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MIT-skywalker

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MIT-Skywalker

Caitlyn Joyce Bosecker, Hermano Igo Krebs, *Senior Member, IEEE*

Abstract – The ability to walk is important for independent living and when this capacity is affected by neurological injury, gait therapy is the traditional approach to re-train the nervous system. The importance of this problem is illustrated by the approximately 5.8 million stroke survivors alive in the US today and an estimated additional 700,000 strokes occurring each year, many requiring gait therapy. This manuscript presents the design and proof-of-concept testing for a novel device to deliver gait therapy. While robotic devices to train gait therapy exist, none of them take advantage of the concept of passive walkers and most of them impose the kinematics of unimpaired gait on impaired walkers. Yet research has found that proper neural input and stimulation is a critical factor for an efficacious therapy program. This novel device might afford a more ecological gait therapy including heel-strike.

I. INTRODUCTION

Stroke is the third leading cause of death and the leading cause of permanent disability in the United States, with about 5.8 million stroke survivors alive today and an estimated additional 700,000 strokes occurring each year [1]. The effect of stroke on the ability to walk is significant with only 37% of stroke survivors regaining the ability to walk within one week post-stroke [2]. In addition to stroke, many other neurological conditions lead to significant gait impairment. For example, over 250,000 people in the United States have a spinal cord injury (SCI) with an estimated 11,000 new injuries each year [3]. In contrast to the stroke population, the average age of SCI individuals is 31. Individuals with incomplete injuries may be able to regain function through physical therapy and this is an active area of research.

In order for an individual to be able to walk, he must be able to do a minimum of four things [4]:

- 1) Each leg in turn must be able to support the full body weight without collapsing.
- 2) Balance must be maintained during single-leg stance.
- 3) The swing leg must be able to advance in order to transition into support stance.
- 4) Sufficient power must be provided to make the necessary limb and forward trunk movements.

Dr. H. I. Krebs and C. J. Bosecker have filed for patent protection. Dr. Krebs holds equity positions in Interactive Motion Technologies, Inc., the company that manufactures this type of technology under license to MIT.

C. J. Bosecker is a Masters student in the Department of Mechanical Engineering, Massachusetts Institute of Technology, Cambridge, MA, USA (email: bousecker@mit.edu).

H. I. Krebs is with the Department of Mechanical Engineering, Massachusetts Institute of Technology, Cambridge, MA, USA, with the Department of Neurology and Neuroscience, Weill Medical College, Cornell University, Burke Medical Research Institute, White Plains, NY, USA, and with the Department of Neurology, University of Maryland, School of Medicine, Baltimore, MD, USA (email H. I. Krebs: hikrebs@mit.edu).

A. Mechanical Assistance

Utilizing mechanical devices to deliver therapy is not a new idea and several have been developed for gait therapy. The most common mechanical device is the treadmill. It reduces the amount of space required for therapy and encourages patients to maintain a constant gait velocity. Numerous studies have been completed with both healthy and impaired subjects to compare gait kinematics between treadmill and overground walking and the effect of body weight supported treadmill training (BWSTT) on functional outcome. Riley et al. studied 33 healthy subjects and compared their overall gait kinematics for overground and treadmill walking and concluded that the kinematics were very similar, but that the ground reaction forces were significantly larger ($P < 0.05$) for the overground walking compared to the treadmill [5]. Similarly, Lee et al., Stolze et al., and Matsas et al. observed small differences in joint kinematics for healthy subjects, but concluded that treadmill walking did not show any negative effects on gait [6-8]. In studies that included hemiparetic patients, Nilsson found no difference in walking ability, balance, or sensorimotor performance between the BWSTT group and the subjects that completed overground training [9]. Hesse et al. found a significant difference ($P < 0.05$) between BWSTT and conventional Bobath physiotherapy concluding that treadmill training was superior [10].

As for body-weight support (BWS), Visintin et al. studied a population of 100 stroke patients that were randomly selected to complete treadmill gait therapy either with 40% BWS or without BWS for six weeks [11]. The results showed that the BWS group scored significantly ($P < 0.05$) higher than the no-BWS group for functional balance, motor recovery, overground walking speed, and overground walking endurance. This significant difference continued at the 3 month follow-up evaluation with the BWS having significantly higher scores for overground walking speed and motor recovery.

An additional advantage of using a treadmill for gait training is the ability to control and increase the speed at which therapy is being completed. Pohl et al. studied 60 stroke patients randomly chosen to receive either structured speed-dependent treadmill training (STT) (with the use of an interval paradigm to increase the treadmill speed), limited progressive treadmill training (LTT) (speed increased no more than 5% per session), and conventional gait training (CGT) (Bobath techniques) [12]. After a 4-week training period, the STT group scored significantly higher ($P < 0.01$) than the LTT and CGT groups for overground walking speed, cadence, stride length, and Functional Ambulation Category scores [12]. Lamontagne et al. also found a

positive result from increasing gait speed without any improvements in body and limb kinematics and muscle activation patterns [13].

In summary, treadmill training has not been found to have a detrimental impact and, in some studies, subjects that completed therapy with a treadmill regained more function compared to traditional physiotherapy techniques. With the added benefit of body-weight support, BWSTT reduces the strain on therapists and it is safe.

B. Robot-Mediated Therapy

While treadmill training improves efficiency, it still requires a therapist to monitor pelvis movement and another to propel the leg forward. Robotic devices were built in an attempt to automate the therapy process further. While several robotic devices already exist or are under development (e.g. MIT's Anklebot, KineAssist, Haptic Walker, UC Irvine's Pam and Pogo, Lopes, Motorika/Healthsouth Autoambulator), only two devices have been used extensively with published reliable outcomes, namely the Gait Trainer I and the Lokomat.

The Gait Trainer I has accumulated the largest body of evidence among these two devices with different RCTs suggesting its benefits to promote better stroke recovery [e.g., see 14, 15]. It is an end-effector based robot with quick set-up time and it incorporates both an adjustable BWS and sliding foot plates that are secured to the patient's feet. While it minimizes the need to use only one therapist for safety and to prevent the knee from over-extension, the planar sliding motion does reproduce heel-strike.

The Lokomat BWSTT system is the most widely adopted device with an estimated 130 rehabilitation centers employing it worldwide [16]. It is an exoskeletal device and it includes a treadmill, an adjustable and active BWS designed to provide a constant level of support throughout the gait cycle, and a robotic orthosis with four degrees of freedom (left and right knee and hip joints) [17]. The device allows vertical movements during gait, but it does not incorporate any means to promote weight shifting from one leg to the other. It imposes a fixed kinematic gait pattern determined from testing with healthy subjects.

While the results employing the Gait Trainer I were quite positive, several studies have been completed with the Lokomat with mixed results. Mayr et al. conducted a randomized blinded pilot study with 16 subjects in an ABA or BAB design (A = 3 weeks of Lokomat training and B = 3 weeks of conventional physical therapy). Both therapy groups improved their walking function significantly over the trial, and most improvement for both groups was observed during the Lokomat training block [18]. However, these findings were not replicated in larger studies by Hornby et al. [19] and Hidler et al. [20]. These studies compared Lokomat training to conventional therapy for stroke patients and found no advantage for the Lokomat training.

One might speculate that the Lokomat experience might not be affording the proper neurological stimulus. For example, Hidler et al. has shown that the muscle activation

observed negative effects and the increased speed induced patterns during Lokomat training differ significantly from normal treadmill walking [18]. Hornby et al. suggested that while the Lokomat reduces the strain on therapists and provides a safe environment for patients to practice walking, it also allows patients to remain completely passive and not actively engaged as it does not offer an interactive experience.

II. DESCRIPTION OF THE MIT-SKYWALKER

A. Dynamics vs. Kinematic Concept

Our novel device is distinct from any of the existing kinematic-based rehabilitation robotic devices for gait. It delivers safe and efficacious gait therapy inspired by the concept of passive walkers [21]. Passive walking devices are purely mechanical devices with no actuators, sensors, or controllers that are able to "walk" down slopes (Figure 1) [21]. The device shown in Figure 2 is able to walk down a 3.1° slope at a speed of about 1.67 ft/sec (0.51 m/s). The MIT-Skywalker implements this elegant dynamic concept to rehabilitation and human gait, creating the required ground clearance for swing and exploring gravity to assist during propulsion. Preliminary tests with a mannequin and unimpaired subjects demonstrated its ability to allow gait therapy without restricting the movement to a rigid, repetitive kinematic profile. It maximizes the amount of weight bearing steps with ecological heel strike. It also promotes active patient participation during therapy, while having a compact design that can be implemented in a variety of settings.

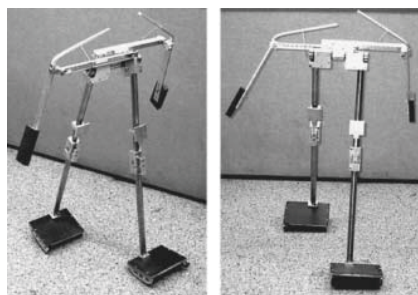


Figure 1. Two-legged passive walker [31].

B. Hardware Concept

We employed TRIZ in the design of the new gait robot. TRIZ is a Russian acronym which translates to, "The Theory of Inventor's Problem Solving," which was developed by Genrich Altshuller beginning in 1946 [22]. Altshuller recognized that one of the keys to the success of inventive ideas was that many problems stem from contradictions between two or more components or desired traits. He defined "inventing" as identifying and eliminating the contradictions. For this project, the contradiction encountered was that while a walking surface is necessary for gait therapy, this surface inhibits the leg during the swing

phase and requires intervention to clear the surface and propel the leg forward.

In conventional stroke physiotherapy, the therapist manipulates the pelvis and transfers the patient's body weight to the stance leg and pushes or slides the patient's swing leg forward. The same is true for the treadmill, where a therapist transfers the patient's weight to the stance leg while another therapist lifts the impaired swing foot of the ground and propels it forward. For the Lokomat, straps lock the ankle in dorsiflexion position, the body-weight system unloads the swing leg, and the orthosis propels the leg forward. Finally the Gait Trainer I is similar in function, but it does not require locking the ankle in dorsiflexion. The foot plates provide both simulated foot clearance and propulsion. These solutions are employed above a horizontal walking surface and do not attempt to alter it.

One of TRIZ principles applicable to this problem is "Do it in Reverse" [23]. For this project, altering the walking surface is the opposite of previous solutions. Instead of lifting the patient's leg manually or mechanically, we lower the walking surface which both provides swing clearance and takes advantage of dynamics to propel the leg forward. Contrary to existing devices, it will not specify a rigid kinematic profile that must be followed and it will address the need of allowing proper neural inputs provided by hip extension and ecological heel strike.

C. K'nex Proof-of-Concept

In order to quickly and inexpensively determine if actuating the walking surface was a feasible solution, a model was made using the construction toy K'nex and a 12 inch (30.5 cm) wooden mannequin (Figure 2). The mannequin was altered so that its legs would swing freely and it was crudely supported above the simulated treadmill. The treadmills were constructed from rubber bands and were hinged and driven by the same motor so the height of one was independent of the other.

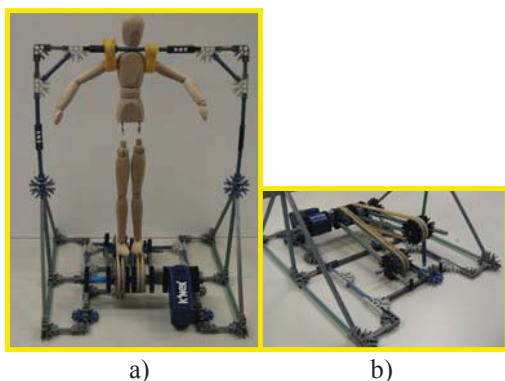


Figure 2. K'nex scaled proof-of-concept, a) system with 12" mannequin, b) close-up of treadmill concept.

The motor drove both "treadmills" at the same speed and they were "actuated" by hand. Even though the mannequin's legs had additional degrees of freedom compared to the average human (namely rotation about the vertical axis), by keeping the treadmill surface parallel to the ground during foot contact and then lowering to allow free swing, the

mannequin was able to maintain a cyclical gain pattern. This concept appeared to be a feasible solution and we proceeded developing a human-scale alpha-prototype.

D. Human-Scale Functional Requirements

To be effective, this device must be adjustable to accommodate a range of patients. We selected a range to cover the 99th percentile adult male and the 1st percentile adult female. Table 1 lists the measurements considered when determining the amount of necessary leg swing clearance and the dimensions of the BWS.

TABLE 1 SUBJECT ANTHROPOMETRICS [34]

	99% Man	1% Woman
Height (in) [cm]	75.6 [192]	58.1 [147.5]
Weight (lbs) [kg]	244 [111]	93 [42]
Chest width (in) [cm]	14.1 [35.6]	8.8 [22.4]
Average Stride (in) [cm]	27.4 [69.6]	20.2 [51.3]
Hip Width (in) [cm]	16.9 [42.9]	11.2 [28.4]
Ground to hip (in) [cm]	40.1 [101.8]	29.6 [75.2]
Ground to armpit (in) [cm]	60.6 [153.9]	45.4 [115.3]
Ground to crotch (in) [cm]	36 [91.4]	27 [68.6]

The MIT-Skywalker must provide a stable walking surface that is parallel to the ground, allow adequate clearance for a leg to swing without knee flexion, and return to the horizontal plane in time for the heel strike of the next stride. The average walking speed of a healthy individual is 3.3 ft/sec (1 m/s), but studies with stroke patients are often completed at speeds between 0.29 and 1.17 ft/sec (0.09 – 0.36 m/s) [24]. An appropriate treadmill must be able to operate through the whole range for use by severely to moderately affected subjects. Figure 3 shows the estimated foot clearance for the 99th percentile male and 1st percentile female for both a 10° and 15° toe-off angles.

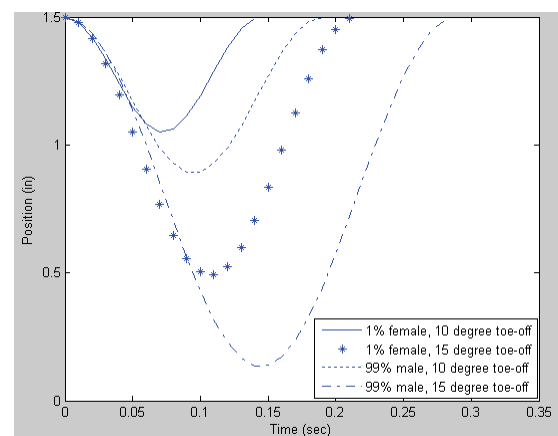


Figure 3. Foot clearance and time requirements for treadmill actuation.

While 10° is the common angle for healthy individuals walking at about 3.3 ft/sec (1 m/s), 15° was included as a factor of safety. These trajectories were calculated by treating the leg and heel as a simple pendulum acting under gravity. For the 99th percentile male with a 15° toe-off, 1.5

inches (3.8 cm) of swing clearance will be adequate. Also, for the 1st percentile female, the treadmill must be actuated in 0.14 seconds in order to provide swing clearance and return to parallel in time for foot strike.

The walking surface of the treadmill must be long enough for the subject to be able to complete a normal stride with an allowance for missteps, and the width must accommodate stance width. The recommended minimum treadmill length is about 60 in (150 cm) with a width of 25 in (60 cm) [35]. It should be noted that the width of the treadmill should not exceed 29.5 in (75 cm) because this will require the therapist to lean forward in order to access the subject's legs [35].

E. Scavenger Hunt

Split Treadmill

To build the first human-scale prototype, we searched for parts in our laboratory and at MIT. For a hinged split treadmill, we found a small footprint exercise machine called the BowFlex TreadClimber. In contrast to its use for exercise where the adjustable cylinders provide resistance as the user steps and pushes the treadmill down so it is parallel to the ground, a patient would be facing in the opposite direction and the treadmills would be externally actuated down or back to horizontal. The TreadClimber treadmill length is 40 inches (101.6 cm), which will be adequate for this alpha-prototype since the body weight support system will prevent the subject from shifting forward or backward along the treadmill. The total treadmill width is 18.5 inches (47 cm), which will allow the therapist easy access to the subject's legs.

Cam System

To actuate the treadmills so that they are parallel to the ground during stance, provide adequate swing clearance of 1.5 inches (3.8 cm), and return to the parallel position for heel strike, we designed a cam system (Figure 4).

While Figure 4 shows a single motor and camshaft for both cams, we employed independent cams controlled via dedicated motors and shafts. Although this would increase the cost of the system, it would nevertheless provide flexibility to actuate each treadmill at a different rate, especially for stroke where often only one side is affected. In order to determine the required speed of the camshaft, we estimated the time required of a stride at different treadmill speeds for subjects of different heights with a 10° toe-off angle and 30° heel strike angle, which is average for healthy gait. At the slowest treadmill speed of 1.03 ft/sec (0.3 m/s), the 99% male with a leg length of 40.1 inches (101.8 cm) has a stride length of 27.44 inches (70 cm) and each foot stance duration of 2.22 seconds. Since 60% of gait is stance, then the gait cycle time is 3.7 seconds. The cam would have to rotate at 16.2 rpm (1.68 rad/sec) to support this gait cycle. At the maximum treadmill speed corresponding to the average healthy walking speed of 3.28 ft/sec (1 m/s), the 1% female with a leg length of 29.6 inches (75.2 cm) has a stride length of 20.24 inches (51.4 cm) and each foot will be in stance for 0.51 seconds. The resulting gait cycle is 0.86 seconds and requires a camshaft speed of 70 rpm (7.33 rad/sec). In addition, altering the speed profile of the cam would change the vertical motion profile of the treadmills.

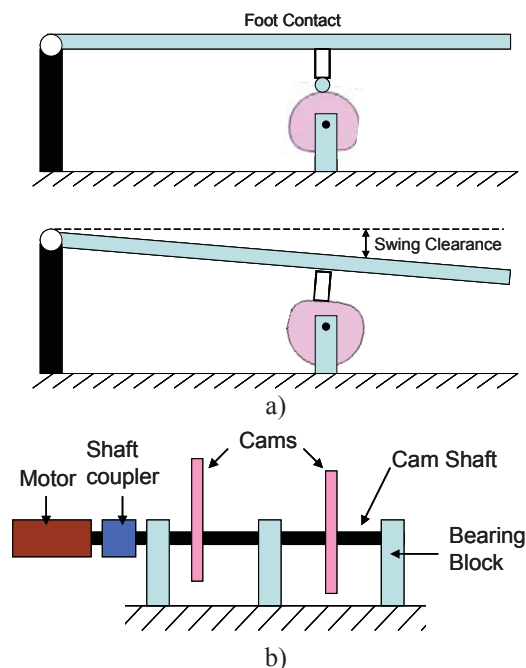


Figure 4. Cam design concept, a) cam design side view, b) front view.

The amount of torque required to turn the camshaft in order to actuate the treadmills must be calculated in order to choose capable hardware. Conservation of power was applied and the load of a track of the split treadmill is 18 lbs (8.2 kg). For the 99% male at the slowest treadmill speed, the cam has to lift the treadmill 1.5 inches in 0.74 seconds. The resulting average vertical velocity is 2 in/sec (5 cm/sec) and the approximate maximum velocity is 4 in/sec (10 cm/sec). The required torque is 42.86 in-lb (4.82 Nm). For the 1% female at normal walking speed of 3.28 ft/sec, the cam has to lift the treadmill 1.5 inches in 0.172 seconds. The average vertical velocity is 8.7 in/sec (22 cm/sec) and the approximate maximum velocity is 17.4 in/sec (44 cm/sec). The required torque for this configuration is 42.73 in-lb (4.83 Nm). At 16.2 rpm, 0.011 HP is required, and at 70 rpm, 0.047 HP is needed. If a factor of safety of 2 is included to account for bearing friction and efficiency, the required torque for both the slowest and fastest camshaft speeds is 85.72 in-lb (9.69 Nm) and 0.095 HP at 70 rpm. We were able to use a system already in our laboratory which employed Kollmorgen brushless servomotors. Table 2 lists the relevant characteristics of these motors.

For the range of users, the camshaft speed will be between about 16-70 rpm, which is much lower than the 3000 rpm speed that the motor is rated at. The continuous rated torque is 33.3 in-lb (3.76 Nm), which is less than the estimated 42.86 in-lb (4.83 Nm) calculated before any safety factor was applied. A gear reduction will be required to increase the available output torque, but contrary to all our other robotic devices backdriveability is not an issue here [25-29], we selected low cost 20:1 gearboxes.

In an average gait cycle, 60% is spent in stance and the remaining 40% spent in the swing phase. During stance, the treadmill should be parallel to the ground to provide a flat

walking surface, and during swing the treadmill must lower to provide swing clearance and then rise before the next heel strike. This 60/40 split will be used to design the radial cam profile with 216° (or 60% of 360°) of high dwell and the remaining 144° for the fall and rise.

TABLE II. KEY MOTOR CHARACTERISTICS

Specification	Value
Peak Stall Torque, T_{ps} (Nm)	9.67
Continuous Rated Stall Torque, T_{cr} (Nm)	3.76
Maximum Speed, ω_{max} (rpm)	6000
Rated Speed, ω_r (rpm)	3000

The dynamics of a cam will be determined by its position (S), velocity (V), acceleration (A), and jerk (J) [30]. These values are often combined in an SVAJ diagram. Although using straight lines to dictate the position profile does not appear to be a bad choice at first, the resulting derivatives encounter problems at the boundaries of the motion. The discontinuities in the velocity result in infinite spikes in acceleration. In order to achieve infinite acceleration, an infinite force would be required. Even though that is impossible, the sharp corners of the displacement diagram will create very large spikes in acceleration which will, in turn, create high stresses in the cam and cause rapid wear. The cam will then not retain the desired profile and the motion will change over time, both of which are unacceptable.

To eliminate discontinuities, the cam-follower function must be continuous through the first and second derivatives of displacement across the entire interval [30]. In addition, the jerk function must be finite across the entire interval. In order to abide by this law, the displacement function must be at least a fifth-degree polynomial [30]. We considered different cam profiles such as the single-dwell cycloidal displacement (sinusoidal acceleration), a double harmonic function, and a polynomial function.

Both the cycloidal and polynomial profiles are almost indistinguishable from each other, while the double harmonic has the steepest slope. Since the polynomial function has the lowest peak velocity and acceleration, and comparable jerk, we selected it to design the cam profile. Once the SVAJ functions have been defined, the next step is to size the cam. This size is affected by the pressure angle and the radius of curvature of the cam. We followed standard cam design practices milled the profile from 0.5 inch thick 6061 aluminum. We employed 4140 “chrome-moly” steel for the shafts. For a 1 inch (2.5 cm) diameter shaft of 4140, the resulting deflection with 1472 lbs (887.7 kg) of load (4*368) is 1.68×10^{-5} inches (4.3×10^{-5} cm). This deflection will not be noticeable for this application. For the same load and a 1 inch diameter circle, the resulting stress is 1874.2 psi (12.9 MPa), which is over 32 times smaller than the yield strength of 60,200 psi.

Figure 5 shows a Solidworks model of the treadmill cam system with the roller followers that are bolted to the

treadmills. Both cam assemblies are identical with the exception of the orientation of the camshaft. Both non-threaded ends must be facing towards the outside of the treadmill in order to couple with the gear reducer and motors. Figure 6 shows a cross-sectional view of a single cam subassembly highlighting the bearing pre-load design. The bearing roller blocks were machined with a press fit for the taper roller bearing cup and a shoulder to prevent axial motion of the cups. The pre-load on the pair of taper roller bearing cones is created by the clamp collar on the motor coupling side of the camshaft and the nut on the opposite threaded end. The position of the nut on the shaft will determine the force on the bearings. The nut was positioned so that the cam was able to spin freely without any lateral motion. A lock nut (not shown in the figure) was added to prevent the pre-load nut from moving. Finally, Figure 7 shows both machined cam assemblies prior to final assembly.

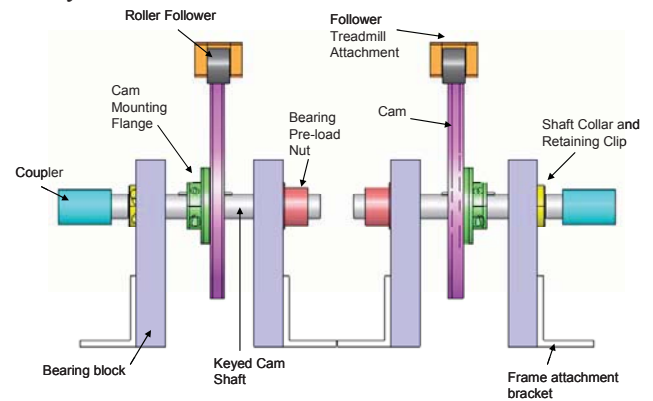


Figure 5. Cam system for treadmill actuation.

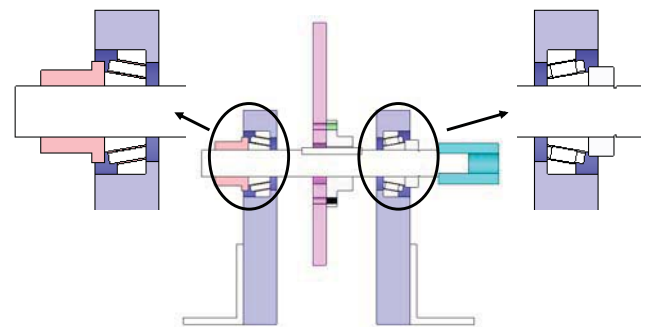


Figure 6. Detail of bearing pre-load design.

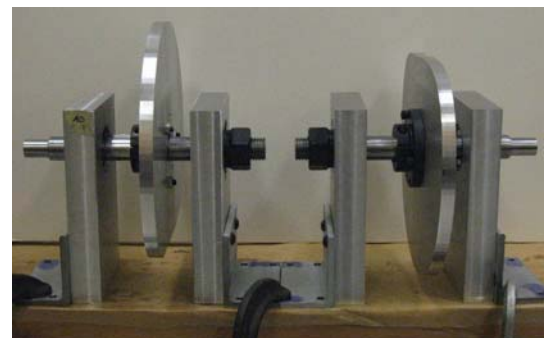


Figure 7. Assembled cam subsystem.

Body- Weight Support

Many patients are not able to support their weight on the impaired leg(s) or they may need assistance maintaining balance. A body weight support (BWS) system needs to provide enough support to unload up to 100% of the patient's weight and keep the patient safe from falls while not interfering with the required ranges of motion. Using an upper limit of a 350 lbs (158.8 kg) patient, and a factor of safety of 3 for fall prevention, we designed a BWS capable of withstanding at least 1050 lbs (476.4 kg). Our BWS allows +/- 2 in (5 cm) of vertical support of the patient's center of gravity to permit normal gait. A study of 25 healthy men and 25 healthy women measured the average vertical displacement as 1.46 +/- 0.35 in (3.7 +/- 0.9 cm) and 1.06 +/- 0.24 in (2.7 +/- 0.6 cm) respectively [35]. Pelvic tilt is also important to maintaining balance and about 5° is required on average. In addition, rotation about the vertical axis of +/- 4° is important for the swing phase and advancing the foot for the next step. This structure must fit around the treadmill assembly which is 24 inches (61 cm) wide, but still be narrow enough to fit through a standard door frame which is 35 inches (89 cm) wide and 83 inches (211 cm) tall so it can be moved easily.

Most of the existing devices employ overhead full-body harnesses, but this has many disadvantages. It greatly increases the height of the device, the harness requires significant don-on and don-off time, and the harness is often very uncomfortable due to the support straps digging into the skin. We employed a different approach as used in Elvis-the-Pelvis (our pelvis robot) which had very positive user feedback. The subject is supported from below the waist with a bicycle seat and the upper body is stabilized with a simple chest harness [31].

F. Assembled Alpha-Prototype

Figures 8 and 9 show the full system as a Solidworks model and as-built, respectively. This configuration was used in both the healthy human subject and passive mannequin testing which will be presented next.

III. ALPHA-PROTOTYPE INITIAL TESTING

We completed an initial evaluation of this alpha-prototype with a small set of four (4) healthy young subjects walking on the device both as a regular treadmill (no actuation of the cams or BWS) and with the BWS and the motors driving the cams. Electromyography (EMG) from leg muscles was collected to measure the muscle activation. This testing was approved by the MIT Committee on the Use of Humans as Experimental Subjects (COUHES) and all subjects gave informed consent. We also tested the system with a passive mannequin.

For the healthy human subjects, surface EMG were recorded during normal treadmill walking (cam system not actuated and subject not using the BWS) and, additionally, when the subject upon being asked relaxed his legs on the BWS with an ankle brace to prevent drop-foot and allow the treadmill to provide the necessary swing clearance. We collected EMG on four muscles using the 16 channel surface

electrode Myomonitor IV wireless datalogger from Delsys (Boston, MA), namely:

- The tibialis anterior (TA) which exhibits peak EMG activity at heel-strike when the foot is dorsiflexed, and no activity during midstance and toe-off.
- The soleus which is located on the lower leg and is activated during foot plantar flexion. McGowan et al. found that while both the soleus and gastrocnemius contribute to body support, the soleus is the primary contributor to forward propulsion [32].
- The rectus femoris which is one of four muscles that make up the quadriceps femoris which is also comprised of the vastus lateralis, vastus medialis, and vastus intermedius. Its highest activation occurs at heel strike.
- The semitendinosus which is part of the hamstrings which also includes the semimembranosus and the biceps femoris. This muscle is located on the back of the leg above the knee and is activated during stance phase and exhibits a triphasic pattern with three peaks. The first peak occurs at heel-strike, the second at 50% of the cycle, and the third at about 90% of the cycle.

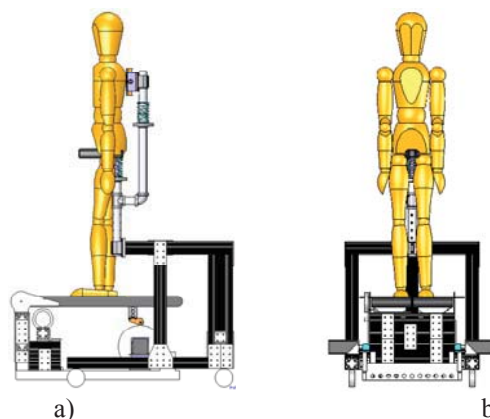


Figure 8. Model of full system. a) side view, b) front view

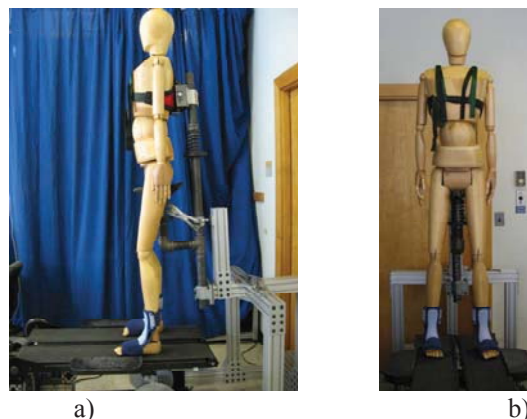


Figure 9. Complete system, a) side view, b) front view.

Figure 10 in the left column shows the raw traces of the EMG signal from the four muscles during normal treadmill walking. For the TA, during normal treadmill walking the peak activation occurs at heel-strike and during stance with little activation during the swing phase. As expected for normal healthy gait, the soleus was activated at toe-off to

provide propulsion, the rectus femoris shows a biphasic pattern and the semitendinosus a triphasic pattern. This panel will serve as a baseline to compare the EMG activity during treadmill actuation with BWS. An ankle brace was secured to the subjects so that they could relax their leg without experiencing foot drop and allow their leg to swing freely. The right column shows the EMG signal from the TA which is very similar to normal treadmill walking. Peak activation was observed at heel-strike and activation was present throughout the stance phase. The soleus and rectus femoris showed almost no activity. Since the soleus is responsible for forward propulsion, it is not surprising that its activity is greatly decreased when gravity and the treadmill actuation are facilitating the swing phase. As patients improve, we will challenge them by controlling the treadmill and cam speeds and requiring them to assist further (for example: increase self-generated propulsion).

To demonstrate that MIT-Skywalker takes full advantage of the dynamics (contrary to all other kinematics-based devices) and that it is capable of training gait to a totally passive “subject,” we tested it with a passive mannequin. Our mannequin, nicknamed Pinocchio, is made of wood, 5’10” (177.8 cm) tall, weighs approximately 20 lbs (9.1 kg), and is shown in the device in Figure 11 and with goniometers during treadmill actuation (Figure 12).

For the mannequin and the treadmill running at about 1 ft/sec (0.3 m/s), the observed hip angle at heel-strike was 5° and at toe-off 0°. This range of movement is much smaller than human gait and it was due primarily to the slow speed and to interference of the bicycle seat assembly which prevented the mannequin’s hip from moving the hip beyond 0°.

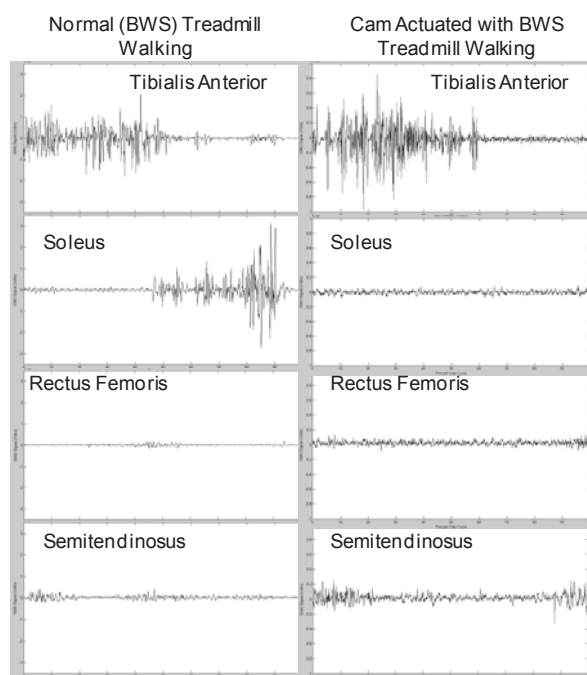


Figure 10. EMG for healthy subjects during treadmill normal walking and during relaxed state.

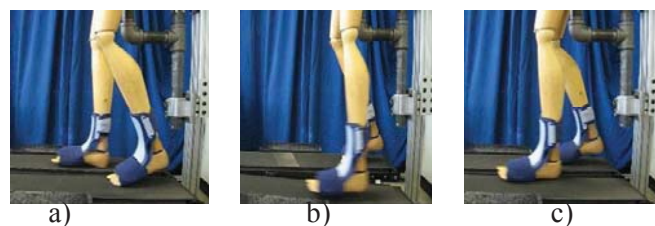


Figure 11. Mannequin test, a) at toe-off, b) swing phase, c) heel-strike.

IV. CONCLUSIONS

This manuscript presents the concept, basic design, and initial testing of the alpha-prototype of the MIT-Skywalker, a novel rehabilitation robotic device to train gait. To our knowledge, it is the only gait robot that takes advantage of human dynamics. While all other gait robotic devices focus exclusively on the kinematics, the MIT-Skywalker facilitates ecological gait training including heel strike. It affords integration with our Anklebot and pelvis robot (Elvis-the-Pelvis), thereby allowing entire lower body training.

Of course further testing will be required with mannequin, healthy subjects, and patients with both unilateral and bilateral impairments. However this initial test of the alpha-prototype with the mannequin and healthy subjects has demonstrated its feasibility and the potential of this unique rehabilitation robotic device.

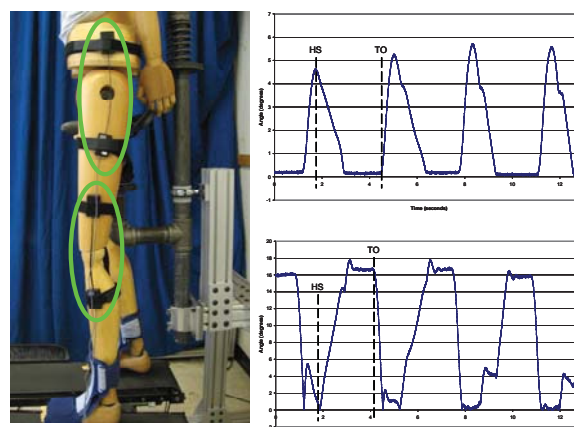


Figure 12. Mannequin goniometer placement, hip and knee angular displacement

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