Functional Outcomes of a Custom, Energy Harvesting "Bullfrog" AFO

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Organization
The study was conducted at Georgia Institute of Technology
within the School of Applied Physiology Clinical Biomechanics Laboratory
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Abstract

Purpose: The purpose of the study was to develop and test a prototype custom orthosis that passively harvests energy and provides an assist to plantar flexion during walking.

Scope: Few orthoses have been designed to provide plantar flexion assistance for walking, yet the greatest power output of the ankle occurs during push-off at terminal stance. Since harvesting power would be dependent on the amount of passive resistance that could be tolerated without perturbing gait, we instrumented an AFO with force and motion sensors that could quantify the dorsiflexion resistive force threshold during walking. Data on the influence of dorsiflexion resistance during walking has important clinical implications in the functional performance of orthoses and the human response to the orthotic constraint of joint movements.

Methods: A conventional metal AFO, attached to a running shoe, was instrumented with: a load cell to measure the dorsiflexion resist force, potentiometers to measure knee and ankle joint angles and force resistance sensors to determine foot orientation to the ground. One healthy subject walked at their self-selected speed on a treadmill in four test conditions that consisted of a control (no resist) and two dorsiflexion resist force conditions of 25.6 and 38.9 N respectively. In addition to these fundamental tests a special gear mechanism was designed to harvest energy during dorsiflexion and plantarflexion motion via a ratchet and catch release system.

Results: The instrumented AFO effectively accomplished its objective of measuring force and joint angles. The pilot data clearly demonstrated that orthotic constraint of the ankle with a resist/assist force will result in compensatory leg movements. Some of the adopted movements that were observed during tests in the pilot study were primarily related to an inherent plantar flexion assist force, which was a byproduct of the dorsiflexion resist. A small change to the instrumented AFO that isolates the dorsiflexion resists force application from midstance to terminal stance with minimal constraint to ankle motion during swing will allow us to refine the measurements for the ongoing studies to determine the dorsiflexion resist perturbation threshold values. Initial tests of our gear mechanism with a 3 dimensional computer-aided-design (CAD) software program (SolidWorks, Concord MA) demonstrate that energy can be harvested from bi-directional movements of the ankle and foot.

Key Words: ankle foot orthoses, ankle joint constraint, energy harvesting, gait perturbations

Introduction

The development and testing of an orthosis that passively harvests energy and promotes plantar flexion during walking was the primary goal of this research project. The underlying concept of the design is based off of the biomechanical advantages present in the ankle of many high performance jumping animals, the mechanism of which has been described in scientific study of the bullfrog, thus our device has been termed the “Bullfrog AFO”. Our initial prototype (i.e., v01) provided a plantar flexion assist during walking but was limited in its range from10 degrees of dorsiflexion to the neutral ankle position of 90 degrees. Whereas the plantar flexion from the neutral 90 degree position to 25 degrees of plantar flexion did not have any plantar flexion assist. Our research team determined that to overcome the limited range of the plantar flexion assist with the AFO (v01), a special gear mechanism would have to be designed. During the process of revaluating the AFO (v01) we also felt that some fundamental biomechanical measures related to calculating the maximum power output of the device would be important in establishing the target design parameters of the redesigned AFO.

Since we were considering the design of a new ankle joint mechanism for the prototype orthosis, we gave thought to some additional features to include with the orthosis. Originally we were interested in just harvesting power from stance phase dorsiflexion. However the range of dorsiflexion during the stance phase of gait is limited to only about 10 degrees. If we could also harvest plantar flexion motion, that would almost double the potential power return produced. So our research team decided to design a mechanism that could harvest the motion of both dorsiflexion and plantar flexion motion during stance with the capability of returning the power to assist late stance push-off. We speculated that our orthosis would have a greater magnitude of power assistance with the addition of harvesting plantar flexion. A significant effort was invested in the design of the gear mechanism to accomplish this objective.

A substantial effort was devoted to the design of an orthosis to quantify the passive dorsiflexion resist threshold during walking. The instrumented AFO incorporated a load cell to measure the dorsiflexion resist force threshold and potentiometers to measure joint angles at the knee and ankle. Force resistance sensors at the heel and forefoot were used to determine the location of the foot with respect to the ground. Our pilot tests with the instrumented AFO were effective at collecting load and angular measures during treadmill walking. Several minor modifications of the orthosis will be made prior to conducting the full study on a subject population. Although the determination of the resistive dorsiflexion threshold is the underlying purpose of the study, the human response to joint motion constraint is perhaps of equal or greater importance in understanding the benefits or limitations of orthotic motion control devices. To our knowledge there are no studies that have looked at dorsiflexion resistive constraint and its effect on walking.

Orthotic Dorsiflexion Resist Perturbation Thresholds During Treadmill Walking

During pilot testing with our first design iteration of the Energy Harvesting AFO, subjects reported that when walking with the AFO the resistance during dorsiflexion, which loaded the spring for the plantar flexion assist power, altered their gait. During these early trials we were required to adjust the magnitude of the dorsiflexion resist, via subject feedback, as to the ideal amount of resistance that they could accommodate without disrupting their walking. From these observations it became clear that the threshold of dorsiflexion resistance would be a critical design parameter for any Energy Harvesting AFO utilizing ankle motion to harvest power. Furthermore, knowledge of the dorsiflexion resist perturbation threshold during walking would also be useful clinically since orthotists often fit AFO’s that inherently resist dorsiflexion as a by-product of their dorsiflexion assist function (e.g., posterior leaf spring AFO). A literature search on the topic did not reveal any information on the resistive dorsiflexion thresholds for orthoses and their influence on gait.

We decided that the resistive dorsiflexion perturbation threshold was critical to the design of our Energy Harvesting AFO and redirected our efforts to determining those values. Furthermore we hypothesized that the threshold of dorsiflexion resistance perturbation would vary with walking speed. Slower walking would have a lower dorsiflexion resistive threshold compared to a faster walking speed due to the relative effects of inertia. To quantify the dorsiflexion resistance perturbation threshold during walking, we designed an experimental ankle foot orthosis that could measure the resistive dorsiflexion force (N) of the spring, the mechanical ankle joint angle of the orthosis and the anatomical knee joint angle during walking. We evaluated one subject walking on a treadmill with the instrumented AFO. The study is continuing and we estimate that we will need approximately 15 subjects to determine the dorsiflexion perturbation threshold.

Methods

Instrumented AFO to Measure Force and Joint Angles

A conventional metal AFO was used as the platform to instrument with sensors to measure load, joint angles and regional foot contact. The AFO structure was construction of: medial and lateral uprights (i.e., 3/16” x 5/8” *aluminum*); calf band (1/8” x 2” *aluminum*), free motion ankle joints and strirrup (*stainless steel*) (Becker Orthopedic, Troy, MI). Lightweight aluminum pieces affixed with BNC connectors for the measurement signals and our power source connection were attached to the lateral uprights and stirrup. A piezoelectric load cell (Kistler Instrument Corp., Winterthur, Switzerland) was used to collect the dorsiflexion resistance spring force measurements. A series of elastic cords (i.e., bungee cords) approximately 4mm in diameter were used to apply the respective orthotic resist/assist constraint force. The compression/tension load cell was affixed distally to a posterior U-shaped heel component and proximally to the calf band. Joint angles were measured with rotary potentiometers (6639 Series, Bourns Inc. Riverside CA). The ankle potentiometer was mounted to the lateral

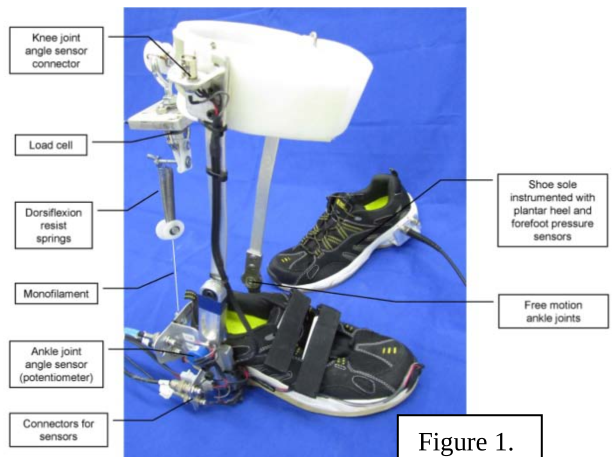
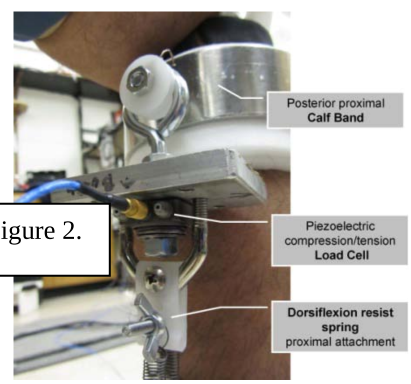


Figure 1.



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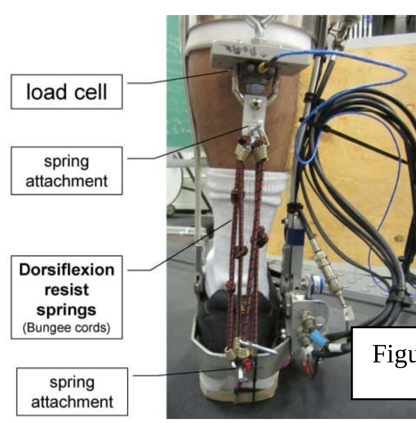
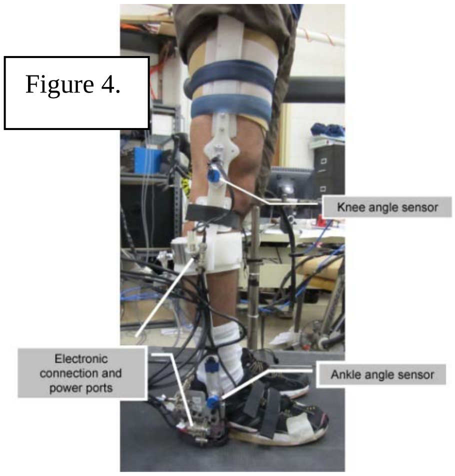


Figure 3.

upright and stirrup in line with the mechanical orthotic ankle joint axis of the AFO. For the knee joint angle measurement the potentiometer was mounted to a custom made polypropylene overlap single axis knee joint. The instrumented knee joint was attached to a polypropylene uprights proximally with a polyethylene foam thigh cuff and a distally attached to the AFO on the lateral upright and secured with a Velcro strap. Force sensing resistors (FSR 402, Interlink Electronics, Camarillo, CA.) situated on the plantar surface of the sole at the heel and the forefoot of both shoes were used to determine the timing of the gait cycle.

Figure 4.



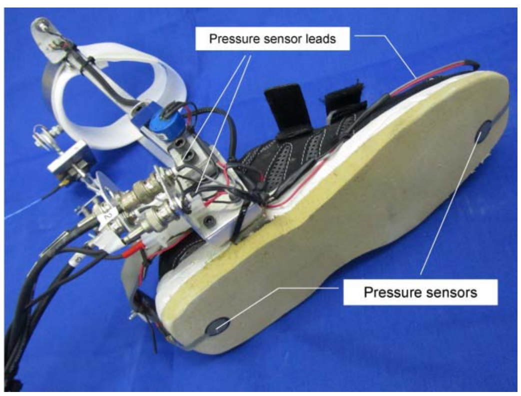


Figure 5.

Test Protocol

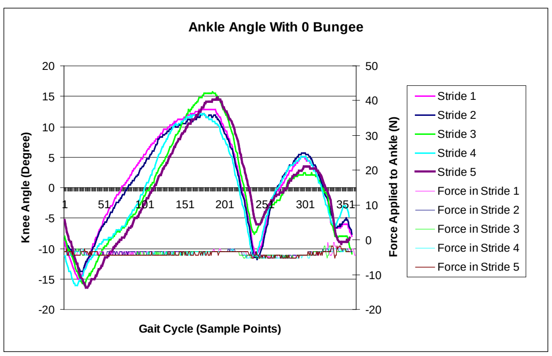
We proposed three subjects for the pilot data (n = 3), however at the time of writing this report, data from only one subject had been collected. A male subject, (age 23y; height 172cm; weight 59kg) was recruited for the study. The test orthosis was custom fitted to the subject, which equated to the minor addition (i.e., 5mm) of some padding attached to the calf band of the orthosis to reduce the circumferential and medial-lateral (M-L) dimension to that of the subject's proximal calf. The proximal suspension straps of the knee joint angle sensor upright had to be modified also to accommodate the subject's proximal and distal thigh girth dimensions. After achieving a satisfactory fit of the test apparatus, the subject walked on a treadmill for approximately eight minutes to determine his self-selected walking speed. During the period of adaptation to the orthosis and walking speed determination, the ankle joint was permitted to articulate freely with no orthotic constraint (i.e., resistive force).

The study consisted of three test conditions using the instrumented AFO: 1) free ankle/no orthotic ankle resist/assist (control) 2) dorsiflexion resist/plantar flexion assist (3 elastic cords) 3) dorsiflexion resist/plantar flexion assist (5 elastic cords). Data collection was conducted for ten seconds once the subject reached his self-selected speed of 1.56 m/s with each of the three conditions. Load (N) from the orthotic resist/assist constraints (i.e., elastic cords), knee and ankle angles were collected via tethered electrical leads to a data acquisition box (DAQ National Instruments, Austin, TX) connected to a computer. Data was processed using MatLab (MathWorks, Natick, MA). Data analysis consisted of evaluating the middle five steps from the 10 seconds of data collection after the subject reached his self-selected walking speed.

Results and Discussion

Knowledge of how the neuromotor control system responds to orthotic constraint of movement is fundamental to predicting therapeutic outcomes for the use of AFO's as well as exploring new treatment regimes. Therefore, the pilot study reported here and our subsequent studies should prove to be invaluable in understanding how current AFO systems influence walking. Since clinicians do not typically take dynamic kinematic measures to evaluate the performance of the AFO's that they prescribe and fit, they have no point of reference to refine an AFO's functional parameters (i.e., magnitude of the orthotic resist/assist). We hope that data such as this will provide some perspective on functional affect orthoses have when an orthotic constrains ankle joint movements.

One of the primary goals of this aspect of the research was to evaluate the influence of different orthotic dorsiflexion resist force magnitudes have on walking. Since our energy harvesting AFO will store power from the passive movements of stance phase walking, the threshold in which the orthotic dorsiflexion resist force altered a subject's normal walking mechanics would have to be quantified. So of particular interest was the effect an increasing dorsiflexion resist force would have on the peak ankle dorsiflexion angle during walking. This value would inherently reflect changes in step length that would be perturbed by a resistive dorsiflexion constraint. We speculated that the peak ankle dorsiflexion angle during mid to late stance would decrease with increased orthotic dorsiflexion resistance. Our pilot data supports that supposition. The graphs in figures 6, 7 and 8 clearly show a decrease in ankle dorsiflexion with increasing dorsiflexion resistance (Figures 6-8). Even though our data clearly shows that when our subject walked at his self-selected speed with dorsiflexion resistance his peak ankle dorsiflexion angle after midstance decreased, he reported no perceptible alterations between the different conditions with regard to dorsiflexion. He did note changes with the five bungee resist but could not detect exactly what alterations he was assuming with his gait. It appears that a gait perturbation threshold may exist with an orthotic dorsiflexion resist and that there is a linear relationship between the resistive force and maximum the maximum dorsiflexion angle.



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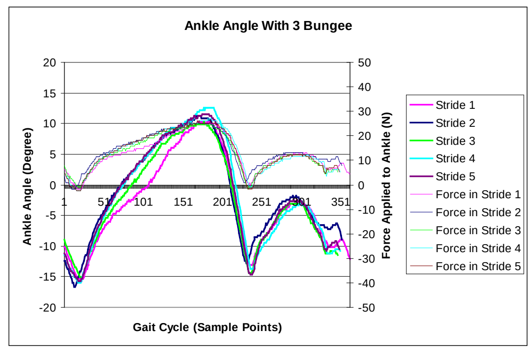

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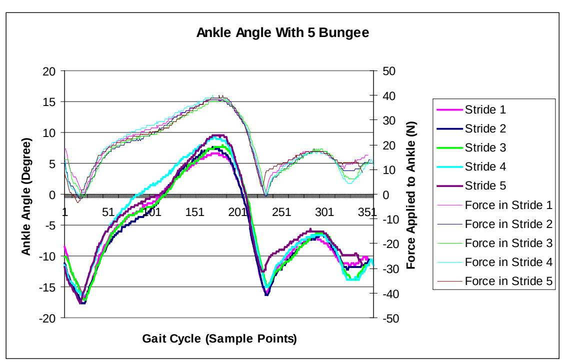

Figure 8.

Table 1 shows the mean values for both force (N) and joint angles collected from five steps. The maximum mean peak orthotic dorsiflexion resist force with three bungee’s was 25.2 N and with five bungee’s was 38.3 N. These values clearly show the linear relationship of the orthotic dorsiflexion resist conditions with increasing the number of bungees from three to five. The graph in Figures 6-8 shows how consistent the measurements were with multiple steps.

0 Bungee							
Stride #	Max PF Assist Force (N)	Max DF Resist Force (N) in Stance	Max Knee Flexion Angle° in Stance	Max Knee Flexion Angle° in Swing	Max Ankle PF Angle° in Loading Response	Max Ankle DF Angle° in Stance	Max Ankle PF Angle° in Toe Off
1	-5.2	-0.9	16.8	74.2	-15.1	12.8	-10.9
2	-5.2	-2.6	15.2	74.6	-13.8	12.2	-11.8
3	-5.2	-1.7	10.0	75.5	-16.0	15.8	-7.6
4	-5.2	-2.6	11.6	74.6	-16.0	12.2	-11.2
5	-5.2	-1.7	6.1	74.6	-16.4	14.8	-6.0
Mean	**-5.2**	**-1.9**	**11.9**	**74.7**	**-15.5**	**13.6**	**-9.5**

3 Bungee							
Stride #	Max PF Assist Force (N)	Max DF Resist Force (N) in Stance	Max Knee Flexion Angle° in Stance	Max Knee Flexion Angle° in Swing	Max Ankle PF Angle° in Loading Response	Max Ankle DF Angle° in Stance	Max Ankle PF Angle° in Toe Off
1	-1.7	25.2	20.7	78.8	-15.7	10.2	-13.1
2	-1.7	25.2	20.4	76.8	-16.7	11.2	-12.8
3	-1.7	24.3	20.7	77.8	-15.4	10.2	-14.4
4	-2.6	26.9	23.6	78.5	-16.0	12.5	-13.8
5	-2.6	26.1	24.6	79.4	-15.7	11.5	-14.7
Mean	**-2.1**	**25.6**	**22.0**	**78.3**	**-15.9**	**11.2**	**-13.8**

5 Bungee							
Stride #	Max PF Assist Force (N)	Max DF Resist Force (N) in Stance	Max Knee Flexion Angle° in Stance	Max Knee Flexion Angle° in Swing	Max Ankle PF Angle° in Loading Response	Max Ankle DF Angle° in Stance	Max Ankle PF Angle° in Toe Off
1	0.0	38.2	27.2	83.7	-16.7	6.7	-16.0
2	-1.7	38.2	24.9	82.4	-17.7	7.6	-16.4
3	-0.9	38.2	27.2	82.7	-17.0	8.0	-15.1
4	-0.9	40.0	30.4	83.3	-16.4	9.3	-15.1
5	-3.5	40.0	31.7	83.7	-17.7	9.6	-12.5
Mean	**-1.4**	**38.9**	**28.3**	**83.1**	**-17.1**	**8.2**	**-15.0**

Table 1.

The results from this pilot study (n=1) showed that an orthotic resist/assist force in an ankle foot orthosis perturbed walking compared the non-orthotic resist/assist condition (i.e, control). This finding is noteworthy because the data suggests that even a relatively nominal orthotic resist/assist force alters gait mechanics and that there is a linear relationship between the magnitude of the orthotic force constraint and the ankle and knee joint kinematics. With no orthotic resist/assist force applied to the ankle, the maximum mean stance phase dorsiflexion angle was 13.6 degrees (Table 1). However when an orthotic dorsiflexion resist force was applied at the ankle of 25.9 N and 38.9 N, the ankle dorsiflexion angle during stance phase decreased in magnitude to 11.2 and 8.2 degrees respectively. Both of the orthotic force resist values perturbed gait, so from this one subject trial we were not able to determine the threshold of orthotic restraint that altered gait but it is obvious it is a lower force value.

We established the dorsiflexion resist force used for the pilot test through trial and error and subject feedback. With our subject blinded as to how many bungees were attached to the AFO, we asked him to indicate what orthotic dorsiflexion resist set-up he perceive affected his walking. In other words, at what point did he detect that his walking pattern was altered by the orthotic ankle dorsiflexion resist or plantarflexion assist configuration. Surprisingly, our subject did not perceive any change walking on a treadmill at his self-selected speed until three bungees were added to resist dorsiflexion, which equated to a mean maximum force of 25.9 N. We speculate that the reason our subject did not report any changes in his walking with an orthotic ankle resistive constraint is because he adopted a compensatory gait pattern. Essentially when the subjects ankle motion was constrained with the orthotic dorsiflexion resist, his stance phase dorsiflexion angle decreased to counteract the force. So a subject's "perceived gait perturbation" may not be a very reliable way to determine if ones walking pattern is altered when ankle joint motions are constrained with an orthosis. Such a consideration has clinical implications because orthotists and prosthetists often use patient feedback and perceptions as a qualitative measure to evaluate the performance of an orthosis or prosthesis. Therefore, a patient's perceived gait perturbation response to orthotic constraint of ankle joint motion may be of limited value for assessing actual changes in walking patterns. The length of time that the orthotic joint constraint is applied may yield different results of "perceived gait perturbations" from subjects.

Compensatory gait movements were also observed during swing phase with the orthotic joint constraint conditions. Because the orthotic dorsiflexion resist on our instrumented AFO applies a resistive force in the entire range of dorsiflexion ankle motion, a plantarflexion assist force was also present. The plantar flexion assist appeared to be of a magnitude that would not permit adequate dorsiflexion during swing and thus creating an orthotic induced pseudo "drop foot".

This observation is supported by the data which showed increased knee flexion during swing phase with increased dorsiflexion resistance (Table 1). Ideally, one would want the orthotic dorsiflexion resist force to be applied from the neutral (i.e., 90º) ankle position to the end range of dorsiflexion so that swing phase would be perturbed minimally. Hence we modified our test orthosis since conducting the pilot study. To minimize perturbing swing phase for subsequent tests we will be able to isolate the orthotic dorsiflexion resist force application from just midstance to terminal stance by a small change to the orthotic control mechanism. This change will also allow us to learn what the gait patterns are adopted with the orthotic constraint of ankle dorsiflexion from mid to late stance. Thus we will have an improved understanding of the body's sensory motor response to ankle joint constraint with an AFO.

We anticipate that our upcoming "full" study will determine the plantar flexion resist perturbation threshold values we need to set the design parameters for our energy harvesting "Bullfrog" AFO. Since we plan on storing our energy by loading a mechanical spring or deflecting a viscoelastic material, the force value from these studies will provide us with the size and capacity of the energy storage spring unit we will eventually be using. However this lengthy process of additional tests will also yield important clinical information on the neuromotor control systems response to orthotic joint constraint and assist controls.

Ongoing Research

Even though the funding cycle is over for the grant this work is continuing. The full study on the orthotic pertubation threshold will begin within a few weeks. Several modifications and refinements to the AFO measurement device are currently being completed for those studies. Two more pilot tests will be conducted using the same test protocol reported. Once we have collected data on two more subjects will perform a power analysis to determine how many subjects we will need to conduct the full orthotic perturbation threshold study.

IRB INFORMATION

The IRB number and certification associated with this project is: H09036

FORTHCOMING SECTIONS OF THE REPORT WILL BE SUBMITTED ONCE THE PRINCIPAL INVESTIGATOR HAS RECEIVED THE NON-DISCLOSURE AGREEMENT

Proposed additions to the final report pending NDA.

Development of a Mechanism that Harvests Power from Dorsiflexion and Plantar Flexion Ankle Motion

Limitations of the Studies

Clinical Relevance of the Studies

Potential Impact of the Studies

Dissemination of the Information

Acknowledgements

APPENDIX