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DESIGN AND PRELIMINARY EVALUATION OF A POWERED PEDIATRIC LOWER LIMB ORTHOSIS

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ABSTRACT

This paper describes the first stages of hardware development and preliminary assessment for a powered lower limb orthosis designed to provide gait assistance and rehabilitation to children with walking impairments, such as those associated with cerebral palsy and spina bifida. The design requirements, including range of motion, speeds, torques, and powers, are investigated and presented based on a target user age range of 6–11 years old. A three stage joint actuator is designed, built, and tested against the design requirements. The 0.6 kg actuator produced 4.2 Nm continuous torque and 17.2 Nm peak torque, and was able to run up to a speed of 480 deg/s. Backdrivability was characterized in terms of rotational friction, which was measured at 1.1 Nm. Finally, a 5.1 kg prototype orthosis was developed consisting of a hip segment, left and right thigh segments, and left and right shank segments, with four identical actuator prototypes installed in the thigh segments to actuate the hips and knees. Control electronics and a basic control structure were implemented to test the joint tracking capability of the orthosis against a predefined set of trajectories which were representative of pediatric gait patterns. Fitted to a dummy, the controlled limb successfully tracked the desired trajectories with a root-mean-square error of 9% and 4% of full scale for the hips and knees, respectively. With the dummy loaded with additional weight to representing a 32 kg child, the limbs also successfully tracked the trajectories with a root-mean-square error of 15% and 6% of full scale for the hips and knees, respectively.

INTRODUCTION

Cerebral palsy (CP) and spina bifida (SB) are two of a number of developmental and neurological conditions which occur in children and are accompanied by varying levels of gait impairment [1]. For every 10,000 children, there are approximately 36 school-age children with CP [1] and approximately 3 children ages 8 to 11 with SB [2].

Traditional overground gait training has been used for the rehabilitation of children with these disorders relying often on the aid of parallel bars, walkers, canes, or crutches [1, 3]. Treadmill-based gait training has also been used for rehabilitation of adult and children [3, 4], allowing for more steps in each training session and more task-specific training [4] which is important for improving performance [5]. Overhead harness systems are sometimes used to provide bodyweight support to the patient in these therapies [1, 3-5]. However, these treatments require one or two therapists to provide manual limb guidance to the patients and they are physically demanding to the therapists, which is often a limiting factor in the training session duration [6]. Also, the bodyweight support harness and the restriction of straight-line walking on a treadmill preclude the possibility of balance training afforded by overground gait training. In addition, these therapies lack objective, quantitative measures of performance improvement [1, 5]. A powered lower limb orthosis designed to enable and enhance overground and treadmill-based gait training would provide a better option to therapists which would not suffer from drawbacks associated with traditional and treadmill-based gait training.

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There are a number of powered orthoses for adults which are commercially available or currently under development. Researchers at the University of Delaware developed ALEX I and ALEX II [7, 8], powered orthoses for gait training with a treadmill. Unlike ALEX I which used linear actuators driven by electric motors, the ALEX II device supplied torques to the subject's hips and knees using servomotors through gearboxes, allowing for a broader span of permitted user limb lengths.

Ankle-foot orthoses (AFO) and knee-ankle-foot orthoses (KAFO) developed by Ferris' group [5, 9] used pneumatically powered artificial muscles to supply torque in the sagittal plane while remaining lightweight. Currently, since the compressor is external and tethered to the device, these devices are not portable [10].

The commercially-available Indego by Parker Hannifin Corporation [11, 12] is a 12 kg powered lower limb orthosis capable of providing 40 Nm torque at the hips and knees in the sagittal plane using brushless DC motors and a transmission system, while the ankle uses an AFO. The modular orthosis has quick disconnects at each joint, allowing for easier donning and doffing and transport of the system.

ReWalk by ReWalk Robotics [13-15] is a 23 kg commercialized battery powered height-adjustable device that uses motors to supply torque at the hips and knees in the sagittal plane while having a passive ankle.

Rex Bionics [16-18] developed a 36 kg self-supporting battery-powered lower limb exoskeleton. The user controls the device with a joystick, though there is some work using electroencephalography (EEG) signals to interface with the device [16].

The Hybrid Assistive Limb (HAL), commercialized by Cyberdyne Inc. in Japan [19-21], is a 15 kg battery powered device which applies torques to the hips and knees using DC motors through harmonic gear drives. Though originally designed as an augmenting device to increase users' load carrying capacity, the exoskeleton has been used for gait training in spinal cord injury and stroke patients [22].

Ekso Bionics developed Ekso GT [23, 24], a 20 kg device for rehabilitation of stroke and spinal cord injury patients. The device is commercially available and is designed to support the re-learning of correct step patterns and weight shifts.

The aforementioned orthoses are designed for use by the adult demographic. A more extensive list and thorough discussion on existing adult orthoses can be found in [6], [10], and [25].

In developing a new powered orthotic system specifically for the pediatric population, it is important to examine the unique needs and challenges presented by this demographic which may be different than those addressed by powered adult orthoses. First, the design of a powered pediatric orthosis must take into account the growth rate of a child, which is at about 3–3.5 kg and 6 cm per year [26]. As such, the device should accommodate a wide range of weights and sizes either by adapting to the child with adjustable components, or by having a customizable design which is easily adjusted or refitted as the child grows. Second, the device must be easy to learn and use by children who are developing in cognition [26], and therefore have poorer concentration skills, and attention spans as compared to adults. Third, the orthosis should ideally consider potential abnormalities in the patient's anatomy or physiology which may be present due to issues like improper posture or deformed bone structure, which often accompany disorders like CP and SB [1]. Lastly, since gait therapy in CP and SB often starts at a young age, as young as 5 years [27], the device should be safe and approachable – it should feel natural and comfortable to use and wear, and be smooth and quiet in operation.

The authors are aware of a few existing powered lower limb orthosis for pediatric patients. Lokomat [28, 29] is a commercially available treadmill-based device developed by Hocoma which is used primarily in gait rehabilitation of adults, but has been adapted more recently to allow gait therapy with children. In conjunction with an overhead bodyweight support system, the pediatric Lokomat is adaptable to the size of the child and supplies torque to the hips and knees using electric drives.

Researchers at MIT developed a pediatric version of their Anklebot [30, 31], an ankle-training device used in the seated position targeting children ages 6 to 10 years with neurological disorders like cerebral palsy. The low friction and backdrivable device is powered by linear actuators using DC motors and can supply a maximum dorsiflexion-plantarflexion torque of 7.21 Nm and inversion-eversion torque of 4.38 Nm, while the remaining internal-external degrees of freedom (DOF) are left passive.

Marsi Bionics developed ATLAS 2020 [32-34], a 10 DOF lower limb orthosis for children ages 3 to 12 with spinal muscular atrophy and similar pathologies. A variable stiffness series-elastic actuator driven by a motor through a harmonic drive actuates the flexion-extension DOF at the hip, knee, and ankle, whereas linear drives powered by brushless DC motors actuates the abduction-adduction DOF at the hip and the eversion-inversion DOF at the ankle. Its dimension can be adjusted to accommodate for the growing child. Clinical trials are underway with the latest prototype, the ATLAS 2030 [35], a similar powered lower limb orthosis weighing 14 kg and targeting children ages 3 to 14.

Copilusi [36, 37] has developed a modular knee orthosis with multiple rotation axes targeting children ages 4 to 7 years. In this system, a single servomotor drives two flexible steel cables through a pulley system to apply flexion or extension torque at the knee.

Canela [38] published the description of an orthosis targeting children ages 7 to 17 with cerebral palsy. The hips, knees, and ankles are powered by brushless DC motors and the segment lengths are adjustable by means of telescopic bars. Androwis et al. [39] converted a passive hip-knee-ankle orthosis to provide 35 Nm of torque at the joints using servomotors and gear transmissions for children with cerebral palsy. Lastly, Giergiel et al. [40] developed a stationary device

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that drives the hips, knees, and ankles using servomotors for children who weigh 25 kg.

In addition to the fact that none of these devices has progressed to commercial reality except the Lokomat treadmill system, each of these pediatric lower limb orthoses is limited in one or more aspect. In particular, of the devices that are mobile (e.g., [32-39]), most have few publications available or appear to not be further researched, or are heavy or bulky which can make them intimidating or difficult to use by children. The stationary devices (e.g., [28-31, 40]) are limited since they only provide rehabilitation capabilities and cannot be used as assistive devices in a community setting. The MIT [30, 31] and Copilusi [36, 37] devices are only concerned with the ankle and knee, respectively. An ideal lower limb orthosis should have either an active or passive joint at each hip, knee, and ankle.

This paper describes the work of Cleveland State University (CSU) in developing an innovative powered lower limb orthosis for children. This is motivated by the importance of early orthotic intervention to avoid or reduce long-term ambulatory issues in the pediatric population [41] and the lack of lightweight orthotic devices for children without the aforementioned limitations. The CSU orthosis takes inspiration from the design characteristics of the Indego exoskeleton [11, 12] and is being developed in collaboration with the Parker Hannifin Human Motion and Control team. The device is being developed to provide assistive and rehabilitative capabilities for pediatric use and is specifically designed to be small, lightweight, and compliant while still providing adequate torque and speed along the sagittal plane at the hips and knees.

In the following sections, the design requirements for the CSU orthosis are presented, including target age range, expected anthropomorphic data, the anticipated biomechanics of walking for this population, and design constraints for the joint actuators and orthosis frame. The design of the actuators and orthosis will be described and preliminary experimental results will be provided. The paper will conclude with lessons learned and directions for future work.

DESIGN REQUIRMENTS

The orthosis presented in this paper is designed to provide walking assistance and gait rehabilitation for children ages 6 to 11 with gait impairments. The age of 6 is selected as the lower bound to ensure sufficient attention span to focus on specific tasks, sufficient communication skills to follow a series of commands, and sufficient motor skills [42], all of which are important for engaged rehabilitation therapy as well as effective use of a powered orthosis. The age upper limit of 11 years is selected because children around that age are tall enough (around 150 to 155 cm on average [43]) to consider use of one of the commercially available adult orthoses. For reference, the Indego exoskeleton reportedly accommodates individuals 155–191 cm in height [44], ReWalk 155–200 cm [45], Ekso Bionic 150–190 cm [46], and Rex Bionics 142–193 cm [47]. Table 1 presents average weight and height characteristics for children of ages 6, 8, and 11 years.

Table 1. Height and weight for target population (average ± standard deviation) [42]

Age (years)	6	8	11
Weight, Male (kg)	24.3±6.5	31.3±10.2	46.6±16.4
Weight, Female (kg)	23.6±6.3	31.9±14.4	47.5±18.9
Height, Male (cm)	119.3±6.3	131.6±9.6	149.9±9.0
Height, Female (cm)	119.2±6.9	131.3±12.0	150.4±8.0

Various clinical trials have been conducted using powered lower limb orthosis for adults which actuate only the hips and knees: for example, ALEX II for stroke patients [48], Indego for spinal cord injury patients [49], and ReWalk for spinal cord injury patients [50]. The successful use of these devices suggest that an orthosis with hip and knee actuation in the sagittal plane is sufficiently capable for assistive and rehabilitative use by those with either complete paralysis or partial gait impairments. The pediatric orthosis design adopts this approach, providing hip and knee actuation in the sagittal plane. The hip abduction-adduction DOF should be sufficiently restrained to provide lateral stability but sufficiently compliant to allow natural walking and balancing to occur. Readily available AFOs can be used for passive support at the ankle.

Since the orthosis is specifically intended for children, there are some special considerations that should be addressed in the design. First, the device should be usable by children of a wide range of ages and therefore accommodate a variety of weights and heights. In addition, the device should account for the growth of the child, both in weight and height. This can be accomplished by designing the orthosis with adjustability or, alternatively, the orthosis frame could be made specific to the child, providing an improved fit with custom contouring of the interfacing components. The latter is selected over the former for the CSU pediatric orthosis because of the added advantage that the device would be able to accommodate anatomical or physiological abnormalities, which can occur in CP and SB [1]. In addition, the electronics and actuators should be designed as modular units so that they can be inserted into the custom made orthosis frame. The orthosis frame could be designed with parametric modeling using inputs of physiological dimensions measured from the child to create a custom frame model. This model could be produced by employing additive manufacturing techniques. Second, the orthosis should be lightweight, compact, and quiet. It should ideally be much lighter than the child using the device (24 kg for children of age 6 years). For reference, the Marsi Bionics ATLAS 2030 device weighs 14 kg [35]; the designed device should be around this weight or lighter. Finally, the device should be usable not only as a gait rehabilitation device but also as an assistive device. As such, it should be usable without a treadmill and should not be tethered.

The natural walking cadence in adults is around 101 to 120 steps per minute while the cadence in children is typically higher, e.g., around 144 steps per minute for children 7 year old [51]. This indicates that, for children on the younger side of the target pediatric population, gait period is expected to be around 0.8 s and slower for older children. However, when the orthosis

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is used in practice, the gait period would probably be slower than this, particularly when used by those with more severe gait impairment. Those with cerebral palsy, for example, tend to have reduced walking speed [52]. Since published data on the biomechanics of walking for children is limited, typical adult hip and knee trajectories, weight-specific torques, and weight-specific powers taken from Winter [51] are shown in Fig. 1. This should be adequately representative of most children within the target age range since children 7 year old have a gait which is similar to that of adults [26]. Extrapolating this data and considering the aforementioned pediatric weight and cadence requirements, the expected minimum and maximum hip and knee range of motion, velocity, torque, and power are presented in Table 2. The sign convention for positions, velocities, and torques used in this paper is positive for flexion and negative for extension. The orthosis should be designed to provide these gait characteristics to allow for its use by a broad range of children.

ACTUATOR DESIGN

For simplicity in the first prototype, a single actuator design, as seen in Fig. 2, is shared for the hips and knees. Each actuator is powered by a 70 W brushless DC motor, which should be adequate, even after transmission losses, to satisfy the maximum power requirement of 46.4 W at the knee. The torque provided by the motor is transferred through a three stage toothed belt transmission system with a total speed-reduction ratio of 40.6:1, which provides a nominal output speed of 375 deg/s, a continuous torque of 5.4 Nm, and a stall torque of 35.7 Nm. This is without considering frictional losses or

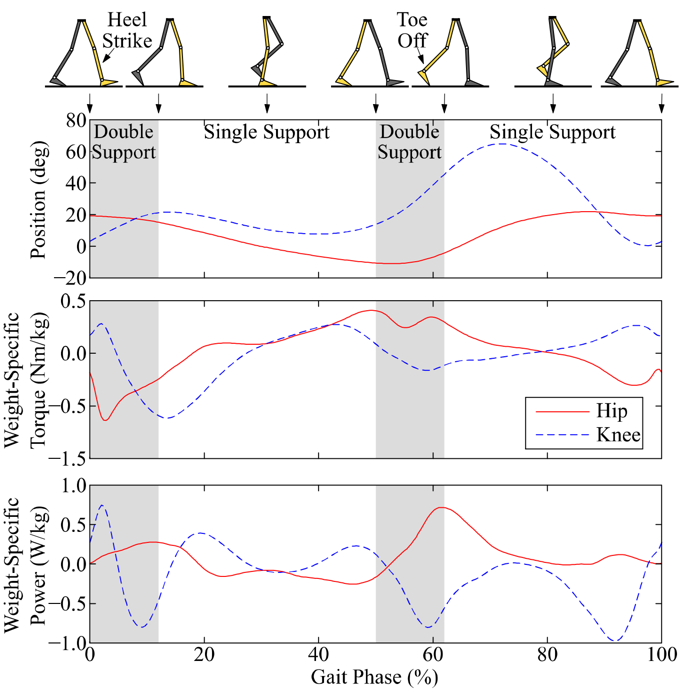

Figure 1. Representative mean data of the hip and knee based on a healthy gait cycle [44]

Table 2. Estimated gait characteristics for a 1.1 second period gait cycle

	Minimum	Maximum
Hip range of motion (deg)	−11	22
Knee range of motion (deg)	0	65
Hip velocity (deg/s)	−82	159
Knee velocity (deg/s)	−369	312
Peak hip torque, 8 years (Nm)	−19.0	13.8
Peak knee torque, 8 years (Nm)	−19.4	8.9
Peak hip torque, 11 years (Nm)	−28.5	19.2
Peak knee torque, 11 years (Nm)	−29.2	13.3
Peak hip power, 11 years (W)	−12.1	33.9
Peak knee power, 11 years (W)	−46.4	35.4

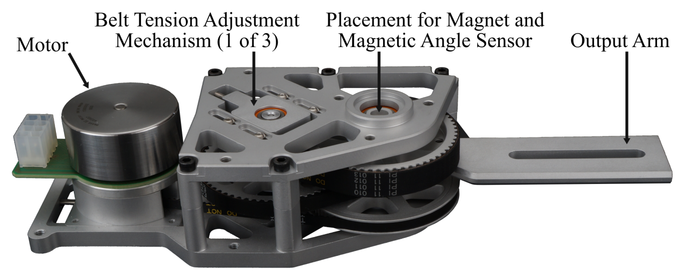

Figure 2. Prototype actuator for the pediatric orthosis

dynamics in the actuator; the actual output power, speed, and torque observed in operation will differ from these numbers in accordance with the actuator's efficiency and compliance characteristics. Belts are used in the transmission to allow compliant behavior under load while also being quiet and lightweight.

Screw-adjustable belt tensioning mechanisms are designed into the actuator to provide the ability to change the tension in the belts, and consequently the level of friction in the transmission and the backdrivability and compliance behaviors. The actuators have a modular design with interchangeability in mind. They are thin and lightweight, with a height of 46 mm and weight of 0.6 kg.

ACTUATOR EVALUATION AND GAIT TRACKING

The actuator's torque capabilities were determined by measuring the actuator's output torque using a force gauge holding the output arm stationary while driving the motor in a constant current mode of control. This was applied incrementally from zero current to four times the continuous current rating of the motor. The result, as shown in Fig. 3, was an approximately linear curve which remains below the ideal transmission line. The actuator achieved a maximum continuous output torque of 4.2 Nm and a maximum peak output torque of 17.2 Nm. Breakaway torque was also measured when operating within the continuous current region in both directions, giving some approximate bounds to the accuracy of the measured output torque, at least for the continuous operating region. Static friction can be estimated as the half difference of the breakaway torques in either direction, calculated as 1.0 Nm.

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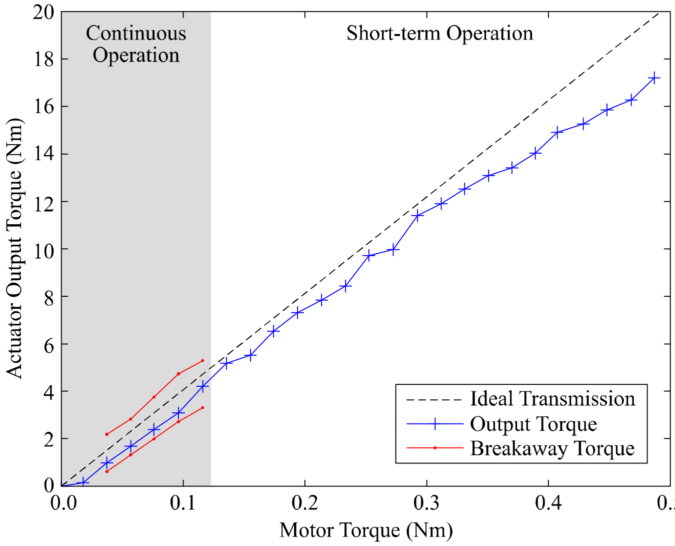

Figure 3. Actuator output torques for constant current controlled motor

The actuator speed was determined by running the actuator unrestricted until it reached its maximum speed of 480 deg/s at the output, tested in both directions.

To quantify actuator backdrivability, which is primarily characterized by frictional and inertial effects, the level of friction was measured in a low and a high belt tension state. Testing was performed with various belt tension levels, ranging from low tension to high tension. However, the lack of means to measure belt tension in this prototype left this without quantification. Friction was calculated based on the electrical current draw of the motor when running the actuator at various constant speeds up to 400 deg/s in both directions, resulting in a friction torque of 0.8 and 1.1 Nm at the output for the low and high tension states, respectively. The measured torque did not vary significantly when operating at different speeds, suggesting that Coulomb friction is dominant over viscous friction when operating at these speeds. The 1.1 Nm Coulomb frictional torque closely matches the previously measured 1.0 Nm static frictional torque, and was within measurement error.

To evaluate the functionality of the actuator in the orthosis, gait trajectory tracking experiments were conducted using a dummy constructed to represent an average 8 year old child [43, 53], as shown in Fig. 4. For an 8 year old child weighing 32 kg, the thigh would be expected to weigh around 3.2 kg with a 344 kg·cm^2 moment of inertia, and the shank and foot combination would be expected to weigh around 2.0 kg with a 471 kg·cm^2 moment of inertia [43, 53]. For the 10.1 kg dummy as configured in this paper, the thigh weighed 1.2 kg and shank weighed 1.2 kg, whereas the orthosis thigh component weighed 1.8 kg and shank interface component weighed 0.3 kg, contributing to the total orthosis weight of 5.1 kg as configured in this paper. Electrical power was sourced externally at this stage of development so this does not include the weight of a battery. The gait tracking experiments were done for two cases:

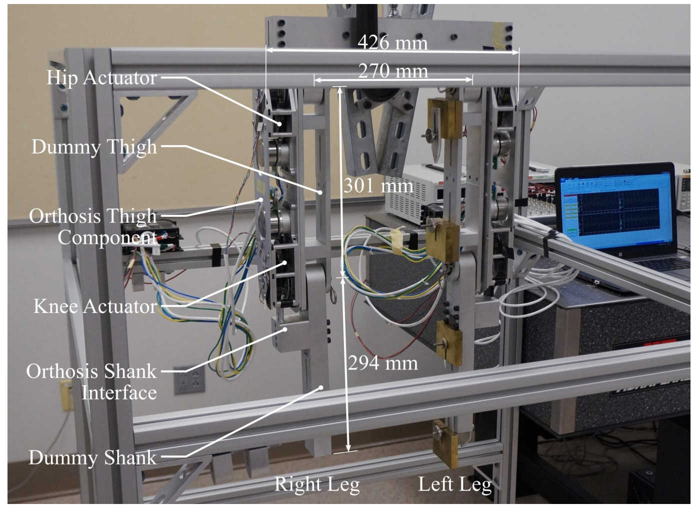

Figure 4. Experimental setup for single leg gait tracking using the prototype orthosis

first, moving the dummy without any additional mass (unloaded case), and second, moving the dummy with weights attached for a total dummy thigh weight of 3.2 kg and shank weight of 2.2 kg (loaded case), close to that of an average child 8 years old. The weights were positioned such that the center of mass and moment of inertia for the dummy body segments approximately matched that of the child.

The hip and knee actuator output torques were controlled using dSPACE MicroLabBox running at 1 kHz through servo-amplifiers powered by a power supply at 38 V. Each joint angle was directly measured using a magnetic angle sensor. The joint velocity was estimated using the measured motor velocity from Hall effect sensors in the motor scaled by the transmission ratio of 40.6:1. Both angular position and velocity signals were conditioned by an experimentally tuned 35 Hz low-pass filter to reduce noise.

A decentralized proportional-derivative (PD) position controller was applied to enable the orthosis to track the desired gait trajectory shown in Fig. 1 with a gait cycle period of 1.1 s. This controller can be described by the equation

$$\begin{bmatrix} \tau_{\text{hip}} \\ \tau_{\text{knee}} \end{bmatrix} = \begin{bmatrix} P_{\text{hip}} + D_{\text{hip}}s & 0 \\ 0 & P_{\text{knee}} + D_{\text{knee}}s \end{bmatrix} \begin{bmatrix} \tilde{q}_{\text{hip}} \\ \tilde{q}_{\text{knee}} \end{bmatrix} \tag{1}$$

where τ_{hip} and τ_{knee} are the joint torques, $\tilde{q}_{\text{hip}}$ and $\tilde{q}_{\text{knee}}$ are the joint position errors, P_{hip} and P_{knee}, are the controller proportional gains, and D_{hip} and D_{knee} are the controller derivative gains. The same controller gains were used for both the unloaded and loaded test cases, tuned experimentally based on the loaded case. The proportional and derivative gains for the hip were 0.349 Nm/deg and 0.0436 Nm·s/deg, respectively, and the proportional and derivative gains for the knee were 0.349 Nm/deg and 0.0262 Nm·s/deg, respectively. The resulting positions, velocities, and applied joint torques for a single gait cycle can be found in Fig. 5 for the unloaded case and Fig. 6 for the loaded case.

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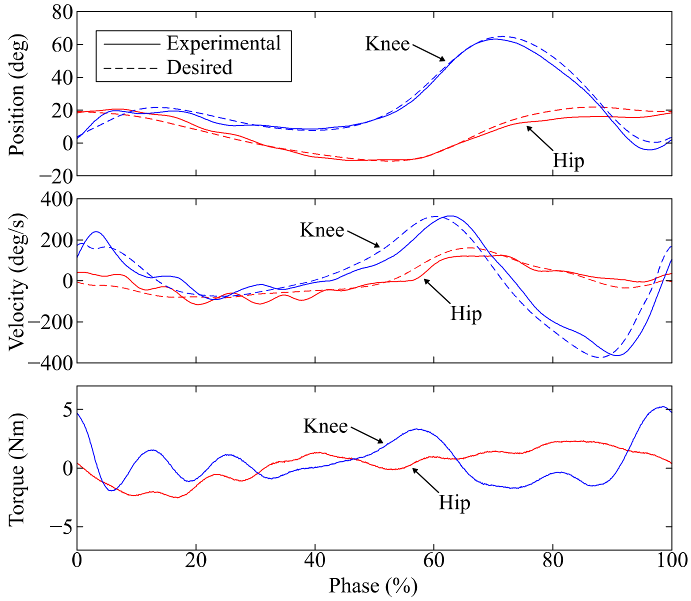

Figure 5. Gait tracking of the unloaded dummy fitted with the powered orthosis using PD control

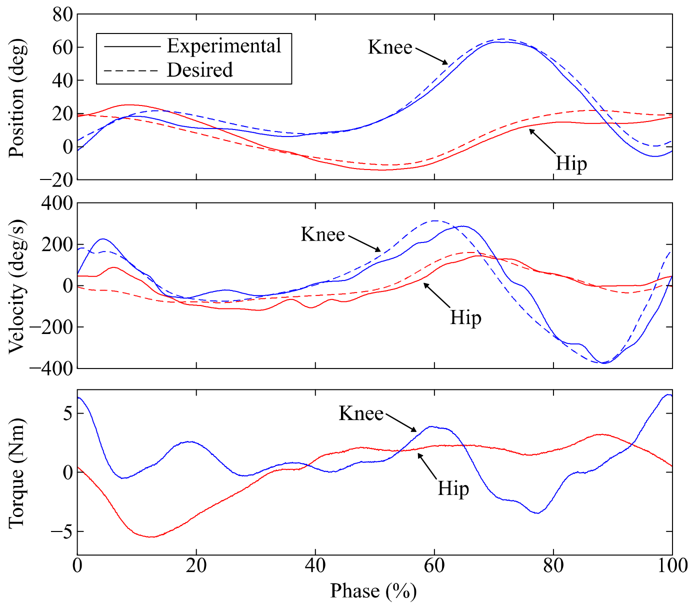

Figure 6. Gait tracking of the loaded dummy fitted with the powered orthosis using PD control

The orthosis successfully tracked the desired gait trajectories for both test cases with the PD controller and showed consistent behavior between gait cycles. The joint trajectories tended to overshoot the desired trajectories when decreasing due to the help of gravity, particularly for the knee when extending after toe-off. In addition, the hip deviated from the desired trajectory at this time and around heel strike due to the cross-coupling of the knee and hip. The root-mean-square (RMS) position errors in degrees and as a percent of full scale (FS) taken over six gait cycles can be found in Table 3. The peak and RMS torques over the same timespan are reported in Table 4. The largest peak torque was less than the actuator's tested peak torque capabilities, and the largest RMS torque was larger than the actuator's tested continuous torque capabilities. This suggests that higher controller gains could be used to improve tracking performance, however erratic behavior limited the gains and gait period selected in this paper.

CONCLUSIONS

There are presently few examples of powered lower limb orthoses which can provide rehabilitation to the pediatric population. Existing solutions are often limited insofar that they cannot be used in a community setting, are often heavy or bulky, or does not consider all lower limb joints. Given there is a need for a device without these limitations, this paper presented the first stages of development and evaluation of such a device.

The design requirements and details of the resulting actuator and orthosis were presented. Prototype actuators were built, each weighing 0.6 kg, and used to power the hips and knees of the prototype orthosis, weighing 5.1 kg. The actuators were tested and shown to be backdrivable with a friction torque around 1.1 Nm at the output. It could provide 4.2 Nm continuous torque and 17.2 Nm peak torque, and can operate up to 480 deg/s. For an unloaded dummy with a 1.4 kg thigh mass and 1.2 kg shank mass fitted with the prototype orthosis, gait tracking performance using root-mean-square position error was 9% of full scale for the hip and 4% for the knee. For a loaded dummy with a 3.4 kg thigh mass and a 2.2 kg shank mass, gait tracking performance was 15% for the hip and 6% for the knee. Based on the presented preliminary experimental results, the actuator and orthosis design show promise for gait assistance and rehabilitation in children around 8 years of age and possibly older.

Various issues were encountered throughout the experimental testing of the actuator and the orthosis. First, the performance of the actuator was significantly dependent on the belt tension. When belt tension was too low, belt slippage can

Table 3. Hip and knee RMS position errors for the unloaded and loaded experimental cases

	Unloaded Case	Loaded Case
Hip RMS error	3.0 deg (9% FS)	5.1 deg (15% FS)
Knee RMS error	2.7 deg (4% FS)	3.9 deg (6% FS)

Table 4. Hip and knee peak and RMS torques for the unloaded and loaded experimental cases

	Unloaded Case	Loaded Case
Hip peak torque	2.5 Nm (15% PT)	5.5 Nm (32% PT)
Knee peak torque	5.3 Nm (31% PT)	6.6 Nm (38% PT)
Hip RMS torque	1.4 Nm (33% CT)	2.6 Nm (61% CT)
Knee RMS torque	1.8 Nm (43% CT)	2.3 Nm (56% CT)

PT: peak torque in the actuator tests (17.2 Nm)
CT: continuous torque in the actuator tests (4.2 Nm)

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take place and potentially cause damage to the belts. When belt tension was too high, the actuator was much less compliant. In addition, the actuator exhibited much higher friction and made otherwise unheard noises, possibly due to the imperfect teeth meshing of the belts and sprockets which is exacerbated by the high belt tension. Second, without means of directly measuring belt tension in the actuators, it proved difficult to set the tensions of the belts to similar levels. For the experiments in this paper, belt tensions were tuned by feel until the resistance was similar when backdriving the actuator by hand. Third, erratic behavior was observed when running the gait tracking control experiments. It is suspected that this is because the joint velocity measurement used in control was not collocated with the joint, the value of which was estimated based on motor velocity. For the estimation of joint velocity to be accurate, the dynamic coupling between the actuator output and motor must be highly stiff, which is a poor assumption due to the compliance in the actuator. In addition to the encountered issues, the gait tracking experimental validation reported here is limited in that there was no environmental interaction. This would be observed in reality through ground reaction forces when walking.

Future work will entail developing a better estimation of joint velocity using the measured joint position and motor velocity, which should eliminate erratic behavior experienced in these experiments. Furthermore, by measuring or estimating the motor position, a more sophisticated controller could take advantage of the inherent compliance in the actuator while also eliminating the erratic behavior. A repeatable method for belt tensioning should be developed to obtain consistency in the actuators, thereby providing more similar joint behaviors and improving the ability of future experiments to produce repeatable results. The orthosis should eventually be tested in a manner which incorporates interaction with the environment such as walking on a treadmill. Finally, further design revisions will allow for additional weight reduction.

The technology described herein is Cleveland State University protected intellectual property with patent pending [54].

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