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Design of Stance-Control Knee-Ankle-Foot Orthotics

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Abstract: "The Knee ankle foot orthosis (KAFOs)" are full leg braces for individuals with knee extensor feebleness, designed to support the person during weight bearing activities by avoiding knee flexion. KAFOs normally outcome in an unusual gait design and are principally used for level ground walking. Mechanical and biomechanical examination established that the project opposed knee motion through stance stage, discharged the knee joint exclusive of confining the knee's range of association, and providing flexion confrontation during stair descent. Strategy presentation and associated testing processes were established to further progress joint functioning and to authorize the strategy prior to testing on persons with knee extensor feebleness.

Keywords: *Knee ankle foot. Orthotics, Stance*

I. Introduction

Throughout the movement, the knee presents shock absorption, maintain stability during the situation, and contribute to the development of the parties during the swing. The knee extensor strengths resist flexing the knee throughout posture, absorbing forces due to body weight, and facilitating the development of limbs. When the moment of external knee flexion occurs on the knee, the extensor muscle group in the knee generates a conflicting extension moment. If the knee ligament ligaments are not strong enough, the knee will collapse due to the external flexion of the knee and the person will drop (Buranfield, 2011).

"The Knee ankle foot orthosis (KAFOs)" are complete leg supports designed for persons with knee extensor feebleness. Conventional "(KAFOs)" designs can lock the knee in an extended position, which keeps the individual stable throughout stance. However, the resulting stiff-legged gait is an unnatural gait pattern that is inefficient for the user and can lead to soft tissue injury (P. Bowker, 2013).

Stance control "(KAFOs)""(SCKAFOs)" support the leg during mass bearing and authorization free knee movement during swing stage. If the "(SCKAFO)" user has adequate hip muscle strength for forward progression, "(SCKAFOs)" improve mobility compared to static knee KAFOs and provide many clinical benefits (A. Cuallell, 2011).

The most functionally advanced "(SCKAFOs)" are microprocessor controlled and provide more reliable mode switching. Microprocessor controlled "(SCKAFOs)" can be designed with multiple settings that allow the user to more easily negotiate different types of terrain (Taaylor, 2016).

Personalities with knee extensor feebleness need a "(KAFO)" to provision the knee during stance, particularly when the knee experiences an external flexion moment. Studies have shown that "(SCKAFOs)" allow for a more natural gait pattern than conventional "(KAFOs)"due to knee joint release during swing, thereby enhancing mobility and walking. "(SCKAFOs)" that are presently accessible on the marketplace use mechanical or microprocessor control. Mechanical devices have the advantage of not requiring external power and are generally smaller,

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more aesthetically pleasing, and less expensive. Microprocessor controlled orthosis have more reliable switching between modes and the ability to sense various terrains. An issue with many orthotic devices is that they require a fully extended knee to engage the locking mechanism, which may not be possible for all individuals with knee extensor feebleness. Such individuals may not have the muscle strength and/or control to fully extend their knee consistently, and the orthosis will not catch an individual who stumbles and lands on a flexed knee. Other prominent issues that determine if an orthosis will be accepted by the user are device weight, size, and noise (K. Shamaaei, 2014).

Ideally an "(SCKAFOs)" should be made with minimal dimensions so that the device can be worn underneath pants, making the brace more aesthetically pleasing. Another issue is that "(SCKAFOs)" are expensive. Individuals with disabilities have reported that cost was a major reason for their unmet assistive device needs (K. Shamaaei, 2014).

Presently the most versatile orthosis available is the Otto Bock C-Brace, which uses a linear hydraulic actuator to control knee flexion. The C-Brace's hydraulic actuator system has many advantages, primarily the ability to provide in-stance dampening, allowing the user to walk with a more natural gait pattern. The C-Brace also has control modes for walking on ramps, stairs, uneven ground, or to sit naturally. Unfortunately, this orthosis is very bulky, and the joint unit on the lateral side is too large to fit beneath clothing (E. D. Laemaire, 2013).

The new knee joint, known as the "Ottawalk Variable Speed(OWVS)", provides variable knee flexion resistance to improve level ground walking and stair descent . A manual, electronic control method was implemented to switch between valve settings, enabling valve and preliminary biomechanical tests (Zachaarias, 2012).

1. Design and progressive

1.1 Mechanical Strategy

The Ottawalk-Variable Speed "(OWVS)" must function as a "(SCKAFO)"for walking, supporting the knee during the stance phase and allowing uninhibited knee flexion and extension through the swing stage. The knee joint's "range of motion (ROM)" is "$0\text{-}70^0$". The maximum knee angular velocity for level ground walking at a normal pace is approximately "$350^0/S$" in flexion and "$350^0/S$" in extension. These angular velocities are for capable - bodied individuals walking at a normal cadence, and individuals requiring a "(KAFO)" will likely walk at a slower cadence. Winter reported that the peak knee flexion angular velocity for slow cadence walking was approximately "$285^0/S$". The extreme exterior extension moment during level ground, normal cadence walking occurs during early stance and can be as large as "1.0 Nm/Kg" (mean plus one standard deviation). Slow cadence walking generates a lower external knee extension moment "$0.416 \pm 0.342\ \mathrm{Nm/Kg}$"(Maartin, 2010).

For level ground walking, an ideal "(SCKAFO)" joint should resist a "1.0 Nm/Kg" flexion moment and maintain knee joint stability during stance. For swing phase, the joint should move freely to achieve uninhibited motion (approximately "$300^0/S$" in flexion and "$350^0/S$" in extension).

1.1.1 Design Principles

Table 1 Design criteria

Function	Criteria
User weight range	50-100 kg
Total joint weight	< 560 g
Connection to orthosis	Modular design that can be used with standard orthotic components
Range of motion	0 - 110° (fully extend limb and sit comfortably)
Function during stance	Lock and prevent knee joint flexion at any angle, allow knee

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	joint extension
Function during swing	Free motion (uninhibited knee flexion and extension)
Function during stair walking	Resist knee flexion to lower body centre of mass smoothly. (support user for 0.7 - 1.0 seconds during the stance phase of stair descent)
Function during stand-to-sit	Resist knee flexion to lower body center of mass smoothly
Freely permit maximum flexion angular velocity	> 325˚/s
Freely permit maximum extension angular velocity	> 350˚/s
Valve reaction time	<100 ms (0.1 s) switching between the open and locked settings

1.1.2 Ottawalk-Speed joint

The "Ottawalk Variable Speed(OWVS)" was based on the (OWVS) design by "Lemnaire" at TOHRC . This single-axis joint mounts on the "(KAFO)" lateral side. As the knee flexes, a connecting rod link pushes hydraulic fluid out of the hydraulic cylinder through the spring-biased one-way valve. The fluid streams accomplished the ball valve into an external, expandable reservoir. If fluid velocity-induced drag on the ball is greater than the spring force opposing the drag, the ball blocks fluid flow to the reservoir and locks the joint. Flow can always enter the cylinder because of the spring bias, allowing free extension as shown in figure1.

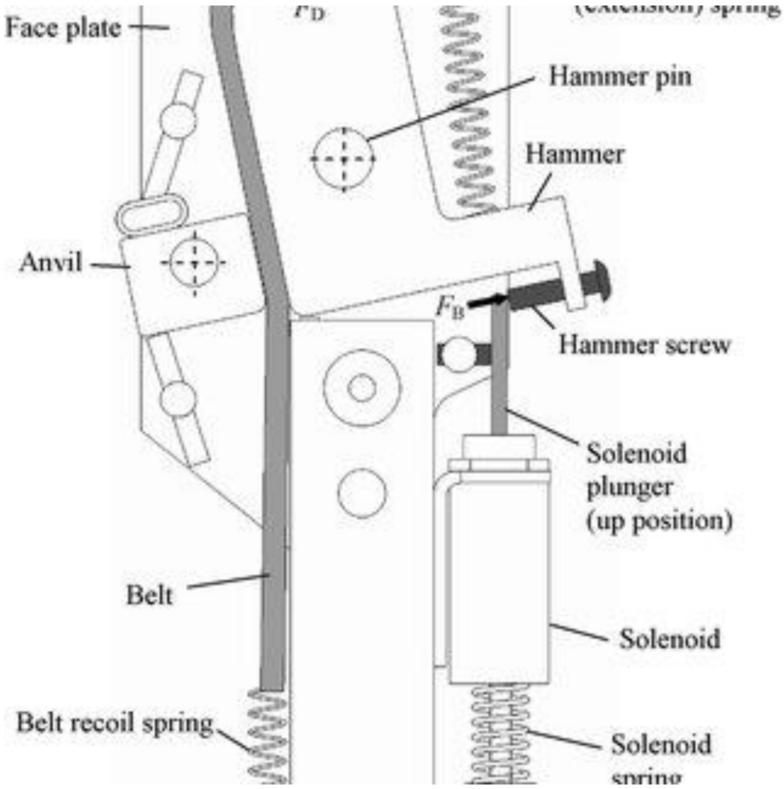

Figure 1 Original OWS SCKAFO joint design

The upright connections and connecting rod link were made from 4140 grade steel and the valve was made of brass. The technically obtainable hydraulic cylinder was used in the design (Inc, 2014).

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The orthotic joint had an average resistance to free-knee motion of "1.06 ± 0.23 Nm", which was considered negligible for human leg movement, and allowed for knee angular velocities over "300^0/S". The theoretical maximum joint supportive moment was calculated as "150 NM", and the joint successfully supported an average load of "144.6 ± 4.8 Nm" over 20 knee collapse trials, with a maximum of "152.1 Nm".

The hydraulic cylinder was changed to the Vektek mini threaded, which has same stroke length but increased capacity (445 kg, 5000 psi maximum). Surgical tubing was used as the fluid reservoir and was positioned round the outside of the piston.

1.1.3 Force analysis

A quasi-static scenario was used for joint force analysis, with the variable resistance valve blocking fluid flow from the cylinder (such as, this represents the largest possible forces because the cylinder has locked the joint from moving) (E. D. Laemaire, 2013). The quasi - static evaluation simulates that the effects of apathy of the links are ignored. The valve is in a closed position blocking fluid flow so the knee joint will remain motionless. This analysis determined forces on the joints and hydraulic cylinder. External flexion moments M_{EXT} were applied to the knee at angles (θ) within the joint's ROM. Only knee flexion/extension was analyzed (Canada, 2014).

The moment M_{EXT} is an external flexion moment acting at the knee via link AB. The knee joint angle (θ) is 0° when fully extended. The link BC angle (φ) is determined from the previous analysis for a given knee angle.

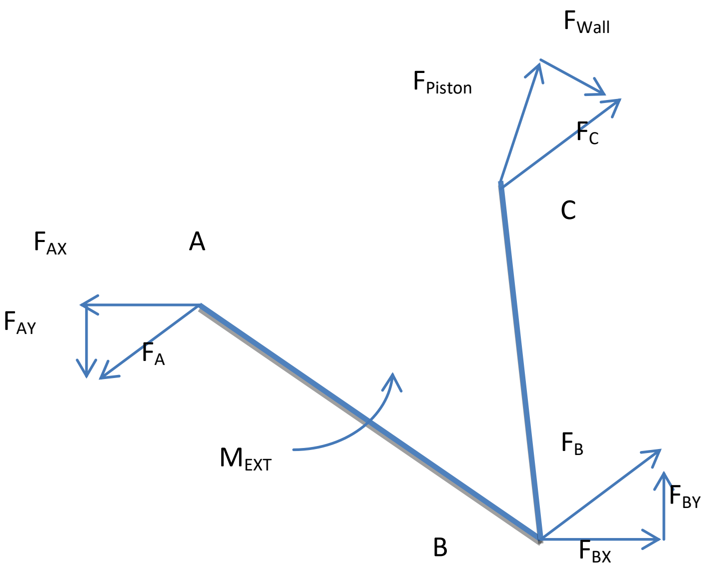

Figure 2 Free body diagrams

The links AB and BC can be separated into two free body diagrams (FBD), where each link is in equilibrium due to the quasi-static analysis. The next Equations are the equilibrium force balances for link AB. The internal forces at B (FB) can be determined by solving these equations. The orientation and length of link AB are known ((i_{AB} = 25.4 mm).

$$\sum F_x = 0 = F_{AX} + F_{BX}$$

$$\sum F_y = 0 = F_{AY} + F_{BY}$$

$$\sum M_A = 0 = M_{EXT} + i_{AB}\, F_B$$

FB is applied perpendicular to link AB in next equation. θ is the knee angle. A 19.27° offset (angle from vertical of the link AB when the knee is fully extended, θ = 0) is added to the knee angle and 90° is added to the equation because FB is perpendicular to link AB.

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$$< F_B = (\theta + 19.27) + 90$$

Since link BC is in equilibrium, the force at joint C (FC) is equal and opposite to the force at joint B (FB). To determine the force that is transferred to the hydraulic cylinder (F_{Piston}) and the upper link (F_{Wall}), FC components need to be rotated +75° shown in next equations (Hiabbeler, 2011).

$$F_{Piston} = F_c \cos(75- < F_B)$$

$$F_{wall} = F_c \sin(75- < F_B)$$

1.1.4 Costs

The variable resistance prototype costs are shown in Table 2. The company or shop where each part was manufactured is also shown. Many of these costs would be decreased if the example were created on a larger balance. The ordered parts (Mc Master and Carr) would decrease in price with larger orders. Cost-effective methods of component manufacturing will be explored if the design proceeds to the manufacturing stage.

Table 2 Cost

Part	Source	Cost
OWS Joint	TOHRC Machine Shop	4200
Maxon Motor/Gearbox/Encoder	McMaster-Carr	231
Blatchford Valve	McMaster-Carr	34
Hydraulic Circuit Block	TOHRC Machine Shop	75
Hydraulic Cylinder Adapter	TOHRC Machine Shop	160
Reservoir Connection	McMaster-Carr	175
Check Valve Seat	TOHRC Machine Shop	85
5 mm Diameter Balls	TOHRC Machine Shop	8.35
Metric Compression Spring	TOHRC Machine Shop	11.25
M5 Machine Screws	Blatchford Model Shop	6.35
M5 Pressure Sealing Washers	McMaster-Carr	14.2
3D Printed Case	McMaster-Carr	123.1

1.2 Biomechanical

Three functional tests were performed: level ground walking, stair descent, and stand-tosit. These tests were performed in the RTL using an 8-camera motion capture system (Vision Inc.) and two force platforms (AMTI) embedded in the floor.

1.2.1 Data collection

Three-dimensional indicator data were collected at 100 Hz, using the 8-camera motions capture system (Vision Inc.). A lower body 6 degree of freedom (DoF) marker set was used, adapted from Collins et al. Thirty reflective markers were placed on the subject's pelvis, thighs, shanks, and feet using two-sided tape. Thigh and shank plates (four marker clusters) were attached to the non-braced side, with the braced side's thigh and shank markers placed directly on the (KAFO) to tracking orthosis movement rather than movement of the limb within the orthosis.

1.3 Orthotic joint control

The OWVS was controlled manually for biomechanical testing. The control software connected to the valve using Bluetooth for remote control by computer or tablet. For the walk test, manual switching between setting 0 (closed) and setting 1 (open) was performed. The operator closely monitored the participant and selected setting 0 from braced limb foot contact to foot off. Just prior to foot off, at the beginning of pre-swing, setting 1 (open) was selected to allow free knee flexion.

The stair descent and sitting tests did not require setting switching. Prior to each stair descent or stand-to-sit sequence, the valve was set to the appropriate resistance setting.

1.3.1 Data analysis

Indicator data were labeled and reconstructed using "Vision Nexus" and analyzed using "Visual 3D".

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Indicator and ground effect force data were categorized with a 5th order low-pass Butterworth filter at 10 Hz and 20 Hz cut-off rates, respectively. A seven section lower body model (2 feet, 2 shanks, 2 thighs, pelvis) was created in "Visual 3D". Kinematic and kinetic statistics from the braced limb were normalized to 100% of the gait cycle (heel attack to heel strike of the braced limb) for the stair, walk, and stair descent tests. For the stand-to-sit tests, braced limb kinematic and kinetic data were normalized to 100% of the stand-to-sit cycle.

2. Results and discussion

2.1 Mechanical Plan

The results for a 60˚ joint flexion range (30˚ to 90˚). An issue that was noticed during testing was that the hydraulic knee joint did not completely resist knee joint flexion at setting 0. This setting was designed to completely resist knee joint flexion, supporting the individual while the braced limb is in contact with the ground. Fluid was not leaking from the circuit (hydraulic cylinder, variable resistance valve hole, reservoir, fill points), so the fluid was likely bypassing either the variable resistance valve or the check valve and collecting in the reservoir. A mechanical test was performed to further investigate where the leak was occurring, why the leak occurred, and methods to correct it. Even though setting 0 did not completely resist knee flexion, flexion resistance was high. At a 200N load (40-75 Nm), the time to flex 60˚ was 18.83 ± 0.82 s (3.2 ˚/s). This resistance should provide safe knee flexion control during stance phase; however, small knee joint flexion during stance may adversely affect gait confidence for people with knee extension weakness as shown in table 3.

Table 3 Setting

Setting	Time for joint to flex 60˚ (30˚ - 90˚) [seconds]			
	51 N	111 N	168 N	200 N
0	141.83 (19.65)	49.25 (1.25)	23.21 (1.52)	17.53 (0.75)
1	0.31 (0.00)	0.21 (0.02)	0.23 (0.01)	0.10 (0.10)
2	0.78 (0.04)	0.25 (0.01)	0.23 (0.01)	0.23 (0.10)
3	0.25 (0.02)	0.26 (0.04)	0.19 (0.02)	0.24 (0.02)
4	0.66 (0.01)	0.39 (0.01)	0.29 (0.01)	0.21 (0.02)
5	0.21 (0.03)	0.31 (0.00)	0.31 (0.06)	0.25 (0.10)
6	1.98 (0.14)	0.55 (0.03)	0.41 (0.01)	0.32 (0.02)
7	0.21 (0.01)	0.64 (0.01)	0.21 (0.02)	0.25 (0.03)
8	1.50 (0.01)	0.48 (0.05)	0.36 (0.14)	0.59 (0.15)
9	0.21 (0.03)	0.25 (0.02)	0.35 (0.04)	0.19 (0.01)
10	5.26 (0.02)	1.36 (0.11)	0.61 (0.02)	0.45 (0.02)

The moments applied to the joint during the drop tests increased as the joint flexion angle increased, due to the increasing moment arm length. For the four weights tested, the moments applied to the joint between 30˚ and 90˚ are shown in Figure 4.6. The 200 N weight basket applies the largest moments of 40 - 75 Nm. The 51 N weight basket only applies moments of 9.5 Nm – 19 Nm over this range.

Setting 1 is the valve's open setting. It is designed for free knee flexion, and has the largest hole diameter of all the settings (2.25 mm). Valve settings 2, 4, 6, 8, and 10 were designed to expose a single hole to the channel leading to the variable resistance valve (Table 3.6). The hole diameters are: 0.50 mm (setting

2), 0.35 mm (setting 4), 0.30 (setting 6), 0.25 mm (setting 8), and 0.20 mm (setting 10). Settings 3, 5, 7, and 9 expose two holes to the channel leading to the variable resistance valve, increasing flow though the valve and thereby decreasing the joint flexion resistance. The hole diameters are: 0.50 mm and 0.35 mm (settings 3), 0.35 mm and 0.30 mm (setting 5), 0.30 mm and 0.25 mm (setting 7), and 0.25 mm and 0.20 mm (setting 9). The joint flexion time results for the valve settings 1-10 over a 60˚ range (30˚ - 90˚), for the 168 N and 200 N drop tests.

As expected, the single hole settings (2, 4, 6, 8, and 10) provided more resistance to flexion than the double hole settings, particularly settings 6, 8 and 10. For the 200 N drop test, 60˚ of joint flexion at settings 8 and 10 required 0.59 ± 0.02 s and 0.53 ± 0.02 s, respectively. Surprisingly, setting 10 provided less resistance than setting 8 for the 200 N drop test. Setting 10 provided the most resistance of all the settings for 51 N, 111 N and 168 N tests. This inconsistency may be because the valve was not able to fully resist motion at the locked setting. At larger weights, more pressure is applied to the variable resistance valve. If a leak occurs at the variable resistance valve, setting 10 may be more susceptible to fluid escaping at higher pressures.

Overall, the resistances were lower than predicted. From the hole selection experiment, a 0.25 mm hole (equivalent to setting 8) required 1.1 ± 0.02 s to flex the joint 50˚ (i.e., approximately twice as long as the time from the resistance setting test). The lower resistance results are likely due to hydraulic fluid leaking past the variable resistance and/or check valves and collecting in the reservoir.

The joint's resistance settings were designed to support individuals for 0.7 - 1.0 seconds during the stance phase of stair descent (approximately 65˚ of knee flexion) on the braced limb to safely lower the body's center of mass to the next step. The settings provided less resistance than expected, but the drop test experiments do not replicate knee motion or loading during stair or stand-to-sit motions. Settings 6, 8 and 10 did resist joint flexion and should be tested functionally to determine performance. These tests will provide a better understanding of whether the resistance settings are sufficient for users, even though resistance moments were lower than expected. From the results, settings 8 and 10 will likely be used as resistance settings for stair descent, and setting 6 may be appropriate for lighter individuals.

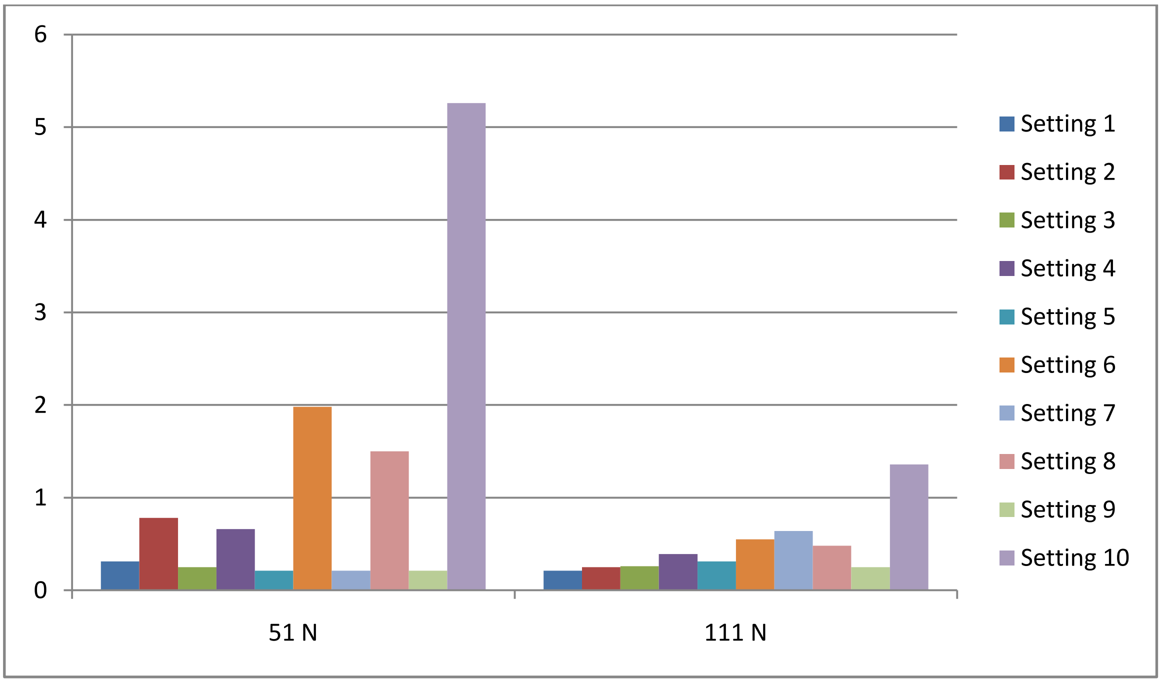

Figure 3 Time for the joint (51 N, 111 N)

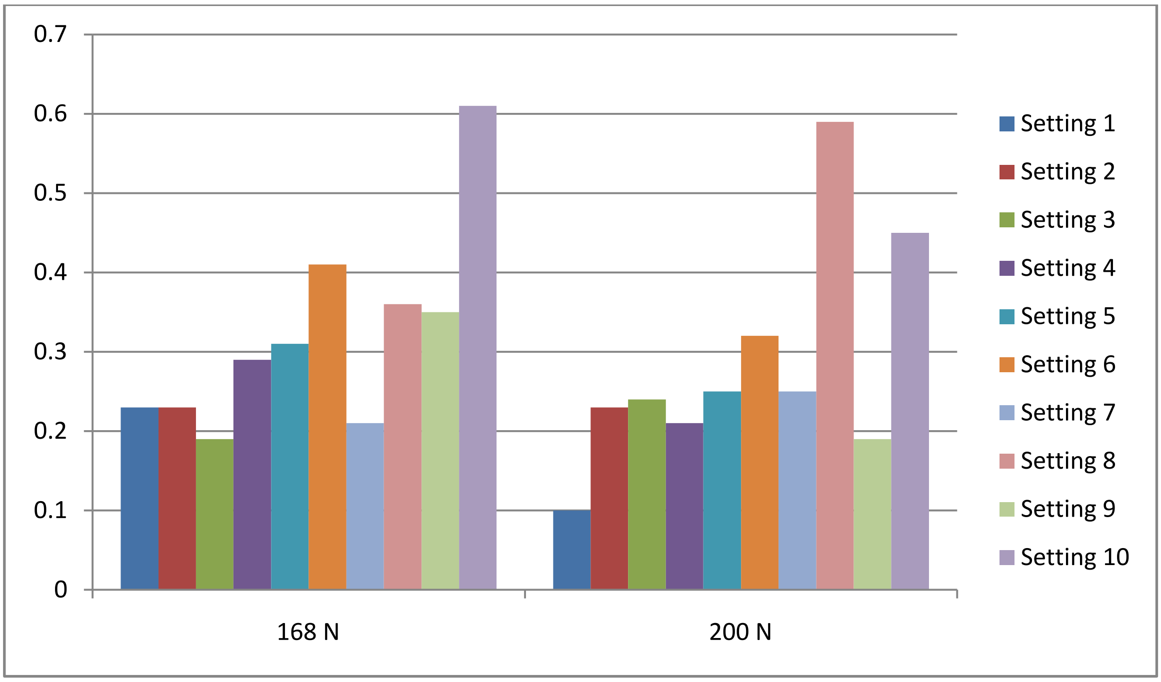

Figure 4 Time for the joint (168 N, 200 N)

The double hole settings (3, 5, 7, and 9) provided little flexion resistance. The 60˚ flexion times were similar to the open setting and did not provide enough flexion resistance for any of the target activities. This is not surprising since these settings use an arrangement of two holes. Having these settings allowed for a larger range of test results from different hole sizes but future OWVS designs may not need to include these double hole settings. Future hole pattern design should be modified to include a range of more appropriate resistance settings.

2.2 Biomechanical

2.2.1 Walking test

the average stride parameters for the three walking tests. The participant was instructed to walk at a comfortable slow pace for the stance-control test, to allow for more accurate switching of settings. The participant had a reduced stride velocity (38.3% slower than walking with the joint open or locked), and a reduced stride length (15.0% shorter than walking with the joint open and 16.8% shorter than walking with a locked knee joint). The stride time increased (43.4% longer than walking with an open joint and 40.0% longer than walking with a locked knee joint). There was no difference in the change in stance (% of gait) between the stance-control mode and the open mode. When walking with a locked knee joint, the participant had a decreased stance/swing ratio (4% less stance than the stance-control and open modes) as shown in table 4.

Table 4 Average stride parameters

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Joint Setting	Stride time (s)	Stride Length (m)	Stride Velocity (m/s)	Stance (% of stride)
Stance-control	1.64 (0.33)	1.52 (0.11)	0.62 (0.16)	62
Open	1.14 (0.04)	1.04 (0.02)	1.32 (0.04)	62
Locked	1.16 (0.02)	1.34 (0.03)	1.24 (0.04)	57

The orthosis had similar ranges of motion for stance-control and open settings walking, with approximately 5% greater range of motion (ROM) while walking with the joint open. The ROM for the locked joint test was 11.65 ± 1.36 ˚, with a large portion of the motion occurring during weight acceptance (8.33 ± 1.27 ˚). LA1 (minimum angle during the weight acceptance task) were similar for stance-control and open tests, and had a more extended knee during the locked testing.

During stance-control walking, the orthosis locked knee flexion during single leg support (i.e., minimal change between angles LA2 and LA3). This is similar to the results reported by Irby et al. for individuals walking The Dynamic Knee Brace Scheme (developed into the Otto Bock Sensor Walk). The Dynamic Knee Brace Scheme used foot shifts to lock the knee through stance and permissible free knee motion during swipe. Differences in angle profiles between the Dynamic Knee Brace Scheme and the OWVS occurred during weight acceptance and when the knee was released for knee joint flexion. The Dynamic Knee Brace Scheme joint angle during the weight acceptance task remained constant, while the OWVS joint angle increased slightly. This indicates that the brace was not completely resisting joint flexion during weight acceptance. The OWVS also released the knee joint for flexion earlier (to prepare for swing) than the Dynamic Knee Brace Scheme did. This difference was due to the different methods used to control the joints. The Dynamic Knee Brace Scheme uses a footswitch to release the knee after toe off, while the OWVS was manually controlled, and the knee was released just prior to toe off.

2.2.2 Stair descent test

the braced limb average stride parameters for stair descent (joint settings 1 (open), 8, 10). The participant walked at a similar pace for all stair descent trials. Different settings had similar swing/stance ratios as shown in table 5.

Table 5 Average Stair descent parameters

Setting	Stride time (s)	Stance time (s)	Swing time(s)	Stance (% of stride)
1	1.09 (0.08)	0.64 (0.01)	0.35 (0.01)	60
8	1.16 (0.21)	0.66 (0.02)	0.39 (0.02)	59
10	1.12 (0.01)	0.55 (0.02)	0.35 (0.01)	61

During stair descent, setting 1 (open) had a larger maximum knee flexion angle during weight acceptance compared to the other resistance settings. The knee also remained at the constant angle prior to lower the body's center of mass during single leg support at setting 1, while the angle constantly increased for the resistance settings.

The joint's ROM decreased with increasing resistance settings. Knee ROM during stair descent is approximately 78 – 90˚ for able-bodied individuals [19]–[21]. The resistance settings ROM were lower than this range (75.2 ± 4.0˚ for setting 8, and 67.7 ± 3.6˚ for setting 10). This suggests that the orthotic joint was resisting knee joint flexion, but it is impossible to eliminate the possibility that the participant was not letting the joint lower their center of area completely to the following step. Peak flexion angular velocities also decreased for the resistance settings, compared to the open joint, indicating that the joint resisted flexion.

2.2.3 Stand-to-sit test

the average sagittal knee angle and angular velocity during the stand-to-sits tests using setting 10 and setting 1 (open) of the OWVS. The average sagittal knee angle and angular velocity profiles were similar when using the open and resistance settings as shown in table 6.

Table 6 Average Stand parameters

	Setting 1	Setting 8	Setting 10
Sit time [s]	1.21 (0.11)	1.32 (0.10)	1.24 (0.02)
Min. Angle [˚]	**17.6 (1.5)**	**13.1 (2.1)**	**11.1 (1.1)**
Max. Angle [˚]	105.3 (0.7)	102.6 (0.1)	102.1 (1.2)
ROM [˚]	89.2 (1.1)	85.5 (1.1)	82.7 (1.0)
Max. Velocity [˚/s]	117.0 (7.3)	119.0 (12.3)	108.1 (9.2)
Max. Moment [Nm/kg]	1.01 (0.01)	0.57 (0.06)	0.77 (0.01)
Min. Power [W/kg]	- 1.11 (0.11)	-1.25 (0.12)	-1.18 (0.11)

The time to perform a stand-to-sit motion was the longest while using setting 10, and fastest while using the open setting (setting 1). The ranges of motion were similar for the standto-sit trials at all joint settings tested. The maximum knee moments were similar for all settings, and only slightly larger for setting 1 "1.06 ± 0.23 Nm/kg".

As expected, stand-to-sit descent times were greatest for setting 10 and shortest for setting 1 (open). The resistance settings resisted joint flexion, lowering the body's center of mass smoothly to the seat. The angle profiles were similar between the open and resistance settings, implying that the joint's flexion resistance did not change the participant's stand-to-sit movement pattern, but merely lengthened the time required to descend.

3. Conclusions

The mechanical examination revealed that the design always allows for the extension of the free knee link, allows the unauthorized knee flexion in the opening position, and that the variable-resistance gearbox / gearbox is sufficient for the design. Unfortunately, OWVS has not completely resisted the flexion of the joint, in closed mode during the mechanical test since the fluid leaks around the top of the valve variable valve. This leakage is allowed to bend the joint in the closed position, and it is likely to reduce the flex resistance in the settings 2-10. Although OWVS has provided large joint flexion resistance in the closed frame (setup 0) and in the higher resistance settings (settings 6, 8 and 10), the fluid leakage problem should be addressed in future designs as the joint must be fully capable of resisting the flexion of joints (Hiabbeler, 2011).

The biomechanics examination showed, in general, that the OWVS design bodes well and supports the participation of 80.74 kg. The knee joint was resisted during the knee position and the knee joint was released without a knee-joint restriction. OWVS provided bending resistance during down stairs and stand-up movements. There was concern observed during walking tests that the joint allowed some flexion of the knee during weight acceptance. Some mechanical are found between 0-15 of bending, allowing the joint to move freely between these angles.

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