REHABILITATION ROBOTICS

Robot-driven downward pelvic pull to improve crouch gait in children with cerebral palsy

J. Kang,[1] D. Martelli,[1] V. Vashista,[2] I. Martinez-Hernandez,[3] H. Kim,[3] S. K. Agrawal[1,3]*

Children with cerebral palsy commonly exhibit an abnormality called crouch gait, which is characterized by excessive flexion of the hips/knees and weak plantar flexor muscles during the stance phase. One of the major reasons for this pathological gait is weakness in soleus muscles. During the mid-stance phase of gait when the toe and heel are both on the ground, the soleus keeps the shank upright and facilitates extension of the knee angle. It also provides propulsive forces on the body during the late stance phase of the gait cycle. We hypothesized that walking with downward pelvic pull will (i) strengthen extensor muscles, especially the soleus, against the applied downward force and (ii) improve muscle coordination during walking. We then tested a robotic training paradigm to improve both posture and gait of children with crouch gait. In this paradigm, participants with crouch gait were subjected to downward pelvic force when walking on a treadmill, provided by a cable-driven robot called Tethered Pelvic Assist Device. Electromyography of soleus and gastrocnemius muscles and walking kinematics of the participants showed the feasibility of this training, enhanced upright posture of the participants, and improved muscle coordination. In addition, walking features of these participants, such as increased step length, range of motion of the lower limb angles, toe clearance, and heel-to-toe pattern, improved. This robotic training method can be a promising intervention for children with cerebral palsy who have a crouch gait.

INTRODUCTION

In the United States, 3.6 children out of every 1000 school-aged children have a diagnosis of cerebral palsy (CP) (*1*). These children show abnormal gait patterns, for example, jump gait, equinus gait, and crouch gait (*2*, *3*). The crouch gait is characterized by excessive flexion of the hips, knees, or ankles (*2*, *4*, *5*). Children with crouch have slow walking speed, reduced range of motion of the joints (ROM), small step length, large body sway, and absence of a heel strike (*6*). In addition to other complications, crouch gait increases the energy cost of walking, causes pain, and results in joint degeneration over time (*7*).

Crouch gait is caused by weak extensor muscles that do not produce adequate muscle forces to keep upright posture under gravity (*7*, *8*). Among the extensor muscles, the soleus plays an important role to prevent knee collapse during the middle of stance phase (*9*). The soleus muscle keeps the shank upright during the mid-stance phase of the gait to facilitate extension of the knee. When dorsiflexion of the foot increases by weak soleus muscles, the upper part of the shank leans forward, and this contributes to the flexion of the knee. In this posture, the body weight creates a larger flexion torque on the knees, which leads to collapse of this joint (*4*). The soleus is also responsible for propulsive forces on the human body during mid-stance to terminal stance phase. The activation timing of the muscles with respect to the gait cycle is also an important issue. If this timing is not appropriately synchronized with the gait, it results in poor kinematics of the leg and ground reaction forces (GRFs). If two muscles with opposite functions activate together, muscle efficiency is reduced, which leads to early fatigue. Coactivation of plantar flexor muscles and quadriceps is a major issue in crouch gait. During normal gait, plantar flexor muscles accelerate the center of mass forward during the late stance phase, whereas the quadriceps decelerate the center of mass during the early stance phase (*8*). Children with crouch gait are often observed to have earlier activation of gastrocnemius and soleus muscles, and this contradicts the function of quadriceps muscles, contributing to inefficiency of crouch gait (*8*). Dietz and Berger have also reported that plantar flexor muscles are impaired in CP children, showing lower amplitudes and poor timing, compared with healthy peers (*10*).

Surgery is often recommended for children with CP to lengthen the tightened flexor muscles. However, previous results show that this does not improve walking because extensor muscles still remain weak (*7*). Although early literature suggested that muscle strengthening of the legs can worsen spasticity and coactivation in the muscles, more recent research disputes these claims (*11*, *12*). Different strength training strategies have shown increased muscle strength without worsening spasticity or muscle coactivation (*12*–*15*). Currently, different resistive strength training schemes are used in clinical practice with leg presses, rubber bands, or loaded sit-to-stand to strengthen the extensor muscles. Whether the effects of strength training translate to overground walking still remains under debate (*16*, *17*). One possible reason suggested by researchers is that the task during the training is not directly related to walking. Another method often used in clinical practice to retrain walking in children with CP is treadmill training with partial body weight suspension (*18*–*20*). However, no studies with partial body weight suspension report postural correction of the crouch gait.

In this study, we took an approach opposite to what is used in conventional therapy with these children. Instead of partial body weight suspension during treadmill walking, participants are trained to walk with a force augmentation. The scientific rationale behind this study is to strengthen the soleus muscles, which is a major pathological cause for the crouch gait. It was previously shown that soleus muscles are activated more strongly among the lower limb muscles when extra weight is added on the human body during gait (*21*). This is because the soleus is the major weight-bearing muscle during the single stance support. From these observations, our study is designed to apply an additional downward force on the human body to intensively retrain the activity of the soleus muscles. A downward force equivalent to 10% of

[1]Department of Mechanical Engineering, Columbia University, New York, NY 10027, USA. [2]Department of Mechanical Engineering, Indian Institute of Technology Gandhinagar, Gujarat 382355, India. [3]Department of Rehabilitation and Regenerative Medicine, Columbia University, New York, NY 10032, USA.
*Corresponding author. Email: sunil.agrawal@columbia.edu

the body weight was chosen on the basis of the results of healthy children carrying backpacks—this was the minimum weight to show notable changes in posture or gait parameters during walking (*22*, *23*).

A cable-driven robot called Tethered Pelvic Assist Device (TPAD) (*24*) was programmed to apply a downward pull force through the center of the pelvis while the participants walked on a treadmill. TPAD consists of a lightweight belt worn by a participant on the pelvis to which several wires are attached. The tension in each wire was controlled in real time by a motor placed on a stationary frame around the treadmill, based on real-time motion capture data from cameras. TPAD is a unique cable-driven robotic device because it applies external forces on the human body during motion, and the training with this device is distinctive because it provides necessary strength and coordination training while walking.

Recently, robotic training paradigms for CP children have been proposed. The most widely used device is Lokomat, which is a rigid robot that moves the lower limbs in the sagittal plane using position control (*25*). Studies with this device have shown mixed outcomes in gait improvements, and this was attributed to lack of the child's active participation in the study (*25*–*27*). An overground gait trainer, "CPWalker," was used to apply partial body weight support with control of hip flexion-extension and knees in position mode (*28*). Training with CPWalker was performed for 5 weeks with three CP children, but the outcomes for the posture were inconsistent (*29*). Another study conducted by Lerner *et al.* (*30*) used a robotic exoskeleton that provides torque assistance to change the posture during walking and enhance the muscle's ability to extend the limbs. Wearing the exoskeleton changed the muscle patterns and joint angles positively, but the training effects without the exoskeleton were not presented.

Compared with other robotic devices that could be potentially used for gait training of children with CP, TPAD offers some unique advantages: (i) TPAD does not add inertia to the human body. TPAD only adds 320 g on the child, which is 10 to 20 times smaller compared with other devices (*24*). This is important for children with small body size and weight. Added inertia on the child's body can result in undesirable motion during gait such as reduced anterior-posterior acceleration of the center of mass (*31*). (ii) TPAD does not add rigid links on the human body. This feature allows free movement of the limbs with no restrictions on any degree of freedom (DOF) and reduces the therapist's effort to align the joint axes of the device with the human limbs or to adjust the size of the rigid links to the child's body. Because gait is a sequential coordination of limbs in three dimensions, restriction of specific DOFs can alter the entire gait pattern. For instance, the restriction of the leg in the sagittal plane, such as what Lokomat provides, can cause issues of lateral balance after the intervention, because the gait pattern is practiced within an artificial environment where the hip abduction/adduction muscles are not actively involved in the motor control (*32*). (iii) TPAD imposes an overall external force during movement and allows the users to coordinate and control the limb DOF. Because CP is caused by impaired central nervous system (*4*), the type and level of lesion are quite different among individual children (*33*). Instead of providing assistance to each joint with a predefined phase-dependent torque or displacement, TPAD promotes variability among joints and provides flexibility to the users in learning, as suggested by theories of human motor learning (*34*).

Six children diagnosed with CP and exhibiting crouch gait underwent 15 training sessions of 16 min each, over a duration of 6 weeks. We hypothesized that, as the study participants raise their center of mass against the applied downward force from TPAD during walking, they will improve control of their ankle plantar flexor muscles, especially the soleus, and improve their crouched posture. The muscle strength and coordination were investigated using electromyography (EMG) data from the first and after last sessions of training. Kinematics and GRF were monitored continuously throughout the training.

RESULTS

In each TPAD training session, participants performed 4 min of free walking before training, 16 min of TPAD walking with continuous

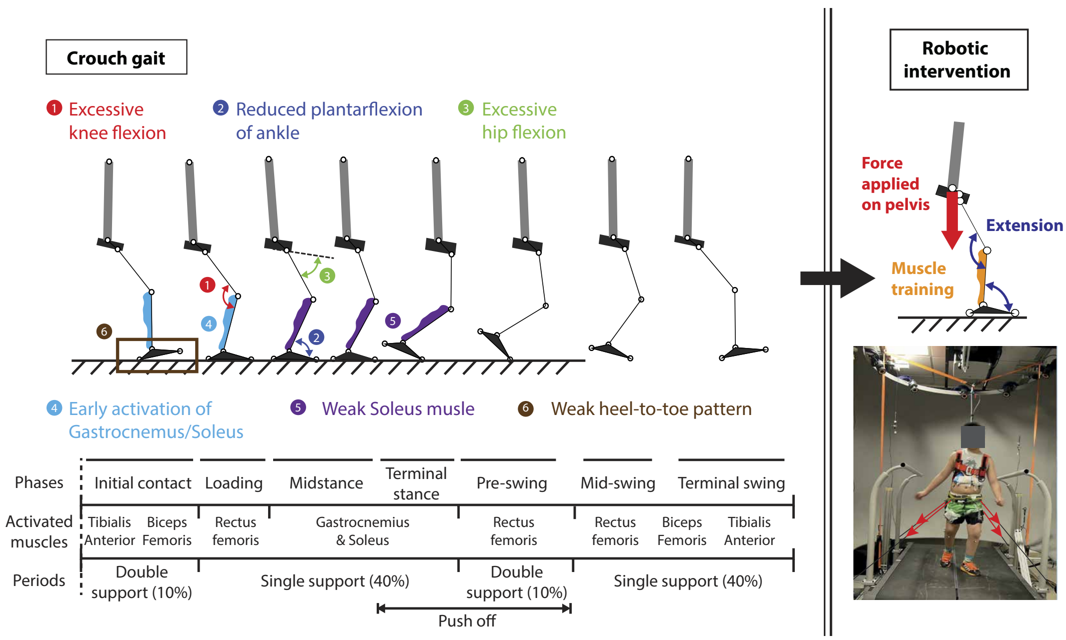


Fig. 1. Gait characteristics of crouch gait and TPAD training. Motion capture data from a child with crouch gait during free walking before the training is illustrated, which was collected using a Vicon system. The gait pattern of the crouch gait can be described as excessively flexed hips/knees and reduced plantar flexion of the ankle. TPAD training applies a downward force on the stance leg to improve the muscle patterns necessary for extension of the limbs while bearing the child's weight. Here, the soleus muscle engaged during the single stance phase is intensively trained to improve the crouch gait.

downward force equivalent to 10% of body weight, and another 4 min of free walking after training. The free walking data before and after training were collected for comparison to investigate the effects of TPAD in the short term. Participants performed 3 training sessions per week over 5 weeks, with a total of 15 sessions.

The study was designed to observe the training effects on different gait parameters. After finishing all 15 training sessions, an additional session, referred to as the 16th session, was added, in which the children walked with the same treadmill speeds as their first sessions. EMG was recorded only during the 1st and 16th sessions. EMGs of soleus and gastrocnemius were monitored for their magnitudes and timing during the gait cycle. The normal EMG pattern of soleus and gastrocnemius in healthy children has a peak around 40 to 50% of the gait cycle after heel strike. However, children with CP often activate these muscles much earlier during the gait cycle, during the initial stance phase. If gastrocnemius muscle activates too early, it hinders the knees from extension because it is a knee flexor (*4*). Earlier studies have reported that, when weight is added to the body during walking, soleus and gastrocnemius are active during the latter part of the stance phase. If these muscles are trained to activate during the stance phase of walking with TPAD, we hypothesize that TPAD training will (i) strengthen soleus muscles and (ii) delay the early activation of gastrocnemius/soleus muscles. Especially, TPAD will intensively train the leg during the single stance phase, that is, middle of the stance phase when the lower limb should be straight, whereas only one leg supports the extra downward force. Besides EMG, kinematics and GRFs were also monitored. For the lower limb kinematics, hip, knee, and ankle angles were monitored to verify if (iii) the children extend the joints more during the stance phase. Each joint angle was computed in sagittal plane, as suggested in (*35*). As described in Fig. 1, children with CP typically do not exhibit distinct heel strike and toe-off during walking, unlike typically developing children (*2*). The heel strike plays a role to absorb the shock during the initial contact, and the push-off facilitates forward propulsion of the body (*9*). For children with CP, these two gait events are often merged together, and the GRF shows up as a single continuous force, which is undesirable. The distinct pattern of the heel and toe is important to achieve a stable stance and a strong push-off.

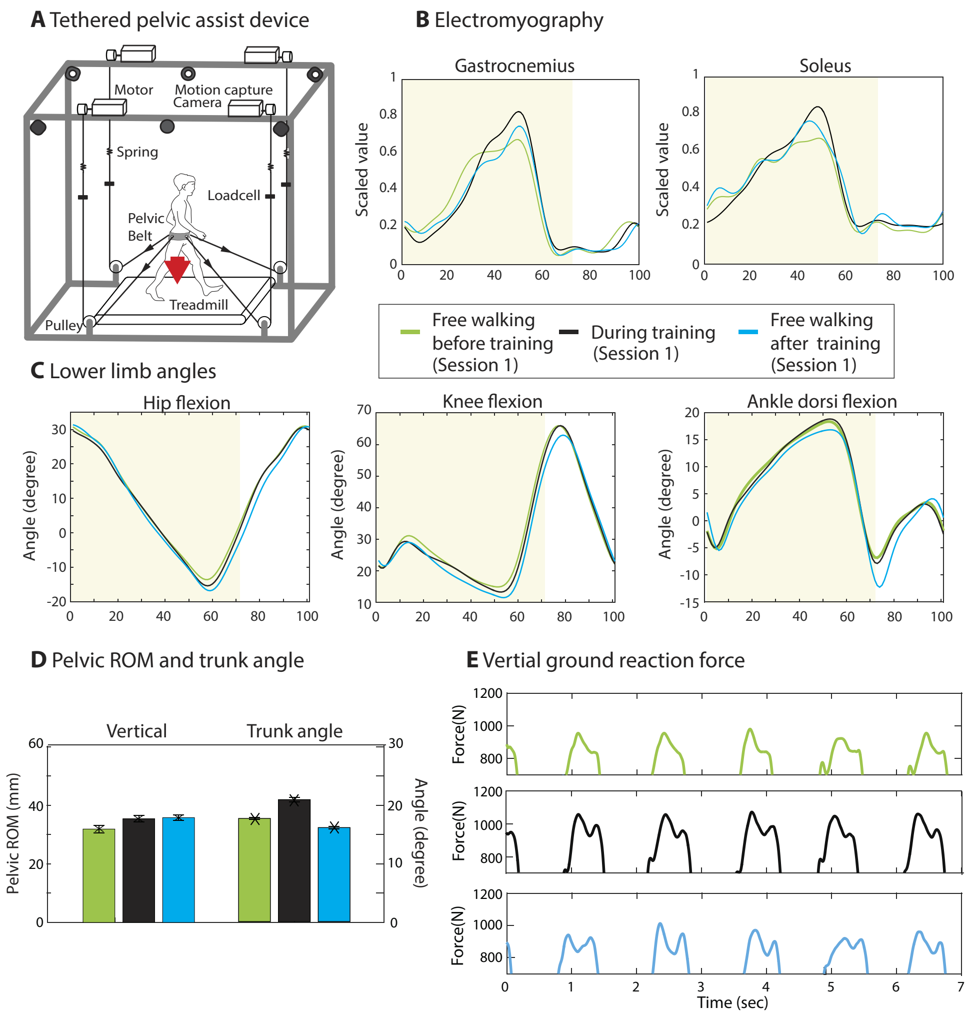


Fig. 2. System setup and results of a representative participant in a single training session. (**A**) TPAD system setup. (**B**) EMG data, (**C**) lower limb angles for hip, knee, and ankle, (**D**) pelvic translational ROM and trunk angle, and (**E**) vertical GRFs during session 1 including free walking before training, during training with force, and free walking after training. Data during training (black line) were recorded at the seventh minute of training, and data after training (blue line) were recorded at the third minute after training. Yellow shaded area presents the stance phase of the gait cycle. Trunk ankle is defined by two markers attached to the seventh cervical vertebra and sacrum, averaged over the gait cycle.

Changes in muscle forces and center of mass motion during a single training session: Representative participant

EMGs of two plantar flexor muscles in the lower limbs were measured to observe the effects of the external force on the pelvis. Muscle activations of these two muscles are displayed over the gait cycle, which starts with the heel strike of the leg. The yellow shaded areas in Fig. 2 (B and C) represent the stance phase of the gait cycle.

Figure 2B shows the EMG signals for a representative participant in session 1 during free walking and when downward forces were applied. EMG signals are scaled with respect to their maximum values during free walking before training. In Fig. 2B, increased peaks of gastrocnemius and soleus muscles were observed during the stance phase when the TPAD is active. Once the tethers were off, the peaks of gastrocnemius and soleus during post-training were still larger than pre-training. During the mid-stance phase (20 to 40%), the gastrocnemius shows smaller activation during the training and after the training. Figure 2C shows that the hip and knee are more extended than baseline during the stance phase. Ankle plantar flexion is also increased, closer to a normal plantar flexion angle of −20°, at push-off.

Figure 2D presents the pelvic translational ROM. Vertical ROM of the pelvis was larger after the TPAD training. The child was able to raise the center of mass more after walking with extra downward force. The second graph in Fig. 2D shows the trunk angle, which is the angle between the vertical and a line on the trunk, defined by two markers attached to the seventh cervical vertebra and the sacrum. Decreased trunk angle reflects improved crouched posture during post-training compared with pre-training. Vertical GRF was also presented in Fig. 2E. During the training and after training, the pattern of the GRF shows two explicit peaks, characteristic of a distinct heel-to-toe pattern.

Changes in gait during treadmill walking after multiple training sessions: Representative participant

Figure 3 shows different parameters during free walking before (green line) and after 15 training sessions (red line). In these plots, the treadmill speeds before and after 15 training sessions were set the same to remove the effect of walking speed on gait pattern (*36*). Figure 3A shows changes in EMG signals. Both EMG signals were scaled with the maximum value of baseline at the first session (before the first training). The soleus shows higher peaks during stance phase of the gait, and the curve of gastrocnemius is shifted more toward terminal stance phase. These trends were also observed during the training and just after training in session 1; however, the changes are more dominant after the 15th training session. The child showed straighter posture after training. His knee and hip are more extended during the mid-stance phase. Ankle plantar flexion is also increased during push-off with the toes against the ground. The increased ROM of the pelvis in the vertical direction shows that the child was able to push up his center of mass against the gravity. The trunk is straighter after the 15th training. In addition, Fig. 3D shows two distinct and higher peaks in GRFs, during heel strike and toe-off. This pattern of the GRF was also observed during the first session data, when the tethers were taken off.

Changes in gait during treadmill walking before and after 15 sessions of training: Group results

Six children with diplegic CP participated in the robotic training with TPAD. All of these children presented crouch gait. Participants of ages 20 years and below were recruited and were classified as Gross Motor Functional Classification System (GMFCS) level II. These children could walk independently but experienced difficulty while walking on level or uneven surface. The mean weight of the group is 65.83 ± 22.69 kg, and the mean height is 161.33 ± 15.04 cm. Each child completed the protocol outlined in Materials and Methods.

Group data of all six children are presented in Fig. 4. In Fig. 4 (A to C), left graphs show free walking over different training sessions, and right graphs show free walking for sessions 1 and 16 with the same walking speed. Nonparametric statistical analysis was conducted only for sessions 1 and 16. Wilcoxon signed-rank test was used with $\alpha = 0.05$ (*37*), because the data did not satisfy the normal distribution due to small number of participants. All lower limb–related parameters are averaged for left and right legs. Figure 4A demonstrates the minimum flexion angle, namely, maximum extension (or maximum plantar flexion), for hip, knee, and ankle joints. These parameters show whether the participants improved extension after being trained with TPAD. All hip, knee, and ankle angles showed decreasing trend toward smaller joint angles. The ankle angle showed a significant decrease before and after 15 sessions of training: for ankle, $P = 0.028$ with the effect size of $r = 0.635$; knee, $P = 0.046$ with $r = 0.575$; and hip, $P = 0.028$ with $r = 0.635$. Because the effect size is larger than 0.5, the intervention has a large effect in the measured variables (*37*). Pelvic translational ROM increased over training in the vertical direction as well ($P = 0.046$ with $r = 0.575$). Step length and step width were also compared before and after training. Step length increased over the training and showed a significant change between sessions 1 and 16 ($P = 0.046$ with $r = 0.575$), whereas step width did not change over training and after 15 sessions of training. Pelvic translational ROM and gait parameters were scaled with the height of the participants. Trunk angle showed a decreasing trend in Fig. 4E, but no significance was found. All these parameters' data during the training with the downward force are included in fig. S3.

Walking ability of children is often evaluated by walking speed, step length, and foot clearance. Significant increases were observed in both overground and treadmill walking speed of children after the training. Overground walking speed was computed

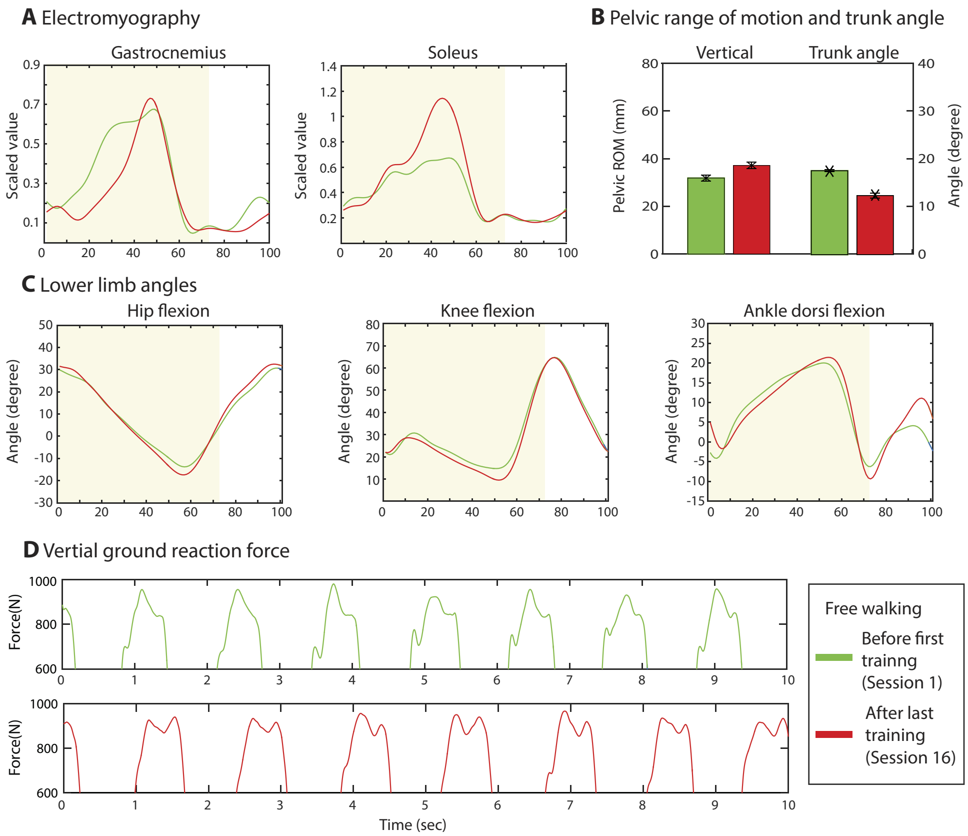
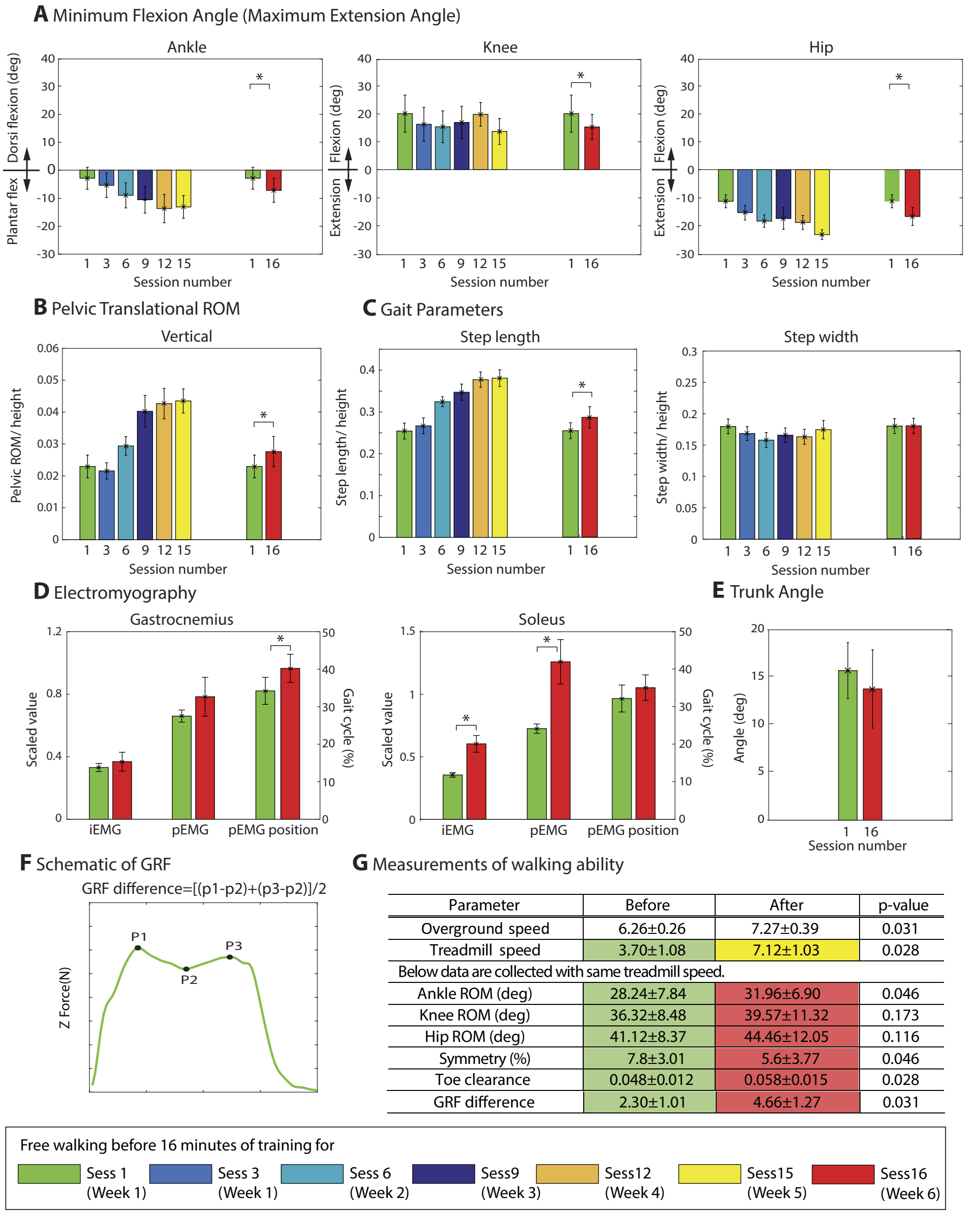


Fig. 3. Results of a representative participant with multiple training sessions. Average and SE are presented for the data before and after 15 training sessions during free walking. (**A**) EMG of muscles. (**B**) Lower limb angles. (**C**) Pelvic translation ROM and the trunk angle. (**D**) Vertical GRFs (please refer to movie S1).



Parameter	Before	After	p-value
Overground speed	6.26±0.26	7.27±0.39	0.031
Treadmill speed	3.70±1.08	7.12±1.03	0.028
Below data are collected with same treadmill speed.			
Ankle ROM (deg)	28.24±7.84	31.96±6.90	0.046
Knee ROM (deg)	36.32±8.48	39.57±11.32	0.173
Hip ROM (deg)	41.12±8.37	44.46±12.05	0.116
Symmetry (%)	7.8±3.01	5.6±3.77	0.046
Toe clearance	0.048±0.012	0.058±0.015	0.028
GRF difference	2.30±1.01	4.66±1.27	0.031

Fig. 4. Group data during free walking between sessions 1 and 16. For (**A**) to (**C**), the left side of the graphs presents data for each week (sessions 1, 3, 6, 9, 12, and 15), and the right side presents data for sessions 1 and 16, which have the same walking speed. From sessions 1 to 15, the treadmill speed was increased every first session of each week, as described in Materials and Methods. (**D**) EMG is scaled with the maximum value of baseline in session 1. (**E**) Trunk angle during walking. (**F**) The definition of the GRF difference is illustrated, and its group value is presented in (**G**). Mean and SEs are presented for six participants (n = 6), except EMG (n = 5).

from the Six Minute Walk test. In addition, treadmill walking after 15 sessions of training showed increased step length, symmetric gait, and better foot clearance, as shown in Fig. 4G. Symmetry of each child was computed by comparing the normalized stance time between right and left legs. According to Eq. 1, the value is close to 0, when the child presents a symmetric gait.

$$\text{Symmetry (\%)} = \left|1 - \frac{\text{right stance time}}{\text{left stance time}}\right| \times 100 \tag{1}$$

EMG of gastrocnemius and soleus is presented in Fig. 4D for the group. We investigated the magnitudes and patterns of the EMG activity in two extensor muscles for ankle, gastrocnemius and soleus, before and after 15 sessions of training. EMG values are filtered, rectified, and then scaled with the maximum value of the baseline in session 1. If the muscles in the 16th session are firing stronger than the first session, the scaled value can be above 1. After all post-processing (see Materials and Methods), three different parameters were computed to extract the characteristics of the EMG pattern:

1) iEMG: mean value of post-processed EMG over the entire gait cycle.

2) pEMG: peak value of post-processed EMG of each gait cycle.

3) pEMG position: position of the peak value in terms of the percentage of the gait cycle.

If the pEMG is larger than iEMG, not only the EMG is increased during the entire phase but also the shape is changed toward a sharper cone, the normal EMG pattern of soleus and gastrocnemius.

Figure 4D shows the magnitudes of integrated and averaged EMG (iEMG) and peak EMG (pEMG), with higher magnitudes after 15 training sessions. Normal EMG patterns of these muscles have sharp peaks around 40 to 50% of the gait cycle. The gastrocnemius peak after training has shifted to the right, closer to 40% of the gait cycle (P = 0.043 with r = 0.640), and the soleus significantly increased the magnitude for both iEMG and pEMG (both P = 0.043 with r = 0.640).

Another important observation is that children develop distinct heel strike and toe-off as a result of this training. Heel-to-toe pattern can be characterized by peaks and valleys of the GRFs (38). Statistical analysis was conducted for sessions 1 and 16 with the same walking speed. We define "GRF difference" to characterize the heel-to-toe pattern of walking using the equation below:

$$\text{GRF difference} = \frac{[\text{first peak (P1)} + \text{second peak (P3)}]}{2} - \text{valley (P2)} \tag{2}$$

The first peak (P1) in Fig. 4F indicates the force during heel strike, and the second peak (P3) is the force during push-off of the stance foot. GRF difference was scaled for each child's weight. GRF difference values for sessions 1 and 16 are presented with the P value in Fig. 4G. The training promoted a heel-to-toe pattern that stabilizes the stance leg during initial contact and propels the body forward.

Clinical evaluations were conducted twice, 1 week before the first training and 2 weeks after the last training. Six Minute Walk test (before: 361 ± 28.56 m; after: 403.67 ± 46.38 m), Pediatric Berg Balance Scale (BBS) (before: 49.67 ± 3.98; after 49.50 ± 3.83), and Timed Up and Go (TUG) test (before: 9.26 ± 2.58; after: 8.97 ± 1.95) were evaluated, but no significant improvement was found after the training for BBS (P = 0.785 with r = 0.079) and TUG test (P = 0.345 with r = 0.272). Six Minute Walk test showed significant improvement (P = 0.028 with r = 0.635). Individual values are presented in table S1.

DISCUSSION

This study shows the feasibility that training with TPAD using downward pull force can improve walking of CP children with crouch gait. The lower limbs have increased extension after the training, showing stronger soleus muscles. Improved vertical pelvic ROM and decreasing trend of trunk angle are also observed. In addition, children who participated in this study showed improvements in walking ability, for example, step length, walking speed, symmetry, and foot clearance. Currently, there is no well-established physical therapy or strengthening exercise for the treatment of crouch gait (39); TPAD training with downward pelvic pull could be a promising intervention for these children.

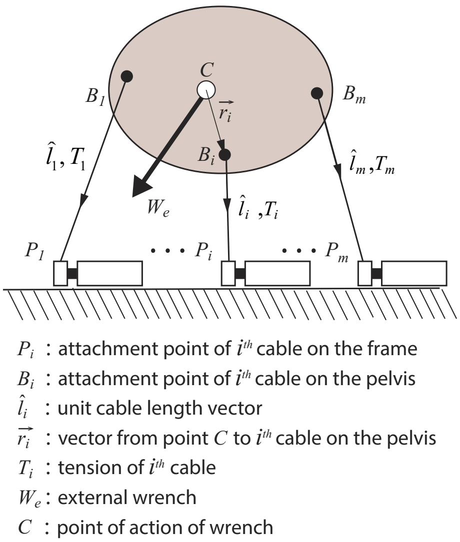


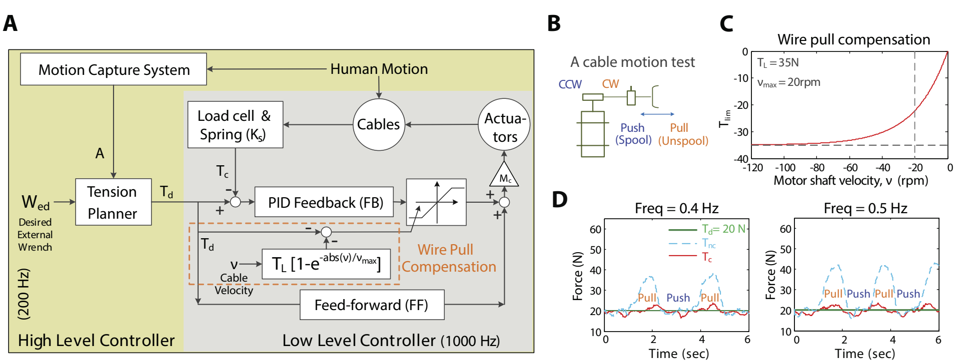
Fig. 5. Cable-driven system diagram for structure matrix. The TPAD is a cable-driven parallel system with m actuated cables connected to the participant's pelvis having n = 6 DOFs. One end of each cable is attached to the pelvis, and the other end is attached to an electrical motor, shown as points B_i and P_i, respectively. These cables together exert an external wrench, W_e, on the pelvis.

After training, children presented increased extension of the knee and plantar flexion of the ankle. This is explained by the increased activity of the soleus muscle, which prevents the progression of the tibia during the stance and allows the knees to extend due to momentum of the center of mass. The improved timing of the gastrocnemius also contributed to knee extension, because the gastrocnemius activated less during the early stance phase to prevent knee flexion of the crouch gait. Improved muscle activation of soleus and gastrocnemius also increased the plantar flexion of the ankle during the push-off phase. However, hip extension cannot be directly explained by those two muscles. Hip extension can increase due to activation of the muscles that were not measured in the present study, such as gluteus maximus, adductor magnus, semimembranosus, and semitendinosus. Another possible explanation is that the changed posture of the knee and ankle enabled the hip to extend more easily. In crouch gait, the hip requires larger torque for extension because the knee is leaning forward, and the body weight vector creates a larger moment arm to flex the hip (9).

In the past, studies were conducted to improve crouch gait of children with CP by strengthening hip and knee extensors using bench press or elastic rubber bands (40–42). These studies targeted knee extension and reported inconsistent outcomes after strength training (7). Instead of focusing only on the knees, TPAD chooses to train all leg extensor muscles, especially the soleus. Strength training of extensor muscles during walking with TPAD might be more effective than conventional strengthening therapy for two reasons. First, loading is an important sensory cue that facilitates activation of the extensor muscles during gait (43–45). With the additional downward force on the pelvis using TPAD, this sensory cue is emphasized to induce activation of the appropriate motor pool for load bearing. Second, the activation is phase-dependent when the limbs support weight (43, 45). This may explain why training with bench press or stretching with rubber bands does not transfer to improvements in gait (17).

Strengthened ankle extensor muscles reinforce the push-off force of the lower limbs. As the leg swings further with increased step length and toe clearance, children attain a better heel strike. Stronger heel strike was also reported with treadmill training on an inclined surface (46, 47). Although this training improved heel strike of children with CP, the underlying mechanism in this study is different from the current study. Walking on an inclined surface strengthens weak tibialis muscles to increase dorsiflexion of the foot and lift the toe for heel strike. Another benefit of the strong push-off is that walking speed

Fig. 6. TPAD controller schematic and testing for WPC. (**A**) Control architecture of low- and high-level controller. (**B**) Schematic of cable motion test. (**C**) WPC function. (**D**) A cable was pulled and pushed about 1 ft by hand at frequencies of 0.4 and 0.5 Hz for tension T_d = 20 N. T_c and T_{nc} are the cable tension values with and without the WPC term. The performance of controller is improved during pulling phase, when WPC is added.



becomes faster with increased step length. Similar results have been reported in conventional treadmill studies with or without the partial weight support (*19*, *20*, *48*, *49*). In these studies, a physical therapist assisted the child with manual corrections and verbal instructions, which led to improvements in walking (*48*). Instead of a physical therapist assisting in walking straight, TPAD can provide a robotic environment for the children to strengthen extensor muscles compared with ordinary walking. This device can potentially reduce the effort and labor of the physical therapists so that they can focus on higher-level management, dosage, and monitoring of the patients.

The BBS and TUG tests were introduced as secondary measures of this study to determine how the training results translate to other tasks. For TUG test, we found that only half of children improved. This might be because children who did not show improvement already had a higher performance on this measure before training. Their TUG values were between the values of typically developing children and CP children with GMFCS level I (*50*). Although these children had high performance in TUG, their posture was different from the typically developing children. Unfortunately, TUG score only measures the total speed and does not capture the quality of each task. TUG is measured from a dynamic task consisting of different movements; however, BBS mostly consists of static tasks (*51*, *52*). This might be the reason why BBS training did not improve after TPAD training. A study over 5 weeks may be insufficient to transfer the training effects to other tasks that are not directly related to the trained task. Among these three clinical scales, we should note that the Six Minute Walk test showed improvements. Six Minute Walk test is a self-paced, submaximal test that assesses functional capacity for walking over a prolonged distance (*53*, *54*). It reflects exercise tolerance and endurance required for the performance of active daily living (*54*). Here, we asked children to walk within a submaximal environment, by augmenting weight, for longer than 6 min. We think that the training effect of the present study is better estimated by the Six Minute Walk test that is more analogous to the task in the TPAD training, rather than BBS or TUG tests.

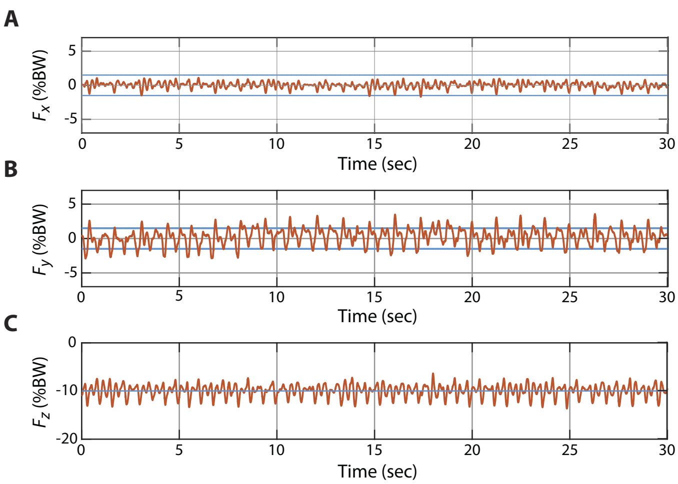
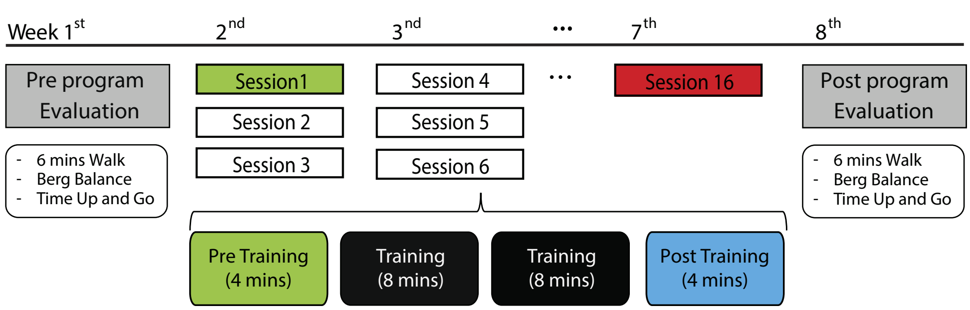


Fig. 7. TPAD controller performance for three force components. (**A**) F_x in medial-lateral direction satisfies to be within ±1.5% BW. (**B**) F_y in anterior-posterior direction is within ±1.5% BW. (**C**) F_z in vertical direction is controlled to be at −10% BW.

Future studies warrant some attention. When recruiting children for TPAD training, independent walkers with GMFCS level I or II are recommended. TPAD training provides a downward force on the pelvis, and any hand holding of the rails will diminish the force delivered to the joints of lower limbs and reduce the training effects. For children who are more severely affected, other robotic interventions with assistance or guidance force might be more desirable (*55*). Second, the present study was designed to compare parameters before and after the training with the same treadmill speed (Fig. 4). The training improvements, as stated here, are conservative because the treadmill speed was kept the same before and after training for scientific comparison. Once trained, the optimal walking speed of the children was higher, and their gait should be measured and documented at that speed. Future design of the experiment should include overground walking data for evaluation of the training. Third, this study does not include a randomized control group. The impact of natural ongoing development of the children should be considered within the result. However, there are studies that show that there are no significant changes due to the natural development of CP children within 4- to 6-week period, unless they undergo an intervention (*18*, *56*, *57*). These studies support this idea by measuring GMFM D (standing function) and GMFM E (walking function) in children with CP. As future study, a larger number of participants with a control group should be organized. Last, this pilot study applied 10% of downward force as a start of this new training paradigm. Different dosage in terms of the amount of the weight and period should be tried and compared to maximize the benefits of this training. Large population of participants with CP will be recruited in the future with different GMFCS levels to increase the power of this study. We are also considering studies targeted at children with hemiplegic/quadriplegic CP.

Feedback from parents and children involved in this study was consistent: They reported improved posture, stronger legs, and faster walking speeds. Our results show that intensive and well-designed paradigms of gait training with robotic devices, such as TPAD, can yield improvements in gait, as has been postulated in previous studies with CP children (*49*, *58*). Our study confirms these observations, as EMG magnitudes and the pattern of muscles improved after 15 sessions of gait training. Future works should investigate more stable and permanent functional recovery through longer interventions.

Fig. 8. TPAD training protocol. Training consists of 15 training sessions over 5 weeks. Pre-evaluation was conducted 1 week before the first session, and post-evaluation was conducted 2 weeks after the last training session. The 16th session was added after the 15th training session to record free walking on the treadmill with the same speed as session 1.



CONCLUSION

The present study showed the feasibility to use this training method to strengthen lower limbs and correct the posture of children with crouch gait. Downward force increased the activation of the soleus muscle and delayed the early activation of the gastrocnemius muscle. This change of muscle pattern enabled straighter lower limb during the middle of stance phase and improved walking. Because there are no well-established physical therapies or strengthening methods for children with crouch gait, our training method may be a promising intervention for these children. This new training could be used solely as a strength training device in the clinic or combined with conventional treatment such as Botox or ethanol injection to reinforce both methodologies.

MATERIALS AND METHODS

Experiment setup

The design details of the TPAD are presented in Fig. 2A. Briefly, four single-phase ac servo motors were mounted on an inertial fixed rigid frame. Low-stretch nylon-coated stainless steel wires, with a breaking strength of 540 N, were routed using pulleys within the workspace. All motors were powered in torque mode, and a continuous torque of 4 N·m was achieved in each motor through a 10:1 gear reduction. Motors, drivers, and gear boxes were from Kollmorgen (Radford, VA). A cable reel with a diameter of 5.08 cm was mounted on the gearbox shaft so that a maximum continuous tension of 157 N was achieved in each cable.

The end of the cable coming from a cable reel was connected to a fabric hip belt (climbing harness from Black Diamond Equipment, UT) worn by a human participant. To measure the instantaneous cable tension, a load cell was installed in series with each cable (MLP-200 from Transducer Techniques, CA). A spring of stiffness 2.5 N/mm was also installed in series with each cable to reduce the output impedance. A 10-camera motion capture system (Bonita-10 series from Vicon, UK) was used as a part of the controller to track the human motion and cable attachment locations. GRF was recorded by the force plate on the split treadmill (Bertec, OH). EMG signals were recorded and received by a wireless desktop unit (DTS Desktop Receiver, Noraxon Inc., AZ). The controller was implemented on a LabVIEW PXI real-time system (National Instruments, TX).

System model

TPAD is a cable-driven parallel system with actuated cables connected to the end effector, which is the human pelvis (Fig. 5). Each cable is modeled as a pure force at the attachment point. These cables together exert a wrench on the pelvis. Suppose $T \in \mathbb{R}^m$ represents the tensions in the m cables, and $W_e \in \mathbb{R}^n$ is the n-DOF external wrench on the pelvis, these are related to each other as

$$AT = W_e \tag{3}$$

where $A \in \mathbb{R}^{n\times m}$ is the structure matrix that depends on the system geometry and can be computed by the coordinates of the cable attachment points. For the case of 6-DOF wrench W_e ($n = 6$) consisting of three-dimensional forces and moments, the matrix A is given by the following expression (Fig. 5):

$$A = \begin{bmatrix} \dots & \hat{l}_i & \dots \\ \dots & \vec{r}_i \times \hat{l}_i & \dots \end{bmatrix}_{6\times m} \tag{4}$$

where $\hat{l}_i$ is the i^{th} unit cable length vector from the end effector B_i toward the fixed routing point P_i, and $\vec{r}_i$ is the vector from the point of application C to the i^{th} cable attachment point B_i on the rigid body.

Tension planner

A tension planner computes desired tension of each cable to achieve the targeted wrench (for the present experiment, a constant downward force) in real time. Cables in a cable-driven robot can only apply a pulling force on the end effector, and therefore, positive cable tensions must be maintained to retain control. For a general case of n-DOF system, at least $n + 1$ cables are required for generating a desired wrench, $W_e \in \mathbb{R}^n$ (*59*, *60*). Because the number of cables m is greater than the number of DOFs of the wrench n, Eq. 3 is underdetermined. Because this leads to infinitely many tension solutions, optimality criteria are needed to find a feasible solution set. Here, a quadratic programming–based optimization scheme with equality and inequality constraints is implemented as below:

$$\begin{gathered} \min \quad [\tfrac{1}{2}(T - T_p)^T(T - T_p)], \\ \text{subject to} \quad AT = W_e, \quad \text{and} \quad T_{\min} \le T \le T_{\max} \end{gathered} \tag{5}$$

where $W_e = [F_x, F_y, F_z, M_x, M_y, M_z]^T$ consists of force and moment in three dimensions; x is medial-lateral direction, y is anterior-posterior direction, and z is the vertical direction. T_p is the optimized tension value of previous time step for continuous tension profile. $T_{\min}$ and $T_{\max}$ are the lower and upper bounds on the cable tension values, respectively.

Ideally, to apply pure downward force with $W_e = [0\ 0\ -10\ \%\ BW\ 0\ 0\ 0] \in \mathbb{R}^6$, a minimum number of seven cables ($m = 7$) is needed to achieve the target wrench. However, only four cables are used in this experiment by using relaxation of Eq. 3. Instead of setting the desired force/moment except the downward force component to be zero, they are limited to small values in the human experiment (*61*). The vertical force component of Eq. 3 is set to an equality constraint, and the other components are set to inequality constraints. The constraints used in the human experiment are as follows to achieve a constant downward force that is equivalent to 10% body weight (BW):

$$\begin{aligned} |F_{x,y}| &\le 1.5\%\ \text{BW} \\ F_z &= -10\%\ \text{BW} \\ |M_{x,y,z}| &\le 5\ \text{N}\cdot\text{m} \end{aligned} \tag{6}$$

Four cables are installed with symmetric configuration as shown in Fig. 2A. This configuration was decided to minimize unwanted force/moment components and to avoid having cables in close proximity of participants while allowing arm movement. Before the human experiment, the feasible workspace was simulated by using free walking pelvic trajectories of healthy individuals. The simulation was conducted for a wide range of participant height (1.22 to 1.95 m) and weight (30 to 100 kg) such that the cable tensions satisfy Eq. 6. The performance of the controller during the actual experiment is presented in Fig. 7. The root mean square of the 10% BW of downward force (Fig. 7C) is 1.26% BW for 1-min data.

Controller

The controller consists of two different strategies. The high-level controller computes the desired tension using tension planner, and the low-level controller achieves the desired tension with feedback and feedforward terms. In the high-level controller, the tension planner runs at 200 Hz to solve the quadratic programming with constraints in Eq. 6. A real-time motion capture system is used in the high-level controller to track the markers placed at cable attachment locations and human anatomical positions. The low-level controller implements the desired cable tension at 1000 Hz using a force mode control scheme. An open-loop reference feedforward (T_{FF}) term was computed by the desired tension value, and a feedback (T_{FB}) term was calculated from a proportional-integral-derivative controller. The motor constant (Mc) is empirically defined for each motor as the following values: motor 1 (18.69 N/V), motor 2 (18.46 N/V), motor 3 (18.83 N/V), and motor 4 (18.89 N/V). Because the feedback term can have negative values, the minimum output of the feedback term was chosen to satisfy Eq. 7 to prevent the cable to be slack.

$$T_{FB} + T_{FF} > 0 \tag{7}$$

However, the dynamics of the human and the actuators add errors in the cable tensions, as demonstrated in Fig. 6D. To account for this phenomenon, wire pull compensation (WPC) was implemented which uses a relaxation of Eq. 7. It changes the lower limit $(T_{FB})_{min}$ for each motor as a function of the cable velocity, $v(t)$, as shown in Eq. 8.

$$\begin{aligned}(T_{FB})_{min} &= -T_{FF} + T_{lim} \\ &= -T_{FF} - T_L\left(1 - e^{\frac{-\|v(t)\|}{v_{max}}}\right)\end{aligned} \tag{8}$$

where T_L and v_{max} are two positive constant parameters. These parameters were tuned during a human walking test to achieve a responsive controller. T_{lim} is zero when the participant is stationary, that is, $v = 0$, and approaches $-T_L$ exponentially as the negative cable velocity magnitude increases. T_L and v_{max} are constant values set by repeated human testing, respectively, $T_L = 35$ N and $v_{max} = 20$ rpm. The controller performance improved significantly during the cable-pulling phase, which is almost similar to the cable-pushing phase as shown in Fig. 6D.

Experiment protocol

Each participant participated in 3 training sessions per week for a total of 15 training sessions. For each training session, participants walked on a force plate–instrumented treadmill for roughly 30 min, with body reflective markers recorded by a motion capture system. During the first training session of each week, the treadmill speed was decided before the training. The treadmill speed was selected by increasing the speed slowly in increments of 0.1 m/s until the child had difficulty walking. Then, the speed was slowed down until the child felt comfortable to walk at that speed. The study protocol consists of free walking before training, training with TPAD, and free walking after training (Fig. 8). Free walking was recorded without tethers attached to the pelvic belt. After 5 weeks of training, participants visited again during the sixth week to characterize their walking on the treadmill with the same speed as the first week. One week before the first training session and 2 weeks after the last training session, the Six Minute Walk test was conducted to measure overground walking speed. Besides walking test, BBS and TUG tests were also performed. EMG was recorded only for the 1st and 16th sessions while walking on the treadmill at the same speed. The study was approved by Columbia University's institutional review board.

Participants

Six CP children with crouch gait participated in the robotic training with TPAD. These children were diagnosed as non-toe walkers, because the present study is focused on the crouch gait. Participants ages 20 years and below were recruited and classified as GMFCS level II. Exclusions criteria were (i) Botox injections within the past 3 months, (ii) ethanol injections within the past 6 months, and (iii) dorsal rhizotomy surgery within 1 year before the study. Detailed information of each participant is shown in table S1.

Data analysis

All data were divided into gait cycles starting with the heel strike on the treadmill. Gait events were detected by the GRFs recorded from the force plate on the treadmill (*62*). The markers were attached on the lower limb, pelvis, and C7, as per (*61*). Detailed explanation of marker placement is presented in the Supplementary Materials. For pelvic ROM, pelvic center was computed from the geometrical mean of anatomical landmarks (left/right anterior superior iliac and sacrum). Pelvic ROM was calculated by subtracting maximum and minimum displacements of the pelvis within a gait cycle. Step width was defined as the maximum medial-lateral distance between the right and left heel markers during double support period after the heel strike. Step length for one leg was defined as the anterior-posterior distance between the heel markers of two legs at the moment of the leg's heel strike. GRF was filtered with fourth-order low-pass filter with a cutoff frequency of 20 Hz and divided to gait cycle as well.

EMG signals were filtered with fourth-order band-pass Butterworth (40 to 450 Hz), rectified, and then filtered again with fourth-order low-pass Butterworth (6 Hz). For each participant, EMG data of each muscle were normalized to the maximum values recorded during the baseline of the first session. Then, EMG data were divided for each gait cycle and time-normalized from 0 to 100%.

SUPPLEMENTARY MATERIALS

robotics.sciencemag.org/cgi/content/full/2/8/eaan2634/DC1
Materials and Methods
Fig. S1. Vertical GRF of sessions 1, 15, and 16 for all six children during free walking.
Fig. S2. Vertical GRF of all six children during free walking and training with downward force.
Fig. S3. Pelvic ROM and gait parameters during the training.
Fig. S4. EMG for five different muscles before and after 15 training sessions.
Fig. S5. Walking speed of individual children with CP.
Fig. S6. System setup.
Fig. S7. Defining motor constant.
Fig. S8. Vertical force evaluation during training sessions.
Table S1. Participant information and clinical measurements.
Table S2. Response of the participant and parents from clinical chart.
Table S3. Spasticity of ankle and knee joints.
Table S4. Cable attachment locations on the frame.
Movie S1. Experiment video including overground walking before/after training.

REFERENCES AND NOTES

1. S. Winter, A. Autry, C. Boyle, M. Yeargin-Allsopp, Trends in the prevalence of cerebral palsy in a population-based study. *Pediatrics* **110**, 1220–1225 (2002).
2. D. H. Sutherland, J. R. Davids, Common gait abnormalities of the knee in cerebral palsy. *Clin. Orthop. Relat. Res.* **288**, 139–147 (1993).
3. J. M. Rodda, H. K. Graham, L. Carson, M. P. Galea, R. Wolfe, Sagittal gait patterns in spastic diplegia. *Bone Joint J.* **86**, 251–258 (2004).
4. J. R. Gage, *Gait Analysis in Cerebral Palsy* (Mac Keith Press, 1991).
5. H. Binder, G. D. Eng, Rehabilitation management of children with spastic diplegic cerebral palsy. *Arch. Phys. Med. Rehabil.* **70**, 482–489 (1989).
6. R. Norlin, P. Odenrick, Development of gait in spastic children with cerebral palsy. *J. Pediatr. Orthop.* **6**, 674–680 (1986).
7. K. M. Steele, D. L. Damiano, M. N. Eek, M. Unger, S. L. Delp, Characteristics associated with improved knee extension after strength training for individuals with cerebral palsy and crouch gait. *J. Pediatr. Rehabil. Med.* **5**, 99–106 (2012).
8. K. M. Steele, A. Seth, J. L. Hicks, M. S. Schwartz, S. L. Delp, Muscle contributions to support and progression during single-limb stance in crouch gait. *J. Biomech.* **43**, 2099–2105 (2010).
9. J. Perry, J. Burnfield, *Gait Analysis: Normal and Pathological Function* (Slack Incorporated, 1992).

10. V. Dietz, W. Berger, Cerebral palsy and muscle transformation. *Dev. Med. Child Neurol.* **37**, 180–184 (1995).
11. D. L. Damiano, C. L. Vaughan, M. E. Abel, Muscle response to heavy resistance exercise in children with spastic cerebral palsy. *Dev. Med. Child Neurol.* **37**, 731–739 (1995).
12. H. Anttila, I. Autti-Rämö, J. Suoranta, M. Mäkelä, A. Malmivaara, Effectiveness of physical therapy interventions for children with cerebral palsy: A systematic review. *BMC Pediatr.* **8**, 14 (2008).
13. S. W. Blundell, R. B. Shepherd, C. M. Dean, R. Adams, B. Cahill, Functional strength training in cerebral palsy: A pilot study of a group circuit training class for children aged 4–8 years. *Clin. Rehabil.* **17**, 48–57 (2003).
14. V. A. Scholtes, A. J. Dallmeijer, E. A. Rameckers, O. Verschuren, E. Tempelaars, M. Hensen, J. G. Becher, Lower limb strength training in children with cerebral palsy—A randomized controlled trial protocol for functional strength training based on progressive resistance exercise principles. *BMC Pediatr.* **8**, 41 (2008).
15. J. F. Morton, M. Brownlee, A. K. McFadyen, The effects of progressive resistance training for children with cerebral palsy. *Clin. Rehabil.* **19**, 283–289 (2005).
16. A. Scianni, J. M. Butler, L. Ada, L. F. Teixeira-Salmela, Muscle strengthening is not effective in children and adolescents with cerebral palsy: A systematic review. *Aust. J. Physiother.* **55**, 81–87 (2009).
17. V. A. Scholtes, J. G. Becher, Y. J. Janssen-Potten, H. Dekkers, L. Smallenbroek, A. J. Dallmeijer, Effectiveness of functional progressive resistance exercise training on walking ability in children with cerebral palsy: A randomized controlled trial. *Res. Dev. Disabil.* **33**, 181–188 (2012).
18. M. R. Schindl, C. Forstner, H. Kern, S. Hesse, Treadmill training with partial body weight support in nonambulatory patients with cerebral palsy. *Arch. Phys. Med. Rehabil.* **81**, 301–306 (2000).
19. D. M. Begnoche, K. H. Pitetti, Effects of traditional treatment and partial body weight treadmill training on the motor skills of children with spastic cerebral palsy: A pilot study. *Pediatr. Phys. Ther.* **19**, 11–19 (2007).
20. K. J. Dodd, S. Foley, Partial body-weight-supported treadmill training can improve walking in children with cerebral palsy: A clinical controlled trial. *Dev. Med. Child Neurol.* **49**, 101–105 (2007).
21. C. P. McGowan, R. Kram, R. R. Neptune, Modulation of leg muscle function in response to altered demand for body support and forward propulsion during walking. *J. Biomech.* **42**, 850–856 (2009).
22. D. H. K. Chow, M. L. Y. Kwok, A. C. Au-Yang, A. D. Holmes, J. C. Cheng, F. Y. Yao, M. Wong, The effect of backpack load on the gait of normal adolescent girls. *Ergonomics* **48**, 642–656 (2005).
23. J. X. Li, Y. Hong, P. D. Robinson, The effect of load carriage on movement kinematics and respiratory parameters in children during walking. *Eur. J. Appl. Physiol.* **90**, 35–43 (2003).
24. V. Vashista, X. Jin, S. K. Agrawal, Active tethered pelvic assist device (a-tpad) to study force adaptation in human walking, in *Proceedings of the IEEE International Conference on Robotics and Automation* (IEEE, 2014), pp. 718–723.
25. A. Meyer-Heim, C. Ammann-Reiffer, A. Schmartz, J. Schäfer, F. H. Sennhauser, F. Heinen, B. Knecht, E. Dabrowski, I. Borggraefe, Improvement of walking abilities after robotic-assisted locomotion training in children with cerebral palsy. *Arch. Dis. Child.* **94**, 615–620 (2009).
26. I. Borggraefe, J. S. Schaefer, M. Klaiber, E. Dabrowski, C. Ammann-Reiffer, B. Knecht, S. Berweck, F. Heinen, A. Meyer-Heim, Robotic-assisted treadmill therapy improves walking and standing performance in children and adolescents with cerebral palsy. *Eur. J. Paediatr. Neurol.* **14**, 496–502 (2010).
27. S. Lefmann, R. Russo, S. Hillier, The effectiveness of robotic-assisted gait training for paediatric gait disorders: Systematic review. *J. Neuroeng. Rehabil.* **14**, 1 (2017).
28. C. Bayón, O. Ramírez, J. I. Serrano, M. D. Del Castillo, A. Pérez-Somarriba, J. M. Belda-Lois, I. Martínez-Caballero, S. Lerma-Lara, C. Cifuentes, A. Frizera, E. Rocon, Development and evaluation of a novel robotic platform for gait rehabilitation in patients with cerebral palsy: CPWalker. *Robot. Auton. Syst.* **91**, 101–114 (2017).
29. C. Bayón, S. Lerma, O. Ramírez, J. I. Serrano, M. D. Del Castillo, R. Raya, J. M. Belda-Lois, I. Martnez, E. Rocon, Locomotor training through a novel robotic platform for gait rehabilitation in pediatric population: Short report. *J. Neuroeng. Rehabil.* **13**, 98 (2016).
30. Z. F. Lerner, D. L. Damiano, H.-S. Park, A. J. Gravunder, T. Bulea, A robotic exoskeleton for treatment of crouch gait in children with cerebral palsy: Design and initial application. *Trans. Neural Syst. Rehabil. Eng.* **25**, 650–659 (2016).
31. J. H. Meuleman, E. H. F. Van Asseldonk, H. Van der Kooij, The effect of directional inertias added to pelvis and ankle on gait. *J. Neuroeng. Rehabil.* **10**, 40 (2013).
32. J. M. Hidler, A. E. Wall, Alterations in muscle activation patterns during robotic-assisted walking. *Clin. Biomech.* **20**, 184–193 (2005).
33. D. L. Damiano, A. S. Arnold, K. M. Steele, S. L. Delp, Can strength training predictably improve gait kinematics? A pilot study on the effects of hip and knee extensor strengthening on lower-extremity alignment in cerebral palsy. *Phys. Ther.* **90**, 269–279 (2010).
34. R. A. Magill, D. Anderson, *Motor Learning and Control: Concepts and Applications* (McGraw-Hill, 2007).
35. D. A. Winter, *Biomechanics and Motor Control of Human Movement* (John Wiley & Sons, 2009).
36. A. Hof, H. Elzinga, W. Grimmius, J. P. K. Halbertsma, Speed dependence of averaged EMG profiles in walking. *Gait Posture* **16**, 78–86 (2002).
37. J. H. Zar, *Biostatistical Analysis* (Prentice Hall, 2009).
38. S. Ajemian, D. Thon, P. Clare, L. Kaul, R. F. Zernicke, B. Loitz-Ramage, Cane-assisted gait biomechanics and electromyography after total hip arthroplasty. *Arch. Phys. Med. Rehabil.* **85**, 1966–1971 (2004).
39. P. Kedem, D. M. Scher, Evaluation and management of crouch gait. *Curr. Opin. Pediatr.* **28**, 55–59 (2016).
40. D. L. Damiano, L. E. Kelly, C. L. Vaughn, Effects of quadriceps femoris muscle strengthening on crouch gait in children with spastic diplegia. *Phys. Ther.* **75**, 658–667 (1995).
41. M. Unger, M. Faure, A. Frieg, Strength training in adolescent learners with cerebral palsy: A randomized controlled trial. *Clin. Rehabil.* **20**, 469–477 (2006).
42. M. N. Eek, R. Tranberg, R. Zügner, K. Alkema, E. Beckung, Muscle strength training to improve gait function in children with cerebral palsy. *Dev. Med. Child Neurol.* **50**, 759–764 (2008).
43. S. J. Harkema, S. L. Hurley, U. K. Patel, P. S. Requejo, B. H. Dobkin, V. R. Edgerton, Human lumbosacral spinal cord interprets loading during stepping. *J. Neurophysiol.* **77**, 797–811 (1997).
44. R. af Klint, N. Mazzaro, J. B. Nielsen, T. Sinkjaer, M. J. Grey, Load rather than length sensitive feedback contributes to soleus muscle activity during human treadmill walking. *J. Neurophysiol.* **103**, 2747–2756 (2010).
45. C. M. Bastiaanse, J. Duysens, V. Dietz, Modulation of cutaneous reflexes by load receptor input during human walking. *Exp. Brain Res.* **135**, 189–198 (2000).
46. J. Lorentzen, H. Kirk, H. Fernandez-Lago, R. Frisk, N. Scharff Nielsen, M. Jorsal, J. B. Nielsen, Treadmill training with an incline reduces ankle joint stiffness and improves active range of movement during gait in adults with cerebral palsy. *Disabil. Rehabil.* **39**, 987–993 (2017).
47. M. Willerslev-Olsen, J. Lorentzen, J. B. Nielsen, Gait training reduces ankle joint stiffness and facilitates heel strike in children with cerebral palsy. *Neurorehabilitation* **35**, 643–655 (2014).
48. N. Chrysagis, E. K. Skordilis, N. Stavrou, E. Grammatopoulou, D. Koutsouki, The effect of treadmill training on gross motor function and walking speed in ambulatory adolescents with cerebral palsy: A randomized controlled trial. *Am. J. Phys. Med. Rehabil.* **91**, 747–760 (2012).
49. M. Hodapp, J. Vry, V. Mall, M. Faist, Changes in soleus h-reflex modulation after treadmill training in children with cerebral palsy. *Brain* **132**, 37–44 (2009).
50. R. D. Nicolini-Panisson, M. V. F. Donadio, Timed "up & go" test in children and adolescents. *Rev. Paul. Pediatr.* **31**, 377–383 (2013).
51. E. N. Williams, S. G. Carroll, D. S. Reddihough, B. A. Phillips, M. P. Galea, Investigation of the timed 'up & go' test in children. *Dev. Med. Child Neurol.* **47**, 518–524 (2005).
52. G. Kembhavi, J. Darrah, J. Magill-Evans, J. Loomis, Using the berg balance scale to distinguish balance abilities in children with cerebral palsy. *Pediatr. Phys. Ther.* **14**, 92–99 (2002).
53. M. A. van Loo, A. M. Moseley, J. M. Bosman, R. A. de Bie, B. L. Hassett, Test–re-test reliability of walking speed, step length and step width measurement after traumatic brain injury: A pilot study. *Brain Inj.* **18**, 1041–1048 (2004).
54. P. Thompson, T. Beath, J. Bell, G. Jacobson, T. Phair, N. M. Salbach, F. Wright, Test–retest reliability of the 10-metre fast walk test and 6-minute walk test in ambulatory school-aged children with cerebral palsy. *Dev. Med. Child Neurol.* **50**, 370–376 (2008).
55. H. J. van Hedel, A. Meyer-Heim, C. Rüsch-Bohtz, Robot-assisted gait training might be beneficial for more severely affected children with cerebral palsy. *Dev. Neurorehabil.* **19**, 410–415 (2016).
56. R.-J. Cherng, C.-F. Liu, T.-W. Lau, R.-B. Hong, Effect of treadmill training with body weight support on gait and gross motor function in children with spastic cerebral palsy. *Am. J. Phys. Med. Rehabil.* **86**, 548–555 (2007).
57. V. Knox, A. L. Evans, Evaluation of the functional effects of a course of bobath therapy in children with cerebral palsy: A preliminary study. *Dev. Med. Child Neurol.* **44**, 447–460 (2002).
58. M. Willerslev-Olsen, T. H. Petersen, S. F. Farmer, J. B. Nielsen, Gait training facilitates central drive to ankle dorsiflexors in children with cerebral palsy. *Brain* **138**, 589–603 (2015).
59. A. Ming, T. Higuchi, Study on multiple degree-of-freedom positioning mechanism using wires. I: Concept, design and control. *Int. J. Jpn. Soc. Prec. Eng.* **28**, 131–138 (1994).
60. S. K. Mustafa, S. K. Agrawal, On the force-closure analysis of n-DOF cable-driven open chains based on reciprocal screw theory. *IEEE Trans. Robot.* **28**, 22–31 (2012).

61. V. Vashista, D. Martelli, S. K. Agrawal, Locomotor adaptation to an asymmetric force on the human pelvis directed along the right leg. *IEEE Trans. Neural Syst. Rehabil. Eng.* **24**, 872–881 (2016).
62. J. A. Zeni Jr., J. G. Richards, J. S. Higginson, Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait Posture* **27**, 710–714 (2008).

Acknowledgments: We thank F. Aprigliano, D. Boiarsky, and L. Bishop for discussions during this work. **Funding:** The personnel on this work were funded with partial support from the following grants: NSF IIS-1339666, NSF CPS-1329363, New York State (NYS) Contract DOH01-C30828GG-3450000, and NYS Translational Grant C31290GG. **Author contributions:** S.K.A., J.K., V.V., D.M., and H.K. designed the experiment. V.V. and J.K. worked on the hardware and software of the system. J.K., D.M., and I.M.-H. performed the human experiment with CP children. I.M.-H. and J.K. conducted the pre- and post-evaluation of the study. H.K. and I.M.-H. recruited and organized participants. H.K. screened participants in terms of inclusion/exclusion criteria. J.K. and S.K.A. analyzed and interpreted the data. J.K. and S.K.A. prepared the manuscript. All authors provide comments and feedbacks on the manuscript. **Competing interests:** The authors declare that they have no competing financial interests. A patent application on the design of cable-driven systems has been filed by Columbia University with U.S. Patent Office, application no. 20170027803. S.K.A., V.V., and J.K. are inventors of this application. **Data and materials availability:** Please contact S.K.A. for data and other materials.

Submitted 25 March 2017
Accepted 30 June 2017
Published 26 July 2017
10.1126/scirobotics.aan2634

Citation: J. Kang, D. Martelli, V. Vashista, I. Martinez-Hernandez, H. Kim, S. K. Agrawal, Robot-driven downward pelvic pull to improve crouch gait in children with cerebral palsy. *Sci. Robot.* **2**, eaan2634 (2017).

Science Robotics

Robot-driven downward pelvic pull to improve crouch gait in children with cerebral palsy

J. Kang, D. Martelli, V. Vashista, I. Martinez-Hernandez, H. Kim and S. K. Agrawal

Sci. Robotics **2**, eaan2634.
DOI: 10.1126/scirobotics.aan2634