

# A Quasi-Passive Compliant Stance Control Knee-Ankle-Foot Orthosis

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**Abstract**— In this paper, we present the design of a novel quasi-passive stance-control orthosis that implements a natural amount of knee compliance during the weight acceptance phase and potentially the entire stance phase of the gait, and allows for free motion during the rest of the gait. We explain that the unaffected knee behaves close to a linear torsional spring in stance and hypothesize that an assistive device that places a linear spring of appropriate stiffness in parallel with the knee can help restore the natural behavior of the joint in stance. We present the design of a friction-based latching mechanism and a control algorithm that engages the spring in parallel with the knee in stance and disengages it during the swing phase of gait, and explain how this module is implemented into a brace in order to create a novel class of compliant stance control orthosis. The device is quasi-passive in that a small actuator serves to lock and unlock the spring module, but the device otherwise requires no actuation and very little power, computation, and control to operate.

## I. INTRODUCTION

ADVANCED bionic leg orthoses can benefit thousands of patients who suffer from injury, stroke, post-polio, multiple sclerosis, SCI, Patellofemoral Pain Syndrome, and others [3-6]. These musculoskeletal disorders often result in lower limb dysfunctions such as impaired quadriceps and other muscles that stabilize the knee. This can result in collapse or pathological hyperextension of the knee from the onset of the stance phase until stance termination. A traditional knee or Knee-Ankle-Foot Orthosis (as described in [7]) involves a thermoformed plastic cast formed around the impaired leg, rigidly supporting (and locking) the knee throughout the gait cycle. These inexpensive devices allow the body mass to be supported, but require compensatory, unnatural, and metabolically expensive movements at the hip, pelvis, and ankle [8-10]. To overcome this issue, a number of stance control orthoses (SCOs) have recently been commercialized and clinically evaluated, and more complex exoskeleton-like devices have been developed by researchers [6, 8-20]. SCOs have been an orthotic option for patients with paresis and paralysis in the lower limb muscles. Current SCOs rigidly stabilize the knee in the stance phase and allow free movement in swing. Compared with the regular knee-ankle-foot orthoses, SCOs have been shown to provide medical benefits to novice users, including increased walking speed and cadence [8, 9]. To both experienced and novice users, SCOs provide increased knee range of motion, stride, step lengths, and user satisfaction [9,

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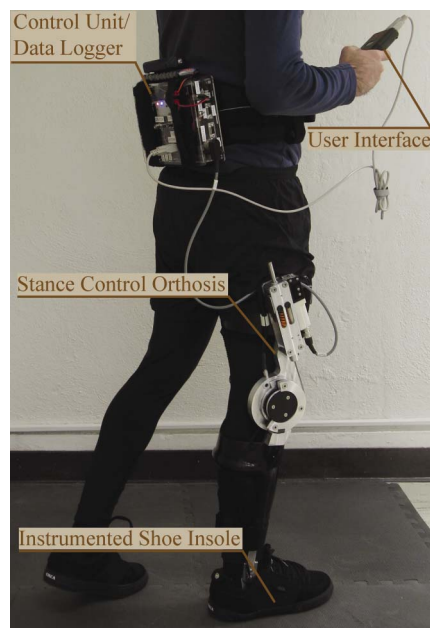


Fig. 1. Ouasi-passive complaint stance control Knee-Ankle-Foot Orthosis.

16, 19]. Moreover, users have demonstrated reduced energy expenditure and gait asymmetry [10, 16]. Previous research has shown kinematic benefits to both affected and unaffected legs [10]. However, since current SCOs rigidly lock the knee in stance, users cannot demonstrate a natural knee flexion. This hinders the shock absorption function of the knee in stance (as outlined in [21, 22]) and could potentially cause increased metabolic cost, user pain and discomfort.

A SCO can mimic the natural damping function of the knee during stance, provided it implements compliant support (instead of rigidly locking) in the weight acceptance phase of the gait [23, 24]. The knee moment-angle relationship in the stance phase reveals a spring-type behavior in the weight acceptance period [23] at the preferred gait speed. This implies that a SCO can employ an accurately sized linear torsional spring in the stance phase to mimic the behavior of an unaffected knee joint at the preferred gait speed.

In this paper, we present the design of a quasi-passive compliant stance control orthosis (CSCO) that supports the knee in the stance phase of the gait by implementing a spring in parallel with this joint (as an improvement to the current devices that rigidly lock the knee in stance), and allows for free motion in the rest of the gait, as shown in Fig. 1. We show how the CSCO exhibits multiple levels of stiffness at different phases of the gait through engagement/disengagement of a linear spring in parallel with the knee. We further discuss how the control algorithm decides whether to engage the spring depending on the gait

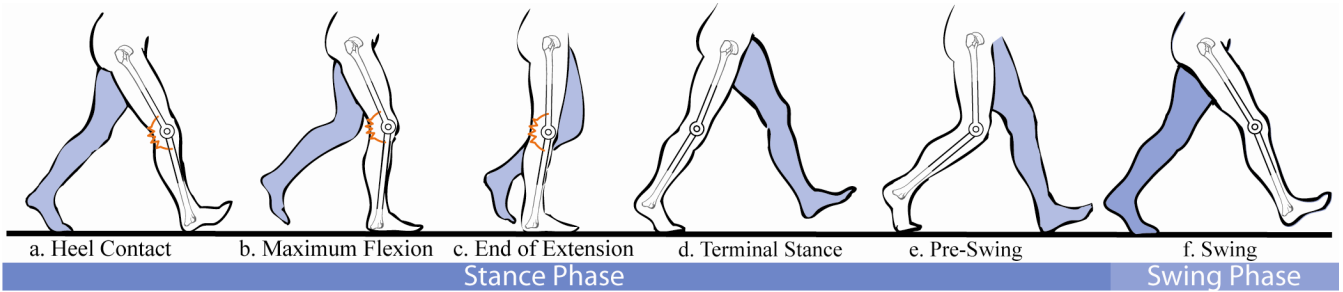


Fig. 2. Schematic posture of the lower limbs during a gait cycle (graphics adapted from [2]).

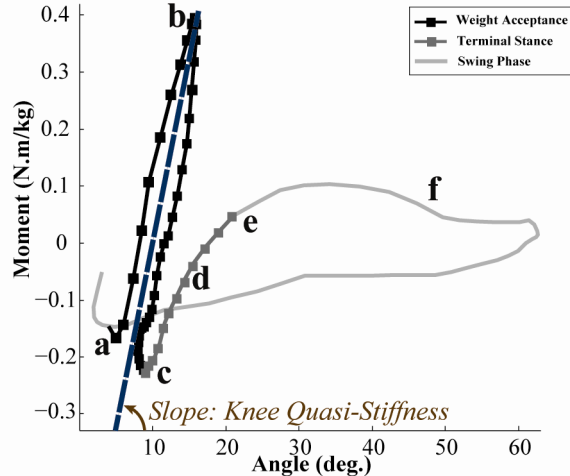


Fig. 3. The average moment-angle graph for the knee of a set of subjects walking at slow cadence of 85step/min (data from [1]). The knee behaves linearly in the weight acceptance phase of stance.

phase, and how the spring stiffness and assistance duration might be adjusted for users with different forms and levels of impairment.

## II. METHODS

### A. Moment-Angle Performance of Healthy Knee

Fig. 2 shows the schematic posture of the lower limbs during a gait cycle and Fig. 3 depicts the moment-angle graph of the knee for a healthy subject walking on level ground with the corresponding schematic configurations labeled. We divide the gait cycle into a stance and a swing phase. We further divide the stance phase into a weight acceptance phase (first  $\sim 40\%$ , as depicted in Fig. 2 and 3 points a to c) and a stance termination phase ( $\sim 40\text{--}63\%$ , as shown in Fig. 2 and 3 points c to e) [1, 2, 23, 25, 26]. As shown in Fig. 3, the knee exhibits a large moment and considerable flexion in the weight acceptance phase that follows a linear moment-angle pattern [23]. Considering the large moment that the knee is required to generate during the weight acceptance phase, it is highly prone to collapse at this phase without proper function of the musculature system or external assistance. In our previous work, we showed that the knee behaves close to a very stiff torsional spring (average  $\sim 3$  Nm/rad.Kg) that is dramatically loaded ( $0.5$  N/m.Kg) at the preferred gait speed [23]. The knee displays substantially smaller moments in the rest of the gait cycle that could imply a less eminent need for external assistance.

TABLE I. Target and Realized Values for the Design Parameters

Design Parameter	Target	Realized
Joint Excursion Under Load	$25^\circ$	$40^\circ$
Angular Resolution	$1^\circ$	$<1^\circ$
Engagement Delay	$\sim 0$ ms	$\sim 30$ ms
Maximum Moment	105 Nm	$>105$ Nm
Stiffness	$<750$ Nm/rad	750 Nm/rad
Angle of Engagement	$6\text{--}32^\circ$	$0\text{--}60^\circ$
Weight	3 Kg	3 Kg
Power	$\sim 0$ W	$\sim 0.3$ W

### B. Design Objectives

The biological moment-angle behavior of the unaffected knee can imply that in order for the CSCO to mimic the behavior of the knee in the weight acceptance phase, it should function close to a linear torsional spring. More specifically, the CSCO should engage a spring (sized based on the subjects' stature and speed) at the onset and disengage it at the end of the weight acceptance phase. In our previous works, we investigated the variability of kinematic parameters as well as the quasi-stiffness and work of the ankle and knee joints [23, 27]. We showed that the knee excursion ranges from  $2$  to  $30^\circ$  in the weight acceptance phase of the gait. This indicates that the CSCO should have a very high angular engagement resolution in order to be able to capture the natural knee flexion in stance. We also showed that the angle of equilibrium (the angle at which the knee moment is zero) of the torsional spring equivalent to the knee can attain values of up to  $32^\circ$ , depending on the gait speed and carried weight, implying that the angle of engagement should also be adjustable. The weight acceptance phase spans  $\sim 40\%$  of the gait that corresponds to a period of  $\sim 450$  ms, implying that the engagement and disengagement of the assistive device spring should ideally be instantaneous to be responsive in the weight acceptance period. We further showed that the knee quasi-stiffness modulates based on the gait speed and carrying weight. We envision the CSCO design to be capable of displaying a stiffness up to  $\sim 750$  Nm/rad and a maximum moment of  $\sim 105$  Nm/rad (selectable by changing out the spring to size for the specific individual) to be able to function for a wide range of adults [23]. *Torsional springs* provide substantially smaller range of stiffness as long as they remain within a reasonable size limit suitable for this application. Alternatively, we can incorporate linear springs tangential with a circular pulley to be able to exhibit the range of torsional stiffness required to support the knee.

The CSCO should also be as light as possible. We target a weight of 3 Kg, similar to that of the commercial SCOs such as the SensorWalk from OttoBock (as measured in our laboratory). Additionally, the device should demonstrate minimal electric power consumption and noise generation, as well as reasonable cosmetic appearance. Table I summarizes the target requirements we envision for the design of the CSCO.

### C. Description of the Stance Control Orthosis

Fig. 1 shows the CSCO worn on a healthy subject. The CSCO is composed of an orthotic device that is cuffed on the leg, and a control and recording module that is belted behind the waist of the subject. Note that this control module is much larger for the experimental prototype in order to allow for a range of telemetry and data collection purposes. In a commercial device, this module can easily be reduced to the size of a stack of business cards.

The orthotic device includes a stiffness switching module and a lateral joint designed for the CSCO, as illustrated in Fig. 4. Fig. 4 also shows the components of the engagement mechanism and the lateral joint. The stiffness switching module engages a spring during the weight acceptance phase of the gait (or whenever support is required including standing), and allows nearly free motion in the rest of the gait. The stiffness switching module is mounted on the orthosis thigh and, through a tendon, harnesses a pulley that is attached to the orthosis shank. As such, the pulley rotates when the knee flexes/extends, which then couples to the shaft of the stiffness switching module. Therefore, the linear stiffness exhibited at the shaft of the stiffness switching module is transformed to a torsional stiffness around the knee joint.

The controller receives the knee angle, the status of the heel and toe contact with the ground (through the FSR insoles), and the status of the device stiffness, and determines the engagement. It further wirelessly transfers data to a host computer for experimental purposes.

### D. Stiffness Switching Module

The stiffness switching module employs a friction-based latching mechanism and primarily comprises a hardened shaft, friction lever, bearing block, shock absorber, and an engagement mechanism that includes a worm-gear set, a DC motor, two push-buttons for feedback on the position of the lever, and structural components, as shown in Fig. 4. The concept of friction-based latching has been exploited before for clamping purposes [28], and has been used in a compliant prosthetic foot for many of the same reasons we chose it [29]. Here, the latching friction lever is being manipulated with a reliable mechanism that can automatically engage/disengage it with a shaft. As shown in Fig. 4, the motor spins a worm-gear that in turn moves a custom-made gear away/toward the lever. The gear can accordingly engage or disengage the friction lever with the hardened shaft.

Fig. 5 shows the stiffness module under two loading conditions: high-stiffness and low-stiffness. To engage the high-stiffness spring, the motor spins counterclockwise that

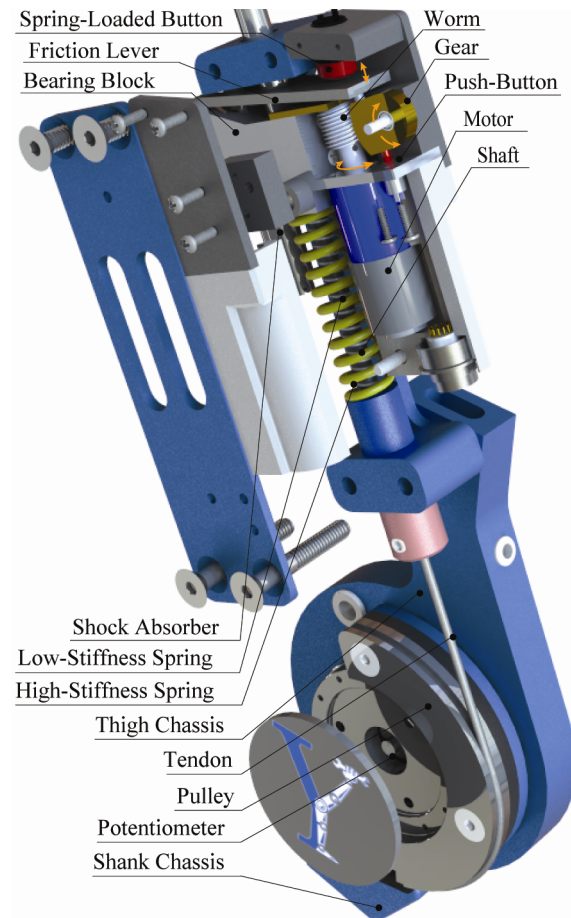


Fig. 4. Exploded view of the orthotic device including the stiffness switching module that harnesses the pulley of the lateral joint of the stance control orthosis.

moves the gear away from the friction lever as shown in Fig. 5-top, to clear behind the friction lever and press the push-button located behind it. When the gear is away from the lever, the spring loaded push-button pushes the friction lever that initiates the engagement of the friction lever to the shaft. Upon a load to the shaft, the engaged friction lever transfers the force of the loading shaft to the bearing block. Therefore, the shaft force tends to move the bearing block that, in turn, compresses the high-stiffness spring. The shaft force always compresses the low-stiffness spring. Therefore, the module exhibits an output stiffness that is equal to the summation of the stiffness of both springs.

In order to disengage the high-stiffness spring, the motor spins clockwise and moves the gear toward the lever as shown in Fig. 5-bottom. The gear pushes the friction lever away from the motor until the spring loaded push-button is pressed. In this configuration, the shaft can slide inside the bearing block and the friction lever. Therefore, any force on the shaft only compresses the low-stiffness spring. As such, the module only displays the low stiffness of the internal spring at the output. The shock absorber is incorporated to dissipate any remaining potential energy in the high-stiffness spring, provided it is disengaged while it is still compressed.

To transform the linear stiffness of the spring to a rotational stiffness around the knee, we assume that a knee flexion of  $\Delta\theta$  causes a compression of  $\Delta L$  in the spring. The

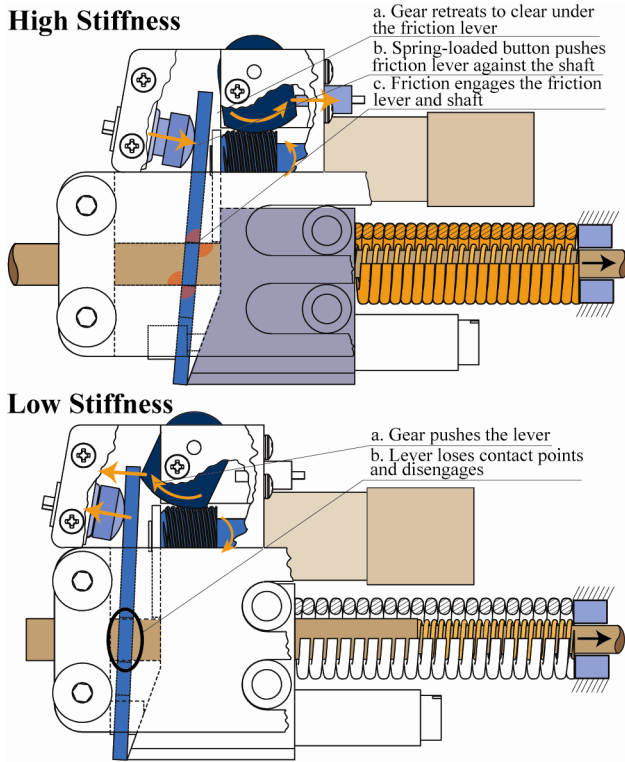


Fig. 5. Top, Engagement of high-stiffness spring: When the friction lever is engaged, flexion of the knee compresses both the low stiffness and high stiffness springs of the variable-stiffness module, Bottom, disengagement of high-stiffness spring: When the friction lever is disengaged, flexion of the knee only compresses the low stiffness spring. Only those parts of the variable-stiffness module that are involved in each mode are colored.

equivalent torsional stiffness around the knee would be calculated as:

$$K_{\theta} \cdot \Delta\theta = K_L \cdot \Delta L \cdot r \quad (1)$$

where  $K_{\theta}$  is the equivalent torsional stiffness exhibited at the knee joint,  $K_L$  is the linear stiffness of the stiffness switching module, and  $r$  is the radius of the pulley. Considering  $\Delta L = r\Delta\theta$ , we obtain:

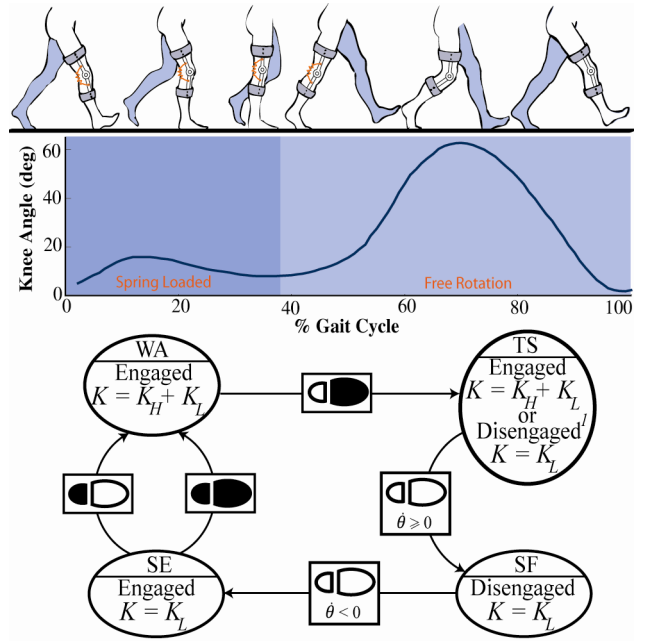
$$K_{\theta} = K_L \cdot r^2 \quad (2)$$

Therefore, to satisfy the range of torsional stiffnesses suggested in Section II-B, the linear spring should have a stiffness of up to  $\sim 290\text{N/mm}$  for a pulley radius of 51mm (2in.). Table I lists the values for the design parameters measured on the device. The device spring can be sized based on our previous works that establish a foundation for the selection of the stiffness of the knee and ankle joints of prosthetic and orthotic devices [24, 30].

#### E. Control Unit and Data Logger

The controller engages the high-stiffness spring during the weight acceptance phase by monitoring the direction of the knee joint velocity and foot contact with the ground, as schematically shown in Fig. 6-top and detailed here. The controller differentiates the knee joint angle obtained from a rotary potentiometer incorporated in the orthosis lateral joint, and senses the heel and toe contact with ground from an instrumented shoe insole.

Fig. 6-middle shows the knee angle in a gait cycle (data from [1]) and the period of engagement, in addition to Fig.



1. Engagement depending on the user's impairment level

Fig. 6. Top: Schematic function of the stance control orthosis, Middle: the knee angle profile for a healthy subject walking at the slow cadence of 85steps/min (data from [1]), Bottom: The state-machine employed for the control of the stance control orthosis in level ground walking.

WA: Weight Acceptance, TS: Terminal Stance

$K_H$ : High-Stiffness Spring Constant,  $K_L$ : Low-Stiffness Spring Constant

6-bottom that shows the state machine implemented to control the level of stiffness of the CSCO. The CSCO switches to high stiffness when the knee is in extension period of the swing phase (pre-stance phase when the heel and toe sensors are off), as a precaution against the mechanism switching delay. Since the knee is in extension mode, the high-stiffness spring is not loaded, because the engagement mechanism only latches in the flexion direction. When the heel contacts the ground, the lever remains engaged. At this point, the knee starts flexing and as a result the high-stiffness spring is loaded. The spring remains engaged when the foot is flat (i.e. both the heel and toe are in contact with the ground). The leg goes into the terminal stance phase when the toe is on the ground. The spring would disengage during the terminal stance for users who require external stabilization only during the weight acceptance phase. For users who require assistance during the entire stance phase (e.g. in case of complete impairment of the knee), the spring would remain engaged until the toe leaves the ground.

The data logger simultaneously records data obtained from the device and wirelessly transfers them to a host computer. The data logger allows for recording and transfer of up to 24 analog and digital channels. The controller transfers the knee angle and velocity, foot sole force, orthotic assistive moment, engagement status, and gait phase. The controller is also equipped with a user-interface that allows for synchronization of the device with the gait lab system, changing the control algorithm, emergency stop, and additional options for experimental purposes.

Fig. 7 shows the data obtained from the device for the treadmill gait of a healthy subject with height of 1.78 m and

weight of 71 Kg, and gait speed of 0.6 m/s. The subject wore the device under four conditions: a. Three different high-stiffness springs: 109, 175, and 333 Nm/rad, and b. Natural walking wearing the device and without the orthotic assistance. Fig. 7-top shows the knee angle, -middle the device stiffness, and -bottom the device assistive moment. Moreover, Fig. 8 shows the gait of a healthy subject wearing the CSCO and walking on a treadmill. Here, the orthosis employs a spring with stiffness of 109 Nm/rad and provides assistance only during the weight acceptance phase of the gait.

### III. DISCUSSION AND FUTURE WORK

This paper presents the mechanical design and control algorithm for the CSCO that can stabilize an impaired knee by implementing a spring in parallel with this joint in the stance and allow for free rotation during the swing phase of the gait. Here, the implemented compliance in the CSCO was biologically inspired from the moment-angle analyses of healthy human knee. We hypothesize that the implemented compliance would improve performance over current SCOs that rigidly lock the knee during the stance phase and allow free rotation during the swing. The potential benefits include a reduction in the compensatory movements of the unaffected joints, restoring the shock absorption function of the knee flexion in the stance, higher gait speed, and longer walking distances/intervals, all of which are being addressed in a human subjects study currently ongoing by the study researchers.

The developed CSCO might be used to restore the natural gait of users with gait impairment following injury, stroke, post-polio, multiple sclerosis, SCI, Patellofemoral Pain Syndrome, or any other musculoskeletal dysfunction that causes quadriceps weakness. The CSCO developed here can employ different levels of torsional stiffness and period of assistance depending on the level of functionality or impairment of the knee.

The current device satisfies nearly all the design criteria that were established based on the biomechanical behavior of a healthy knee joint, as outlined in Table I. Those criteria include high angular engagement resolution, instantaneous engagement, high angular stiffness, wide range of motion, variable angle of engagement, low noise generation, light weight, and low power consumption. However, the current prototype can be additionally improved in terms of endurance, weight, and engagement delay. Our initial experiments show that the CSCO can reliably operate on a healthy subject for at least 40,000 cycles without any systematic failure. In the next step, the CSCO would undergo many more loading cycles (e.g. ~500,000 cycles observed in a 6 months period, as suggested by [31]) to ensure the required mechanical reliability. The current design employs high-speed steel for the material of the friction lever, and the endurance of the device can be substantially improved using steel with higher hardness. Moreover, the mechanical system exhibits a 30 ms delay

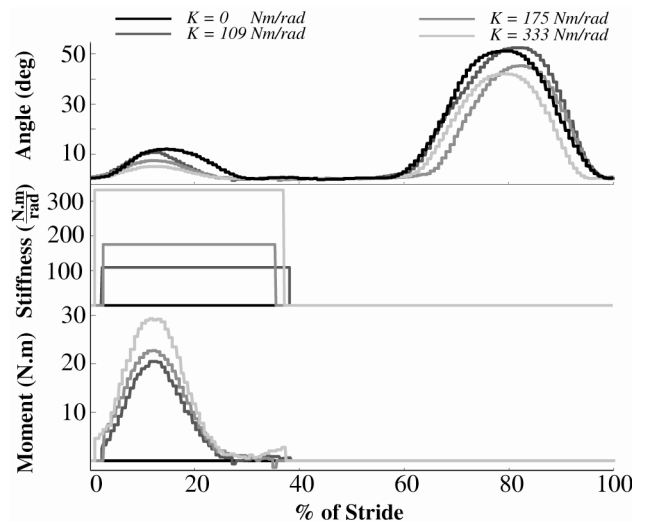


Fig. 7. Kinetic and kinematic data obtained from the stance control orthosis for a healthy subject with the height of 1.78 m and weight of 71 Kg, walking at speed of 0.6m/s on a treadmill. The orthosis employed three levels of stiffness including 109 Nm/rad, 175 Nm/rad, and 333 Nm/rad as well as natural walking (0 Nm/rad), Top: Knee joint angle, Middle: Device stiffness, and Bottom: Device assistive moment. The stiffness of the low-stiffness spring has been neglected.

during for the engagement, which is higher than the target value of ~0 ms. While the engagement delay has been overcome by a premature engagement signal during the pre-stance phase, a more careful selection of the motor and gear ratio might decrease the engagement and disengagement delay.

The CSCO tends to suppress the knee flexion of the healthy subject both in the stance and swing phases of the gait, hypothetically because a healthy knee can only accommodate the external stiffness of up to a certain level. This change of trajectory would be of a lesser extent for an impaired knee.

Additionally, both the orthosis and controller are currently oversized and can be made smaller and lighter. The spring module was built to accommodate a variety of gaits and movements, including uphill/downhill walking and stair ascent/descent, providing a range of motion under load of ~40°. During level ground walking, a healthy knee only requires ~15-20° of flexion during stance. Moreover, the current design is capable of operating under loads (more than 105 Nm) that are greater than what is observed for the average subjects in normal walking (~0.5 Nm/kg [26]). The control unit can be made significantly smaller and more efficient by eliminating the data logging and transfer functionality, such that the controller and battery pack can be implemented and embedded inside the lateral joint of the CSCO, saving space and reducing the power consumption of the device.

For the next steps in the project, we intend to conduct both impaired and unimpaired human walking experiments to evaluate the functionality of the CSCO, as well as the potential benefits that incorporation of compliance in these devices can introduce to the field of SCOs. We intend to conduct additional research to evaluate the timing and level

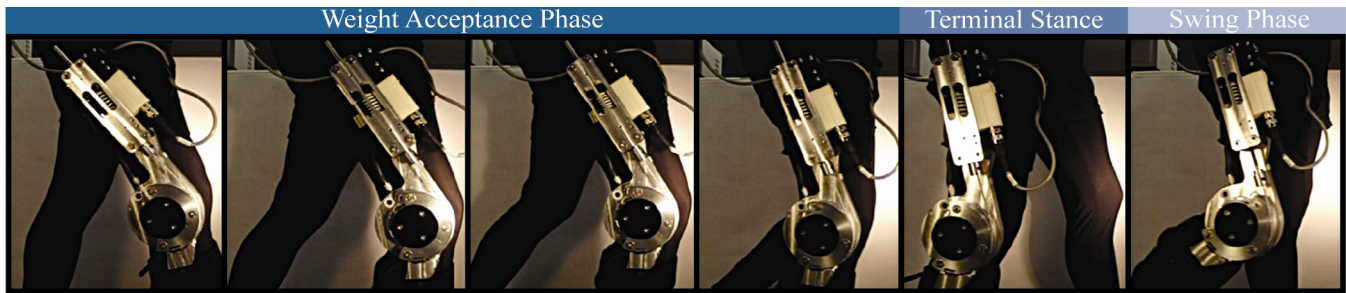


Fig 8. A healthy subject wearing the stance control orthosis and walking on a treadmill. The orthosis stabilizes the knee by implementing a sized spring in parallel with it during the weight acceptance phase of the gait and allows for free (low-stiffness) rotation during the rest of the gait.

of assistance for users with different gait impairments. We also intend to identify the target users for whom the device is most suitable.

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#### REFERENCES

- [1] D. A. Winter. (1983, Dec.). Biomechanical motor patterns in normal walking, *Journal of Motor Behavior*, 2(3), 302-330.
- [2] J. Rose, and J. G. Gamble, " Human walking," 3rd ed., Philadelphia: Lippincott Williams & Wilkins, 2006.
- [3] W. Robert, and M. Leslie. (1987) A physiologic rationale for orthotic prescription in paraplegia, *Clinical Prosthetics and Orthotics*, 11( 2), 66-73.
- [4] E. A. Hurley. (2006) . Use of KAFOs for patients with cerebral vascular accident, traumatic brain injury, and spinal cord injury, *Journal of Prosthetics and Orthotics*, 18(7), 199-201.
- [5] M. K. Taylor. (2006). KAFOs for patients with neuromuscular deficiencies, *Journal of Prosthetics & Orthotics*, 18(7), 202-203.
- [6] J. E. Earl, S. J. Piazza, and J. Hertel. (2004). The protonics knee brace unloads the quadriceps muscles in healthy subjects, *Journal of Athletic Training*, 39(1), 44-49.
- [7] J. B. Redford, J. V. Basmajian, and P. Trautman, "Orthotics : clinical practice and rehabilitation technology," New York: Churchill Livingstone, 1995.
- [8] S. Irby, K. Bernhardt, and K. Kaufman. (2005, Dec.). Gait of stance control orthosis users: the dynamic knee brace system, *Prosthetics and Orthotics International*, 29(3), 269-282.
- [9] S. Irby, K. Bernhardt, and K. Kaufman. (2007, Dec.). Gait changes over time in stance control orthosis users, *Prosthetics and Orthotics International*, 31(4), 353-361.
- [10] B. Zacharias, and A. Kannenberg. (2012). Clinical benefits of stance control orthosis systems: an analysis of the scientific literature, *JPO: Journal of Prosthetics and Orthotics*, 24(1), 2-7.
- [11] A. A. Rasmussen, K. M. Smith, and D. L. Damiano. (2007). Biomechanical evaluation of the combination of bilateral stance-control knee-ankle-foot orthoses and a reciprocating gait orthosis in an adult with a spinal cord injury, *JPO: Journal of Prosthetics and Orthotics*, 19(2), 42-47.
- [12] T. Yakimovich, E. D. Lemaire, and J. Kofman. (2006). Preliminary kinematic evaluation of a new stance-control knee-ankle-foot orthosis, *Clinical Biomechanics*, 21(10), 1081-1089.
- [13] T. Yakimovich, E. Lemaire, and J. Kofman. (2009) Engineering design review of stance-control knee-ankle-foot orthoses, *Journal of Rehabilitation Research and Development*, 46(2), 257-267.
- [14] P. Davis, T. Bach, and D. Pereira. (2010, Jun.). The effect of stance control orthoses on gait characteristics and energy expenditure in knee-ankle-foot orthosis users, *Prosthetics and Orthotics International*, 34(2), 206-215.
- [15] A. Cullell, J. C. Moreno, E. Rocon, A. Forner-Cordero, and J. L. Pons. (2009). Biologically based design of an actuator system for a knee-ankle-foot orthosis, *Mechanism and Machine Theory*, 44(4), 860-872.
- [16] A. McMillan, K. Kendrick, J. Michael, J. Aronson, and G. Horton. (2004). Preliminary evidence for effectiveness of a stance control orthosis, *JPO: Journal of Prosthetics and Orthotics*, 16(1), 6-13.
- [17] E. D. Lemaire, L. Goudreau, T. Yakimovich, and J. Kofman. (2009). Angular-velocity control approach for stance-control orthoses, *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 17(5), 497-503.
- [18] K. Bernhardt, S. Irby, and K. Kaufman. (2006, Dec.). Consumer opinions of a stance control knee orthosis, *Prosthetics and Orthotics International*, 30(3), 246-256.
- [19] A. Zissimopoulos, S. Fatone, and S. Gard. (2007). Biomechanical and energetic effects of a stance-control orthotic knee joint, *Journal of Rehabilitation Research and Development*, 44(4), 503-513.
- [20] A. Dollar, and H. Herr. (2008, Feb.). Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art, *IEEE Transactions on Robotics*, 24(1), 144-158.
- [21] S. Gard, and D. Childress. (2001). What determines the vertical displacement of the body during normal walking?, *Journal of Prosthetics and Orthotics*, 13(3), 64-67.
- [22] R. J. Ratcliffe, and K. G. Holt. (1997, Apr.). Low frequency shock absorption in human walking, *Gait & Posture*, 5(2), 93-100.
- [23] K. Shamaei, and A. M. Dollar. "On the mechanics of the knee during the stance phase of the gait." *IEEE International Conference on Rehabilitation Robotics (ICORR)*, Zurich, Switzerland, 2011.
- [24] K. Shamaei, G. S. Sawicki, and A. M. Dollar. (2013) Estimation of quasi-stiffness of the human knee in the stance phase of walking, *PLoS ONE*, 8(3), e59993.
- [25] J. Perry, "Gait analysis : normal and pathological function," Thorofare, NJ: SLACK, 1992.
- [26] D. Winter, "The biomechanics and motor control of human gait : normal, elderly and pathological," 2nd ed., Waterloo, Ont.: University of Waterloo Press, 1991.
- [27] K. Shamaei, M. Cenciari, and A. Dollar, "On the mechanics of the ankle in the stance phase of the gait." *Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, Boston, Massachusetts 2011.
- [28] Quick action bar clamp/spreader, by J. A. Sorensen, and L. Nebr. (1991, Mar.). Patent No. D333,602.
- [29] S. Collins, and A. Kuo. (2010, Feb.). Recycling energy to restore impaired ankle function during human walking, *PLoS One*, 5(2), e9370.
- [30] K. Shamaei, G. S. Sawicki, and A. M. Dollar. (2013) Estimation of quasi-stiffness and propulsive work of the human ankle in the stance phase of walking, *PLoS ONE*, 8(3), e59935.
- [31] T. Yakimovich, J. Kofman, and E. D. Lemaire. (2006) Design and evaluation of a stance-control knee-ankle-foot orthosis knee joint, *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 14(3), 361-369.