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Design and Evaluation of a Powered Hip Exoskeleton for Frontal and Sagittal Plane Assistance

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***Abstract*— Powered exoskeletons have the potential to improve ambulation for millions of individuals who struggle with mobility. Most powered exoskeletons aim to improve walking economy and increase speed by generating propulsive torque in the sagittal plane. However, individuals with mobility impairments typically have limited mediolateral balance, which requires assistance in the frontal plane. Here we present the design and preliminary evaluation of an autonomous powered hip exoskeleton that can generate torque in both the frontal and sagittal planes. The exoskeleton leverages a unique parallel actuator to produce up to 30 Nm of torque while achieving a compact and lightweight design that adds only 3 cm posterior and 8 cm lateral to the user and weighs only 5.3 kg. Preliminary validation tests with two healthy subjects show that the proposed powered hip exoskeleton can successfully assist gait by controlling the frontal plane torque to alter step width and providing sagittal plane torque to assist with hip flexion. A device with these characteristics has the potential to improve both gait economy and balance in clinical populations.**

I. Introduction

Powered autonomous exoskeletons have shown the ability to improve ambulation in individuals with mobility impairments. For example, hip exoskeletons providing assistive flexion/extension torques have shown increased level-ground self-selected walking speed in stroke survivors [1], [2], and decreased the metabolic cost of transport in above-knee amputees and elderly individuals [3], [4]. Similarly, ankle exoskeletons and exosuits have increased walking speeds in stroke survivors [5], [6] and children with cerebral palsy [7]. These exoskeletons have mainly focused on improving walking economy and speed by providing assistive torques in the sagittal plane. However, individuals with mobility impairments also have limited balance and are at high risk of falling [8], [9].

To maintain anterior-posterior balance, healthy individuals can harness the passive dynamics of walking [10] and exoskeletons can improve on that by providing assistance in the sagittal plane [11], [12]. In contrast, mediolateral balance requires active control strategies that involve controlling joint torques in the frontal plane [10]. Thus, exoskeletons may improve balance in individuals with mobility impairments by assisting the user's lower limbs in the frontal plane.

Healthy individuals use their hip abductor and adductor muscles to regulate balance in the frontal plane [13], [14]. Using these muscles, healthy individuals can control their balance by applying torques to the lower limb during stance to shift their center of mass with respect to their center of pressure [14]. Alternatively, healthy individuals can apply torques to the lower limb during swing to reposition the foot before it contacts the ground, thus increasing the base of support in the subsequent step, which also improves mediolateral balance [14], [15]. However, most people affected by mobility impairments have weakened lower limbs [16]–[18], which limits the effectiveness of these active control strategies, reducing balance and increasing the risk of falls [19], [20]. Based on this analysis, we propose using a powered hip exoskeleton to assist the user's hips in the frontal plane by providing abduction/adduction torques.

To the best of our knowledge, few autonomous exoskeletons can provide hip abduction/adduction torques in addition to assistive hip flexion/extension torques [21]–[25]. The first clinically motivated exoskeleton to show this capability is the MindWalker [21], a 28-kg hip-knee-ankle exoskeleton developed for people with spinal cord injury. This device provided the first demonstration of an exoskeleton that actively modifies the lateral foot placement in response to a perturbation. More recently, a 9.2-kg powered hip-only exoskeleton extended the control scheme of the MindWalker to include admittance-based control [22]. Unfortunately, these exoskeletons are quite heavy and bulky which can increase the metabolic cost of walking [26]. Moreover, the large size of these exoskeletons interferes with the user's ability to perform functional tasks, such as sitting on a normal chair and swinging their arms while walking. These limitations may further exacerbate the metabolic cost of wearing the device [27], decrease recovery after a perturbation [28], and reduce usability in the real world.

To avoid this problem, researchers have developed exoskeletons that assist the user's hip only in the frontal plane [29], [30]. These exoskeletons have shown promising results, including increased measures of balance and decreased metabolic consumption in healthy subjects [30]. However, exoskeletons assisting the user's hip only in the frontal plane cannot help with body propulsion and anterior-posterior balance, which require assistance in the sagittal plane. Thus,

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these exoskeletons address the issues of excessive mass and size at the cost of reduced functionality.

In this paper, we present a lightweight and compact autonomous hip exoskeleton that can assist the user's hip both in the frontal and sagittal plane. This new exoskeleton leverages a parallel kinematic actuation system that assists both hip flexion/extension and abduction/adduction. The device weighs 5.3 kg, adds only 3 cm posteriorly, and 8 cm laterally, and can provide up to 30 Nm of torque during gait. This paper presents the design of the hip exoskeleton and a preliminary evaluation with two healthy subjects. This preliminary evaluation demonstrates the ability of the proposed exoskeleton to provide assistance both in the sagittal and frontal planes and to modify step width. As a result of its compact size and small mass, the proposed powered hip exoskeleton has the potential to improve balance and reduce metabolic cost in individuals with lower-limb impairments in the real world.

II. Powered Hip Exoskeleton

A. Mechanics

The powered exoskeleton incorporates two powered hip actuators (one for each leg) that connect to the user through a pelvis brace and two thigh braces (Fig. 1).

Each powered hip actuator comprises two parallel underactuated five-bar mechanisms. The parallel mechanisms share a grounded pelvis frame, two orthogonal revolute joints, and a distal thigh frame. The pelvis frame and thigh frame connect through two shared revolute joints which accommodate hip motion in the frontal and sagittal planes (Fig. 1 a). Two parallel powered prismatic joints attach between the pelvis frame and the thigh frame through 2-degree-of-freedom (2-DoF) joints. Thus, the two underactuated chains combine such that force generated by the two powered prismatic joints creates torque about the frontal and sagittal revolute joints (Supplementary Video 1). In this system, the powered prismatic joints (i.e. linear actuators) comprise a brushless motor (Maxon), a helical gear stage, and a ball screw. Similar to our prior work, an iterative design framework drives the selection of each dimension of the hip actuator [31].

The hip actuator pelvis frame connects to a compliant pelvis interface (Fig. 1 b, c) worn by the user. Similar to our previous work [32], we built the pelvis interface from a lower spine orthosis (Ottobock) and compliant thermoplastic pads. The pads beneath the orthosis connect to the pelvis frame through a rigid crossbar (Fig. 1 b-d). Similar to our previous work [31], the thigh frame connects to a thigh brace through a self-aligning mechanism (Fig. 1 b, d) which allows for dynamic alignment of the powered joints to the human hip joint center of rotation [33], thus reducing spurious forces and torques between the exoskeleton and the user [34], [35]. The thigh brace comprises a rigid bar which connects to a mesh band and BOA lacing system (Click-Medical) [31].

B. Embedded Electronics and Sensing

The embedded electronic system consists of a high-level motherboard, two low-level boards (one in each powered hip actuator), four motor drivers (one on each linear actuator), and a suite of digital sensors (Fig. 2). A 1200 mAh 8-cell lithium polymer battery powers the exoskeleton.

The high-level motherboard contains a single board computer and a microcontroller. The single-board computer (Raspberry Pi Compute Module 4) calculates desired joint torques and streams data wirelessly to a host computer which runs a graphical user interface (GUI). This interface enables the experimenter to monitor exoskeleton data and modify control parameters. A microcontroller (PIC32) communicates

Fig. 1 (a) The hip actuator can generate flexion/extension and abduction/adduction torques. (b) The realized exoskeleton on a subject. (c-d) The hip actuator is connected to the subject's pelvis through a pelvis wrap with a rigid crossbar and to the subject's thigh through a rigid thigh brace with a flexible cuff. Both the pelvis wrap and thigh cuff can be tightened with a BOA closure system. (e) The high-level electronics and battery are placed posteriorly to the user. Per side, the hip actuator increases the user's lateral dimension by 8 cm. The high-level electronics and battery add 3 cm posterior to the user.

between the single-board computer and the two low-level boards (located in each hip actuator). The pelvis interface (Fig. 1 e) houses the battery and high-level electronics.

Each low-level electronics board contains a microcontroller (PIC32) which communicates over SPI with physical sensors, two motor drivers, and the high-level board. The physical sensors include two 18-bit absolute joint encoders (iC Haus iC-MU with 16 Pole, 1.28 pitch Nonius encoder) and one inertial measurement unit (XSENS MTi-3). The motor drivers (Ingenia Capitan Core) conduct motor commutation using a 12-bit absolute encoder (RLS RM08) and run closed-loop field-oriented current control. One low-level board is located on each exoskeleton frame and one motor-driver and commutation sensor are mounted to each linear actuator.

C. Control

We adapt the hierarchical control structure from our previous research into hip exoskeletons with sagittal plane assistance [31] to control both sagittal and frontal plane joint torques. Specifically, the high-level controller generates a real-time gait phase estimate for each hip actuator based on a gait timer and a gait cadence estimate [36], [37]. The gait timer is reset every cycle by a state machine that identifies peak hip flexion. A mid-level torque planner uses the phase estimate along with experimenter tunable parameters (i.e. duration, timing, and magnitude) to calculate desired gaussian-shaped assistive torque profiles for both the sagittal and frontal plane exoskeleton joints. A low-level controller converts the desired joint torques to motor currents based on a kinematic model.

III. Human Experiments

A. Methods

We tested the performance of the control algorithm and the ability of the unilateral hip exoskeleton to modify step width with two healthy young subjects (subject 1: 22 years, 54 kg, 168 cm; subject 2: 96.4 kg, 25 years, 180 cm) that were familiar with the exoskeleton operation. The University of Utah Institutional Review Board approved the experimental protocol (IRB_00120712). The subjects provided written informed consent before the experiment took place and consented to disseminate pictures and videos of the experiment.

Each subject tested three experimental conditions designed to assess the controller and the relationship between frontal plane assistance and step width. The experimental conditions were transparent mode (i.e., no exoskeleton assistance), and sagittal plane assistance with either positive or negative frontal plane assistance during the swing phase of walking (Supplementary Video 1). We randomized the order of the experimental conditions.

In each experimental condition, subjects walked on a split belt treadmill for one minute while wearing the exoskeleton and reflective markers (Fig. 3). This time period was selected based on pilot studies and prior work [29] which revealed that adaptions in step width occur within a few steps. Reflective markers were placed on the lower-limb segments based on a modified Newington-Helen Hayes gait model (Vicon Nexus 2.12) with two additional markers added to each shank and thigh segment. The exoskeleton was set to transparent mode, or to provide assistive torques. The flexion assistance magnitude was set to 0.32 Nm/kg and peaked after toe off, while the extension assistance magnitude was set to 0.21 Nm/kg and peaked after heel strike (Fig. 4 a). Frontal plane assistance during swing was set to 0.11 Nm/kg and tuned to peak immediately before heel strike to have the largest impact on step width (Fig. 4 b). During each experimental condition, subject kinematics and ground reaction forces were measured by an optical motion capture system (Vicon 3D Motion) and an instrumented treadmill (Bertec).

For each subject and condition separately, we segmented the data into individual strides using the ground reaction force for the right leg, as measured by the instrumented treadmill. After segmentation, we averaged the last ten strides of each trial to calculate the mean trajectory for the exoskeleton

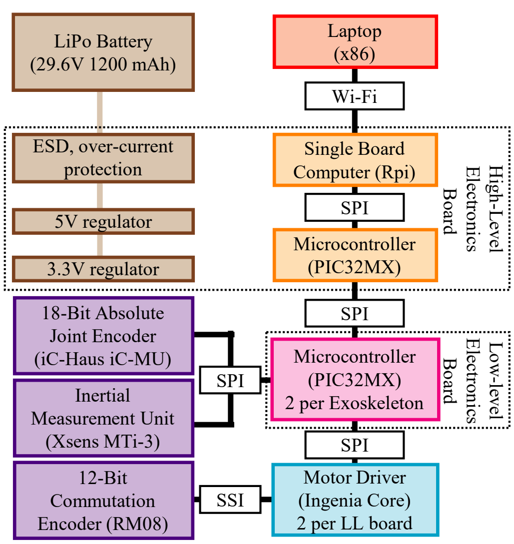

Fig. 2 Schematic architecture of electrical system. A laptop computer runs a graphical user interface that displays exoskeleton data and high-level control parameters. These parameters are sent via Wi-Fi to the high-level electronics board. The high-level board communicates with the low-level boards. The low-level boards collect sensor information, run low-level control routines, and communicate with the motor drivers.

Fig. 3 Subject S1 walked on an instrumented treadmill with the powered exoskeleton in a unilateral configuration. Reflective marker trajectories were recorded with motion capture cameras. Ground reactions forces were recorded by the instrumented treadmill.

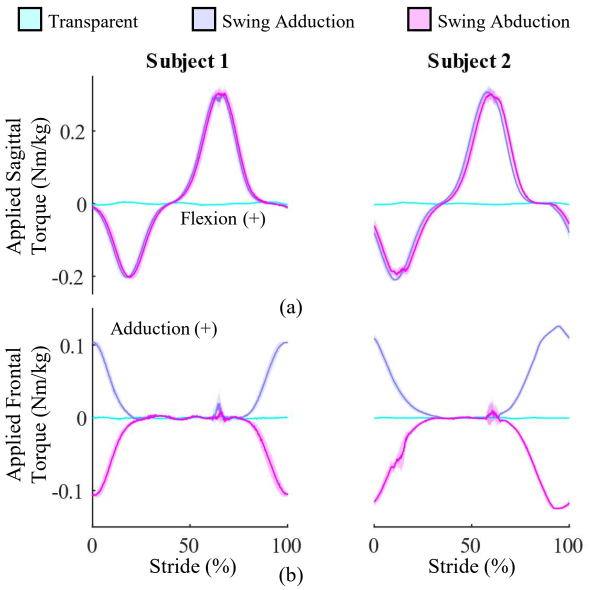

Fig. 4 The exoskeleton generated assistive torques in the (a) sagittal and (b) frontal plane. For each subject and condition, the mean trajectory is shown by a solid line and the standard deviation of the mean is shaded.

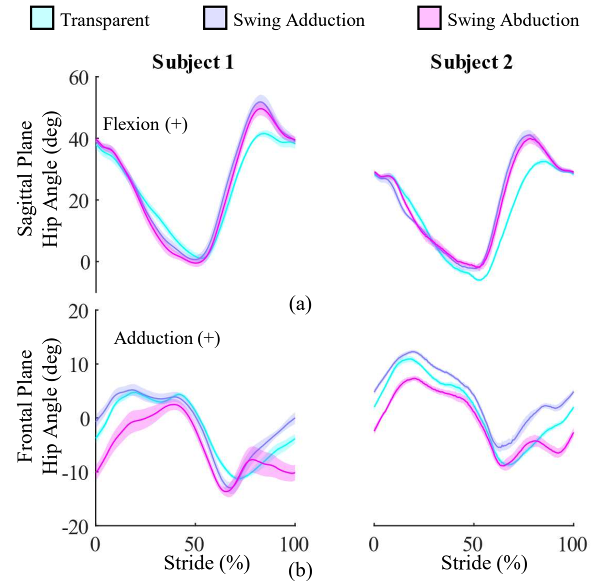

Fig. 5 The hip joint kinematics for the two subjects in the (a) sagittal and (b) frontal plane. For each subject and condition the mean trajectory is shown by a solid line and the standard deviation of the mean is shaded.

applied torque and for the anatomical hip angle. The last 10 strides were included in the analysis to account for adaptations that occur early in the gait cycle. For each of the last ten strides, we also found the average maxima and minima of the applied torque, the maxima and minimum of the anatomical hip angle in the sagittal plane, the anatomical hip angle in the frontal plane at heel strike, and the average step width during double support. We reported the results for individual experimental conditions as the mean ± standard deviation. For results that combine multiple experimental conditions (e.g. peak flexion torque during the two powered conditions), we reported the mean of means.

B. Results

The two subjects walked at 1.07 m/s for all experimental conditions. During the two powered trials, the exoskeleton provided an average normalized peak flexion assistance of 0.32 and 0.31 Nm/kg and an average normalized peak extension assistance of 0.21 and 0.21 for subjects 1 and 2, respectively (Fig. 4 a). The average normalized peak frontal plane assistance during swing was 0.11 and 0.13 Nm/kg for subjects 1 and 2, respectively (Fig. 4 b). For the heaviest subject, the exoskeleton provided an average peak of 29.9 Nm in the sagittal plane and 12.3 Nm in the frontal plane.

Exoskeleton assistance modified the human hip joint kinematics in the sagittal and frontal planes. For subject 1, the average peak flexion angle during the transparent condition was 41.6º ± 1.0º which increased to 52.0º ± 2.4º and to 49.8º ± 2.1º during sagittal plane assistance with adduction or abduction torques during swing. Similarly, for subject 2, the average peak flexion angle during the transparent condition was 32.8º ± 0.8º which increased to 41.3º ± 1.6º and to 40.0º ± 1.7º during two powered conditions. In contrast, for subject 1 the peak extension angle remained relatively the same across all trials (0.9º ± 1.3º, 0.6º ± 1.3º, -0.6 ± 1.2º, respectively). For subject 2, the magnitude of peak extension decreased slightly from -6.1º ± 0.4º (i.e. extended) during the transparent condition to -2.8º ± 1.0º and -2.4º ± -1.0º during the two powered conditions. In the frontal plane, the hip angle at heel strike during the transparent condition was -3.8º ± 0.9º (i.e. abducted) and 2.0º ± 0.4º (i.e. adducted) for subject 1 and 2 which increased to (i.e. adducted) to 0.0º ± 1.1º and 4.9º ± 0.5º under adduction assistance and decreased (i.e. abducted) to -10.1º ± 1.5º and -2.7º ± 0.8º under abduction assistance (Fig. 5 b).

Exoskeleton assistance in the frontal plane modified the base of support (Fig. 6). Without exoskeleton assistance, the step width during double support for subject 1 and 2 was 203 ± 9.8 mm and 269 ± 8.3 mm. Under adduction assistance, step width for subject 1 and 2 decreased by 24% to 154 ± 20 mm and by 25% to 202 ± 9.1 mm, respectively. In contrast, under abduction assistance, step width increased by 40% to 285 ± 14 mm for subject 1 and by 15% to 309 ± 15 mm for subject 2.

IV. Discussion

Powered hip exoskeletons have the potential to improve gait balance and efficiency in clinical populations by generating torques in the frontal and sagittal planes. To the best of our knowledge, few autonomous powered hip exoskeletons can generate torques in both planes concurrently [21]–[25]. However, these devices are heavy and bulky. The actuators, electronics, power sources, and braces protrude significantly from the user, both laterally and posteriorly. The lightest of these exoskeletons is 9.2 kg. The excessive weight

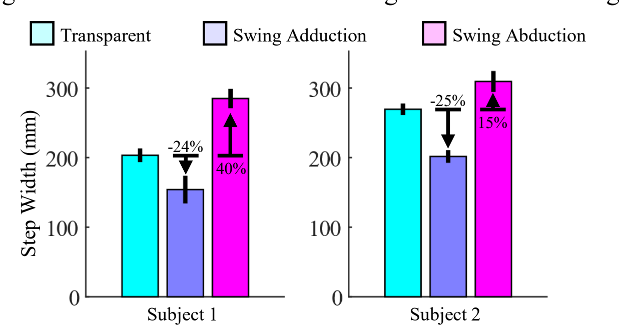

Fig. 6 Average step width during double support for the three conditons and two subjects. Error bars indicate standard deviation of the mean.

and size of these devices has negative effects on the metabolic cost of walking [26] and may prohibit functional tasks like sitting on a normal chair and swinging the arms while walking. Thus, existing powered exoskeletons with frontal and sagittal plane assistance have clear limitations for the real world, especially for clinical populations with limited strength and balance.

In this paper, we present a powered hip exoskeleton that is lightweight (5.3kg), compact (adds 8 cm laterally and 3 cm posteriorly), and capable of generating high torque (up to 30 Nm of assistance measured during walking). The substantial reduction of weight and size compared to previous designs is largely the result of the unique parallel kinematic design used for the exoskeleton hip actuators. Compared to previous exoskeletons where one actuator powers joint motion in one plane [21]–[24], the proposed parallel kinematic design enables both linear actuators to generate hip torque in both the frontal and sagittal planes. Thus, this exoskeleton can generate similar levels of torque to previous designs while using small actuators. Therefore, compared to the predicate device, the proposed exoskeleton is much lighter than previous designs (e.g., 5.3 kg vs 9.2 kg, a 42% reduction) [22] while being able to provide similar torque (30 Nm vs. 34 Nm) [38]. Moreover, this parallel kinematic design places the whole actuators lateral to the user's thigh, which is preferable to the posterior placement used in previous exoskeletons [21]–[24] because it does not interfere with the user's ability to sit on a normal chair while wearing the exoskeleton. Pilot studies with two human subjects show that the proposed exoskeleton can assist in the sagittal plane while providing frontal plane torques to alter step width, which is an important indicator of balance [39].

Level ground walking with two human subjects shows that the exoskeleton can produce up to 30 Nm of sagittal plane torque and 12 Nm of frontal plane torque concurrently (Fig. 4). These torque peaks correspond to 33% and 12% of the biological sagittal and frontal plane torque of a 95th percentile male [40], [41]. Studies with clinical populations and healthy individuals indicate that torques of similar magnitude can reduce metabolic consumption and improve measures of balance in over-ground walking through applying either frontal or sagittal plane torques [3], [4], [11], [30], [42] This comparison suggests that our exoskeleton has the potential to improve metabolic consumption and balance in healthy and clinical populations.

Assistive hip exoskeletons can improve measures of gait function and balance in clinical populations [1]–[3], [11]. In agreement with these studies, our human testing shows that sagittal torque assistance can increase maximum hip flexion angle and frontal plane assistance can selectively increase or decrease the hip frontal plane angle at heel strike (Fig. 5). Moreover, a recent study using an autonomous exoskeleton with only powered frontal plane hip assistance demonstrated the ability to increase step width by an estimated 57% and decrease step width by an estimated 31% [29]. In our study, abduction assistance increased step width by 40% and 15% while adduction assistance decreased step width by 24% and 25% for subjects 1 and 2, respectively. The difference in human response may be due to the difference in control strategy. Notably, their exoskeleton was controlled bilaterally and used admittance control throughout the duration of the gait cycle while our exoskeleton provided unilateral assistance to only the right leg, and only during the swing phase of walking. Thus, assistance magnitude, duration, and control strategy may impact the relative change in step width.

Although the two subjects received similar levels of bodyweight normalized assistive torques, the individual response to assistive torques differed between the two individuals. Specifically, Subject 1 experienced a greater increase in peak flexion angle under exoskeleton assistance compared to subject 2 (9.3º vs. 7.8º). Furthermore, subject 1 exhibited a greater increase in both abduction angle at heel strike (6.3º vs. 4.7º) and step width (40% vs. 15%) during abduction assistance compared to subject 2. This comparison suggests that heavier or taller subjects may require greater bodyweight normalized torques to achieve the same kinematic changes compared to shorter or lighter subjects.

The results of this study are limited in their interpretation. This study demonstrates the ability of the proposed exoskeleton to modify the kinematics of expert users over a short time period (e.g. one minute). It is unclear what adaptions may occur over a longer period of exoskeleton use. Similarly, the impact of the exoskeleton assistance on metabolic consumption and measures of gait function and balance should be directly studied with this device in healthy novice subjects and clinical populations. Furthermore, work should quantify the actuator performance (e.g. bandwidth and back drivability) and present the kinematic model of the transmission system.

V. Conclusion

Lightweight and compact powered hip exoskeletons that assist the user in both the sagittal and frontal plane are critical to improving gait efficiency and balance in individuals with mobility impairments. This paper introduces a new hip exoskeleton with a unique parallel actuation design in which two linear actuators concurrently generate torque in the sagittal and frontal planes. The exoskeleton is lighter and more compact than previous devices. Human tests with two healthy adults show that the proposed powered hip exoskeleton can alter hip kinematics and step width by generating torques in both the frontal and sagittal planes. Future work will focus on testing the proposed powered hip exoskeleton with clinical populations.

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