Passive Orthosis Linkage for Locomotor Rehabilitation

Peter Berkelman, Peter Rossi, Timothy Lu, and Ji Ma

Abstract—Foot drop is a condition which results from injury to the central nervous system (stroke and head injury) or injuries to the peroneal nerve and is characterized by insufficient dorsiflexion of the ankle during the swing phase of walking gaits, resulting in dragging or slapping of the forefoot or toe during walking. We propose a novel type of ankle-foot orthosis based on a passive four-bar linkage for the rehabilitation, prevention, and treatment of the foot-drop disorder.

The linkage is attached to the calf and foot of the patient and passively couples together the motion of the knee and ankle during locomotion. As the knee is bent at the beginning of the stride phase of a walking gait, a curved bar contacts the back of the thigh and the linkage and spring mechanism produces a force to assist the flexion of the ankle at the initiation of the stride phase. An initial wearable prototype has been fabricated and is ready for testing on test subjects to demonstrate the feasibility of the concept and refine the design of the mechanism.

I. INTRODUCTION

Due to the aging demographics of the United States population, there is an increasing need for physical therapy and rehabilitation for disabled patients including stroke and injury victims. Many of these patients may require extensive therapy, yet the availability of one-on-one physical therapy is limited.

Robotic rehabilitation devices have the potential to enhance the impact of physical rehabilitation due to their potential greater availability to patients. Robotic rehabilitation to address locomotor disability would have a particularly significant impact. Accordingly, active robotic ankle-foot orthoses are currently an active area of research and development.

A major cause of gait abnormality is the condition referred to as ‘foot-drop’. In the normal walking gait, the front of each foot must be lifted to allow the leg to stride forward during the swing phase of gait. Patients with muscle paresis in dorsiflexion or spasticity in plantarflexion often have difficulty raising the foot from the ankle while stepping forward. Gait compensation for insufficient dorsiflexion leads to ‘toe drag’ at the beginning of the stride as the foot is lifted or ‘foot slap’ as the foot is brought down prematurely at the end of the stride. Patients often overcompensate by leaning to the contralateral side and hip hiking while circumducting the leg forward leading to imbalance. The foot drop condition thus leads to greatly reduced walking speeds, excessive fatigue, reduced mobility, and poor balance.

This work was supported by the University of Hawaii College of Engineering.

Peter Berkelman, Timothy Lu, and Ji Ma are with the Department of Mechanical Engineering, University of Hawaii-Manoa, 2540 Dole Street, Holmes Hall 202, Honolulu, HI 96822. peterb@hawaii.edu

Peter Rossi is with the Department of Neurology, UCLA, 710 Westwood Plaza, Los Angeles CA 90095. prossi@mednet.ucla.edu

The ‘foot-drop’ gait abnormality is often present in victims of central or peripheral nerve disorders. Rigid ankle braces may be used to restrain the foot drop motion, however these braces also result in an unnatural gait. The condition may improve through patient motor adaptation, but rehabilitation by manual assistance is labor intensive and requires a skilled therapist. Flexible ankle-foot orthoses attempt to correct this condition by supporting the foot and guiding its motion while walking with the aid of springs or other elastic elements incorporated in the orthosis, however these flexible AFOs may not provide appropriately timed assistive forces to the ankle dorsiflexion.

Standard ankle-foot orthoses [AFOs] for foot drop consist of a single moulded piece of material which encloses and immobilizes the ankle. Although this approach does prevent foot drop during the leg swing phase, it also restricts free motion of the foot throughout the entire gait cycle, resulting in an unnatural gait. As the ankle flexion is not stimulated or exercised at any time, the use of rigid fixed AFOs in rehabilitation is limited. Active AFOs use an actuator to assist the dorsiflexion of the ankle only as required, during the swing phase of the leg during each gait cycle. The advantage of active AFOs is that the ankle is free to move normally while the actuator is not active, however a large, heavy actuator is typically required to generate the necessary forces, which can be cumbersome to the disabled user.

To eliminate the bulk of a large actuator on an active AFO, we propose an AFO in which the assistive force provided to the ankle dorsiflexion is produced by a linkage mechanism which couples the ankle flexion to knee flexion as the knee is flexed during the swing phase of the walking gait, without restricting the free motion of the ankle at other times.

II. BACKGROUND

Powered, active ankle-foot orthoses to correct gait abnormalities are currently an active area of development. An active orthosis may address shortcomings of fixed braces and passive flexible orthoses. The dynamic behavior of the active orthosis can be digitally controlled by computer and its motion response data can be recorded to be used for patient evaluation and diagnosis.

A. Active Ankle-Foot Orthoses

Yamamoto et al developed an experimental ankle-foot orthosis to easily adjust the assistance moments and angles [1]. It was found that assistance during plantarflexion was generally unnecessary. A subsequent ankle-foot orthosis with dorsiflexion assistance controlled by a spring was then developed, with favorable clinical evaluations [2].
Ferris et al at the University of Michigan have developed an ankle-foot orthosis powered by an antagonistic pair of artificial pneumatic muscles [3], [4] each myoelectrically controlled by the leg muscles of the patient. The artificial muscle forces are adjusted to be proportional to low-pass filtered signals obtained from electromyography of the tibialis anterior and soleus muscles of the patient. Each pneumatic artificial muscle [5] consists of latex tubing enclosed by a woven textile sheath so that increasing the internal pneumatic pressure causes the artificial muscle to shorten. The overall device is lightweight with a moulded carbon fiber shell to conform to the calf and foot of the patient. The active orthosis has also been used with a footswitch and a pushbutton control to initiate plantarflexion [6] instead of the myoelectric sensors.

A two degree-of-freedom (DOF) ankle-foot orthosis is currently in development at the University of Delaware [7]. The first degree of freedom is the dorsiflexion/plantarflexion angle of the foot and is actively controlled using a DC servomotor. The second degree of freedom is the inversion/eversion rotation angle of the foot. The 2nd DOF of the orthosis is passive and its dynamic response is determined by a torsion spring and a damper built into the mechanism of the orthosis. The structure of this orthosis consists of an articulated frame around the ankle attached to braces for the shank and foot of the patient. Aluminum blocks with faces machined to specified angles are used to correctly locate and orient the dorsiflexion/plantarflexion and inversion/eversion rotation axes of the foot.

The “Rutgers Ankle” consists of a 6-DOF pneumatically-actuated Stewart platform which is fixed to the foot of the patient [8]. The device can be used with a virtual environment simulation for ankle and/or leg rehabilitation for walking and has been evaluated in case studies [9]. Two platforms have been used together with a virtual environment to compose a system for walking simulation [10].

Another active ankle-foot orthosis has been developed at the Massachusetts Institute of Technology. This orthosis uses a brushless motor actuator in series with a variable impedance spring, which allows the stiffness of the ankle-foot orthosis joint to be actively adjusted throughout each gait cycle [11]. Furthermore, the motor is isolated by the spring from physical shocks due to foot impacts. An adaptive variable-impedance control strategy using a finite state machine has been implemented and experimentally compared to constant stiffness joint control for two drop-foot patients wearing the orthosis. The use of a series elastic actuator to control stiffness is described in [12] by Pratt et al. The inversion/eversion motion is rigidly constrained by the orthosis and the total mass is 2.6 Kg. The developers report positive feedback from the users of the orthosis.

The active ankle-foot orthoses that have previously been developed have shown the feasibility and potential benefits of robotic rehabilitation devices for foot-drop gait abnormalities. Further improvements in device design would address deficiencies in the weight, bulk, and complexity of current devices, however.

B. AFO Actuation

It is mechanically difficult to generate the forces and motions of the human ankle joint during walking with a lightweight actuator due to the high torques required and the shocks of foot-ground impacts. The artificial pneumatic muscles used for actuation in the University of Michigan orthosis have several advantages over conventional electric motor actuators, such as high force capabilities in a lightweight actuator without any transmission or gear reduction needed, and passive compliance. However its disadvantages are that an external source of air pressure is needed, the force exerted by the actuator is not independent of its length, the response of the actuator is delayed by the inflation time of the inner tube, and the actuator tube may easily rupture due to wear.

The MIT orthosis uses a compliant spring in series with a motor, which is referred to as a ‘series elastic actuator’. A robotic tendon using a "Jack Spring" (TM) has been proposed as a means of storing energy and reducing the required peak motor power during ankle dorsiflexion [13]. The "Jack Spring" is also an actuator in series with an elastic element with variable stiffness which can be adjusted quickly during operation by a lead screw.

A novel type of orthosis for foot drop is described by Zancan et al in [14], consisting of a harness and sling of straps fastened around the foot, knee, waist, and opposite shoulder of the patient. The straps are arranged so that the motion of the affected lower limb and the opposite shoulder of the patient are linked together, producing a lifting force on the forefoot as the leg is swept forward. A case study was done on a single patient to compare walking gaits while wearing the sling, wearing a standard flexible AFO, or neither. Of the three cases, the patient gait while wearing the sling was closest to symmetrical between the affected and unaffected leg, the walking speed was highest, and only the sling produced the desired result of heel-strike before forefoot contact after swinging the leg forward.

In [15] the authors present an above-knee prosthesis mechanism based on a mechanical linkage rather than a single fixed knee pivot point. A six-bar linkage mechanism is used which couples the angle trajectories of the prosthetic knee and ankle, producing a more natural gait. Kinematic and dynamic analysis was performed to optimize the linkage design parameters.

III. PASSIVE LINKAGE ORTHOSIS DESIGN

The proposed active ankle-foot orthosis is designed to be portable, reliable, safe, and easy to use so that it could be used by patients at home without direct supervision by physical therapists. A major design priority is to minimize the total mass of the device so that the patient can walk naturally without additional effort to while wearing the device attached to the calf and foot. Furthermore, it is preferable to have a device which is non-obtrusive, such as a device which could potentially be worn underneath loose clothing.
A. Concept

Our initial prototype design for an ankle-foot orthosis consists of a four-bar linkage to couple together the flexion of the knee and ankle, providing an assistive force to lift the foot as it is brought forward during the swing phase of walking gaits. The four-bar linkage consists of 4 rigid links joined together at pivoting joints. The linkage can be fabricated from composite material tubing to minimize mass, resulting in a total mass of approximately 200 g for the linkage.

The typical hip, knee, and ankle joint angle trajectories during a single stride of a normal walking gait are shown in Fig. 1. In the foot-drop pathology, the patient has difficulty to flex the ankle at the initiation of the stride phase of the gait, indicated by the dotted line in the figure. This stage of the gait cycle corresponds to an increasing knee angle.

In the proposed orthosis linkage, contact between the linkage and the back of the thigh is initiated when the knee is flexed past a given angle between 5 and 20 degrees. This contact force produces a lifting force on the ankle through the action of the four-bar linkage. The assistive force is generated only when needed. If the knee is not flexed or when the ankle is sufficiently flexed without assistance, no force is produced. This concept is similar to the ankle-foot orthosis made from wearable straps as described in [14], where the natural motion of the wearer’s body produces a force to assist the flexion of the ankle during walking. The knee angle to engage the ankle flexion assistance and the degree of assistance are adjustable to the needs of each user, however the device does require the knee of the wearer to be flexed to some degree to lift the foot as it strides forward.

B. Device Prototypes

Prototypes of the linkage foot-drop orthosis mechanism have been fabricated. The links consist of lightweight aluminum bars, where one link is attached to the calf and one link is attached to the foot of the wearer by hook-and-loop straps. The initial linkage is pictured by itself in Fig. 2 and worn in use on the leg in Figs. 3 and 4.

After initial use testing, the first basic prototype was modified by substituting a pair of springs for the fixed connection between the thigh link and the top link of the four-bar linkage in front of the knee. The connecting springs are shown in Fig. 5. The added compliance in the linkage provided by the springs produces a more gradual degree of assistance to the flexion of the ankle during locomotion and prevents the sudden impacts of the thigh link against the thigh of the wearer which may occur while wearing the linkage orthosis as pictured in Figs. 2 through 4. The additional compliance in the linkage also greatly improves the comfort of the wearer.

When the knee of the wearer is unflexed and the linkage is in the rest position, one of the two springs is prestressed and the other is in full compression to prevent unconstrained motion of the thigh link during locomotion. Without the prestressed spring, the thigh link would tend to bounce and oscillate in an uncontrolled manner as the knee is straightened during walking gaits. The elasticity of the two
springs and their attachment points may be changed easily by hand to adjust the timing and degree of ankle flexion assistance according to the needs of the wearer. The mass of the current prototype is 625 g.

C. Kinematic Synthesis

It is possible to design the kinematic behavior of a four-bar linkage to approximate a given function relating two of the joint angles of the linkage. In Fig. 6, a four-bar linkage is shown schematically with link lengths \( a, b, c, d \) and two adjacent joint angles \( \theta \) and \( \phi \). In the linkage orthosis, the length \( d \) corresponds to the distance between the knee and ankle joints and the remaining linkage lengths \( a, b, c \) may be chosen to satisfy a desired relationship between the knee angle \( \theta \) and ankle angle \( \phi \) during the walking gait. Link \( a \) contacts the back of the thigh during walking gaits and link \( c \) is attached to the foot so that the angle \( \theta \) corresponds to the knee angle plus a constant and the angle \( \phi \) corresponds to the ankle angle plus a constant.

Freudenstein’s method [16], [17] is a set of equations which can be used to calculate the link lengths of a four-bar linkage to satisfy a set of requirements expressed as pairs of joint angles. The basis of Freudenstein’s method is as follows:

With the link lengths of a four bar linkage \( a, b, c, d \) as shown, \( K_1, K_2, K_3 \) can be defined as follows:

\[
K_1 = \frac{d}{a}, \quad K_2 = \frac{d}{c}, \quad K_3 = \frac{a^2 - b^2 + c^2 + d^2}{2ac}.
\]

The joint angles \( \theta \) and \( \phi \) can then be related as follows from the linkage geometry:

\[
K_1 \cos(\phi) - K_2 \cos(\theta) + K_3 = \cos(\theta - \phi).
\]

Then, given a set of three desired joint pairs \( (\theta_1, \phi_1), (\theta_2, \phi_2), (\theta_3, \phi_3) \) for the linkage, three constraint equations can be written:

\[
\begin{bmatrix}
\cos(\phi_1) & -\cos(\theta_1) & 1 \\
\cos(\phi_2) & -\cos(\theta_2) & 1 \\
\cos(\phi_3) & -\cos(\theta_3) & 1
\end{bmatrix}
\begin{bmatrix}
K_1 \\
K_2 \\
K_3
\end{bmatrix} =
\begin{bmatrix}
\cos(\theta_1 - \phi_1) \\
\cos(\theta_2 - \phi_2) \\
\cos(\theta_3 - \phi_3)
\end{bmatrix}.
\]

This matrix equation can be solved for \( K_1, K_2, K_3 \), which can then be used to calculate the link lengths \( a, b, c \) required to satisfy the \( (\theta_1, \phi_1), (\theta_2, \phi_2), (\theta_3, \phi_3) \) constraints on the linkage angle trajectories when \( d \) is given by the knee-to-ankle distance of the user.

IV. EXPERIMENTAL RESULTS

Experiments were performed first to measure the ankle flexion assistance torque as a function of the angular deflection between the link which contacts the thigh of the wearer and the link attached to the calf, then to demonstrate
Ankle flexion assistance torque and link deflection angle relationship

\[ y = 0.011x + 1.3 \]
\[ y = -0.00018x^2 + 0.019x + 1.3 \]

the function of the linkage while being worn by test subjects during locomotion. The current modified linkage with springs to provide compliance in the assisting link as shown in Fig. 5 was used for all experiments shown.

Due to the geometry of the orthosis linkage and the configuration of the two springs, the ankle flexion assistance torque is not proportional to the deflection angle of the link as it contacts the thigh. A series of torque and angle measurements were taken to show this relationship; the data are shown in Fig. 7. The actual relationship is a trigonometric function of the linkage parameters, but a quadratic least squares curve fit follows the experimental data closely.

To demonstrate the function of the current prototype orthosis linkage, preliminary trials were undertaken with unimpaired wearers and using an optical position sensing device\(^1\) to track the motions of the links of the orthosis as the wearer walked across the sensing range.

To track the motions of the individual links of the orthosis prototype, infrared LED markers were attached to linkage at the pivot joints. The LEDs are detected by 3 directional sensors in the motion tracker, which calculates the motions of the LEDs in 3D by triangulation at a given sample rate. The dimensions of the prototype orthosis linkage and the positions of the 5 motion markers used are shown in Fig. 8.

During the trial shown here, the subject walked with a normal gait through the sensor volume range while wearing the linkage orthosis. The approximate 2.0 meter range of the sensor was traversed in 3.5 seconds while completing approximately 1.5 stride cycles, for a walking speed of 0.67 m/sec and stride frequency of 0.43 Hz, a period of 2.33 Hz, and a stride length of 1.33 m. The motion of the linkage was recorded at a 30 Hz data sample rate, with 107 data samples taken during the 3.5 seconds.

The horizontal and vertical motions of the individual markers are shown in Figs. 9 and 10 against time. Using these motion data, the leg motion can be calculated and pictured as shown in Fig. 11, in which the sampling rate has been reduced to 6 Hz for clarity by plotting only the first of every 5 data samples. The locomotion shown is from right to left.

The joint angles of the linkage and the knee and ankle joint angles of the wearer can be calculated from the recorded LED positions by vector analysis. During this trial, the ankle dorsiflexion assistance provided by the linkage is demonstrated by the angle plots of Figs. 12, 13, and 14, and the corresponding ankle flexion assistance torque shown in Fig. 15. The assistance torque shown in Fig. 15 was calculated by combining the recorded knee angle data with the quadratic assistance torque data from Fig. 7. The assistance torque provided by the linkage prevents plantarflexion of the ankle as the knee is flexed and the leg is extended forward during the swing phase of the walking gait.

The assist link angle follows the knee angle when the knee flexion angle surpasses a threshold of approximately 5 degrees, as determined by the linkage dimensions, and a bar on the link contacts the back of the thigh of the wearer. In this way the dorsiflexion of the ankle is coupled to the knee flexion through the compliance of the springs in the linkage. The assist link angle is related to the assistance torque as shown in Fig. 7. As the springs attached to the linkage are prestressed, the linkage assistance compliance

\(^1\)OptoTrak Certus, Northern Digital Inc.
behavior is only for assistance torques over 1.3 N·m; when the assistance torque is below 1.3 N·m it cannot be directly measured with our current prototype. Direct measurement of the ankle torque would require a torque sensing load cell in the structure of the linkage.

The initiation of the swing phase of the gait cycle is not directly detected from the tracking data figures, however it can be inferred that the swing phases of the leg are from approximately 0.5 to 1.5 seconds and from 2.8 seconds onward as shown. The direction of the ankle joint angles in Fig. 13 and Fig. 1 as shown are measured in opposite directions.

The timing and degree of the ankle dorsiflexion assistance may be adjusted by changing the lengths of the orthosis links, and the stiffness and attachment points of the springs. It is also necessary to adjust the linkage to the ankle to knee joint distances of different individual wearers.

V. FURTHER WORK

The initial prototype orthosis linkage design described here is completely passive and is intended to demonstrate the feasibility of the linkage approach against the foot-drop condition in locomotion in volunteer test subjects. Successive prototypes of the linkage orthosis will include instrumentation at the joints to eliminate the need for infrared position markers and the optical localizer, which is limited to a 2.0 m sensing range. The linkage joints will be instrumented with angle resolvers and torque sensors, so that the forces and motions of the wearer’s leg can be monitored in real time and recorded directly during trials. Miniature motors may also be used in the linkage to modify the link lengths and joint angles of the device during operation to adapt to the individual locomotion characteristics of the wearer. Any motors to be used would only be to modify the kinematics of the linkage and not to directly provide locomotion assistance torques, which would require much larger and faster motors instead. The addition of compact, lightweight motors in the device would enable adaptation of the linkage kinematics during actual use, through a feedback loop under real-time computer control. Lightweight composite materials can be used to reduce the mass of the orthosis, and the lengths of the two shortest links in the linkage could be reduced to make the orthosis less obtrusive and easier to wear without affecting the ankle flexion assistance function.

Further validation and refinement of the device will be done through trials with disabled patients.

VI. CONCLUSIONS

A four-bar linkage prototype used as an orthosis has been designed and fabricated to treat the foot-drop pathology in locomotion in stroke or accident victims. The function of the linkage orthosis has been demonstrated with the use of a motion tracking device. The linkage device is passive and couples the joint motion of the ankle to that of the knee so that dorsiflexion of the ankle is assisted at the initiation of the leg stride. The initial device prototype is wearable and suitable for testing and further development.

ACKNOWLEDGMENT

The authors thank the University of Hawaii Department of Mechanical Engineering for its provided support.
REFERENCES


