

## Stance control knee mechanism for lower-limb support in hybrid neuroprosthesis

Curtis S. To, PhD;<sup>1\*</sup> Rudi Kobetic, MS;<sup>1</sup> Thomas C. Bulea, MS;<sup>2</sup> Musa L. Audu, PhD;<sup>2</sup> John R. Schnellenger, MS;<sup>1</sup> Gilles Pinault, MD;<sup>1</sup> Ronald J. Triolo, PhD<sup>1-3</sup>

<sup>1</sup>Louis Stokes Cleveland Department of Veterans Affairs Medical Center, Cleveland, OH; Departments of <sup>2</sup>Biomedical Engineering and <sup>3</sup>Orthopaedics, Case Western Reserve University, Cleveland, OH

**Abstract**—A hydraulic stance control knee mechanism (SCKM) was developed to fully support the knee against flexion during stance and allow uninhibited motion during swing for individuals with paraplegia using functional neuromuscular stimulation (FNS) for gait assistance. The SCKM was optimized for maximum locking torque for body-weight support and minimum resistance when allowing for free knee motion. Ipsilateral and contralateral position and force feedback were used to control the SCKM. Through bench and nondisabled testing, the SCKM was shown to be capable of supporting up to 70 N-m, require no more than 13% of the torque achievable with FNS to facilitate free motion, and responsively and repeatedly unlock under an applied flexion knee torque of up to 49 N-m. Preliminary tests of the SCKM with an individual with paraplegia demonstrated that it could support the body and maintain knee extension during stance without the stimulation of the knee extensor muscles. This was achieved without adversely affecting gait, and knee stability was comparable to gait assisted by knee extensor stimulation during stance.

**Key words:** assistive technology, closed-loop control, exoskeleton, functional neuromuscular stimulation, gait, hybrid neuroprosthesis, hydraulics, orthosis, paraplegia, rehabilitation engineering, spinal cord injury.

### INTRODUCTION

A considerable effort has been made to develop and integrate a controllable knee mechanism in lower-limb bracing that supports the knee during the stance phase of gait and allows for free movement during swing. These stance control knee mechanisms (SCKMs) have utilized a myriad of design approaches, including bail locking [1], pin locking [2–4], a ratchet/pawl [5–6], cam locking [7], belt clamping [8], hydraulics [9], a magnetic particle brake [10], a wrap-spring clutch [11], a dog clutch via circular ratchet plates [12], a roller clutch [3], lever locking, and spring stiffness switching [13]. Many of these mechanisms are difficult to unlock under load [3,8,14], which may be necessary during the transition from the stance phase to the swing phase of gait. Users with weak or fatigued knee extensors may have difficulty extending the knee to unload the joint to unlock the mechanism.

**Abbreviations:** AFO = ankle-foot orthosis, FNS = functional neuromuscular stimulation, FSKC = finite state knee controller, FSR = force sensitive resistor, HNP = hybrid neuroprosthesis, IRGO = isocentric reciprocal gait orthosis, KAFO = knee-ankle-foot orthosis, SCI = spinal cord injury, SCKM = stance control knee mechanism, UECU = Universal External Control Unit.

\*Address all correspondence to Curtis S. To, PhD; 2700 Greenstone Boulevard, Apt 1404, Auburn Hills, MI 48326; 216-401-6444. Email: [cst@cwru.edu](mailto:cst@cwru.edu)

DOI:10.1682/JRRD.2010.07.0135

This is also a problem when bracing is combined with functional neuromuscular stimulation (FNS) in a hybrid neuroprosthesis (HNP) [15] for gait assistance in individuals with paraplegia [2]. The commercially available SCKMs do not provide sufficient control for this user population because they either require consistent full extension or a pre set orientation of the thigh to operate. Others only lock at discrete angles [14], which may result in a minimal degree of unsupported knee flexion after locking and thus complicate the contralateral swing leg clearance.

The objective of this study was to develop a new SCKM to provide improved reliability and functionality over existing mechanisms specifically for individuals with spinal cord injury (SCI) whose gait is powered by electrical stimulation of their paralyzed muscles. This article describes the development of the knee mechanism—from the design concept, through bench testing, to safety testing with nondisabled individuals—and the effectiveness of stance knee control during walking in paraplegia [16].

## DESIGN

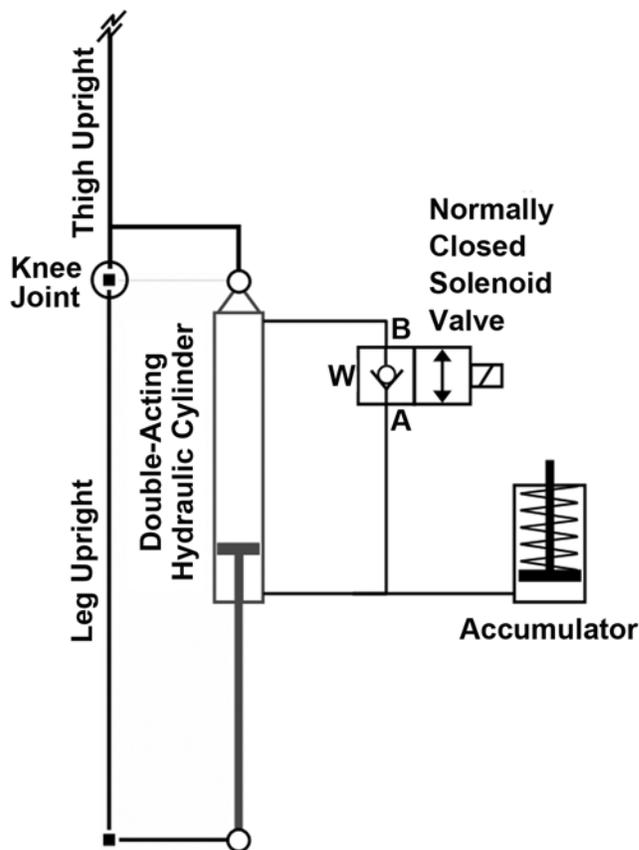
### Objective

The objective of the SCKM is to fully support the knee during standing and the stance phases of gait while allowing unhindered knee movement during stepping. The primary goal is to eliminate the need for muscle activity during static load supporting tasks or to reduce the duty cycle of electrical stimulation to the knee extensor muscles in an HNP system. This should delay the onset of fatigue by prolonging the rest periods between successive contractions [10] and, therefore, extend operating times and walking distances. Consequently, the SCKM was specified to have high mechanical impedance during stance to prevent falling from knee collapse or buckling resulting in insufficient foot-to-floor clearance of the contralateral leg during swing. On the other hand, low mechanical impedance is critical during swing to minimize the torque generated by electrical stimulation of the paralyzed muscles necessary to drive the mechanism and thereby reduce muscle fatigue. Finally, the SCKM must be capable of transitioning between states of high and low impedance responsively and consistently according to the dynamic requirements of gait.

### Conceptual Mechanism Design

The design of the SCKM consists of a miniature hydraulic system attached across the knee joint to the thigh and leg uprights of the knee-ankle-foot orthosis (KAFO) via revolute joints in a four-bar linkage arrangement for linear-to-rotary transmission (**Figure 1**). A two-way, two-position, normally closed solenoid valve inline between the ports of a single rod, double-acting hydraulic cylinder locks the knee mechanism without consuming power. The mechanism was designed to be locked only against knee flexion. Knee extension, corresponding to cylinder extension, can cause the pressure at port A relative to port B to exceed the valve cracking pressure, which forces the valve to open and thereby ineffective to lock against extension.

A single-acting, spring-loaded cylinder was employed as an accumulator to take up the fluid volume of the piston rod when flow is directed from the blind to the rod



**Figure 1.** Schematic of hydraulic stance control knee mechanism. A = valve output port, B = valve input port.

side of the cylinder during knee flexion. The accumulator also affects system behavior during knee extension. With an unpowered (closed) valve, fluid will flow into the accumulator as the cylinder extends, decreasing the pressure and increasing the volume (due to the expansion of existing air bubbles) of the system blind side. Consequently, the knee mechanism cannot effectively lock against flexion until the resting volume of the blind side is restored. The accumulator keeps the differential pressure across the valve below the cracking pressure when an unpowered knee mechanism is extending, preventing the valve from opening passively. Since the hydraulic fluid must be transferred to the blind side to effectively lock against knee flexion, feedback control must be used to open the solenoid valve during knee extension.

### Component Selection, Optimization, and Fabrication

The design goal was to maximize the locking torque of the SCKM at/near full knee extension by optimizing the moment arm as a function of knee angle while minimizing the torque that is required to drive the knee during the swing phase of gait when the mechanism is unlocked. The minimum locking torque specified was at least 50 N·m of knee flexion torque [17] during stance, and the maximum passive resistance from pressure losses through the opened valve was specified to not exceed 1 N·m. The off-the-shelf hydraulic components shown in the **Table**

were selected based on (1) minimal cylinder bore and stroke to maintain low flow and low weight, (2) maximal operating pressure to allow for minimal component size, (3) high valve flow coefficient to minimize pressure losses, and (4) minimal power consumption of the solenoid valve to prolong usage.

A simple mathematical model of the system shows that the maximum operating torque of 70 N·m at full extension decreases with increased knee flexion (**Figure 2**). Thus, for slow cadence walking, such as in pathological gait, the expected maximum extension torque required during stance [18], for an average male weighing 83 kg [19], should be easily accommodated by the SCKM with an approximate safety factor of three. With a linkage mechanism, a singularity occurs at 77° when the moment arm and the operating torque are equal to zero. Attempts at removing the singularity from the knee range of motion resulted in an overall decrease in the maximum operating torque of the mechanism. With the range of knee motion observed in walking to be 0° to 65° [17], the strategy used in the current design was to move the singularity to a point where knee locking is least likely to occur or be needed.

The passive resistance in the model was represented as the applied knee torque necessary to overcome the maximum pressure differential across the opened valve induced by a maximum knee angular velocity of 330 °/s,

**Table.**

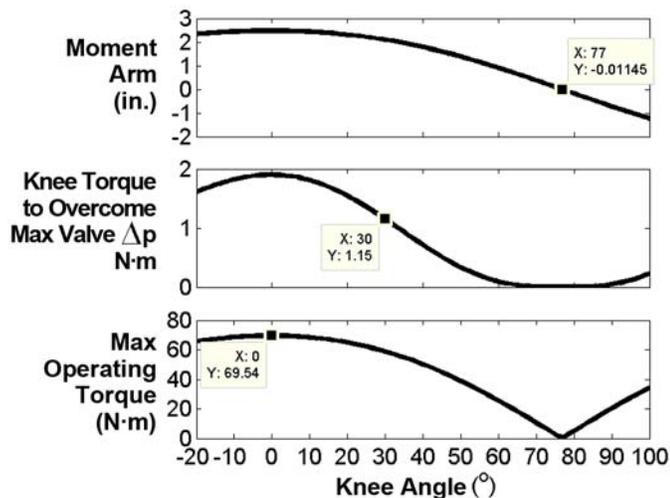
Knee mechanism hydraulic components.

Specification	Component		
	Cylinder	Valve	Accumulator
Manufacturer	Clippard Minimatic*	Allenair†	Clippard Minimatic*
Type	Double acting	Solenoid 2/2	Single acting; spring return
Bore	9/16 in.	—	3/4 in.
Port	1/16 in. NPT	1/8 in. NPT	1/8 in. NPT
Orifice	—	2.38 mm	—
Stroke	3 in.	—	1 in.
Rod Diameter	0.25 in.	—	0.25 in.
Voltage	—	12 VDC	—
Power Consumption	—	7 W	—
$C_v$	—	0.176 B→A; 0.166 A→B	—
Response Time (no load)	—	12 ms (on); 43 ms (off)	—
Maximum Operating Pressure	2,000 psi	—	250 psi
Spring Force	—	—	3 lb installed; 6 lb compressed
Cracking Pressure	—	46 ± 7 psi	—

\*Clippard Minimatic; Cincinnati, Ohio.

†Allenair Corporation; Mineola, New York.

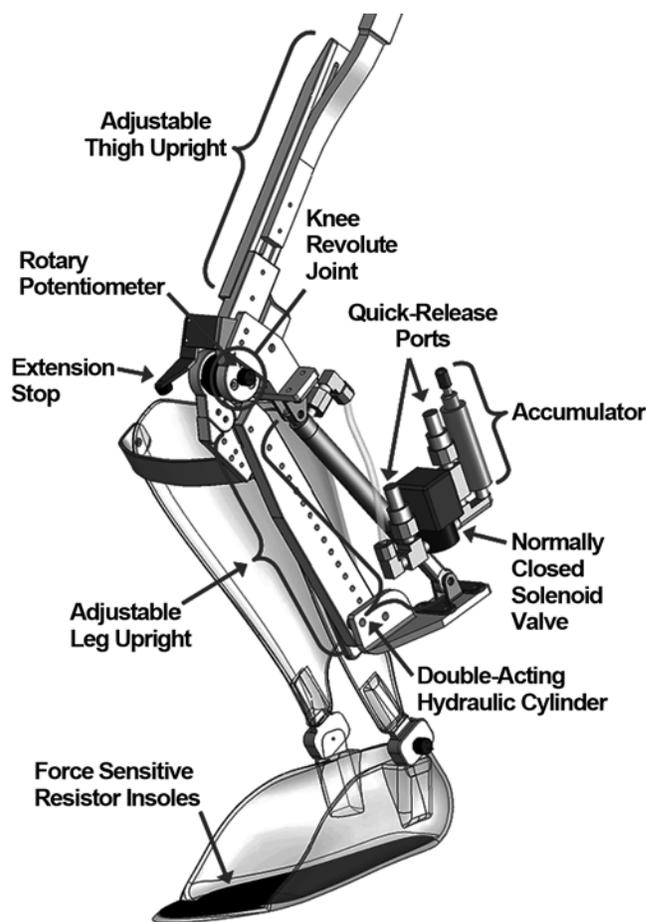
$C_v$  = valve flow coefficient, NPT = National Pipe Thread Taper.



**Figure 2.** Design parameters of stance control knee mechanism as function of knee joint angle. Max = maximum.

which is characteristic of nondisabled gait with a trunk-hip-knee-ankle-foot orthosis (with all joint constraints freed in the sagittal plane). Since the maximum angular velocity of the knee typically occurs at approximately 30° of knee flexion during swing [17], the maximum passive resistance contributed by the energy losses through the valve has been constrained to approximately 1 N·m (Figure 2).

The SCKM attached to the KAFO, as shown in Figure 3, weighs approximately 3.3 kg (7.3 lb). Hydraulic oil, ISO VG 46, was used as the fluid media. A Reynolds number of 721 calculated at an estimated maximum internal fluid velocity of 13.92 m/s indicates laminar flow. The structural components of the KAFO were fabricated from a combination of 6061 and 6063 aluminum alloy and 4142 and galvanized low-carbon steel alloys. The range of motion of the SCKM in flexion was limited to 106° by the off-the-shelf clevis components used to simplify construction. A mechanical extension stop was incorporated to prevent hyperextension of the knee. The KAFO was made adjustable for fitting different users. Hydraulic quick-release nipples were installed at each cylinder port for pressure measurements and system fluid priming. The normally closed solenoid valve has an override lever for manual operation of the valve during test or emergency conditions. Custom circuitry was developed to drive the valves for a pair of SCKMs. A 12 VDC (volts of direct current) supply, boost



**Figure 3.** Computer-aided design representation of stance control knee mechanism.

converted from a Sony NP-F970 47.5 Wh lithium ion rechargeable battery (Sony Corporation; Tokyo, Japan), was used to power the bilateral SCKMs.

### Closed-Loop Controller Design

A closed-loop controller was developed to unlock the SCKM during the swing phase of gait and lock the mechanism in extension during stance. The controller was designed as a finite state machine based on feedback signals including (1) contralateral knee valve state control signal, (2) contralateral forefoot and heel ground contact, (3) ipsilateral heel ground contact, (4) ipsilateral knee angle, (5) ipsilateral knee angular velocity, and (6) a synchronization signal from the baseline muscle stimulation pattern. The finite state knee controller (FSKC) and data acquisition software were developed and implemented in

the MATLAB/Simulink/xPC Target real-time environment (The Mathworks, Inc; Natick, Massachusetts).

#### *Sensors for Signal Feedback*

Feedback control signals for forefoot and heel contact were detected by force sensitive resistors (FSR) (B&L Engineering; Santa Ana, California) embedded in the soles of the shoes to measure foot-to-floor contact under the first metatarsal, great toe, fifth metatarsal, and heel of each foot. The knee angle was measured with a precision rotary potentiometer (Vishay Spectrol; Malvern, Pennsylvania) and angular velocity was obtained by signal differentiation. All sensors were connected to custom-designed signal processing circuitry and 12 VDC power supply. All signals were sampled at a frequency of 200 Hz and low-pass filtered through software. The potentiometer signal was low-pass filtered (5th-order Butterworth) at a cutoff frequency of 10 Hz and the FSR signals were low-pass filtered (7th-order Butterworth) at a cutoff frequency of 20 Hz.

#### *Finite State Knee Controller*

The finite state machine uses four rules that can act independently or mutually to unlock the SCKM for flexion. The thresholds for the FSKC were determined empirically during bench and human testing.

Rule 1 states that while the knee is flexed and extending, the mechanism is unlocked. An extending knee is indicated by an angular velocity under a preset threshold (extension is negative). This threshold was set to be two standard deviations above the steady state angular velocity signal at  $-6^\circ/\text{s}$ . The knee can extend even in a locked state since the rod side of the system is always open to the accumulator. The knee flexion is detected by two thresholds—the input knee angle must have extended below a first threshold ( $=3^\circ$ ) for the knee to be considered fully extended and flexed beyond a second threshold ( $=13^\circ$ ) for the knee to be considered flexed—to accommodate the mechanical compliance of the SCKM when locked and loaded in flexion. Rule 1 was established to prevent SCKM extension when the valve is unpowered, which can compromise the responsiveness of the device locking against flexion.

Rule 2 states that the SCKM unlocks if all three conditions are met: (1) the contralateral knee valve is closed unless the contralateral knee is extending and the valve is open to meet conditions of Rule 1; (2) the ipsilateral heel is off the ground (i.e., FSR is low), indicating either terminal

stance or preswing; and (3) either the contralateral forefoot or heel is in contact with the ground (i.e., FSR is high), indicating stance. Rule 2 coordinates locking/unlocking of the SCKM with gait events.

Rule 3 states that the SCKM must be unlocked during the swing phase of gait initiated by the timing signal from the preprogrammed stimulation pattern of the ipsilateral knee flexors and extensors muscle during swing. Rule 3 was established to prevent the knee mechanism from locking (due to Rule 2) during mid swing in the event of foot drag resulting in a high FSR signal.

Rule 4 states that once the stimulation pattern signal is low, the SCKM is unlocked until the knee has returned to a fully extended position (as determined by knee angle thresholds of Rule 1) or if both Rule 1 and Rule 2 dictate that locking should occur.

## EXPERIMENTAL METHODS

The performance of the prototype SCKM was first characterized in bench testing to verify the design criteria. Testing on three nondisabled individuals was conducted to validate the system response. The effectiveness of SCKM to support the knee during stance and allow the knee to move freely during swing when driven by electrical stimulation of the paralyzed muscles was tested in one individual with paraplegia implanted with a multi-channel FNS system [20].

### System Characterization

Bench testing was conducted on the SCKM to (1) quantify the passive resistance with respect to angular velocity, (2) verify that the mechanism can resist at least 50 N·m of flexion torque at/near full extension, (3) test whether the knee mechanism can reliably unlock under relatively high loads, and (4) quantify the mechanical compliance of knee flexion when the knee mechanism is locked. All bench testing was completed with the SCKM, as represented in **Figure 3**, secured to the actuator of a Biodex System 3 (Biodex Medical Systems, Inc; Shirley, New York) robotic dynamometer. In addition to the feedback sensor and valve control signals, torque was collected from the dynamometer, and cylinder blind and rod side pressures were measured with pressure transducers (Gems Sensors & Controls; Plainville, Connecticut) installed via the quick-release couplers. The dynamometer torque and cylinder pressure signals were low-pass filtered at a cutoff

frequency of 10 Hz (5th-order Butterworth) and 20 Hz (7th-order Butterworth), respectively. The experimental setup limited the range of motion of the SCKM to 90°.

#### *Passive Resistance*

Passive resistance of the knee mechanism actuated by the dynamometer was measured at angular velocities, ranging from 5 °/s to 150 °/s. The inertial component of the measured torque necessary to accelerate the mass of the dynamometer attachment and mechanism was subtracted from the total measured torque to obtain the passive resistance torque.

#### *Dynamic Characterization*

The dynamometer was set to apply a different maximum torque for each trial to test locking torque, unlocking response, and mechanical compliance. The mechanical compliance is the change in knee angle into flexion from the angle in which the valve of the mechanism transitioned to a closed state. The flexion torque contributed by gravity (12 N·m), from the mass of the dynamometer attachment and SCKM, was added to the measured torque applied by the dynamometer to determine the total applied torque to the locked SCKM.

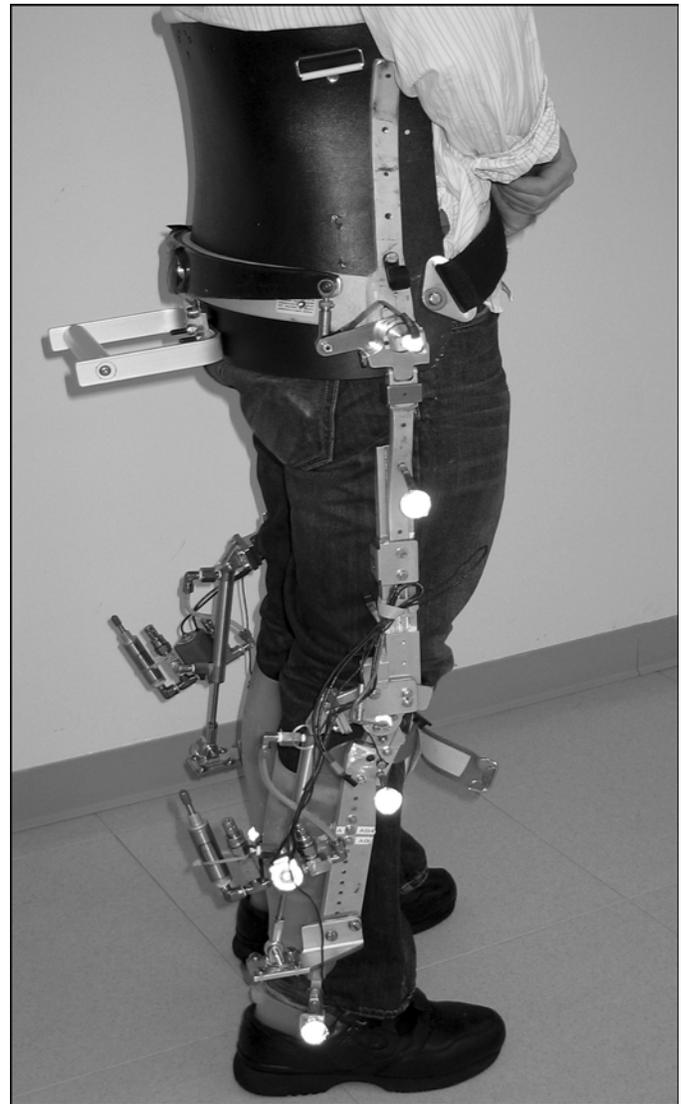
### **Clinical Evaluation**

#### *Study Participants*

The reliability of the FSKC for controlling the SCKM was first evaluated with three nondisabled individuals with an average weight of  $69 \pm 2$  kg. The prototype HNP system consisting of FNS to the hip, knee, and ankle musculature combined with bilateral SCKMs was evaluated in a participant with thoracic SCI (T9, American Spinal Injury Association A). He was 1.57 m tall and weighed 70 kg. He had received 24 percutaneous intramuscular electrodes targeting the hip flexors [21] (tensor fasciae latae, sartorius, and iliopsoas), hip extensors (posterior portion of adductor magnus, hamstrings, and gluteus maximus), knee flexors (gracilis and sartorius), knee extensors (vastus medialis, lateralis, and intermedius), ankle dorsiflexors (tibialis anterior), and ankle plantar flexors (gastrocnemius and soleus). He had been using his FNS-only system for exercise and walking with standby assist for more than 24 years. User-specific muscle stimulation patterns were set up for standing up, sitting down, and walking [22].

#### *Evaluation*

The SCKMs were installed on an isocentric reciprocating gait orthosis (IRGO) custom fitted to each participant (**Figure 4**). This orthosis weighed approximately 11 kg. For the participant with paraplegia, the IRGO was configured to reciprocally couple hip flexion with contralateral hip extension, thus maintaining upright trunk posture [23]. The ankle-foot orthosis (AFO) constrained the ankle joint to neutral. Donning of the IRGO consisted of fastening a strap across the lower torso, pelvis, and just below the knee and wearing the shoes, with embedded



**Figure 4.** Individual wearing pair of stance control knee mechanisms.

FSRs insole, over the AFOs while the participant was seated in a chair.

A target/host system was used to implement the HNP controller and collect data. The target computer ran an xPC Target kernel that facilitated the real-time implementation of the FSKC, FNS, and data acquisition. Since no FNS was needed for nondisabled volunteers, only Rule 1 and Rule 2 of the FSKC were implemented to control the SCKM. The target computer was equipped with data acquisition boards (National Instruments; Austin, Texas) for sensor signal acquisition and the output of control states to the exoskeleton. All communication between the target PC and exoskeleton was at a frequency of 200 Hz. The host computer ran a MATLAB-based graphical user interface developed to simplify the building, calibration, implementation, and testing of the target application.

The muscle stimulator unit, the Universal External Control Unit (UECU), delivered the FNS to drive limb motion [24]. The UECU contained two 12-channel stimulation output boards. The stimuli were biphasic charge-balanced asymmetric pulses with the stimulus current amplitude set at 20 mA. The target PC sent the instantaneous stimulus pulse width and interpulse interval parameters to the UECU. Baseline stimulus patterns were used to activate the muscles for the hip and ankle joints while the stimulus to the knee extensors was modulated from the baseline stimulation. Electrical stimulation for each consecutive step was manually triggered by the user via a finger switch.

Load cells (AMTI, Inc; Watertown, Massachusetts) were integrated into each handle of a two-wheel rollator to measure the vertical component of the forces applied by the upper limbs during gait. The load cell signals were low-pass filtered online (7th-order Butterworth) at a cut-off frequency of 20 Hz. A pressure transducer attached at each side of the valve measured the pressure differential, which along with the instantaneous geometry of the SCKM, was used to calculate the applied torque on the SCKM. Gait parameters were determined through the motion capture of markers, positioned on the participant and exoskeleton, using the Vicon MX40 (Vicon, Inc; Oxford, UK) motion analysis system. Average speed was determined from tracking of the forward velocity of a marker positioned approximately at the participant's center of mass. Step length was determined by measuring the distance between calcaneus markers upon heel strike.

Cadence was determined by averaging the duration of each step.

The participants walked with the exoskeleton along an 8-meter walkway. All participants walked with a two-wheel rollator and a spotter for safety. Additionally, the participant impaired by SCI walked with the knee extensor stimulation either turned off during stance, to determine whether the SCKM was capable of fully supporting the knee, or turned on (i.e., baseline stimulation) during stance, as a control case. Each case was randomized over six trials. Approximately 20 strides were analyzed for each case. Analysis of variance with 95 percent confidence ( $p < 0.05$ ) determined statistically significant differences in user effort and gait parameters between stance supported by only the SCKM versus stance supported by stimulated muscles.

## RESULTS

### System Characterization

#### *Passive Resistance*

For all angular velocities, the mean passive resistance torque magnitude did not exceed 2.0 N·m in flexion and 1.0 N·m in extension. A statistical difference in resistance was found between flexion and extension ( $p < 0.001$ ). The passive resistance is larger in flexion because the accumulator pressure opposes SCKM flexion but assists in extension due to the volume differential between cylinder sides. In flexion, passive resistance was independent of angular velocity. In the extension, passive resistance was independent of angular velocities up to 120 °/s ( $p = 0.35$ ). However, a statistical difference in resistance was found between the low angular velocities and 150 °/s ( $p < 0.001$ ). Extrapolated from first-order least squares regressions, the mean applied passive resistance was 2.1 N·m and 0.9 N·m at a maximum knee angular velocity of 350 °/s [17] for flexion and extension, respectively.

#### *Dynamic Parameters*

A maximum flexion torque of 71 N·m was applied on the locked SCKM. The duration to unlock the SCKM by opening the valve was approximately 0.2 s with a valve pressure differential of up to 700 psi corresponding to a flexion torque of 49 N·m at full extension. When unloaded,

the SCKM required only 12 ms to unlock; however, once loaded, it required a minimum of 180 ms to unlock.

**Figure 5** shows the knee compliance ( $\Delta$  angle) with respect to the applied flexion torque. At the minimum specified locking torque of 50 N·m, 2.5° to 4.5° of compliance was observed.

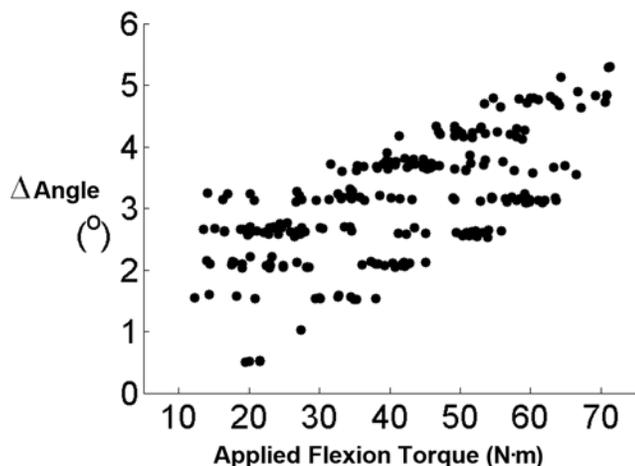
## Clinical Evaluation

### Nondisabled Testing

During evaluation with nondisabled subjects, unlocking of the SCKM began at the end of midstance (38% gait cycle) with the majority occurring during preswing phase (49% to 64% gait cycle). With inadequate foot-to-floor clearance, the SCKM remained locked during swing. This shortcoming motivated implementation of Rule 3 of the FSKC. The transition to a locked state always occurred consistently upon heel strike. The average power consumption per stride of the bilateral SCKMs at walking speeds between 0.22 m/s and 0.67 m/s was quantified to be approximately 8.5 W, thus allowing for more than 5 hours of walking in the current configuration.

### Spinal Cord Injury Testing

**Figure 6** shows knee angle, baseline quadriceps stimulus pulse width for walking with only FNS, actual quadriceps stimulus pulse width with stance supported only by the SCKM, calculated torque applied to the SCKMs, and the vertical component of the upper-limb

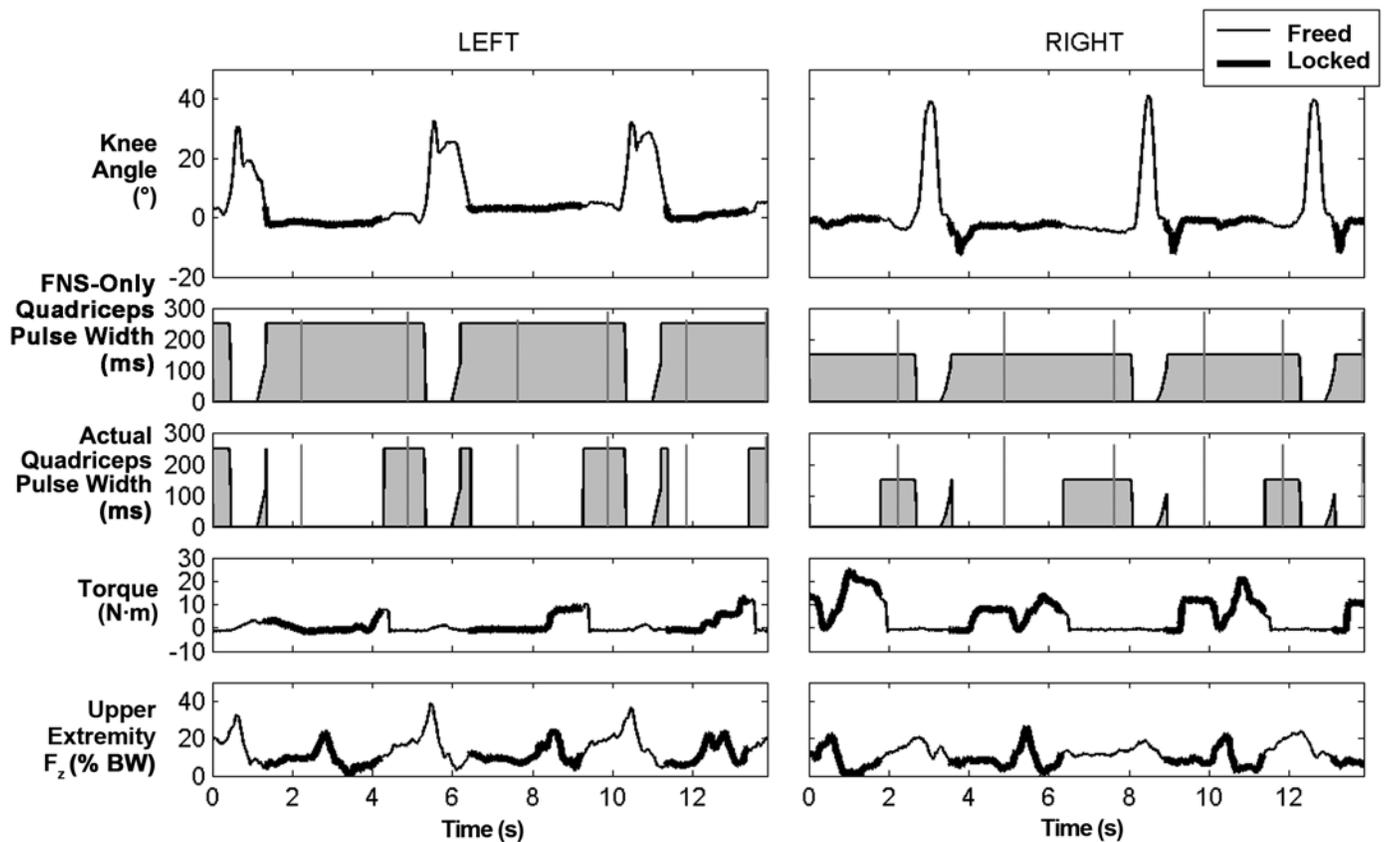


**Figure 5.** Change in knee flexion angle when stance control knee mechanism is locked (compliance) with respect to applied flexion torque.

forces applied to the walking aid for three typical strides of the participant with paraplegia walking with bilateral SCKMs for the left and right legs. The thick regions on the curves indicate the periods in which the SCKM was locked against knee flexion. Note that the knee could extend when the SCKM was locked against flexion. The asymmetry in the kinematics between left and right knees may be attributed to both variations in muscle response and voluntary user actions on the walker to accommodate each step. In the case of the right knee, impact of the posterior portion of the heel caused hyperextension after heel strike. This was less apparent in the left knee. Step initiation via finger switch is indicated by the pulses in the stimulus pulse width plots. Stimulation to the knee extensors was turned off during stance with the SCKM providing full support (**Video**). Thus, knee extensor activity is substantially reduced relative to walking with only FNS. Flexion torque applied to the SCKM was consistently significantly larger when the knee extensor stimulation was turned off during stance as compared with when the knee extensor stimulation was turned on (i.e., baseline stimulation) during stance (left:  $p < 0.001$ ; right:  $p < 0.001$ ). At the average walking speed of 0.15 m/s the average maximum torque on the SCKM with stimulation turned off was  $9.0 \pm 2.6$  N·m (left) and  $19.8 \pm 4.6$  N·m (right), whereas only  $2.8 \pm 1.0$  N·m (left) and  $8.4 \pm 6.1$  N·m (right) were measured with stimulation turned on. The compliance measured in the SCKM when locked was not statistically different with stimulation as compared with no stimulation (left:  $p = 0.29$ ; right:  $p = 0.36$ ), with the average SCKM compliance being less than 1°. The range of knee motion during swing was not statistically different between cases (left:  $p = 0.83$ ; right:  $p = 0.24$ ), with the average maximum knee flexion of  $28^\circ \pm 4^\circ$  (left) and  $36^\circ \pm 5^\circ$  (right) for the stimulation off case.

The mean and average maximum upper-limb support forces over strides were approximately 12 percent ( $p = 0.28$ ) and 32 percent ( $p = 0.19$ ) of the participant's body weight on the left arm, respectively, with and without knee extensor stimulation during stance. Similarly, the mean and average maximum upper-limb support forces were approximately 11 percent ( $p = 0.77$ ) and 23 percent ( $p = 0.65$ ) of the participant's body weight on the right arm, respectively, with and without knee extensor stimulation during stance. Thus, no significant additional upper-limb effort was required when knee support was provided by only the SCKM with muscles stimulation.

Between cases, no differences ( $p > 0.05$ ) were observed in speed, cadence, and step length.



**Figure 6.**

Knee angle, baseline quadriceps stimulus pulse width for walking with only functional neuromuscular stimulation (FNS), actual stimulus pulse width with only stance control knee mechanism (SCKM) supporting knee during stance, calculated torque resisted by SCKMs, and upper-limb forces ( $F_z$ ) applied to walker represented as percentage body weight (% BW) for three strides. Step trigger pulses are indicated in stimulus pulse width plots.

## DISCUSSION AND CONCLUSIONS

The results show the feasibility of utilizing a hydraulic approach in the development of a controllable mechanism for providing knee stability during stance when restoring gait after paralysis from SCI. The design objectives of this hydraulic SCKM were to (1) fully support the knee in extension during stance, (2) allow for unimpeded knee flexion during swing, and (3) consistently unlock under an applied flexion load. The development of the SCKM consisted of the simulation and optimization of the conceptual design, bench testing to verify design specifications, nondisabled testing to confirm operational response and safety, and system validation on a user with paraplegia.

The design specifications were verified through bench testing. First, the SCKM was capable of supporting more than 50 N·m of applied flexion torque with less than 5° of

mechanical compliance. Second, the average passive resistance of the SCKM was less than 2 N·m and 1 N·m, respectively, during flexion and extension. For this passive resistance during gait to be overcome, approximately 13 percent of knee flexion and only 1 percent of the knee extension torque generated by FNS was required [22]. Finally, SCKM was shown to have improved on existing designs in that it was capable of consistently unlocking under an applied flexion torque of up to 49 N·m.

Evaluation of the prototype HNP with bilateral SCKMs with an individual with paraplegia demonstrated that the SCKM was capable of fully supporting the user during the stance phase of gait without the need for electrical stimulation of knee extensors or additional arm support. No statistical differences were observed in the mechanism compliance, upper-limb effort, gait speed, cadence, and step length between walking with knee

extensor stimulation during stance and SCKM only supported stance without knee extensor stimulation.

The modest knee range of motion observed with the individual with paraplegia relative to nondisabled gait may be contributed to weak knee flexors, mechanism passive resistance, and/or mechanism mass [17]. The torque generated by the knee flexors can be augmented by the addition of an active component to the SCKM. This would, however, increase system mass and power consumption. The passive resistance as quantified in this study was only from the energy losses within the hydraulics. Future work will focus on examining the influence of the mass and inertial forces of the SCKM on knee joint kinematics. Currently, the hydraulics comprise 26 percent (=0.85 kg) of the total mass of the SCKM and KAFO. Thus, structural components of the mechanism (i.e., uprights, mounting brackets, and AFO) weigh approximately 2.48 kg. When these structural components were fabricated, the mass and overall mass distribution over the limb were not principal considerations. Many of these parts were overengineered using steel alloys to assure structural integrity and safety during mechanical load testing and human evaluation. Full optimization of the SCKM in relation to the materials utilized and mass distribution to minimize inertial effects has yet to be accomplished. To fully take advantage of the reduction in muscle activity as a result of the SCKM, future research will focus on the effect of system mass on the energy costs. The objective is to understand whether the energy saved from reducing muscle activity outweighs the energy lost from moving the mass of the SCKM and whether this net saved energy is functionally relevant in positively affecting walking distance and duration. As a portable controller has yet to be developed, the use of the SCKM was confined to the laboratory. Future work will focus on minimizing the weight and geometry of the SCKM as well as the development of a wearable/portable control unit to perform metabolic and walk-to-fatigue studies outside the laboratory.

The development of the hydraulic knee mechanism from conception to implementation was expedited by use of a simplified transmission solution. This compromise reduced the range of motion in which the mechanism can effectively support the user. Future work will include the selection of an optimal transmission type. A rack-and-pinion solution could facilitate a constant moment arm with respect to knee angle. Thus, the constraint could support against a high torque at any angle in which the knee is locked. However, the gears may increase the size

and weight of the device. Another solution may be to use a more complex linkage or a cam design to optimize the profile of the moment arm with respect to the knee angle. Similar to the current design, larger moment arms will occur at small knee angles that require high impedance while smaller moment arms will occur at larger knee angles coincident with high angular velocities during gait to minimize mechanism passive resistance. With this approach, a larger cylinder bore can be used to increase the maximum operating torque without substantially increasing passive resistance.

## ACKNOWLEDGMENTS

### Author Contributions:

*Study concept and design:* C. S. To, R. Kobetic, R. J. Triolo.

*Acquisition of data:* C. S. To, R. Kobetic, T. C. Bulea, J. R. Schnellenger, M. L. Audu.

*Analysis and interpretation of data:* C. S. To, R. Kobetic, T. C. Bulea, M. L. Audu.

*Drafting of manuscript:* C. S. To, R. Kobetic.

*Critical revision of manuscript for important intellectual content:*

C. S. To, R. Kobetic, T. C. Bulea, M. L. Audu, R. J. Triolo.

*Statistical analysis:* C. S. To.

*Obtained funding:* R. Kobetic, R. J. Triolo.

*Administrative, technical, or material support:* T. C. Bulea, J. R. Schnellenger, M. L. Audu, G. Pinault.

*Study supervision:* R. Kobetic, R. J. Triolo.

**Financial Disclosures:** The authors have declared that no competing interests exist.

**Funding/Support:** This material was based on work supported in part by the Department of Veterans Affairs, Rehabilitation Research and Development Service (grant A6404R) and the Department of Defense, Congressionally Directed Medical Research Program (grant PR043074).

**Additional Contributions:** We thank Drs. Robert F. Kirsch, Patrick E. Crago, and Roger D. Quinn for their advice and contributions. Dr. To is currently working at Chrysler Group, LLC, Auburn Hills, Michigan.

**Institutional Review:** All volunteers signed a consent form approved by the institutional review board of the Louis Stokes Cleveland Department of Veterans Affairs Medical Center.

**Participant Follow-Up:** The authors plan to inform participants of the publication of this study.

## REFERENCES

1. Lehmann JF, Stonebridge JB. Knee lock device for knee ankle orthoses for spinal cord injured patients: An evaluation. *Arch Phys Med Rehabil.* 1978;59(5):207–11. [\[PMID: 655831\]](https://pubmed.ncbi.nlm.nih.gov/655831/)

2. Kagaya H, Shimada Y, Sato K, Sato M, Iizuka K, Obinata G. An electrical knee lock system for functional electrical stimulation. *Arch Phys Med Rehabil.* 1996;77(9):870–73. [PMID: 8822676] DOI:10.1016/S0003-9993(96)90272-5
3. Harrison R, Lemaire E, Jeffreys Y, Goudreau L. Design and pilot testing of an orthotic stance-phase control knee joint. *Orthopadie Technik.* 2001;3:2–4.
4. Kim G, Kang S, Kang S, Ryu J, Mun M, Kim K. Unlockable knee joint mechanism for powered gait orthosis. *Int J Precision Eng Manufacturing.* 2009;10(3):83–89. DOI:10.1007/s12541-009-0051-y
5. Van Leederdam NG, Kunst EE. New UTX-swing orthosis: Normal gait and safe standing. *Orthopadie Technik.* 1999;50:506–15.
6. Nijenbanning G, Goudsmit JA, inventors. Gravity operated locking hinge. United States patent US 20,030,153,854. 2003 Aug 14.
7. Hatton BJ, Hatton DL, Wallace ZG, inventors. Articulating knee supports. United States patent US 6,635,024. 2003 Oct 21.
8. Yakimovich T, Kofman J, Lemaire ED. Design and evaluation of a stance-control knee-ankle-foot orthosis knee joint. *IEEE Trans Neur Syst Rehabil Eng.* 2006;14(3):361–69. [PMID: 17009496] DOI:10.1109/TNSRE.2006.881578
9. McGhee RB, Tomovic R, Yang PY, MacLean IC. An experimental study of a sensor-controlled external knee locking system. *IEEE Trans Biomed Eng.* 1978;25(2):195–99. [PMID: 640706] DOI:10.1109/TBME.1978.326247
10. Goldfarb M, Duffee WK. Design of a controlled-brake orthosis for FES-aided gait. *IEEE Trans Rehabil Eng.* 1996;4(1):13–24. [PMID: 8798068] DOI:10.1109/86.486053
11. Irby SE, Kaufman KR, Wirta RW, Sutherland DH. Optimization and application of a wrap-spring clutch to a dynamic knee-ankle-foot orthosis. *IEEE Trans Rehabil Eng.* 1999;7(2):130–34. [PMID: 10391582] DOI:10.1109/86.769402
12. Sclater N, Chironis NP. Mechanisms and mechanical devices sourcebook. 3rd ed. New York (NY): McGraw-Hill; 2001.
13. Moreno JC, Brunetti F, Rocon E, Pons JL. Immediate effects of a controllable knee ankle foot orthosis for functional compensation of gait in patients with proximal leg weakness. *Med Biol Eng Comput.* 2007;46(1):43–53. [PMID: 17926076]
14. Yakimovich T, Lemaire ED, Kofman J. Engineering design review of stance-control knee-ankle-foot orthoses. *J Rehabil Res Dev.* 2009;46(2):257–67. [PMID: 19533539] DOI:10.1682/JRRD.2008.02.0024
15. Marsolais EB, Kobetic R, Polando G, Ferguson K, Tashman S, Gaudio R, Nandurkar S, Lehnert HR. The Case Western Reserve University hybrid gait orthosis. *J Spinal Cord Med.* 2000;23(2):100–108. [PMID: 10914350]
16. To CS. Closed-loop control and variable constraint mechanisms of a hybrid neuroprosthesis to restore gait after spinal cord injury [dissertation]. [Cleveland (OH)]: Case Western Reserve University; 2010.
17. Perry J. *Gait analysis: Normal and pathological function.* Thorofare (NJ): SLACK; 1992. p. 91–92.
18. Winter DA. *The biomechanics and motor control of human gait: Normal, elderly and pathological.* 2nd ed. Waterloo (Canada): Waterloo Press; 1991.
19. Human weight [Internet]. [updated 2011 Apr 15; cited 2010 Nov 20]. Available from: [http://www.articleworld.org/index.php/Human\\_weight](http://www.articleworld.org/index.php/Human_weight)
20. Agarwal S, Kobetic R, Nandurkar S, Marsolais EB. Functional electrical stimulation for walking in paraplegia: 17-year follow-up of 2 cases. *J Spinal Cord Med.* 2003;26(1): 86–91. [PMID: 12830975]
21. Nandurkar S, Marsolais EB, Kobetic R. Percutaneous implantation of iliopsoas for functional neuromuscular stimulation. *Clin Orthop Relat Res.* 2001;(389):210–17. [PMID: 11501813] DOI:10.1097/00003086-200108000-00030
22. Kobetic R, Marsolais EB. Synthesis of paraplegic gait with multichannel functional neuromuscular stimulation. *IEEE Trans Rehabil Eng.* 1994;2(2):66–79. DOI:10.1097/00003086-200108000-00030
23. Dall PM, Müller B, Stallard I, Edwards J, Granat MH. The functional use of the reciprocal hip mechanism during gait for paraplegic patients walking in the Louisiana State University Reciprocating Gait Orthosis. *Prosthet Orthot Int.* 1999;23(2):152–62. [PMID: 10493143]
24. Trier S, Vrabec T, Weisgarber J. Using functional electrical stimulation to restore movement to individuals with neuromuscular disabilities. *MATLAB Digest: Academic Edition.* 2008. p. 1–4.

Submitted for publication July 20, 2010. Accepted in revised form January 20, 2011.

This article and any supplementary material should be cited as follows:

To CS, Kobetic R, Bulea TC, Audu ML, Schnellberger JR, Pinault G, Triolo RJ. Stance control knee mechanism for lower-limb support in hybrid neuroprosthesis. *J Rehabil Res Dev.* 2011;48(7):839–50.

DOI:10.1682/JRRD.2010.07.0135

ResearcherID: Curtis S. To, PhD: C-8101-2011.



