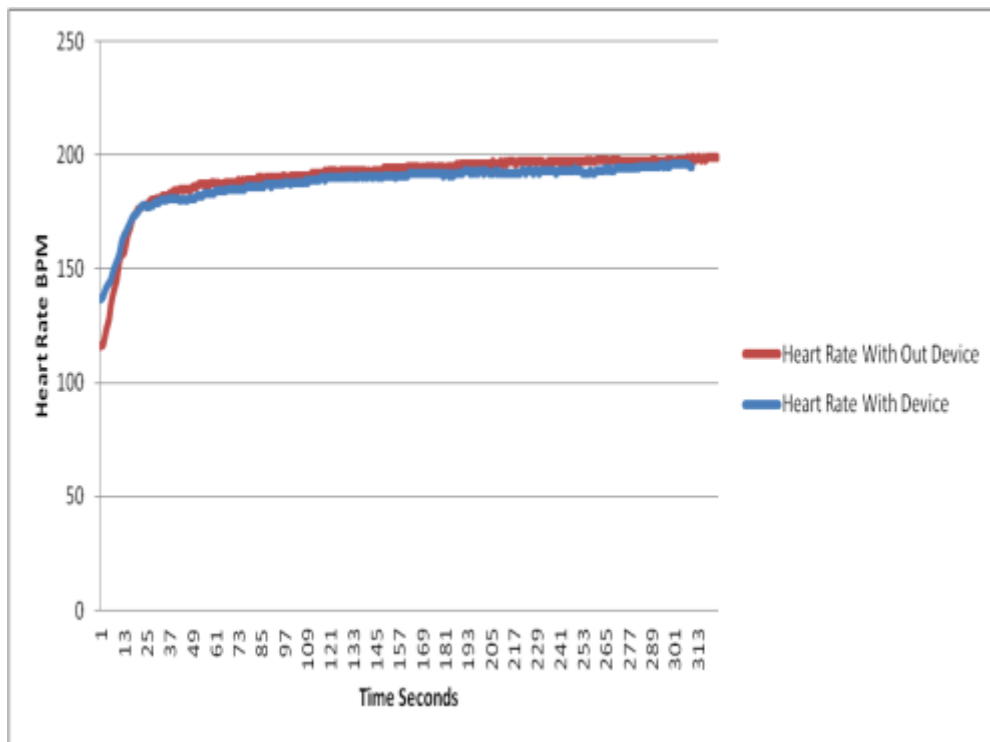
Figure 5.4. Heart Rate on 1 Mile Test With and Without Jet Pack

Table 5.2

Comparison of Speed and Heat Rate With and Without Jet Pack During 1 Mile Test

No Device		
Average Speed	5.06	m/s
Maximum Speed	6.41	m/s
Average Heart Rate	193	Bpm
Maximum Heart Rate	199	Bpm
With Device		
Average Speed	5.22	m/s
Maximum Speed	6.57	m/s
Average Heart Rate	187	Bpm
Maximum Heart Rate	196	Bpm

Data from the table above results in a 2.5% increase in maximum speeds, 3.1%
increase in average speed and a 5.6% decrease in time over one mile and a 3.1% decrease
in average heart rate. This essentially demonstrates, without the use of VO2 data, that the
Jet Pack makes it easier to run because the user's heart rate is actually lower on average,
and with lower peaks.

CHAPTER 6

CONCLUSIONS

Prior research has demonstrated that the metabolic cost of running could be
decreased with a forward propulsive force. Kram's research of applying pulling forces
on a treadmill was also experimentally tested and validated in situations where subjects
ran over-ground and not on a treadmill. The following concepts were successfully
implemented into the designs.

- An exoskeleton device consisting of pneumatic actuators connected to the legs and ankles allows for running at high speeds while reducing the metabolic cost of running.

- A soft-suit exoskeleton using cables connected to pneumatic actuators at the users back and knees will assist human locomotion and reduce metabolic cost for running while allowing for unrestricted motion.

- A propulsive device at the users back will assist human locomotion and reduce the metabolic cost for running

Three devices were designed and fabricated to successfully augment running by either adding a pure external force or by transferring mechanical energy through the legs of a user. In all cases, the devices demonstrated that successful augmentation is possible if the devices do not hinder the user with over complicated actuators and mechanisms. They also demonstrate the need for a robust and precise controller such as one designed around the phase oscillator. Furthermore, machines that are successful at augmenting human locomotion help only the muscle groups that expend the most energy and do not try to help all possible movement.

Over all, each device has different characteristics which result in different performance abilities. AirLegs V1 can not only help reduce the metabolic cost of high speed running, but it can help users push off the ground with even greater force by using an artificial tendon. AirLegs V2 allows users to gain assistance with the stance phase and push off while also offering a very flexible, unrestrictive method of attaching to the human. Also, The Jet Pack is capable of assisting nearly any user without modification or reconfiguration because of its simplicity. Furthermore, this research thesis

demonstrated that robotic assistive devices can have off the shelf components, with few sensors and still operate as intended. This is a very important aspect because it shows the potential for assistive devices to become universal, assisting nearly anyone. If research is continued in a manner such as this thesis, focusing on minimalistic designs that are efficient and precise, the industry of wearable robotics and robotic assistive devices will soon be available to the masses.

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APPENDIX A

JETPACK IRB PROTOCOL AND CONSENT DOCUMENTATION

IRB Submission Protocol
Wearable Robot for Running Augmentation

1. Background
The purpose of this device is to augment a runner's ability to run at high speeds.

2. Objectives
The objective of the proposed investigation is to collect data (heart rate, and running speeds) and video of a person running with and without the robot. The purpose of the robot is to decrease the metabolic cost and increase overall performance of an able-bodied runner by means of propulsion from ducted fans. The energy source for the device is two lithium polymer batteries that are self-contained within the device and power the ducted fans. The primary purpose of this pilot study is to determine the effectiveness of the device.

Device weighs 11 lbs
Noise level 84 dB
Electric Brushless DC motors
Robot is mounted to a pack frame
Emergency switch mounted at the waist of the back pack frame to turn off the system

Figure 1 - Device

3. Study Design
The proposed study will involve 4 tests on each subject. The purpose of these tests will be to monitor the heart rate and the time it takes to run set distances with and without the robot. The test subject will be monitored running on a 200 meter track and a 1 mile track. The test subject will also be recorded while running to gain information on their stride length, gait cycle, and flight time.

All trials will be conducted on Arizona State University at the Polytechnic Campus within the robotics research lab and the campus sports track located outside of the Student Recreational Center.

4. Study Population

The test subjects that will be used in the proposed investigation will be comprised of individuals that are capable of performing the tests, running at speeds greater than 10 mph without assistance, and are in good health.

5. Recruitment Methods

Test subjects will be recruited from within Arizona State University's student body, active members of athletic programs, and others who meet the physical requirements. Test subjects will be evaluated based upon their physical abilities and the information they provide to the research staff.

As part of the screening process, test subjects will be required to sign a consent form indicating that they have been fully briefed on all of the possible physical dangers associated with the testing and the entire testing protocol.

6. Test Subject Inclusion/Exclusion Criteria

INCLUSION CRITERIA
1. Between the ages of 18-45 years of age
2. Can run the distance of 1 mile within 6 minutes
3. Full range of limbs including arms, knees, legs, and hips.
5. Ability to sprint 200 meters.
6. Ability to wear test equipment and heart rate monitors.
7. Ability to follow simple instructions.
8. Must have proof of medical insurance
9. Must answer no to all of the questions in the Par-Q questionnaire

EXCLUSION CRITERIA
1. Excessive body mass or lack of stamina
2. Restricted joint movement in arms and legs.
4. Inability to wear test device.
5. Inability to meet inclusion criteria.
6. Any past medical problems of heart disease

Four to six individuals will be screened for eligibility

Four participants will be enrolled in the study. It is a pilot study to determine if wearing the device versus not wearing the device increases running speed and reduces heart rate.

The participants in the proposed study will be comprised of individuals that are capable of performing the tests, running at speeds greater than 10 mph without assistance, and are in good health.

7. Roles of investigators and research procedures

Dr. Thomas Sugar

Dr. Thomas Sugar is the overseeing mentor on the project and is in charge of verifying all documents and subject information is handled appropriately according to IRB protocol. He may assist in collecting data and or analyzing data. He is the advisor on the project to the students.

Dr. Sugar will obtain informed consent from each subject and will have the subjects sign the consent form.

Jason Kerestes

Mr. Jason Kerestes is a MS student in engineering with expertise in robotic systems and fabrication. He will provide technical advice, services, attend training sessions, recruit, and participate in administering tests and collecting data, and analyzing de-identified data.

Testing at Arizona State University

Before any physical tests are conducted, the subject will be provided an opportunity to gain familiarity with the device. Basic training on the operation of the device and user controls will be administered by one of the research team. The subject will be allowed to gain familiarity with the device while jogging or running at their own set pace. This training session will include step-by-step instructions covering the basic operation of the device; thrust control, device shut down protocol and emergency power-down procedures. Any time the subjects wear the device they will be required to wear all safety equipment provided.

Test Subjects must provide and wear the following safety items:
1.   Close toed running shoes.
2.   Flexible running clothing such as running pants/shorts and shirt.

Test Subjects must wear the following safety items provided by the research team:
1.   Full Face helmet.
2.   Single use foam ear plugs

Figure 2 - Full Face Helmet

Figure 3 - Single Use Foam Ear Plugs

Testing will be conducted over 4 days, one day for each test subject. The test duration will last approximately 5 hours.

Device training will consist of a 10 minute information session in the robotics lab on how the device works, what to expect, and how to use the controls. After this period, the subject will be taken to the running track where they will have the opportunity to wear the device with protective equipment and become familiar with how it feels. They will be allowed to jog and run at their own pace while wearing the device for approximately 5-10 minutes.

After the subject has become comfortable with the device, they will rest for approximately 30 minutes to 1 hour while the device batteries are recharged to full capacity.

During testing, the subject will run over flat ground, either on an athletic running track or over a predetermined smooth dirt or grass path. Before operating the device on the track or path, the user will first attempt to run at an average speed of 10 mph for a short distance to warm up, then they will be asked to stretch to help prevent strained muscles.

There will be 4 tests administered per subject; 2 tests running 200 meters (with and without the device) and 2 tests running 1 mile (with and without the device).

- Test subjects 1 and 3 will first perform the 200 meter test without the device, rest for an additional 30 minutes, then perform the 200 meter test with the device.
- Test subjects 2 and 4 will first perform the 200 meter test with the device, rest for an additional 30 minutes, then perform the 200 meter test without the device.
- There will be a 1 hour resting session between the 200 meter test and 1 mile test.
- Test subjects 1 and 3 will then perform the 1 mile test with the device, rest for 30 minutes, then perform the 1 mile test without the device.
- Test subjects 2 and 4 will then perform the 1 mile test without the device, rest for 30 minutes, then perform the 1 mile test with the device.

While performing the tests, the subjects will be monitored for their heart rate via a wireless Bluetooth heart rate wrist watch. They will also pass through a section of track which is in the view of a high speed camera. The high speed camera will be used to capture their running stride, gate frequency, and stride length.

Figure 4 – 200 Meter Course

8. Parameters Measured
The test parameters that will be collected will be
1) The test subjects run time.
2) The test subjects heart rate.
3) Video footage of their stride length, gait cycle, and flight time.

9. Data Analysis
The focus of this test protocol will be to evaluate potential benefits of the device concerning the subjects running ability. Tests performed in which the subject's duration of the test are unchanged or similar will show either a decrease or increase in metabolic cost based on their

heart rate. Tests performed in which the subject's duration of the test are vastly different will show either a increase or decrease in their maximum running speed.

10. Benefits
The benefit of the device is similar to the project objective, that the device shall improve an individual's physical performance in the area of running. The primary goal of the device is to either increase the runners speed up to 15 mph, or decrease their metabolic cost so they are able to run longer distances that previously capable.

There will be no personal or long term benefit to the test subject outside of the testing conditions.

11. Risks and Risk Management
There are certain risks and discomforts that may be associated with this research. They include but are not limited to:
1. Physical risks associated with running such as tripping and falling either due to the runner's ability or lack thereof, or the device adding additional force to the subject.
2. Physical risks associated with running at high speeds such as increased impact forces at the user's joints, (knees, ankles, hips)
3. Physical risks associated with intense exercise such as strained or pulled muscles.
4. Hearing risk associated from high noise levels produced by the ducted fans may have a short term effect on the runner's hearing.
5. Fatigue or increased stress levels associated with elevated cardiovascular exercise.
6. Running at increased speeds could reveal underlying heart disease
7. There is a chance for the two batteries to overheat and catch fire
8. There is a chance of tripping and falling and breaking a bone

To minimize potential risks, you will be required to wear and utilize a protective full face helmet. Should you fall while running, the helmet will protect your face and head. You will also be required to wear foam ear plugs to prevent any hearing damage. The current device has a noise level of 84dB which is below the threshold for 8 hours of use.

The batteries will be mounted in an aluminum box to minimize the risk of batteries catching fire and containing a fire to protect the user. Fire-proof material will be mounted behind the battery box to minimize burns.

Before each test you will be asked to stretch to reduce the possibility of strained or pulled muscles. Rest periods will be provided between tests and as needed to alleviate fatigue and stress, however you may remain fatigued during or after the testing procedure. If any part of the test does not go accordingly, the device will utilize a remote emergency switch, allowing the research staff to remotely and instantaneously disable the device. The user will also be able to push an emergency switch on the backpack to turn off the system at any time.

You will be asked to wear a heart rate monitor. If your heart rate goes above the maximum level, the heart rate monitor will beep and you will be asked to stop.

From CDC, "*This maximum* rate is based on the person's age. An estimate of a person's maximum age-related heart rate can be obtained by subtracting the person's age from 220. For example, for a 50-year-old person, the estimated maximum age-related heart rate would be calculated as 220 - 50 years = 170 beats per minute (bpm)."

Participation in this study may cause all or some of the side effects listed above. In addition, there is always the risk of developing previously unknown side effects.

If mild side effects or discomforts do occur, the research staff will attempt to minimize these effects by asking you to rest. If any serious side effects occur, the research staff will attempt to minimize these by directing you to the emergency room.

If there are any major side effects, the research staff will call 911 immediately.

The investigator is willing to discuss any questions you might have about these risks and discomforts.

12. Incentives and research related costs
All tests and procedures are research related. There will be no fee to test subjects wishing to participate, however in the event of side effects the subjects alone will be responsible for any medical expenses. Test procedures will have no monetary benefit to the test subject.

13. Alternatives
Taking part in this research study is strictly voluntary. A subject may choose not to take part in this research study or may withdraw their consent at any time. The commitment of the research staff is to provide help and assistance and will not at any time be affected by an individual's choice not to participate. If subjects decide to end their participation in the study, they will not complete any additional testing nor will they be asked to continue testing. There will be no penalty or loss of benefits if the subjects decide to end testing prematurely.

14. Research Materials, records, and confidentiality
All data, videos, and photographs will be modified as to not contain any information relating to the subjects identity. All data, video, and photographs and will be kept in a locked storage cabinet and all unused data, video, and photographs will be destroyed two years after testing is concluded. It is estimated that the documents will be destroyed January 1, 2015. All data about the subjects will not include any names, addresses, or other identifying information.

15. Test Subject Consent
The research staff will provide a consent form for each subject. The subjects will be required to read, understand, and sign the consent form. All known risks and benefits will be discussed with the subject verbally and also included on consent form. An unsigned copy of the consent form

may be mailed to the potential subject prior to testing so they may have the time to study it and discuss it with their family or friends before making the decision.

Consent Form: Bioscience

Title of research study: Wearable Robot for Running Augmentation

Investigator: Dr. Thomas Sugar

Why am I being invited to take part in a research study?

We invite you to take part in a research study because we are determining the effectiveness of a running augmentation device.

Why is this research being done?

Dr. Thomas G. Sugar and his student are designing a running augmentation device. The device has the potential to increase a runner's top speed or the distance they are able to run. A device such as this could have potential benefit in military applications where soldiers are required to run at high speeds, potentially in life threatening situations.

How long will the research last?

We expect that individuals will spend about 5 hours on a single day participating in the proposed activities.

How many people will be studied?

We expect about 4 people will participate in this research study.

What happens if I say yes, I want to be in this research?

It is up to you to decide whether or not to participate.

If you decide to participate, then as a study participant you will join a study involving research of a powered wearable robot device to assist running.

Your participation will last for 1 day at the Human and Machine Integration Laboratory at Arizona State University at the Polytechnic Campus in Mesa Arizona.

You will interact with 2 people involved in the research study, Dr. Thomas Sugar and Mr. Jason Kerestes.

You will be asked to come to Arizona State University at the Polytechnic Campus to perform the tests. The testing day will last for 5 hours which includes four tests, a 10 minute training session, 10 minute familiarization with the device, a one hour resting period before tests begin, and three 30 minute resting sessions between tests. The schedule of events is broken down as follows:

1.   You will be issued and asked to wear safety equipment required for the test (helmet and hearing protection)

2.   You will go through a 10 minute training demonstration of how the device works and how to use it. You will have the opportunity to use the device for 5 to10 minutes before testing to become familiar with the controls and operation of the device.

3.   After training, you will have the opportunity to rest for up to 1 hour, and will then be asked to run on a set path at a speed of 10 mph to warm up.

4.   You will then be asked to stretch your muscles before beginning each test.

5.   (Test 1) To begin testing, you will be asked to run on a designated 200 meter track either with or without the device. While running, your heart rate and time will be recorded as well as a high speed video.

6.   You will then have a 30 minute rest time before your next test.

7.   (Test 2) You will then be asked to run the same 200 meter track again, either with or without the device and your heart rate and time will be recorded as well as a high speed video.

8.   You will then have a 60 minute rest time before your next test.

9.   (Test 3) You will then be asked to run on a designated 1 mile track either with or without the device. While running, your heart rate and time will be recorded as well as a high speed video.

10.   You will then have a 30 minute rest time before your next test.

11.   (Test 4) Then you will be asked to run the same 1 mile track either with or without the device. While running, your heart rate and time will be recorded as well as a high speed video.

12.   After completing the 4 tests, you will be asked to give your feedback on the device.

What happens if I say yes, but I change my mind later?

You can leave the research at any time it will not be held against you.

If you decide to leave the research, there will be no adverse consequences, and no one will try to convince you otherwise or change your mind. If you decide to leave the research, contact the investigator so that the investigator can prepare another candidate.

Is there any way being in this study could be bad for me?

There are certain risks and discomforts that may be associated with this research. They include but are not limited to:

1.   Physical risks associated with running such as tripping and falling either due to the runner's ability or lack thereof, or the device adding additional force to the subject.

2.   Physical risks associated with running at high speeds such as increased impact forces at the user's joints, (knees, ankles, hips)

3.   Physical risks associated with intense exercise such as strained or pulled muscles.

4.   Hearing risk associated from high noise levels produced by the ducted fans may have a short term effect on the runner's hearing.

5.   Fatigue or increased stress levels associated with elevated cardiovascular exercise.

6.   Running at increased speeds could reveal underlying heart disease

7.   There is a chance for the two batteries to overheat and catch fire

8.   There is a chance of tripping and falling and breaking a bone

To minimize potential risks, you will be required to wear and utilize a protective full face helmet. Should you fall while running, the helmet will protect your face and head. You will also be required to wear foam ear plugs to prevent any hearing damage. The current device has a noise level of 84 dB which is below the threshold for 8 hours of use.

The batteries will be mounted in an aluminum box to minimize the risk of batteries catching fire and containing a fire to protect the user. Fire-proof material will be mounted behind the battery box to minimize burns.

Before each test you will be asked to stretch to reduce the possibility of strained or pulled muscles. Rest periods will be provided between tests and as needed to alleviate fatigue and stress, however you may remain fatigued during or after the testing procedure. If any part of the test does not go accordingly, the device will utilize a remote emergency switch, allowing the research staff to remotely and instantaneously disable the device. The user will also be able to push an emergency switch on the backpack to turn off the system at any time.

You will be asked to wear a heart rate monitor. If your heart rate goes above the maximum level, the heart rate monitor will beep and you will be asked to stop.

From CDC, "This maximum rate is based on the person's age. An estimate of a person's maximum age-related heart rate can be obtained by subtracting the person's age from 220. For example, for a 50-year-old person, the estimated maximum age-related heart rate would be calculated as 220 - 50 years = 170 beats per minute (bpm)."

Participation in this study may cause all or some of the side effects listed above. In addition, there is always the risk of developing previously unknown side effects.

If mild side effects or discomforts do occur, the research staff will attempt to minimize these effects by asking you to rest. If any serious side effects occur, the research staff will attempt to minimize these by directing you to the emergency room.

If there are any major side effects, the research staff will call 911 immediately.

The investigator is willing to discuss any questions you might have about these risks and discomforts.

What happens to the information collected for the research?

Efforts will be made to limit the use and disclosure of your personal information, including research study, to people who have a need to review this information.

What else do I need to know?

If you agree to participate in the study, then consent does not waive any of your legal rights. However, no funds have been set aside to compensate you in the event of injury.

If mild side effects or discomforts do occur, the research staff will attempt to minimize these effects by asking you to rest. If any serious side effects occur, the research staff will attempt to minimize these by directing you to the emergency room.

Who can I talk to?

If you have questions, concerns, or complaints, or think the research has hurt you, talk to the research team: Dr. Thomas Sugar at 480-727-1127.

This research has been reviewed and approved by the Bioscience IRB ("IRB"). You may talk to them at (480) 965-6788 or research.integrity@asu.edu if:

- Your questions, concerns, or complaints are not being answered by the research team.
  - You cannot reach the research team.
  - You want to talk to someone besides the research team.
  - You have questions about your rights as a research participant.
  - You want to get information or provide input about this research.

Signature Block for Capable Adult

Your signature documents your permission to take part in this research.

_____	_____
Signature of participant	Date
_____	
Printed name of participant	
_____	_____
Signature of person obtaining consent	Date
_____	
Printed name of person obtaining consent	