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Analysis of a Periodic Force Applied to the Trunk to Assist Walking Gait

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***Abstract*— Past research has shown that a horizontal force applied to the trunk greatly assists the user's gait by reducing metabolic cost and reducing the horizontal ground reaction force. Different from the literature which describes a constant applied, horizontal force, the authors hypothesized that a horizontal tether force will not be constant but will oscillate due to the periodic nature of human walking gait. This hypothesis was tested by analyzing the tether force, ground reaction forces, and gait kinematics and kinetics of six able-bodied human participants. An assistive device was designed by attaching a spring-tether to the user's trunk and the other end was affixed to the treadmill. The user naturally found a position on the treadmill to stretch the spring tether. Multiple tethers with different stiffnesses were used. The subjects were asked to walk on a treadmill at 1.2 m.s^{-1} while wearing the assistive device. Motion data, volumetric rate of oxygen consumption ($\dot{V}O_2$) data, and the tether force data were collected. Three iterations of the tests were performed per participant, and the data was averaged. The horizontal assistive force was found to be periodic with twice the frequency of the gait cycle. Also, $\dot{V}O_2$ was found to be lower when wearing the device. The tether with the lowest stiffness was found to be the most effective in terms of reducing the metabolic rate. The authors concluded that the assistive device supplied power just at push-off for each foot while reducing the metabolic rate by 17% on average.**

I. Introduction

Understanding how to assist walking gait is crucial in designing rehabilitation devices for stroke therapy, assistive devices for the elderly, devices for people with weak muscles, and devices created to replace a joint or limb. How best to apply a force to assist gait is still not perfectly understood. Some systems apply joint torques while other systems apply external forces. The timing of these forces/torques is very important because if the assistive action is not in synchrony with a particular gait motion, the joint or leg motion is perturbed and metabolic cost increases. In literature, applying a constant external force to the trunk is described as quite beneficial. However, when we applied a constant pushing force with a jetpack to a person that was walking, we did not see any beneficial results. However, we did see reduced heart rate and metabolic cost with an assistive jetpack when running. Our goal was to analyze the timing and frequency of a horizontal tether force that was applied to assist a user's walking gait.

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II. Literature Review

Many research teams have studied the effect of external forces on human walking and running. Donovan and Brooks [1] established that horizontal impeding forces were directly proportional to the amount of energy spent while walking. Bijker et. al. [2] studied the effects of various factors on running and cycling, and found that a horizontal opposing force led to higher metabolic cost in both cases. Cooke et. al. [3] and Lloyd and Zacks [4] observed a rise in energy consumption with an increase in the horizontal impeding force. Bastien et. al. [5] found a direct relation between walking speed, energy spent, and the amount of load carried, where the energy spent increased with the load carried, until a certain speed. Chang and Kram [6] studied the effects of horizontal forces on running using the rate of oxygen consumption ($\dot{V}O_2$). They observed that horizontal forces constituted more than one third of the total metabolic cost of running. Similar observations were made by Kerestes and Sugar [7] while applying a constant thrust force at the trunk using a jetpack while running. They noted a 3.5% decrease in heart rate and a 2.5% increase in running speed while using the jetpack. In their article from 2003, Gottschall and Kram [8] proposed that an optimized, constant, horizontal force acting on the human body would result in a reduction in the metabolic cost even while walking. They applied a force that aided as well as opposed the walking gait motion. Their conclusion was that an aiding force of 10% of the subject's body weight was optimal and reduced the net metabolic cost of walking by 53%. However, in the literature articles, all groups assumed a constant horizontal tether force is required. However, human walking gait is known to be periodic based on the gait cycle.

Multiple researchers modeled the human gait as an inverted pendulum. For instance, Buczek et. al. [9] compared an inverse pendulum model to the gait of 24 healthy children and found the model to be reliable. Sun et. al. [10] developed a human gait model with higher efficiency than a simple inverted pendulum but with similar characteristics ie. periodicity. In their article, Tesio et. al. [11] noted the energy fluctuation of the CG (center of gravity) of their test subjects. This fluctuation was noted to be periodic in nature.

Because the CG for walking gait on a treadmill oscillates about a fixed position, and the inverted pendulum model describes an oscillation between kinetic energy and potential energy, the authors decided to examine the applied tether force at the trunk. None of the previous research tested the effects of an oscillating force on human walking gait. Researchers have applied oscillating forces to the joint such as the work by Collins et. al. [12]. They developed a clutch-based exoskeleton for the ankle which used a spring to apply a periodic braking force at the ankle while walking. This exoskeleton reduced the metabolic cost of walking by about 7.2%. In this article, the authors aim to analyze and describe the assistive force applied to the trunk and determine if a periodic assistive force is beneficial.

III. Methods

To test the hypothesis that the assistive force required at the trunk is periodic in nature, the authors measured the tether force, and gait kinematics and kinetics. All the tests were conducted under the protocol as per the guidelines of the Institutional Review Board at Arizona State University (STUDY 00009416).

A. Subjects

Six healthy participants volunteered for the experiment. Their age, weight, height, and ground to hip height was measured. The participants were also asked to grade their daily activity level from 0 to 10 (0 denoting no activity at all and 10 denoting a healthy amount of exercise daily). The measured values can be found in Table 1.

TABLE I. Physical Characteristics Of The Subjects (N=6)

Physical Characteristics	Mean	SD	Range
Age (years)	28	8.81	18-46
Weight (kg)	91.37	12.1	61.32-107.5
Height (m)	1.78	0.04	1.75-1.87
Hip to ground distance (m)	1.03	0.01	1.02-1.04
$\dot{V}O_{2\ peak}$ (mL.kg^{-1}.min^{-1}) (Resting Metabolic Rate)	3.32	0.51	2.636-3.967
Daily Activity Level (0-10)	3	2	1-6

B. Protocol

Each participant was asked to first walk on the treadmill to become adjusted to treadmill-walking before any experimental procedure began. Three trial sets were designed. In each set, the participants were asked to sit for 10 minutes and relax while their Resting Metabolic Rate (RMR) was measured. For the tests, they were asked to walk on a treadmill a fixed 1.2 m.s^{-1} in order to make consistent comparisons of the metabolic rate. The first test was done without the device to collect the control data. The participants then performed four trials (walking) with different tethers that varied in stiffness. The sequence of different tether stiffness values were randomized for each set of the trials. Kinematic, kinetic and ground reaction force data for the subject's gait were collected and compared.

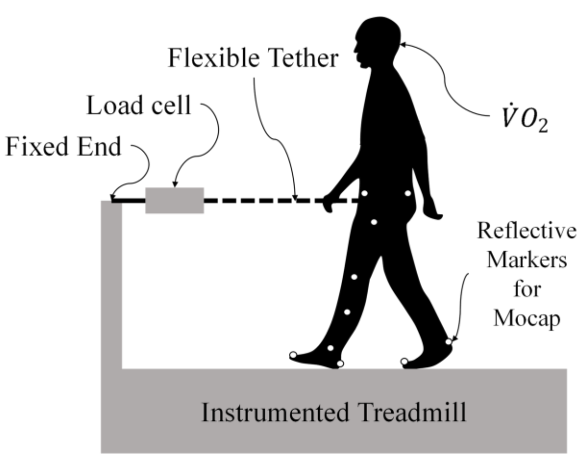

Figure 1. Experimental setup for the tests performed using the tethered exo-suit; Mocap, Motion Capture; $\dot{V}O_2$, Volumetric rate of oxygen uptake

C. Exosuit Apparatus

The exosuit consisted of a hip brace with a flexible tether attached to its front using metal carabiners. The stiffnesses of these tethers were 1600, 2100, 2600, and 3200 kg.m^{-1}. The other end of the flexible tether was attached to a load cell. The load cell was then attached to a fixed bar on the treadmill. The brace was designed to allow the tether to be changed as per the test requirement. The subject was equipped with a harness to prevent any fall related injury. It was made sure that the harness didn't provide any vertical force to the subject when standing. The tether force was designed to assist the user by applying a horizontal aiding (assistive) force during gait (assisted gait). During the tests, it was observed that the tethers were never slack and always applied some force to the test subject.

D. Data Collection Devices

Various instruments were used to measure the necessary data during the experiment. The volumetric rate of oxygen consumption ($\dot{V}O_2$) was measured using a Fitmate Pro (a desktop metabolic device that used a flowmeter to measure the amount of air inhaled/exhaled). The subject's gait kinematic and kinetic data were collected using a Vicon motion capture system. Multiple reflective markers were placed on the subject's legs as instructed in the Vicon Nexus user documentation (Lower body modeling with Plug-in Gait). Ground Reaction Forces (GRF) were measured using a Bertec Instrumented Treadmill which had different sensors for the right and left foot GRF. The experimental setup is shown in Figure 1. The subjects were asked to wear the flexible tether exosuit. The load cell used was a FUTEK LSB300 sensor with a FUTEK amplifier. The load cell was attached to a rigid frame bolted to the treadmill. Data was collected from this load cell using an Arduino micro-controller.

E. Metabolic cost measurement

$\dot{V}O_2$ for the subjects were measured for each trial. The equations for metabolic cost demand $\dot{V}O_2$ and $\dot{V}CO_2$. However, in their recent article, Kipp et. al. [13] tested ten published equations to measure metabolic cost. They found the equations to be varying but comparable. In all the cases mentioned by Kipp et. al., the metabolic cost was shown to be proportional to $\dot{V}O_2$. Hence we decided to consider $\dot{V}O_2$ as a measure of the metabolic cost.

Data was collected for three minutes and averaged to determine a metabolic rate.

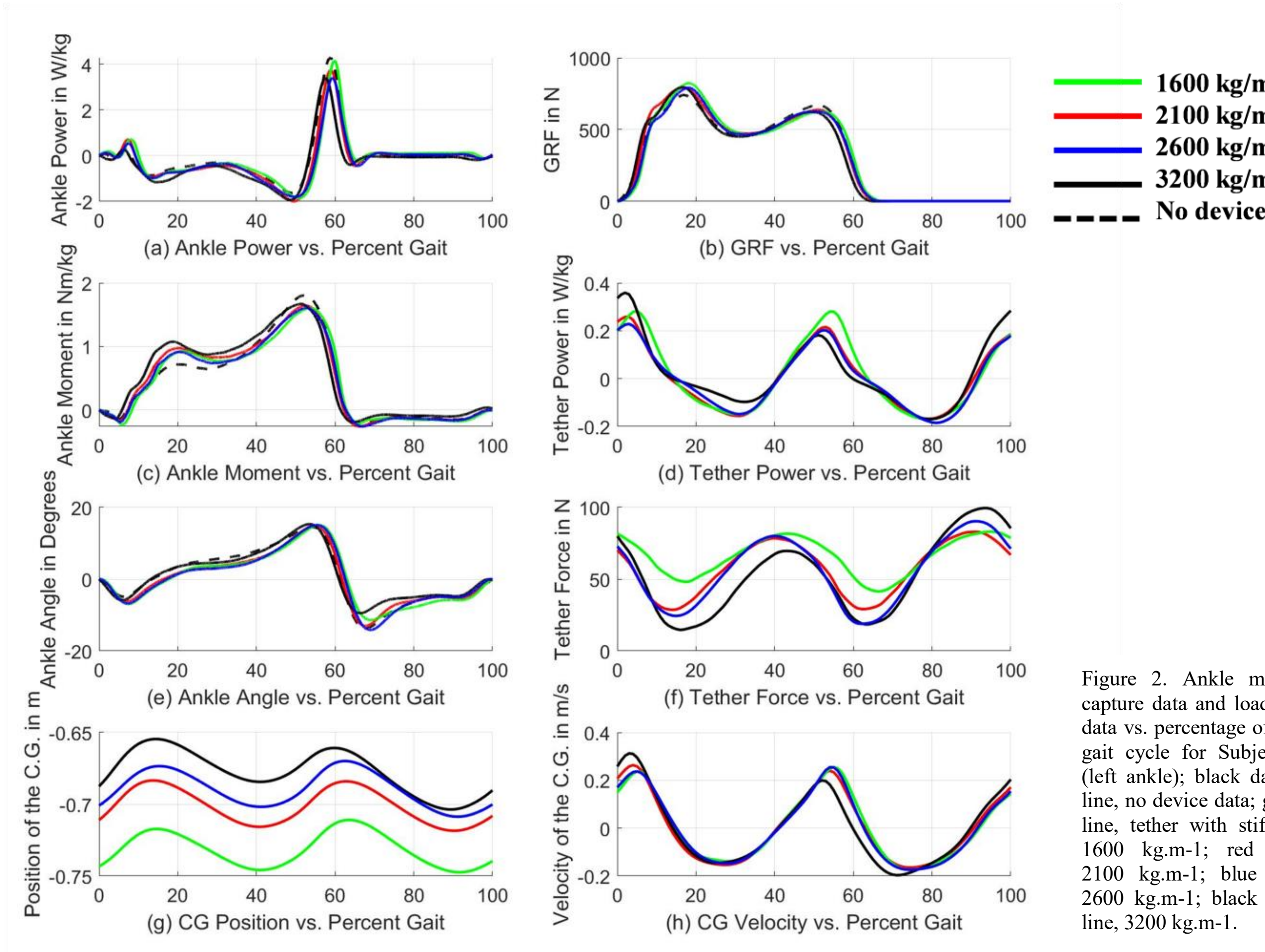

Figure 2. Ankle motion capture data and load cell data vs. percentage of the gait cycle for Subject D (left ankle); black dashed line, no device data; green line, tether with stiffness 1600 kg.m-1; red line, 2100 kg.m-1; blue line, 2600 kg.m-1; black solid line, 3200 kg.m-1.

F. Post-processing

The data collected from the tests were post-processed as required. The Vicon motion capture data yielded trajectories and joint angle outputs. These trajectories were filtered using a Butterworth filter (Low-pass with cut off frequency of 3Hz). The joint angles were obtained using the plug-in gait post-processing pipeline within the Vicon Nexus software. The data collected from the load cell were smoothed using a smoothing spline operation in Matlab. The RMR data for each subject was subtracted from the $\dot{V}O_2$ data. This was done to calculate the actual amount of oxygen consumed for walking in each trial. All the data collected from the tests were divided into individual gait cycle data. These individual gait cycles were averaged for each participant to obtain the results discussed below.

We decided not to average the group of gait cycles for all participants because of individual differences. Individual peaks can be reduced and timing can be shifted when all data is averaged. We wanted to clearly determine how the tether power aligned with different characteristics in the gait cycle for each participant.

IV. RESULT

All the data collected from the tests confirmed the main hypothesis: the force applied to the human body while performing gait was not constant but periodic in nature. The tether forces, as shown in Figure 2 (f), were periodic in a given gait cycle. The force delivered by the flexible tether oscillated at an average frequency of 1.92 Hertz. The average gait cycle for the test subjects lasted for about 1.03 seconds. On average, the gait cycle frequency was 0.97 Hertz. Hence the horizontal force applied to the human body during gait was about twice the walking gait cycle frequency. The tether force oscillated at double the gait frequency to assist both the left and right legs. The other data collected from the tests showed similar characteristics.

Figure 2 showed the data collected for subject D's left ankle. The peak power (59% gait cycle) delivered at the subject's left ankle while wearing the device was observed to be 14.2% lower on average than when wearing no device (Figure 2 (a)). The 3200 kg.m^{-1} tether (black line) showed the least peak ankle power with a reduction of 20.85% compared to no device data. Figure 2 (b) showed the vertical GRF data for all the tests conducted on subject D. The GRF showed an 8.13% increase on average during the weight-acceptance phase (specifically 19% gait cycle) and a 13.65% decrease on average during push-off (51% gait cycle). Figure 2 (c) showed the normalized ankle moment generated by the subject's left ankle. On average, an increase of 32.89% (19% gait cycle) and a decrease of 9.38% (52% gait cycle) was observed as compared to the no device data.

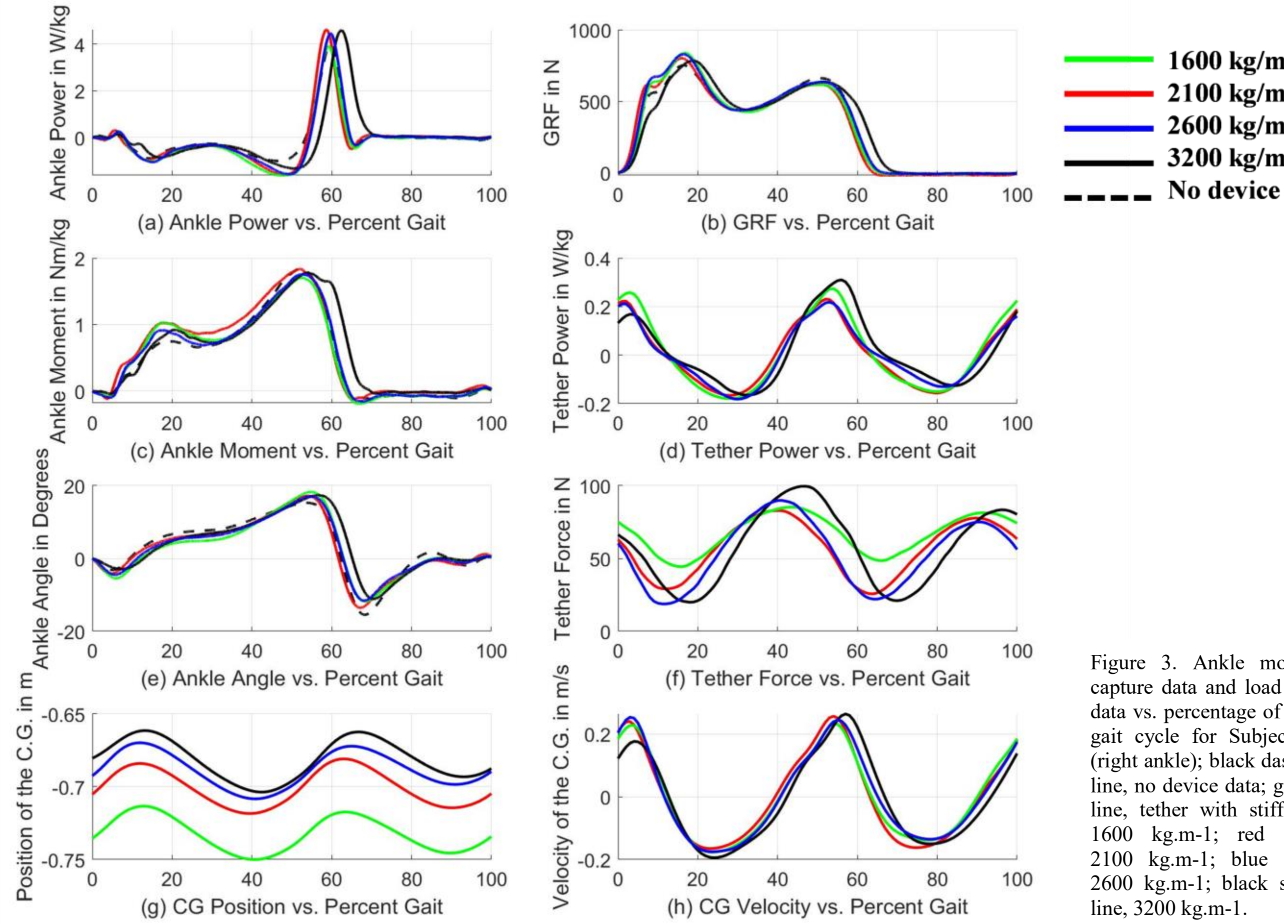

Figure 3. Ankle motion capture data and load cell data vs. percentage of the gait cycle for Subject D (right ankle); black dashed line, no device data; green line, tether with stiffness 1600 kg.m-1; red line, 2100 kg.m-1; blue line, 2600 kg.m-1; black solid line, 3200 kg.m-1.

Figure 2 (d) showed the amount of normalized power delivered by the tether to the trunk of the participant. A spike was observed in the tether power during push-off. The 1600 kg.m^{-1} tether (green line) delivered the most power at 6.58% of the peak ankle power while wearing no device. All the other tethers delivered a power of 4.18% of the peak power on average. Average plantarflexion decreased while wearing the device by 14.6% (67% gait cycle) (Figure 2 (e)). The average forces delivered by the 1600, 2100, 2600, and 3200 kg.m^{-1} tethers were 66.04, 57.39, 55.55, and 52.28 Newtons respectively as seen in Figure 2 (f). These forces were 11%, 9.56%, 9.26%, and 8.71% the subject's body weight respectively.

Figure 3 showed similar data for Subject D's right ankle. The peak ankle power (59% gait cycle) reduced on average by 10.74% when wearing the device (Figure 3 (a)). The 3200 kg.m^{-1} tether did not perform as well on subject D's right ankle as in the left ankle. As seen in Figure 3 (b), the GRF showed an 8.08% increase on average during the weight-acceptance phase (specifically 19% gait cycle) and a 5.2% decrease on average during push-off (51% gait cycle). The subject's ankle moment showed an increase of 28.56% (19% gait cycle) and a decrease of 4.14% (52% gait cycle) (Figure 3 (c)). In the right ankle, the 1600 kg.m^{-1}delivered a power of 6.91% of the peak ankle power while wearing no device. The other tethers delivered 5.38% the peak power on average (Figure 3 (d)). The average forces delivered by the 1600, 2100, 2600, and 3200 kg.m^{-1} tethers were 66.93, 56.57, 53.45, and 57.19 Newtons respectively as seen in Figure 2 (f). These forces were 11.15%, 9.43%, 8.91%, and 9.53% the subject's body weight respectively.

The metabolic cost results showed a reduction when wearing a tether. Figure 4 showed the box and whisker style plot of $\dot{V}O_2$ data for all the tests conducted on the six subjects. The $\dot{V}O_2$ median with no device was 10.65 ml.kg^{-1}.min^{-1} while the medians for the assisted gait were lower at 8.61, 8.86, 9.01, and 9.19 ml.kg^{-1}.min^{-1} respectively. The interquartile range of the no device data was also higher than the data while wearing the device. Also, Table 2 defines the means of the $\dot{V}O_2$ data. The % force per body weight provided by the tether was around 10% on average. The mean $\dot{V}O_2$ was higher for gait without the device and lowest for the 1600 kg.m^{-1} tether. All metabolic reductions were significant at the 0.01 level or better. This proved that the device reduced the user's oxygen consumption while walking.

V. Discussions

All the results indicate that when an aiding force was applied to a human subject during gait, the metabolic cost (which was considered to be proportional to $\dot{V}O_2$) decreased significantly. This conclusion was consistent with Gottschall and Kram's [8] results that showed that a 10% body weight

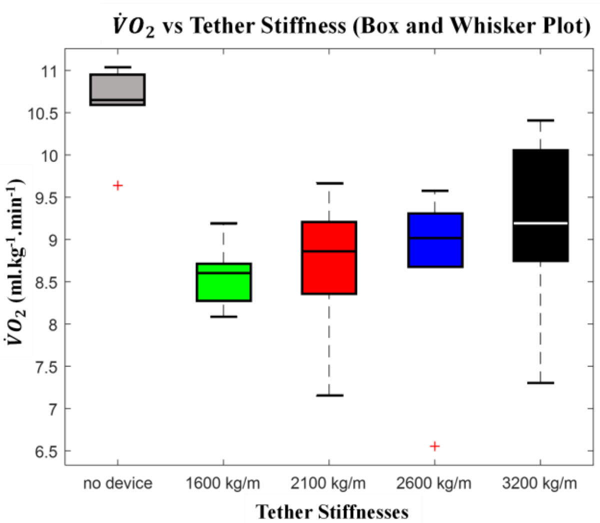

Figure 4. $\dot{V}O_2$ boxplot for all tether stiffnesses used during the tests

assistive force reduced the metabolic cost of gait. Their tests did not account for or describe a periodic force for walking. As can be seen in Table 3, the frequency of gait was about twice that of the force applied by the 1600 kg.m^{-1} tether.

TABLE II. MEAN $\dot{V}O_2$ WITH THE MEAN ASSISTANCE PROVIDED BY THE TETHERED EXO-SUIT. (* P < 0.01 AS PER ANOVA)

Tether Stiffness (kg.m^{-1})	$\dot{V}O_2$ (ml.kg^{-1}.min^{-1}) (Mean ± SD)	% Reduction in $\dot{V}O_2$ Compared to no device	% Assisting Force Per Body Weight (Mean ± SD)
0 (no device)	10.58 ± 0.45	N/A	N/A
1600	8.57 ± 0.34	19.0%*	9.7 ± 3.1
2100	8.68 ± 0.80	18.0%*	10.73 ± 2.34
2600	8.69 ± 1	17.9%*	10.73 ± 2.041
3200	9.14 ± 0.99	13.6%*	10.57 ± 2.59

The tether applied 9.7% of the user's body weight on average. Other tethers showed a similar behavior. The periodicity in the assistive force required could be accounted for in the braking and propulsive phases of walking gait. As evident from the data in Figure 2, the force applied decreased during the braking phase (10% - 25% gait cycle), increased before the push-off phase (around 45% gait cycle) and the tether delivered power during the push-off/propulsive phase (55% - 65% gait cycle). This was due to the flexible tether releasing its stored spring potential energy during push-off. The released energy was transformed into less effort required by the user's ankle. It was also observed that the subject's ankle applied higher moment during the braking phase to resist the tether and lower moment during the push-off/propulsive phase because the tether supplied push-off power. A decrease in the subject's GRF on push-off was evident in all subjects. This reduction in GRF was also shown by Gottschall and Kram [8].

The behavior of the tether and the subjects CG was found to be interesting. The position of the CG, as seen in Figure 2 (g) and figure 3 (g), was closer to the fixed end of the tether with a higher value of tether stiffness. This meant the subject could pull the tether with lower stiffness more easily than the higher stiffness tethers.

The CG position amplitude was very consistent as a percentage of the gait cycle. The CG velocity also had less variability in terms of frequency and magnitude as compared to the tether force. The tether forces varied considerably as a percentage of the gait cycle for the different stiffness values. However, once the tether force was multiplied by the CG velocity to determine tether power, the variability in timing was reduced. The magnitude of power delivered varied but the tether powers in plots (d) match with the push-off force in plots (b). The frequency of the tether power was consistent, and the tethers released power precisely during push-off.

Four out of six participants mentioned the device as being helpful. They stated that walking with the device was easier and less strenuous than without the device. While most of them felt the device assisted their gait, one subject expressed their discomfort while wearing the device. They felt the device barely helped them and that walking without the device felt more natural. In the case of this subject (subject B), the % assisting force per body weight was lower than the other subjects. Also, a smaller decrease in $\dot{V}O_2$ was observed in this participant's case. The authors believe more testing is required to investigate the factors involved. This participant may have needed more time to train with the device and "accept" a larger tether force.

TABLE III. EFFECT OF THE 1600 KG.M^{-1} STIFFNESS TETHER ON ALL SUBJECTS (MEAN ± SD)

Subjects	Average Tether Force Amplitude (Newtons)	Average Tether Force Frequency (Hertz)	% Assisting Force Per Body Weight	Average Gait Frequency (Hertz)
A	15.69 ± 0.69	2.17 ± 0.01	7.98624	1.09 ± 0.01
B	24.66 ± 1.58	1.83 ± 0.01	4.700679	0.91 ± 0.01
C	13.76 ± 0.9	1.96 ± 0.02	12.54512	0.97 ± 0.01
D	23.87 ± 4.05	1.82 ± 0.01	9.849237	0.91 ± 0.01
E	23.69 ± 6.15	1.79 ± 0.02	14.27931	0.91 ± 0.01
F	18.84 ± 1.79	1.88 ± 0.03	8.819339	0.95 ± 0.03
Average	20.08 ± 4.27	1.91 ± 0.13	9.7 ± 3.1	0.96 ± 0.06

It was also noted that during the experiment, the subjects leaned back to maintain equilibrium between the gravitational force and the assistive force. This behavior is similar to downhill walking. The $\dot{V}O_2$ data collected during the experiment conformed to the observations of Margaria [14] in relation to uphill and downhill walking. This similarity was noticed by Gottschall and Kram as well. Also, they noted that a comparison between an aiding horizontal force and downhill walking was debatable as the effects of change in kinetic and gravitational potential energy would be different in both the mentioned cases.

Interesting, reducing the braking force in early stance has been shown to reduce metabolic cost by Collins et. al. [12], but the tether increased the braking force. In our case, the overall metabolic cost was reduced probably due to a reduction in metabolic cost while assisting push-off.

VI. Conclusions And Future Work

All the results indicate that when an assistive force is applied to a human participant during gait at the right moment, the metabolic cost (which is proportional to $\dot{V}O_2$) is reduced. The results from this work can be used in the design of a novel rehabilitative exosuit. An exosuit would need to aid the user's joints at the right time (proper frequency tuning with respect to the user's gait). The joint torque or force would have to be oscillatory in nature. A soft impedance force would work better than a hard and stiff force. In this work, a force aiding the user to move forward at a resonant frequency to the user's gait was found to be the most beneficial.Future work will continue the same study and test tethers with even lower stiffness as well as apply a constant force similar to prior research.

Appendix

Figure 5 shows the left ankle data for Subject F.

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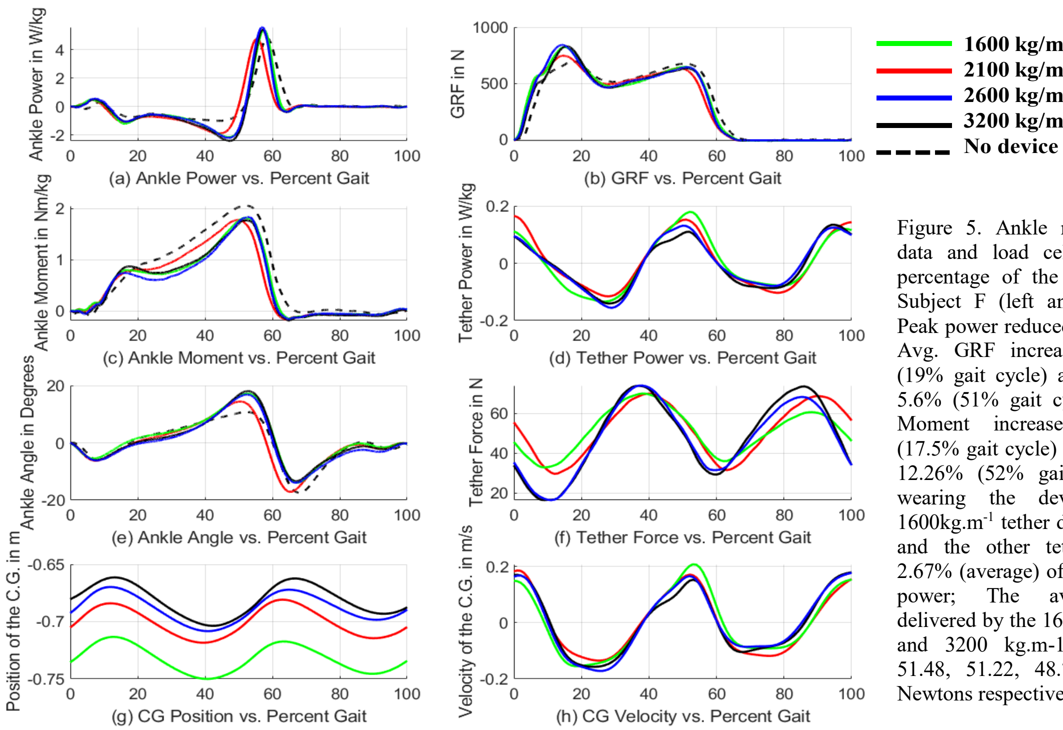

Figure 5. Ankle motion capture data and load cell data vs the percentage of the gait cycle for Subject F (left ankle); (a) Avg. Peak power reduced by 9.68%, (b) Avg. GRF increased by 16.6% (19% gait cycle) and reduced by 5.6% (51% gait cycle), (c) Avg. Moment increased by 4.78% (17.5% gait cycle) and reduced by 12.26% (52% gait cycle) when wearing the device; (d) the 1600kg.m^{-1} tether delivered 3.71% and the other tethers delivered 2.67% (average) of the peak ankle power; The average forces delivered by the 1600, 2100, 2600, and 3200 kg.m-1 tethers were 51.48, 51.22, 48.75, and 47.68 Newtons respectively