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An experimental device for investigating the force and power requirements of a powered gait orthosis

Brent J. Ruthenberg, MSME; Neil A. Wasylewski, BSME; John E. Beard, PhD
Department of Mechanical Engineering and Engineering Mechanics, Michigan Technological University, Houghton, MI 49931

Abstract—The Powered Gait Orthosis (PGO) is a powered exoskeleton developed as an experimental device to provide bipedal locomotion to individuals with physical impairment. The current prototype consists of a single degree of freedom (DOF) system for each leg, providing power and proper displacement required for bipedal locomotion. It is the goal of this research to obtain the forces that are present in the device while it is in normal operation. In addition, the time ratio of the hip function generator has been varied to determine the effect that different time ratios have on system forces and required user energy. The time ratio is the relationship between the time period that the thigh is in swing phase and when it is in support phase. Knowing the forces in the system and the optimal time ratio will allow for the design and construction of a feasible device for the rehabilitation and assistance of individuals who have lost the ability to walk.

Key words: *bipedal locomotion, gait, ground reaction forces, hip torque, multiple sclerosis, orthosis, powered walking exoskeleton.*

INTRODUCTION

For a powered exoskeleton to be used as an aid in walking it must be lightweight, easy to don and doff, simple to maintain, inexpensive, unobtrusive, quiet, and have sufficient walking distance on a single charge. To design such a device, one of the requirements is an understanding of the forces transmitted through the device to the user so the weight and power requirements can be minimized. The forces and power that occur during gait can be divided into two components: those transmitted through the individual (skeletal loading), and those transmitted through the brace. This article will focus on a study conducted at Michigan Technological University (MTU) to determine quantitatively the amount of force that such a device must support during gait and the power needed as a function of step cycle.

Background

Currently, researchers at MTU are studying the design of a device known as the Powered Gait Orthosis (PGO) as shown in **Figure 1**. This particular configuration of the PGO is the fourth generation of prototypes. **Table 1** details the significance of each prototype used in the research program. The long-term goal of the program is to develop an orthosis that would be used in a home or work environment as an alternate to the wheelchair for certain tasks. The physical profile of the users would probably fit characteristics defined by Seireg and Grundman (1): "full upper extremity strength capability; moderate to full trunk stability; the patient can shift body mass with conscious effort; the patient can stand with bracing."

Many researchers have investigated the feasibility of exoskeleton type walking braces, including externally powered, patient-powered, and hybrid systems. A brief history of these devices would include *Handyman* and

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Address all correspondence and requests for reprints to: John E. Beard, PhD, Department of Mechanical Engineering and Engineering Mechanics, Michigan Technological University, 1400 Townsend Drive, Houghton, MI 49931-1295; email: jebeard@mtu.edu.

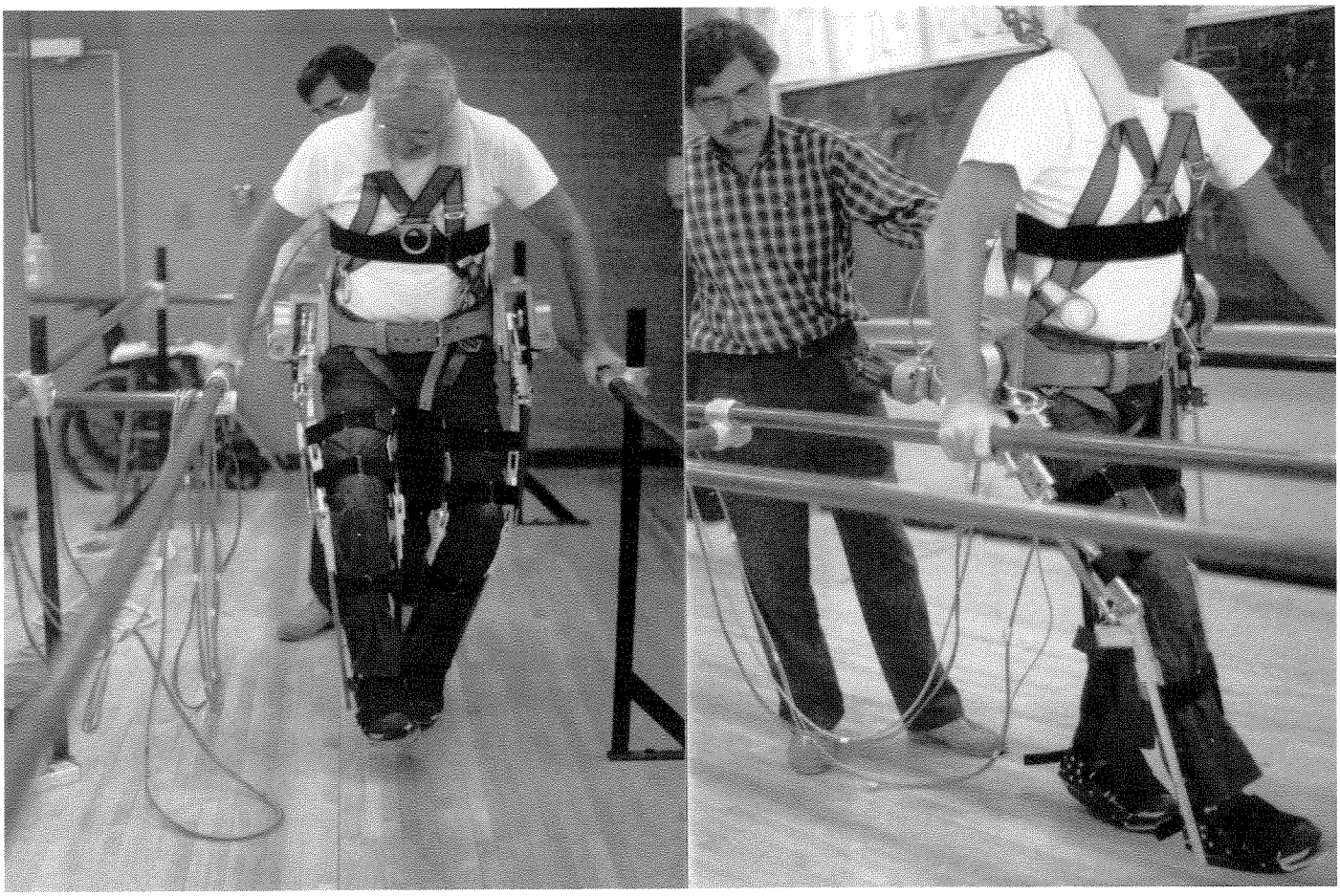

Figure 1.
Powered gait orthosis.

Table 1.
Prototype progression of PGO.

No.	Highlights/Accomplishments
1	Swing gait, pendulum-like device with fixed knee investigated power requirements tested different function generators established state testing approval helped in recruiting of test subjects
2	Added power for knee flexion found that minimal power is needed at knee, as predicted by Inman (2) found no advantage in varying motor speed, decreased battery life
3	Hydraulically actuated design provided an accurate, semi-programmable, repeatable gait system friction prevented torque determination (many variables) found that a large percentage of forces were carried by skeletal loading
4	Mechanized hip and knee design added cam-modulated linkage for knee function generator able to vary time ratio to investigate effect on gait performed complete force analysis

Hardiman, cybernetic anthropomorphous machines developed by General Electric for the military, to be worn as exoskeletons for strength multiplication (3). Kato et al. did extensive research in powered prosthetics for persons with transfemoral amputation (4). Thring built an exoskeleton for arthritics to transfer the load from the knee joints to a leg brace through a bicycle type seat attached to the leg braces (5). As reported by Vukobratoivic, powered exoskeleton systems that were capable of climbing stairs were developed in Warsaw, Poland (6).

Popovic and Schwirtlich, with their self-fitting modular orthosis (SFMO), proposed a nonmotorized device to

restore locomotion (7). Through his earlier work, Popovic concluded that a totally powered external brace was an inappropriate technique for gait restoration in subjects with spinal cord injury (SCI). The ORLAU ParaWalker™ was developed as a practical patient-powered device and is established with an extensive patient performance base (8,9). The Louisiana State University (LSU) reciprocating gait orthosis (RGO) was initially developed for use by children and is now used as a patient-powered device for both children and adults (10). Stallard et al. added functional neuromuscular stimulation (FNS) to the ORLAU ParaWalker for a hybrid system with success in decreasing the energy requirements of the user (11). Solomonow has also been successful in adding FNS to the RGO (12). Seirig and Grundman have concluded, in contrast to Popovic, that an externally powered device is feasible, although their device was primarily used as a research tool (1).

METHODS

Design

Each leg of the current PGO prototype is a one degree of freedom (DOF) system that produces a repeatable gait. A one DOF system requires a single input (motor) to control the positions of all of the links in the system. In the case of the PGO, the input for each leg is created by a 13.2 volt DC motor that turns the input gear, Link 2 (**Figure 2**). The battery pack and control circuit for the motors can be fastened to the back of the corset.

The walking cycle is initiated when the user or lab assistant pushes the start button on the control pad. Once motion for a particular leg is initiated, the control circuit allows it to go through exactly one cycle (toe-off to heel-strike), at which time the button for other leg is pushed. This causes the leg initially at rest to proceed to heel-strike from toe-off. As familiarity increases with the device, it is possible for the user to initiate the motion of the second leg before that of the first is complete. This process is very similar to natural gait in form. The controls for automatic repetitive motion will be used during the next testing stage to insure that users will take a continuous, full stride as they transverse the length of the laboratory.

The hip is controlled by a class one, four-bar (i.e., crank-rocker) mechanism that consists of Links 1, 2, 3, and 4, where 1 is attached to the user's torso (which acts as the reference link or kinematic ground for the mechanism). The knee is actuated with the use of a cam-modulated linkage, the cam profile being machined into the face of Link 8 (lower gear) and the follower attached to Link 5. Link 2 (upper gear) is the crank of the four-bar mechanism, and it also causes the rotation of the lower gear, which in turn drives the cam-modulated linkage. Since the cam follower is captured in a slotted cam profile, it always remains in contact with the cam during flexion and extension. This produces a *fixed* effect: though the knee function is always changing, at each instant in time (percent of cycle) the knee is at a specific *fixed* point, unable to change from its intended function.

Since it is the primary goal of this project to mechanically approximate natural human gait, it is necessary to have a quantitative definition of natural gait. Inman provided the knee and thigh angles needed as boundary conditions for the evolution of the PGO's hip and knee function generators (1). Inman's work considered bipedal locomotion of many nondisabled test subjects. **Figures A-1** and **A-2** of the Appendix compare the gait angles defined by Inman and those produced by the PGO. Notice in **Figure A-2** that the knee output of the PGO shows no discernible difference to that defined by Inman. Due to its simplicity and nature, the four-bar link-

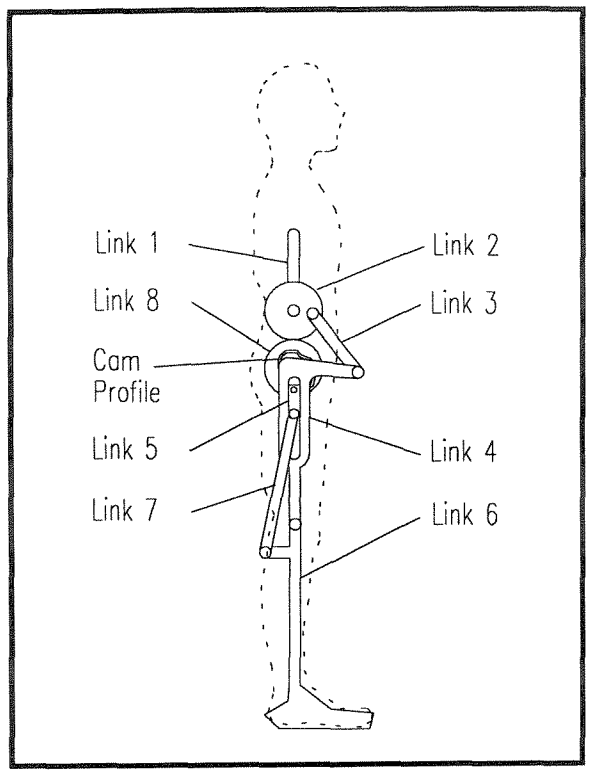

Figure 2.
Kinematic sketch of the PGO.

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age used for the hip function is not able to achieve the same degree of replication found at the knee. It was determined that a compact cam modulated linkage was not feasible for the hip due to the large power requirements there. A feasible cam mechanism, and hence better accuracy, was possible at the knee because of the relatively low power requirements of that joint.

Subjects

Two test subjects have participated in the experiments conducted on the PGO. Testing began at LSU with an individual who has an incomplete SCI at the C2 level and was a quadriplegic in the rehabilitation stage. The majority of the work conducted at LSU was in determining power requirements of the PGO (13). The other test subject participated at MTU; he is in the advanced stages of multiple sclerosis and has minimal use of his legs. These subjects may not have been the ideal candidates for testing of the PGO, but both volunteered, lived in close proximity to the testing labs, and had experience walking with the RGO or leg braces. Additional candidates with disabilities better suited for use of the PGO are available and will be tested when local living arrangements can be made.

Force and Torsional Analysis

Weight reduction of the PGO requires that all forces and torques induced in the system during normal operation be known. It was determined that six full bridge axial transducers and two torsional transducers would be sufficient to completely define the loading conditions at all of the joints in each leg (14). Since the PGO is symmetric about the sagittal plane, it was assumed that each leg experiences the same type of loading; therefore, only the right leg of the PGO was strain gaged.

Once the strain gages were installed and calibrated, a multichannel data acquisition board was used to simultaneously record all force and torsional data. This was done repetitively for both the swing and support phases to assure accuracy. After the data were compiled, a representative average was found and used for further analysis.

Comparison of Results Obtained to Previous Work

A considerable amount of research has been performed to determine the ground reaction forces that occur in natural human gait. Most of these studies have involved having nondisabled test subjects walking or running across a force platform. The platform is used to measure the vertical and horizontal components of force in the direction of forward motion. By comparing force platform results to PGO forces, an understanding as to how the weight of the PGO affects the function of the device can be asserted.

Using the data obtained by Alexander and Jayes (15), a comparison can be made with the PGO forces. In this study, the reaction forces were recorded for the time period from heel-strike to toe-off and the data reported as the magnitude of the vertical and horizontal forces measured. Normalizing their data to the weight of their subjects yields the dashed curve of **Figure 3.** Note that the forces during the swing phase are not shown in that figure.

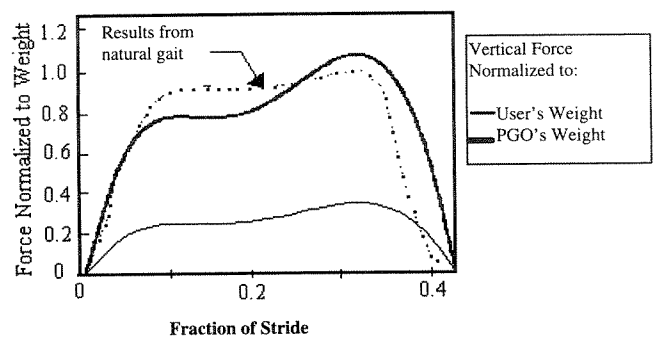

Figure 3.
Ground reaction forces in natural gait and those supported by the PGO.

Using only the data from the MTU PGO study where the foot is in contact with the ground, the two additional curves of **Figure 3** were formulated. In order to compare the PGO results to those of Alexander and Jayes, the MTU results needed to be normalized to weight. The lower curve has been normalized to the weight of the user, which, except for its basic shape, does not reflect the force platform results. Therefore, in an effort to better understand the force distribution between the PGO and user, the results were normalized solely to the weight of the PGO. This curve, shown as the heavy line in the figure, closely follows the results of Alexander and Jayes' force platform. This would suggest that the PGO carries its own weight while the user's weight is carried through skeletal loading (i.e., the user is not carrying the weight of the device).

Effect of Varying the Time Ratio

In the initial design of the PGO, along with the thigh and knee angles mentioned above, the time ratio was used as an invariant for design comparisons. The time ratio of gait can be defined as the ratio of the time the foot is unsupported, the swing phase, to the time the foot is in

contact with the floor, the support phase (16). For optimization purposes, it was important to know how this affected the forces induced in the system and the energy required by the user.

It was decided to vary the timing ratio between 1.0 and 1.2 for this study. **Figure A-3** of the Appendix is a sketch showing the four vectors that compose the thigh function generator of the PGO that controls the time ratio. An attribute of the design illustrated in **Figure A-3** is that the thigh input link is perpendicular to the vector r_4. This allows the torque at the hip joint to be measured easily, since it is simply the product of a constant scalar multiple (the length of r_4) and the axial load on the thigh input link.

RESULTS

Figure 4 is a plot of the hip torque determined by multiplying each of the force measurements by their respective moment arm lengths and normalized to the user's weight. The curves follow the same general trends, but the maximum values decrease with increasing time ratio.

The angular velocity of the motor that drives the PGO is relatively constant, varying by only ±1.6 percent throughout the cycle. Since power is the product of torque and angular velocity, it is, therefore, a reasonable assumption that the average power consumption of the PGO is proportional to the average hip torque.

Given that the average hip torque increases, and that the thigh transducer forces decrease with increasing time ratio, there should be an optimal value that best satisfies the choice of power reduction or force reduction. If the trends described are plotted so that their scales are compatible, there will be a point of intersection providing the time ratio that would give the minimum forces with the lowest power consumption. In an attempt to determine the optimal time ratio, the data obtained were plotted in the fashion shown in **Figure 5.**

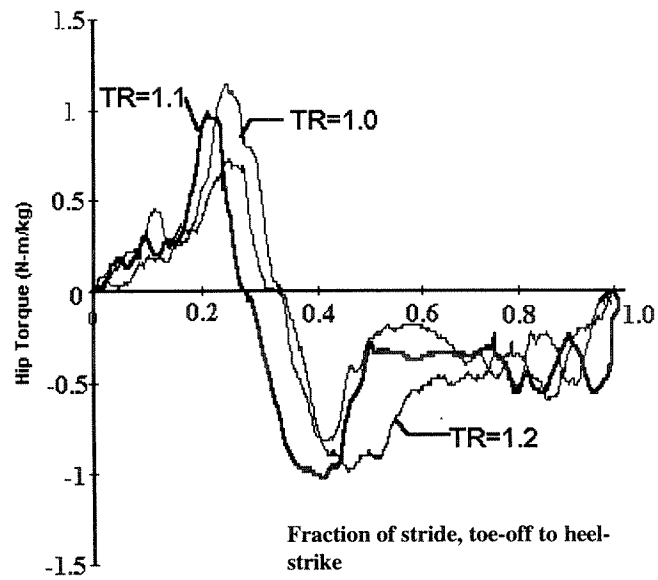

Figure 4.
Hip torque for three different time ratios.

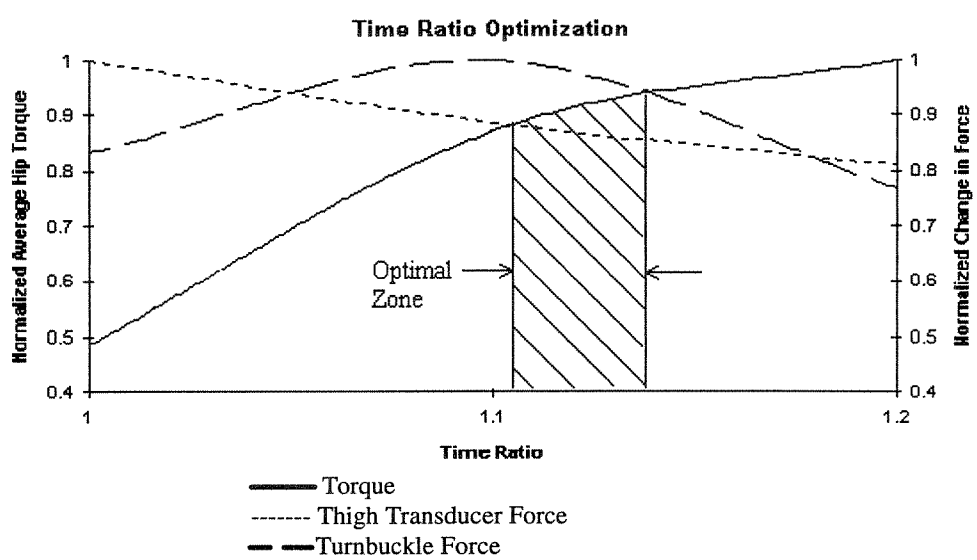

Figure 5.
Time ratio optimization plot.

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By considering both the hip and knee function generators, an optimal zone can be established in **Figure 5.** The optimal zone includes time ratios from 1.106 to 1.139, where:

- a time ratio of 1.106 corresponds to minimum forces in the hip function generator with minimum power requirements
- a time ratio of 1.139 corresponds to minimum forces in the knee function generator with minimum power requirements.

Patient Energy Expenditure

The torque results above are not significant if they are considered independent of other variables. Because the PGO is a device that is designed to assist people, the amount of energy that is needed from the user is of great significance. Low forces and torques could mean a very light, inexpensive design, but if it requires excessive energy from an already weak user, that design will not be feasible. As an easy method to monitor the exertion of the human body, pulse and blood pressure data were recorded before and after each of the time ratio tests. Although it is not the best method for energy expenditure determination, it has been shown by Stallard (17) that there is a well-defined linear relationship between oxygen uptake and heart rate. Therefore, for the initial phases of this study, the intra-subject comparison of pulse and blood pressure data was used to evaluate the user's energy input. This method would not be used as a comparison between patients, and eventually, oxygen consumption will be used to quantify the user's energy expenditure.

Figure 6 plots pulse and blood pressure data. As one would expect, the pulse did increase after each test was completed. It can also be seen that the change was independent of the time ratio used, and the change in pulse was 6 beats/min for each test.

Although there does appear to be a change in the blood pressure with changing time ratios, the randomness of this change may suggest that it is within the natural variation associated with blood pressure. Even though a detailed relationship can not be determined from this data between energy expenditure and time ratio, one important observation can be made: there is not an apparent negative relationship between user energy input and time ratio selected; therefore, this selection can be based on the kinematics and power concerns of the system.

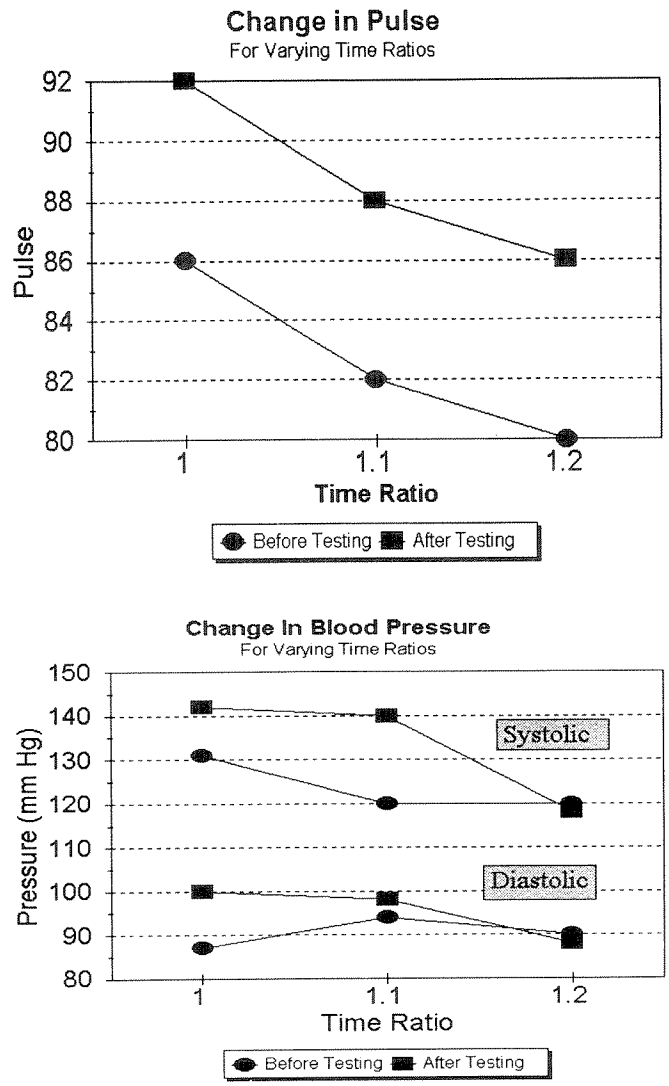

Figure 6.
Pulse and blood pressure before and after each test.

DISCUSSION

A complete force and power analysis of the PGO has been achieved. The forces were obtained with proven strain gage techniques, making it possible to accurately measure specific force and moment components in regions of combined loading. The forces obtained agree with the results published by Alexander and Jayes, suggesting valid results. These force and moment results suggest that a lightweight design of the PGO is possible, making it seem even more feasible for rehabilitative and

assisted walking. However, since it was determined that the PGO carries its own weight and this weight is not transferred to the user, so long as the device does not become overly cumbersome, in terms of this research, a larger device will only affect the motor selection, not user energy requirements. For that reason, the device was purposely over-designed in areas not considered in this study, such as the torso interface. This minimizes the chance that these areas will influence results in more critical areas. Obviously, when the device leaves the laboratory for home use, the size and weight of the device will be more important, especially during donning and doffing.

The effect of varying the time ratio of the hip function generator has been studied. The results show that the time ratio of the hip function generator is of significant importance on both the power requirements of the system and the forces introduced into the system. In addition, the optimal time ratio zone was determined to be between 1.106 and 1.139. It was also witnessed that the required energy expenditure of the user was not strongly correlated to the time ratio chosen. This would suggest that the design can incorporate a timing ratio that proves feasible kinetically and kinematically, not one based on difficult-to-determine energy requirements of users.

During the course of the torque measurements on the PGO, it was found that the power requirements of the device were not much different from the power necessary for natural gait. It has been shown that natural human gait requires a peak power of approximately 300 watts (2) where the PGO was found to require approximately 23 amps at 13.2 V DC (304 watts) at its most demanding point when the user had inadvertently shifted his weight onto the nonload-bearing leg just prior to toe-off. **Figure 4** shows the hip torque measured on the PGO (normalized to user weight). The maximums observed in this figure are approximately 20 percent higher than those determined for natural gait by Winter (18). This difference may mostly be attributable to the lack of mobility inherent to an exoskeleton type device. The human body adjusts both upper and lower limbs in an effort to minimize energy, but the PGO restricts these adjustments and minimizes the feedback for balance adjustments, increasing the power necessary for gait. Also, the PGO greatly simplifies a very complicated process. In nonimpaired gait, 11 joints in the pelvis and legs (lumbosacral, bilateral hip, knee, ankle, subtalar, and metatarsophalangeal joints) are used for movement (19). In each leg, motion is controlled by 57 muscles. However, in each leg of the PGO, motion is provided through two joints (knee and hip) and by only two force generators. This simplistic approach, although effective, does increase the required power in the system.

The next step in the project will be to use the knowledge obtained through this and previous research to design the next generation of the PGO. The intended goal for the next prototype is to obtain a significantly lighter device that will provide a better replication of natural human gait with greater adaptability. The projected weight of the next prototype is approximately 98.1 Newtons (10 kg). This adaptability will allow the researcher and the user to vary the gait functions for different situations.

Whatever form the next generation prototype takes, it will be constrained by the strength and required energy requirements of the user. In addition to the energy requirements while using the device, the ease of donning and doffing the PGO must be addressed. In order to evaluate these quantities, many methods will be used in an effort to quantify them. The following methods are available to evaluate user energy inputs:

- Oxygen consumption ($\dot{V}O_2$)
- Carbon dioxide consumption ($\dot{V}CO_2$)
- Respiratory exchange ratio ($\dot{V}CO_2/\dot{V}O_2$)
- Heart rate
- Blood pressure
- O_2 debt with recovery time
- Basal metabolic rate in M.E.T.S.

These will allow a better, more efficient device to be developed that will aid in the rehabilitation of individuals who have temporarily lost the ability to walk or assist those with a permanent loss of bipedal locomotion.

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APPENDIX

Kinematic and Dynamic Model of the PGO

Kinematic Analysis

The PGO consists of an eight-link mechanism for each leg, attached to a rigid hip corset (**Figure 2**). The eight-link mechanism has nine revolute joints, one roll-slide joint (cam) and one gear pair (which kinematically counts as a roll-slide joint) in the sagittal plane; therefore, each leg is a one DOF system. The position, velocity, and acceleration of all links may be determined from a single input position, velocity, and acceleration for a single DOF system.

A four-bar mechanism (crank-rocker type) is attached to the torso and utilizes Link 1 as the reference frame. Link 1 translates with the torso, and may rotate, and it is used as the moving reference frame. The input to the system is Link 2. The output of the four-bar is Link 4. Link 4 is attached to the torso at the axis of rotation of the hip joint, both on the orthosis and on the patient. The frame of the orthosis connecting the hip and knee joint is rigidly connected to Link 4 and rotates as shown in **Figure A-4.**

The power to rotate Link 2, and thus Link 4, is provided by a 13.2 volt DC motor, operating at 12,380 RPM when attached to the orthosis and without the subject in the PGO. The motor speed is reduced to 406.2 RPM by 4 pairs of spur gears and reduced to 20.30 RPM by a 20/1, double lead worm gear reduction for a total gear reduction of 609.6/1. This provides an input speed at the hip joint of 20.3 RPM. Neither the spur gear pairs for speed reduction nor the worm gear pair are shown in the diagram. The worm gear reduction is also a self-locking device that prevents unwanted joint rotations when the motor is off and the user is standing in the PGO.

To provide power for the knee flexion, gear 2 was attached to Link 2 with the center of the gear located at the center of the pivot between Links 1 and 2. The mating gear (Link 8), gear 3, has its center at the center of the hip joint on the orthosis[1]. Gear 3 rotates at the same angular speed as gear 2 and in the opposite direction. Although gear 3 and Link 4 have a common pivot (or instant center) their rotations are independent. By locating gear 3 at the hip joint, both motion and power can be obtained for flexion of the knee independent of the hip angle. The motion of the knee angle is provided by the cam machined into (integral with) gear 3. Because the angular position of Link 4 is known as a function of the input angle using well-known kinematic analysis (20,21) and because the desired angle of the knee is known from the Fourier series describing the knee in Equation 3, the cam shape can be calculated for a given value of the cam base circle, r_{offset}, r_5, r_6, r_7, and r_8 link lengths. The cam base circle diameter was maximized to minimize the pressure angle between the cam and the follower (22). The cam follower is directly attached to Link 5. Links 4, 5, 6, 7, and 8 are classified as a cam-modulated linkage (22). The specific values for this particular PGO are given in **Table A-1.**

Table A-1.
Link lengths of PGO mechanism.

$r_1=2.500$	$r_2=0.643$	$r_3=2.088$	$r_4=1.518$
$r_{offset}=3.875$	$r_6=0.500$	$r_7=0.750$	$r_8=16.500$

Dynamic Analysis

A dynamic model of the PGO was developed using Lagrange's equations of motion (23) with the horizontal displacement of the hip joint as the generalized input. Lagrange's equations of motion state that the work done on a system is equal to the change in kinetic and potential energy of that system. An energy balance can be written using this postulate as represented below:

$$W = \Delta T + \Delta U + W_f \qquad [1]$$

where

W = *(work in) − (work out)*
ΔT = *change in kinetic energy of moving parts*
ΔU = *change in potential energy stored in the mechanism*
W_f = *energy dissipated through friction*

The differential equations of motion were obtained by taking the derivative of the energy balance. The derivatives for each mechanical link and each limb were obtained using the chain rule with the horizontal displacement of the hip as the generalized

[1]The numbering system for the gears starts at 2 for convenience.

coordinate that is directly related to time. Because the PGO is a mechanism, and the motions of all the links can be determined as a function of the hip angle Θ_{Hip}, a complete set of equations that describe the motion of the system as a function of the position, velocity, resisting torque, and so forth were obtained.

The Fourier series describes the vertical displacement of the hip joint and the angular rotation of the torso as a function of the position of the hip joint shown in the equations below.

$$\Theta_{Hip}=293.5°-18.18\cos(2\pi X)-2.74\cos(4\pi X)-0.67\cos(6\pi X)-0.942\cos(8\pi X)+0.16\cos(10\pi X) +10.096\sin(2\pi X)-5.28\sin(4\pi X)+1.43\sin(6\pi X)-0.98\sin(8\pi X)+1.46\sin(10\pi X) \quad [2]$$

$$\Theta_{Knee}=271.455°-8.07\cos(2\pi X)+6.79\cos(4\pi X)-3.86\cos(6\pi X)+2.22\cos(8\pi X)+19.04\sin(2\pi X) +2.22\sin(4\pi X)-0.98\sin(6\pi X)-0.59\sin(8\pi X) \quad [3]$$

In the initial analysis (23), the desired hip and knee function were both included in the dynamic model to predict the torque and power necessary for the prescribed motion. Both the thigh and calf were assigned individual masses and moments of inertia. Resisting torque for the hip and knee joints of the user were added as a function modeled with a Fourier Series (23,24) similar to Franken et al. (25). Included in the model were the frictional forces from predicted coefficients of friction between the floor and shoe. The kinematic analysis was used to predict toe-off and to correct the dynamic model for unsupported stance during swing-through. Once the initial torque and power requirements for the desired gait were calculated, the actual torque and power for the motion *generated* by the four-bar linkage at the hip were incorporated in the dynamic model. Because the knee motion was generated by a cam designed to accurately reproduce the desired motion, the actual dynamic model and desired dynamic model were the same and no correction was necessary. Assumptions in the basic model are:

1. The mass of each leg of the user can be represented as two lumped masses, one on the thigh and one on the shank. It is assumed that the masses have the same motion as that portion of the PGO.
2. The resisting torque of the user's leg due to muscle tone can be added as an external torque at the hip and knee joint.
3. The reaction force between the floor and the foot can be modeled as a normal and friction force.
4. Power lost due to friction is experimentally measured and a Fourier series or polynomial approximation is used to represent this function in the differential equation.
5. The upper torso is "attached" to the upper PGO frame in the model.
6. The movement of the skeletal structure in response to compression of the soft tissue by movement of the orthosis is minimal compared to the overall motion of the orthosis.

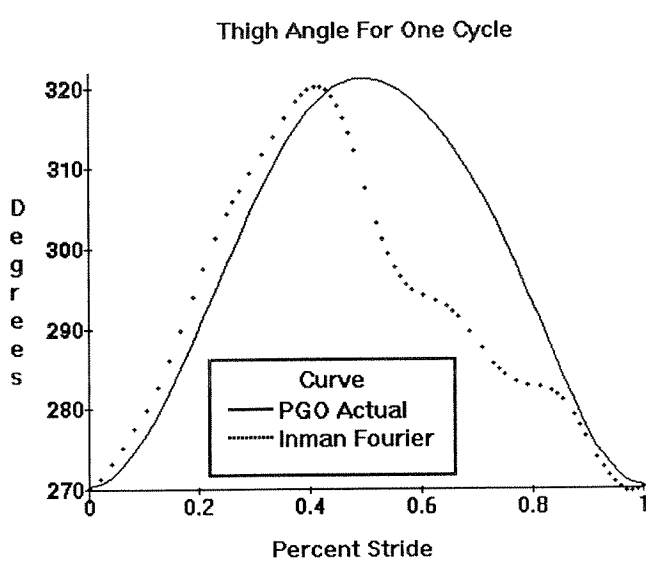

Figure A-1.
Hip angle provided by Inman and that produced by the PGO.

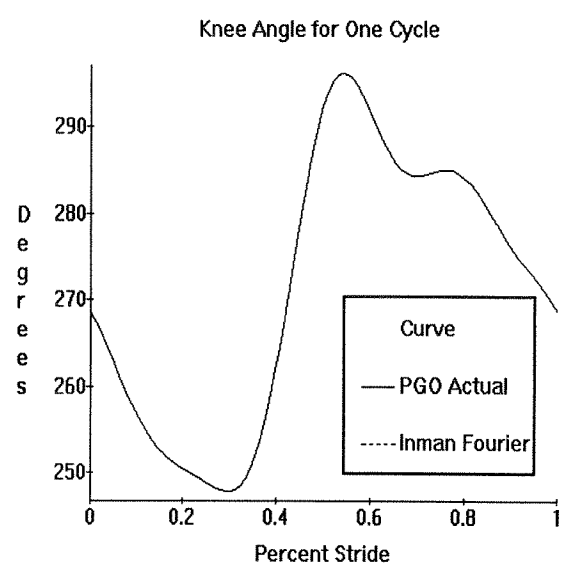

Figure A-2.
Knee angle provided by Inman and the essentially identical one produced by the PGO.

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The total weight of this version of the PGO (including batteries) is 265 newtons (26.75 kg) as compared to the 117.72 newton weight (12 kg) of the previous version. The significant increase in weight is the result of replacing the composite hip band with an aluminum hip band and increasing the thickness of the gears and linkage housing. The hip band and housing members were significantly stiffened to remove any measurable deflection. This insures that the strain measurements, and therefore the force calculations, reflect the deflections caused only by the power into the system and not a combination of the input power and "twist" of the orthosis frame by the user. A "stiff" and lightweight prototype PGO was possible to construct, but at an expense that exceeded the budget allocation for orthosis components.

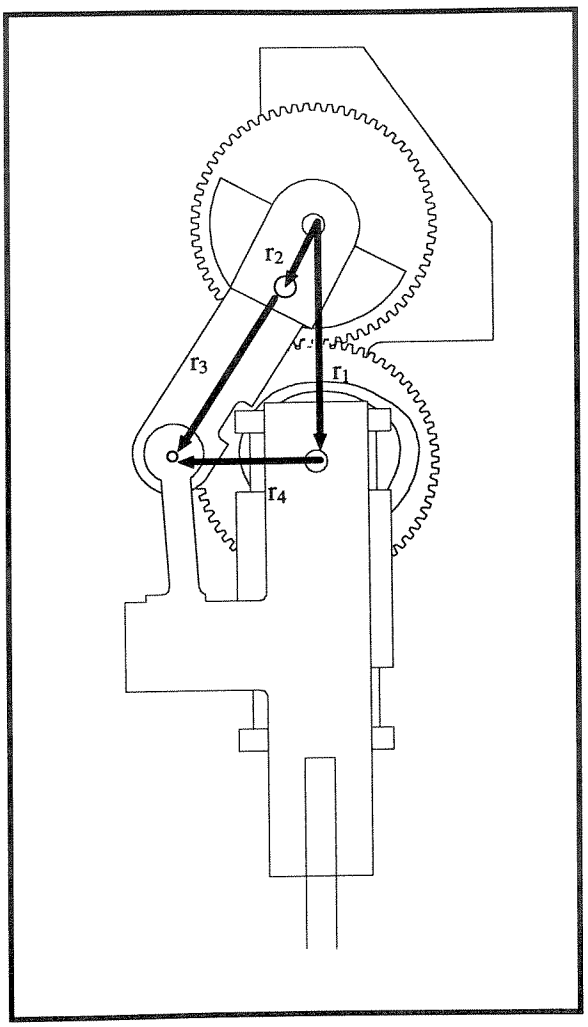

Figure A-3.
Sketch of vectors composing hip function generator.

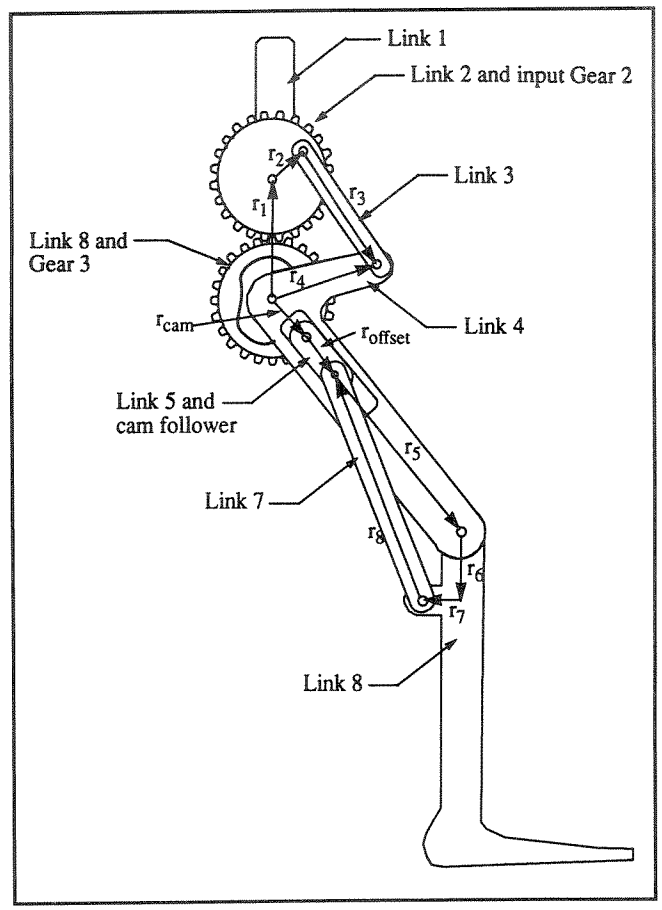

Figure A-4.
Vector loops used for analysis.

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