Abstract—Stroke often results in hemiplegia, which greatly affects the walking ability of the patients. We propose a multi-functional portable ankle exoskeleton for use in preventing foot-drops, assisting propulsion, and stabilizing inversion/eversion during walking to help gait rehabilitation of stroke patients. The portable ankle exoskeleton was fabricated by 3D printing a soft/rigid hybrid structure. The device was able to prevent foot-drop and assist propulsion with a bi-directional cable-driven actuation system. It also showed a capability of stabilizing inversion/eversion motions using a counter-electromotive force of two small, lightweight gear motors. The device was controlled by a microcontroller based on real-time feedback from one inertial measurement unit and a customized force sensitive resistor. The device is fully untethered with all the components integrated on-board, with a total weight of less than 1 kg. Five healthy subjects performed over-ground walking tests with the proposed ankle exoskeleton for three different walking situations (normal walking, walking with simulated foot-drop, and walking on an uneven terrain) and three walking conditions (without the exoskeleton, with the exoskeleton powered off, and with the exoskeleton powered on). From the test results, we confirmed the feasibility of the proposed ankle exoskeleton for foot-drop prevention, propulsion assistance, and inversion/eversion stabilization. The ankle exoskeleton showed a potential for wearable gait rehabilitation for stroke patients with high mobility and portability.

I. INTRODUCTION

Stroke is one of the most common causes of disability all over the world [1], and places a great burden not only on the individual but also on the society [2]. Stroke often results in hemiplegia, and stroke patients accompanied by hemiplegia experience muscular weakness in their ankle joints, which causes deficits in propulsion during the late stance phase [3], foot-drop during the swing phase [4], and unstable inversion/eversion during the stance phase [5]. These impairments in gait would not only result in deteriorated performances, such as decreased walking speed [6] and increased metabolic cost [7], but also increase risk of falls [8]. The decrease in walking ability significantly affects the quality of life, which makes gait rehabilitation essential and highly beneficial to post-stroke patients. For this reason, a variety of assistive devices have been proposed for gait rehabilitation after stroke. There are currently three general classes of assistive devices to help ankle rehabilitation: unpowered ankle-foot orthosis (AFO), powered platform robotic devices, and powered portable robotic devices.

Unpowered AFOs are most widely used devices for ankle rehabilitation due to their low cost and easy accessibility. For example, Hesse et al. have tested the gait function in hemiparetic patients walking barefoot, with a shoe, and with an AFO, and found that an AFO improved the walking speed, the cadence, and the stride length [9]. De Wit et al. have reported the effect of an AFO on the walking ability in chronic stroke patients, showing improvement of the walking speed during timed-up-and-go tests and stairs tests [10]. Danielsson and Sunnerhagen have found that an AFO increased the walking speed and also decreased the energy consumption during walking in stroke patients [11]. Despite the benefits of AFOs, they often limit the natural degrees of freedom (DOFs) as well as the existing range of the ankle motion, which would result in abnormal gait patterns [12]. Although an AFO with a hinge joint may allow more DOFs [13], it may provide only limited assistance due to its passive design and function.

Powered platform robotic devices transmit forces and augmented motions to the ankle joint when the body is in a stationary position, or when the body weight is supported by the external structure of the platform. For example, Jamwal et al. have designed a robotic ankle orthosis with pneumatic muscle actuators for ankle rehabilitation with the robot operated on predefined trajectories commonly adopted by the therapists [14]. Saglia et al. have developed an actuated parallel mechanism with redundancy for ankle rehabilitation using linear actuators, providing assistive and resistive force for rehabilitation [15]. Freivogel et al. have performed a six-week gait training program with an electromechanical gait device (LokOHelp) and found improvements in the walking ability in terms of lower limb strength and postural control [16]. Žanotto et al. have designed a robotic platform (ALEX III) with 12 DOFs for human gait training [17]. Many other research efforts have also been made for ankle rehabilitation [18]. While powered platform robotic devices have shown promising results, the benefits are still confined to people who have access to those expensive, large and stationary devices.
One possible solution to address the aforementioned challenges and support medical rehabilitation and training of walking is a wearable robotic system designed for foot-drop prevention and propulsion assistance. Park et al. have proposed an active, soft robotic orthosis, weighing less than 1 kg (not including the air source) powered by pneumatic artificial muscles for assisting three-dimensional (3D) ankle motions [19]. Awad et al. have developed a 4 kg portable robotic exosuit made of a textile-based cable-driven system with a waist-mounted actuator and a battery that helped ankle plantarflexion and dorsiflexion for post-stroke patients [20]. Kwon et al. have designed a cable-driven portable robotic orthosis of 1.5 kg in weight, made of soft materials and flexible structures with a waist-mounted controller and a battery for assisting ankle plantarflexion and dorsiflexion [21]. Although these approaches have shown the feasibility of gait rehabilitation with relatively lightweight devices, there has been no portable robotic device that can assist ankle motions in both sagittal and mediolateral planes, such as dorsiflexion/plantarflexion and inversion/eversion, respectively, reported for gait rehabilitation to the best of our knowledge.

Therefore, we propose a multi-functional portable ankle exoskeleton that has three main functions of foot-drop prevention, propulsion assistance, and inversion/eversion stabilization during walking for stroke patients. The device is lightweight (0.98 kg) and fully untethered with all the components integrated on-board, thus easy to wear. Before directly implementing such a portable robotic device to the target clinical population (e.g. stroke patients) for gait rehabilitation, it is beneficial to first test the feasibility with a sample of young, healthy subjects. The objectives of this study are thus to evaluate the feasibility and the efficacy of the proposed ankle exoskeleton for foot-drop prevention, propulsion assistance, and inversion/eversion stabilization.

II. METHODS

A. Soft Ankle Exoskeleton Design

The soft ankle exoskeleton (Fig. 1) is mainly made by 3D printing and consists of four major parts: the body structure, the foot-drop prevention and propulsion assistance module, the inversion/eversion stabilization module, and the control hardware. The body structure contains a foot brace, two shank pads, and two foot-shank connectors. The foot brace and the shank pads were made of soft thermoplastic polyurethane (TPU) by a 3D printer (Cubicon Single Plus, Cubicon). The foot-shank connectors were made of a rigid plastic material (Onyx, Markforged) by a 3D printer (Mark Two, Markforged). Although the material for the foot-shank connectors was rigid, the simultaneous flexibility and rigidity in the sagittal and mediolateral planes, respectively, was achieved by a patterned beam flexure (Fig. 1e). The foot brace and the shank pad were anchored to the foot and the shank with wraps of hook-and-loop fasteners (e.g. Velcro). The foot-shank connector was coupled with the foot brace and the shank pad by a hinge joint and a slide joint, respectively. This structure transfers the rotational ankle motion for inversion/eversion to the linear motion of the foot-shank connector.

The module for foot-drop prevention and propulsion assistance contains a direct-current (DC) servo motor (MX-64T, ROBOTIS), a bi-directional pulley, two Bowden cables, a motor housing, and a rechargeable lithium polymer battery (14.8 V, 2000 mAh). The bi-directional pulley and the motor housing were 3D printed (Object30 Prime, Stratasys) with rigid plastic (VeroBlack Plus, Stratasys). A single motor with the bi-directional pulley (Fig. 1a) was enough to control both foot-drop prevention and propulsion assistance instead of two separate motors, contributing to minimizing the size and the weight of the device [21]. The motor and the housing were mounted on the lateral side of the shank pad to avoid interference with the other leg during walking.

The inversion/eversion stabilization module contains two small DC motors (RA114WGM, DNI), two rack-and-pinions and their housings (Fig. 1b). The rack-and-pinions were made of acrylic by laser cutting (Speedy 300, Trotec). The motor housing was 3D printed (Cubicon Single Plus, Cubicon) with rigid plastic (PLA, Cubicon). The motors and the housings were mounted on both the lateral and the medial side shank pads. The motors were not actuated by electricity to rotate. Instead, they were operated as generators driven by the rack-and-pinions [22]. The rack and the pinion were mounted on the foot-shank connector and the shafts of the motors, respectively. The rack-and-pinion converts the linear motion of the foot-shank connector into a rotational motion of the gearmotor. When the two electric terminals of the motors are not connected (no load), the rack-and-pinions can rotate the motors easily with only small friction in the gearboxes, allowing almost free inversion/eversion. When the two electric terminals of the two motors are connected to each other, the motors work as generators with each other being the load, generating a counter-electromotive force (CEMF), which impedes the motors and slows down the

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Fig. 1. Custom designed soft ankle exoskeleton. (a) Bi-directional cable-driven system for foot-drop prevention and propulsion assistance. (b) Rack-and-pinion to transfer linear motion of foot-shank connector to rotation motion of the gearmotor. (c) Side view of the ankle exoskeleton wearing on a prosthesis. (d) Inversion/eversion stabilization achieved by the CEMF of gearmotors. (e) Foot-shank connector that achieves flexibility in sagittal plane (in black) and rigidity in mediolateral plane (in purple) simultaneously. (f) Custom-designed FSR sole for gait phase detection.
linear motion of the foot-shank connector, eventually restricting ankle inversion/eversion by stiffening (Fig. 1d).

The control module is composed of a microcontroller (MKR-1000, Arduino), an inertial measurement unit (IMU) (EBIMU-9DOFV4, E2BOX), a force sensitive resistor (FSR) sole (Fig. 1f), a metal-oxide-semiconductor field-effect transistor (MOSFET) (TK3R1E04PL, TODHIBA), a rechargeable lithium-ion battery (3.7 V, 500 mAh). The MOSFET operates as an electric switch to control the activation of the CEMF braking mechanism. The total weight of the proposed soft ankle exoskeleton is 0.98 kg including batteries, and the batteries last for three hours of continuous use between charges.

B. Real-Time Gait Detection and Control

The proposed soft ankle exoskeleton required only one IMU and one FSR to detect the gait cycle, which greatly reduced the complexity of both the hardware system and the control algorithm (Fig. 2). The custom-designed FSR sole (Fig. 1f) beneath the foot brace detects the ground reaction force to determine the stance/swing phases and thus control the activation of generative braking. The microcontroller turns on and off the MOSFET to control generative braking of the motors. The IMU mounted on the medial shank calculates the Shank angle in the sagittal plane. The starting time for propulsion assistance is detected from the negative peak of the shank angle, and the starting time for foot-drop prevention is detected from the negative peak of the Shank angle. During the stance phase of walking, the generative braking engages to stabilize the ankle inversion/eversion, and the generative braking is deactivated to allow free motions of inversion/eversion during the swing phase. The propulsion assistance is activated by pulling the posterior cable to lift the heel during the late stance when the foot pushes off the ground, and the foot-drop prevention is activated by pulling the anterior cable to lift the toe during the swing phase right after the toe-off (Fig. 3).

For controlling the servo motor for propulsion assistance and foot-drop prevention, a current-based torque control strategy was employed to directly generate desired pulling forces. This control method ensures the safety in human-robot interactions as it keeps the same output force upon a given command, regardless of sudden position changes in the gait that may occur due to unexpected disturbances. In this study, the maximum torque of 7.3 Nm from the motor was used for both propulsion assistance and foot-drop prevention, and the maximum speed of the motor was 78 rpm.

C. Experimental testing

Over-ground walking experiments were conducted to explore the feasibility of the soft ankle exoskeleton for propulsion assistance, foot-drop prevention, and inversion/eversion stabilization (Fig. 4). Five healthy male subjects (age: 29.4 ± 4.7 years, height: 1.76 ± 0.07 m, weight: 76.7 ± 6.9 kg) participated in this study with informed consent. The device was worn on the dominant limb only. In order to simulate the gait condition that people with stroke have, a spring was used to connect the heel and the Shank to induce foot-drop (Fig. 4b). Low-height obstacles with a round surface were attached to the ground to make an uneven terrain to induce inversion/eversion of the ankle during walking (Fig. 4c). Subjects were asked to walk at a relatively slow speed around 0.7 m/s similar to the average walking speed of stroke patients [23]. Prior to formal data collection, participants practiced walking at a slower speed to become familiar with the device.

Each subject participated testing by walking with three different situations: normal walking, walking with simulated foot-drop, and walking on uneven terrain. For each situation, three different conditions were also applied to the subjects: without the exoskeleton (Bare), with the exoskeleton powered off (Exo Off), and with the exoskeleton powered on (Exo On). Total nine different sets of trials were performed with five repetitions for each trial. The walkway was 5.7 m long with the latter half (2.8 m) section within the motion capture volume, which allowed enough space for the subject to attain a stable gait pattern before entering the motion capture volume. The subjects were allowed to rest if needed between trials. Kinematic data were collected using eight infrared cameras (Prime 13, NaturalPoint Inc.) with a sampling frequency of 100 Hz. 20 reflective markers were placed on the following landmarks on both lower limbs: anterior and posterior superior iliac spine, greater trochanter, medial and lateral epicondyle, medial and lateral malleolus, heel, first metatarsal, and fifth metatarsal. The muscle activation was measured using surface electromyography (EMG) sensors (Trigno™ Avanti, Delsys Inc.) at a sampling frequency of 2000 Hz, with six sensing nodes placed on the
following muscles: gastrocnemius (plantarflexion), tibialis anterior (dorsiflexion), and peroneus longus (inversion/eversion) on both lower limbs. It was expected that the Exo-On condition would bring the ankle kinematics closer to the Bare condition compared to the Exo-Off condition for propulsion assistance and foot-drop prevention. It was also expected that Exo-On condition would have lower activation of the peroneus longus muscle compared to the Bare condition for stabilizing inversion/eversion.

D. Data Analysis

One complete step of the dominant limb in the motion capture volume was analyzed, and thus each condition had five steps for each subject in the data analysis. The raw coordinate data from the markers were filtered using a zero-lag low-pass Butterworth filter with a cut-off frequency of 10 Hz, and a seven-segment model of the lower limb was built to calculate joint kinematics using a biomechanics analysis software tool (Visual3D v6™, C-Motion). The EMG data were rectified and filtered using a zero-lag band-pass filter with a cut-off frequency between 20-350 Hz and processed in MATLAB (Version 2106b, MathWorks). The heel-strike and the toe-off gait events were detected using a previously published method [24]. The lower limb joint angles were computed in the sagittal plane. The walking speed was calculated by the speed of the center of mass of the lower limb in the walking direction. The EMG data were integrated for the stance and the swing phases. Statistical analyses were not performed in this study due to small number of subjects.

III. RESULTS

The average walking speed for all the subjects across all the conditions was $0.72 \pm 0.08$ m/s. For kinematics data, qualitatively, the ankle exoskeleton showed a promising result for preventing foot-drop and assisting propulsion in Subject 1; not significant improvement in Subjects 3 and 5 with Exo_On similar to Bare; and an offset between Exo_On and Bare in the other two subjects (Fig. 5). The subjects exhibited qualitatively different muscle activation during the Bare, Exo_Off and Exo_On conditions for all the walking situations (Fig. 6).

IV. DISCUSSION

This paper presented a multi-functional portable ankle exoskeleton that can assist ankle motions in both sagittal (plantarflexion/dorsiflexion) and mediolateral (inversion/eversion) planes for gait rehabilitation. Kinematics and the EMG data qualitatively showed that the device was effective in carrying out the three target functions. The simulated foot-drop walking particularly showed that the device improved foot-drop with clear increases in dorsiflexion angle from the Bare condition to the Exo_On condition (Subject 1 in Fig. 5), and also increased the activation level of the tibialis anterior muscle (Fig. 6) during the swing phase. For
healthy participants, foot-drop was simulated with a spring connecting the heel and the shank, and the activation of the device increased the isometric contractions, thus increasing the muscle activation. For normal walking, with the ankle exoskeleton powered off, the device acted as a passive AFO in fact and limited the plantarflexion. The Exo_On condition brought the peak plantarflexion close to the Bare condition for Subjects 1, 4 and 5 (Fig. 5), which indicates the device has a benefit of assisting propulsion. Also, the decreased activation level of the gastrocnemius muscle during the stance phase for Subject 1 and 4 (Fig. 6) indicates that the device could help the forward propulsion. It is important to note that the increase of the activation levels of gastrocnemius muscle may be due to the interaction between the lower limb and the ankle exoskeleton, though further research will be necessary to confirm. Our result aligns with the walking test result of the previous portable ankle exoskeleton. In the single-patient over-ground walking study by Kwon et al., the device improved paretic propulsion during the stance phase and alleviated the symptom of foot-drop during the swing phase [21]. In the treadmill walking test with patients by Awad et al., the device increased dorsiflexion angle by 5.3° during the swing phase and improved paretic propulsion by 11% [20]. For uneven terrain walking tests, the Exo_On condition showed a lower activation level of the peroneus longus muscle than the Bare condition in Subjects 3, 4 and 5 (Fig. 6), which may indicate that the device helped inversion/eversion.

The device did not show significant improvement in two subjects (Subject 3 and 5) with the Exo_On condition similar to the Bare condition. One possible reason is that the device did not tightly fit the feet of the two subjects and was too loose to assist the ankle motions. The device showed offsets in the other two subjects (Subject 2 and 4) with the Exo_On condition, which was worse than the normal ankle motion with the Bare condition. This might be because the device was too tight on those two subjects, which disrupted the normal ankle motions. Five subjects had different heights with a large standard deviation while the size of the ankle exoskeleton was fixed. Therefore, future versions should be adjustable for subjects with different heights. Furthermore, since how the device was worn significantly affected the results, further research will be necessary to provide a clear instruction on wearing and also to standardize the calibration process to maximize the benefit of the device.

Unlike previous designs with waist-mounted components, such as an actuator, a controller, and a battery [20],[21], which required long signal and power lines, causing difficulty in wearing, our ankle exoskeleton integrated all the components on-board only on the lower leg, making the device compact and easy to wear. Since it has been reported that the metabolic cost during walking increased with a load mass added especially to more distal location [25], we located most of the weight (the actuator and the battery) at the top of the ankle exoskeleton (Fig. 1c), thus making the weight closer to the center of mass of the wearer. It is important to note that the total weight of our device (0.98 kg) is comparable to the weight located at the lower limb alone of similar portable devices [19]–[21].

One limitation of this study is that a relatively small sample size of the subjects participated in testing, and the results could be strengthened with more subjects in future testing. Additionally, we only tested healthy participants, and the effect of the proposed device for actual stroke patients still remains unknown. However, we believe it is important to test the feasibility first with a sample of young, healthy individuals before moving on to the target clinical population. Another limitation of the study is that we did not measure the effect of the ankle exoskeleton on the metabolic cost, which is another important parameter to be considered in rehabilitation [26]. Further research is needed to explore the performance of our device for different populations and its effect on the metabolic cost. Future work will also include design optimization and development of standard processes for wearing and calibration of the device, as well as investigation on the long-term rehabilitation effect.

![Figure 5: Ankle kinematics for all five subjects. Mean (sagittal plane) ankle joint angles of the dominant leg from initial foot contact (0%) to the subsequent initial foot contact (100%) for walking with normal, simulated foot-drop, and uneven terrain situations. Main differences could be found at the point of peak plantarflexion and 100% stride.](image-url)
Another area of future work will be implementation of soft actuators, such as custom-designed pneumatic artificial muscles [27],[28], instead of electric motors so that the device can easily conform the shape of the wearer’s ankle and leg and allow more freedom in natural motions. This will also facilitate testing of a wide range of subjects since the device can be customized more effectively to different sizes and shapes of the subjects’ bodies.

V. CONCLUSION

This paper presented a portable ankle exoskeleton for foot-drop prevention, propulsion assistance, and inversion/eversion stabilization. Human walking tests were conducted to evaluate the feasibility and the efficacy of the proposed system. The result showed a potential of the proposed device for gait rehabilitation of post-stroke patients, not requiring expensive stationary equipment. Insights of the kinematics of walking and muscle activation also helped establishing the foundation for a more effective and practical gait rehabilitation system design for future implementations.

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