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Design and Preliminary Evaluation of a Multi-Robotic System with Pelvic and Hip Assistance for Pediatric Gait Rehabilitation

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Abstract— **This paper presents a modular, computationally-distributed "multi-robot" cyberphysical system designed to assist children with developmental delays in learning to walk. The system consists of two modules, each assisting a different aspect of gait: a tethered cable pelvic module with up to 6 degrees of freedom (DOF), which can modulate the motion of the pelvis in three dimensions, and a two DOF wearable hip module assisting lower limb motion, specifically hip flexion. Both modules are designed to be lightweight and minimally restrictive to the user, and the modules can operate independently or in cooperation with each other, allowing flexible system configuration to provide highly customized and adaptable assistance. Motion tracking performance of approximately 2 mm root mean square (RMS) error for the pelvic module and less than 0.1 mm RMS error for the hip module was achieved. We demonstrate coordinated operation of the two modules on a mannequin test platform with articulated and instrumented lower limbs.**

I. Introduction

Developmental delay due to premature birth is a major public health problem in the United States [1]. One and a half percent of the more than 4.25 million infants born annually in the U.S. have a very low birth weight [2], and of the 90% that survive, 25-50% suffer a brain injury resulting in development of locomotor delays [3]–[5]. Cerebral palsy (CP) is one manifestation of these locomotor delays and the most common childhood disability that causes lifelong physical impairment, affecting an average of 3.1 per 1000 children in the US [6].

As fundamental motor skills are typically acquired in the early stages of development [7], early intervention is critical, and may help prevent or mitigate development of abnormal movement patterns that lead to secondary musculoskeletal problems common in children with CP, such as hip dislocation, contractures, and bone deformities [6]. In addition, it is important to improve mobility and encourage a more active lifestyle to help children with CP interact with their peers, achieve developmental milestones, and experience better quality of life.

Gait interventions for CP may include surgery, functional electrical stimulation, orthotics, and medication, but the central component of the strategy for managing locomotor disorders is physical therapy [6]. Robotic devices have appeared in the clinic as tools to facilitate and augment physical therapy. For example, treadmill-based lower-extremity rigid exoskeleton robots such as the Lokomat [8], and ALEX [9] provide gait training by moving or guiding the lower limbs in a physiological gait pattern, and have been shown to improve the gait and balance of and increase the likelihood of achieving independent ambulation in patients with impairments stemming from stroke or incomplete spinal cord injury. Other robots such as the KineAssist [10] can provide dynamic body weight and postural support. Outside of the clinic, portable lower-extremity rigid exoskeleton robots such as the H2 [11], ReWalk, Ekso, and Indego systems [12] have enabled patients with total loss of lower-limb function to walk again. For patients with partial gait impairment, such as stroke survivors, recent developments include textile-based, Bowden-cable actuated soft exosuits [13]–[15] which provide only moderate levels of assistance compared to their rigid counterparts, but are much lighter, lower-profile and wearable. These exosuits work in parallel with the motion of the user to induce more normal kinematics, reduce walking effort, and increase mobility.

More recently, as the benefits of earlier intervention have been recognized, there has been increased interest in pediatric applications of assistive technologies [16]. This interest has manifested in a variety of projects, from passive exoskeletal garments such as the Playskin Lift to assist upper extremity function [17], driving robots for infant mobility [18], [19], single-joint robotic modules such as the MIT pediatric Anklebot [20] or a knee exoskeleton to alleviate crouch gait, bio-inspired active orthotics for infants [21], treadmill-based gait trainers such as the pediatric Lokomat, and pediatric rigid exoskeletons [22].

Children learning to walk face multiple challenges of stabilizing medio-lateral body sway, developing a dynamic gait that exploits potential and kinetic energy exchange, and coordinating of multiple degrees of freedom. As such, a device that assists the child with several of these sub-tasks of walking may be more effective than a device that focuses on only one. This combined assistance approach has been used in adults; for example, the PAM and POGO pneumatic robots

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work together during treadmill-based gait training to assist both the leg motion and the pelvic motion [23]. In children, the recent CPWalker [24] project built an active robotic platform for rehabilitation by motorizing the base of an existing pediatric passive assistive device for CP, the NF-Walker, for forward propulsion, adding controllable body weight support, and integrating it with a rigid lower-limb exoskeleton.

Taking a similarly multi-faceted approach to addressing the challenges of learning to walk, we present a modular, computationally-distributed cyberphysical system that enables a multi-modal approach to addressing these challenges in a lightweight and minimally-restrictive platform. This "multi-robot" (Fig. 1) consists of (a) a tethered cable pelvic module with up to 6 DOF, which can modulate the motion of the pelvis in three dimensions, and (b) a 2 DOF hip module to assist lower limb motion, specifically hip flexion. As these modules are designed to be used individually or in combination, the system may be flexibly configured to provide assistance tailored to the specific needs of the user, as well as keep pace with the user as they progress and change their walking behavior.

II. Design Requirements and System Concept

As the goal is early intervention, the system was designed for toddlers and young children 2-5 years of age. The key functional aim is not to control how the children walk, but rather, enable them to explore their own behavior, while supported by a programmable and safe structured environment that provides the appropriate level of guidance at different times to encourage normal walking.

A. *Functional Requirements*

Given the above objective, it is important that the system interferes as little as possible with the user's natural walking dynamics. Therefore, the weight and profile of body-worn components of the system should be minimized. Heavy actuation, power and control components are located off-board on a supporting frame behind the user. The body-worn components are designed to avoid restricting joint motion, while actuation components can go slack or have built-in mechanisms for becoming highly transparent to the user when not active.

In addition, due to the young age and vulnerable nature of the target patient population, the system incorporates safety mechanisms at multiple levels to ensure system safety and reliability, including mechanical hard stops, hard-wired emergency stop switches, fuses for current limiting, and programmable position/force limits in software. While not shown with the mannequin in Fig. 1, any children using the system would wear a harness suspended from the top frame to prevent falls. The harness may also aid upper body posture and/or provide partial body weight support as necessary.

B. *Modular Design*

The system uses a division of labor among multiple, coordinated robotic modules for the separate learning tasks of (a) redirecting the body center of mass (CoM) so that it moves forward while maintaining stability, and (b) guiding behavior of the legs in a gravitational field (stepping). These tasks are similar to what adults do when they lift a child's hands above their heads and pull in such a way that the child pivots on one foot as their other leg swings in a forward stepping motion. The first robot is a cable-driven pelvic module connected to the toddler's pelvis, based on the A-TPAD [25] developed at Columbia University, which provides dynamic stabilization of the toddler. The second robot is a hip module worn on the legs and trunk, which provides sagittal-plane assistance to the hip joint in order to help the child learn how to initiate gait and coordinate motion of the legs with the full body movement. A unique aspect of the system is that each module can operate on its own, or in coordination with the other.

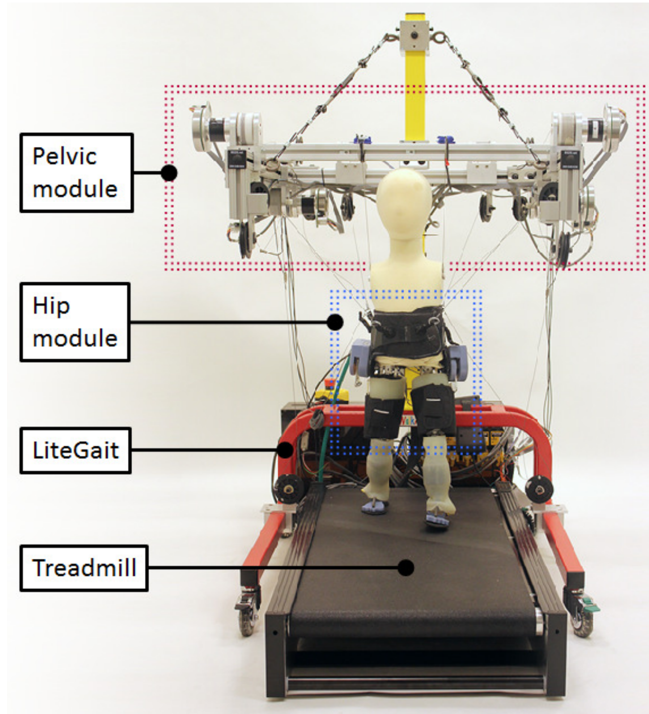

Figure 1. Overview of full system on mannequin test platform, showing the major components: the pelvic module (up to 6 DOF), the hip module (2 DOF), the supporting LiteGait frame, and the base treadmill.

This modular architecture enables a high level of customization of therapy to meet the individual needs of each toddler. Some toddlers may have more trouble with balance than with limb motion, or vice versa, and so the learning assistance provided by each robot can progress at different rates to match the individual's development. Furthermore, the system may help simplify the complex task of walking for the child by providing maximum support for one aspect of walking while relaxing support for the other, allowing the child to focus on one task at a time. In addition, at any time, a physical therapist or a parent may substitute for either robotic module to provide therapeutic interventions, while the other robot continues to perform its role. Finally, if therapy is effective, and the child progresses to no longer need the support of the pelvic module, the child may be able to take the hip module outside of the clinic to walk independently in other environments, such as at home or at school.

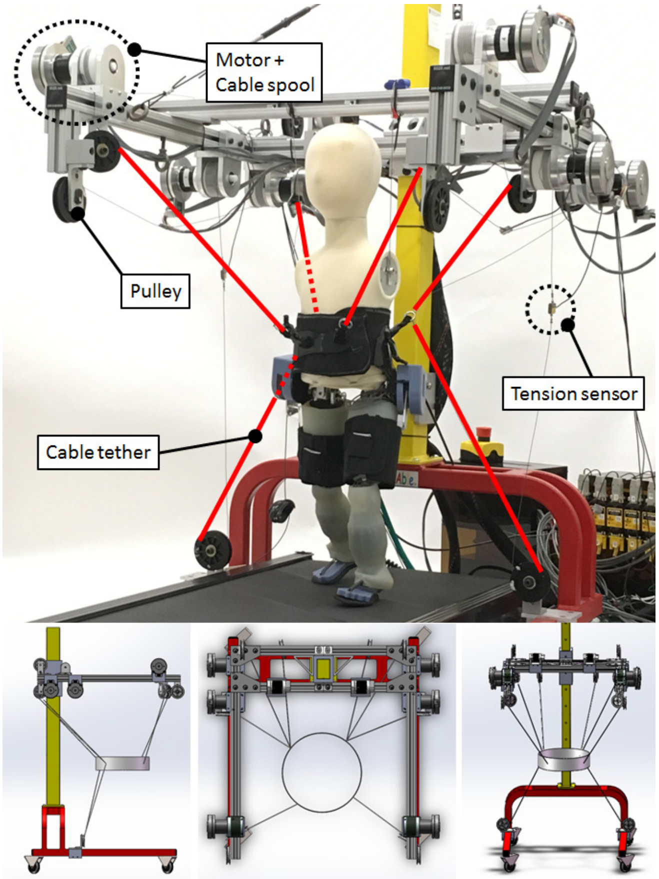

Figure 2. Top image: cable-driven pelvic module mounted on the LiteGait frame. Bottom row: side, top, and front CAD views showing one possible cable configuration.

III. Design of Pelvic Module

The cable-driven pelvic module is intended to assist the balance and gait of children. A child in the device will wear a waist belt connected to multiple cables. The cables apply forces/torques on the child's pelvis to guide, support, and assist gait. Additionally, these forces/torques can be used to provide partial body weight support, weight augmentation for strengthening, or to control the pelvis according to the training strategy.

A. Design Requirements

Target subjects for this device are children between two to five years old who weigh under 25 kg. The pelvic module is able to apply up to 25 N (10% of body weight) of force on the pelvis in any direction. This specification is based on pilot results from an ongoing study measuring interaction forces between a parent's hands and their healthy toddler's body, using instrumented gloves, as the parent stabilizes the toddler's standing posture prior to the toddler initiating gait and walking forward. The maximum speed of the pelvis in this design was chosen as 1.0 m/s. This is from center of mass data recorded from healthy toddlers walking toward their parents using a motion capture system [26].

B. Hardware Description

The cable-driven pelvic module consists of motors, pulleys, passive cable winches, and a controller box mounted on a support system already used in therapy, the pediatric LiteGait (LG MX100, Mobility Research, USA). Thus, a significant consideration in the module design is mounting it on the LiteGait to allow the system to be portable, enabling positioning over a treadmill as well as the possibility of over-ground use if the frame is pushed or motorized. In this cable-driven system, six electrical motors and two passive winches are used as shown in Fig. 2. Each electrical motor (EC 90 flat, Maxon AG, Switzerland) can apply up to 65 N and uses 4.3:1 gearing. A tension sensor (LSB200 50 lb, FUTEK, USA) is installed along the cable length to monitor its tension.

The system is designed with the flexibility to mount the motors in any location on the LiteGait frame. Besides the motors, movable pulleys are also attached to the frame to change the direction of the cables and the corresponding forces. A total of six active cables are used in this system. While four cables are sufficient to generate a three-dimensional force [27], [28], two additional cables are used to distribute the end-effector load among the cables. This arrangement also allows us to avoid positioning the cables close to the child's face.

Two passive cable winches are installed to obtain additional cable length measurements. The passive modules include an encoder to measure the change in rotation of the cable reel to measure the cable lengths. This information is used to solve for the forward kinematics, eliminating the need for motion capture required by the previous A-TPAD and enabling the system be used in a wider range of settings. The passive cables are kept in tension by torsional springs installed inside of the cable reel.

IV. Design of Hip Module

The hip module is intended to assist hip flexion of the toddler to initiate the swing phase. This assistance is designed to ensure ground clearance by the foot, and to alleviate ankle deficits. However, rather than completely controlling the user's own movement and locking the legs to the robot's trajectory, the module supplements the user's motion by providing impulses and cues to help initiate movement of the legs.

A. Design Requirements

As the purpose of the module is to guide, not control, hip flexion, it needs to provide only partial assistance. Therefore, we targeted 30% of the biological hip flexion torque. As the target population is from two to five years old, the module can provide up to 2.3 Nm of torque, based on an average biological peak flexion torque of 7.6 Nm in 5-6 year old children [29]. The required velocity is up to 600 degrees/second, derived from the angle data published in [26]. In addition, it is important for the robot to be as transparent to the toddler as possible. The body-worn components of the system must not restrict natural motion of the user, both in the sagittal plane in which the hip module actuates, as well as in the other degrees of freedom including adduction/abduction. The weight of these body-worn components must be minimized so as to not significantly impact the toddler's natural walking dynamics.

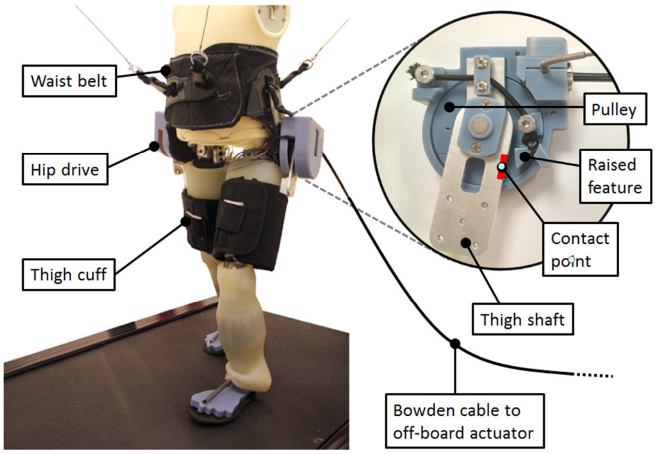

Figure 3. Overview of wearable hip module, showing body-worn hip drives on a waist belt and thigh cuffs, with the actuating Bowden cable leading from the hip drive to the off-board Maxon EC60 motor (not shown). A detailed view of the internal hip drive mechanism is shown.

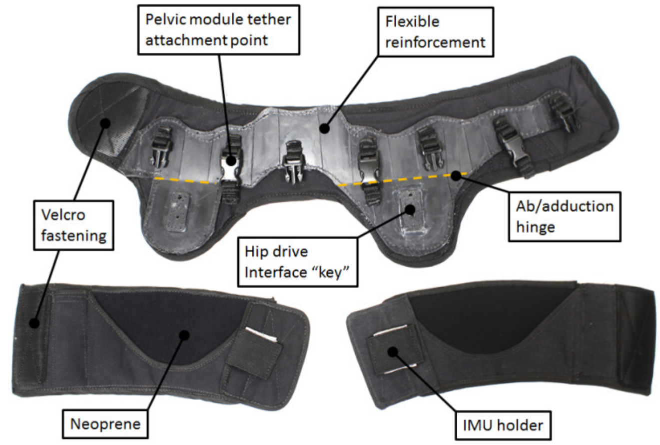

Figure 4. Components of the human interface, laid flat. Top: waist-belt with attachment points for cable tethers and mounting keys for hip drive. Bottom: thigh cuffs with integrated neoprene section and pockets for drive unit thigh shafts and IMUs.

B. Hardware Description

The hip module hardware consists of two motors located off-board behind the toddler, and two "hip drive" units mounted on a belt worn around the toddler's waist (Fig. 3). The motors are connected to the drive units via a Bowden cable transmission. The drive unit mechanism, described in greater detail below, converts the motor-driven linear pull of the Bowden cable to rotational motion of a shaft aligned with the toddler's thigh.

The two brushless DC motors (EC60 flat, Maxon AG, Switzerland) are mounted behind the toddler to reduce bending and frictional losses in the Bowden cable transmission. The motors are not geared and are capable of 0.32 Nm continuous torque, and up to 5 Nm maximum torque. When an off-board motor turns, it pulls on one end of the inner cable of the Bowden transmission, causing the hip drive pulley to rotate. The diameter of this driven pulley matches that of the driving pulley on the motor, resulting in a 1:1 transmission ratio which allows the module to be back-drivable. As the pulley rotates, a raised feature on its surface contacts the thigh shaft, engaging it and pushing it forward. This mechanism ensures that the hip drive cannot hold back the user should they choose to move their leg faster than the module's actuation profile. When the motor is not active, an elastic cord retracts the Bowden cable and the pulley rotates back to its initial position out of the way of the thigh shaft so that it does not block the leg during extension. This enables the user to walk freely without feeling any interference from the mechanism, the transmission, or the motor. Hard stops on the pulley frame limit the allowable motion within a safe range from 30° extension to 90° flexion.

A small magnetic position sensor (AS5600, ams, Austria) tracks the absolute angle of the thigh shaft. A through-hole compression load cell (LTH300, FUTEK, USA) measures the tension in the Bowden cable by indirectly measuring the reaction force of the cable sheath. As $\tau = F \times r$ where r is the pulley radius, 3 cm, this measured tension is used to estimate the torque being applied to the toddler's hip joint.

C. Human Interface

The function of the interface is to transfer the forces applied by the pelvic module cables and hip module drive units to the body of the toddler in a comfortable manner. The interface consists of a belt worn around the toddler's waist, and two cuffs worn around the thighs (Fig. 4). The waist belt serves as the interface between the wearer, the pelvic module cables, and the hip drive units. The belt has integrated buckles for quick attachment and release of the pelvic module's cables. The thigh cuffs transfer the forces between the user's thigh and the thigh shaft coming out of the hip drive unit, as well as hold the inertial measurement units (IMUs) (VN-100, VectorNav, USA) on the thigh for gait sensing. The thigh cuffs have an integrated section of neoprene which increases comfort as torque is being applied to the thigh by providing a small amount of elasticity.

To enable the waist belt to resist the counter-torque created by the actuation of the hip drive unit, it is reinforced with medical splinting thermoplastic (Omega Black, North Coast Medical, USA) sewn directly onto the textile layer. The hip drives are located on two drop-down sections which more closely co-locate the centers of rotation of the hip drive and the toddler's hip joint. The thermoplastic is scored to create a passive hinge that allows for ab/adduction.

All the body-worn components are secured with hook and loop fastener, enabling adjustable fit and quick donning and doffing. The total mass of the human interface components is 0.35 kg. If the hip module's drive units are worn, the total mass of the body-worn components becomes 0.89 kg.

V. Assistance Strategy and Control System

For the robotic assistance strategy, we take a bio-inspired approach that is based on the underlying principles of parent-child interaction biomechanics and the walking biomechanics of healthy toddlers.

A. Pelvic Motion Assistance Control

Postural control is a fundamental prerequisite for gait stabilization and critical for developmentally-delayed children when learning to walk. These children typically have

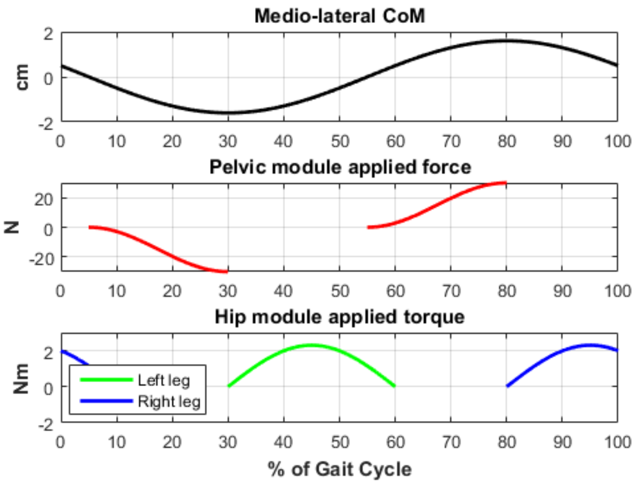

Figure 5. Desired force and torque reference trajectories for the pelvic module and the hip module with respect to the gait cycle. Medio-lateral force applied by the pelvic module induces lateral motion of the CoM and weight shifting onto the stance leg, followed by actuation of the swing leg by the hip module.

difficulty stabilizing medio-lateral body sway [30], [31]. Thus, the first application of the pelvic module is to assist frontal plane motion to aid weight shifting. As the center of mass lies within the pelvis [32], assisting the pelvic motion directly modulates the body center of mass.

The onset and offset timing of the assistance is chosen to facilitate the medio-lateral motion of the center of mass, and the position profile with timing and magnitude is illustrated in Fig. 5. We are targeting a range of 3 cm of medio-lateral motion, based on data from a previous motion capture study of toddler walking biomechanics.

The basic controller of the pelvic module consists of two layers. The low level motor controller (ESCON 70/10, Maxon AG, Switzerland) runs closed-loop speed control, while at the high level a proportional–integral–derivative (PID) controller closes the position-tracking loop.

B. Hip Flexion Assistance Control

To facilitate gait initiation and ground clearance, hip flexion assistance is provided. We aim to provide 1/3 of the biological hip flexion moment and maintain a similar range of sagittal hip motion as the toddler biomechanical data reported by Hallemans et. al. in [33], where the average maximum thigh angle is approximately 40° at 85% of the gait cycle and average minimum thigh angle is approximately 10° at 55% of the gait cycle.

The system uses a gait cycle-based closed-loop position control system. IMUs attached to the thigh cuffs of the hip module measure thigh angle and calculate the current position in the gait cycle. The controller can detect maximum hip flexion and stride time is estimated as the time between two consecutive maximum hip flexion events. Using this information, the controller determines the timing for applying the next assistive impulse and generates the appropriate command position trajectory.

To control the hip flexion motion with the required range, both the Bowden cable travel distance and the desired motor rotation motion were calculated based on the pulley diameters at the actuation side and thigh cuff side. A PID

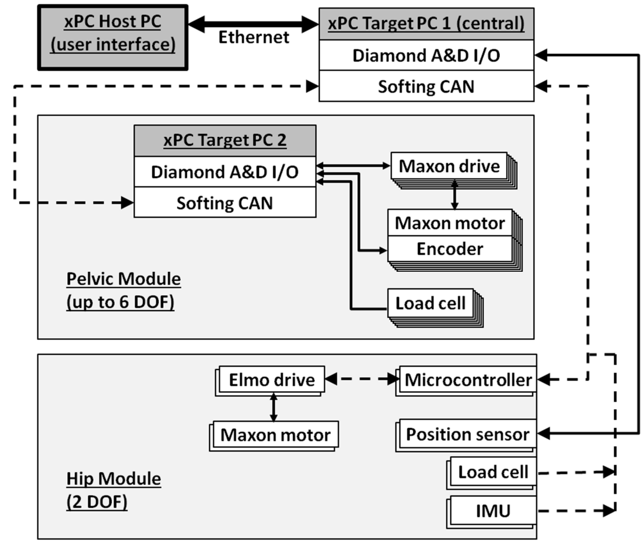

Figure 6. Control and communication architecture of the system showing connections between modules and components within modules. Dotted lines indicate CAN bus signals.

controller was implemented in the low level motion control system (Gold Twitter, Elmo Inc., USA).

C. Control System Architecture

The high level control of the system is handled by a compact rugged central computer (ARS-1200, Vecow Inc., China) running MATLAB Simulink Real-Time for rapid development and evaluation of control algorithms. The computer has a PC-104 bus enabling flexible hardware configuration, including the addition of a Controller Area Network (CAN) interface card (CAN-AC2-104, Softing, Germany) and a digital and analog I/O board (DMM-32DX-AT, Diamond Systems, USA). As the pelvic module has a large number of controllable axes, up to 6 cable tethers, it has its own dedicated computer with an identical setup (Fig. 6). By running the Simulink model in external mode, selected parameters can be modified in real-time via the model's block diagram, which is displayed on the host computer and essentially acts as the user interface for the system. The host computer is connected to the central target computer by an Ethernet cable.

Communication between modules and sensors is primarily handled through CAN buses. CAN is a message-based serial communications bus that is robust and easily expandable. Modules can easily be connected to or disconnected from the bus, greatly simplifying the wiring compared to if each sensor was directly wired to a central I/O board.

For the hip module, the central computer sends position reference commands via CAN messages to the module's microcontrollers, which in turn relays the commands to the motor drivers. For the pelvic module, the central computer sends a position reference command via CAN to the pelvic module's dedicated control computer, which in turn controls the module's motor drivers with analog reference signals, using position feedback from the motor encoders and force feedback from the in-line cable load cells.

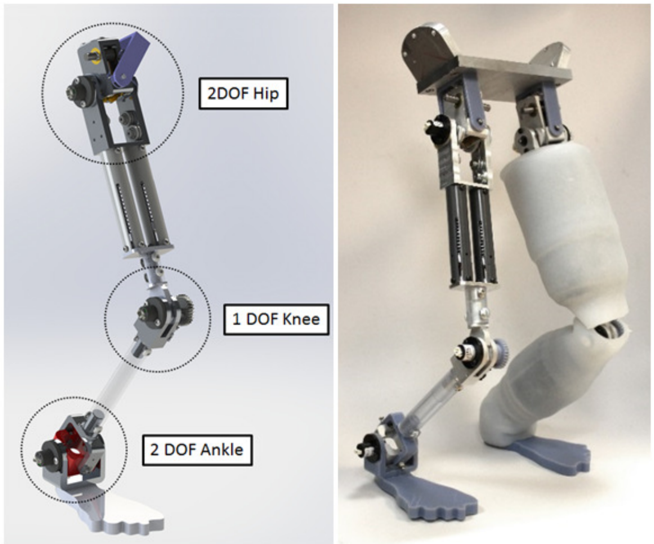

Figure 7. Left: CAD model of one mannequin leg highlighting the three modeled joints and DOF. Right: Assembled legs with and without foam/silicone covering.

VI. System Evaluation

Due to the complexity of the system and the young age of the target population, preliminary testing was necessary before proceeding with human subjects. For this purpose, we constructed a passive mannequin test platform. Below, we show the low-level tracking performance of the individual modules, followed by the results of the combined system operation on the mannequin model.

A. *Test Platform*

A mannequin test platform was built with articulated and instrumented lower limbs. Three joints were modeled: hip, knee and ankle (Fig. 7). The dimensions and masses of the limb segments were approximated from measurements from 2-year-old typically developing children.

The hip joint uses a universal joint mechanism allowing 2 DOF: flexion-extension and abduction-adduction. A self-locking differential mechanism enables adjustment of the joint stiffness in the sagittal plane. The knee is a pin joint with 1 DOF, flexion-extension, and is connected to a damper. The effective damper coefficient can be changed by using different pairs of gears. The ankle also uses a 2 DOF universal joint mechanism, allowing plantarflexion-dorsiflexion and inversion-eversion, with both DOFs spring-loaded with torsional springs. All 3 joints are equipped with absolute rotary encoders (MAE3, US Digital, USA) to measure joint angles in the sagittal plane.

The mechanical skeleton was covered with an inner foam core, then an outer sleeve molded from soft silicone elastomer (Ecoflex 30, Smooth-On Inc., USA), to provide the appropriate volume and mimic the skin and soft tissue of a real leg.

B. *Position Tracking Performance of Pelvic Module*

To provide medio-lateral pelvic motion assistance to the toddler, the pelvic module was configured as shown in the photo in Fig. 9, with two horizontal tethers attached to the left and right sides of the waist belt. Pulleys were used to route these cables from the motors above the mannequin. The motors actuating these two tethers were commanded to provide approximately 3 cm of cable travel. Fig. 8 shows a plot of the position tracking performance, comparing the reference position to the actual position as measured by the motor encoder. The plotted position is represented as millimeters of equivalent cable travel. The RMS error of the right-side motor is 1.97 mm and the RMS error of the left-side motor is 0.63 mm. As the motors and drivers are all identically tuned, the discrepancy between the left and right sides may be due to unequal loading of the two tethers by the mannequin, which is not self-supporting and therefore can lean towards one side or the other.

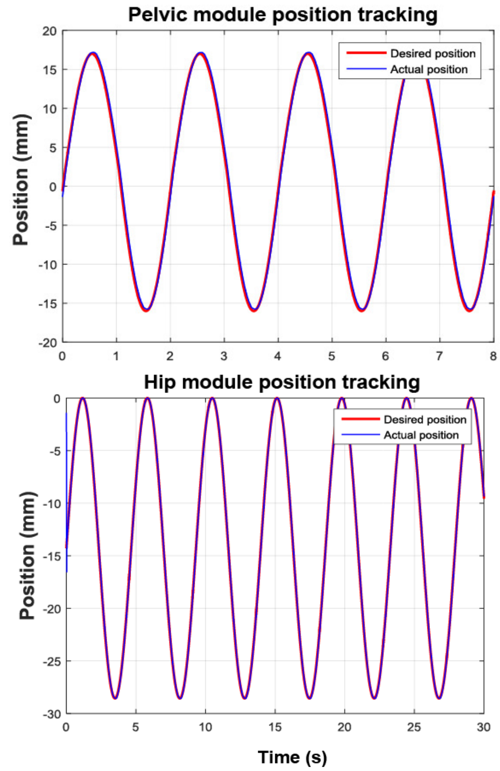

Figure 8. Top: Position tracking performance of the left-side motor of the pelvic module. The RMS error is 1.97 mm for the right motor and 0.63 mm for the left. Bottom: Position tracking performance of the right-side motor of the hip module. The RMS error is 0.08 mm for the right motor and 0.09 mm for the left.

C. *Position Tracking Performance of Hip Module*

To evaluate the tracking performance of the hip modules, they were tested with the mannequin model and the motors were commanded with a sinusoidal reference trajectory. The position tracking result is shown in Fig. 8, where the position units are millimeters of travel of the Bowden inner cable resulting from the rotation of the off-board motor. Negative travel corresponds to retraction of the cable, and this retraction in turn causes the body-worn hip drives to apply flexion assistance. We observed an RMS error of 0.08 mm for the right-side hip module and 0.09 mm for the left-side hip module.

D. *Motion Tracking and System Interoperability Evaluation*

In order to demonstrate the coordinated operation of the two modules, the system was tested on the mannequin model

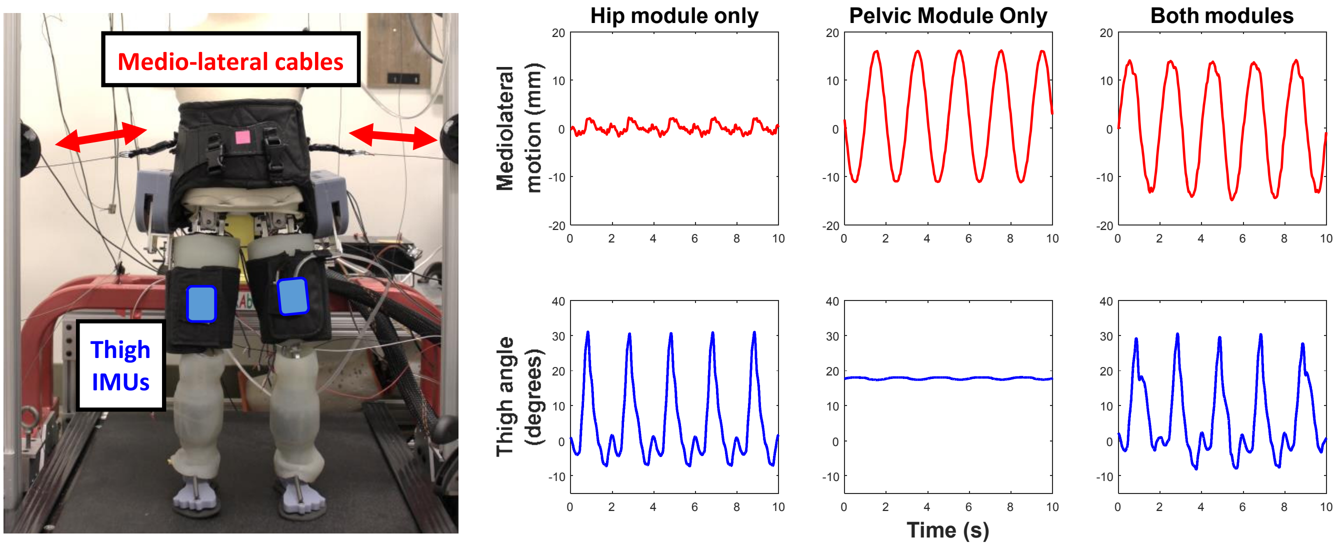

Figure 9. Experimental setup (left). Two cables attaching to the sides of the waist belt provide medio-lateral motion of the pelvis (denoted by red arrows). A colored marker placed in the middle of the waist belt was used to track this motion, using image processing in MATLAB. Hip flexion/extension angles of the legs were measured with IMUs on the front of the thigh (indicated in blue). Data was recorded during three modes of operation: hip module only (treadmill on), pelvic module only (treadmill off), and both modules combined (treadmill on).

with the goal of simulating gait. The test mannequin was suspended at a height such that its feet were flat on the treadmill (GK Mini, Mobility Research, USA) with its knees slightly flexed, prior to turning on the system actuation. The system was then operated in three different modes: hip module only, pelvic module only, and both modules operating together.

In this experiment, the hip module was always operated together with the treadmill. This allowed the mannequin to simulate walking in place without needing to move the entire system forward. In addition, as the hip module only provides flexion assistance, the treadmill facilitated hip extension to complete the gait cycle: the foot that was in contact with the treadmill surface would be pulled backwards by the treadmill, causing the stance leg to go into extension.

The hip module was programmed to induce sagittal hip motion similar to actual toddlers, with maximum flexion of up to 40°. The commanded position trajectory is trapezoidal in shape, with a rapid ramp up to maximum flexion shortly followed by a rapid ramp down to retract the drive unit pulley out of the way of the thigh shaft as the leg was extended by the treadmill.

The pelvic module was programmed to apply approximately 3 cm of medio-lateral sway. This was achieved by commanding the two motors with 180° out-of-phase sinusoidal reference trajectories, such that one cable pulled on the mannequin while the other cable was fed out. However, the trajectories were modified so that the leading cable pulled ahead slightly faster, in order to help take up slack in the cable and accommodate the flexing of the waist belt when switching the direction of motion. The pelvic motion was timed such that weight was shifted onto the stance leg before the hip module actuated the swing leg.

Hip flexion/extension angles of the legs were measured with IMUs placed on the thigh in pockets on the front of the thigh cuffs. To measure the medio-lateral motion of the pelvis, a brightly colored marker was placed in the middle of the waist belt and video of the test was processed in MATLAB to track its position. Pixel-to-distance conversion was determined from a ruler placed in-frame.

Fig. 9 displays the motion of the mannequin during the three different modes of operation of the system, with the hip module in the first column (treadmill on), the pelvic module in the second (treadmill off), and the combined operation of both in the third (treadmill on). Only a small amount of coupling between the two modules is observed. For both the hip module only case and the combined case, thigh angle ranged from -7° to 31°. This differs slightly from the average trajectory of the toddler hip reported in [33], but is still within the normal range of motion. The mediolateral motion data shows that approximately 2.8 cm translation was achieved, which is very close to the target range of 3 cm.

VII. Discussion and Future Work

In this paper we have presented a multi-robot system that can assist both pelvic motion and hip flexion, and showed an initial proof-of-concept demonstration on an instrumented mannequin model.

Future work on the hip module will focus on adding capabilities to actuate additional degrees of freedom and miniaturizing the design prior to human subject testing. Actuation of the more distal knee and ankle joints via Bowden cables is possible as well. In addition, for the initial evaluation experiment described in this paper, the pelvic module only applied mediolateral forces using 2 cable tethers. However, the system can have up to 6 tethers, allowing application of forces on the pelvis in any direction within the workspace. Future experiments will take advantage of the full capabilities of the pelvic module to explore what type of functionality is most useful, as this is still unclear. Finally, to ensure the system is as safe and robust as possible for children, human subjects studies will initially be conducted with healthy adult subjects.

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