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Design of a customizable, modular pediatric exoskeleton for rehabilitation and mobility

David Eguren, *Student Member, IEEE*, Manuel Cestari, Trieu Phat Luu, *Member, IEEE,* Atilla Kilicarslan, *Member, IEEE*, Alexander Steele and Jose L. Contreras-Vidal, *Fellow, IEEE*

***Abstract*— Powered exoskeletons for gait rehabilitation and mobility assistance are currently available for the adult population and hold great promise for children with mobility limiting conditions. Described here is the development and key features of a modular, lightweight and customizable powered exoskeleton for assist-as-needed overground walking and gait rehabilitation. The pediatric lower-extremity gait system (P-LEGS) exoskeleton contains bilaterally active hip, knee and ankle joints and assist-as-needed shared control for young children with lower-limb disabilities such as those present in the Cerebral Palsy, Spina Bifida and Spinal Cord Injured populations. The system is comprised of six joint control modules, one at each hip, knee and ankle joint. The joint control module, features an actuator and motor driver, microcontroller, torque sensor to enable assist-as-needed control, inertial measurement unit and system monitoring sensors. Bench-testing results for the proposed joint control module are also presented and discussed.**

I. Introduction

Cerebral Palsy (CP), Spina Bifida (SB) and Spinal Cord Injury (SCI) are major causes of motor function disability in the pediatric population. The most common pediatric motor disability is CP, which occurs in 2-5 per 1,000 live births worldwide [1, 2], and has an estimated global prevalence of approximately 17 million people [3]. Spina Bifida occurs in 1 per 1,000 live births each year in the U.S [4] and approximately 3% to 5% of annual SCI cases occur in the pediatric population resulting in about 1455 pediatric hospital visits in the US each year [5]. These conditions can result in reduced mobility and can interfere with a child's ability to achieve critical developmental milestones [6].

A. Target Populations

CP is a group of permanent, non-progressive conditions caused by injury to the brain of a developing fetus or infant child [7]. The initial injury can result in reduced mobility due to secondary neuromuscular deficits such as muscle weakness, shortened muscles and tendons, spasticity and impaired selective motor control [8]. While the primary causes of the conditions are non-progressive, the secondary effects due to the musculoskeletal defects can worsen over time [9]. It is important to note that while the primary causes are difficult to alter, the secondary abnormalities are more amenable to treatment [10]. In the most common form of SB, myelomeningocele (MMC), individuals often display deficits of the motor and sensory systems, which can result in weakness or paralysis of the lower-limbs causing significantly reduced mobility [11]. While studies utilizing powered mobility or rehabilitation for this population are lacking, a recent study on the use of progressive resistance exercises for an adolescent with MMC reported improved gait speed, walking endurance and functional lower extremity strength [12]. SCI in children can result in significant motor and sensory limitations. It also increases the risk of secondary conditions such as pressure sores, scoliosis, pain, bowel and bladder dysfunction, respiratory issues and immobilization hypercalcemia [13].

B. Pediatric Robotic Assisted Gait Therapy

The last decade has seen a growing number of powered devices and increased research efforts to investigate the benefits of Robotic Assisted Gait Therapy (RAGT) in children. Some reported benefits include improvements in gait speed and performance on the 6-minute walk test [14-15], improvements in overall walking and standing ability [16], improved hip kinematics and 10-meter walk test performances [17], reduction in the metabolic cost of walking [18], reduction of crouch gait and postural improvements with outcomes equivalent to those reported after invasive orthopedic surgery [19], and improvements in exercise induced motor learning despite the presence of neurological impairments [20].

While powered lower limb exoskeletons for mobility and rehabilitation are currently available for the adult population, the commercially available options for children in the U.S. are limited to large, treadmill or footplate-based systems, such as the Lokomat (Hocoma AG, Switzerland) or the Gait Trainer GT-1 (RehaStim, Germany). But due to the size and cost of these devices, these systems are primarily constrained to clinical settings. Attempts to reduce the size and cost of assistive and rehabilitative devices are growing however [21-23] and developments may soon enable children to receive the benefits of RAGT outside a rehabilitation clinic or research facility. For example, the Trexo Home (Trexo Robotics, Canada) is a lower limb exoskeleton device featuring passive and active assist modes that can be used for children with mobility limiting conditions inside and outside the clinical setting. The device is attached to a walker for stability, which in some cases may limit the overground capability of the device, or its ability to navigate enclosed spaces such as those encountered in a home or classroom. The device is controlled by a tablet. Another device, which is currently being used in a research setting at the National Institute of Health's *Functional and Applied Biomechanics* facility [27] contains two actuated degrees of freedom in the sagittal plane, at the knee joint, and

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J.L. Contreras-Vidal, D. Eguren, A. Kilicarslan, T.P.Luu, M. Cestari and A. Steele are with the NSF IUCRC BRAIN Center at the University of Houston, Houston, TX 77204 USA (phone: 713-743-4429; fax: 303-555-5555; e-mail: jlcontreras-vidal@uh.edu).

has been shown to be effective at reducing crouch gait in children with cerebral palsy [19].

The overall effectiveness of a rehabilitation regime involving robotic assisted gait therapy depends largely on the device's ability to assist the user in executing movements according to therapy requirements [24]. Depending on the therapy stage, a patient may require active or passive rehabilitation exercises. In this case, passive mode exercises are responsible for assisting the user in tracking a predefined trajectory to improve movement ability. This mode is often employed in early stages of rehabilitation or to prevent muscle atrophy. Active mode assistance encourages patients to initiate assistance from the robot by providing some form of volitional effort [25]. Our device is capable of operating in both passive and active modes. A description of the control modes and performance tests is included in the *Control Architecture and Hardware Implementation* section.

Within the target populations mentioned, the severity of neuromuscular effects and prescribed intervention may vary widely. Therefore, we have sought to design a rehabilitation and mobility assistance device that is highly customizable, lightweight and modular to aid children with the aforementioned mobility challenges.

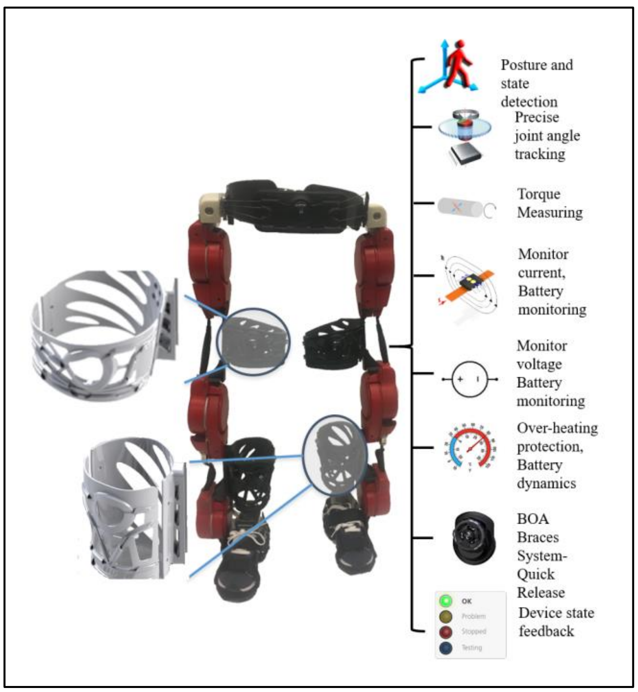

Figure 1. Exoskeleton, sensors and custom braces

II. Pediatric Exoskeleton Device

Figure 1 shows a depiction of the pediatric exoskeleton device, lists sensors located at each joint and displays the location of the custom braces. P-LEGS contains six actuated degrees of freedom (hip, knee, and ankle joints) in the sagittal plane. Customized braces can be used interchangeably with the same actuator units and are located at each thigh and shank. The braces are 3D printed, based on 3D scans of the child's legs, and can be reinforced with a variety of materials to ensure strength in a lightweight form factor [26]. This way we are able to ensure an optimal fit, increase torque transfer from the user's legs to the device and improve comfort. A ribbed design allows for increased breathability, reducing perspiration and pressure sores from rubbing. This method of brace fabrication allows the device to grow with the child by rescanning the child's lower body and printing a new set. This also allows numerous children to use the same device while maintaining personalization. Accurate tracking of joint angles and position control for the production of customizable gait trajectories are done by means of a rotary encoder in each joint. A Wheatstone bridge-based torque sensor measures interaction forces between the user and the device, enabling assist-as-needed control of the device to ensure the child remains engaged in the rehabilitation process. Force sensors in the user's insole provide gait phase identification and pressure distribution information. Inertial measurement units on the device may provide a measure of balance and a way to monitor spasm onset. Table I contains additional information about the user and device as well as the methods of control and feedback mechanisms. The targeted age, height and weight are children between the ages of 4-8, of height (1 –1.23 m.) and weight (16 – 28 Kgs.). The device weighs 8 Kgs.

TABLE I. Device and User Specifications

Users	
User Height	1-1.23 m.
User Weight	16 – 28 Kgs
Device	
Weight	8 Kg
User Feedback	Sound, LED, Display Screen, Haptic (vibration)
Control	Joystick, Tablet, Joint, assist-as-needed, Position, myoelectric (EMG), brain-machine interface (EEG)

Figure 2 is an image of an able-bodied boy wearing a prototype of the pediatric exoskeleton during a data collection session.

Figure 2. Able-bodied boy wearing pediatric exoskeleton prototype

P-LEGS can be used with a walker or crutches depending on the stability needs of the child. P-LEGS contains an open architecture and can receive inputs from a crutch-based controller. It can also be interfaced to myoelectric (EMG) and EEG-based neural interface shared control, as demonstrated in previous studies with other devices [28, 29]. Arrangements are underway to collaborate with two clinical centers in the Texas Medical Center for feasibility testing of the device within the target populations.

A. Mechanical Description

The device contains six actuated degrees of freedom (DOF) in the sagittal plane and two passive DOF at the hip, enabling external rotation, and abduction/adduction providing the ability for weight shifting. The system is comprised of six joint control modules, one at each hip, knee and ankle joint. The joint control module, features an actuator and motor driver, microcontroller, torque sensor to enable assist-as-needed control, inertial measurement unit and system monitoring sensors. The joint control modules each have an identical design to increase modularity and manufacturability. The actuator housing at each joint contains a 24V Maxon motor, crossed roller bearing and gearbox with a 161:1 ratio. The nominal torque at each joint is 13.5 Nm with a maximum momentary peak torque of 76 Nm. The joints contain redundant mechanical hard stops to prevent the joint from rotating beyond the predefined range of motion limits.

B. Electrical Description

The distributed architecture of the system electronics is designed to increase the modularity of the system. Each joint contains an ARM Cortex M-4 MCU, motor controller and sensors for voltage, temperature and current sensing. It also contains a 9 DOF inertial measurement unit, signal acquisition electronics for insole force sensitive resistors, rotary encoder for joint tracking and gait trajectory production and a Wheatstone bridge-based torque sensor. The system communication is based on the CAN bus communication protocol. Custom printed circuit boards were designed and built for the joint control modules. The compact design seeks to further increase the overall modularity of the system and to reduce the system size and weight.

C. Torque Sensing

The joint control module contains an embedded torque sensor consisting of two strain gauges installed at 180° apart, and 90° from the output link. These shear gauges form a full Wheatstone Bridge (WB) that can translate small deformations on the joint housing into torque measurement. By implementing the WB directly into the joint housing, no extra components are needed contributing to the joint's compact design.

Each modular joint is capable of operating independently of the other joints. All instrumentation and control hardware is self-contained in each joint control module. Figure 3 presents the experimental setup used for testing each module independently. A section of the joint is fixed, while the output is connected to an output link where the load will be exerted during testing. The WB formed by the strain gauges on the joint housing is also shown.

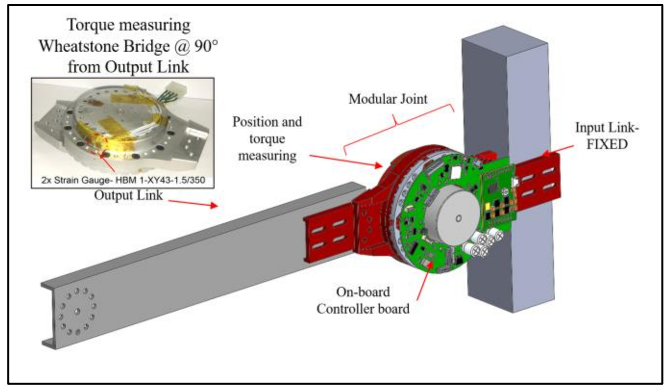

Figure 3. Testbed for bench testing the adaptive impedance control for assist-as-needed (ANN) control of a pediatric exoskeleton

D. Control Architecture and Hardware Implementation

Control strategies for robotic assisted gait therapy (RAGT) must manage the physical interaction between the user's lower limbs and the robot. The assist-as-needed control strategy is a rehabilitation paradigm that encourages patients to actively participate in tasks and takes the user's intention into consideration rather than imposing a rigid control strategy [25]. In a pilot study involving a post-stroke individual, greater improvements in functional walking ability following an assist-as-needed RAGT paradigm compared to a RAGT with continuous assistance were demonstrated [30]. Additionally, cooperative control strategies using Functional Electrical Stimulation (FES) and robotic controllers have been implemented in the adult SCI population [31]. The device could be adapted to receive control inputs from FES signals, and the drive to provide an open architecture design is that additional researchers would have access to low-level control mechanisms to explore alternative control strategies and improve outcomes for the target populations in need of highly customizable rehabilitation. Our device is capable of both passive and active control modes. The passive control strategy uses encoders for joint angle and gait phase tracking. The active mode uses an adaptive impedance controller to allow for volitional control inputs from the user. Figure 4 shows the implemented Assist-As-Needed (AAN) control scheme of the Pediatric Exoskeleton.

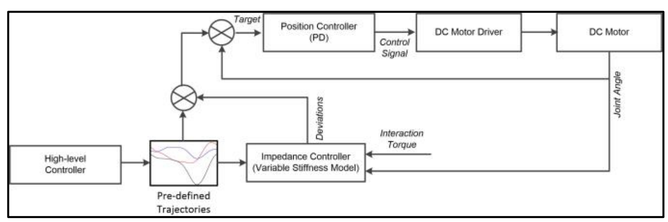

Figure 4. Adaptive impedance control for Assist-As-Needed (AAN) control of a Pediatric Exoskeleton

The low-level joint controller includes a Proportional-Derivative (PD) position and a velocity controller. The control parameters of the embedded velocity controller were independently tuned for each joint control module. The hierarchical control architecture is inspired by the theory of a shared-control paradigm in human-robot interaction systems [32]. The high-level controller is designed to estimate the state of the overall exoskeleton, generate gait patterns, receive user commands and determine user intent. The low-level controller collects and processes sensor data at the joint level (encoder angle, interaction torque and current), and drives the actuator to follow the desired trajectories dictated by the high-level controller. Feedback from interaction torque, current, and angle position is monitored to generate the desired output at the joint based on the commands given by the high level controller, and the specific mode of operation. The impedance controller at the joint operates as a virtual spring with exponential behavior as seen in Figure 5.

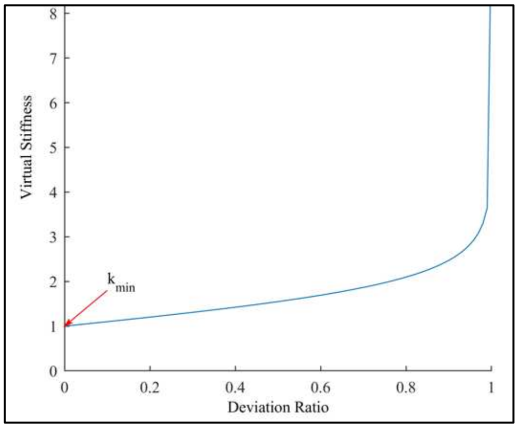

Figure 5. Virtual spring constant versus deviation ratio plot

Based on the desired stiffness of the joint, a relationship between the desired set point, actual position, and interaction torque can be set. The virtual stiffness is modified preventing the joint from deviating beyond a desired position and torque threshold defined by the operator. This feature allows the device to operate in a range of compliancy, which can be adjusted based on the needs of the user and mode of operation. A high degree of stiffness is desirable for position tracking while varying degrees of compliance would be useful for assist-as-needed control. CAN Bus was implemented for data transmission and communication between the high- and low-level controllers. The system architecture is depicted in Figure 6.

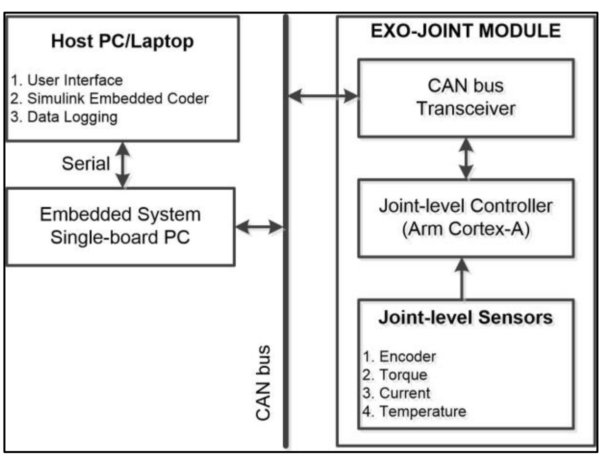

Figure 6. Control architecture for the pediatric exoskeleton

III. Results and Discussion

A series of three tests were performed on the joint control module to evaluate controller performance. In the first test, the controller was tasked with tracking the hip, knee and ankle joint trajectories with various gait cycle times ranging between 4 - 10 seconds. The performance of the position tracking can be seen in Figure 7.

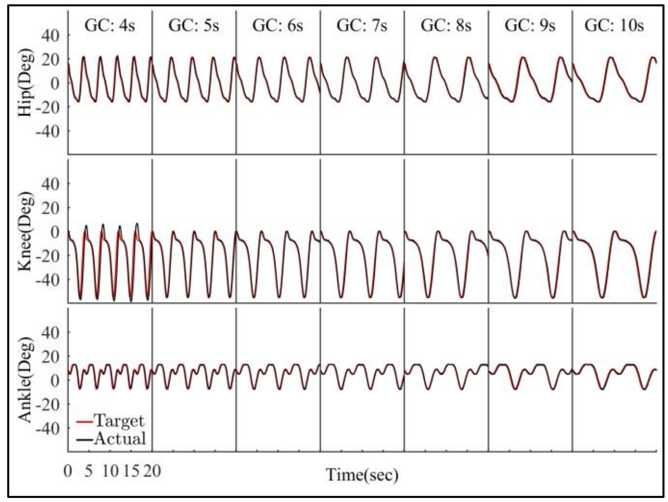

Figure 7. Position tracking for gait cycle speeds between 4-10 seconds

The results of the second test can be seen in Figure 8. The controller was set to passive mode (position control) for the first 20 seconds of testing and active mode (variable stiffness impedance control) after the 20 second mark and remained in active mode until the end of the test. Interaction forces were applied in both directions during passive and active mode control testing. Interaction torque can be seen as spikes on the plot during active mode control, however, during the passive (position control) mode, the joint angle maintained the set point position despite the interaction torque and no deviation was observed. During the active mode (impedance control), the joint position deviated from the target when the interaction torque was applied. The amount of deviation is proportional to the interaction torque applied and bounded within pre-set values, seen as dotted red lines in the plot.

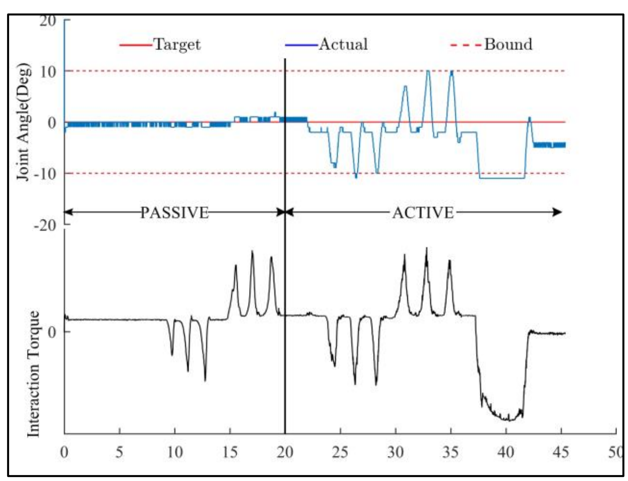

Figure 8. Active versus passive mode reaction to interaction torques

Similar to the second test, in the third test (results in Figure 9) the controller was set to passive mode (position control) for the first 20 seconds of testing and active mode (variable stiffness impedance control) after 20 seconds until the end of the test. The controller was tasked with tracking a hip joint gait trajectory during the third test while interaction forces were applied to the joint in both directions. In the position control mode, the joint angle maintained the set point and no deviation was observed even when the interaction torque was applied. During the active mode, the joint position deviated from the target when the interaction torques were applied. The amount of deviation is proportional to the interaction torque applied and bounded within pre-set values. It can be seen that the controller in active impedance mode allows deviation from the commanded trajectory when in presence of external load. However the overall trend of the motion tracks the commanded trajectory without evidence of overshooting or blocking.

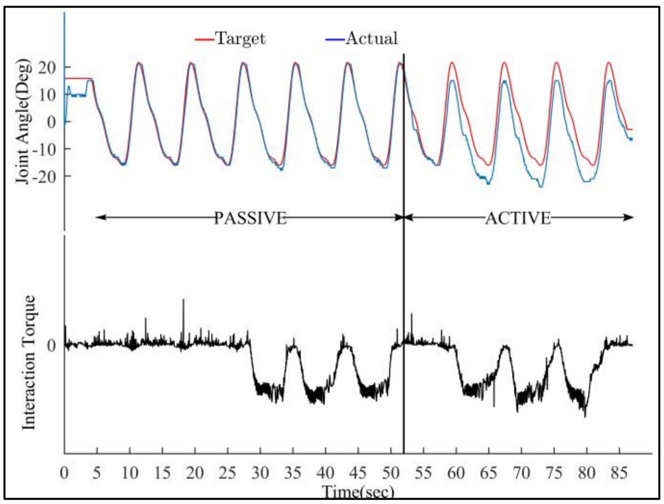

Figure 9. Controller tracking hip joint trajectory in passive and active modes with interaction forces applied

IV. CONCLUSION

Rehabilitation and mobility assistance devices capable of providing customizable RAGT outside clinical settings are lacking for the pediatric population, therefore we have developed a multifunctional, modular, assist-as-needed powered pediatric exoskeleton. The system's control strategy has been designed to operate with multiple control modes, including position, and variable stiffness impedance control. Gait intention detection can be accomplished from inertial sensors, myoelectric (EMG) or neurological (EEG) sensors to further engage the children. The proposed device is multifunctional, and can act as a diagnostic tool, a gait rehabilitation platform and/or a mobility assistance device. Importantly, we have integrated a Wheatstone bridge-based torque sensor to measure the interaction forces between the user's lower limbs and the device and have demonstrated that the joint is capable of accurately tracking a reference trajectory from joint angle data in the presence of interaction forces. The device is capable of operating in passive and active mode. In the passive mode, the joint closely tracks the input trajectory despite the presence of interaction torque, while in the active mode, the controller allows the joint angle to deviate from the reference trajectory within a set bound, enabling the device to respond to volitional torque contributions from the user. The wearable exoskeleton, due to its high degree of modularity and ability to be customized, can be adapted to meet the physical and neurological needs of children with mobility limiting conditions such as cerebral palsy, spina bifida and childhood stroke.

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