Test of A Customized Compliant Ankle Rehabilitation Device in Unpowered Mode

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Abstract— **Presented is the design, implementation, and initial gait testing of a lightweight, compliant robotic device for ankle rehabilitation. Many patients with neuromuscular disorders suffer deficits in sensorimotor control of the ankle joint, leading to an abnormal walking pattern. Robotic devices have been used to assist ankle rehabilitation. However, these devices are usually heavy and rigid, which can deviate a natural gait pattern. To address these issues, our team has developed a light weight, compliant ankle robotic device actuated by artificial pneumatic muscles. A total of 3 healthy subjects were recruited to test whether the mechanical structure of the device deviates gait. We used a 3-dimensional (3D) motion analysis system to record and analyze subjects' ankle kinematics during gait while walking barefoot and while wearing the device unpowered. The preliminary results suggest that the device caused some, but minimal changes in ankle kinematics during gait. The changes were mainly caused by the device's rigid footplate, used to support the foot and connect to the pneumatic muscles. The preliminary results will be used for future improvement of the device.**

I. Introduction

Many patients with neuromuscular disorder show deficits in sensorimotor control of the ankle joint, which significantly limit the walking ability and quality of life. For example, patients with stroke demonstrate drop foot during the swing phase of gait due to weakness of the Tibialis Anterior and/or spasticity of the calf muscle [1]. The drop foot reduces ground clearance during initial swing, causing an increased risk of falling. In addition, patients with chronic ankle instability (CAI) walk with an overly inverted ankle due to impaired propioception, which is a position predisposed to ankle sprains, explaining why patients with CAI recurrently sprain the ankle joint [2].

There is a growing interest of using robots to assist ankle rehabilitation, and a number of devices have been developed to achieve this goal. The Anklebot, developed at MIT, is a 2 DOF robot which uses brushless DC motors in a linear actuation system to actuate plantarflexion, dorsiflexion, inversion, and eversion [3]. The device utilizes backdrivable actuation and has amongst the best range of motion for any ankle robot. It can actuate to 45° plantarflexion and 25° dorsiflexion 20° inversion, 20° of eversion, and 15° of external rotation. However, the device is heavier than other ankle exoskeletons, with a mass of 3.6 kg [4].

The bio-inspired soft wearable robotic device for ankle-foot rehabilitation, developed by Carnegie Mellon University and Harvard University, is a robot which utilizes McKibben pneumatic muscles to assist inversion, eversion, plantarflexion and dorsiflexion. The muscles are arranged to emulate the functions of the tibialis anterior, extensor digitorum longus, peroneus tertius and gastrocnemius. A custom strain sensor is used in conjunction with IMUs to measure the deflection of the ankle. The device is quite exciting in its use of compliant materials, making the device lightweight (950 g) and less intrusive to the wearer. However, the device has a relatively low range of motion compared to other exoskeletons, with human subject tests showing 14° dorsiflexion and 13° plantarflexion from rest, while the range of motion for inversion and eversion is unreported [5].

The ankle-foot orthosis powered by artificial pneumatic muscles is a device developed by researchers from University of Michigan, University of Washington, and VA Puget Sound Healthcare System. The device utilizes 2 McKibben pneumatic muscles to provide actuation to -plantarflexion and dorsiflexion [6]. However, these devices do not provide actuation in the frontal plane.

The portable powered ankle-foot orthosis is an untethered device developed at University of Illinois. The device uses a pneumatic rotary actuator to actuate plantarflexion and dorsiflexion, while allowing inversion and eversion passively. While the device only provides 1 DOF, it is completely untethered, making it one of the most promising devices for rehabilitation, or as a daily use device [7]. The device is slightly heavier than other pneumatic exoskeletons (3.1 kg), but that is largely due to having the power and air supply onboard.

From analyzing the previous literature, the authors found that there is a need for further study into a device that can offer actuation to the 4 functional motions plantarflexion, dorsiflexion, inversion, and eversion, while maintaining a light weight to ensure the device does not affect user gait. Therefore, the author's sought to develop a device that can combine the light weight of pneumatic muscles with the full functionality provided by devices with electromechanical actuators.

Presented in this paper is the design and initial testing of a new lightweight robotic ankle orthotic for rehabilitation. Using 3 pneumatic muscles, the device actuates plantarflexion, dorsiflexion, inversion and eversion, while

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allowing ankle abduction and adduction passively. The device has a mass of 1.1 kg, making it comparable in mass to other pneumatic muscle based ankle devices, and much lighter than electromechanically powered ankle exoskeletons. Initial testing of the device's actuation shows that the device can actuate to 12.8° plantarflexion, 15.2° dorsiflexion, 17.5° inversion and 10.3° eversion, all measured from resting position. This paper will discuss the device's design, and an initial study to determine how the mechanical structure of the device will affect ankle kinematics during natural gait.

II. DEVICE DESIGN

A. Device Overview

The device consists of 3 main components: the actuation system, the footplate, and the sensors. A belt houses the heaviest components, including the pneumatic valves and electronics. An off the shelf knee brace (Mueller HG80-Reg) provides the attachment points and mechanical ground for the 3 air muscles. The knee brace was chosen as it provides a comfortable means to anchor the actuators, and is not expected to cause noticeable deviation to user gait. The pneumatic muscles attach from the knee brace to the footplate.

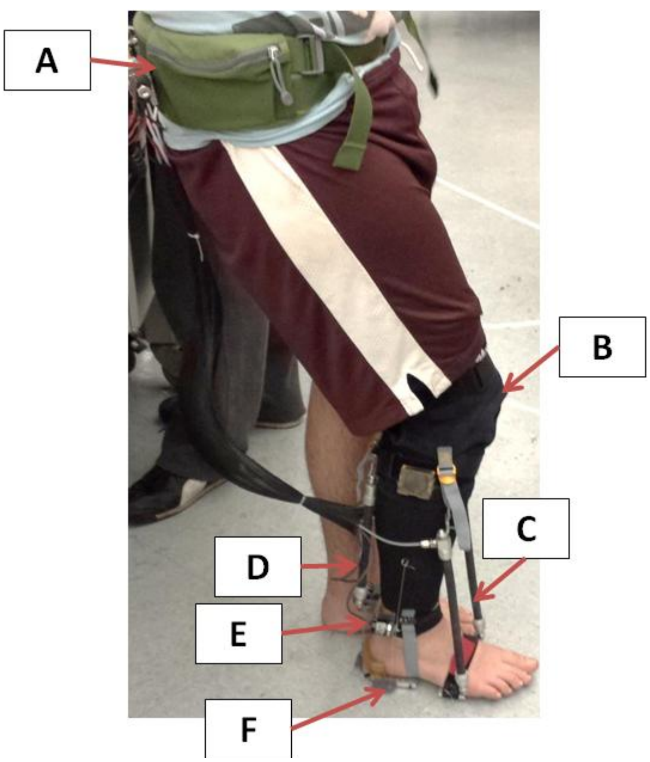

Figure 1: Device overview. A) Belt housing electronics and pneumatic valves. B) Knee brace providing mounting for air muscles. C) Distal air muscles. D) proximal air muscle. E) Encoders mounted at base of the shank. F) Footplate

B. Actuation

The device is actuated by 3 pneumatic muscles. The muscles attach from the knee brace to the footplate. The muscles are attached in the following locations: 2 at the distal end of the footplate, 1 on the lateral side and 1 on the medial side (Festo DMSP-10-230-RM-CM). The 3rd muscle is attached to the heel of the footplate (Festo DMSP-10-160-RM-CM). The layout of the artificial muscles can be seen in Figure 1.

The lateral muscle contracts to actuate eversion, while the medial muscle contracts to actuate inversion. The 2 muscles contract in unison to provide dorsiflexion. A third muscle attaches to the heel of the footplate to provide plantarflexion. The functional movements delivered by each muscle are shown in Figure 2.

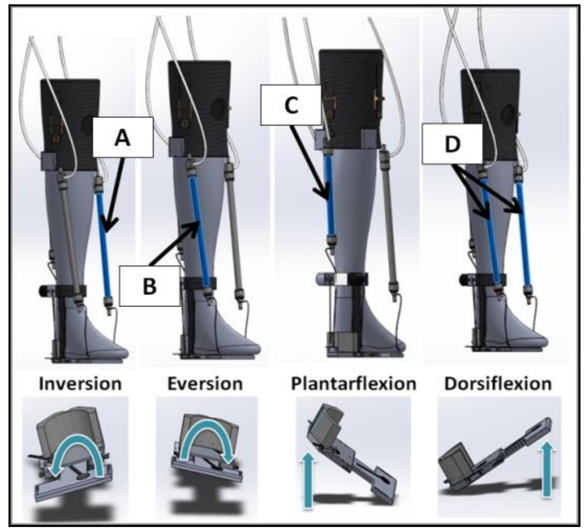

Figure 2: Functional motions delivered by each muscle. A) Distal-medial muscle delivers inversion. B) Distal-lateral muscle delivers eversion. C) Proximal muscle actuates plantarflexion. D) Both distal muscle contract to actuate dorsiflexion

C. Footplate

The footplate plays an important role in this device as it provides the interface between the user and actuator. To maximize the device's ability to deliver the desired force trajectories, a rigid aluminum footplate was used, seen in Figure 3 below. The footplate is adjustable to a wide range of sizes, and contacts the foot from the heel to the metatarsal head, designed to not interfere with the user's ability to extend the toe during terminal stance phase.

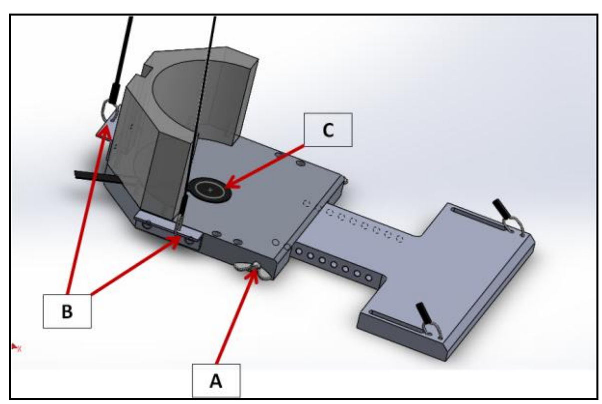

Figure 3:A) Size adjustment B) Muscle and encoder attachment points. C) Force sensor.

D. Sensing

The device is outfitted with rotary encoders to sense the position of foot rotation about the sagittal plane (plantarflexion/dorsiflexion) and the frontal plane (inversion/eversion). To ensure a minimal impact on gait, the encoders are mounted at the base of the shank, connected to the footplate via a tensioned cable and to the knee brace via a retractable cable, ensure that they can sense the foot's full range of motion. A force sensitive resistor (Sparkfun SEN-09375) mounted on the heel of the plate is used as a switch to determine whether the user is in the stance or swing phase of gait, as seen in Figure 3. This particular sensor was chosen due to its extremely low profile, allowing integration of the sensor with virtually no effect on user comfort. The pressure of the muscles is controlled via an electronic pressure regulator (SMC ITV2050).

The device is controlled via a National Instruments PCI-6221 Data Acquistition board (DAQ). With 2 counters (80 Mhz operation), this DAQ board is capable of the necessary position sensing in the device, as well as controlling the pressure regulator via analog voltage output.

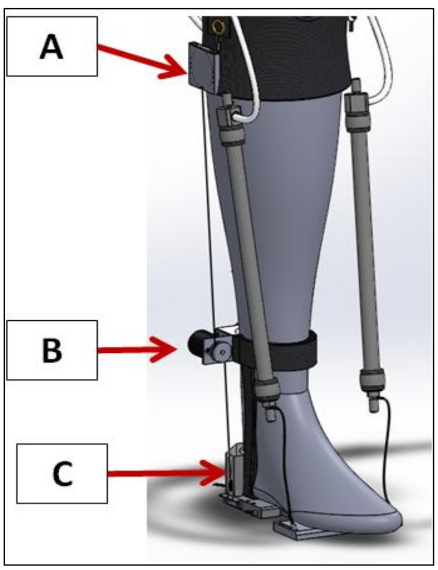

Figure 4: CAD of encoder implementation. A) retractable cable connected at knee brace. B) Encoders mounted at base of shank. C) Tensioned cable connecting encoder to footplate.

III. GAIT STUDY

The device is designed to facilitate motor learning on the ankle control during gait. To achieve this goal, it is important that the mechanical structure of the device itself does not interfere with ankle kinematics during gait. A pilot study was conducted to determine how the mechanical structure of the device affects ankle kinematics during gait in healthy individuals.

A. Protocol

A total of 3 healthy male subjects were recruited and underwent the following two trials: (a) first a control trial walking on a treadmill without wearing the ankle device for approximately 50 cycles of each leg; (b) a device trial walking on a treadmill while wearing the device unpowered for approximately 50 cycles of each leg. Subjects wore the device on the right (dominant) leg. The walking speed for both test trials was set at each subject's comfortable speed while walking without wearing the ankle robotic device.

A 3D motion analysis system (Qualisys AB, Sweden) and visual 3D software (C-motion, MD) were used to record and analyze the ankle joint kinematics during walking. To track/measure the ankle kinematic trajectory using the 3D motion analysis system, shank and foot models were created. The foot model was created based on reflective markers placed on the second and fifth metatarsal heads and medial and lateral malleoli, while the shank model was created based on reflective markers placed on the lateral and medial femoral condyles and medial and lateral malleoli. One additional marker was placed on the heel in order to determine gait events. Heel contacts were determined as the time when the heel marker changes its moving direction from forward to backward. Toe offs were determined as the time when the second toe marker changes its moving direction from backward to forward [8].

B. Results and Discussion

The mean ankle kinematic trajectories of one typical subject (subject 1) are presented in Figures 5 (sagittal plane) and 6 (frontal plane). 0% of the gait phase represents right heel strike, and the stance phase occurs from 0% to approximately 60% of gait. 60% and after represent the swing phase of the right leg during gait.

During the trial, the subjects experienced peak dorsiflexion at the end of the stance phase (50-55%) and peak plantarflexion during the swing phase (60-70%). Peak inversion occurred during the swing phase (60-70%) and peak eversion is experienced during the plant phase (30-40%).

Table 1 below displays the peak values of ankle kinematics seen throughout each subject's mean gait cycle. From analysis of each subjects mean trials, and peak values seen during each trial, a few trends became apparent. In

Table 1: Peak Values of Subject Mean Gait Cycle

	Plantarflexion		Dorsiflexion		Inversion		Eversion	
Subject	Control Trial Peak (°)	Device Trial Peak (°)	Control Trial Peak (°)	Device Trial Peak (°)	Control Trial Peak (°)	Device Trial Peak (°)	Control Trial Peak (°)	Device Trial Peak (°)
1	9.6	11.3	9.1	7.0	15.8	13.7	3.1	7.1
2	3.7	9.2	17.5	14.1	14.7	8.7	8.8	9.9
3	18.6	19.5	5.8	3.5	7.6	5.7	7.4	12.7
Mean	10.7	13.3	10.8	8.2	12.7	9.4	6.4	9.9
St. Dev	7.5	5.5	6.0	5.4	4.5	4.1	3.0	2.8

each subject, the peak values of plantarflexion and eversion increased when the subject wore the device. The peak values of dorsiflexion and inversion decreased when the subject wore the device. This led to two key deductions about the footplate design.

First, in the sagittal plane, the footplate caused each subject to increase plantarflexion during both the stance and swing phase of gait. The deviation during stance phase could result from the rigid foot plate. In normal gait, dorsiflexion during stance phase is facilitated by heel, ankle, and forefoot rockers [9]. As the footplate is rigid and flat, it limits the effect of heel and forefoot rockers, leading to a decrease in dorsiflexion. During the swing phase, the weight of foot plate exerted additional force on the foot towards the ground and thus increase plantarflexion.

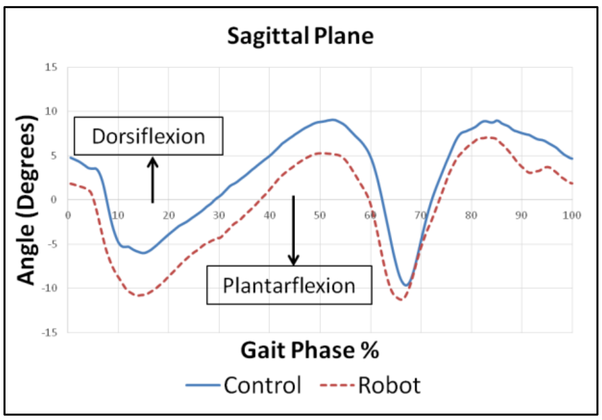

Figure 5. Mean subject ankle rotation in sagittal plane for control and device trial of representative subject

In the frontal plane, the footplate caused each subject to increase eversion during both the stance phase and end of swing phase (before heel contact). At the end of the swing phase, the subject's tendency was to perform heel strike with a flat surface of the footplate, rather than a corner. Therefore, they everted their right foot more than normal to ensure the footplate contacted the floor while horizontal. During the stance phase, because the footplate sat horizontal on the ground, rather than a shoe or bare foot which can rotate slightly, the user everted slightly more than normal to maintain contact between the lateral side of the foot and the footplate.

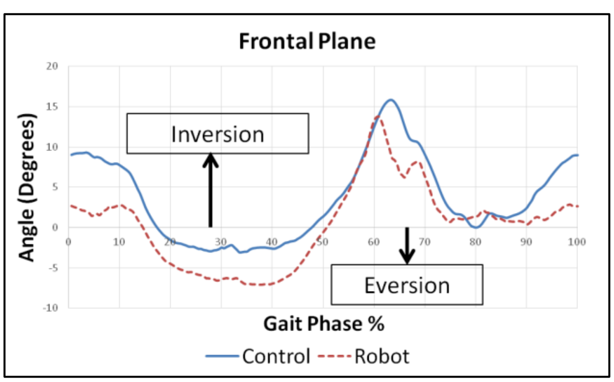

Figure 6: Mean subject ankle rotation in frontal plane for control and device trial of representative subject

CONCLUSIONS AND FUTURE WORK

A 2 DOF actuated ankle rehabilitation device is presented. The device is lightweight, actuating rotation about the sagittal and frontal planes, while allowing passive motion about the transverse plane. Initial testing shows that the mechanical structure causes minor deviations to ankle kinematics during natural gait. Future work includes design alterations to the footplate to further decrease device effect on gait, and implementation of closed loop control algorithms to allow dynamic rehabilitation.

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