THE EFFECT OF DYNAMIC ANKLE-FOOT ORTHOSES ON ANTHROPOMETRICS AND BIOMECHANICS AS AN INTERVENTION OVER TIME IN MILITARY PERSONNEL WITH LOWER EXTREMITY NEUROPOTHY

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Abstract

Musculoskeletal trauma to the extremities accounts for over 50% of injuries sustained in recent military conflicts and may result in partial to full paralysis of the foot and ankle. A dynamic ankle-foot orthotic (DAFO) has been successful in returning service members to military duty. Six service members (29.3±7.2 years) with drop foot were fitted with a DAFO; following six months of use, ankle and knee strength and walking gait were compared to a group of healthy matched controls. Ankle and knee joint power and vertical ground reaction force while walking without the DAFO increased in concert with knee and ankle muscular strength. Ankle propulsive power in injured service members at study onset was significantly lower than control subjects, but improved to a level not significantly different than controls. Improvements were likely attributable to increased knee extension strength allowing greater knee flexion and improved confidence during loading with increased walking velocity.

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Introduction

Musculoskeletal injuries to the extremities account for 49.4%-54% of injuries sustained in Operation Iraqi Freedom and Operation Enduring Freedom [1, 2]. Wound incidence has been evenly distributed between upper and lower extremities, with the lower comprising 49% of injuries [3]. Improvised explosive devices alone produce 80.7% of all musculoskeletal combat casualties [4], while penetrating soft tissue wounds have accounted for 53% of extremity injuries [3]. These injuries may result in nerve and musculoskeletal damage, causing partial to full paralysis of the foot and ankle or even amputation.

Military members with lower extremity injuries often face the difficult decision between limb salvage and amputation based on considerations of function during activities of daily living, length of recovery, pain, and the ability to return to military duty [5]. Advances in assistive device technology have provided amputees the ability to walk, run, bike and perform many other vigorous activities, but at the cost of multiple different prosthetics [6, 7]. However, activities of daily living may be more complex for amputees, such as donning the prosthetic at night to use the restroom [8] and only 16.5% of amputees have returned to active duty during recent conflicts [9, 10].

Patients with limb salvage have also demonstrated the potential to return to active lifestyles with improved bracing devices, though this technology is not currently the military medical standard of care. New dynamic ankle-foot orthotics (DAFO) are multipurpose, and may provide these patients the potential to have more functional lives and return to military duty. Custom-fitted DAFO have shown promise in returning other patients including military personnel with drop foot to more normal functional gait patterns by aligning the limb to pre-injury positions and unloading painful segments of the extremity to allow for maximum comfort and propulsion [11, 12]. This correction may decrease their pain during weight-bearing to enhance recovery, while improving muscle function and reducing compensatory movements.

Gait deficits in limb salvage patients often manifest as decreases in vertical ground reaction forces (GRF) and joint powers, specifically during weight acceptance and propulsion [13-15]. Possible sources of these deficits include slower walking velocities and momentum losses associated with instability and drop foot gait compensations such as steppage gait or hip hiking [13, 16, 17]. Loss of momentum may also be the result of muscle atrophy such as weakened quadriceps, causing a reduction in power absorption during loading response and consequently affecting energy return at toe-off [16]. Similar gait deficits are common in other conditions with associated neurological damage such as cerebral palsy, stroke and other trauma [4, 11, 18]. Previous biomechanics research has examined the effect of ankle-foot orthoses (AFO) in improving gait for individuals with these conditions; however the ability for these AFO to facilitate long term changes in function, even when not wearing an AFO is not evident. To our knowledge, no studies exist examining the effects of DAFO use for improving strength and gait biomechanics in a military population. Unlike populations studied previously, active duty military may have greater need to return to running and vigorous activity, and are likely to benefit from the capabilities of a DAFO. Additionally, the young age and high pre-morbid levels of fitness in the limb salvage population may increase the likelihood that the use of a DAFO might facilitate increases in lower extremity strength leading to improvements in gait that are maintained, even when DAFO use is discontinued. Therefore, the purpose of this study was to determine the effect of wearing a DAFO for six months on lower extremity strength and gait biomechanics in military personnel who had sustained lower extremity neuropathy. We hypothesized that the use of the DAFO would result in an increase in ankle and knee strength after six months of use, leading to increases in GRF and joint power while walking without the DAFO.

Methods

Research Design

This prospective repeated measures design compared changes in the strength and biomechanics of subjects' involved lower extremity over six month after wearing a DAFO. The independent variables were the DAFO intervention and time (study onset and study completion). The dependent variables included ankle and knee strength, ankle (propulsive) and knee (absorptive and propulsive) joint power and peak vertical ground reaction force in the no-brace condition following six months of DAFO use.

Participants

Six military service members with a mean age of 29.3 ± 7.2 years, from all military branches and diagnosed with drop foot or other lower limb peripheral neuropathies affecting normal gait served as subjects. Eligibility for the present study included:

1. Military personnel with permanent lower extremity neuropathy
2. Military personnel who had been using a traditional AFO for more than one month
3. Military personnel who could walk continuously for 10 minutes

Instruments and Procedures

All study procedures were approved by the University of Hawaii Committee on Human Studies and the Tripler Army Medical Center Human Use Committee (HUC)/Internal Review Board prior to data collection. Service members completed the approved written Informed Consent document (Appendix A), and were scheduled for fitting of the dynamic ankle-foot orthosis (DAFO) by a Tripler Army Medical Center (TAMC) Physical Medicine and Rehabilitation physician. The same TAMC physical therapist measured muscle strength for ankle plantarflexion and dorsiflexion and for knee flexion and extension with a Microfet 2 handheld dynamometer (Draper, Utah, USA) using procedures described by Kendall [19] on the involved limb only. Three measurements were taken for each variable and the averages of these measurements were recorded on the "Anthropometric Data Collection Sheet" (Appendix B).

Service members' gait parameters were assessed prior to issue of the DAFO during one two-hour session per subject at the University of Hawaii Human Performance and Biomechanics Laboratory. Service members were asked to wear physical fitness attire during data collection. Upon arrival, the same research team members collected anthropometric measurements, including height, weight, leg length, and joint width. Height was determined using a standiometer (Model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA), and weight was assessed using a Befour PS6600-ST scale (Befour, Inc., Saukville, Wisconsin, USA). The research team recorded these measurements on the "Biomechanics Data Collection Sheet" (Appendix C) and marked the subjects' trunk and lower extremities in preparation for kinematic measurements. Twenty-seven (27) retroreflective markers were applied to each subject (list and photo in Appendix D), and subjects were instructed to walk at a maximum comfortable self-selected velocity (not to exceed 4.0 m/s ± 20%) down an 18-meter runway. A 13-camera Vicon MX 3D motion capture system (Vicon, Inc., Centennial, Colorado, USA), two Basler high-speed digital video cameras (Basler, Inc., Exton, PA, USA) and Vicon Nexus software (Vicon, Inc., Centennial, Colorado, USA) were used to capture, reduce, and analyze kinematic data. Two force plates (Advanced Mechanical Technology Incorporated, Boston, Massachusetts, USA) embedded flush with the floor surface were used to collect kinetic data during walking trials. Kinematic data were collected at 240Hz and time synchronized with digital video collected at 60Hz and kinetic data collected at 480Hz, then smoothed using a Butterworth filter with an 8Hz cut-off [20-22]. Speedtrap II (Brower Timing Systems, Draper, Utah, USA) infrared sensors placed four meters apart, in the middle one-third of the runway were used to ensure consistent walking velocity, defined as each subjects' maximum comfortable self-selected pace for the initial trial ± 20%. Walking gait was assessed over three successful trials for each foot in the shod no-brace condition. Kinematic and kinetic variables were calculated as the mean of three successful trials for each foot in each condition. Mean values of only a few trials have been determined by previous authors as sufficient for assessing gait data due to the high reliability between trials [23, 24]. A successful trial was defined as completion of the pass through the field at a consistent walking velocity and landing with one foot completely on the force plate with no obvious change in stride [23-25].

Gait analysis procedures were repeated without the DAFO, in shoes alone, at three and six months following DAFO issuance, in conjunction with a complimentary study examining gait changes monthly while wearing the DAFO. The final strength measures were repeated at six months. Service member control subjects, matched by age, height and weight to each injured service member, completed a one-time gait assessment in shoes to serve as normative values of healthy, uninjured service member controls.

Statistical Analysis

All data analyses were completed using SPSS (IBM Version 19) with an alpha level set at $p<0.05$. Non-parametric Mann-Whitney *U* Tests were used to assess changes in gait velocity and kinetics between the injured service member group and healthy service member controls. Wilcoxon signed-ranks test were used to assess changes in strength, anthropometrics and gait kinetics in the no-brace condition prior to and following six months of DAFO use. Injured group service members' age, height and weight were compared to control group members using a t-test. Reliability of strength testing measures was examined using the Intraclass Correlation Coefficient (ICC).

Results

Descriptive data for the injured service members and controls are presented in Table 1.0. The injured group consisted of five service members: three Army, one Navy, and one Air Force. Descriptive and gait analysis data were collected on the healthy service member control group, which consisted of one Army, one Navy, and three Air Force service members. There were no significant differences in age (p=0.75), height (p=0.63), or weight (p=0.95) between the injured and healthy service member groups. The injured service members sustained various injuries which lead to the common pathology of partial lower extremity paralysis and subsequent drop foot. Two of the five soldiers suffered gunshot wounds, one to the back of the knee, and one through the upper thigh. One Army physician suffered a stroke, a Navy seaman endured injuries from a motorcycle accident, and the Airman was injured while running during combat readiness training. Each injured service member was in a different stage of recovery, six to fourteen months post-trauma. Two of the service members had just completed surgeries at study onset. All participants met the inclusion criteria of permanent lower limb nerve palsy, had been using a TAFO for longer than one month and less than two years, and could walk continuously for ten minutes.

Table 1.0. Descriptive Data for All Service Members

	Injured Mean (SD)	**Control Mean (SD)**
N	5	5
Age (years)	29.33 (7.20)	30.67 (7.10)
Height (cm)	180.87 (6.06)	179.32 (7.06)
Weight (kg)	86.57 (15.89)	86.00 (12.71)

Intra-rater reliability testing of the strength scores produced ICC's ranging from 0.94 – 0.99 (ICC 3, k) with a standard error of the mean (SEM) ranging from 0.50 – 2.20 pounds of force (2.63 – 12.11% SEM) during manual muscle testing with the Microfet2 Hand-held Dynamometer. Table 1.1 presents the means, standard deviations (SD) and medians for all anthropometric, spatial-temporal and kinetic gait variables of interest. All injured service members values increased over time and kinetic gait variables approached values seen in healthy controls. Table 1.2 includes the results of a Wilcoxon Signed-Ranks Test which was utilized to assess the strength and kinetic gait variables within the injured service member group over time. There were no significant differences in strength, joint power and vertical GRF. Knee extension strength displayed the greatest improvement over time, although not at a level reaching statistical significance.

A Mann-Whitney *U* Test was used to compare kinetic gait variables between injured service members and healthy matched controls. Table 1.3 includes the medians, ranges, *Z*-scores, *p* values, and effect sizes for the variables of interest. Ankle power values at study onset were significantly different from the healthy matched control subjects prior to DAFO use (p=0.029). Following six months of DAFO use, the injured service members' ankle power had increased to a level no longer different from the controls (p=0.057) while not wearing the brace. Walking velocity markedly improved over the study period; velocity at study onset approached significantly lower values than controls and improved to levels similar to controls by study completion.

Table 1.1. Descriptive Statistics for Anthropometrics and Gait Variables in the No Brace Condition

	Pre		Post	
Injured Service Members (n=5)	**Mean (SD)**	**Median**	**Mean (SD)**	**Median**
Walking Velocity (m/s)	1.43 (0.14)	1.39	1.70 (0.20)	1.80
Ankle Power (W/kg)	3.35 (1.48)	3.89	4.26 (1.74)	4.37
Knee Power (K1) (W/kg)	-0.79 (0.82)	-0.44	-1.01 (0.55)	-1.03
Knee Power (K2) (W/kg)	1.04 (1.02)	0.66	1.56 (0.78)	1.51
Vertical GRF (N/kg)	11.22 (2.15)	10.23	12.07 (2.10)	11.55
Knee Flexion Strength (lbf)	22.01 (9.05)	20.70	31.37 (15.25)	25.17
Knee Extension Strength (lbf)	28.75 (8.68)	27.83	41.90 (12.38)	37.00
Ankle DF Strength (lbf)	2.98 (4.51)	0.00	8.92 (10.75)	9.00
Ankle PF Strength (lbf)	26.75 (14.51)	29.83	35.00 (21.63)	42.17
Service Member Controls (n=5)				
Walking Velocity (m/s)	1.86 (0.22)	1.92	n/a	n/a
Ankle Power (W/kg)	6.79 (0.97)	6.93	n/a	n/a
Knee Power (K1) (W/kg)	-3.24 (1.42)	-3.33	n/a	n/a
Knee Power (K2) (W/kg)	2.35 (1.19)	2.50	n/a	n/a
Vertical GRF (N)	12.88 (0.83)	12.98	n/a	n/a

DF: Dorsiflexion; PF: Plantarflexion; GRF: Ground Reaction Force; K1: Knee absorptive power; K2: Knee propulsive power

Table 1.2. Wilcoxon Sign Rank Test for Pre-Post Measures of Injured Service Members (n=5)

	Pre Median (Range)	Post Median (Range)	Z	*p*	Effect Size
Knee Flexion (lbf)	20.70 (11.70- 36.67)	25.17 (20.00 - 58.00)	-1.75	0.13	0.62
Knee Extension (lbf)	27.83 (20.40 - 43.00)	37.00 (29.67 - 59.00)	-2.02	0.06	0.72
Ankle DF (lbf)	0.00 (0.00 - 10.17)	9.00 (0.00 - 26.33)	-0.73	0.63	0.26
Ankle PF (lbf)	29.83 (1.93 - 38.93)	42.17 (0.00 - 54.50)	-1.21	0.31	0.43
Ankle Power (W/kg)	3.89 (1.18-4.47)	4.371 (2.05-6.26)	-1.83	0.13	0.65
Knee Power (K1) (W/kg)	-0.44 (-2.00 --.280)	-1.03 (-1.64--.33)	-0.73	0.63	0.26
Knee Power (K2) (W/kg)	0.66 (.281-2.54)	1.51 (.65-2.56)	-1.83	0.13	0.65
Vertical GRF (N/kg)	10.23 (9.98-14.55)	11.55 (10.26-14.92)	-1.83	0.13	0.65

Significant at $p \leq 0.05$; DF: Dorsiflexion; PF: Plantarflexion; K1: Knee absorptive power; K2: Knee propulsive power

Table 1.3. Mann Whitney *U* Test Comparing Injured versus Control Subjects (n=5)

	Median (Range)	Z	*p*	Effect Size
(Pre) Walking Velocity (m/s)	1.66 (1.32 – 2.13)	-2.02	0.057	-0.71
(Post) Walking Velocity (m/s)	1.80 (1.53 – 2.13)	-1.02	0.40	-0.36
(Pre) Ankle Power (W/kg)	5.04 (1.99 - 7.66)	-2.31	0.029*	-0.82
(Post) Ankle Power (W/kg)	5.94 (2.05 - 7.66)	-2.02	0.057	-0.72
(Pre) Knee Power (K1) (W/kg)	-1.78 (-4.72 - (-0.28))	-2.02	0.057	-0.72
(Post) Knee Power (K1) (W/kg)	-1.60 (-4.72 - (-0.33))	-2.02	0.057	-0.72
(Pre) Knee Power (K2) (W/kg)	1.35 (0.28 - 3.38)	-1.73	0.11	-0.61
(Post) Knee Power (K2) (W/kg)	1.61 (0.65 - 3.38)	-1.16	0.34	-0.41
(Pre) Vertical GRF (N/kg)	12.37 (9.98 - 14.44)	-1.16	0.34	-0.41
(Post) Vertical GRF (N/kg)	12.67 (10.26 - 14.92)	-0.87	0.49	-0.31

**Significant at* $p \leq 0.05$

Discussion

Service members with peroneal neuropathy in the present study showed improvements in strength and gait mechanics while shod with no assistive device after six months of DAFO use. Ankle power was significantly lower than controls at study onset, however increased to levels similar to able-bodied gait following six months of DAFO use. Walking velocity and knee power also increased, which indicated confidence in limb stability during weight acceptance and toe-off. Strength measurements of all lower extremity muscle groups increased over time, likely associated with the increased ability to exercise facilitated by the DAFO. The consistent use of a DAFO may benefit the rehabilitation of the lower extremity to a level where functional improvements are evident without the DAFO.

Six injured service members met the inclusion criteria for the present study and were able to complete a portion or all of the six month protocol. One service member will complete the study protocol in late April 2013. Two were medically retired prior to the end of the study protocol, one at five months following DAFO issue, and one less than halfway through data collection. Both met medical evaluation boards which necessitated military moves off-island. One of these service members was able to attend post-study data collections, and therefore was included in post-test analysis, whereas the other was unable to attend, and therefore his data was removed from statistical analysis.

Comparisons to healthy service members were important to accurately identify deviations from normal gait. Strength data was not collected for control subjects in the present study; therefore all strength comparisons for the injured service member group were directly compared to manual muscle testing values in published literature. Normative strength values for adult males (ages 20-39) at the knee and ankle included 129.22 pounds of knee extension force and

86.19 pounds of dorsiflexion force [26]. Normal ankle plantarflexion strength measures for males (ages 23-28) averaged 52.43 pounds of force [27].

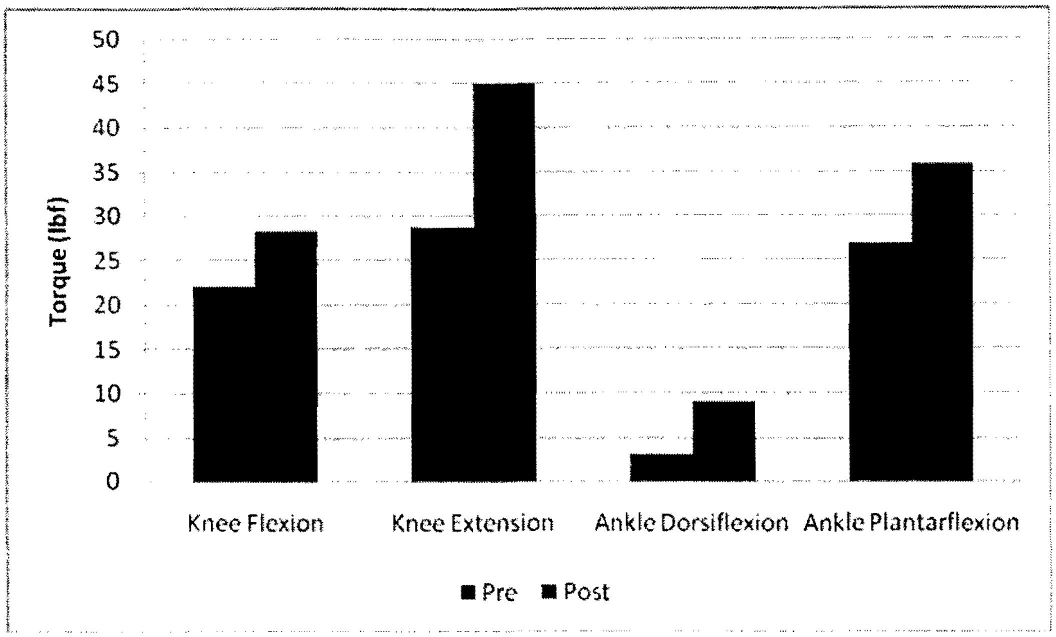
Differences in gait variables between the injured and healthy service member groups were considered pathologic deviations; to confirm the normalcy of the uninjured sample, control values were compared to published literature for normal walking gait biomechanics. Kinetic gait variables of the healthy service member controls represented normative values for vertical GRF and joint power [18, 28-30]. Peak vertical GRF during walking reached 1.4 times control subjects' body weight (12.88N); similar to values reported by Keller et al. (1996) (11.00N) and ranges by Ounpuu (1994) (1.3 - 1.5%) at comparable walking velocities [28, 29]. Maximum ankle propulsive power was 6.78 W/kg which was supported by Vardaxis et al. (1998) who measured power in young healthy males in ranges from 3.0 to 5.0 W/kg at slightly slower velocities than controls in the present study [30]. Similarly, knee power absorption and generation from initial contact through midstance was in line with values measured by Ounpuu (1994) and Whittle et al. (2010) [13, 28]. Overall, healthy service member controls were indicative of the normative population for kinetic gait parameters.

Strength measurements collected on the injured service member group at study onset revealed considerably decreased levels compared to normative values cited for knee extension, ankle dorsiflexion, and plantarflexion [26, 27, 31] likely related to atrophy following injuries (Figure 1.0) associated with nerve damage [32]. Knee extension strength, which is important for stabilizing the knee during full weight acceptance in early stance [14], improved at levels approaching significance (p=0.06) over the six months in the injured service member group. The DAFO restored the initial heel rocker, which allowed proper tibial advancement and stance phase knee flexion. This correct alignment promoted quadriceps activity, and likely contributed to the increase in knee extension strength over time. Although these increases were not significantly improved nor reached normative levels reported by Andrews and Bohannon [26, 31], knee extension increased by 31% (28.75 to 41.90 pounds of force) which indicated the brace may have aided in the observed strength gains by providing proper alignment and therefore allowing increased activity and rehabilitation.

The quality of eccentric quadriceps contraction is also evident in the level of absorptive knee power at loading response [13, 15] during tibial advancement. Knee power may be affected with proximal injuries to the peroneal nerve [16]. Drop foot patients with proximal injuries and related proximal atrophy have presented with weakened quadriceps and hyper-extension of the knee as a protection mechanism against buckling during loading response [16]. The strength deficits among injured service members in the present study were likely responsible for the observed deficits in knee absorption at loading response. Adequate improvements in knee extension strength over time would likely lead to improved absorptive knee power and contribute to increased confidence during weight acceptance.

Prior to the knee power generation in mid-stance, a period of zero power exists during knee stabilization by the advancement of body weight and soleus activation, characterized by the first vertical ground reaction force peak curve [15, 16]. The injured service member group exhibited lower knee power generation during mid-stance likely related to a more extended knee, which has been reported as the most stable weight-bearing condition in the absence of a heel rocker [14]. This has also been noted in research by Kim et al. (2004) [33], where post-stroke patients displayed a knee extension thrust pattern, which resulted in decreased power generation during midstance. Increases in knee extension strength ultimately allow for more propulsion during mid-stance through terminal stance for forward advancement. As with the absorptive knee power, improvements in knee extension strength appear to have produced a greater confidence in the involved limb during stance phase which allowed for a greater knee propulsive power (K2).

Figure 1.0. Anthropometric Strength Measurements at Study Onset and Study Completion

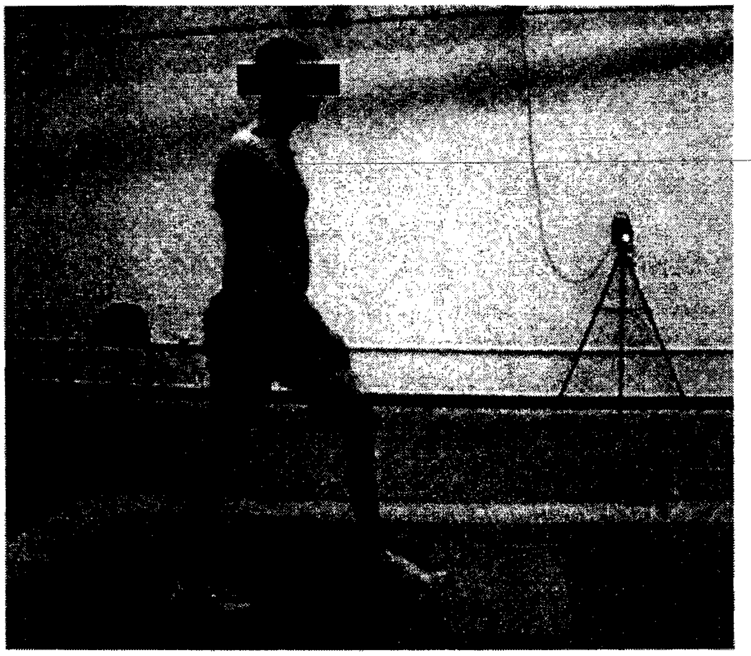


Published research on normal walking gait has reported a linear increase in vertical GRF with increased velocity [29]. The initial peak vertical GRF seen while walking is associated with acceptance of full body weight at initial contact through loading response [13, 14, 34]. Literature on patients with limb salvage have cited slower walking velocities are related to pain, musculoskeletal weakness and inadequate bracing, resulting in an antalgic gait and subsequent decreased initial vertical GRF [14, 35].

Cautious loading of body weight was observed in the injured service members group in the present study (Figure 1.1), and was reflected in decreased initial walking velocity and vertical GRF values at study onset. Following six months of DAFO use, the injured service member group reported increased confidence which translated to improvements in walking velocity and initial vertical GRF to nearly identical measures seen in healthy service member controls [29]. These improvements in vertical GRF were likely attributed to the improved walking velocity and may also be related to self-assurance during walking and improved biomechanics following six months of DAFO use.

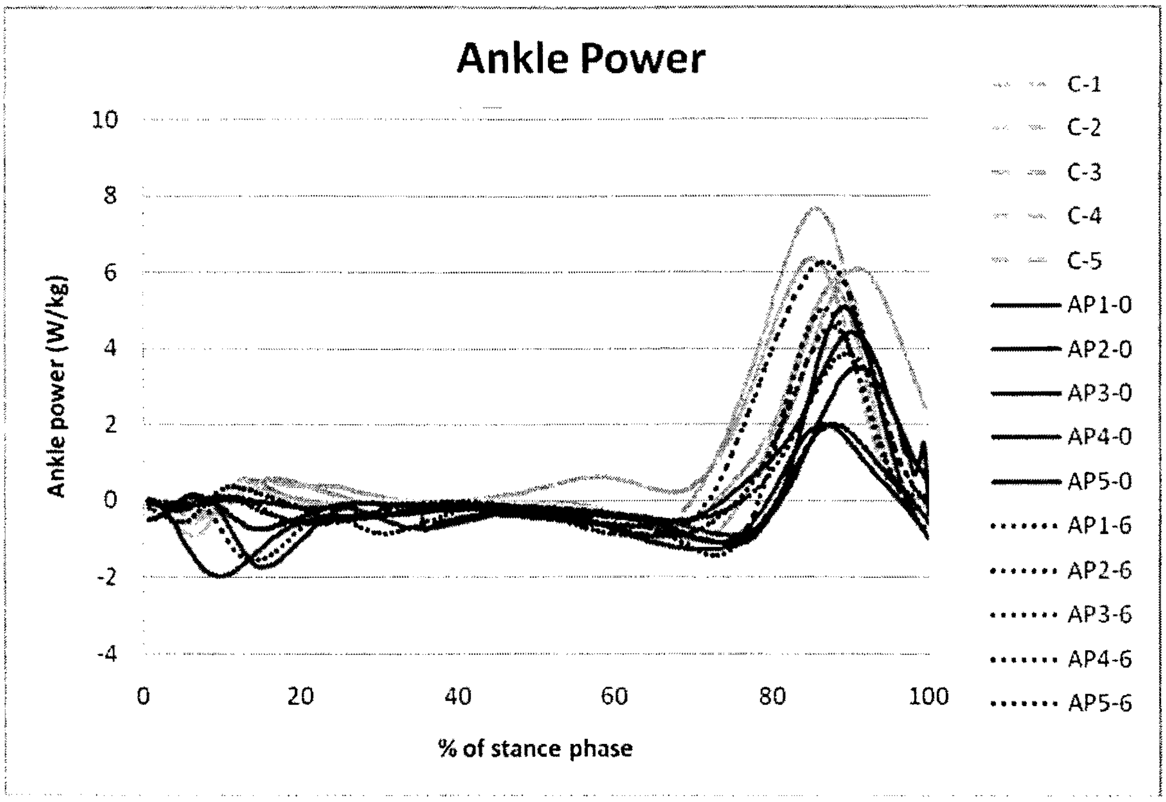
Kinetic waveforms, specifically joint power, are directly related to changes in walking velocity [33, 36]. Joint power has been described as the rate at which energy is either generated or absorbed through muscle and soft tissue expressed in forward muscle movement [13, 15]. A double wave of power absorption at the ankle exists through tibial advancement, due to the eccentric contraction of the soleus and gastrocnemius muscles during mid and terminal stance [13]. During pre-swing a large peak of positive energy exists corresponding to a propulsive concentric push of plantar flexors [13, 15]. Ankle propulsive power in the injured service member group was significantly lower than controls at study onset (p=0.03). The inability to activate lower extremity muscles due to injury resulted in loss of power generation in gait [17, 18]. This observation of low ankle power was also reported by Olney et al. (1991) who found near zero ankle power generated by the plantarflexors during toe-off in stroke patients at slow walking velocities [18].

Figure 1.1. Walking Body Posture of Service Member with Drop Foot



Propulsive ankle power improved to levels not significantly different than healthy controls upon study completion (p=0.057) (Figure 1.2). Proper alignment reinstated a heel strike at initial contact which eliminated the prominent foot slap or steppage gait observed in the injured service member group, and likely reduced energy requirements at the hip and energy loss at the ankle. This energy savings combined with returned stance phase knee flexion may have translated to greater push-off ability during pre-swing, with the added effort of plantarflexors. Injured service members reported feeling contraction of the triceps surae while wearing the DAFO which may have contributed to increased plantarflexion strength measured at study completion. These increases in strength and proper alignment within the DAFO improved propulsive ankle power during walking even when the DAFO was removed.

Figure 1.2. Ankle Power during Walking without an AFO at Study Onset and Study Completion Compared to Controls



Legend: Control (gray dashed line); Injured service member at study onset (gray solid line); Injured service members at study completion (black dotted line)

Limitations in the present study included differences in the nature of injury and level of recovery between service members, patient compliance, and the low number of available limb salvage patients. Despite the generally homogenous nature of groups of military service members, the injured service members in the present study were markedly different in injury type, severity and level of function at study onset. This disparity in function also affected the extent to which individual subjects were able to utilize the DAFO. Additionally, other related physical and psychological stressors may have affected their motivation towards rehabilitation and compliance with wearing the DAFO. Finally, the subject sample in the present study was restricted to limb salvage members within the Pacific region which limited the potential subject pool.

Conclusion

The DAFO demonstrated potential as a rehabilitation tool to increase kinetic variables including ankle joint power, knee joint power and vertical ground reaction force as well as knee and ankle muscular strength. These kinetic increases may be attributed to improvements in walking velocity as well as an increase in knee extension strength. The injured service members' vertical GRF increased to levels seen in healthy matched control subjects, and ankle joint power improved to a level not significantly different than controls. Future studies should include dynamic electromyography during gait to directly assess muscle function during walking, and continue to measure long-term functional outcomes of DAFO use in service members with lower extremity neuropathy as meaningful improvements in function when not wearing an assistive device may require more than six months to manifest.

Literature Review

Ankle-foot orthoses (AFO) are continually improving due to new injuries and advancements in research. Currently the United States Military issues an off-the-shelf brace that has been anecdotally regarded as minimally functional for an active duty service member. A new carbon fiber dynamic ankle-foot orthotic (DAFO) initially created for polio patients has shown promise to improve gait to functional levels. Because this is a new bracing system, there is little research on its affects on gait biomechanics and anthropometric measures after wearing the DAFO for any period of time in an active population.

AFO Studies

A number of studies have examined different types of AFO's and their effects on gait. Romkes et al. (2002) [11] assessed 12 children (mean age:11.9 ± 4.9 years) with hemiplegic cerebral palsy, nine with right and three with left-sided limb involvement. A control group of ten healthy subjects was included as well, with a mean age of 26.9 ± 6.3 years. They compared a dynamic ankle-foot orthotic (d-AFO) trimmed just above the tip of the malleoli to a hinged AFO (h-AFO), which extended to just below the knee and its flat foot-plate extended to the tip of the toes to provide control of the ankle. Both braces were compared to a no-brace condition. The patients walked at a self-selected velocity for 10 meters. The researchers reported none of the patients had heel-toe gait patterns on the involved side when walking barefoot.. The h-AFO produced a heel-toe gait in all twelve patients while only four of the patients walked with a heel-toe gait wearing the d-AFO. The h-AFO was found to reduce ankle plantarflexion angle at the heel strike to a normal value, creating a heel strike in all patients. These positive changes also resulted in increased step and stride lengths.

Prosthetics

The AFO is a major factor when deciding between limb salvage and amputation. Mackenzie et al. (2007) [6] conducted a two-year health care cost estimate comparing amputee versus limb salvage patients. Over a three year period, 601 patients were enrolled in this comparative study. Subjects were followed prospectively in person for the first three, six, twelve, and twenty-four months after injury. After approximately eighty-four months researchers attempted to locate the subject for follow-up. Healthcare costs were estimated by six factors: the initial hospitalization, all re-hospitalizations for acute care related to the limb injury, inpatient rehabilitation, outpatient doctor visits, outpatient physical and occupational therapy, and prosthetic devices and related services. Costs were adjusted to a "cost-to-charge" ratio for financial comparison created by Medicare, or by reimbursement rates for insurance companies. The total life-time healthcare costs for amputation were more than three times the costs for limb salvage. The large difference was attributed to the purchase and maintenance of a new prosthetic device. Initial treatment costs were highest for through–the-knee amputation. The amputee group costs were greater even without the specific costs of multiple devices for different activities or the increase in costs of newer devices due to advancements in technology and materials.

Amputees often cannot perform activities without the use of their prosthetics. Legro et al. (2001) [7] investigated activities low-limb amputees took part in and whether they needed their prostheses. Ninety-two individuals with low-limb prosthetics responded to the survey. Each survey asked a series of questions about their recreational activities, including naming two of their favorite activities. A coding system was developed to categorize each activity of particular interest; "Energy Level" to describe the energy cost of the activities and "Lower-Limb

Impact Load" to designate the load of the residual limb. The ability to perform an activity was much greater while wearing a prosthetic than not, as hypothesized, with an activity score of 66.5 (out of 0 to 100). The activity score without a prosthetic for the same activity was approximately 30.0 on the same scale. Some activities were reportedly performed better without a prosthetic, such as swimming, wheelchair basketball and sex.

Another study completed by Dougherty et al. (2003) [8] reviewed Vietnam veterans who had sustained transfemoral amputations during combat. This two-part study included a medical record review and a follow-up survey. Ninety subjects fit the inclusion criteria but only 46 entered the study. Medical records were reviewed for mechanism of injury, initial surgery in Vietnam, procedures performed, how long until they were fitted for their permanent prosthetic, etc. The survey included a Short Form-36 Health Survey which was administered to each participant by mail or phone. The survey was used to obtain information on other types of surgeries, other medical problems, marital status, employment status and prosthetic usage. Researchers reported most wounds were caused by multiple fragments from exploding munitions. Other injuries sustained consisted of lower extremity bone fractures, abdominal wounds, wounds to the head or face, and chest wounds. The average number of prostheses worn since their initial fitting was 7.7. The participants stated they wore the devices on average 14.1 hours per day, and about half stated they had changed the device or any major component. Forty-one participants had been employed for 20.1 years and 43 had been married for an average of 23.1 years.

Severe lower leg injury treatments take serious thought and decision. Shawen et al. (2010) [5] compared the foot and limb salvage to complete amputation, finding medical progress has made limb salvage more advanced as well as improved prosthetic designs.

Amputees and limb salvage patients are able to participate in recreational activities that were previously impossible. Amputee counselors are also being provided to aid the injured both physically and mentally. Patients can confide in individuals who have suffered similar experiences help with difficult decisions. Military personnel have an even more difficult time with these injuries because they are just trying to return to a productive member of society, or even military duty.

Military Statistics

Recent conflicts during the war on Terrorism have had high rates of combat injuries and casualties. Two articles by Owens et al. (2007, 2008) [1, 3] and two by Belmont et al. (2010, 2011) [2, 4] looked at the statistics of combat wounds, injury epidemiology, and incidence in Operation Iraq Freedom and Operation Enduring Freedom; two main conflicts in the current war. Both authors compared location of wounds; including head and neck, extremity, thorax, and abdomen, as well as types of injuries; open fractures, closed fractures and musculoskeletal injuries. Owens et al. (2008) found 11% of injuries were to the abdomen, and 54% were to the extremities. The proportion of these injuries occurring from gunshot wounds was 18% and improvised explosion device (IED) accounted for 78%. Musculoskeletal combat casualty wound rate per 1,000 combat per year were 2.1 for major amputation, 0.6 for minor amputation, and 32.8 soft tissue/neurovascular injuries[4]. Returning to activities of daily living can become a complex task for injured service members, let alone returning to duty.

Advancements in prosthetics are allowing for increases in daily activities as well as return to more athletic participation. Owens et al. (2011)[12] studied ten high energy lower extremity trauma (HELET) limb salvage patients while wearing the Intrepid Dynamic Exoskeletal Orthosis (IDEO). The IDEO is a custom fit, carbon fibe,r energy storing AFO. All patients went through an aggressive rehabilitation program focusing on strength, plyometrics, power and agility after being fitted for an IDEO. The program was broken up into two phases, one wearing the brace known as "out-of-frame" and one without the brace knows as "in-frame". The strength portion included squats, lunges, and dead lifts, increasing the weights and reps as the patient began to show gains. The running portion was started when the patient could perform the agility training without any increases in pain. The run phase was performed on a treadmill and track and was defined as the ability to run on a treadmill without stopping for two consecutive miles. After training, eight of the ten subjects returned to running, three returned to basketball, two to softball and three patients have re-deployed again in combat roles. The results reported that by using an aggressive sports-medicine style rehabilitation program along with the IDEO, limb salvage patients are able to return to higher level of functional and athletic activities. The IDEO helps to unload specific segments of the lower extremity to allow for pain free weight bearing which allows for increased activity level. A higher activity level can increase the rate at which these injured service members can return to active duty.

Cross et al.(2012) [9] and Stinner et al. (2011) [10] examined the return to duty rate in soldiers sustaining tibial fractures and amputee soldiers. Tibial fractures had a return to duty (RTD) rate of 18%, while individuals with salvaged extremities and amputees had RTD rates of 20.5% and 12.5%, respectively. Stinner et al. [10] did not study specific injuries, but the overall amputee population during the conflicts in Iraq and Afghanistan. Three hundred and ninety five soldiers from 2001 to 2006 met the inclusion criteria and were reviewed. Within the 395 participants, sixty five returned to active duty, or 16.5%. Researchers noted that officers returned at a higher rate (35.5%) when compared to junior enlisted personnel (7.0%), and double amputee had lowest return to duty at only 3%.

Normal Walking

Normal walking is a method of locomotion involving the use of the two legs, alternately to provide both support and propulsion[13]. Novacheck (1999) [36] reviewed the biomechanics of running, and discussed many variables of gait in able-bodied individuals. Walking, running and sprinting were all compared the difference being, double support during walking and a short period of flight during running and sprinting. Velocity was also a factor differentiating types of gait. Researchers noted that the amount of ankle power was directly related to gait velocity, because the power being generated gives energy for forward propulsion. The eccentric quadriceps contraction is limited due to the plantarflexors absorbing most of the shock, therefore limiting power that is absorbed by the knee, which differs from walking. Overall, the total amount of power generated increased as the velocity increased. The contribution from each of the muscle groups varied relative to the power increases.

Ounpuu (1994) [28] discussed different aspects of normal walking and running gait. Variables that were reviewed included joint kinematics, kinetics, muscle activation and energy consumption. Subjects tested were normal children at the Newington Children's Hospital in their gait analysis laboratory. Kinematic data was collected using reflective markers on joints of the lower extremity, and kinetic data was collected using two force plates embedded in the floor. Euler angles and Newtonian mechanics were used to compute these variables. Notable results included vertical ground reaction forces ranging from 1.3 to 1.5 times the body weight during loading response and push off. Understanding what is considered normal gait assists in quantifying pathological gait and how to resolve it.

Ground Reaction Force

Ground reaction forces are forces exerted by the ground at foot contact[28]. Gait compensations due to injury can affect the amount of vertical GRF that an individual may exert during walking. The study performed by Keefer et al. (2008) [37] compared the ground reaction forces (GRF) produced different types of short leg walkers. Two short leg walkers were used, a gait walker and equalizer, and always worn on the right foot. Participants performed five level walking trials in six randomized conditions; lab shoes, gait walker, gait walker with heel insert on shoe side, gait walker modified with insert on walker side, equalizer walker, and equalizer walker with heel insert on shoe side. The main variables of focus included initial peak vertical GRF, peak vertical GRF with load acceptance, peak vertical GRF with push-off, minimum GRF during mid-stance, peak anteroposterior breaking GRF, peak anteroposterior propulsive GRF and time. Vertical GRF peaked earlier than normal peaks while walking for most subjects. The propulsive peak GRF for the right side walker conditions was significantly lower when compared to shoes alone, however the same values for the left side were significantly higher for gait walker, gait walker with heel insert, equalizer, and equalizer with heel insert than the right side.

Keller et al. (1996) [29] compared vertical GRF to the velocities of walking and running in healthy active males and females. All subjects were tested in the same type of running shoes, and were asked to walk and jog down a six meter runway with an embedded force plate. Four walking velocities and four running velocities were collected for both men and women. Foot patterns were also collected by videotape recordings. Researchers reported that most subjects were rear-foot strikers, but above speeds of 3 m s^{-1} there was an increase of mid-foot and forefoot striking. The vertical GRF presented with a double peak at both walking and slow jogging velocities. As running velocity increased, the vertical GRF consisted of a single peak at about 40-50% of total stance time. The results present a positive linear relationship between velocity and vertical GRF.

Chiu et al. (2007) [38] studied the effect of velocity and gender on rating of perceived exertion (RPE), muscle activity, joint motion of the lower extremity, vertical GRF and heart rate during normal walking. Thirty healthy adults (15 males, 15 females) with no history of musculoskeletal disorders participated in the study. All subjects were randomly assigned to three different walking groups in efforts to minimize between-group variance. A sex-camera motion capture system was used to capture kinetic data, while a force platform was used to collect kinematic data. All participants walked on a treadmill for five minutes at a self selected pace, while a metronome was used to keep the individual's pace. After the five minutes was complete the subject was asked to walk on an eight meter walkway, using the metronome to maintain the same pace for nine trials. Results reported that velocity had a significant effect on vertical GRFs. Fast walking velocities (4 and 5 km/hr) generated a higher initial peak vertical GRF during loading response. Mid-stance however, had a lower vertical GRF with increasing speeds which has also been noted in previous research done by White et al. (1996)

Strength Measurements

Decreased mobility due to injury can cause atrophy almost immediately. Immobilization, paralyzing nerve damage or soft tissue injury can often cause atrophy. Increased levels of daily musculoskeletal stress can reverse the affects of muscle wasting. A study performed by Staron et al. (1994) tested the skeletal muscle adaptations after an eight week resistance training regimen in both men and women [39]. The researchers' main focus was to examine the time for specific muscular changes during the early phases of resistance training as well as compare differences between untrained and previously trained individuals. Thirty three healthy individuals, 21 untrained subjects and 12 trained control subjects participated in the study. Body composition and girth measurements were taken before and every two weeks during the 8-week training regimen. Muscle biopsy, and blood serum were also taken to test muscle fibers and hormone levels during the study. The end of the 8-week resistance training revealed no significant changes in anthropometric measures. However, maximal dynamic strength values increased after just four weeks for leg extension, leg press and squat for men. Women also increased after four weeks with leg extension and squat, but increased only after two weeks with leg press. Both genders continued to increase in strength as the study progressed throughout the eight weeks. Although gender differences were seen in different studies, these differences did not contribute to any apparent adaptive difference in early phase heavy resistance training.

Wiley et al. (1998) [40] compared lower extremity muscle strength in spastic diplegia and spastic hemiplegia. Thirty children between the ages of 5-12 with spastic cerebral palsy (CP) were age-matched to a group of 16 healthy children with no known neurological disorders, and strength measures were taken with a hand-held dynamometer. The weakest muscle groups were the hip extensors and the ankle dorsiflexors and plantarflexors. Four significant differences between the involved and uninvolved hemiplegic sides were identified. The dorsiflexors and plantarflexor muscle groups were weaker on the involved side with the knee flexed and with the knee extended. Strength on the uninvolved side of hemiplegia patients was still significantly lower than the age matched control subjects. However a disclaimer was made that CP patients are weaker than their "normal" peers. Even patients with hemiplegia, their uninvolved side should still not considered "normal".

Blaya et al. (2003) [16] studied the biomechanical effects of drop foot and tested a new active ankle-foot orthotic to improve gait. This manuscript included a review of gait biomechanics literature looking at the phases of gait and stability. During loading response the knee accepted body weight producing and external flexion moment. The quadriceps reacted to prevent excessive flexion which is then assisted by the soleus and forward motion of the body weight. If the quadriceps are weak, an individual will protect them by reducing the heel rocker in loading response.

Andrews et al. (1996) [31] tested one hundred and fifty six healthy subjects age ranging from 50 to 79 to obtain normative values for muscle force using a hand-held dynamometer. There were at least 25 males and 25 females in each testing decade. The three testers were the authors who have tested thousands of patients in their more than eight years of employment. Eight upper extremity and five lower extremity actions were isometrically tested on both dominant and non dominant sides. Each action was measured in a gravity neutral position, and they subjects were allowed at least one practice trial so they were familiar with pushing against the dynamometer. Each action was tested twice, with each trial lasting seven seconds.

The correlations between muscle force and gender, weight and height were a moderate to high significance, while the correlations between force and age were significant but weak. The mean values found can provide a reasonable estimate of normal if looking at a subject with assumed impairments. Any force that is less than two standard deviations below the mean value found can be considered "below normal force".

Bohannon et al.(1997) [26] tested 106 men and 125 women from ages 20 to 79 years of age to obtain reference values for muscle strength using a handheld dynamometer. One tester, with ten years of experience tested each subject twice in gravity eliminated positioning for each motion tested bilaterally. The subjects were instructed to build to maximum contraction for two seconds and then hold for five more. The max force was recorded for each subject, with 21 subjects exceeding the max of 650 N with knee extension. Results reported a strong correlation between muscle strength and age, weight, and sex. However age did not correlate as highly which was surprising to researchers. Values reported are quite similar to that of previous research with similar age groups. This study includes a greater number of participants and age ranges; however may not the perfect representation of all populations due to being a convenience sample.

Fakel et al. [27] performed a study testing adults, adolescents and children's plantarflexion strength using a cybex isokinetic dynamometer. One hundred and twenty subjects from Duke University and local schools were split up into three categories by age (6-8 years, 14-16 years, and 23-28 years). Subjects met the inclusion criteria if they did not regularly participate in athletics or activities such as running, and they were in the 15 percent or lower of their height and weight standard as established by the Metropolitan Life Insurance Company. Each subject was put in the long sitting position and their dominant leg, which was determined by having the individual kick a soccer ball, was placed in a footplate of the machine. A strap was placed at the tibial tuberosity to insure only movement at the ankle joint. After instruction, each subject performed a few sub-maximal contracts to become familiar with the resistance and speed by the device. Three trials of both isokinetic and isometric contractions were recorded and then averaged for their final score. Results reported that weight and age were significant factors when determining strength while height and sex did not. The 6-8 year old female group had a significantly higher mean than the males did; the 14-16 year old males had slightly higher means than the females did; and the adult men were significantly stronger than the adult females were.

Further research needs to be done incorporating all major muscle groups and joint ranges of motion. Strength measurements are not only an important clinical test but biomechanically they can have effects on other variables such as velocity and joint power as well.

Joint Power

Walking and running velocity affects many kinetic patterns in gait biomechanics, specifically joint power. Kim et al. (2004) [33] looked at the relationship of kinetic and kinematic gait profiles of stroke victims and compared them to walking velocity. Over time, stroke victims developed a compensatory gait to allow for an efficient method of walking. Efficiency was tested in three planes: frontal, transverse and sagittal, and then groups were divided into fast walkers and slow walkers. Each participant was advised to walk down an eight-meter walkway over three force plates at their most comfortable velocity. Five trials were collected for each leg of each participant. Knee power during the first phase in the sagittal plane represented negative work meaning eccentrically contraction by the flexors. There was also a lack of plantarflexion after initial contact due to an absence of normal dorsiflexion moment.

A study performed by Chen et al. (1997) [17] compared the affect of walking velocity on joint powers in the hip, knee and ankle. Ten subjects with the average age of seven walked down a ten-meter walkway at three different velocities. Slow velocity was considered less than 0.9 m/s, normal velocity was considered 0.9-1.2 m/s, and fast was greater than 1.2 m/s. Power was measured at each joint including the hip knee and ankle. Power was positive in the first half of the stance phase at the hip, and then became negative in late stance. Power mainly stayed negative at the knee while at the ankle the power was negative in early stance followed by a large positive peak in late stance. Chen also found that the ankle helped with slow walking propulsion while the knee and hip musculature are responsible for fast walking acceleration. This shows there is a correlation between walking velocity and joint power.

Olney et al. (1991) [18] studied thirty adults with hemiplegia able to walk, 19 of which were males. The inclusion criteria comprised of patients suffering one stroke, being treated in a rehabilitation facility, were able to follow directions and were willing and physically able. Subjects were allowed to walk with a straight cane or ankle-foot orthotic. Nine cameras and retroreflective markers were used to collect kinematic data in conjunction with a force plate collecting kinetic data. Three trials for each leg was recorded and then averaged. Researchers reported that positive ankle power by the plantarflexors was near zero on the affected side at the slowest speed. This shows a positive correlation between both velocity and ankle joint power.

Sadeghi et al. (2000) [41] studied 19 adult males with no recent injuries or apparent gait disorders. All subjects were right hand and right leg dominant. Even though dominance was not studied, it was uniform so it did not make a difference. Each subject was set up with a 20 marker set-up, and instructed to walk through an 8-camera set up with two force plates in the center of the walkway. Then a 2 step approach was used to identify which muscle contributed to the each respective power during walking gait. 52 different gait variables were reviewed, including spatiotemporal variables. Results reported nine of the 52 variables were significant, and 6 of the 9 significant variables were related to muscle power. Both limbs generated similar amounts of energy and power, but only hip extensor and internal knee rotation were the same on both limbs. During walking, when power generation was present, the propulsion was secondary to control activities. They may have even occurred due to correcting actions from the leading limb propulsion. The peak hip extensor power has been associated with control of the forward acceleration. Considering the strong hip activity and the absences of ankle power variables when run in the principle component analysis, researchers inferred that the ankle plays a passive role in propelling the limb forward.

Vardaxis et al. (1998) [30] tested nineteen able bodied young men from the University of Montreal. They were all right hand dominant to avoid any sidedness bias. Twenty four reflective markers were placed over joint centers and anatomical landmarks to help determine 3D kinematics. Four video cameras and two force plates were used to collect the walking trials performed by each subject walking at their own self selected pace. Cluster analysis was used as the multivariate process for detecting natural grouping in the data, grouping the 3-D peak powers into "families". Once divided, five groups were created each with different characteristics. Family 1 had a total of three subjects and was considered the propellers. Family 2 contained only two subjects and had the highest K1 knee power due to the abduction moment stabilization during weight transfer. Following, was the highest K2 power bringing the knee back into extension. Family 3 included six subjects and were considered middle of the road performance while maintaining a walking speed. Family 4 also contained six subjects, developing the second highest hip activities at heel-strike. Family 5 comprised of the last two subjects which had the slowest normal walking speed when compared the other families.

Based on the information presented, research on the effect a dynamic ankle-foot orthotic (DAFO) has on both anthropometric strength gains and biomechanical gait changes in military personnel with lower extremity neuropathy is needed. The use of a DAFO as a treatment for gait rehabilitation has not been tested in this specific population and new DAFO's have shown promise in returning individuals back to daily activities.

Appendix A. Informed Consent

(Separate attachment due to formatting)

Appendix B. Anthropometric Data Collection Sheet

Passive Range of Motion	Measure 1	Measure 2	Measure 3	Mean
Hip Flexion				
Hip Extension				
Knee Flexion				
Knee Extension				
Ankle Dorsiflexion				
Ankle Plantarflexion				
Calcaneal Inversion				
Calcaneal Eversion				

Strength	Measure 1	Measure 2	Measure 3	Mean
Hip Flexion				
Hip Extension				
Hip Abduction				
Hip Adduction				
Knee Flexion				
Knee Extension				
Ankle Dorsiflexion				
Ankle Plantarflexion				
Ankle Inversion				
Ankle Eversion				

Leg Segment Girth	Measure 1	Measure 2	Measure 3	Mean
10cm above patellar superior pole				
10cm below patellar inferior pole				

Appendix C. Biomechanics Data Collection Sheet

Date: __________ Subject ID: __________

Data Collection Session: 0-0 1 2 3 4 5 6

Biomechanics Data Collection Sheet 3

Resting HR: ___________

	Shoes Only	DBS-AFO
Height (mm)		
Weight (kg)		
Left leg length (mm)		
Left knee width (mm)		
Left ankle width (mm)		
Right leg length (mm)		
Right knee width (mm)		
Right ankle width (mm)		

Self-Selected Walking Speed (4m time): __________ ± 20% _________

Total Trials:
1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20

SHOES

1	L R
2	L R
3	L R
4	L R
5	L R
6	L R

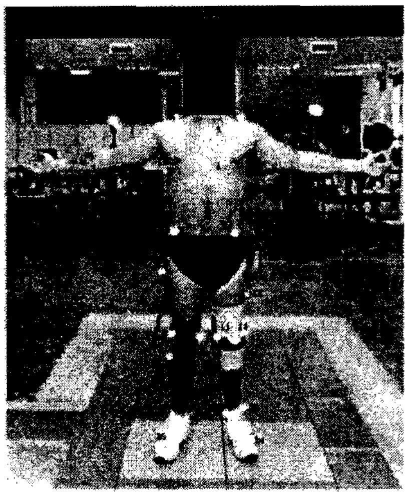
Total Trials:
1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20

DBS-AFO

1	L R
2	L R
3	L R
4	L R
5	L R
6	L R

Appendix D. Marker Set and Placement

Vicon Plug-In-Gait Lower-Limb & Thorax Marker Set



1. C7 Spinous Process
2. T10 Spinous Process
3. Clavicle
4. Sternum
5. Left Acromio-clavicular (AC) Joint
6. Right Acromio-clavicular (AC) Joint
7. Right scapula
8. Left Anterior Superior Iliac Spine (ASIS)
9. Right Anterior Superior Iliac Spine (ASIS)
10. Left Posterior Superior Iliac Spine (PSIS)
11. Right Posterior Superior Iliac Spine (PSIS)
12. Left Thigh (slightly lower than right)
13. Right Thigh (slightly higher than left)
14. Left Lateral Knee
15. Left Medial Knee
16. Right Lateral Knee
17. Right Medial Knee
18. Left Shank (slightly lower than right)
19. Right Shank (slightly higher than left)
20. Left Lateral Malleolus
21. Left Medial Malleolus
22. Right Lateral Malleolus
23. Right Medial Malleolus
24. Left Base of Second Metatarsal
25. Left Heel
26. Right Base of Second Metatarsal
27. Right Heel

Appendix E. Raw Data

Pre	Ankle Power	Knee Power (K1)	Knee Power (K2)	Vertical GRF	DF Strength	PF Strength	Knee Flexion Strength	Knee Extension Strength
Mean	3.357	-0.792	1.036	11.218	4.859	26.858	20.617	36.500
SD	1.485	0.815	1.018	2.151	5.622	17.403	7.200	12.430
Min	1.183	-2.002	0.281	9.975	0.000	1.933	11.700	23.500
Max	4.468	-0.280	2.535	14.440	10.167	42.167	29.333	50.500

Post	Ankle Power	Knee Power (K1)	Knee Power (K2)	Vertical GRF	DF Strength	PF Strength	Knee Flexion Strength	Knee Extension Strength
Mean	4.261	-1.008	1.555	12.070	10.017	40.375	34.959	31.958
SD	1.742	0.549	0.784	2.102	11.482	13.166	16.867	4.065
Min	2.045	-1.639	0.646	10.256	0.000	28.667	20.000	27.833
Max	6.256	-0.328	2.562	14.916	26.333	54.500	58.000	37.000

Control	Ankle Power	Knee Power (K1)	Knee Power (K2)	Vertical GRF
Mean	6.785	-3.235	2.346	12.882
SD	0.968	1.422	1.186	0.829
Min	5.615	-4.719	0.999	11.775
Max	7.656	-1.559	3.383	13.786

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