THE EFFECT OF DYNAMIC ANKLE-FOOT ORTHOSES ON GAIT BIOMECHANICS, FUNCTION AND QUALITY-OF-LIFE IN SERVICE MEMBERS WITH PARTIAL LOWER EXTREMITY PARALYSIS

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Melanie M. Presuto

Dissertation Committee:
Christopher D. Stickley, Chairperson
Iris F. Kimura
Ronald K. Hetzler
Seongah Im
Michael F. Zupan

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ABSTRACT

Lower extremity injuries account for over half of combat casualties in Operation Iraqi Freedom and Operation Enduring Freedom, and often involve nerve damage resulting in partial lower extremity paralysis. Traditional ankle foot orthoses (TAFOs) used to manage lower limb peripheral neuropathy are reportedly insufficient for the physical demands of active service members including a return to running. Recently developed dynamic ankle-foot orthoses (DAFOs) have been successful in returning service members to vigorous activity and ultimately a return to active duty. This cases series measured anthropometric, walking and running gait biomechanics, and quality-of-life changes in six service members (age 29.3±7.2 years) diagnosed with drop foot while wearing a DAFO over a six-month period, and compared gait characteristics to an age and anthropometrically matched healthy service member control group. Walking spatial-temporal, kinetic, and kinematic gait parameters at the knee and hip improved to closely resemble normative levels. Walking velocity in the DAFO exceeded controls (DAFO: 1.96 vs. Controls: 1.86 m/s) while controlling dorsiflexion velocity (No brace: 200.74°/sec; DAFO: 69.77°/sec; Control: 116.16°/sec) to increase ankle stability. Improvements in running gait were apparent with DAFO use (one-third of subjects were unable to run without the DAFO), which provided better comfort, stability and energy return during push-off and translated to continual gait improvements over the six month study period. Increases in running velocity (No Brace: 3.18 m/s vs. DAFO: 3.61 m/s) and ground reaction force, reported increased confidence during limb loading, and observed reductions in proximal kinetic chain compensations including decreased forward trunk lean and increased knee flexion moments at loading response indicated improved running capability. These improvements translated to increases in strength in all lower extremity muscle groups measured via manual muscle testing, as well as a significantly

improved ability to ascend stairs measured via the SF-36 quality-of-life questionnaire. The improvements in service member quality-of-life contributed to a ‘very cost-effective’ rating of the DAFO despite the increase in price over traditional models. Functional outcomes from this study may be used to improve the standard of care for service members with limb salvage and provide evidence-based outcomes for the optimal cost-benefit AFO.

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CHAPTER I - INTRODUCTION

Operation Iraqi Freedom (OIF) and Operation Enduring Freedom (OEF) are the largest scale armed conflicts since the Vietnam War. During these operations, over 50,500 soldiers have sustained combat injuries not resulting in death [1] including 3,102 combatant casualties with 6,609 inflicted wounds, of which 54% were in the extremities [2-5]. These injuries were evenly distributed between the upper (51%) and lower extremities (49%) [6], and combat explosions accounted for 78% of the mechanisms of injury [3].

Treatment of significant limb injuries in the field often requires military physicians to quickly decide between limb-salvage and amputation. The advancement of primary vascular repair, management of extremity injuries and soft-tissue infection have allowed for decreased rates of primary amputations and increased the number of limb-salvage patients [7]. Compared to amputees who initially have a greater range of motion, less pain, more prosthetic options, and appear to have a larger support group [8], limb-salvage patients tend to recover slower [9] and have significantly fewer orthotic options [10]. These service members with limb salvage, in consultation with their physicians, often opt for secondary amputation due to non-functionality [11, 12]. Though observed functional outcomes appear similar, lifetime cost for amputees are higher and quality-of-life scores tend to be lower when compared to limb-salvage patients [9, 13-15].

Previous quality-of-life research on service members returning from war with lower extremity trauma has cited prolonged hospitalization and recumbency as producing negative physical and psychological effects on the patients [16]. Rehabilitation for limb-salvage typically requires an average of eighteen months [17] and may necessitate re-hospitalization for major complications [9, 12, 18] whereas primary amputees are typically able to return to work within six months. Currently 17-22% of lower limb amputees are able to return to active duty status [19, 20]. However, despite complications associated with limb-salvage, patients overwhelmingly prefer a salvaged limb to amputation [11, 21].

Despite successful limb salvage surgery, severe lower extremity trauma may result in loss of peroneal nerve function leading to a lack of muscle control, inability to dorsiflex and evert the foot, and extend the toes producing an altered gait pattern [22, 23]. Patients with complete drop foot may also present with the ankle in slight inversion, associated with the loss of dorsiflexion capacity [24]. This condition, commonly called "drop foot," is characterized by the slapping of the foot after heel strike and dragging of the toe during the swing phase [22, 25-27]. Anthropometric examinations may reveal weakness in ankle eversion, ankle and toe dorsiflexion, and more proximal deficits depending on injury origin.

Traditional methods of standard care to mechanically treat injuries resulting in residual lower extremity neuropathy include the application of an ankle-foot orthotic (AFO) [22, 23, 27-35]. These AFO are capable of successfully correcting drop foot but do not promote energy return, appropriate range of motion (ROM) or adequate comfort and stability needed for return to high level activity [10, 23, 36]. Previous AFO research has been limited to populations such as children with neuromuscular disorders, post-polio patients and post-stroke patients and have only investigated walking gait [28, 29, 31-33, 37-40]. Research involving young, pre-morbidly fit military service members who have undergone limb salvage and desire a return to running is lacking. Consequently, versions of the AFO currently issued to military patients are likely inadequate to allow return to running and high-level military activities due to limitations in comfort and dynamic movement [23]. While past outcome goals of successful limb salvage have included ambulation without pain or assistive devices [10], this may not be adequate for military members pursuing a return to vigorous activity or combat duty.

Recently, dynamic AFOs (DAFO) have been introduced as an alternative to traditional versions (Figure 1.1). Designers claim these devices are capable of providing energy return from stance phase to push-off [37, 41-43], offering medial-lateral and rotational stability [34, 42], and improving ambulation over varied terrain at variable running velocities [42, 43]. Individual case studies have supported these claims and provided preliminary evidence that DAFO may improve function in multiple lower extremity pathologies and allow a return to vigorous physical activity (see Chapter V). Further investigation is needed to define specific parameters optimal for return to normal walking and running gait, as well as which AFO characteristics are essential for the rigorous activities found in combat.

Figure 1.1. Dynamic Ankle-Foot Orthoses

Despite the proposed benefits of DAFO for limb salvage patients, the costs associated with these devices have precluded their use as standard care. However, the ability of a DAFO to allow a service member to return to combat readiness may warrant the added costs. The level of improvement in quality-of-life alone may justify the cost difference between traditional models and the DAFO. Cost-utility analyses have been useful in the medical field to provide justification for more expensive equipment and devices based on improvements in quality-of-life [44]. Cost comparisons based on currency alone cannot adequately depict long-term clinical health outcomes, which if improved may provide a savings in total future health expenditures. The costs of returning wounded service members to duty using a DAFO versus the long-term cost of medically retiring these soldiers have not been thoroughly studied, though researchers agree costs of treatment, rehabilitation, and lifetime costs of devices are important in guiding treatment decisions [12].

Based on the increased incidence of drop foot among physically active service members returning from war and the limitations of published research, the purpose of this three-part study was to examine changes in biomechanical, anthropometrics and quality-of-life, as well as provide cost-benefit data necessary to guide AFO prescription for service members with drop foot. Part one assessed six injured service members over the initial six months of DAFO use in terms of biomechanics, strength and quality-of-life, and compared pre- and six-month biomechanics data to a healthy, uninjured service member control population; age- and anthropometrically-matched to each injured subject. Differences in gait biomechanics between no AFO, a traditional AFO and a DAFO were also investigated. Part two (see Chapter V) descriptively evaluated three case studies of service members who have been wearing a DAFO for longer than six-months, to assess long-term functional outcomes and quality-of-life. Part three (see Chapter VI) analyzed the costs between the traditional and dynamic AFOs, associated changes in quality-of-life and repercussions of these changes on return to duty versus medical retirement.

The following research hypotheses were considered:

1. There would be an increase in anthropometric measurements (range of motion, muscle strength, thigh and lower leg girth) between brace fitting and at six months after brace issue.
2. There would be improvements in walking and running gait mechanics in the DAFO over the traditional AFO.
3. There would be improvements in walking and running gait mechanics in the DAFO that allow for more normal gait when compared against an age- and anthropometrically-matched service member control population.
4. There would be an improvement in the Health Related Quality-of-life Short Form-36 Questionnaire score between DAFO fitting and at six months after brace issue.
5. There would be an increased cost associated with medical retirement of a wounded service member compared to returning to active duty with the proper AFO device.
6. The DAFO would be found more cost-effective than the traditional AFO when compared using cost-utility analysis.

CHAPTER II - METHODOLOGY

Research Design

This prospective, repeated-measures case series examined anthropometric, biomechanical and quality-of-life changes of military members diagnosed with drop foot before and during six months of using a custom dynamic AFO (DAFO) (Dynamic Bracing Solutions AFO, San Diego, CA). Changes in the biomechanics of injured service members were compared to an anthropometrically matched, non-injured service member control group to identify improvements in walking and running gait.

Participants

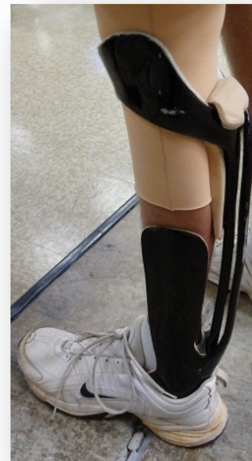
Subjects included six military personnel (mean age: 29.3 ± 7.2 years) from all branches of service who had been diagnosed with drop foot or other lower limb peripheral neuropathy which affected normal gait. Subject eligibility for the study included:

1) Active-duty military personnel
2) Permanent partial lower extremity paralysis
3) Traditional AFO use for more than one month
4) Ability to walk continuously for 10 minutes without aid

Control subjects included age- and anthropometrically-matched military personnel from all branches of service. The study protocol was approved by the Human Use Committee at Tripler Army Medical Center and the University of Hawaii, Manoa Institutional Review Board's Human Studies Program. Investigators adhered to policies for protection of human subjects as prescribed in 45 Code of Federal Regulation 46.

Testing Devices

Figure 2.1. DAFO



Two testing devices were used in this study: a traditional AFO (TAFO) and a dynamic AFO (DAFO). Subjects used their most recently issued TAFO for this study; therefore these TAFOs were not identical but were grouped as non-custom, non-dynamic orthotics, and were used for comparison in gait analyses. Traditional AFOs were generally made from a thermoplastic sheet that paralleled the posterior leg and statically supported the heel. The DAFO (Figure 2.1) utilized a continuous carbon fiber composite design, incorporating a cuff around the proximal tibia, two tibial struts, and a foot component. This commercially available device met the FDA criteria for custom devices under rule 21 C.F.R §812.3(b), and was exempt from FDA approval.

Fitting and fabrication of the DAFO was a lengthy process. The certified orthotists and manufacturer of the DAFO custom-molded the brace to fit each individual based on correct alignment and injury severity, and fabrication took an average of one month. Injured service members then returned for brace fitting and adjustment of the new DAFO, which was used in conjunction with a rubber sleeve for the knee as padding for the proximal cuff, a hard foam spacer attached to the posterior strut which could be modified to accommodate swelling associated with trauma or gastrocnemius hypertrophy, and a heel lift for the sound limb shoe due to the angle of the foot component. The injured service members required a 1.5 increase in shoe size of the involved limb to accommodate the DAFO.

Instrumentation and Protocol

Injured service members and controls completed the approved written Informed Consent document (Appendix I), and the injured subjects were scheduled for fitting of the DAFO by a Tripler Army Medical Center (TAMC) Physical Medicine and Rehabilitation physician. Prior to DAFO fitting, subjects received a physical therapy consult for treatment related to the device. Physical therapy treatment included the same generalized range of motion, flexibility, strength, gait, and proprioceptive training for all study participants. The same physical therapist measured range of motion, muscle strength, and leg segment girth on the involved limb only. Measurements were repeated three times, and the average of the three trials was recorded on the "Anthropometric Data Collection Sheet" (Appendix I). Joint range of motion was measured with a universal goniometer using procedures outlined by Daniels and Worthingham [45]. Muscular strength was measured using a Microfet 2 handheld dynamometer (Draper, Utah, USA) using procedures described by Kendall [46]. Leg segment girth was measured ten centimeters above the superior patellar pole and ten centimeters below the inferior patellar pole using a Gulick tape measure.

Injured subjects' quality-of-life and gait parameters were assessed prior to issue of the DAFO. These assessments involved one two-hour session per subject at the University of Hawaii Human Performance and Biomechanics Laboratory. Healthy service member control subjects' gait biomechanics were assessed during a single, one-hour session. All injured service members were asked to wear physical fitness attire during data collection, and to wear or bring their traditional AFO. Previous researchers have cited the appropriate baseline for determining the effectiveness of an AFO is to compare the device with walking gait in footwear, and concluded that footwear had a significant positive contribution to AFO function [26]. Further, research has shown that an individual's adaptive capability exceeds the variability of shoe type [47]. Therefore, subjects and controls wore their own running shoes for all gait biomechanics assessments.

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 I) and marked the subjects' lower extremities with retroreflective marker locations in preparation for kinematic measurements. Twenty-seven (27) retroreflective markers were applied to each subject (list and photo in Appendix I), and each subject was 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 3-D 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 were used to collect kinetic data during walking and running trials. Kinematic data were collected at 240Hz and time synchronized with digital video collected at 60Hz and kinetic data collected at 480Hz and smoothed using a Butterworth filter with an 8 Hz cut-off [48, 49]. 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, which was defined as each service member's maximum comfortable self-selected velocity for the initial trial ± 20%.

Gait biomechanics were assessed as the mean of three successful trials for each foot in each condition (shoes alone and TAFO). 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 [50, 51]. 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 [50-52]. Subjects repeated this procedure for running trials in shoes alone and wearing their TAFO. Service member control subjects, matched by age, height and weight to each injured service member, completed a one-time walking and running gait assessment in shoes to serve as normative values of healthy, uninjured service members. Gait analysis procedures were conducted within two weeks of DAFO issue and repeated monthly for six months with service members wearing their DAFO. At three and six months, injured service members also completed gait analysis in shoes alone to examine whether use of the DAFO affected gait parameters when not wearing the brace.

Following each gait analysis session, injured service members completed the SF-36 health-related quality-of-life survey (Appendix I), which quantified eight health topics: physical limitations, social limitations, physical work or role limitations, bodily pain, mental health, mental work or role limitations, vitality, and general health perceptions [53]. Subjects also answered a brief questionnaire regarding the use of their TAFO during initial data collection. Injured service members also repeated the SF-36 questionnaire at the monthly data collection sessions. The anthropometric measures were repeated at six months as standard care for drop-foot patients. All testing procedures are outlined in Table 2.1.

Table 2.1. Procedural Timetable for Six-Month Protocol

Procedure	Period (days)		Follow-up (months)					
	1-28 days before brace issue	1 to 14 days after brace issue	1	2	3	4	5	6
Medical History	X							
Physical Exam	X							X
Anthropometrics	X							X
Gait Analysis: Controls								X
Gait Analysis: Shoes Only	X				X			X
Gait Analysis: TAFO	X							
Gait Analysis: DAFO		X	X	X	X	X	X	X
SF-36 Version 2	X		X	X	X	X	X	X

Data Analyses

All data analyses were completed using SPSS (IBM version 19) with an alpha level set a $p<0.05$. Non-parametric Friedman's Tests were used to compare biomechanical gait differences in injured service member group at study onset (no brace, TAFO and naïve to the DAFO brace at study onset). Separate Mann-Whitney *U* Tests were used to compare biomechanical gait differences between healthy service member controls and limb salvage service members in the no brace condition, the traditional AFO, the DAFO at study onset, and the DAFO at study completion. Separate Wilcoxon Signed-Ranks Tests were used to compare the differences between the no brace condition and DAFO parameters at study completion to assess overall improvements in gait biomechanics among injured service members over the study period. Additional Friedman's Tests with post hoc Wilcoxon Signed-Ranks Tests and Bonferroni corrections (p =0.017) were completed on the SF-36 quality-of-life questionnaires to ascertain whether there was a significant change in responses over time for each category of responses, and for each individual question over time. Separate Wilcoxon Signed-Ranks Tests assessed differences in anthropometric range of motion and strength measurements prior to and following the six month study period.

CHAPTER III - SIX-MONTH PROTOCOL RESULTS

Participants

Descriptive data for the injured service members and healthy age and anthropometrically-matched service member control group are presented in Table 3.1. The injured service member group consisted of three Army soldiers, one Marine, one Navy seaman, and one Airman. Only descriptive and gait analysis data were collected on the healthy service member control group, which consisted of one Army soldier, one Navy seaman, three Airmen, and one Army Reserve Officer Training Corps cadet. 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 suffered a variety of injuries leading to the common pathology of partial lower extremity paralysis and subsequent drop foot. Two of the six soldiers suffered gunshot wounds, one to the back of the knee, and one to the upper thigh. One Army physician suffered a stroke, the Marine was injured by an improvised explosive device (IED), the 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 service members had just completed surgeries at study onset, and two were scheduled for additional surgeries related to their lower extremity injuries upon study completion. All participants met the inclusion criteria of permanent partial lower extremity neuropathy, had been using a TAFO for longer than one month, and could walk continuously for ten minutes. Case study service members' demographics are discussed individually in Chapter V.

Table 3.1. Descriptive Data for All Service Members

	Injured Mean (SD)	Control Mean (SD)
N	6	6
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)

Quality-of-Life

The Short-Form 36 (SF-36) quality-of-life questionnaires were scored into eight health domains and psychometrically-based physical (PCS) and mental component summaries (MCS). Group means were averaged for each time period: prior to DAFO issue and monthly for six months, totaling seven SF-36 questionnaires administered to each injured service member. Friedman's Test was used to determine differences in the repeated measures for each health domain for injured service members who completed the entire six-month protocol, and results are presented in Table 3.2.

Table 3.2. Short-Form 36 Quality-of-Life Questionnaire Health Domains

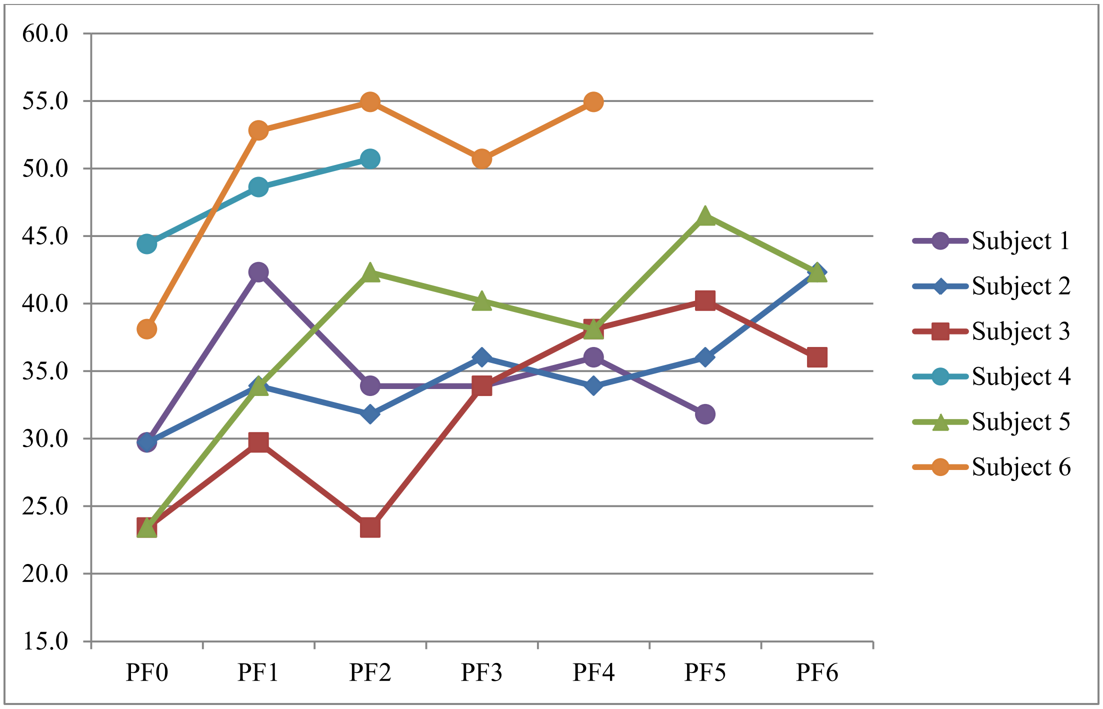
Health Domain	χ^2	*df*	*p*
Physical Functioning (PF)	13.35	6	0.01*
Role-Physical (RP)	1.41	6	0.98
Bodily Pain (BP)	5.04	6	0.61
General Health (GH)	2.16	6	0.93
Vitality (VT)	4.91	6	0.61
Social Functioning (SF)	5.45	6	0.54
Role-Emotional (RE)	1.88	6	0.97
Mental Health (MH)	6.32	6	0.43

**Significant at $p \leq 0.05$*

Scores from the Physical Functioning (PF) health domain, which measured difficulty in accomplishing physical tasks of daily living, improved significantly over time ($\chi^2(6) = 13.35$, p=0.01) (Figure 3.1). An additional Friedman's Test conducted on the individual

36 questions revealed significant improvements in climbing a single flight of stairs ($\chi^2(6) = 12.14$, p=0.03), which contributed to the improved physical functioning health domain.

Figure 3.1. Individual Injured Service Member Short-Form 36 Physical Functioning (PF) Domain over Study Period



Anthropometrics

Intra-rater reliability testing of the strength scores determined an intraclass correlation (ICC) range of 0.94 – 0.99 (ICC 3, k) with a standard error of the mean (SEM) range of 0.50 – 2.20 pounds of force (2.63 – 12.11% SEM) during manual muscle testing with the Microfet2 Hand-held Dynamometer. The Wilcoxon Signed-Rank test, a non-parametric equivalent of the dependent samples *t*-test, was used to compare anthropometrics (ROM, strength, and girth measurements) prior to and following six months of DAFO use. Range of motion measurements did not significantly change following use of the DAFO, and despite marked improvements in all strength measures, no significant differences existed in any pre and post-study measurements; however, hip strength in all three planes and knee extension strength improved at a level approaching significance (p=0.06) in spite of the limited sample size in the present study. Table 3.3 includes the median, range, Z score, p-value and effect size of the pre and post-anthropometric measurements.

Walking Gait Biomechanics

Spatial-temporal (ST) gait parameters, kinematics and kinetics of the hip, knee and ankle were measured in the no brace, TAFO and DAFO conditions. Spatial-temporal parameters included velocity, step and stride lengths, stance and swing times of the involved and sound limbs, and comparisons of symmetry. Over 150 biomechanical gait variables about the ankle, knee, hip, pelvis, and thorax were included in the analysis. Gait variables were measured in the no brace condition at study onset, and at three and six months following issue of the DAFO; all participants completed gait analysis in the no brace walking condition. Five out of six injured service members were able to walk in the TAFO condition. Traditional AFO (TAFO) measurements were considered only at study onset for comparison against the no brace and DAFO conditions, as most service members discarded the TAFO due to lack of function and comfort. One service member did not test in the TAFO condition due to a deep vein thrombosis attributed to TAFO use.

Table 3.4 contains the results of the Mann-Whitney U test for walking ST parameters, including means, standard deviations (SD), U test statistic, Z score, p-value, and effect size. This test was used to compare initial walking ST parameters of the injured service members in the no brace condition to controls at study onset.

Table 3.3. Pre and Post-Study Anthropometric Measurements

Anthropometric Measure	Pre Median (Range)	Post Median (Range)	*Z*	*p*	Effect Size
Hip Flexion ROM (°)	120.33 (107.00 - 126.67)	120.00 (90.00 - 125.67)	-0.14	1.00	-0.06
Hip Extension ROM (°)	10.00 (5.33 - 15.00)	12.33 (7.00 - 14.00)	-1.48	0.19	-0.66
Knee Flexion ROM (°)	129.33 (74.67 - 144.00)	135.67 (78.33 - 147.33)	-1.21	0.31	-0.54
Knee Extension ROM (°)	0.00 (-2.00 - 0.00)	0.00 (0.00 - 0.00)	-1.34	0.50	-0.60
Ankle Dorsiflexion ROM (°)	0.67 (-5.33 - 1.33)	0.00 (-5.00 - 1.00)	-0.14	1.00	-0.06
Ankle Plantarflexion ROM (°)	44.67 (42.33 - 45.33)	48.33 (36.67 - 56.00)	-0.94	0.44	-0.42
Calcaneal Inversion ROM (°)	25.67 (15.00 - 30.00)	24.333 (14.333 - 35.667)	-0.14	1.00	-0.06
Calcaneal Eversion ROM (°)	20.00 (15.67 - 23.00)	19.00 (10.67 - 22.00)	-0.14	1.00	-0.06
Hip Flexion Strength (lbf)	39.33 (29.33 - 49.00)	46.83 (44.17 - 58.67)	-2.02	0.06	-0.91
Hip Extension Strength (lbf)	30.00 (18.87 - 58.67)	38.00 (33.33 - 76.00)	-2.02	0.06	-0.91
Hip Abduction Strength (lbf)	39.83 (25.83 - 50.67)	56.67 (30.33 - 66.50)	-2.02	0.06	-0.91
Hip Adduction Strength (lbf)	30.50 (21.50 - 50.83)	46.67 (26.00 - 75.67)	-1.75	0.13	-0.78
Knee Flexion Strength (lbf)	20.70 (11.70 - 36.67)	25.17 (20.00 - 58.00)	-1.75	0.13	-0.78
Knee Extension Strength (lbf)	27.83 (20.40 - 43.00)	37.00 (29.67 - 59.00)	-2.02	0.06	-0.91
Ankle Dorsiflexion Strength (lbf)	0.00 (0.00 - 10.17)	9.00 (0.00 - 26.33)	-0.73	0.63	-0.33
Ankle Plantarflexion Strength (lbf)	29.83 (1.93 - 38.93)	42.17 (0.00 - 54.50)	-1.21	0.31	-0.54
Ankle Inversion Strength (lbf)	8.80 (0.00 - 15.27)	23.33 (0.00 - 34.67)	-1.83	0.13	-0.82
Ankle Eversion Strength (lbf)	4.20 (0.00 - 15.70)	20.33 (0.00 - 45.33)	-1.83	0.13	-0.82
Leg Girth - Above Patella (cm)	44.27 (39.63 - 53.60)	44.33 (40.47 - 56.93)	-1.21	0.31	-0.54
Leg Girth - Below Patella (cm)	38.00 (28.77 - 38.93)	36.97 (28.37 - 39.67)	-0.67	0.63	-0.30

lbf = pounds of force production

Walking swing time of the involved limb, step length of the involved and sound limbs, stride length and velocity were significantly different between the injured service members and controls.

Donning the TAFO improved involved limb swing time and walking velocity, but step length of both limbs and stride length remained significantly shorter than controls (Table 3.5). Following DAFO issue but while still naïve to the brace, walking ST parameters were also compared to controls by the Mann-Whitney *U* Test and are reported in Table 3.6. Protocol completion comparisons between DAFO at study completion and controls are found in Table 3.7. There were no significant differences in walking ST parameters between injured service members and controls, either naïve to DAFO use or following six months of DAFO use.

Table 3.4. Walking Spatial-Temporal Gait Parameters for No Brace versus Controls at Study Onset, Mann-Whitney *U* Test, n=12

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	12	0.41 (0.37 - 0.45)	5.00	-2.08	0.04*	-0.60
Sound Limb Swing Time (s)	12	0.39 (0.34 – 0.42)	12.00	-0.96	0.39	-0.28
Involved Limb Stance Time (s)	12	0.58 (0.51 – 0.67)	13.00	-0.80	0.49	-0.23
Sound Limb Stance Time (s)	12	0.62 (0.52 – 0.74)	8.00	-1.60	0.13	-0.46
Involved Limb Step Length (m)	12	0.83 (0.77 – 0.96)	0.00	-2.88	0.002*	-0.83
Sound Limb Step Length (m)	12	0.81 (0.66 – 0.94)	0.00	-2.88	0.002*	-0.83
Stride Length (m)	12	1.64 (1.47 – 1.89)	0.00	-2.88	0.002*	-0.83
Velocity (m/s)	12	1.62 (1.32 – 2.13)	4.00	-2.24	0.03*	-0.65

**Significant at $p \leq 0.05$*

Table 3.5. Walking Spatial-Temporal Gait Parameters for TAFO versus Controls at Study Onset, Mann-Whitney *U* Test, n=10

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	10	0.41 (0.36 – 0.42)	10.50	-0.42	0.73	-0.13
Sound Limb Swing Time (s)	10	0.39 (0.33 – 0.42)	7.00	-1.15	0.31	-0.36
Involved Limb Stance Time (s)	10	0.61 (0.51 – 0.82)	6.00	-1.36	0.22	-0.43
Sound Limb Stance Time (s)	10	0.63 (0.52 – 0.84)	5.00	-1.57	0.15	-0.50
Involved Limb Step Length (m)	10	0.85 (0.73 – 0.96)	2.00	-2.19	0.03*	-0.69
Sound Limb Step Length (m)	10	0.82 (0.73 – 0.94)	1.00	-2.40	0.02*	-0.76
Stride Length (m)	10	1.65 (1.46 – 1.89)	2.00	-2.19	0.03*	-0.69
Velocity (m/s)	10	1.61 (1.22 – 2.13)	4.00	-1.78	0.10	-0.56

**Significant at $p \leq 0.05$*

Table 3.6. Walking Spatial-Temporal Gait Parameters for DAFO versus Controls at Study Onset, Mann-Whitney *U* Test, n=12

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	12	0.41 (0.37 – 0.46)	7.500	-1.69	0.10	-0.49
Sound Limb Swing Time (s)	12	0.39 (0.36 – 0.42)	16.00	-0.32	0.82	-0.09
Involved Limb Stance Time (s)	12	0.57 (0.51 – 0.67)	14.00	-0.64	0.59	-0.19
Sound Limb Stance Time (s)	12	0.59 (0.52 – 0.74)	9.00	-1.44	0.18	-0.42
Involved Limb Step Length (m)	12	0.92 (0.68 – 1.05)	14.00	-0.64	0.59	-0.19
Sound Limb Step Length (m)	12	0.91 (0.67 – 1.00)	15.00	-0.48	0.70	-0.14
Stride Length (m)	12	1.84 (1.35 – 2.05)	14.00	-0.64	0.59	-0.19
Velocity (m/s)	12	1.91 (1.19 – 2.13)	17.00	-0.16	0.94	-0.05

Significant at $p \leq 0.05$

Table 3.7. Walking Spatial-Temporal Gait Parameters for DAFO versus Controls at Study Completion, Mann-Whitney *U* Test, n=8

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	8	0.39 (0.32 – 0.43)	8.00	0.00	1.00	0.00
Sound Limb Swing Time (s)	8	0.37 (0.34 – 0.41)	5.00	-0.87	0.49	-0.31
Involved Limb Stance Time (s)	8	0.57 (0.50 – 0.63)	8.00	0.00	1.00	0.00
Sound Limb Stance Time (s)	8	0.60 (0.52 – 0.67)	3.50	-1.31	0.23	-0.46
Involved Limb Step Length (m)	8	0.92 (0.83 – 1.01)	4.00	-1.16	0.34	-0.41
Sound Limb Step Length (m)	8	0.99 (0.82 – 1.00)	5.00	-1.16	0.34	-0.41
Stride Length (m)	8	1.83 (1.65 – 2.01)	4.00	-1.16	0.34	-0.41
Velocity (m/s)	8	1.92 (1.60 – 2.23)	7.00	-0.29	0.89	-0.10

A Wilcoxon Signed-Ranks Test was used to assess differences in walking ST parameters between the no brace condition and following six months of DAFO use (Table 3.8). Significance values were reported via one-tailed approach based on the hypothesis that parameters would improve in the DAFO; all parameters approached significance despite a small sample of injured service members who had completed the six month protocol. Friedman's Test, a non-parametric repeated-measures analysis of variance (ANOVA), was used to determine changes in walking ST parameters between the three conditions in the injured service member group at study onset; results for the no brace, TAFO, and DAFO conditions are included in Table 3.9. Most walking ST parameters did not significantly change over the three conditions. Post hoc Wilcoxon signed-ranks (WSR) tests were used to follow up the significant main effect of sound limb swing time with a Bonferroni correction. Sound limb swing time exhibited the greatest change between the no brace and DAFO conditions at study onset.

Table 3.8. Walking Spatial-Temporal Gait Parameters for No Brace versus DAFO at Study Completion, Wilcoxon Signed-Ranks Test, n=4

	N	No Brace Median (Range)	DAFO Median (Range)	Z	*p*	Effect Size
Involved Limb Swing Time (s)	4	0.44 (0.42 – 0.45)	0.39 (0.32 – 0.43)	-1.83	0.06	-0.65
Sound Limb Swing Time (s)	4	0.40 (0.35 – 0.41)	0.37 (0.34 – 0.40)	-1.83	0.06	-0.65
Involved Limb Stance Time (s)	4	0.65 (0.57 – 0.66)	0.57 (0.50 – 0.61)	-1.83	0.06	-0.65
Sound Limb Stance Time (s)	4	0.71 (0.60 – 0.74)	0.62 (0.56 – 0.67)	-1.83	0.06	-0.65
Involved Limb Step Length (m)	4	0.80 (0.78 – 0.82)	0.95 (0.86 – 1.01)	-1.83	0.06	-0.65
Sound Limb Step Length (m)	4	0.72 (0.66 – 0.81)	0.91 (0.84 – 1.00)	-1.83	0.06	-0.65
Stride Length (m)	4	1.51 (1.47 – 1.63)	1.87 (1.70 – 2.01)	-1.83	0.06	-0.65
Velocity (m/s)	4	1.39 (1.32 – 1.63)	1.90 (1.72 – 2.30)	-1.83	0.06	-0.65

Significant at $p \leq 0.05$ based on exact test (one-tailed); Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

The means reported for each condition also demonstrated that involved and sound limb step lengths, stride length, and velocity improve with immediate use of the DAFO over the TAFO, without previous practice using the dynamic device.

Analysis of the kinematic and kinetic gait variables began with a comparison between the injured service member group and the matched healthy service member controls, to determine significant gait deviations seen without an AFO at study onset. Table 3.10 includes twelve significant gait deviations about the ankle and knee during walking. Friedman's Tests were used to determine how these deviations changed between no brace, TAFO, and DAFO walking gait at study onset (Table 3.10). Two gait variables involving the ankle were significantly different between the three conditions at study onset (p=0.008). Wilcoxon signed-ranks tests were used to follow-up significant main effects; a Bonferroni correction as applied, adjusting the alpha level to 0.017.

Dorsiflexion excursion (*T*=0, *p*=0.008, *r*=-0.90*)* and mean ankle dorsiflexion velocity (*T*=0, *p*=0.008, *r*=-0.90*)* were significantly lower in the naïve DAFO condition compared to no AFO.

Table 3.9. Walking Spatial-Temporal Gait Parameters for No Brace vs. TAFO vs. DAFO at Study Onset, Friedman's Test, n=5

	DB Median (Range)	TAFO Median (Range)	DAFO Median (Range)	χ^2	*df*	*p*
Involved Limb Swing Time (s)	0.42 (0.38 – 0.45)	0.42 (0.36 – 0.42)	0.41 (0.38 – 0.46)	2.80	2	0.37
Sound Limb Swing Time (s)	0.36 (0.34 – 0.41)	0.39 (0.33 – 0.41)	0.39 (0.37 – 0.42)	8.40	2	0.01*
Involved Limb Stance Time (s)	0.65 (0.54 – 0.66)	0.68 (0.54 – 0.82)	0.58 (0.57 – 0.67)	1.20	2	0.69
Sound Limb Stance Time (s)	0.71 (0.57 – 0.74)	0.71 (0.57 – 0.84)	0.61 (0.58 – 0.74)	0.40	2	0.95
Involved Limb Step Length (m)	0.79 (0.77 – 0.81)	0.76 (0.73 – 0.93)	0.91 (0.68 – 1.05)	2.80	2	0.37
Sound Limb Step Length (m)	0.70 (0.66 – 0.76)	0.76 (0.73 – 0.83)	0.90 (0.67 – 1.00)	3.60	2	0.18
Stride Length (m)	1.47 (1.47 – 1.57)	1.52 (1.46 – 1.75)	1.81 (1.35 – 2.05)	2.80	2	0.37
Velocity (m/s)	1.43 (1.32 – 1.72)	1.44 (1.22 – 1.81)	1.77 (1.19 – 2.10)	2.80	2	0.37

**Significant at $p \leq 0.05$ for main effect; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)*

An overall summary of significantly different variables during walking between the no brace condition, TAFO condition, DAFO at study onset and completion, and control group is represented in Table 3.11. Twenty-two gait variables were significantly different between the injured service members and controls while wearing an AFO, regardless of type. The variables measured wearing the TAFO accounted for 68% (15/22 variables) of these differences. After donning the DAFO for the first time, an immediate improvement of 13% was seen in gait variables. Although some variables improved from TAFO use, other new gait deviations appeared while initially wearing the DAFO (naïve). After six-months of DAFO use, six of the original 22 variables were significantly different than controls subjects, improving 41% over TAFO use. These remaining gait deviations variables during walking were related to restriction at the ankle based on DAFO design.

Table 3.10. Significant (*p≤0.05*) Walking Kinematic and Kinetic Gait Variables for No Brace vs. Controls at Study Onset, Mann-Whitney *U* Test, n=12

	Median (Range)	*U*	*Z*	*p*	Effect Size
Ankle Position, Initial Contact (IC) (°)	0.46 (-24.93 – 12.44)	0.00	-2.88	0.002	-0.83
Dorsiflexion (DF) Excursion (°)	13.48 (0.98 – 45.92)	0.00	-2.88	0.002	-0.83
Max Ankle DF Velocity (m/s)	132.91 (88.17 – 314.73)	5.00	-2.08	0.04	-0.60
Mean Ankle DF Velocity (m/s)	30.90 (-26.07 – 83.63)	0.00	-2.90	0.002	-0.84
Max Ankle Internal Rotation (IR) (°)	6.70 (-23.54 – 29.00)	5.00	-2.08	0.04	-0.60
Mean Knee Varus Velocity (m/s)	23.34 (3.41 – 54.07)	2.00	-2.56	0.01	-0.74
Thorax Angle Excursion (°)	6.10 (2.18 – 8.25)	5.00	-2.08	0.04	-0.60
Max Vertical Ground Reaction Force (GRF) (Nm)	10.60 (9.98 – 13.79)	3.00	-2.40	0.02	-0.69
Max Propulsive GRF (N/kg)	2.52 (1.44 – 3.25)	3.00	-2.40	0.02	-0.69
Timing of Max Knee Extension (EXT) Moment (% Stance)	8.17 (3.37 – 99.03)	3.00	-2.40	0.02	-0.69
Timing of Max Knee Adduction (ADD) Moment (% Stance)	23.58 (18.31 – 98.38)	1.00	-2.72	0.004	-0.79
Max Knee IR Moment (Nm/kg)	0.20 (0.04 – 0.28)	4.00	-2.24	0.03	-0.65
Ankle Power: Generation (A2) (W/kg)	5.74 (1.18 – 7.66)	4.00	-2.24	0.03	-0.65
Knee Power: Absorption during Loading Response (K1) (W/kg)	-2.28 (-4.65 – (-0.13))	4.00	-2.24	0.03	-0.65
Hip Power: Generation during Loading Response (H1) (W/kg)	0.76 (0.34 – 2.16)	5.00	-2.08	0.04	-0.60

Significant at p≤0.05; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Table 3.11. Walking Kinematic and Kinetic Gait Variables for No Brace vs. TAFO vs. DAFO at Study Onset, Friedman's Test with post hoc Wilcoxon Signed-Ranks Test, n=5

	NB Median (Range)	TAFO Median (Range)	DAFO Median (Range)	χ^2	*df*	*p*
Ankle Position, IC (°)	-8.23 (-24.93 - (-1.88))	0.45 (-6.25 – 7.91)	6.64 (1.97 – 7.56)	6.40	2	0.04
Dorsiflexion Excursion (°)	32.60 (15.36 – 45.92)	20.70 (11.03 – 27.69)	6.94 (3.55 – 7.51)	8.40	2	0.008*
Max Ankle DF Velocity (m/s)	148.07 (121.21 – 312.48)	148.50 (71.57 – 1339.47)	53.61 (35.08 – 75.71)	7.60	2	0.02
Mean Ankle DF Velocity (m/s)	62.30 (33.48 – 83.63)	33.10 (17.97 – 60.34)	12.87 (7.06 – 15.77)	8.40	2	0.008*
Max Ankle IR (°)	16.87 (-4.21 – 29.00)	23.01 (-168.21 – 31.48)	8.26 (-3.64 – 26.93)	0.40	2	0.95
Mean Knee Varus Velocity (m/s)	34.71 (22.22 – 54.07)	25.96 (7.69 – 43.71)	19.21 (-19.35 – 35.59)	1.20	2	0.69
Thorax Angle Excursion (°)	6.53 (5.04 – 8.25)	7.73 (4.75 – 10.23)	5.51 (4.97 – 6.84)	2.80	2	0.37
Max Vertical GRF (N/kg)	10.25 (9.98 – 12.33)	10.89 (10.03 – 11.72)	11.74 (9.85 – 13.47)	2.80	2	0.37
Max Propulsive GRF (N/kg)	1.98 (1.44 – 2.83)	1.65 (0.74 – 2.09)	1.86 (1.31 – 2.84)	4.80	2	0.12
Timing of Max Knee EXT Moment (% Stance)	57.33 (7.57 – 99.03)	6.95 (3.39 – 97.93)	49.03 (5.62 – 76.46)	1.20	2	0.69
Timing of Max Knee ADD Moment (% Stance)	24.15 (23.36 – 98.38)	27.90 (23.73 – 97.34)	23.25 (19.90 – 28.14)	5.20	2	0.09
Max Knee IR Moment (Nm/kg)	0.16 (0.04 – 0.21)	0.14 (0.10 – 1.91)	0.09 (0.07 – 0.18)	2.80	2	0.37
Ankle Power: Generation (A2) (W/kg)	4.47 (1.18 – 7.24)	2.79 (1.18 – 6.05)	1.70 (0.77 – 3.12)	2.80	2	0.37
Knee Power: Absorption during Loading Response (K1) (W/kg)	-0.34 (-2.98 – (-0.13))	-0.30 (-0.92 – (-0.17))	-0.81 (-1.61 – (-0.08))	1.20	2	0.69
Hip Power: Generation during Loading Response (H1) (W/kg)	1.11 (0.40 – 2.16)	0.98 (0.72 – 1.73)	2.11 (0.55 – 2.29)	3.60	2	0.18

Significant at p≤0.017 with Bonferroni Correction; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Table 3.12. Comparison of Significant (*p≤0.05*) Walking Kinetic and Kinematic Variables for TAFO, DAFO at Study Onset and DAFO at Study Completion; compared to Controls

	No Brace		TAFO		DAFO Time 0		DAFO Time 6		Controls
	Mean (SD)	***p***	**Mean (SD)**	***p***	**Mean (SD)**	***p***	**Mean (SD)**	***p***	**Mean (SD)**
Ankle Position, IC (°)	-10.39 (8.39)	0.002*	1.71 (5.59)	0.10	6.78 (3.18)	0.82	4.98 (1.81)	0.49	7.46 (3.75)
Max Ankle DF (°)	19.44 (4.35)	0.13	21.99 (3.66)	0.03*	12.63 (3.69)	0.24	13.71 (2.67)	0.49	15.06 (2.84)
Time to Max Ankle DF (% Stance)	77.75 (4.97)	0.07	82.28 (3.72)	0.008*	77.80 (4.30)	0.04*	68.64 (18.98)	0.06	64.60 (20.81)
Max Ankle Plantarflexion (PF) (°)	-20.85 (6.76)	0.18	-8.50 (5.99)	0.008*	3.24 (2.49)	0.002*	-11.25 (13.39)	0.03*	-24.23 (2.49)
Time to Max Ankle PF (% Stance)	68.22 (49.30)	0.70	12.51 (2.74)	0.008*	26.17 (36.34)	0.02*	62.63 (44.06)	0.03*	99.70 (0.64)
DF Excursion (°)	29.83 (10.36)	0.002*	20.29 (7.34)	0.008*	5.85 (1.64)	0.09	7.41 (3.24)	1.00	7.60 (3.61)
Ankle Position at Toe-Off (°)	-15.55 (11.36)	0.09	-2.18 (7.99)	0.008*	7.13 (2.90)	0.002*	-9.29 (15.170)	0.03*	-24.12 (2.40)
Max Ankle DF Velocity (m/s)	200.74 (89.52)	0.04*	376.64 (541.70)	0.55	53.25 (17.16)	0.002*	97.04 (34.61)	0.03*	116.16 (24.58)
Mean Ankle DF Velocity (m/s)	61.35 (19.40)	0.002*	37.25 (19.22)	0.008*	11.84 (3.17)	0.13	12.48 (16.39)	0.69	11.21 (18.92)
Max Ankle PF Velocity (m/s)	-699.35 (514.86)	0.31	-1192.70 (1993.40)	0.55	-109.87 (31.04)	0.002*	-292.69 (174.71)	0.03*	-464.90 (41.03)
Ankle Eversion (EV) Excursion (°)	28.40 (12.949)	0.70	23.55 (20.76)	0.42	10.89 (9.40)	0.04*	17.02 (9.23)	0.06	23.61 (5.73)
Max Ankle IR (°)	15.82 (12.36)	0.04*	-16.15 (85.46)	0.22	11.21 (11.33)	0.13	7.58 (12.14)	0.34	-1.85 (13.07)
Mean Knee Varus Velocity (m/s)	37.59 (13.08)	0.009*	26.10 (13.66)	0.22	14.08 (20.48)	0.94	35.00 (38.23)	0.89	14.90 (9.73)
Knee IR Excursion (°)	14.39 (5.33)	0.59	16.82 (14.97)	0.31	4.59 (2.28)	0.002*	11.60 (8.27)	0.49	17.46 (7.01)
Max Knee IR Moment (Nm/kg)	0.14 (0.06)	0.03*	0.49 (0.80)	0.31	0.14 (0.06)	0.03*	0.17 (0.08)	0.11	0.23 (0.50)
Thorax Angle Excursion (°)	6.54 (1.07)	0.04*	7.53 (1.99)	0.03*	5.97 (0.82)	0.18	6.30 (1.25)	0.49	4.57 (1.81)

Table 3.12 (Continued)	**No Brace**		**TAFO**		**DAFO Time 0**		**DAFO Time 6**		**Controls**
	Mean (SD)	***p***	**Mean (SD)**	***p***	**Mean (SD)**	***p***	**Mean (SD)**	***p***	**Mean (SD)**
Max Vertical GRF (N/kg)	10.88 (0.91)	0.02*	10.86 (0.73)	0.03*	11.64 (1.18)	0.24	13.08 (1.43)	0.89	12.58 (0.92)
Max Braking GRF (N/kg)	-2.34 (0.59)	0.13	-1.85 (0.43)	0.02*	-2.03 (0.67)	0.04*	-2.76 (0.63)	0.89	-2.93 (0.58)
Time to Max Braking GRF (% Stance)	16.28 (2.20)	0.39	23.49 (5.03)	0.03*	19.04 (3.62)	0.59	19.09 (6.23)	1.00	17.36 (1.09)
Max Propulsive GRF (N/kg)	2.12 (0.48)	0.02*	1.59 (0.54)	0.008*	1.86 (0.45)	0.002*	2.13 (0.50)	0.06	2.90 (0.35)
Max DF Moment (Nm/kg)	1.24 (0.41)	0.07	1.18 (0.38)	0.03*	1.36 (0.27)	0.13	1.56 (0.17)	0.49	1.61 (0.16)
Time to Max DF Moment (% Stance)	79.88 (2.60)	0.82	82.65 (5.40)	0.15	81.89 (2.01)	0.03*	81.94 (0.80)	0.03*	79.04 (1.17)
Time to Max Plantarflexion Moment (% Stance)	69.17 (47.77)	0.28	42.70 (40.53)	0.008*	14.70 (2.58)	0.004*	12.77 (1.35)	0.06	10.85 (1.31)
Ankle Power: Generation (A2) (W/kg)	4.34 (2.00)	0.03*	3.08 (1.84)	0.02*	1.77 (0.81)	0.002*	2.70 (0.61)	0.03*	6.63 (0.84)
Timing of Max Knee EXT Moment (% Stance)	43.46 (38.10)	0.02*	24.49 (41.10)	0.31	41.52 (30.22)	0.03*	22.49 (35.22)	0.69	10.92 (16.10)
Max Knee ADD Moment (Nm/kg)	0.74 (0.18)	0.13	2.77 (4.90)	0.31	0.66 (0.24)	0.04*	0.85 (0.48)	0.49	1.10 (0.34)
Time to Max Knee ADD Moment (% Stance)	41.90 (30.27)	0.004*	43.24 (30.81)	0.02*	23.62 (2.97)	0.18	22.12 (2.15)	0.49	21.17 (1.86)
Knee Power: Absorption during Loading Response (K1) (W/kg)	-1.05 (1.17)	0.03*	-0.48 (0.34)	0.004*	-0.85 (0.63)	0.004*	-2.10 (1.57)	0.34	-3.35 (1.29)
Knee Power: Generation during midstance (K2) (W/kg)	1.03 (0.94)	0.07	0.59 (0.28)	0.009*	1.38 (0.84)	0.18	2.50 (1.83)	0.89	2.25 (1.04)
Max Hip IR Moment (Nm/kg)	0.57 (1.02)	0.70	0.33 (0.45)	0.22	0.13 (0.05)	0.02*	0.17 (0.07)	0.49	0.20 (0.04)
Hip Power: Generation during Loading Response (H1) (W/kg)	1.25 (0.63)	0.04*	1.18 (0.44)	0.02*	1.86 (0.81)	0.02*	1.69 (0.53)	0.03*	0.61 (0.17)

Significant at p≤0.05; Shaded variables represents significant variables for each group versus control subjects; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Running Gait Biomechanics

Running spatial-temporal (ST) gait parameters, kinematics and kinetics of the hip, knee and ankle were measured in the same brace conditions previously reported while walking. Table 3.13 contains the results of the Mann-Whitney *U* test for running ST parameters, including means, SD, *U* test statistic, *Z* score, *p*-value, and effect size. This test compared initial running ST parameters of the injured group in the no brace condition to controls at study onset. Significant differences in the no brace running condition included decreased swing time and step length in the sound limb, as well as decreases in total stride length and running velocity. Donning a TAFO improved involved limb swing time, but significantly shorter step and stride lengths remained (Table 3.14). While running with a TAFO, stride length and velocity were slower than the no AFO condition, possibly due to pain levels reported by five of the injured service members. Following DAFO issue but still naïve to the brace, running ST parameters were compared to controls by the Mann-Whitney *U* Test; differences in sound limb swing and stance times were found between the DAFO and controls (Table 3.15). After six months of DAFO use, sound limb swing time improved to normative levels; only sound limb stance time was significantly increased from controls (Table 3.16).

Friedman's Test was used to determine changes in running ST parameters between the three conditions at study onset (Table 3.17), and revealed none of the running ST parameters differed over the three conditions in the three service members able to complete all conditions at study onset. Pain and post-surgical requirements precluded half of the service members from running in one or more of the conditions at study onset, requiring descriptive versus inferential statistical measures. Notable improvements in stride length and velocity can be seen from the means of the three service members able to complete the running condition.

Table 3.13. Running Spatial-Temporal Gait Parameters for No Brace versus Controls at Study Onset, Mann-Whitney *U* Test, n=8, 10, 11

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	8	0.49 (0.39 – 0.53)	5.00	-0.75	0.57	-0.26
Sound Limb Swing Time (s)	10	0.47 (0.39 – 0.53)	0.50	-2.47	0.01*	-0.78
Involved Limb Stance Time (s)	11	0.25 (0.22 – 0.59)	8.00	-1.28	0.25	-0.39
Sound Limb Stance Time (s)	11	0.26 (0.22 – 0.58)	7.00	-1.46	0.18	-0.44
Involved Limb Step Length (m)	10	1.43 (0.93 – 1.49)	4.00	-1.71	0.11	-0.54
Sound Limb Step Length (m)	10	1.36 (0.83 – 1.55)	0.00	-2.56	0.01*	-0.81
Stride Length (m)	10	2.79 (1.76 – 3.04)	0.00	-2.56	0.01*	-0.81
Velocity (m/s)	10	3.88 (2.78 – 4.12)	0.00	-2.56	0.01*	-0.81

**Significant at $p \leq 0.05$*

Table 3.14. Running Spatial-Temporal Gait Parameters for TAFO versus Controls at Study Onset, Mann-Whitney *U* Test, n=6, 8

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	6	0.49 (0.37 – 0.50)	1.00	-1.53	0.20	-0.62
Sound Limb Swing Time (s)	6	0.49 (0.32 – 0.53)	0.00	-1.88	0.13	-0.77
Involved Limb Stance Time (s)	8	0.26 (0.21 – 0.45)	5.00	-0.87	0.49	-0.31
Sound Limb Stance Time (s)	8	0.27 (0.21 – 0.45)	4.00	-1.16	0.34	-0.41
Involved Limb Step Length (m)	8	1.43 (0.99 – 1.67)	4.00	-1.16	0.34	-0.41
Sound Limb Step Length (m)	8	1.35 (0.79 – 1.55)	1.00	-2.02	0.06	-0.71
Stride Length (m)	8	2.71 (1.90 – 3.04)	0.00	-2.31	0.03*	-0.82
Velocity (m/s)	8	3.71 (2.10 – 4.12)	0.00	-2.31	0.03*	-0.82

**Significant at $p \leq 0.05$*

Stride length and velocity decreased when wearing the TAFO compared to shoes alone, and the highest disparity in symmetrical step length was noted in the TAFO condition. Unlike the walking condition, a Wilcoxon Signed-Ranks Test to assess differences in running ST parameters between the no brace condition and the DAFO at study completion could not be accomplished due to the combination of an inadequate number of subjects able to run at study onset and the subjects able to run and still enrolled in the study following six months.

Table 3.15. Running Spatial-Temporal Gait Parameters for DAFO versus Controls at Study Onset, Mann-Whitney *U* Test, n=8, 10

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	8	0.47 (0.40 – 0.52)	5.00	-0.89	0.51	-0.31
Sound Limb Swing Time (s)	10	0.47 (0.36 – 0.53)	2.00	-2.21	0.02*	-0.70
Involved Limb Stance Time (s)	10	0.24 (0.21 – 0.29)	10.00	-0.52	0.69	-0.17
Sound Limb Stance Time (s)	10	0.27 (0.25 – 0.34)	2.00	-2.19	0.03*	-0.69
Involved Limb Step Length (m)	10	1.42 (0.97 – 1.50)	5.00	-1.57	0.15	-0.50
Sound Limb Step Length (m)	10	1.39 (0.84 – 1.55)	6.00	-1.36	0.22	-0.43
Stride Length (m)	10	2.80 (1.81 – 3.04)	5.00	-1.57	0.15	-0.50
Velocity (m/s)	10	3.88 (2.55 – 4.24)	5.00	-1.57	0.15	-0.50

**Significant at $p \leq 0.05$*

Analysis of the running kinematic and kinetic gait variables again began with the same comparison between the injured service member group and the matched healthy service member controls as noted during walking. Few differences were seen at the ankle during running; most differences were kinetic and revolved around the knee and hip (Table 3.18). Friedman's Test was used to determine significant kinematic and kinetic gait differences between no brace, TAFO, and DAFO running gait at study onset (Table

3.19), and none of the running gait variables were significantly different between brace conditions.

Table 3.16. Running Spatial-Temporal Gait Parameters for DAFO versus Controls at Study Completion, Mann-Whitney *U* Test, n=6, 8

	N	Median (Range)	*U*	Z	*p*	Effect Size
Involved Limb Swing Time (s)	6	0.48 (0.46 – 0.50)	4.00	-0.22	1.00	-0.09
Sound Limb Swing Time (s)	6	0.48 (0.45 – 0.53)	1.00	-1.41	0.20	-0.58
Involved Limb Stance Time (s)	8	0.23 (0.22 – 0.26)	7.50	-0.15	0.94	-0.05
Sound Limb Stance Time (s)	8	0.26 (0.23 – 0.30)	0.00	-2.31	0.03*	-0.82
Involved Limb Step Length (m)	8	1.42 (1.07 – 1.49)	3.00	-1.44	0.20	-0.51
Sound Limb Step Length (m)	8	1.36 (1.18 – 1.55)	2.00	-1.73	0.11	-0.61
Stride Length (m)	8	2.78 (2.25 – 3.04)	1.00	-2.02	0.06	-0.72
Velocity (m/s)	8	3.95 (3.00 – 4.12)	1.50	-1.89	0.09	-0.67

**Significant at* $p \leq 0.05$

Table 3.17. Running Spatial-Temporal Gait Parameters for No Brace vs. TAFO vs. DAFO at Study Onset, Friedman's Test, n=3

	NB Median (Range)	TAFO Median (Range)	DAFO Median (Range)	χ^2	*df*	*p*
Involved Limb Swing Time (s)	0.45 (0.39 – 0.51)	0.44 (0.40 – 0.49)	0.43 (0.40 – 0.46)	0.00	2	1.00
Sound Limb Swing Time (s)	0.42 (0.39 – 0.45)	0.35 (0.32 – 0.38)	0.43 (0.36 – 0.49)	1.00	2	0.83
Involved Limb Stance Time (s)	0.26 (0.22 – 0.26)	0.27 (0.21 – 0.45)	0.24 (0.21 – 0.24)	4.67	2	0.19
Sound Limb Stance Time (s)	0.28 (0.22 – 0.30)	0.29 (0.21 – 0.45)	0.28 (0.26 – 0.30)	0.55	2	0.89
Involved Limb Step Length (m)	1.37 (0.93 – 1.45)	1.28 (0.99 – 1.67)	1.36 (0.97 – 1.50)	0.67	2	0.94
Sound Limb Step Length (m)	1.18 (0.83 – 1.29)	0.92 (0.79 – 1.37)	1.43 (0.84 – 1.44)	2.67	2	0.36
Stride Length (m)	2.63 (1.76 – 2.66)	2.46 (1.90 – 2.65)	2.79 (1.81 – 2.93)	2.00	2	0.53
Velocity (m/s)	3.40 (2.92 – 3.64)	3.25 (2.10 – 3.56)	3.76 (2.90 – 4.24)	0.67	2	0.94

An overall comparison of significantly different variables for running between the no brace, TAFO, and DAFO conditions at study onset and completion, and control subjects revealed a greater number of significant gait deviations compared to walking. A total of 54 kinematic and kinetic gait variables differed significantly between healthy controls and the injured service members while wearing an AFO, regardless of type (Tables 3.20 and 3.21). Over fifty percent (28/54 variables) of the total gait deviations were seen in the TAFO. When the injured soldiers first donned the DAFO, these differences increased to 70% (38/54 variables). After six months of DAFO use, however, differences had self-corrected back to 50% of gait variables.

Table 3.18. Significant Running Kinematic and Kinetic Gait Variables for No Brace vs. Controls at Study Onset, Mann-Whitney *U* Test, n=10

	Median (Range)	*U*	Z	*p*	Effect Size
Mean Foot Progress Angle During Stance (°)	5.10 (1.95 – 11.61)	1.00	-2.40	0.02	-0.76
Timing of Max Knee Flexion (% Stance)	37.63 (31.49 – 100.00)	1.00	-2.40	0.02	-0.76
Knee Position at Toe-Off (°)	17.15 (5.25 – 56.08)	0.00	-2.61	0.008	-0.83
Mean Knee Flexion Velocity (m/s)	269.57 (85.03 – 365.06)	1.00	-2.40	0.02	-0.76
Hip Angle Excursion - IC to Max EXT (°)	-48.89 (-54.49 – (-31.00))	0.00	-2.61	0.008	-0.83
Mean Hip EXT Velocity - IC to Max EXT (°)	-191.70 (-226.74 – (-78.04))	0.00	-2.61	0.008	-0.83
Max Braking GRF (N/kg)	-3.16 (-3.66 – 1.74)	0.00	-2.45	0.02	-0.78
Timing of Max DF Moment (% Stance)	58.43 (50.36 – 59.79)	0.00	-2.45	0.02	-0.78
Max Knee Flexion Moment at Loading Response (Nm/kg)	2.72 (-0.06 – 3.81)	0.00	-2.45	0.02	-0.78
Max Knee Flexion Moment at Toe-Off (Nm/kg)	2.72 (1.22 – 3.82)	0.00	-2.45	0.02	-0.78
Knee Stiffness (Nm/rad)	10.15 (0.68 – 20.89)	1.00	-2.21	0.03	-0.70
Max Knee ADD Moment (Nm/kg)	2.10 (0.71 – 2.52)	0.00	-2.45	0.02	-0.78
Timing of Max Hip IR Moment (% Stance)	75.76 (28.01 – 100.00)	1.00	-2.21	0.03	-0.70
Ankle Power: Generation (A2) (W/kg)	14.17 (11.16 – 20.90)	0.00	-2.31	0.03	-0.82
Knee Power: Absorption during Loading Response (K1) (W/kg)	-9.96 (-22.20 – (-2.66))	0.00	-2.31	0.03	-0.82

Significant at $p \leq 0.05$; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Table 3.19. Running Kinematic and Kinetic Gait Variables for No Brace vs. TAFO vs. DAFO at Study Onset, Friedman's Test with post hoc Wilcoxon Signed-Ranks Test, n=4

	NB Median (Range)	TAFO Median (Range)	DAFO Median (Range)	χ^2	*df*	*p*
Mean Foot Progress Angle During Stance (°)	3.18 (3.05 – 3.22)	2.71 (0.18 – 6.19)	1.64 (0.43 – 4.64)	1.50	2	0.65
Timing of Max Knee Flexion (% Stance)	38.38 (37.38 – 61.61)	43.71 (39.33 – 100.00)	51.57 (47.98 – 53.24)	0.13	2	1.00
Knee Position at Toe-Off (°)	25.52 (17.42 – 25.86)	24.11 (16.35 – 41.68)	21.72 (20.01 – 31.12)	0.50	2	0.93
Mean Knee Flexion Velocity (m/s)	177.07 (124.55 – 299.39)	183.22 (69.34 – 238.15)	151.53 (51.55 – 216.82)	3.50	2	0.27
Hip Angle Excursion - IC to Max EXT (°)	-33.32 (-41.54 – (-31.00)	-43.96 (-47.16 – (-37.24))	-41.72 (-45.82 – (-32.00))	6.50	2	0.04
Mean Hip EXT Velocity - IC to Max EXT (°)	-153.53 (-161.18 – (-132.40)	-179.28 (-182.86 – (-160.42))	-159.56 (-198.33 – (-133.42))	3.50	2	0.27
Max Braking GRF (N/kg)	-2.53 (-2.67 – (-1.74)	-2.44 (-3.51 – (-1.35))	-2.49 (-2.56 – (-1.71))	0.00	2	1.00
Timing of Max DF Moment (% Stance)	54.54 (50.36 – 58.10)	53.33 (49.67 – 59.34)	57.16 (44.94 – 65.11)	0.67	2	0.94
Max Knee Flexion Moment at Loading Response (Nm/kg)	0.27 (-0.06 – 0.95)	0.33 (-0.07 – 1.27)	0.31 (0.11 – 0.49)	0.67	2	0.94
Max Knee Flexion Moment at Toe-Off (Nm/kg)	2.13 (1.22 – 2.23)	2.31 (1.79 – 2.64)	2.11 (0.92 – 2.46)	2.00	2	0.53
Knee Stiffness	3.22 (0.68 – 5.86)	3.22 (1.12 – 8.25)	5.37 (0.64 – 9.84)	0.67	2	0.94
Max Knee ADD Moment (Nm/kg)	1.07 (0.71 – 2.03)	0.81 (0.75 – 2.12)	1.18 (0.32 – 1.56)	0.67	2	0.94
Timing of Max Hip IR Moment (% Stance)	79.74 (75.76 – 100.00)	64.04 (62.67 – 96.92)	67.25 (63.73 – 94.02)	4.67	2	0.19
Ankle Power: Generation (A2) (W/kg)	13.24 (11.16 – 13.38)	9.73 (1.88 – 12.11)	4.52 (3.74 – 5.65)	4.67	2	0.19
Knee Power: Absorption during Loading Response (K1) (W/kg)	-8.62 (-9.20 – (-2.66))	-5.07 (-10.98 – (-0.95))	-4.46 (-6.11 – (-3.98))	0.67	2	0.94

Significant at p=0.017 with Bonferroni Correction

Table 3.20. Comparison of Significant (*p≤0.05*) Running Kinematic Variables for No Brace, TAFO, DAFO at Study Onset and DAFO at Study Completion; compared to Controls

	No Brace		TAFO		DAFO Time 0		DAFO Time 6		Controls
	Mean (SD)	*p*	Mean (SD)	*p*	Mean (SD)	*p*	Mean (SD)	*p*	Mean (SD)
Max Ankle DF (°)	25.34 (7.55)	0.22	21.18 (4.79)	0.06	15.71 (5.06)	0.008*	16.68 (3.16)	0.03*	30.51 (5.28)
Max Ankle PF (°)	-21.13 (7.92)	0.22	-4.34 (6.75)	0.03*	7.13 (2.55)	0.008*	5.55 (1.04)	0.03*	-27.98 (5.48)
Time to Max Ankle PF (% Stance)	99.72 (0.63)	1.00	44.21 (49.42)	0.14	32.74 (30.20)	0.008*	91.29 (16.04)	0.89	100.00 (0.00)
DF Excursion (°)	30.82 (10.54)	0.31	21.41 (7.13)	0.89	7.83 (3.33)	0.008*	8.92 (3.96)	0.03*	23.56 (7.12)
Ankle Position at Toe-Off (°)	-18.64 (4.82)	0.06	-2.72 (8.00)	0.03*	8.05 (2.94)	0.008*	5.61 (1.05)	0.03*	-27.98 (5.48)
Max Ankle DF Velocity (m/s)	1029.46 (1296.00)	0.42	299.92 (112.96)	0.34	136.98 (54.13)	0.008*	170.07 (67.65)	0.03*	405.59 (107.56)
Mean Ankle DF Velocity (m/s)	235.11 (128.58)	0.84	157.37 (96.81)	0.69	65.36 (35.29)	0.008*	97.11 (42.27)	0.03*	200.00 (90.04)
Max Ankle PF Velocity (m/s)	-763.42 (540.28)	0.15	-318.35 (102.01)	0.03*	-118.64 (22.50)	0.008*	-140.11 (37.63)	0.03*	-759.91 (87.79)
Max Ankle EV (°)	24.97 (16.37)	0.55	22.86 (15.28)	0.20	6.69 (13.79)	0.03*	15.56 (7.86)	0.06	37.63 (15.44)
Ankle EV Excursion (°)	32.62 (14.81)	0.42	18.46 (6.41)	0.06	11.73 (6.67)	0.008*	11.42 (7.84)	0.03*	37.69 (8.47)
Max Ankle External Rotation (ER) (°)	-15.74 (12.06)	0.22	-3.37 (13.71)	0.03*	-10.24 (14.06)	0.10	-14.48 (13.40)	0.11	-28.06 (9.54)
Mean Stance Foot Progression Angle (°)	3.52 (1.58)	0.02*	3.36 (2.55)	0.03*	3.03 (2.93)	0.10	2.249 (4.73)	0.11	8.12 (2.80)
Time to Max Knee Flexion (% Stance)	55.50 (26.81)	0.02*	70.76 (33.81)	0.03*	54.35 (7.48)	0.008*	51.14 (6.18)	0.03*	35.28 (2.29)
Knee Flexion Excursion (°)	26.70 (14.70)	0.42	25.36 (9.26)	0.34	15.85 (7.74)	0.02*	17.86 (9.18)	0.20	28.62 (3.65)
Time to Max Knee Varus (% Stance)	59.95 (36.24)	0.31	79.72 (29.10)	0.03*	34.84 (6.99)	0.31	59.70 (47.55)	0.43	9.31 (5.31)
Knee Varus Position at Toe-Off (°)	1.45 (3.39)	0.008*	7.79 (9.49)	0.03*	-5.16 (6.65)	0.008*	-3.65 (2.75)	0.03*	3.97 (5.90)

Table 3.20 (Continued)	No Brace		TAFO		DAFO Time 0		DAFO Time 6		Controls
	Mean (SD)	p	Mean (SD)	p	Mean (SD)	p	Mean (SD)	p	Mean (SD)
Max Knee Flexion Velocity (m/s)	2506.46 (4763.97)	0.69	421.91 (110.95)	0.11	272.01 (63.22)	0.008*	303.54 (35.15)	0.03*	567.92 (51.99)
Mean Knee Flexion Velocity (m/s)	180.03 (82.95)	0.02*	144.00 (80.46)	0.03*	132.88 (65.93)	0.008*	167.47 (77.19)	0.03*	341.41 (29.17)
Max Knee IR (°)	15.12 (17.15)	0.55	7.56 (2.97)	0.03*	13.53 (6.87)	0.03*	14.91 (4.58)	0.06	25.43 (6.09)
Mean Knee IR-ER Velocity (m/s)	133.25 (82.83)	0.84	236.41 (76.33)	0.11	41.73 (25.10)	0.008*	215.04 (129.76)	0.11	226.43 (86.09)
Max Hip Extension (°)	-3.90 (6.72)	0.42	-1.25 (4.30)	0.11	3.78 (4.77)	0.03*	1.18 (4.69)	0.06	-8.97 (5.89)
Hip Excursion - IC to Max EXT (°)	-38.201 (6.74)	0.008*	-43.48 (4.36)	0.03*	-40.33 (6.25)	0.008*	-39.98 (4.47)	0.03*	-51.71 (1.87)
Hip Sagittal Position at Toe-Off (°)	0.004 (4.82)	0.10	3.99 (9.77)	0.06	3.81 (4.79)	0.03*	1.52 (4.41)	0.06	-8.67 (5.81)
Mean Hip Extension Velocity - IC to Max EXT (m/s)	-142.54 (41.10)	0.008*	-168.64 (14.83)	0.03*	-175.02 (28.03)	0.03*	-188.21 (18.26)	0.03*	-218.68 (11.40)
Hip Frontal Position at Toe-Off (°)	-6.93 (3.77)	0.42	-0.46 (4.31)	0.49	-3.10 (3.75)	0.84	-0.40 (6.79)	0.03*	-4.27 (3.47)
Time to Max Hip ADD Velocity (% Stance)	36.70 (18.83)	0.15	38.67 (16.72)	0.20	42.71 (17.90)	0.008*	33.15 (12.61)	0.06	16.55 (6.27)
Hip Transverse Position at Initial Contact (°)	-9.82 (13.87)	0.55	-20.29 (9.69)	0.03*	-7.77 (17.09)	0.69	-7.27 (27.82)	0.69	0.13 (14.02)
Hip Excursion - IC to Max IR (°)	27.22 (27.51)	0.15	28.34 (9.17)	0.03*	14.74 (11.04)	0.31	12.28 (14.66)	0.69	6.94 (5.59)
Hip Excursion - Max IR to Max ER (°)	-32.84 (24.45)	0.55	-31.13 (10.77)	0.03*	-24.07 (6.88)	0.10	-23.94 (7.39)	0.20	-17.10 (3.41)
Time to Max Hip IR Velocity (% Stance)	48.61 (46.41)	0.22	50.31 (23.86)	0.20	62.83 (29.77)	0.008*	56.08 (14.83)	0.03*	14.07 (13.31)
Max Hip ER Velocity (m/s)	-3151.18 (6156.85)	0.31	-257.33 (217.04)	1.00	-406.23 (104.30)	0.02*	-238.51 (118.66)	0.49	-244.47 (54.56)
Max Spine Angle (°)	16.13 (3.65)	0.42	18.13 (0.75)	0.03*	17.92 (3.51)	0.69	4.70 (1.03)	0.06	7.03 (0.93)

Significant at $p \leq 0.05$; Shaded variables represents significant variables for each group versus control subjects; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Table 3.21. Comparison of Significant (*p≤0.05*) Running Kinetic Variables for No Brace, TAFO, DAFO at Study Onset and DAFO at Study Completion; compared to Controls

	No Brace		TAFO		DAFO Time 0		DAFO Time 6		Controls
	Mean (SD)	*p*	Mean (SD)	*p*	Mean (SD)	*p*	Mean SD	*p*	Mean (SD)
GRF Impulse – Normalized	0.34 (0.02)	0.56	0.31 (0.03)	0.06	0.30 (0.02)	0.008*	0.28 (0.03)	0.03*	0.35 (0.02)
Max Braking GRF (N/kg)	-2.26 (0.42)	0.02*	-2.09 (1.11)	0.11	-2.11 (0.39)	0.008*	-2.02 (0.45)	0.03*	-3.43 (0.18)
Time to Max Braking GRF (% Stance)	25.72 (5.02)	0.19	27.88 (5.15)	0.89	38.50 (5.07)	0.008*	34.08 (8.34)	0.34	29.93 (3.11)
Max Propulsive GRF (N/kg)	2.72 (0.27)	0.06	1.65 (0.73)	0.03*	1.66 (0.62)	0.008*	1.51 (0.69)	0.03*	3.43 (0.53)
Timing of Max DF Moment (% Stance)	54.65 (3.22)	0.02*	53.14 (4.44)	0.11	56.95 (10.11)	0.69	53.16 (6.04)	0.20	58.87 (0.68)
Max Ankle IR Moment (Nm/kg)	0.42 (0.27)	0.29	0.21 (0.12)	0.03*	0.35 (0.15)	0.15	0.19 (0.12)	0.03*	0.52 (0.08)
Max Knee Flexion Moment - Loading Response (Nm/kg)	0.67 (0.70)	0.02*	0.64 (0.62)	0.03*	0.98 (1.56)	0.10	1.07 (1.42)	0.20	3.14 (0.52)
Max Knee Flexion Moment – Toe-off (Nm/kg)	1.95 (0.49)	0.02*	1.96 (0.68)	0.34	2.24 (1.08)	0.15	2.31 (0.76)	0.34	3.14 (0.52)
Time to Max Knee Flexion Moment – Toe-Off (Nm/kg)	36.28 (6.15)	0.73	38.60 (2.05)	0.89	52.93 (4.85)	0.008*	47.58 (4.78)	0.06	36.10 (5.08)
Max Knee EXT Moment (Nm/kg)	-0.52 (0.21)	0.56	-0.43 (0.13)	0.11	-0.49 (0.23)	0.69	-0.33 (0.07)	0.03*	-0.58 (0.12)
Max Knee ADD Moment (Nm/kg)	1.39 (0.61)	0.02*	1.03 (0.74)	0.06	0.86 (0.56)	0.008*	0.92 (0.75)	0.03*	2.46 (0.42)
Knee Stiffness (Nm/°)	4.98 (4.04)	0.03*	4.05 (3.01)	0.03*	9.46 (6.58)	0.55	5.51 (6.75)	0.20	12.19 (4.39)
Max Hip Flexion Moment (Nm/kg)	1.33 (0.52)	0.92	1.00 (0.36)	0.20	1.03 (0.52)	0.69	0.59 (0.22)	0.03*	1.33 (0.15
Max Hip EXT Moment (Nm/kg)	-2.14 (0.56)	0.29	-1.65 (0.31)	0.03*	-1.68 (0.62)	0.03*	-1.88 (0.45)	0.06	-2.83 (0.75)
Max Hip IR Moment (Nm/kg)	0.05 (0.06)	0.41	0.04 (0.02)	0.11	0.03 (0.01)	0.03*	0.05 (0.04)	0.34	0.09 (0.08)

Table 3.21 (Continued)	**No Brace**		**TAFO**		**DAFO Time 0**		**DAFO Time 6**		**Controls**
	Mean (SD)	***p***	**Mean (SD)**	***p***	**Mean (SD)**	***p***	**Mean SD**	***p***	**Mean (SD)**
Time to Max Hip EXT Moment (% Stance)	62.09 (2.19)	0.29	30.38 (12.51)	0.03*	40.49 (6.70)	0.02*	37.22 (12.11)	0.03*	29.61 (5.53)
Ankle Power: Generation (A2) (W/kg)	12.49 (1.04)	0.03*	6.20 (5.54)	0.03*	3.89 (1.39)	0.008*	5.03 (1.02)	0.03*	19.27 (2.58)
Knee Power: Absorption during Loading Response (K1) (W/kg)	-7.05 (2.99)	0.03*	-5.15 (4.24)	0.03*	-4.24 (2.15)	0.008*	-5.66 (3.51)	0.03*	-16.49 (4.94)
Knee Power: Generation during Midstance (K2) (W/kg)	8.97 (4.49)	0.06	6.73 (4.28)	0.03*	5.96 (3.23)	0.008*	7.68 (4.09)	0.06	16.56 (3.75)
Hip Power: Absorption during Midstance (H2) (W/kg)	-8.46 (3.65)	0.06	-6.50 (2.03)	0.03*	-3.74 (4.33)	0.008*	-7.55 (2.72)	0.03*	-17.83 (5.62)

Significant at $p \leq 0.05$; Shaded variables represents significant variables for each group versus control subjects; Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

CHAPTER IV - DISCUSSION

Injured service members with partial lower extremity paralysis may benefit from a dynamic AFO over a traditional AFO for return to vigorous activity. Individual injuries, stages of rehabilitation and patient compliance, however, must be addressed in the AFO prescription regardless of type in order to achieve optimal success in ambulation. Some AFO types may delay or inhibit functional rehabilitation especially if the limb presents with swelling, secondary injuries or fragile skin grafting. Improvements in walking and running gait biomechanics were apparent during initial use of the DAFO, which provided improved comfort, stability and energy return during push-off and translated to continual gait improvements over the six month study period. The improvements in service member quality-of-life as measured by the SF-36 contributed to a 'very cost-effective' rating of the DAFO despite the increase in price over traditional models. The quantified improvements may provide military leadership and healthcare professionals the justification needed to improve the standard of care for military patients with limb salvage.

Despite the shared diagnosis of drop foot among the six service members studied in this case series, the levels of trauma and subsequent side-effects during individual rehabilitations were very different. The complex nature of these injuries and the broad range of clinical sequelae made standardization of data collection for all service members difficult and necessitated flexibility in the interpretation of analyses. One service member suffered a sniper gunshot to the upper thigh, paralyzing all muscles below the wound in his left lower extremity. This subject felt the risk of falling outweighed the risk of attempted running without an AFO, and therefore did not complete the no-brace running condition throughout the study period. The service member involved in the motorcycle accident had recently endured a failed anterior cruciate ligament (ACL) reconstruction, and subsequently rehabilitation continued with a lower extremity containing several pieces of protective metal structure and a knee with severe anterior translation. He was unable to run in the no-brace, TAFO, and DAFO conditions at study onset, and did not complete a successful three trial average until halfway through the study period. The stroke survivor discarded his TAFO prior to the study, having developed a deep vein thrombosis attributed to TAFO use. Consequently no data in the TAFO condition was collected on this individual. Two of the six service members were medically retired during the six-month protocol, forcing a disenrollment from the study due to a military move off-island. Their data is included only in analyses appropriate for their portion of study completion.

Additionally, each service member experienced different levels of stress related to their rehabilitation progress, pending military career and family situations. All members were facing medical evaluation boards for separation or retirement from military service, and half were being treated for post-traumatic stress-disorder related to combat or their injuries. Two soldiers were on medication to help with sleep disorders. These added stressors may or may not have complicated rehabilitation and study participation and affected their levels of functional improvement and increased fitness.

Quality-of-Life

The importance of quantifying changes in quality-of-life has increased in recent decades within the healthcare system. Clinician and health care administrators have prioritized health-related quality-of-life as a component of patient care and policy development [54]. The SF-36 questionnaire was utilized in this study to assess changes in injured service members' quality-of-life over time following use of a DAFO; higher scores indicated better health status. The significant change in the "Physical Functioning" domain over time (p=0.009) provided direct quantification of improvements in patients' ability to perform physical tasks such as vigorous exercise, moderate activities such as house work, lifting or carrying groceries, climbing stairs, kneeling and walking distances. The specific task contributing to this change was the ability of service members to climb one flight of stairs, which improved between study onset and one month after they began wearing the DAFO. Other studies which have assessed quality-of-life of patients with peroneal nerve damage also reported difficulty in climbing stairs associated with musculoskeletal weakness of the quadriceps and gluteus maximus, as well as the limited use of the plantarflexors [55]. This finding was supported by research from De Brujin et al. (2007) [56], who retrospectively followed drop foot patients and noted over two-thirds of subjects perceived limitations in climbing stairs, which also contributed to a significantly lower physical function domain compared to their healthy reference sample population.

Quality-of-life changes over time measured by the SF-36 fluctuated in the present study, but the majority improved between study onset and completion. Two exceptions included vitality, which returned to the same level after six months, and general health perception, which decreased slightly over the study period. Changes in these domains have been highlighted in other research involving service members with limb salvage. Tekin et al. (2007) [57] compared the quality-of-life between members of the Turkish

Armed Forces who suffered severe lower limb trauma resulting in limb salvage or amputation. The general health and vitality domains were significantly lower for limb salvage when compared to amputee quality-of-life scores. Authors theorized this difference may have been attributed to complications resulting in re-hospitalization common in limb salvage patients that may affect them physically, psychologically, and socially [57]. This theory was supported by findings in the present study, as some service members underwent continuous surgical procedures and doctor visits throughout the study duration, and delayed additional procedures due to the psychological fatigue of re-hospitalization.

The ability to resume pre-injury well-being and quality-of-life has been identified as the best measure of successful treatment [11]. Despite improvements in SF-36 scores over time, the quality-of-life of the injured service member group in the present study was significantly lower than normative data from healthy populations. Jenkinson et al. (1993) [58] studied a large working age population, and those not reporting long standing illnesses scored 92.5 in physical functioning, 78.8 on general health perceptions, and 64.0 on vitality. Average total scores in Jenkinson's sample were 2.6 times greater over the eight domain average of 32.5 among service members in the present study. The outcomes of this study and the comparison with healthy normative data indicated the severity of lower extremity trauma is long-term, and the use of the DAFO over six months did not return quality-of-life to normative levels, but improved quality-of-life over time in six of eight SF-36 domains. The long-term effects of DAFO use on quality-of-life in patients with drop foot pathologies warrants further examination.

Anthropometrics

Drop foot pathologies commonly present as muscular deficits with proximal atrophy in the kinetic chain due to injury origin and/or lack of activity. Ankle dorsiflexion and knee extension strength measurements were considerably lower than published normative values of healthy adult males ages 20-29 even at study completion [59], but comparable to the hemiparetic population studied by Wang et al. (2007) [60]. Despite a lack of significant difference in range of motion and strength measurements prior to and at the end of the study period, all tested muscle groups demonstrated improvements in strength with relatively unchanged ranges of motion. The diminished dorsiflexion strength in the present study was expected due to partial peroneal neuropathy, however service members improved by nine pounds of force over the study period.

Limb instability and weakness of the quadriceps has been associated with peripheral neuropathic disorders [55], and may lead to knee collapse or hyperextension during gait. Consistent associations have been reported between decreased knee extension strength and asymmetrical gait parameters combined with decreased walking velocity [61]. Knee extension strength improved over 12 pounds of force during the study period, possibly related to the return of a heel strike at initial contact, which promoted anterior stabilization of the quadriceps and continued shock absorption during loading response [62]. The DAFO also supported a return to more vigorous activities and rehabilitation which was reflected in marked increases in all measured muscle groups over time. Whereas most strength measures were lower compared to normative values, hip flexion and hip abduction in the present study were noticeably higher [59]. This increased strength at the hip may be attributed to the learned compensatory mechanisms of excessive hip flexion (steppage gait) and hip abduction (circumduction) commonly seen in individuals with drop foot pathologies.

Walking Gait Biomechanics

Healthy Service Member Controls

Each injured service member was age and anthropometrically matched by height and weight to a service member control with no previous lower extremity injuries. Comparisons to this uninjured population were considered pathologic deviations from normal, thus it was appropriate to first ensure the biomechanics of the uninjured group adequately matched published literature for walking gait parameters. Spatial-Temporal (ST) parameters for control service member walking gait displayed symmetrical step length, stance time, and swing time (Appendix III). Stance time (58%) and swing time (39%) closely resembled measures reported by Ounpuu (1994) (59% and 41%, respectively) [63]. The maximum comfortable walking velocity achieved by the controls slightly exceeded that of published "fast" walking reference data by Oberg (1993) [64], and also reflected a slightly longer step length (0.89m versus 0.76m). Sagittal kinematics at the hip, knee, and ankle at initial contact, mid-stance, and toe-off closely resembled sagittal waveforms in the literature [65]. Maximum ROM and timing of gait cycle events were also similar, which made it reasonable to assume the same of joint velocities. Coronal plane measurements at the pelvis were consistent with the eight degrees of ROM cited by Ounpuu [63], as were hip abduction and adduction peaks during stance. Foot progression angles also fell within the normal range (0-10 degrees) [63].

Kinetically, the service member controls achieved peak walking vertical GRF measures of 12.58N, or 1.43 times their body weight. This also falls within normative thresholds of published data where subjects reached 1.3 to 1.5 times their body weight in GRF during walking [63]. Joint moments and powers for the hip, knee, and ankle in the sagittal plane were all consistent with Ounpuu with the exception of maximum hip extension moments which were slightly higher than published norms, more resembling those seen in running [63, 66]. Overall, uninjured service member controls demonstrated nearly identical gait measurements to those reported previously for normal walking gait biomechanics.

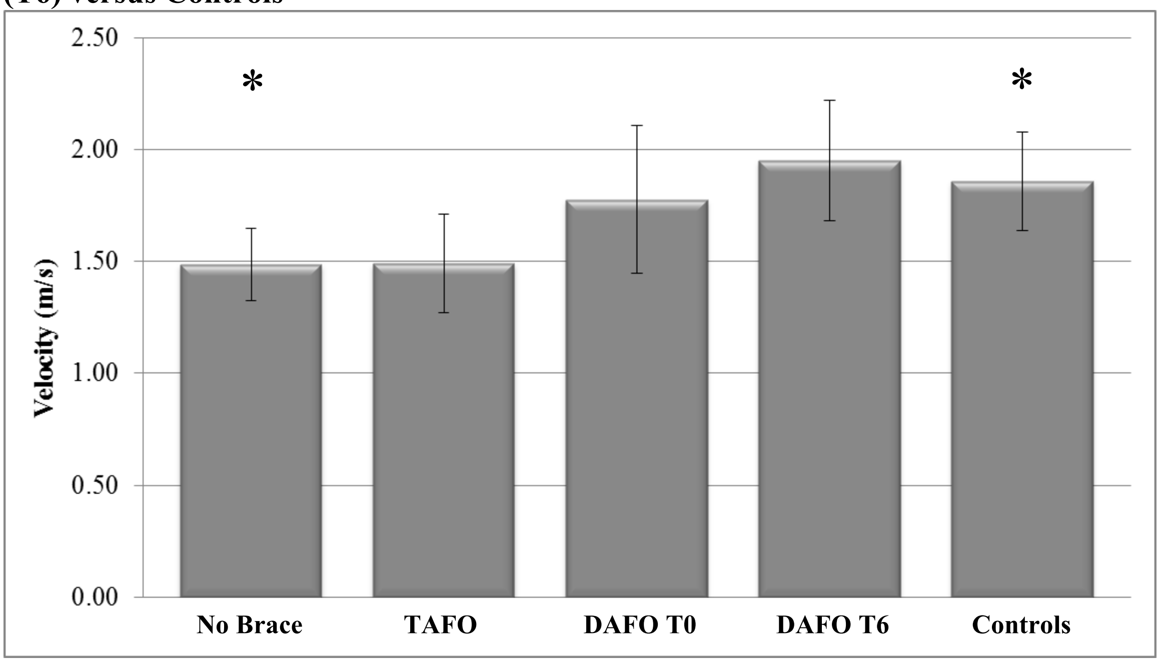
Spatial-Temporal Parameters

Asymmetric step length, longer stance length and a shorter swing time of the sound limb have been uniquely attributed to hemiplegic gait [67, 68]. The walking ST gait parameters in the non-braced injured service member group revealed significantly shorter swing times of the sound limb, longer stance times, slower velocity and shorter stride length as well as shorter and asymmetrical step lengths in the involved and sound limbs (Table 3.4). Researchers of studies involving successful limb-salvage surgeries reported similar findings of longer sound limb stance time, claiming the sound limb must provide support long enough to allow swing phase and assume a portion of the loading function of the involved limb [69]. Stance time has been reported to be a more reliable clinical measure when disability is severe compared to measures of vertical loading [62]. Shorter step length of the sound limb may also be the result of limited progression of the involved limb due to the loss of the ankle rocker during midstance [70]. This adaptation assists in balance, but results in a net loss of energy when compared to able-bodied walking [71].

The use of a TAFO improved stance and swing times and increased velocity as it related to an untreated drop foot, however significantly shorter stride lengths (p=0.032), and shorter involved (p=0.032) and sound limb step lengths (p=0.016) remained. Despite some improvements over the no brace condition, opinions concerning comfort of the TAFO types varied. The traditional designs used in this study included two hard plastic hinged versions and three flexible carbon fiber composites with dorsiflexion assist versions, touting flexibility and strength [72]. The anecdotes from the injured service member group as to the positive aspects of these traditional models included the ability to fit inside a dress shoe, and the dorsiflexion assist allowed movement of the limb between the clutch and the brake with greater ease during motor vehicle operation. The injured service members listed activities they were able to perform in their issued TAFO which included "walking very short distances" and "typical activities of daily living". Some service members with different levels of trauma and pathology found "sitting down was even painful" in the TAFO, or that they could "perform just as well without it" during low impact cardio activities. One service member suffered a deep vein thrombosis while using his plastic hinge TAFO, and attributed this life-threatening side-effect to the issued device.

The mechanics of the DAFO brace were designed for an efficient, secure walking and running gait by providing mechanical energy return from the spring tension on the posterior tibial struts, as well as medial-lateral and rotational stability from the cuff and foot components [34]. Once donning the DAFO, the previously significant ST parameters improved and were no longer different than controls, indicating an immediate improvement effect of the DAFO without practice. Six months after DAFO use, injured service member walking velocity with the DAFO exceeded that of control subjects (Figure 4.1). While wearing the DAFO, service members exceeded the published normative values for fast walking gait in healthy males ages 30-39 (1.77 m/s) as well as step length (0.76 m) [64]. The ability to produce a fast walking gait is especially important for the injured service member hoping to return to active service, as marching in formation is among the everyday duties required in some branches of the military.

Figure 4.1. Walking Velocity Comparisons for Injured Service Members without AFO, with TAFO, with DAFO at Study Onset (T0), and DAFO at Study Completion (T6) versus Controls



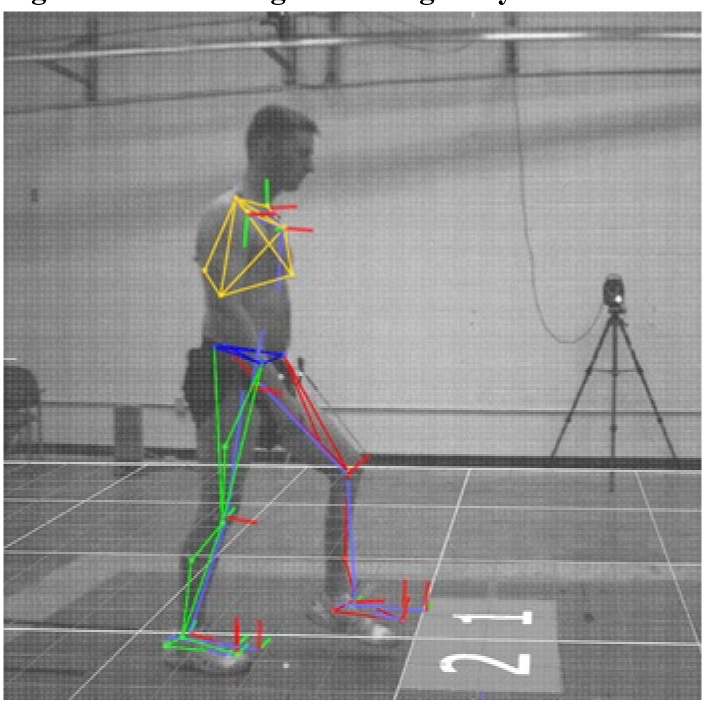
**Significant at p≤0.05*

Pathologic Gait Deviations

The most notable kinematic and kinetic gait abnormalities in patients with drop foot during walking have occurred at initial contact through loading response and during the swing phase [62]. Biomechanical differences between the injured and control groups in the present study were found during swing phase and loading response, with additional variations found during later stance. Service members contacted the ground in excessive plantarflexion without an AFO, often with a forefoot strike pattern resulting in a loss of momentum normally preserved by the heel rocker [62]. Other significant ankle deviations related to drop foot included increased dorsiflexion excursion and a greater maximum dorsiflexion velocity during loading response, commonly referred to as "foot slap", which caused disruption in proper limb advancement [22, 62]. Despite normal dorsiflexion moments in terminal stance, the prior disruptions contributed to a decreased propulsive GRF (p=0.015) and reduced propulsive ankle power by half compared to controls (p=0.026). These gait abnormalities were especially obtrusive to this premorbidly fit population who experienced ankle instability and the increased risk of ankle sprains in early stance, as well as degraded propulsive power at push-off needed for activities requiring a faster gait. Consequently greater conscious control of gait was required, noted by altered body posture (Figure 4.2) and was reflected in a decreased maximum vertical ground reaction force (GRF) during loading response (p=0.015) (Figure 4.3).

Lower extremity joints normally control momentum acquired during the swing phase through joint flexion [66]. Proximal deviations in the no brace condition at the knee were more related to timing of kinematics over joint range of motion. Five out of six injured service members' peroneal nerve damage were related to trauma at the knee including shrapnel inside the joint resulting in a notable varus alignment, and gunshot wounds which caused damage either directly at the posterior knee or proximally, rendering distal segments paralyzed. Structural damage to the knee, along with the aforementioned ankle joint deviations caused by peroneal nerve injury likely led to delays of peak varus velocity, knee adduction moment and peak knee extension moment until later in stance.

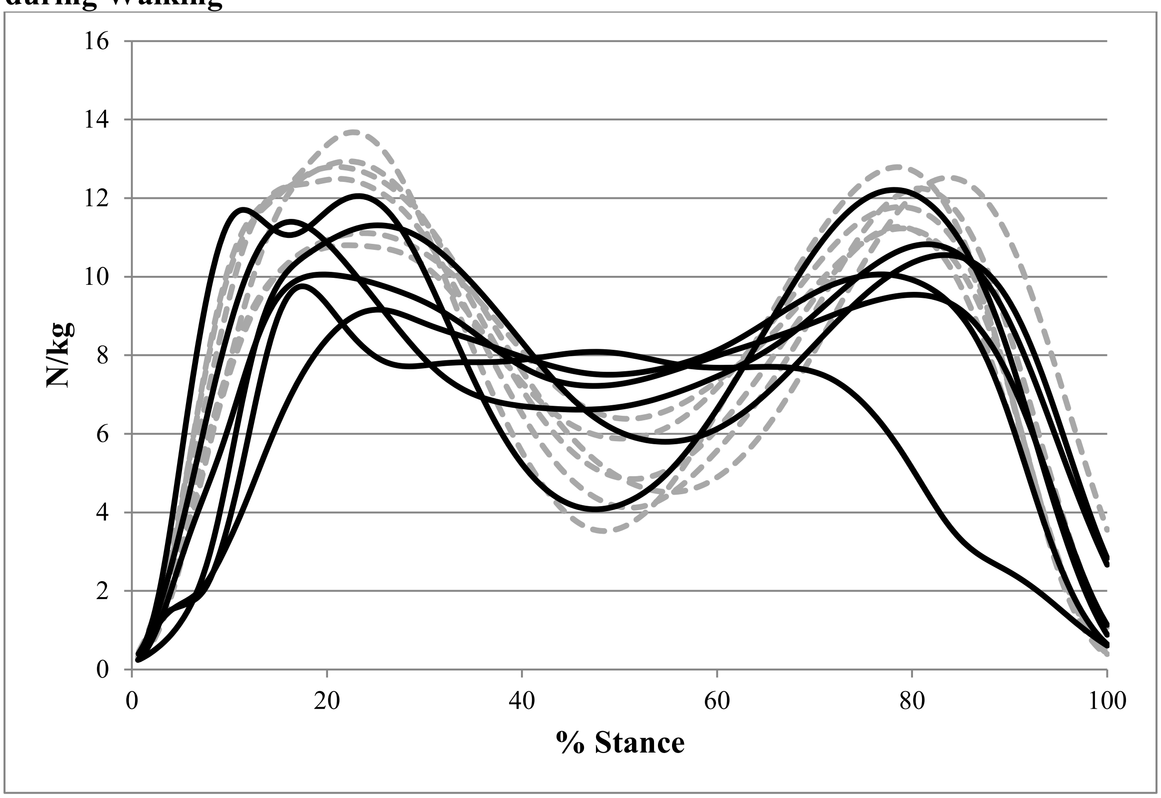
Figure 4.2. Pathologic Walking Body Posture



Power at the hip and knee were also affected, with substantially less eccentric knee extensor absorption compared to controls (*p*=0.004). The knee extension strength below normative levels likely contributed to limited knee flexion during loading response (Figure 4.4), as the most stable weight-bearing condition in the absence of a heel rocker and decreased knee extension strength is to walk with a more fully extended knee [62].

Stiff knee gait was also noted in a case report of a post-stroke patient regardless of DAFO use by Nolan et al. (2010) [73]. This compensation, however, reduced shock absorption and may contribute to secondary micro-trauma injuries. The decrease in knee flexion at loading response was accomplished by increased hip extensor activity, limiting the degree of knee flexion that occurred and slowing the increased tibial advancement instigated by excessive dorsiflexion velocity [62]; this compensation was evident by power generation of the hip extensors during stance, which reached levels twice that of controls (p=0.041).

Figure 4.3. Vertical GRF: Injured Service Members without AFO vs. Controls during Walking

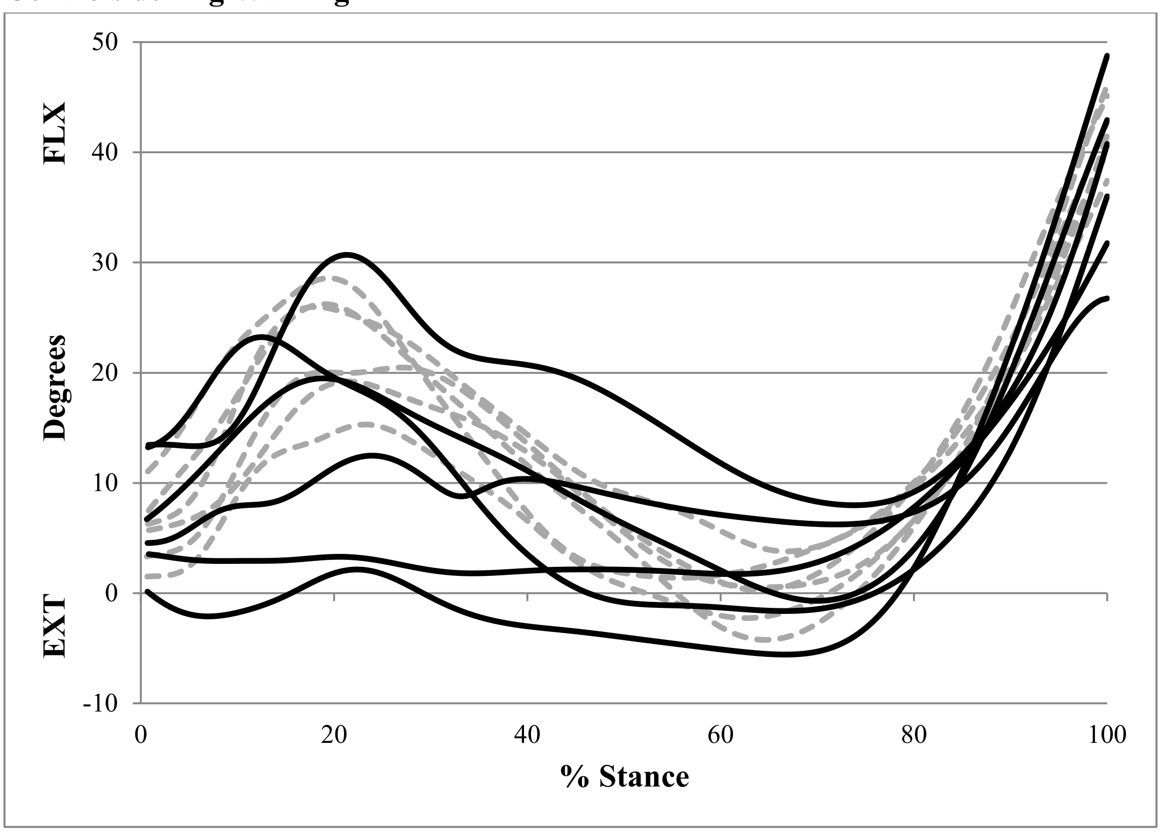


Legend: Without AFO: Black Solid Line; Controls: Gray Dashed Line

Traditional AFO Changes in Walking Gait

Ankle-foot orthotics have often been prescribed to minimize pathological gait deviations and improve walking ability [67], however some designs may create additional gait deviations compounding those already present [74]. Use of the TAFO improved ankle position at initial contact to near neutral, and reduced the dorsiflexion excursion by 44% compared to the no brace condition, despite still significantly exceeding that of healthy controls (p=0.008). The improved control of plantar flexion during loading was reflected in the reduced peak dorsiflexion moment during terminal stance (p=0.032).

Figure 4.4. Stance Phase Knee Flexion: Injured Service Members without AFO vs. Controls during Walking



Legend: Without AFO: Black Solid Line; Controls: Gray Dashed Line; EXT: Extension, FLX: Flexion

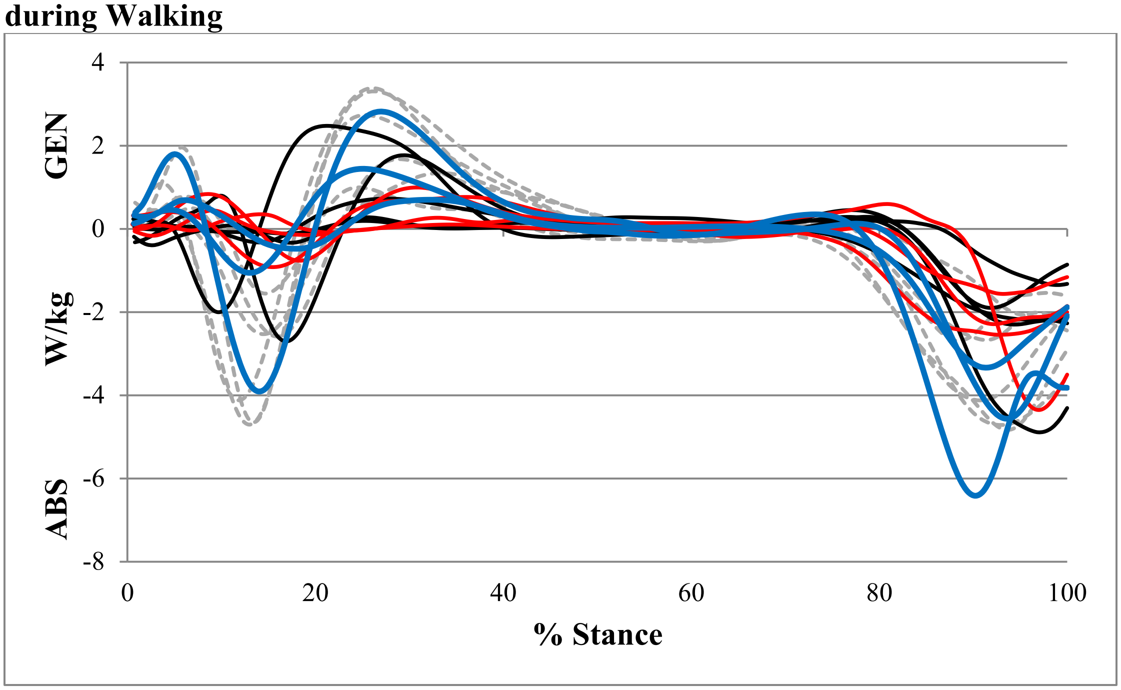
The flexible hinged TAFO design issued to most of the service members did not control dorsiflexion excursion or reduce the dorsiflexion velocity to enhance stability while walking. The average ankle dorsiflexion velocity during mid-stance seen during TAFO use increased over the no brace condition as the design allowed dorsiflexion beyond passive movement, likely augmented by a higher walking velocity. Although increased walking velocities have been correlated to increased joint power at the hip and knee, power at the ankle is insensitive to these changes [75]. A decreased dorsiflexion moment in terminal stance while wearing the TAFO resulted in a significantly lower propulsive power (p=0.017).

The TAFO overly restricted plantarflexion such that time to max plantarflexion was significantly earlier (p=0.008) and the ankle attitude at toe-off was significantly lower (p=0.008) and near neutral compared to controls. The resting state of TAFO designs were in acute dorsiflexed attitudes and likely contributed to greater levels of forced dorsiflexion in the injured group than seen without the brace. Maximum dorsiflexion (p=0.032) and dorsiflexion velocity throughout stance significantly increased (p=0.008), which delayed time to maximum range of motion (p=0.008) and maximum plantarflexion moment (p=0.008), and resulted in accelerated tibial advancement and increased demand placed on the quadriceps [70]. The increased stress on the knee joint was reflected in other knee kinematics and kinetics compared to no brace deviations. Similar to the no brace condition, maximum knee adduction moment was delayed to terminal stance. Power absorption during loading response was less than the no brace condition, suggesting an even more extended knee position during loading. This trend continued through midstance, with an equally low concentric power of the knee extensors

(p=0.009) (Figure 4.5). The compensation of increased hip extensor power improved slightly over the no brace condition, but remained significantly higher than controls (p=0.017).

The more extended knee position may have also contributed to a significantly decreased vertical GRF (p=0.032), decreased peak braking GRF (p=0.016), and delayed time to maximum braking GRF (p=0.032), as compensations that avoid limb loading lowers GRF peaks [62]. The contributions to increases in vertical GRF with increases in walking velocity are overshadowed in this case by the avoidance of loading while wearing the TAFO. Despite corrections made to ankle positioning and knee variables, the design of the TAFO allowed excessive dorsiflexion ROM beyond passive levels, which coupled with weak knee extensor muscles began a cascade of additional gait deviations beyond those seen in the no-brace condition.

Figure 4.5. Knee Power Generation during Midstance: Injured Service Members (without AFO vs. with TAFO vs. with DAFO at Study Completion) vs. Controls during Walking

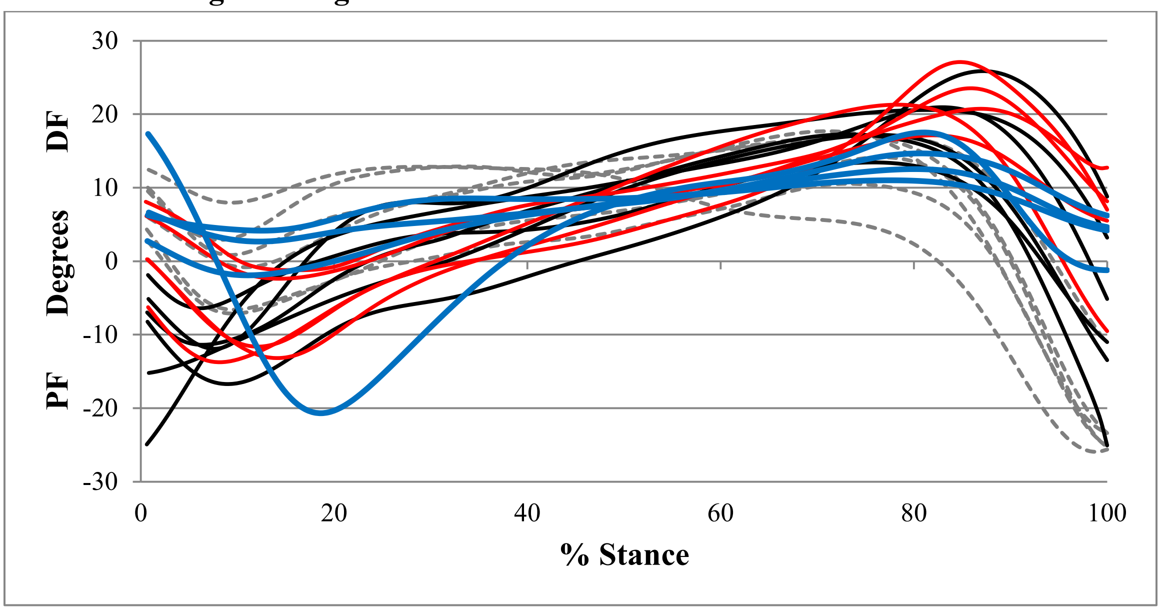


Legend: Without AFO: Black Solid Line; TAFO: Red Solid Line; DAFO: Blue Solid Line; Controls: Gray Dashed Line; ABS: Absorption, GEN: Generation

Dynamic AFO Changes in Walking Gait Over Six Months

The intricacies of the DAFO necessitated a break-in period during which the orthotist recommended no more than four hours of brace wear and one-quarter mileage incremental increases [76]. The period of familiarization needed to adjust to the DAFO was reflected in the number of gait deviations, which in total fell between the no brace and TAFO conditions. Similar to the TAFO, some normalization of original gait deviations were achieved, while other gait deviations surfaced. Ankle position at initial contact was improved to near identical means and standard deviations seen with controls, due to the restrictions imposed by the posterior strut and foot component. The foot position never reached plantarflexion levels at toe-off (p=0.002) compared to 24 degrees of toe-off plantarflexion seen in controls. This loss of plantarflexion range of motion was necessary to maximize stability and exploit the energy-storing capabilities of the carbon fiber strut to assist in propulsive return during pre-swing. The restriction in plantarflexion (p=0.002) was also reflected in premature maximum plantarflexion during loading response as opposed to toe-off (p=0.022). Dorsiflexion in the DAFO design was non-assisted and returned max dorsiflexion angle and dorsiflexion excursion to near-normal levels (Figure 4.6), but restricted both maximum dorsiflexion and plantarflexion velocities (p=0.002). Inversion-eversion excursion was conservatively controlled with a lateral ankle component fitted to service members with severe ankle instability, although one service member had the ankle piece removed due to increased abrasions between the component and fragile skin grafting. The proximal cuff was custom-molded to each individual, and also restricted some movement at the knee. Knee internal-external rotation excursion was significantly decreased (p=0.002), which was likely also related to a decreased knee internal rotation moment (*p*=0.026), however this restriction helped return knee varus velocity to levels identical to controls. Kinetic deviations while first using the DAFO were mainly in the sagittal plane, with reduced braking and propulsive GRF (*p*=0.041 and *p*=0.002, respectively), as well as inadequate power generation at the ankle during push-off (*p*=0.002). Vertical GRF returned to normative values, however, this was likely related to the substantial increase in walking velocity.

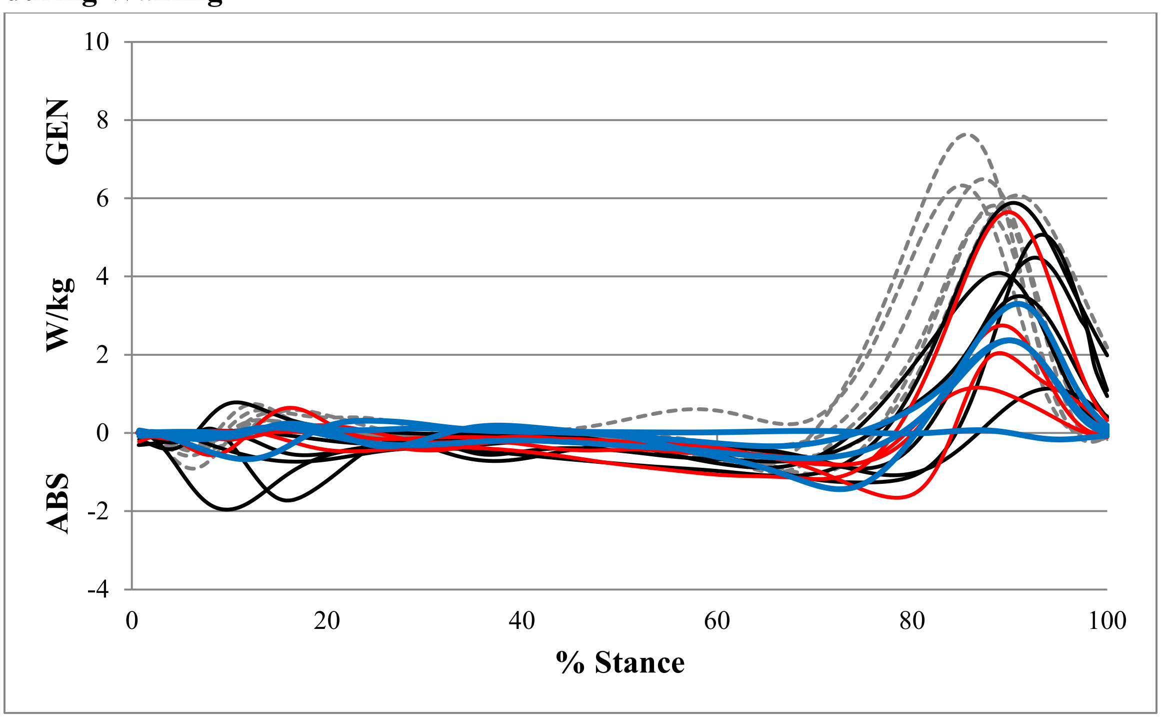
Figure 4.6. Stance Phase Ankle Dorsiflexion/Plantarflexion: Injured Service Members (without AFO vs. with TAFO vs. with DAFO at Study Completion) vs. Controls during Walking



Legend: Without AFO: Black Solid Line; TAFO: Red Solid Line; DAFO: Blue Solid Line; Controls: Gray Dashed Line; PF: Plantarflexion, DF: Dorsiflexion

Completion of the six-month protocol revealed an improvement in all 31 reported kinematic and kinetic walking gait deviations, with only eight remaining significantly different from controls. A majority of these remaining gait deviations were due to DAFO design, unchanged by familiarization and six months of brace use. These included maximum ankle plantarflexion and the time it took to reach this level, ankle position at toe-off and maximum ankle dorsiflexion and plantarflexion velocities. Additionally, ankle power generation at push-off improved by 14% but did not return to published norms (Figure 4.7). This improvement, though small in appearance, may be understated due to the inability to measure the energy return capabilities within the brace. This measure may help explain the increased physical functioning (PF) domain score of the SF-36 questionnaire regarding stair ascent, as propulsive ankle power has been associated with vertical acceleration of the body during this activity [77]. Despite the limiting plantarflexion range of motion, ankle position at toe-off did reach over nine degrees of plantarflexion, suggesting the DAFO is more flexible than it appears.

Figure 4.7. Ankle Power Generation during Preswing: Injured Service Members (without AFO vs. with TAFO vs. with DAFO at Study Completion) vs. Controls during Walking



Legend: Without AFO: Black Solid Line; TAFO: Red Solid Line; DAFO: Blue Solid Line; Controls: Gray Dashed Line

Running Gait Biomechanics

The ability to ambulate without pain and recovery of functional gait has been considered a successful outcome of limb salvage rehabilitation [61]. Ankle-foot orthotics (AFO) have been a conventional approach to treat associated pathologies with limb salvage including drop foot to improve walking gait [22]. Running has sometimes been presented as a natural extension of walking, but variations in velocity, joint ROM, forces and muscle activity, and joint moment and power differences are readily apparent [63]. New requirements sought by a premorbidly fit population of limb salvage patients to allow a return to vigorous activity have challenged the design and necessitated enhancements in the AFO to provide the stability, flexibility, and power needed for running gait.

Without the aid of a DAFO, the injured service members' running gait presented with substantially more gait deviations than seen while walking. Fifty-four different running gait deviations were seen across the no brace, TAFO, naïve DAFO, and DAFO at study completion conditions combined, as compared to controls (Tables 3.20 and 3.21). Only 15 of these 54 significant gait deviations between the injured service member group and controls were present in the no brace condition. This number increased to 28 significantly different gait variables compared to controls while running in the TAFO. When the injured service members first attempted to run in the DAFO the number of significant gait deviations increased to 38, but then decreased to 27 deviations by the end of the study period after adjusting to the brace.

These numbers may suggest that compensation mechanisms without a brace produce similar outcomes to using an AFO. However, a simple comparison of the number of gait deviations between conditions does not adequately represent the effect of an AFO since one-third of the non-braced service members were unable to run at all at study onset. The inability to run could not be quantitatively assessed within the limitations of statistical procedures and cannot reflect the gait improvements facilitated by use of an AFO. The clinical significance an AFO enabling a return to running should not be overlooked. Additionally, gait deviations also present with compensatory mechanisms that could lead to secondary injuries, increased energy expenditure [78], and safety concerns related to ambulating with drop foot pathologies [79].

Healthy Service Member Controls

Uninjured service member running gait variables were also evaluated against published norms to ensure an adequate representation of normal gait. Kinematic and kinetic variables were compared against sprint running biomechanics reported by Novacheck (1998) [66] due to a matched running velocity of 3.99 m/s. All variables revealed high symmetry between both control limbs. Stride lengths, velocities, and swing and step times were consistent with other studies involving male runners with no previous injuries [80]. Much less time was spent in stance than walking, with toe-off occurring at 25% of the gait cycle, similar to values reported previously [66].

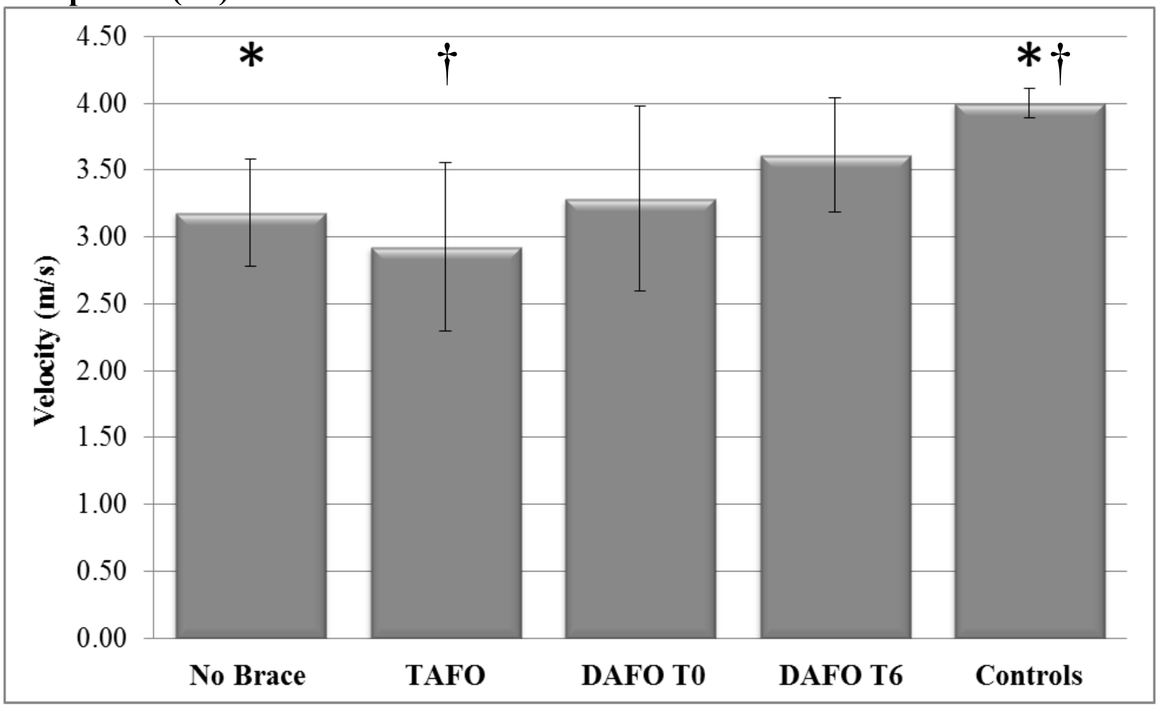
Peak kinematic variables and ranges of motion for the pelvis, hip, knee and thigh were similar to those reported in a normative sample [66]. Ground reaction forces, running velocity and stance time were also nearly identical to values reported by Munro (1987) [81]. The thrust maximum, equivalent to the second peak following the impact peak in a running GRF, measured 2.75 times the body weight of the service member controls, consistent with the 2.5-2.8 range measured by Munro. This peak occurred at

47% of stance in accordance with the 35-50% of stance peaks found in the same study [81]. Joint moments and powers patterns were consistent with Novacheck's findings, although the magnitudes were greater in all cases [66]. Overall, this sample adequately represents normative measures for running gait.

Spatial Temporal Parameters

Differences in ST parameters between controls and the injured service members while running without an AFO were similar to those seen during walking (sound limb swing time, longer stance times, slower velocity, shorter stride length, and shorter, asymmetrical step lengths; see Table 3.12). Stance time, swing time, and step lengths improved significantly while running with the TAFO, however pain levels reported by some service members may have led to the significantly shorter stride lengths (p=0.029) and slower velocities (p=0.029) than controls. Dynamic AFO use significantly improved these differences, although the sound limb swing time remained significantly shorter (p=0.024) and the sound limb stance time significantly longer (p=0.032) than able-bodied gait. These differences correspond with findings of several AFO walking studies [60, 67, 68], however, currently the only published studies pertaining to an AFO in a running condition addressed DAFO design and functional outcomes associated with AFO specific rehabilitation [10, 23], not biomechanical gait improvements. The injured service member group reached an average velocity of 3.61 m/s by study completion corresponding to a 7:25 minute per mile pace (Figure 4.8), which if maintained could produce a passing score on the aerobic component of the physical fitness test for every branch of service [82]. Often a passing score on a military fitness test is one key component to remain on active duty for an injured service member facing a medical evaluation board (MEB).

Figure 4.8. Running Velocity Comparisons for Injured Service Members without AFO, with TAFO, with DAFO at Study Onset (T0) and with DAFO at Study Completion (T6) vs. Controls



**† Significant at p≤0.05*

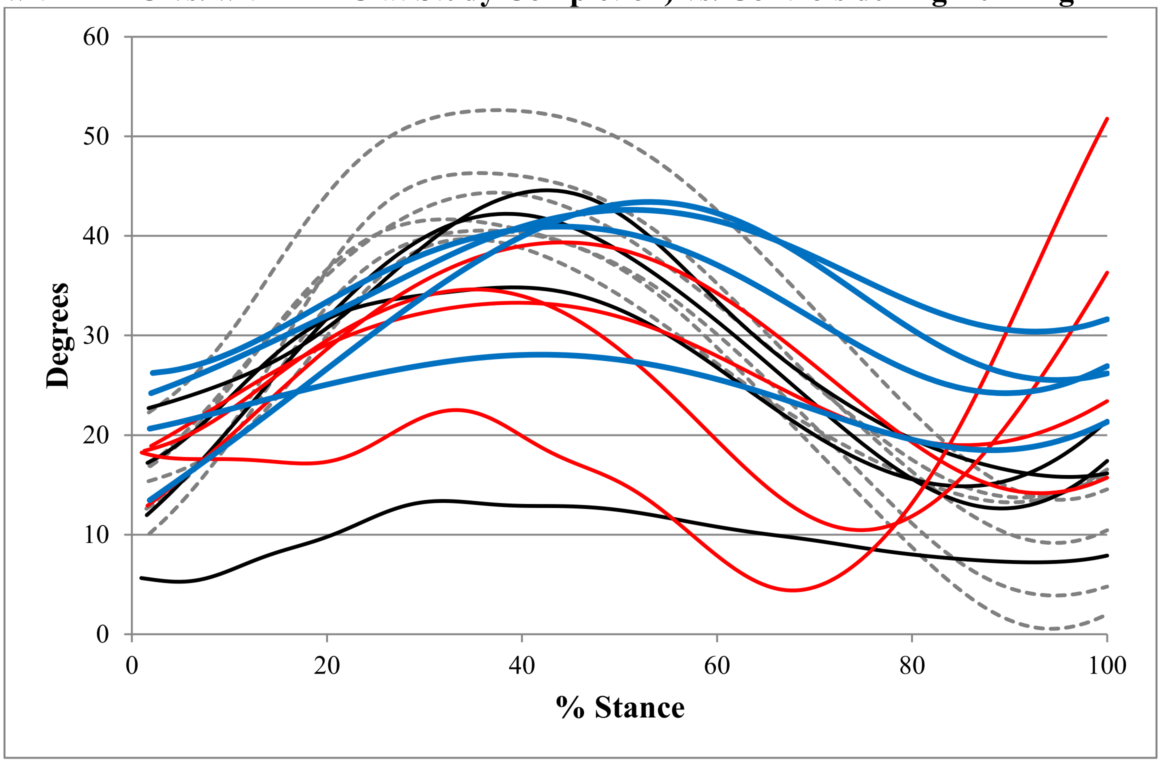
Pathologic Gait Deviations

Published normative values, to our knowledge, do not exist for running gait deviations associated with drop foot pathologies. A majority of gait deviations seen between the no brace condition and controls were found at the knee and hip in the sagittal plane, as greater plantarflexion values at the ankle are common during running. As velocities increase in able-bodied gait, the body's center of mass lowers and notable increases in joint flexion occur [63, 66]. Injured service members ran with a gait that resembled the "Groucho" pattern as described by McMahon et al. (1987). This running pattern has been characterized by excessive knee flexion in an effort to reduce the effective vertical stiffness of the body, ultimately reducing mechanical shock [83]. Knee flexion at initial contact resembled normal levels but with a significantly lower braking GRF (p=0.016) suggesting a cautious loading of the injured limb. This tentative running pattern continued with a slower than normal knee flexion velocity (p=0.016) and resulted in a delay of gait events. Less absorptive knee power during loading response (p=0.029) and a significantly low maximum knee flexion moment during this phase (p=0.016) was compounded by weak quadriceps muscles. Time to maximum knee flexion was delayed from midstance to terminal stance (p=0.016), ending with greater knee flexion at toe-off which resulted in a diminished knee flexion moment and ankle power generation (Figure 4.9). This "Groucho" positioning allowed the injured service member group to remain on the ground for a longer stance period, nearly eliminating the aerial phase of running gait [84]. Knee stiffness reflected this posture, measuring only 41% of the stiffness seen in the literature for healthy runners and the service member controls (p=0.032) [85]. Evidence links low knee stiffness with increased soft tissue injury risk and decreased running economy by as much as 50% [83, 86, 87]. Deviations at the knee with "Groucho" style running may have also contributed to changes at the hip. Hip angle excursion from initial contact to max extension was reduced (p=0.008) which was consistent with McMahon's findings that the hip follows a lower and smoother trajectory with less vertical motion in "Groucho" gait [83].

Traditional AFO Changes in Running Gait

Use of traditional AFO (TAFO) adequately corrected the uncontrolled movements associated with drop foot by aiding a neutral ankle at initial contact, but were overly restrictive with ankle motion into plantarflexion during running, reflected by lower maximum plantarflexion angle and velocity, and ankle position at toe-off (all p=0.029).

Figure 4.9. Stance Phase Knee Flexion: Injured Service Members (without AFO vs. with TAFO vs. with DAFO at Study Completion) vs. Controls during Running



Legend: Without AFO: Black Solid Line; TAFO: Red Solid Line; DAFO: Blue Solid Line; Controls: Gray Dashed Line

A few running gait deviations associated with the "Groucho" pattern improved with TAFO use; hip excursion from initial contact to maximum extension increased as well as the hip extension velocity during stance. Knee flexion moment at toe-off increased; however these improvements remained significantly lower than controls (p=0.029) (Figure 4.9). Some "Groucho" related variables remained unchanged or worsened; timing events at the knee remained delayed, mean knee flexion velocity, absorptive knee power and associated flexion moment during loading response decreased, and the knee reached an even greater flexion position at toe-off, further restricting propulsive ankle power at push-off. Knee stiffness decreased further with use of the TAFO, suggesting injured service members continued compensating to reduce mechanical shock and the aerial phase of running [83, 84]. Additional proximal gait deviations surfaced, including a three-fold increase in spine angle, or forward trunk lean, possibly to compensate for weak quadriceps consistent with previous gait deviations. This movement may have been instigated by the ankle stabilization provided by the TAFO, allowing the center of gravity vector to move nearer to or in front of the knee joint center and relieve the demands placed on the knee extensors [62]. Despite the TAFO providing adequate correction at the ankle for drop foot, associated deviations at the knee related to musculoskeletal weakness resulted in an excessive externally rotated hip at initial contact (p=0.029) likely to provide a wider base of support. Greater excursion from initial contact to maximum internal rotation of the thigh was also present in the TAFO condition compared to no AFO, possibly to resist the flexion moment of the knee as an additional compensation for quadriceps weakness [62].

Dynamic AFO Changes in Running Gait Over Six Months

Donning the DAFO for the first time was awkward for some subjects, as the brace is larger than most TAFO models, a heel lift is required in the shoe of the sound limb, and the posterior strut requires a 1.5 shoe size increase of the involved foot. Despite its robust design, an acclimation period was typically required for service members to trust the stability provided by the DAFO and learn how to utilize the energy storing capabilities of the carbon fiber strut. This may have contributed to kinematic and kinetic deviation increases to 70% of the total significantly different variables when compared to controls. However, 20% of these differences improved to normative values following six-months of DAFO use, with a considerable number of the remaining differences related to the stabilizing features of the DAFO.

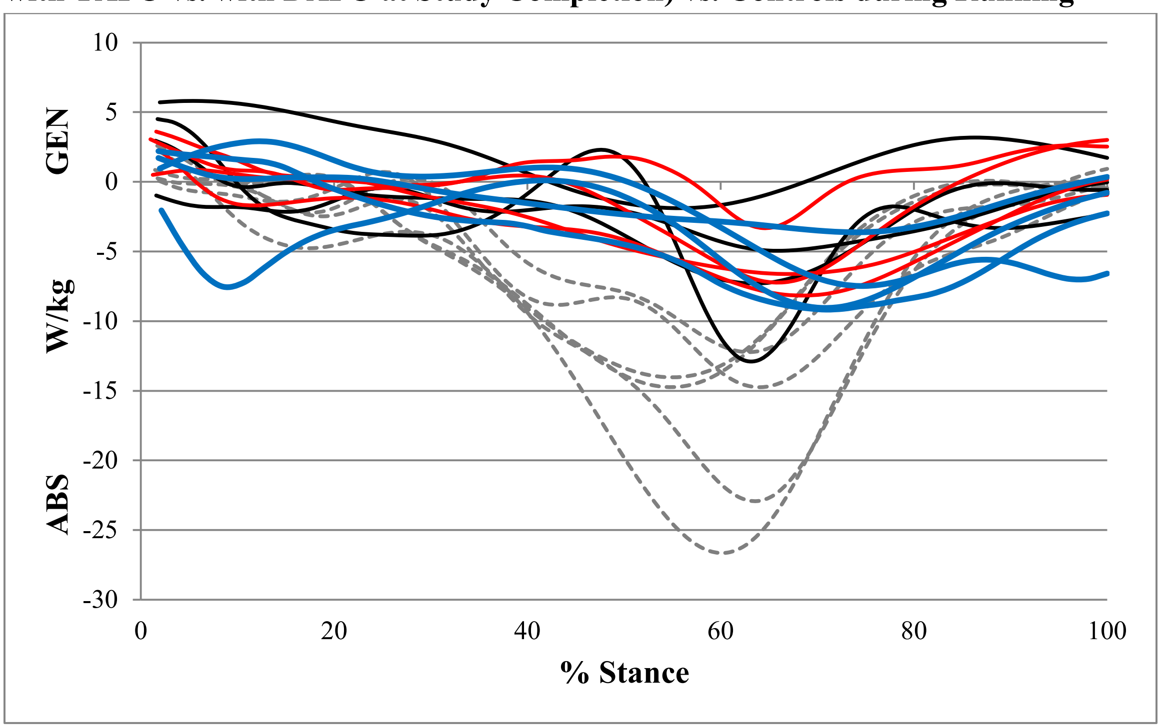
The rigidity of the carbon fiber material in the DAFO maintained the ankle in a position that corrected the drop foot pathology, but limited the amount of plantarflexion, dorsiflexion, and eversion allowed during the gait cycle (all p=0.008). A reduction in dorsiflexion excursion assisted with ankle stability, and controlled the velocities associated with these movements. Over time, however, plantar flexion and dorsiflexion velocities increased by 3 and 15%, respectively, but remained significantly less than those exhibited by controls at study completion (p=0.029). Time to maximum plantarflexion during midstance was premature during initial use of the DAFO but returned to pre-swing timing consistent with controls by the end of the study period. Service members reported greater trust in the DAFO and increased comfort with testing the boundaries of brace flexibility over time. Running gait initiated with a heel-strike pattern that remained stable over six months, was not significantly different from controls and was consistent with published research on normal running gait biomechanics at initial contact [63].

Knee flexion at initial contact and toe-off while wearing the DAFO was not significantly different than controls and within normal ranges seen by Novacheck (1998) [66] unlike values for these variable exhibited while wearing the TAFO (Figure 4.9). Knee stiffness increased with DAFO use to values closer to controls, in conjunction with faster running velocity and related increases in vertical GRF, possibly to accommodate the greater demands of the activity [87]. Increases in stiffness have been associated with reduced excursions and increased peak forces such as those outlined above [87].

Although not significantly different, loading rates in the DAFO at study onset and completion exceeded rates seen in controls, which combined with the increased stiffness and GRF may place the service member at greater risk for bony injuries [87].

The significantly reduced power absorption at the hip in the DAFO at study completion (p=0.008) combined with an increased velocity suggested increased power generation in the distal kinetic chain (Figure 4.10). However, the expected energy-return capabilities of the DAFO were not reflected in the kinetics during propulsion after six months of DAFO use. Ankle power generation only reached 25% and propulsive GRF reached less than half the magnitudes seen in controls.

Figure 4.10. Stance Phase Hip Power: Injured Service Members (without AFO vs. with TAFO vs. with DAFO at Study Completion) vs. Controls during Running



Legend: Without AFO: Black Solid Line; TAFO: Red Solid Line; DAFO: Blue Solid Line; Controls: Gray Dashed Line; ABS: Absorption, GEN: Generation

The calculation of ankle power includes angular velocity requiring range motion at the ankle, which was markedly restricted by the DAFO in order to provide stability and prevent drop foot. Therefore, calculation of ankle joint power does not reflect the energy storing capabilities of the DAFO. The improvements in hip power absorption to more normal levels combined with the increases in running velocity over time in the DAFO demonstrate greater power generation distally in the kinetic chain, and enhanced the ability to run. The distal power generation was likely created by the DAFO to offset the limitations in ankle power caused by the injury.

Limitations

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, medications, post-traumatic stress disorder, depression, difficulties sleeping, injuries to other sites in the body, relationship issues and the stress related to medical evaluation boards and medical retirement 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. The sample size available for statistical analysis was further reduced due to the inability of individual subjects to complete certain conditions or medical retirement.

Conclusions

Walking and running performance were improved via the use of a DAFO over a six month period in service members with limb salvage and subsequent drop foot. The DAFO demonstrated the ability to correct running gait kinematic deviations at the ankle related to drop foot pathologies, provided improvements in stability and comfort over TAFO designs and increased maximum comfortable walking velocity beyond that seen in controls. Increases in running velocity and GRF, reported increases in confidence during limb loading, and observed reductions in proximal kinetic chain compensations indicated improve running capability. These improvements translated to increases in strength in all lower extremity muscle groups over time, as well as significantly higher quality-of-life scores pertaining to an improved ability to ascend stairs. Future studies should assess long-term outcomes of walking and running gait in a DAFO, as six months was not sufficient to return the injured service member group in the present study to consistent vigorous activity.

CHAPTER V – CASE STUDIES

Case Study One

Long-term Outcomes of a Dynamic Ankle-Foot Orthosis on Gait Characteristics of a Service Member with Partial Lower Extremity Neuropathy

(Formatted for and submitted to Military Medicine)

ABSTRACT

This case study reports a five-year follow-up of a 32-year-old male service member who suffered poly-trauma in 2007 following a Humvee rollover in Afghanistan. The service member's injured left lower extremity was salvaged, but injuries to the lumbar sacral plexus, hips, femur, and severe nerve damage resulted in near total paralysis and drop foot of the left lower limb. Two years of multiple substandard ankle-foot orthotic devices pushed him to investigate a dynamic ankle-foot orthotic (DAFO) with energy storing capability, which allowed him to remain on active duty and deploy for a second tour while wearing the device. The anecdotal improvements described by this service member prompted a biomechanical analysis of walking and running gait, comparing a shoes only condition to the DAFO. Results of gait analysis demonstrated an improvement in spatial-temporal parameters in both walking and running, improved sagittal angles and moments at the ankle, knee, and hip, greater ankle stability through decreased dorsiflexion excursion, and a marked increase in ankle power while running. Most notably, the service member credits this device for substantial improvement in quality-of-life including total cessation of pain medication and return to regular vigorous activity.

INTRODUCTION

Lower extremity injuries have accounted for 50-52.8% of combat casualties in Operation Iraqi Freedom and Operation Enduring Freedom [2, 3, 6]. The advancement of primary vascular repair and management of extremity injuries and soft-tissue infections have resulted in decreased rates of primary amputations and increased the number of limb salvage patients [7]. Lifetime costs for amputees are higher and quality-of-life scores lower when compared to their limb salvage counterparts, although functional outcomes appear similar [13, 14]. However, compared to amputees who initially have a greater range of motion, less pain, more prosthetic options and appear to have a larger support group [8], limb salvage patients tend to recover more slowly [9] and have significantly fewer orthotic options. Restoring pre-injury function, strength and range-of-motion to service members who have undergone limb reconstruction can be challenging, and in many cases, lasting disability is common.

Complex injuries to the lower extremity and spine often involve nerve damage resulting in partial paralysis, manifesting in loss of ankle dorsiflexion and eversion known as a 'drop foot' pathology [22]. The resulting motor and proprioceptive deficits lead to walking and running difficulties as the toes drag and cannot clear the ground during the swing phase of the gait cycle [23]. Central nerve damage may also cause a loss of plantarflexion power, resulting in insufficient forward propulsion with each step on the involved side [88]. Ankle-foot orthoses (AFO) are the conventional treatment for drop foot and other nerve palsies of the lower extremity [22]. Wounded service members returning from Iraq and Afghanistan requiring AFO represent a cohort of young, pre-morbidly fit and goal oriented individuals whose functional and mechanical needs are unique compared with other individuals suffering from drop foot, such as stroke or cerebral palsy patients. Traditional AFO issued to service members following these injuries improved walking velocity and bring gait characteristics closer to functional ambulation, but in many cases will not provide enough stability and dynamic support to allow a complete return to normal function or athletic activities. For service members outfitted with traditional AFO, it would be difficult to meet fitness standards, remain on active duty or be deployed.

The advent of dynamic AFO may provide an invaluable tool for improved rehabilitation and long-term function of limb salvage patients. Dynamic carbon fiber AFO store potential energy during the stance phase of the gait cycle and return this energy for a normal push-off [29, 41]. This carbon-fiber design potentially offers more efficient, secure walking and running by providing mechanical energy return via posterior struts and medial-lateral and rotational stability from the cuff and foot components [34]. While initial examinations of these devices have demonstrated improved functional outcomes in soldiers with limb salvage [42], to our knowledge, the kinetic and kinematic gait changes associated with dynamic AFO use facilitating the ability to return to combat duty have not been reported.

CASE REPORT

On 24 August 2007, a 27 year old male active duty infantry officer suffered poly-trauma with loss of consciousness following a Humvee rollover in western Afghanistan. He suffered a posterior left hip dislocation with intra-articular fragmentation and acetabular fracture, left sacroiliac joint diastasis, right superior and inferior rami fractures, right wrist fractures, multiple facial fractures, a left pneumothorax, bladder extravasation in his abdomen, and severe nerve damage resulting in near total paralysis and drop foot of the left lower limb. One month post injury the service member's physical examination noted no movement of his left lower extremity with the exception of his left quadriceps (2/5 strength with manual muscle testing), and decreased light touch sensation throughout L4-5 and S2-S5 dermatomes. He spent two months in intensive care at Walter Reed and Brooks Army Medical Centers, and was then transferred to the Tampa Polytrauma Rehabilitation Center. Over the course of eight months, the service member progressed from being bedridden to electronic and then manual wheelchairs, and began assisted walking in a heated therapy pool. The service member was actively engaged in physical therapy twice daily, five days per week, where he was initially prescribed an off-the-shelf AFO while awaiting a custom solid plastic AFO. Ten months post-operatively he was able to ambulate with an antalgic gait using a straight single point cane and his custom plastic AFO. One year post-operatively the service member had significant left lower extremity atrophy of his quadriceps, hamstrings, and anterior and posterior compartments of his lower leg. He also exhibited decreased light touch sensation in the left lateral thigh, and anterior and posterior lower leg. Manual muscle testing of his left knee was 3/5 strength for flexion and extension, and left ankle dorsiflexion and plantarflexion was 2/5. Left hip extension was 0/5 during 30° of hip abduction and 10° of adduction, and passive left hip flexion range of motion was 85° with the knee flexed. Heterotopic ossification of his left hip was successfully treated by radiation oncology with a single radiation dose after failed conservative management. The service member was diagnosed with complex regional pain syndrome, with 8/10 pain in his left lower extremity which was nonresponsive to maximum tolerated doses of

Neurontin, Lyrica, Clonidine, and anti-inflammatories. He was treated with series of three lumbar interlaminar epidural steroid injections and two lumbar sympathetic chain ganglion blocks which provided 70% pain relief. Follow-up electrodiagnostic studies performed approximately 18 months post-injury continued to show no volitional motor unit action potential in the tibialis anterior and intrinsic foot muscles and additional findings consistent with lumbosacral plexus injury. Due to his significant left lower extremity pain, his worsening left lower extremity atrophy, and his inability to participate in functional and recreational activities compared to his amputee counterparts, the service member and his surgeons were actively considering an elective below the knee amputation.

In February 2009, the service member was prescribed a dynamic ankle-foot orthotic (DAFO) designed to provide energy storing capability on the rear shank (Dynamic Bracing Solutions, San Diego, CA) for his left lower extremity drop foot. Twelve days after being fitted with the device, he was able to run a 40-yard dash in six seconds without any running practice. Since he began wearing the DAFO in May 2009, the service member has obtained a level of fitness required to pass the Army Physical Fitness Test, raced in triathlons, remained on active duty, taken command, and redeployed to Iraq in 2010 with the device. The use of a DAFO and subsequent return to an active lifestyle has allowed him to discontinue the use of all pain medication and resulted in marked return of gastrocnemius muscle definition, requiring modification of the DAFO to accommodate muscle mass. The purpose of this case study was to examine differences in gait mechanics with and without the DAFO after wearing the DAFO for approximately three years.

METHODOLOGY

Study procedures were approved by the University of Hawaii Committee for Human Studies, and the service member completed an approved written Informed Consent. Anthropometric data (height, weight, leg length and joint width) were collected; height was determined using a stadiometer (model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and body mass measured using a Befour PS6600-ST scale (Befour, Inc., Saukville, WI, USA). A three-dimensional (3D), 13-camera Vicon MX 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, MA, USA) embedded flush with the floor surface were used to collect kinetic data during walking and running trials. Kinematic data was collected at 240Hz; time synchronized with digital video collected at 60Hz and kinetic data collected at 480Hz, and then smoothed using a Butterworth filter with 8 Hz cut-off. Following anthropometric measurements, 27 reflective markers were applied using the plug-in-gait lower limb and thorax model. The service member was instructed to walk and run at maximum comfortable self-selected velocities not to exceed 4.0 m/s ± 10% down an 18-meter runway. To ensure consistent velocity, Speedtrap II (Brower Timing Systems, Draper, Utah, USA) infrared sensors were placed four meters apart in the middle one-third of the runway. Walking and running gait were assessed using three successful trials for each foot in each condition (with and without DAFO). 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 [50, 51]. Kinematic and kinetic variables were calculated as the mean of three successful trials for each foot in each condition. A successful trial was defined as completion of the pass through the field at a consistent walking and running velocity, and landing with one foot completely on the force plate with no obvious change in stride [50-52]. Walking and running trials were completed in running shoes only and the DAFO with running shoes as footwear has been shown to influence gait pattern [26, 41, 89].

RESULTS

Mean spatial-temporal variables are reported in Table 5.1. The kinetic and kinematic values from shoes only and DAFO conditions for walking and running are displayed in Table 5.2. All moments reported in this study are external moments. Velocity and stride length increased with the use of a DAFO. Step length measurements more closely resembled the uninvolved leg in the running condition, but step length differences were greater during walking. Double-stance time decreased while wearing the DAFO and the amount of time spent in float during running increased.

Table 5.1. Case Study One: Mean Spatial-Temporal Variables

	Walking		**Running**	
	Shoes Only	**DAFO**	**Shoes Only**	**DAFO**
Velocity (m/s)	1.51	1.83	3.08	3.47
Cadence (steps/min)	121	118	160	162
Step Length of involved leg (m)	0.79	0.98	1.26	1.35
Step Length of uninvolved leg (m)	0.75	0.88	1.07	1.26
Stride Length (m)	1.54	1.86	2.33	2.61
Double-Stance / Double-Float Time (% of gait cycle)	19.92	17.69	27.05	30.11

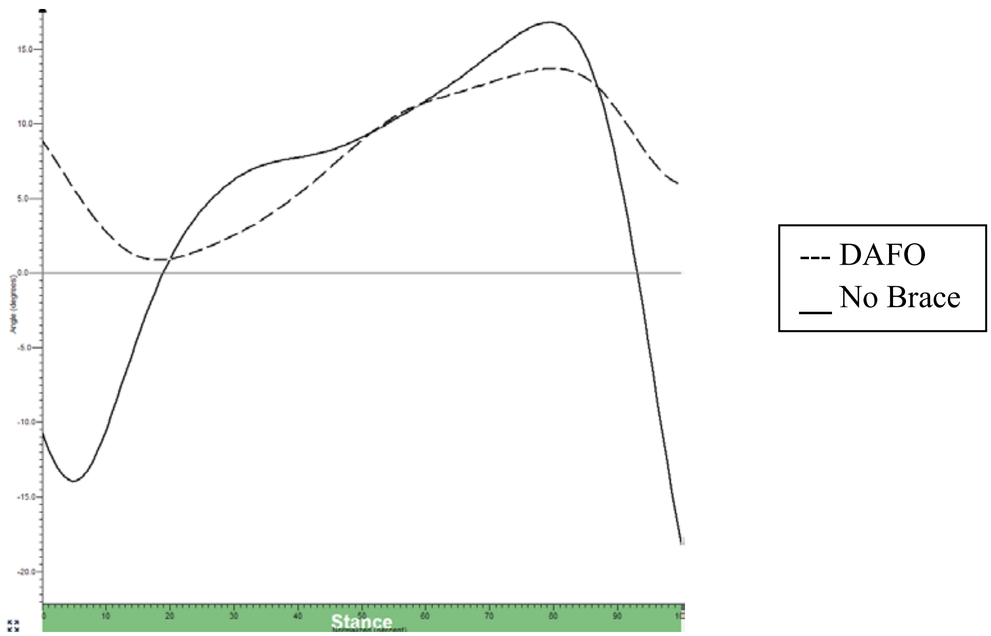
Drop foot was successfully corrected at heel strike, with 8.64° of dorsiflexion while walking in the DAFO, versus 8.99° of plantarflexion while walking in shoes. The involved limb remained in dorsiflexion throughout remaining gait phases due to configuration of the DAFO (Figure 5.1). Maximum ankle plantarflexion and knee flexion moments increased with the use of the DAFO while walking, although a substantial decrease in knee flexion moment at loading was noted during running. Ankle power generation was reduced with the DAFO while walking, despite an increase in walking velocity. However, the service member was able to generate greater power in the DAFO while running (Figure 5.2). Hip flexion moment during early stance was substantially reduced in both walking and running. The service member never reached hip extension during any gait trial, and exhibited a high degree of anterior tilt regardless of brace condition (see Figure 5.3).

Table 5.2. Case Study One: Kinetic & Kinematic Values for Shoes Only & DAFO in Walking and Running Conditions

	Walking		Running	
	Shoes Only	**DAFO**	**Shoes Only**	**DAFO**
Ankle Position at Initial Contact (°)	-8.99	8.64	-2.31	5.66
Ankle Position at Toe Off (°)	-17.35	8.2	-13.67	3.91
Dorsiflexion Excursion (°)	25.57	5.91	49.28	13.76
Eversion Excursion (°)	30.31	15.12	35.39	21.05
Min Dorsiflexion of Swing Leg (°)*	-28.48	6.05	-31.99	2.21
Max Plantarflexion Moment (Nm/kg)	-0.10	-0.48	-0.10	-0.17
Max Ankle Power Generation (W/kg)	3.68	2.95	7.18	10.93
Max Ankle Power Absorption (W/kg)	-1.23	-0.57	-7.68	-3.52
Knee Position at Toe Off (°)	48.15	33.49	19.6	15.94
Max Knee Varus Velocity (°/sec)	65.49	131.66	665.98	132.03
Knee Flexion Moment – Loading (Nm/kg)	0.68	0.96	1.46	0.15
Hip Position at Initial Contact (°)	42.71	45.91	47.02	45.12
Hip Position at Toe Off (°)	15.97	6.33	4.97	4.29
Hip Flexion Moment (Nm/kg)	1.25	0.76	8.30	0.88
Anterior Pelvic Tilt (°)	28.55	27.66	32.97	36.55
Pelvic Internal Rotation (°)	4.34	4.76	4.01	5.97
Pelvic External Rotation (°)	-18.13	-21.74	-14.73	-11.94
Vertical Ground Reaction Force (N/kg)	11.99	13.3	23.98	31.22
Loading Rate (N/s)	9756.14	9379.45	17598.37	21769.27

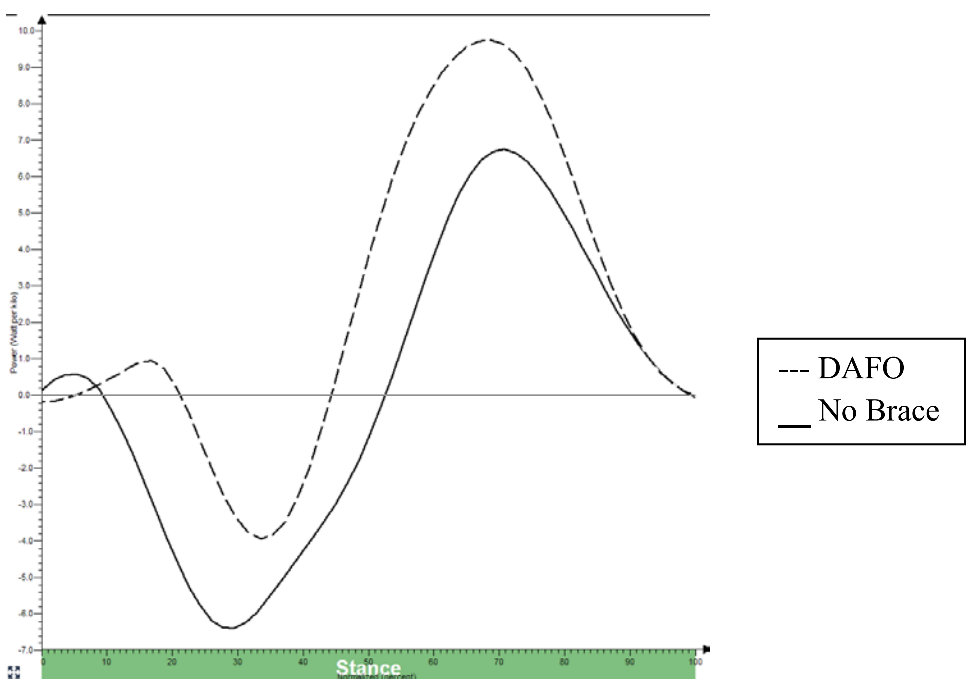
Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-)
**Swing Leg was involved side*

Figure 5.1. Dorsiflexion / Plantarflexion Angles during Stance



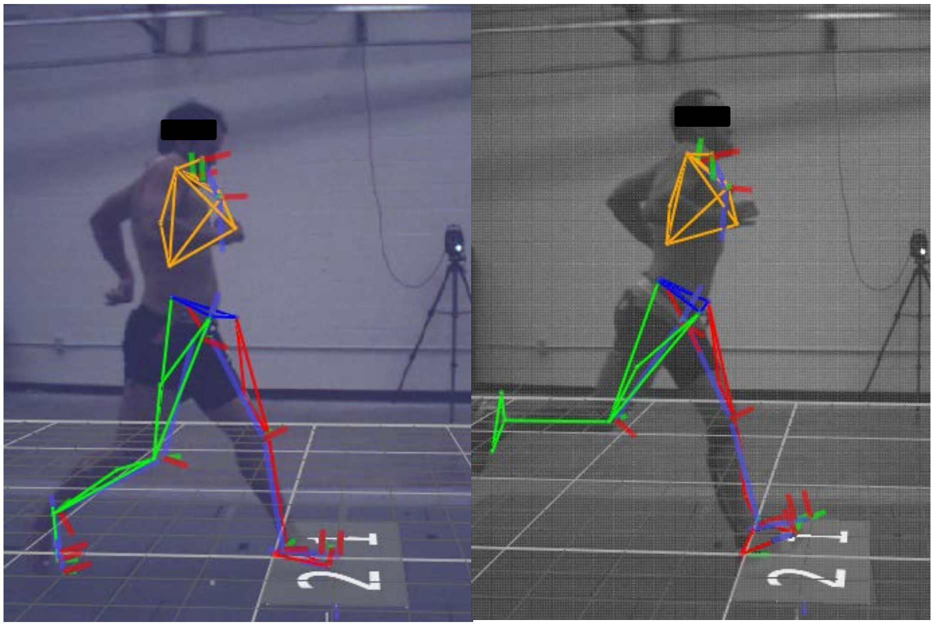
Dorsiflexion (+), Plantarflexion (-)

Figure 5.2. Running Ankle Power during Stance



Absorption (-), Generation (+)

Figure 5.3. Running Body Posture: Service Member without AFO vs. with DAFO



DISCUSSION

This service member returned to vigorous activities with use of the DAFO, continued on active duty and returned to combat. This was likely facilitated by the observed improvements in spatial-temporal parameters and kinematic and kinetic variable improvements about the ankle, knee, and hip while wearing the DAFO. Individuals with severe lower extremity injuries with loss of peroneal nerve function lack muscular control, resulting in the inability to dorsiflex and evert the foot, as well as extend the toes [23]. This 'drop foot' condition is characterized by the slapping of the foot after heel strike or no existing heel strike, and dragging of the toe during the swing phase [22, 25-27]. Hemiparetic gait deficits in general include a decrease in velocity, shorter step and stride length, and reduced stance time [40]. The service member exhibited these deficits in the shoes only condition, values which improved with the use of his DAFO. Walking velocity increased by 31.70 cm/sec, which was greater than the clinically relevant difference of 20 cm/sec used by de Wit [32]. While wearing the DAFO, the service member was able to perform a running pace of 3.47 m/s, equating to a 7:43 minute per mile pace, which if maintained would result in a passing score for the Army Physical Fitness Test two-mile aerobic component [90]. Cadence, stride, and step lengths were similar to the normative measures reported during fast walking by Oberg et al. (1993) [64]. Despite an increase in the length of step for both the involved and uninvolved legs, symmetry between the two steps did not return, possibly due to the weight of the DAFO. Similar findings of improved velocity and cadence but not step length symmetry were reported by Tyson and Thornton (2001), whose subjects indicated the weight of the AFO as potentially affecting function [79].

The DAFO assisted in producing biomechanical gait values that more closely resemble those exhibited by an uninjured individual. The cuff around the proximal tibia produced controlled tibial advancement in the absence of triceps surae force production [91]. The service member has also experienced sufficient gastrocnemius hypertrophy in the last year to warrant modification of the cuff. The DAFO corrected the inability to dorsiflex, enabling the service member to complete a successful heel-strike pattern during both walking and running as evidenced by the increased plantarflexion moment prior to midstance. The design of the brace prevented toe drag during swing: the involved limb finished toe-off in a dorsiflexed attitude during both walking and running, and remained in dorsiflexion throughout the swing phase of gait. Despite never reaching plantarflexion values at toe-off, dorsiflexion angles during swing were maintained close to neutral, resembling values of an uninjured runner [62, 63].

The DAFO also controlled the amount of dorsiflexion and eversion excursion in stance. Dorsiflexion excursion in both walking and running conditions decreased by 72-76% to resemble measures seen in a healthy population (19.24°) [92]. Increases in dorsiflexion/plantarflexion excursion are related to decreased ankle stiffness [93] and an increase in lateral ankle ligament strain. Larger excursions of ankle motion in the frontal and sagittal planes create greater ankle instability [94-96]. The service member emphasized his perception of ankle instability as one of the greatest barriers to return to running, and felt that the support the DAFO provided the necessary stability to allow higher intensity activities.

Increased ankle power also contributed to the service member's return to running; ankle power generation increased to levels seen in the uninjured runners as described by Novacheck (1998) [66]. Although ankle power is directly related to an athlete's velocity[66], a decrease in ankle power generation while walking was noted despite the clinically relevant increase in walking velocity [97]. This was likely due to the energy-return capabilities of the DAFO posterior strut producing power generation with less effort.

Changes in ankle push-off have been found to be inversely related to hip joint moments achieved by a trade-off between ankle and hip muscle requirements during gait [98]. Substantial decreases in hip flexion moment were evident in both walking and running conditions while wearing the DAFO, in concert with the improved ankle power generation. Very little change occurred in the degree of hip flexion at initial contact, which has been documented in other studies without explanation [99]. The service member's hips remained in flexion at toe-off, and never reached extension during the gait cycle, most likely a result of the pelvic trauma, heterotopic ossification and subsequent lordotic nature of the pelvis. Additionally, the change in alignment of the ankle and knee caused by the DAFO influenced the GRF effect on the knee. The slight dorsiflexion alignment with a flexible heel lift insert caused the lateral axis of the knee to lie more anteriorly, increasing the amount of time the GRF vector lies posterior to the knee, noted by the increase in knee flexion moment at loading response during walking [66, 91].

A limitation for this case study included the lack of baseline testing of this service member's functional capabilities in our laboratory setting prior to beginning use of the DAFO in 2009. Gait analysis was completed at a different testing site with non-standardized techniques relative to the present analysis. Spatial-temporal parameters noted in the previous gait analysis report for the DAFO were similar to three-year outcomes, suggesting the DAFO provides immediate ability to return to fast walking velocities; however running gait was not assessed during the baseline analysis.

The most important finding of this case study is that pre-morbidly fit populations, such as military personnel suffering from trauma to the lower extremity, have an alternative to amputation that provides the ability to return to vigorous activity. This brace has allowed the service member in this case study to regain combat readiness, remain on active duty, and deploy to a combat environment as an infantry officer in the Army. The service member credits the use of a DAFO for a substantial overall improvement in quality-of-life, both professionally and personally. The advancement of surgical techniques in combat medicine and subsequent increase in limb salvage cases in the military population makes design and testing essential to advancing current ankle-foot orthoses to optimal functionality, in order to assist service members' return to duty.

Future studies should continue to assess varying types of DAFO in this pre-morbidly fit and goal oriented population, and longitudinal studies are needed evaluate long-term physiological, biomechanical, and quality-of-life outcomes of a device that allows this cohort to return to their pre-injury level of activity.

Case Study Two

Bilateral Dynamic Ankle-Foot Orthoses and Forearm Crutched Gait Analysis of Service Member with Incomplete Lower Extremity Paralysis as a Result of Viral Meningitis

KEYWORDS
Viral Meningitis, Ankle-Foot Orthosis, Forearm Crutches, Gait

ABSTRACT

This case study examined the biomechanical gait analysis of a young, medically retired Army combat medic with bilateral peroneal nerve palsy following a severe case of viral meningitis. Lingering effects of her illness included memory loss, urinary and fecal incontinence, severe muscle wasting, lower extremity neuropathy and bilateral drop foot resulting in the need to ambulate with bilateral dynamic ankle-foot orthoses (DAFOs) and forearm crutches. Baseline biomechanical kinetic and kinematic gait measures of a bilateral DAFO and forearm crutch gait were established and then compared to published healthy gait norms. Kinematic and kinetic walking gait trials on this combat medic were collected using a 13-camera Vicon Motion capture system and two AMTI force plates. There were notable differences in the level of pathology between the veteran's left and right lower extremities, as well as the gait deviations in hip kinematics seen with the use of forearm crutches and musculoskeletal weakness. Maximum walking velocity fell between slow and normal gait velocities of women of similar age, with below normal and slightly asymmetrical step length, as the peroneal neuropathy was more severe on the patient's left side. Despite successful correction of drop foot with the bilateral DAFOs, gait deviations likely due to severe muscle atrophy at the hip were evident throughout the kinetic chain. Marked lordosis likely due to dependency of the forearm crutches was

noted, as well as hyper-flexion and external rotation forming a circumduction gait as well as mal-alignment of the left and right extremities. Severe muscle atrophy and learned compensatory gait patterns post-illness may have contributed to this veteran's pathologic gait despite improved nerve conduction throughout the kinetic chain. Future studies should examine this veteran and other patients with peroneal neuropathy after a rigorous strength and gait re-training rehabilitation program, as early rehabilitation of these foci may minimize compensatory motor patterns compounded by atrophy and immobilization.

INTRODUCTION

Viral or "aseptic" meningitis is typically introduced to a host by an enterovirus, producing symptoms in 10-15 million patients per year in the United States [100]. Second only to viruses producing the common cold, sixty different types of enteroviruses can cause illness and disease in humans [101]. Enteroviruses account for 85% of all viral meningitis cases, but only one in 1000 cases ends in meningitis [102, 103]. The viruses travel through both direct and indirect modes of transmission, including respiratory and throat secretions, and fecal contamination [103], creating an inflammatory response to the meninges and sparking a subsequent hallmark of symptoms including headache, fever, stiff neck, and sensitivity to light [100, 104]. Viral meningitis, though less severe than its bacterial counterpart with less than one percent of patients experiencing lasting effects, can sometimes present rare complications and permanent debilitating conditions in association with extended hospital stay [102]. Due to the extensive list of factors and agents associated with viral meningitis, less than 10% of cases are etiologically identified due to marked difference in clinical manifestations between virus types [102].

CASE REPORT

A 20-year-old active duty female combat medic initially presented on 4 June 2009 with a combination of symptoms including headache, photophobia, fever, and stiff neck which led to a diagnosis of aseptic meningitis. Severe persistent symptoms added additional diagnoses of post spinal tap cephalgia and eventual viral encephalomyelitis over a one-week period, which were exacerbated by respiratory failure requiring mechanical ventilation and complications related to gastric ulcer and reflux esophagitis. Lingering effects of the illness included persistent bilateral lower extremity paralysis and areflexia along with short term memory deficits. In a three-year follow-up, the medically retired veteran denoted severe muscle weakness to the pelvis, proximal, and distal lower extremity, and now ambulates with bilateral dynamic ankle-foot orthoses (DAFO) (Dynamic Bracing Solutions, San Diego, CA) and forearm crutches. Previously wheelchair-bound, she has advanced to the current stage of ambulation and has partially recovered conductivity to the distal lower limb.

Prior to receiving bilateral DAFOs, the veteran was unable to accomplish tasks of daily living (TDL) and successful ambulation with traditional solid plastic ankle-foot orthotics (AFO). The advent of dynamic AFO may provide an invaluable tool for improved rehabilitation and long-term function of patients with lower extremity neuropathy. Dynamic carbon fiber AFO store potential energy during the stance phase of the gait cycle and return this energy for a normal push-off [37, 41]. This design potentially offers more efficient and secure walking and running by providing mechanical energy return via posterior carbon-fiber struts and medial-lateral and rotational stability from the cuff and foot components [34]. While initial examinations of these devices have demonstrated improved functional outcomes in service members with lower extremity trauma [42], to our knowledge, the kinetic and kinematic gait changes associated with DAFO use resulting from viral meningitis and associated side-effects have not been reported.

Based on a recent clinical evaluation of the service member's strength levels and motor function, the Physical Medicine and Rehabilitation physician working with this service member prior to medical retirement believed improvements could be made to the patient's ambulation, TDL, and quality-of-life if she strengthened the muscles of her pelvis and lower extremity. Therefore, the purpose of this case study was to establish baseline biomechanical gait measures with bilateral DAFO and forearm crutches, with the intent to compare to a future repeated measures case study involving a one-year rehabilitation program intervention for this retired Army combat medic.

METHODOLOGY

The case study procedures were approved by the University of Hawaii Human Studies Program. The veteran volunteered to participate and completed an approved written Informed Consent. Anthropometric data (height, weight, leg length and joint width) were collected; height was determined using a stadiometer (model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and weight was measured using a Befour PS6600-ST scale (Befour, Inc., Saukville, WI, USA). A three-dimensional (3D), 13-camera Vicon MX 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, MA, USA) embedded flush with the floor surface were used to collect kinetic data during walking trials. Kinematic data was collected at 240 Hertz (Hz), time synchronized with digital video collected at 60 Hz and kinetic data collected at 480 Hz, and then smoothed using a Butterworth filter with 8 Hz cut-off. Following anthropometric measurements, 27 reflective markers were applied using the Plug-in-Gait Lower Limb and Thorax model. The veteran was instructed to walk at a maximum comfortable self-selected velocity not to exceed 4.0 m/s ± 10% down an 18-meter runway. To ensure consistent velocity, Speedtrap II (Brower Timing Systems, Draper, Utah, USA) infrared sensors were placed four meters apart in the middle one-third of the runway. Forearm crutches were required for ambulation, but reflective surfaces were taped and did not disrupt kinematic data collection within the testing field. Walking gait was assessed using three successful trials for each foot with the use of both bilateral DAFO and forearm crutches. 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 [50, 51]. Kinematic and kinetic variables were calculated as the mean of three successful trials for each foot. 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 [50-52]. Walking trials were completed in the DAFO with walking shoes and forearm crutch condition only, as ambulating without DAFO was currently not possible for the patient. Walking shoes were worn with the DAFO based on device design and because footwear has been shown to influence gait pattern [26, 41, 89].

RESULTS

This case study reported on a female veteran with the following characteristics: 23 years of age, height: 161.5 cm, and weight: 65.7 kg with two DAFO. Table 5.3 includes spatial-temporal parameters from the average of three walking trials for the left and right extremity sides.

Table 5.3. Case Study Two: Walking Spatial-Temporal Parameters in Bilateral DAFO with Forearm Crutches

Velocity (m/s)	1.16
Cadence (steps/min)	107.54
Left Step Length (m)	0.67
Right Step Length (m)	0.62
Stride Length (m)	1.29
Left Step Time (s)	1.38
Right Step Time (s)	1.41

The service member veteran favored her right side for both walking step length and time. Notable differences in kinetic data for the pelvis and left and right lower extremities are reported in Table 5.4, and kinematic parameters for the pelvis and both left and right lower extremity are reported in Table 5.5. All moments discussed in this study are external. Figure 5.4 provides a visual schematic of step length differences. Differences in the brace design to support the level of pathological involvement may account for some differences seen in kinematic measures between the left and right lower extremities, particularly in positions and velocities about the ankle.

Table 5.4. Case Study Two: Walking Kinetics in Bilateral DAFO with Forearm Crutches

	Left	Right
Maximum Ground Reaction Force (N/kg)	9.71	9.33
Maximum Braking Force (N/kg)	-0.86	-0.47
Maximum Propulsion Force (N/kg)	0.64	0.80
Maximum Ankle Plantarflexion Moment - Loading Response (Nm/kg)	-0.32	-0.14
Maximum Knee Flexion Moment - Loading Response (Nm/kg)	0.49	0.15
Maximum Knee Flexion Moment - Push-off (Nm/kg)	0.49	0.16
Maximum Knee Extension Moment (Nm/kg)	-0.38	-0.56
Maximum Knee Adduction Moment (Nm/kg)	0.27	0.35
Knee Stiffness (Nm/rad)	0.38	0.04
Maximum Hip Flexion Moment - Loading Response (Nm/kg)	0.98	1.26
Maximum Hip Adduction Moment (Nm/kg)	0.76	0.59

Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

Table 5.5. Case Study Two: Walking Kinematics in Bilateral DAFO with Forearm Crutches

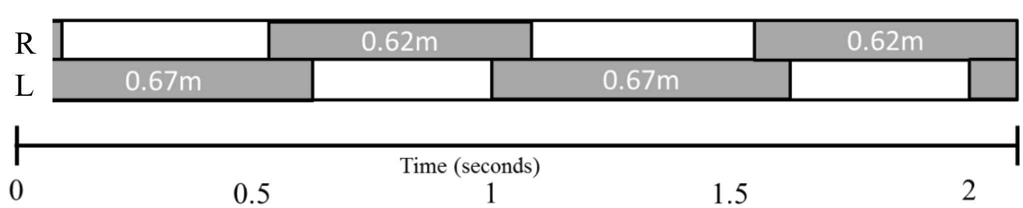
	Left	Right
Ankle Position Dorsiflexion/Plantarflexion Position at Initial Contact (°)	7.24	1.94
Ankle Position at Toe Off (°)	3.46	1.55
Ankle Maximum Plantarflexion Velocity (m/s)	-94.25	-63.49
Ankle Timing of Maximum Plantarflexion Velocity (% of Stance)	1.71	93.50
Ankle Inversion/Eversion Position at Initial Contact (°)	-6.31	1.25
Ankle Maximum Eversion Velocity (m/s)	114.36	61.33
Ankle Internal/External Rotation Position at Initial Contact (°)	28.88	-15.92
Ankle Internal/External Rotation Excursion (°)	28.24	10.79
Mean Foot Progress Angle During Stance (°)	5.76	-1.89
Knee Flexion/Extension Position at Initial Contact (°)	7.49	-0.92
Knee Flexion/Extension Excursion (°)	11.57	23.51
Knee Flexion/Extension Position at Toe Off (°)	19.06	22.59
Knee Mean Varus Velocity (m/s)	11.58	48.66
Knee Internal/External Rotation Position at Initial Contact (°)	15.36	-5.64
Knee Maximum Internal Rotation (°)	21.04	-4.69
Knee Mean Internal/External Rotation Velocity (m/s)	20.91	2.43
Hip Flexion/Extension Position at Initial Contact (°)	47.59	39.48
Hip Flexion/Extension Position at Toe Off (°)	2.54	4.52
Hip Abduction/Adduction Position at Initial Contact (°)	-9.18	3.70
Hip Internal/External Rotation Position at Initial Contact (°)	-59.97	10.00
Hip Internal/External Rotation Excursion (°)	37.72	9.67
Hip Internal/External Rotation Position at Toe Off (°)	-23.88	18.14
Pelvis Angle Excursion (°)	10.42	11.31

Plantarflexion (-), Dorsiflexion (+), Flexion (+), Extension (-), Internal Rotation (+), External Rotation (-), Abduction (-), Adduction (+)

DISCUSSION

Notable differences were apparent between the levels of pathology of the left and right extremities, as well as the gait deviations in hip kinematics seen with musculoskeletal weakness and the use of forearm crutches. Baseline measures established spatial-temporal (ST), kinematic, and kinetic gait deviations and were then compared to normative values from the literature. The maximum comfortable walking velocity achieved by this service member with the forearm crutches fell between the slow and normal gait walking velocities of women ages 20-29 as reported by Oberg et al. (1993) [64]. This 68.5% of normal velocity matched the findings of Vankoski et al. (1997) [105] who reported young adults with myelomeningocele ambulating under the same conditions measured 68.3% of normal pediatrics in their laboratory. Cadence of this veteran also fell below established normative values, with 17 steps less per minute than uninjured women of her same age [64]. Figure 5.4 displayed an asymmetry of step length and time with a slight decrease in length of the right side, which indicated the attempt to decease time spent in left single limb support, likely due to a greater level of neuropathy and subsequent atrophy on the veteran's left side. Walking gait with bilateral DAFO and forearm crutches was the only condition collected during this case report, as current levels of neuropathy and related musculoskeletal atrophy precluded this service member veteran from walking without the aid of these assistive devices.

Figure 5.4. Variation in Step Length Parameters



Injuries to the lower extremity and spine associated with illness can involve nerve damage resulting in partial paralysis, manifesting in bilateral loss of ankle dorsiflexion and eversion known as a 'drop foot' pathology [22]. The resulting motor and proprioceptive deficits lead to walking and running difficulties as the toes drag and cannot clear the ground during the swing phase of the gait cycle [23]. Ankle-foot orthoses (AFO) are the most common treatment for drop foot and other nerve palsies of the lower extremity [106]. This veteran was custom-fitted with two carbon-fiber dynamic ankle-foot orthoses (DAFO), which had slight design differences based on the structural alignment of each lower extremity [76]. Prior to DAFO fitting, she wore a knee-ankle-foot orthosis (KAFO) on her left side and a traditional AFO on her right side, but the capabilities of the DAFO were able to provide more successful ambulation with less assistance [76]. The devices corrected the inability to dorsiflex on both left and right lower extremities, enabling the veteran to complete a successful heel-strike pattern during walking, although the left ankle exhibited five degrees greater dorsiflexion at initial contact than the right. The design of the brace also prevented toe drag during swing on both sides with slightly dorsiflexed ankle angles at toe-off that remained throughout the swing phase. Despite not reaching normal plantarflexion values at toe-off, dorsiflexion angles during swing remained close to neutral, resembling values of an uninjured walker [62].

Gait deviations and subsequent compensatory movements seen in this case study may have been partially attributed to muscle weakness, joint position or muscle contracture [107], particularly in the hips. The interdependency of gait parameters caused a cascade of gait changes in the kinetic chain [73]. Motion of the hip was measured in relation to the pelvis, and pelvic sagittal range of motion approached three times the normal range during the gait cycle in this veteran (11° versus 4°, respectively) [107]. This anterior pelvic tilt was most likely related to the patient's forward posture with the forearm crutches. Resembling a controlled fall, the patient reached forward with the forearm crutches alternating with each step, and used gravity to assist with frontward progression. The pelvis and hip are normally internally rotated during early stance [107], but the excessive hip flexion on the veteran's left side coupled with a high degree of hip external rotation at initial contact initiated a circumduction gait of the contralateral limb (Figure 5.5). She contacted the ground during left stance with a near neutral knee flexion position but with an internally rotated foot (28.88°). Initial contact on the right side, however, reveals an internally rotated hip with an externally rotated (15.92°) ankle position. The subsequent toe-out gait seen on the right lower extremity has been related to a reduced hip external rotation moment in research with healthy subjects, reducing torsional load on the femur [108], a finding that was supported by the moments seen in this patient. A lesser knee varus velocity was denoted on the left side versus the right, which can be attributed to the opposing alignment seen in each limb. The anatomical position of an externally rotated hip and internally rotated ankle contributed to a lower varus velocity due to the inability for the knee to move into varus within that alignment position. The right extremity mean knee varus velocity, however, reached levels four times greater than the left due to the coupling of the internally rotated hip with an externally rotated ankle.

The veteran's gait posture in terminal stance (Figure 5.6) uncovered a tendency to increase anterior tilt of the pelvis with the placement of weight onto the forearm crutches [105] such that the hips remained in a biomechanically flexed position, a degree of knee flexion normally seen at push-off, and dorsiflexion supported by the DAFO [109]. This posture precluded sufficient forward propulsion as noted in the near zero level of propulsive force for both extremities and combined with hip flexor weakness when the knee was flexed resulted in circumduction to ensure clearance of the foot during swing.

Figure 5.5. Hip and Ankle Joint Internal/External ROM during stance

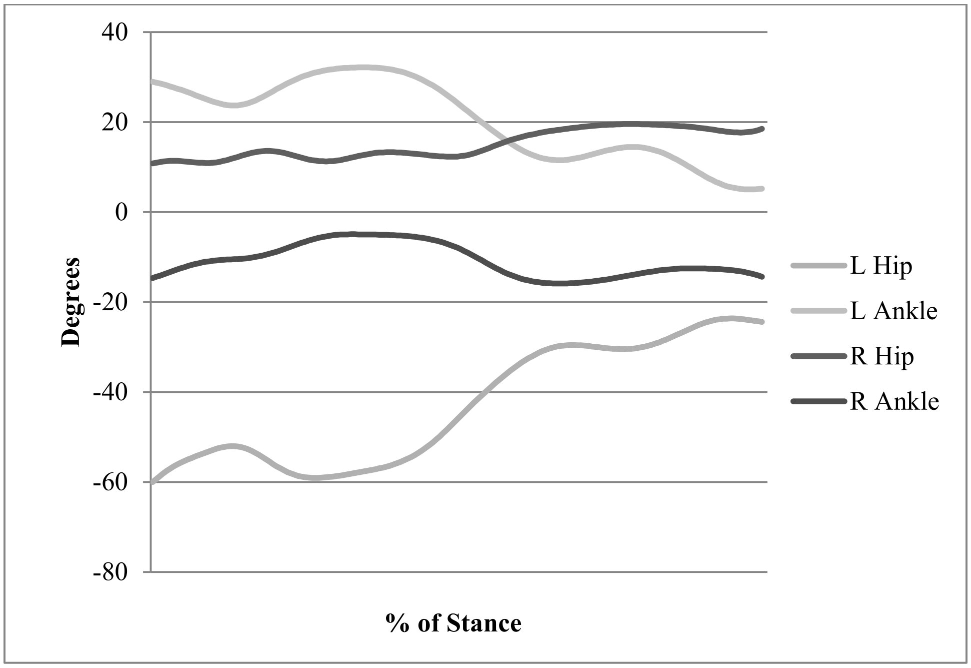


Figure 5.6. Gait Posture with Bilateral AFO and Forearm Crutches

This baseline gait analysis revealed several gait deviations needed to ambulate for activities of daily living, despite the successful correction of bilateral drop foot with the

DAFO. Energy requirements required to maintain these gait compensations were dramatically increased, based on the level of fatigue reported by the veteran to complete tasks of daily living. Despite neurological deficits, she had adequate nerve conduction during evaluation to successfully move her muscles proximally and distally in the kinetic chain, and may benefit from intensive rehabilitation to strength and retrain these muscles for more efficient ambulation. One limitation of this case study was a lack of comparison of gait without the forearm crutches of this individual, due to the level of atrophy and musculoskeletal weakness she has experienced over the last three years. The future rehabilitation of this veteran should focus on all major muscle groups of the lower extremity, with particular attention paid to the muscles of the hip and their contributions to normal gait. Improvements in hip flexion during knee flexion may strengthen the iliopsoas and quadriceps group to improve sagittal movements at the hip and eliminate circumduction movements induced by weakness in these muscle groups. Strengthening the muscles of the core, gluteus maximus, and hamstrings groups may assist in postural control, potentially reducing anterior tilt seen in the pelvis. Improving this area may facilitate disuse of the forearm crutches when tolerated by the veteran, and promote increases in propulsive ankle power at push-off.

In conclusion, the gait deviations seen in this veteran are likely due to learned gait patterns post-illness, combined with musculoskeletal weakness and neurological deficits. Future studies involving neuropathies should focus on early rehabilitation and gait retraining to prevent incorrect learned motor patterns compounded by atrophy and immobilization.

Case Study Three

A Comparison of Walking and Running Gait between Service Members with Combat-Related Injuries resulting in Limb Reconstruction or Transfemoral Amputation

ABSTRACT

This case study examined the biomechanical gait parameters of two service members injured in combat; one female Army Security Forces officer required a unilateral transfemoral amputation of the right lower extremity and a male Army Infantry officer with poly-trauma and subsequent nerve damage suffered near total paralysis and drop foot of the left lower extremity. Despite originally considering amputation of his non-functional limb, the service member with drop foot investigated a dynamic ankle-foot orthotic (DAFO) with energy storing capability, which allowed him to remain on active duty and deploy for a second tour while wearing the device. The service member amputee was provided two carbon fiber prosthetics, one for walking and a non-articulated running model, and has remained on active duty with continued modification of her prosthetics and socket. Results of gait analysis demonstrated normal and near-normal spatial-temporal parameters in both walking and running when compared to their individual gender published normative values, as well as ankle kinematics comparable to published able-bodied gait parameters. The service member with drop foot had improved sagittal angles and moments at the ankle, knee, and hip, greater ankle stability through decreased dorsiflexion excursion, and a marked increase in ankle power while running. The service member amputee exhibited less than normal power at the ankle, compensated for at the hip and likely the sound limb. Both service members displayed decreased knee flexion than normal during midstance while running, which resulted in decreased knee

moments. The service member amputee's non-articulated limb restricted knee flexion, and required a circumduction gait for running. Most notably, both service members credit their devices for substantial improvement in quality-of-life including total cessation of pain medication and return to regular vigorous activity.

INTRODUCTION

The number of lower extremity injuries sustained in the War on Terror has surpassed 50,000 since 2001[1]. Despite the improvements in combat medicine and subsequent increases in limb salvage surgeries, an estimated 1,653 amputations have been performed by military surgeons during this conflict [110]. Above and below the knee prosthetic technologies have continually advanced over the last decade to provide the best rehabilitation for amputees, and subsequent cost of devices have also increased, cited as $82,251 for projected five-year costs of maintaining a carbon fiber prosthetic limb [111]. These projections include devices for both activities of daily living and specialty devices for return to vigorous sports. The current total cost for one prosthetic device system is $45,563.17 [111]; the five-year projections are based on more than one type of device.

The advancement of primary vascular repair and management of extremity injuries and soft-tissue infections have resulted in decreased rates of primary amputations and increased the number of limb salvage patients [7]. Multiple device options for limb salvage patients are currently evolving and improving. Restoring pre-injury function, strength and range-of-motion to service members who have undergone limb reconstruction can be challenging. Although functional outcomes appear similar, lifetime costs for amputees are higher and quality-of-life scores are lower when compared to their limb salvage counterparts [13, 14]. Despite long-term equivalent functional outcomes, some limb salvage patients with severe injuries and projected lasting disability of their limb have consulted with their physicians regarding secondary amputation [11, 12].

Few studies have compared limb salvage patients to amputees. The largest research effort to date was conducted by the Lower Extremity Assessment Project (LEAP) Study Group, who examined both cost and long-term functional outcomes. Unfortunately the premorbidly fit military population was excluded from their study, and authors noted that conclusions drawn from the LEAP project may not adequately reflect outcomes in a military population [15]. The LEAP studies also did not include long-term biomechanical gait comparisons between limb salvage patients and amputees, which may prove an important factor when considering late amputation.

Newly designed dynamic prosthetic and ankle-foot orthotic devices store potential energy during the stance phase of the gait cycle to be returned later for a normal push-off [29, 41]. Both articulated and non-articulated prosthetic models are available, although amputees are encouraged to begin with the non-articulated model with a circumduction gait prior to trying an articulated device, in an effort to prevent inadvertent knee flexion [112]. There is a lack of research comparing biomechanical gait characteristics in the military transfemoral amputee and limb salvage populations. Therefore, the purpose of this study is two-fold: to analyze the walking and running gait characteristics of a transfemoral amputee using non-articulated prostheses, and compare these characteristics to the gait of a limb salvage service member wearing a dynamic ankle-foot orthosis (DAFO). All comparisons made between these service members that differ between genders are based on their respective deviations from published normative gait values, as the two service members differ in gender and stature.

CASE REPORTS

The service members in this case study sustained combat-related injuries during deployments to Iraq and Afghanistan. A 27-year-old male active duty Army infantry officer and focus of case study number one suffered poly-trauma with loss of consciousness following a Humvee rollover in western Afghanistan. He endured severe injuries proximal to the knee: a posterior left hip dislocation with intra-articular fragmentation and acetabular fractures, left sacroiliac joint diastasis, bladder extravasation in his abdomen, as well as severe nerve damage resulting in near total paralysis and drop foot of the left lower extremity. The second case study subject was a 25-year-old female active duty security forces Army officer who dismounted an IED resulting in a right near complete traumatic above the knee (AKA) amputation completed downrange, a contralateral superior femoral artery and vein laceration requiring shunting, reverse saphenous grafting, multiple prophylactic left lower extremity fasciotomies and residual blooming artifact from metallic shrapnel near the left medial meniscus.

METHODOLOGY

Study procedures were approved by the University of Hawaii Committee for Human Studies, and the service members completed an approved written Informed Consent. Anthropometric data (height, weight, leg length and joint width) were collected; height was determined using a stadiometer (model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and body mass measured using a Befour PS6600-ST scale (Befour, Inc., Saukville, WI, USA). A three-dimensional (3D), 13-camera Vicon MX 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, MA, USA) embedded flush with the floor surface were used to collect kinetic data during walking and running trials. Kinematic data was collected at 240Hz; time synchronized with digital video collected at 60Hz and kinetic data collected at 480Hz, and then smoothed using a Butterworth filter with an 8Hz cut-off. Following anthropometric measurements, 27 reflective markers were applied using the Plug-in-Gait Lower Limb and Thorax model. The service member amputee was instructed to walk in her walking prosthetic and run in her non-articulated running prosthetic at maximum comfortable self-selected velocities not to exceed 4.0 m/s ± 10% down an 18-meter runway. To ensure consistent velocity, Speedtrap II (Brower Timing Systems, Draper, Utah, USA) infrared sensors were placed four meters apart in the middle one-third of the runway. Walking and running gait were assessed using three successful trials for each side 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 [50, 51]. Kinematic and kinetic variables were calculated as the mean of three successful trials for each foot in each condition. A successful trial was defined as completion of the pass through the field at a consistent walking and running velocity, and landing with one foot completely on the force plate with no obvious change in stride [50-52]. Gait characteristics were then compared to previously collected walking and running gait data on the service member with limb reconstruction in shoes, with and without the DAFO.

RESULTS

A male service member with limb salvage (age: 32, height: 178.3 cm, body mass: 78.03 kg) outfitted with a DAFO and a female service member amputee (age: 29, height: 164.7 cm, body mass: 65.1 kg) equipped with walking and running prostheses using a suspension system, airtight socket, and carbon fiber nylons were compared in this case study. Mean spatial-temporal variables for both service members are reported in Table 5.6. Kinetic and kinematic measurements of the service members for both walking and running are displayed in Table 5.7.

Table 5.6. Case Study Three: Mean Spatial-Temporal Variables

	Walking		**Running**	
	DAFO	**Prosthetic**	**DAFO**	**Prosthetic**
Velocity (m/s)	1.83	1.54	3.47	2.72
Cadence (steps/min)	118	123	162	173
Step Length of involved leg (m)	0.98	0.76	1.35	0.95
Step Length of uninvolved leg (m)	0.88	0.74	1.26	0.93
Stride Length (m)	1.86	1.50	2.61	1.89
Double-Stance / Double-Float Time (% of gait cycle)	17.69	21.84	30.11	23.95

The service member amputee exhibited the most symmetrical step length, with 0.02 meter difference between the left lower extremity and right prosthetic. According to published normative spatial-temporal parameters established by Oberg et al (1993) [64], the male service member with limb salvage exceeded the average fast walking velocities for males of the same age, while the female service member amputee fell just below fast walking velocities for uninjured women of the same age. Both service members fell short of the normative cadence, or step frequency published for fast walking gait, with uninjured men ages 20-29 stepping an average of 140 steps per minute, while uninjured women ages 20-29 stepped at 153 steps per minute [64]. Despite the low cadence in both cases, both service members exceeded the normative values of step length for the gender-specific fast walking velocities [64].

Differences in the DAFO and walking and running prosthetics occurred throughout the kinetic chain, with more pronounced deviations seen at the ankle during walking and at the hip during running. The DAFO provided limited dorsiflexion and eversion excursions when compared to the walking prosthetic. During the stance phase of walking, the foot progression angle in the DAFO maintained an internally rotated attitude, while the prosthetic was positioned at a neutral angle. The DAFO provided a greater maximum plantarflexion moment during loading response with the controlled movement to foot-flat. Despite a greater absorption of power noted with the walking prosthetic, both the DAFO and prosthetic provided a near equal level of power generation during propulsion. Knee position was similar during initial contact and toe-off for both conditions, but knee varus velocity was substantially lower in the prosthetic due to device design. Despite the large range between peak varus velocities, the mean varus velocity during stance was not different. A greater amount of hip abduction during swing of the involved limb was noted with the running prosthetic, followed by a substantial lesser amount of hip flexion at initial contact. The involved hip of the service member amputee was externally rotated at the initiation of stance, while the service member with limb salvage struck the ground with the hip internally rotated. Due to the level of pelvic injury and subsequent lordosis of the service member with limb salvage, his hip never reached extension throughout the stance phase of gait. All moments reported in this study are external moments.

Table 5.7. Case Study Three: Walking and Running Gait Variables for DAFO and Prostheses

	Walking		Running	
	DAFO	**Prosthetic**	**DAFO**	**Prosthetic**
Ankle Position at Initial Contact (°)	8.64	-3.57	5.66	9.81
Ankle Position at Toe Off (°)	8.20	0.36	3.91	7.32
Dorsiflexion Excursion (°)	5.91	18.49	13.76	10.22
Eversion Position at Initial Contact (°)	9.22	31.94	10.30	-4.33
Eversion Position at Toe Off (°)	10.30	32.75	7.58	-3.77
Eversion Excursion (°)	15.13	58.35	21.05	5.53
Ankle Internal/External Rotation Position at Initial Contact (°)	14.86	-2.10	6.13	7.84
Foot Progress Angle, 30% of Stance (°)	18.02	-15.85	21.29	4.65
Min Dorsiflexion of Swing Leg (°)*	6.05	-8.76	2.21	6.89
Max Plantarflexion Moment (Nm/kg)	-0.48	-0.27	-0.17	-0.08
Max Ankle Power Generation (W/kg)	2.95	2.73	10.93	7.01
Max Ankle Power Absorption (W/kg)	-0.57	-1.09	-3.52	-3.96
Knee Position at Initial Contact (°)	-0.44	3.10	11.94	-0.93
Knee Position at Toe Off (°)	33.49	40.34	15.94	1.45
Max Knee Varus Velocity (°/sec)	131.66	53.15	132.03	39.38
Knee Flexion Moment – Loading (Nm/kg)	0.96	0.53	0.15	0.61
Hip Position at Initial Contact (°)	45.91	35.69	45.12	14.72
Hip Position at Toe Off (°)	6.33	-6.52	4.29	-4.51
Hip Rotation Position at Initial Contact (°)	-5.47	-17.39	8.59	-15.82
Hip Rotation Position at Toe Off (°)	5.67	-2.87	6.56	-15.42
Hip Abd/Adduction of Swing Leg (°)	-8.42	-9.12	-7.21	-28.58
Max Hip Flexion Moment (Nm/kg)	0.76	0.38	0.88	0.005
Max Hip Extension Moment (Nm/kg)	-1.44	-1.92	-2.56	-2.17
Max Hip Power Generation (W/kg)	1.18	2.13	1.20	0.11
Max Hip Power Absorption (W/kg)	-1.41	-3.55	-14.84	-2.16
Vertical Ground Reaction Force (N/kg)	13.3	12.22	31.22	25.11

*Plantarflexion (-), Dorsiflexion (+), Inversion (-), Eversion (+), Flexion (+), Extension (-), Abduction (-), Adduction (+), Internal Rotation (+), External Rotation (-), *Swing Leg was involved side*

DISCUSSION

The service members in this case study have both been able to return to vigorous activity with their respective assistive devices, as well as remain on active duty military status. The total cost difference between these soldiers' devices was approximately $70,000.00. Total device cost (two DAFOs) of the service member with limb salvage was $30,000.00, while the service member amputee had two prosthetics specific for walking and running totaling $100,000.00. The service member with limb salvage obtained two DAFOs: one streamlined version for active sports and one more robust DAFO design refined for return to a combat environment. This soldier is one of the first service members to return to combat with a DAFO as an Army infantry officer, and the device withstood the rigors of deployment and maintained structural integrity. Both service members credit their respective devices for reduction in pain levels and large improvements in quality-of-life.

The most important finding of this case study comparison was the extent to which current assistive devices, DAFO and prostheses, have the ability to provide more normative gait measures during fast walking, and allow for return to running even with variable injury pathologies and severities. The degree to which some gait variables mirror published norms was remarkable considering device engineering and construction for these devices must account for alignment, soft tissue reconstruction, proximal limb damage and comfort in the DAFO, and distribution of residual limb pressures relating to socket contours and interface, suspension system, prosthetic characteristics and alignment of the system in the transfemoral prostheses [113]. The service member with limb salvage contemplated secondary amputation of his involved limb early in his recovery due to the non-functionality and limited options for advanced prosthetic devices available at the time. Only after finding the DAFO did the soldier decide to keep his limb and continue with rehabilitation. Five years after his combat-related injuries, this soldier has ceased taking all pain medication, displayed gastrocnemius hypertrophy substantial enough to require modifications to his DAFO tibial cuff, and credits the DAFO for vast improvements in his life both personally and professionally.

Both service members were able to run at a pace that if maintained would result in a passing score on the aerobic component of the Army physical fitness test for their equivalent gender and age [90]. While wearing the DAFO, the service member was able to perform at a velocity of 3.47 m/s, equating to a 7:43 minute per mile pace. The service member amputee performed at a velocity of 2.72 m/s, equating to a running pace of 9:53 minutes per mile. Both service members have since passed an alternative Army physical fitness test consistent with the capabilities identified in their medical profiles, meeting the physical requirements to remain on active duty.

Despite the individual capabilities of each soldier and some biomechanical gait variables measuring similar to each other and near normal, there were some very distinguishable biomechanical differences in their walking and running gait. Each service member throughout this discussion was identified by their device for ease of explanation; “DAFO member” for the service member with limb salvage, and “prosthetic member” for the service member amputee. The design distinctions between the DAFO and walking prosthetic were evident by the ankle position at initial contact. The drop foot pathology of the DAFO member was successfully corrected at heel strike with the DAFO, based on the dorsiflexed position between the tibial strut and foot components.

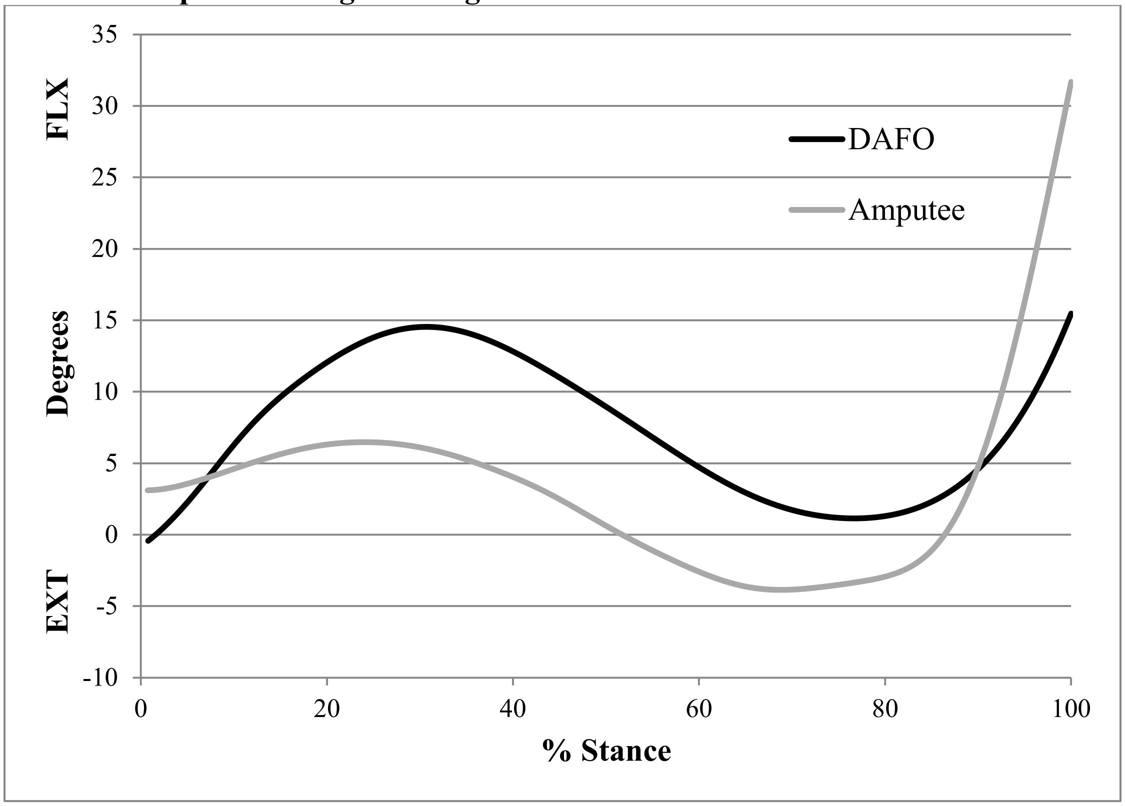
The involved limb remained in dorsiflexion throughout remaining gait phases due to the DAFO configuration, only allowing one-third the amount of walking plantarflexion/dorsiflexion excursion as compared to the walking prosthetic. Eversion excursion was also controlled with the DAFO design, providing support for an ankle with limited stability, possibly due to mechanoreceptor damage during nerve trauma to the lower limb [114, 115].

Both the DAFO and walking prosthetic provided maximum ankle plantarflexion moments prior to midstance that were consistent with normative data during walking [63], ensuring smooth and controlled movements from initial contact to foot-flat in both cases. Foot progression angles at midstance, however, varied greatly between the DAFO and walking prosthetic. The DAFO member exhibited 18.02° of internal rotation during midstance at the foot, while the prosthetic member completed midstance in 15.85° of external rotation. Normal foot progression moves into external rotation between loading response and terminal stance by 8-10° and returns to four degrees of internal rotation by pre-swing [63]. This execution is highly dependent upon stabilizing movements of the hips in the frontal plane, emphasizing the need for proximal hip strength to control distal segments during gait [116]. The varying levels of hip trauma and subsequent hip strength between these two service members may have accounted for the deviations in foot progression.

Knee positions at initial contact and toe off were similar between the DAFO and walking prosthetic; both devices assisted in initial contact at near-neutral knee flexion-extension positions, resembling those seen in healthy walking gait [66]. Knee flexion levels during the first 40% of stance in the prosthetic member barely exceeded five degrees, whereas the DAFO member (15°) approached more normative values (20°) [66]. Other researchers have noted this reduction in stance phase knee flexion during amputee gait, hypothesizing that by keeping the knee extended, the likelihood of knee buckling is reduced (Figure 5.7) [113]. Maximum knee varus velocity, seen in both individuals during the latter part of stance, was much greater in the service member with limb salvage. Knee varus velocity in the prosthetic member only reached 40% of that of the DAFO member, mainly due to the mechanically fixed nature of the walking prosthetic knee joint. Substantial differences between the service members were seen in external knee moments in all three planes. The knee extension moment during loading response was greater in the DAFO member, countering the muscular force of the quadriceps. He also produced greater knee flexion moments than normative values, resembling magnitudes approaching that of a runner's gait [66]. Insufficient muscle activation due to the proximal level of the prosthetic member's dissected quadriceps may explain the substantially lower knee moments in all planes, similar to conclusions of a musculoskeletal modeling study of amputees by Bae et al. (2007) [77].

Hip flexion values were similar in both service members and reflected normative values at initial contact [63], but differences during walking were most pronounced at toe-off, as the DAFO member never reached hip extension throughout the gait cycle likely due to subsequent lordosis following repair of severe injuries sustained in the pelvis. Transverse hip kinematic deviations from normal gait were also notable, as both service members contacted the ground in external rotation at the hip, contrary to published normative data by Novacheck [66] and Ounpuu [63]. These deviations reflect previously explained pathological distal movements in the kinetic chain.

Figure 5.7. Stance Phase Knee Flexion: Service Member with DAFO vs. Service Member Amputee during Walking



EXT: Extension, FLX: Flexion

Kinetic motions about the hip, however, were closer to normative values in the DAFO member when compared to the prosthetic member. Internal hip flexion moments (external hip extension moments) seen in the healthy walker during late stance only reach 1.0 Nm/kg [66]. The prosthetic member reached almost twice that level in late stance (-1.92 Nm/kg). This value coupled with a lower ankle power at toe-off is also noted research by Lewis et al. (2008) [98] and Czerniecki et al. (1996) [113], who reported that the reduction in push-off power at the ankle is partially compensated by an increase in mechanical work by the hip extensors. Stance phases hip joint power generation of the prosthetic member exceeded that of normative values to compensate for the power loss at the plantarflexors despite a mass of only 30% of a normal lower extremity [113].

Running gait differences between the two service members were even more apparent in variables at the hip and knee, as the service member amputee currently used a non-articulated prosthetic for running. This design is often used in the rehabilitation of transfemoral amputees to gain running confidence and improved conditioning prior to adding the challenge of a mechanical knee articulation component, and requires a circumduction technique for successful execution [117]. The prosthetic member had only recently returned to running at the time of this case study and had been issued an articulating knee component for her running prosthetic, but had not yet obtained success with its use. The running prosthetic, like the DAFO, maintained a dorsiflexed attitude throughout the gait cycle. Despite never reaching plantarflexion values at toe-off, dorsiflexion angles during swing for both service members were maintained close to neutral, resembling values of uninjured runners [62, 63].

Foot progression in the healthy runner allows for a greater degree of external rotation than while walking, with 10-12° progressive external rotation during midstance, and returning to neutral during propulsion [63]. Both service members presented with the involved foot or prosthetic in an internally rotated attitude. This progression in the prosthetic member, however, was likely due to the forced circumduction gait of the non-articulated running prosthetic. The DAFO member's large degree of internal rotation at midstance was consistent with his internally rotated hip at initial contact through midstance as well as an internally rotated knee during midstance. The deviations favoring internal rotation throughout the kinetic chain were likely related to the reconstruction of his crushed pelvis and hip dislocation. Despite the large rotational differences between the DAFO member and a healthy runner, he experienced no pain associated with running or balance issues.

Increased ankle power contributed to the DAFO member's return to running; ankle power generation increased to levels seen in the uninjured runners as described by Novacheck (1998) [66]. Changes in ankle push-off have been found inversely related to hip joint moments [98] achieved by a trade-off between ankle and hip muscle requirements during gait [118]. Substantial decreases in hip flexion moment were evident in the DAFO member's running gait in concert with the improved ankle power generation. The prosthetic member's ankle and hip power were lower than normative values during propulsion, as the circumduction gait required compensatory movements of the hip abductors and likely additional compensation from the intact limb [119]. A larger degree of variation at the sagittal knee motion was seen in the running condition between the two service members. The DAFO member contacted the ground with slight knee flexion consistent with other studies of healthy runners [47, 66], while the running prosthetic maintained full extension due to the non-articulated prosthetic design. Neither service member reached normative values of max knee flexion moments during loading response due to lesser degrees of knee flexion in both subjects while their respective limbs were loaded [66].

Limitations in this case study involved the inherent differences between the cases examined. These service members differed by gender, suffered dissimilar combat injuries, and ambulated with the assistance of unique devices. In spite of these differences, a comparison of gait between a service member amputee and a soldier who once considered amputation before he discovered a suitable assistive device was important and notable. Service members returning from current conflicts with combat or physical-training related injuries resulting in the need for an assistive device may benefit from this type of case study to learn more about their newly developed gait, including compensatory mechanisms they may be relying upon, and gait changes which may reduce pain, improve functionality, and increase confidence levels needed to return to vigorous activity. Military healthcare professionals may use this case study to continue learning and researching new techniques to assist our wounded warriors with improvements in function and quality-of-life.

This biomechanical comparison between two service members using assistive ambulation devices concluded several gait variables have returned to normal or near-normal published gait characteristics with the use of newer, high-tech devices. Improvements in spatial-temporal characteristics for both members were evident; symmetry between the prosthetic and intact limb was nearly identical, crediting the engineering capabilities of new prosthetics/orthotics. The service members exceeded published norms for step length while walking, and were close to normative fast-walking velocities. Producing velocities that qualify as passing scores for the Army physical fitness test, both service members have pushed the limits with their devices and have set a high standard for the physical capabilities of wounded warriors. Kinematic and kinetic measures at the ankle returned to within normal limits during walking, and despite compensatory mechanisms at the hip and knee, both service members were comfortable with their current ambulation level in their respective devices. The DAFO member was back to running in his brace and had completed modifications with the orthotist to accommodate substantial return of gastrocnemius muscle mass. The prosthetic member experienced hip pain when running in the non-articulated prosthetic, and plans to incorporate the new mechanical knee accessory as soon as feasible to relieve hip and groin pain caused by running with circumduction. Compensatory gait patterns may be necessary with this prosthetic, but can lead to low back pain, degenerative joint disease, and other maladies [112]. Future studies should continue to address design challenges that contribute to compensatory running movements, and note that gait re-training with these types of patients may be beneficial to maximize function and comfort in both prosthetics and ankle-foot orthotics.

Lastly, the improvements in gait made by these two particular service members were credited largely to their intrinsic motivation and positive attitudes. Despite the trauma these two experienced, their enthusiasm to improve and devotion to remaining on active duty drove their rehabilitation. They were an inspiration to the research team, students at the university, and service members alike. We thank them for their continued service to their country, and for motivating all of us to be better stewards every day in our own lives.

CHAPTER VI – COST ANALYSES

Return to Duty versus Medical Retirement – A 20-Year Perspective

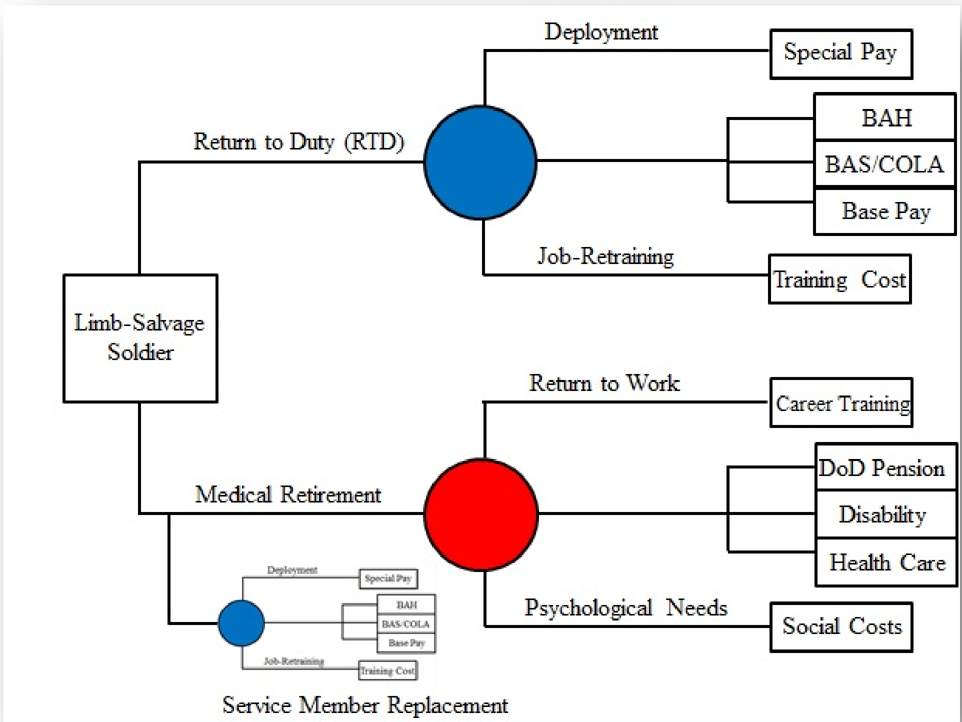
The long-term costs of returning wounded warriors to duty versus medical retirement have not been thoroughly studied, though researchers agree costs of treatment, rehabilitation, and lifetime maintenance of devices are important in guiding treatment decisions [12]. A comparative analysis of these costs may prove particularly important for the wounded warrior population, since the ability to return a limb-salvage patient to duty may result in both patient improvement in quality-of-life and substantial cost savings to the government. Some of these savings may include delayed medical retirement costs, quality-of-life-related health care and the avoided cost of training a new military member to fill the position of the veteran, despite the cost of advance technology AFOs being substantially higher than traditional models. Military leadership, physicians and legal counsel play a considerable role in determining fitness standards to return a soldier to duty and the level of disability compensation should a soldier be found unfit for duty due to permanent injury severity [120]. Decisions are made based on Department of Defense (DoD) guidelines for each branch of service, and the Veterans Affairs (VA) compensation is based on limitations to earning potential in the civilian job sector [120, 121].

The cost of caring for our nation's veterans is the fourth largest budgetary expenditure in the United States [122], and in addition to disability compensation and medical retirement pay, retired veterans are eligible for a range of additional benefits such as home loan guarantees, job reintegration training or additional compensation from Social Security if work reintegration outside the military is not an option [123]. A cost analysis was needed to make a thorough case for each service member's career. This information could be the impetus for a paradigm shift within the leadership regarding the retainment value of the limb-salvage population, if found to be more cost-effective and in the interest of the limb salvage soldier to return to duty. Additionally, the commitment to providing these soldiers with the most effective device may contribute to faster rehabilitation, improvement in motivation and quality-of-life, and an earlier decision to strive for continuation on active duty. This analysis provided the methodology and execution of a cost analysis for two Army soldiers of differing rank over a period of 20 years, one officer and one enlisted soldier, for medical retirement and continuance on active duty. Pay rates were established regardless of career field, and location for Basic Allowances for Housing and Subsistence (BAH/BAS) and Cost of Living Allowances (COLA) were based on assigned duty in Hawaii, which most adequately represents the population sample currently studied. All included figures were justified and cited from the most recent government cost estimates for Fiscal Year (FY) 2012.

METHODOLOGY

When conducting a cost analysis in a health care field, it is essential to specify the viewpoint for the analysis when determining inclusion of specific costs [44]; in this case the viewpoint is from the financial perspective of the federal government. The decision tree in Figure 6.1 pictorially represented the costs included in these scenarios. Costs to the federal government of the training and operation of a soldier on active duty included basic and special pay (hazardous duty pay and combat pay), housing, subsistence, and cost of living allowances, and the average cost of training a soldier regardless of career field. For comparative equivalence, these costs were estimated for a soldier without dependents, though it should be noted that such estimates would change dramatically if dependents were included. Because comparisons of medical retirement costs were specific only to the soldier and not their family, the same comparisons were made for a soldier's return to duty (RTD).

Figure 6.1. Cost Analysis Decision Tree



Costs for medical retirement of a wounded soldier included a retirement pension from the Department of Defense (DoD), Veterans Administration Disability pay, career reintegration training, and the cost of replacing that soldier on active duty. Compensation for social costs such as Post-Traumatic Stress Disorder (PTSD) were removed from the analysis based on the current estimates of documented PTSD rates (10-18%) among soldiers returning from Iraq and Afghanistan [124]. These estimates do not currently represent a majority of limb-salvage patients as reported, although it is important to note the associated VA benefits for a wounded warrior would have been substantially higher if two criteria are met: if the soldier experienced a significant stressor such as threat of military or terrorist activity, and if a documented condition of PTSD is made by a physician with a specialty in psychiatry [125]. Treatment for PTSD is complex and is currently the topic of much psychological research in the military, since currently over half of soldiers returning from war with PTSD symptoms do not seek treatment, or drop out of therapy early [126].

All costs included in this analysis were tallied over a twenty-year period following a specified date in time of a soldier's injury, which included scenarios for an enlisted (E-4) with three years of service, and an officer (O-3) with three years of service, determined by the age range of combat injuries in Iraq and Afghanistan (18-57 years), an average age of 26 years [3] , and the stipulation that service under three years of service may result in separation and severance pay versus medical retirement [127]. Overlapping costs of both scenarios were mentioned but not included in the total cost, as they did not affect the choice between the decisions to RTD or medical retirement. Once relevant costs were determined, each was measured and valued by calculating an 'equivalent annual cost' (reflected in **bold**), and the sources and methods of evaluation were clearly stated [44]. Financial figures in this analysis were extracted from valid government sources, including the Veteran Benefits and Health Administrations (VBA/VHA), the Congressional Budget Office (CBO), the Government Accountability Office (GAO), Department of Defense (DoD) financial documents, and from peer-reviewed literature evaluating the military population. Disability compensation from the VBA is the congressional intent in "recognition of effects of injury aggravated during active military service" or "provide compensation for an impairment in earnings capacity" [123]. Every effort was made to include the most recent costs that reflect present day value. Costs for inflation and cost of living increases were not included in this analysis, and a sensitivity analysis was not performed because none of the estimates were made by informed guesses; only by published values, avoiding methodological controversy [44].

Associated Costs of Medical Retirement

Caring for our wounded veterans includes significant costs associated with long-term healthcare, career reintegration counseling, disability compensation and medical retirement [123]. Long-term healthcare is a benefit to both soldiers who RTD and those medically retired through Tricare, a major component of the military healthcare system [128]. Career reintegration counseling is offered to veterans through the 'Vow to Hire Heroes Act of 2011'[129]. Eligible veterans may receive up to twelve months of full-time payment rates of $1,473 per month or **$17,676** for one year following military retirement for participation in the entire reintegration program. The program provides full-time monthly pay for participants enrolled in an education program through a community college or technical school approved by the VA, leading to an associate, non-college degree or certificate [129]. Soldiers are not eligible for this program if they are receiving benefits under the Montgomery GI-Bill.

Disability compensation payments are unique to each soldier, and may be assigned differently from the Department of Defense (DoD) and the Veterans Affairs (VA). The Physical Evaluation Board (PEB) deems fitness of soldier duty based on recommendations of military physicians, leadership, and legal counsel [120]. Conditions found unfit for duty are assigned a percentage, and multiple conditions are calculated by the largest percentage first (e.g. 60% disabled leaves a soldier 40% efficient). Any remaining conditions assigned a lower percentage are compared to the remaining efficiency (e.g. additional 30% disability rating is taken from original 40% efficiency, equating to an additional loss of 12% efficiency) and added to the original disability percentage (e.g. 60% + 12% = 72% disabled) [122]. The combined rating is then rounded to the nearest ten, giving this example a combined rating of 70% DoD retirement disability. Retired pay is calculated based on the disability percentage assigned. The pay base is the average of the highest 36 months of base pay received in the service member's career. A minimum of 30% disability is required to retire a military member prior to 20 years of active duty service [122]. A 30-70% disability rating equates to that percentage of retired pay base, and 80-100% disability rating equals 75% of retired pay base. Disability retirement pay cannot exceed 75% of a service member's retired pay base [122].

Congress established a standard for rating VA disabilities entitled the Veterans Affairs Schedule for Rating Disabilities (VASRD) [120]. Established ratings were based on residual effects of health impairments on the loss of earning capacity in the civilian sector [120, 121]. Published benefit rate tables for veterans alone or with a spouse and children were published based on percentage of disability. Table 6.1 reflects the monthly compensation rates for 10-100% disability for a single veteran with no dependents, regardless of rank.

Table 6.1. Veteran Monthly Compensation Rates: 10-100% Disabled

10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
$127	$251	$389	$560	$797	$1,009	$1,272	$1,478	$1,661	$2,769

Disability compensation from the VA is different than military medical retirement, though you are currently required to relinquish military retirement pay dollar-for-dollar in order to receive VA disability pay [122]. Other benefits of VA disability include nontaxable compensation, priority admittance for medical treatment, tax free allowances for dependents, and annual cost of living increases [122].

In 2004, legislation activated concurrent retirement and disability pay (CRDP) and combat-related special compensation (CRSC) programs that will phase in by January 2014. These programs replace the offset of retired pay reduced by the VA compensation under Section 641 of the National Defense Authorization Act [130]. Service members qualifying for CRDP have disability ratings of at least 50%, are retired for longevity and are receiving retirement pay [130]. Under the CRDP program, these qualified service members will be back paid for the reduced offset from the VA disability, and will be able to receive both DoD and VA retirement and disability benefits concurrently starting in 2014. To qualify for CRSC, the service member must be retired either for medical reasons or for longevity, and their disability must be combat-related [131]. The method to compute non-taxable CRSC pay is based on years of service and the amount they are currently receiving. The purpose is to account for time lost by early medical retirement, such that the service member receives benefits based on a full 20-year military career. Service members are unable to receive both CRDP and CRSC, but will be automatically paid the higher of the two entitlements through Defense Financial and Accounting Services (DFAS) [122]. Costs for this analysis are based on the instituted CRDP program.

The Defense Finance and Accounting Service developed a “Retirement Disability Pay Estimator” to assist service members facing the possibility of medical retirement [131]. This calculation accounts for the highest 36-month average monthly base pay, disability percentages from both the DoD and VA, CRDP off-sets, and years of service [131]. The medical retirement and VA disability cost estimates for this analysis were based on personal communication with a recently medically retired wounded soldier with a limb salvage injury [132]. This soldier received 70% in DoD medical retirement as well as a 70% VASRD score in June 2012. Although he had symptoms of PTSD, for personal reasons he did not pursue additional disability compensation for mental health concerns. The top 36-months of base pay were averaged and 70% of the average was calculated for the monthly DoD medical retirement pay [131]. Table 6.2 presents the cumulative monthly, annual, and 20-year tallies of military retirement pay for the ranks of O-3 and E-4 based on the 70% compensation [120]. The 70% monthly VA benefit entitlement of $1,272 (Table 6.1) was multiplied by 12 months to achieve an annual VA benefit package, and multiplied again by 20 to represent a cumulative VA benefit over 20 years [131]. Table 6.3 presents the cumulative monthly, annual, and 20-year tallies of VA benefit entitlements for the ranks of O-3 and E-4 based on the 70% compensation.

Table 6.2. Military Medical Retirement Pay for Ranks O-3 and E-4

Rank	Monthly Base Pay	Monthly DoD Retirement (70%)	Annual DoD Retirement Pay	20-year Cumulative Pay Tally
O-3	$5,390	$3,773	$45,277	**$905,540**
E-4	$2,157	$1,510	$18,120	**$362,400**

Associated Costs of Returning to Duty

Training a military soldier in any career field takes time, organization, and money. The Department of the Army publishes annual budget estimates to organize, equip, and train the all-volunteer force of 547,400 soldiers [133].

Table 6.3. VA Disability Entitlement for Ranks O-3 and E-4

Rank	Monthly VA Disability Entitlement (70%)	Annual VA Disability Entitlement	20-year Cumulative VA Disability Entitlement
O-3	$1,272	$15,264	**$305,280**
E-4	$1,272	$15,264	**$305,280**

Current average estimates for recruiting and training a new Army soldier to conduct prompt and sustained operations on land total **$63,455**, including basic military training and subsequent specialized training, focusing on individual skills and leadership education, and combined arms training strategy warfighter exercises [133]. Soldiers able to return to duty but require a change in military specialty due to their limitations will require additional military training. Based on the "Training and Recruiting" fiscal year 2012 budget estimates of $4.873 Billion, which includes accessing training, basic skill and advanced training, other training and education for a 547,400 soldier end strength, career retraining for an individual soldier estimates **$8,902** [133].

Factors that affect a service member's pay include an annual pay raise (1.7% for 2012), longevity raises every two years based on time in service (TIS), promotions, basic allowance for housing increases (BAH), basic allowance for subsistence increases (BAS), cost of living allowances based on location, and special pay such as hazardous duty pay, flight pay, or family separation pay based on occupation and deployments [128]. An Army commissioned officer at the rank of Captain (O-3) currently receives a base pay of $64,681. This rank is obtained in the Army after three years of commissioned service. The average time spent as a company grade officer (Lieutenant and Captain) in the Army is 10 ± 1 years upon promotion to the rank of Major (O-4). If promoted on-time, the soldier will become a Lieutenant Colonel (O-5) by 16 years, and obtain the rank of

Colonel (O-6) by 22 years [134]. The basic pay for this officer, excluding any percentage of pay raise proposed annually by the Appropriations Committee, will total **$1,657,968** by the time the officer reaches 23 years of commissioned service, or 20 years after their injury for this scenario. The pay scale for each rank is represented in Table 6.4.

Table 6.4. Commissioned Officer Pay Scales, Ranks O-3 to O-6 [135]

Rank	Time in Grade	Basic Pay Allowance	Total Pay in Rank	Cumulative Pay
Captain (O-3)	7 years	$64,681	$452,767	$452,767
Major (O-4)	6 years	$82,327	$493,962	$946,729
Lieutenant Colonel (O-5)	6 years	$98,389	$590,334	$1,537,063
Colonel (O-6)	1 year	$120,905	$120,905	**$1,657,968**

An enlisted soldier makes E-4 by two years' time-in-service (TIS), and the basic pay at this rank totals $25,882. Soldiers promoted in the primary zone will spend at least eight months' time-in-grade (TIG) and three years TIS before promotion to the rank of Sergeant (E-5). Dependent upon job performance, the average time in grade for a Sergeant (E-5) is four years, a Staff Sergeant (E-6): five years, Sergeant First Class (E-7): four years, Master Sergeant or First Sergeant (E-8): four years, and Sergeant Major (E-9): two years before reaching 20 years past the date of injury. Cumulative basic pay for an enlisted Army soldier by this point will total **$924, 513**. The pay scale for each enlisted rank is represented in Table 6.5.

Table 6.5. Enlisted Soldier Pay Scales, Ranks E-4 to E-9 [135]

Rank	Time in Grade	Basic Pay Allowance	Total Pay in Rank	Cumulative Pay
Specialist/ Corporal (E-4)	1 year	$25,882	$25,882	$25,882
Sergeant (E-5)	4 years	$32,135	$128,540	$154,422
Staff Sergeant (E-6)	5 years	$39,005	$195,025	$349,447
Sergeant First Class (E-7)	4 years	$49,010	$196,040	$545,487
Master Sergeant/ First Sergeant (E-8)	4 years	$57,966	$231,864	$777,351
Sergeant Major (E-9)	2 years	$73,581	$147,162	**$924,513**

Basic allowance for housing (BAH) is provided based on geographic location, pay grade, and dependency status [136]. These figures of equitable housing compensation are based on civilian housing market information collected by Runzheimer International, and cross-checked by military financial sources for accuracy and reliability [128]. An O-3 and E-4 assigned to Hawaii are allotted monthly BAH amounts of $2,274 and $1,461, respectively, without dependents. These values increase with rank, but change based on location. Tables 6.6 and 6.7 represent the BAH estimates over 20 years for an O-3 and E-4 living in Hawaii and progressing in rank, but does not reflect re-assignment to other bases.

Service members are also provided basic allowance for subsistence (BAS) and cost of living allowances (COLA) in addition to the housing allowance. The BAS continues the military tradition that rations are included in a service member's pay [128]. Enlisted service members receive $348.44 per month, the maximum benefit based on the requirement that a certain number of meals are eaten at the dining facility on-post, while officers receive $239.96. This monthly entitlement is readjusted annually based on the price of food measured by the United States Department of Agriculture (USDA) food cost index [128]. The COLA is a non-taxable allowance to off-set the cost of higher overseas prices of non-housing goods and services, to equate the ability to purchase the same level of goods as service members stationed in the Continental United States (CONUS). This allowance fluctuates intermittently based on the Living Price Survey (LPS) conducted every three years, and the Retail Price Schedule (RPS) survey conducted annually [136]. Tables 6.8 and 6.9 reflect the cumulative estimates of BAS and COLA for each service member scenario, officer and enlisted.

Table 6.6. Basic Allowance for Housing in Hawaii, Ranks O-3 to O-6 [136]

Rank	TIG	Monthly BAH	Annual BAH	BAH in Grade	Cumulative BAH
Captain (O-3)	7 years	$2,274	$27,288	$191,016	$191,016
Major (O-4)	6 years	$2,655	$31,860	$191,160	$382,176
Lieutenant Colonel (O-5)	6 years	$2,751	$33,012	$198,072	$580,248
Colonel (O-6)	1 year	$2,907	$23,884	$23,884	**$604,132**

Table 6.7. Basic Allowance for Housing in Hawaii, Ranks E-4 to E-9 [136]

Rank	TIG	Monthly BAH	Annual BAH	BAH in Grade	Cumulative BAH
Specialist/ Corporal (E-4)	1 year	$1,461	$17,532	$17,532	$17,532
Sergeant (E-5)	4 years	$1,701	$20,412	$81,648	$99,180
Staff Sergeant (E-6)	5 years	$1,866	$22,392	$111,960	$211,140
Sergeant First Class (E-7)	4 years	$1,977	$23,724	$94,896	$306,036
Master Sergeant/ First Sergeant (E-8)	4 years	$2,100	$25,200	$100,800	$406,836
Sergeant Major (E-9)	2 years	$2,235	$26,820	$53,640	**$460,476**

The bulk of deployment burden falls with the mid-grade soldiers; Army enlisted ranks E5-E9 and officer ranks O-3 and O-4 hold the highest cumulative deployment times [137]. Enlisted rank E-6 had an average of two deployments with 18 months cumulative enlisted time during their career. During deployments, soldiers are also entitled to combat pay, hazardous duty incentive pay (HDIP) for certain career fields, and family separation pay.

Table 6.8. Basic Allowance for Subsistence (BAS) and Cost of Living Allowance (COLA) in Hawaii, Ranks O-3 to O-6 [136]

Rank	TIG	TIG BAS	COLA per Month	Annual COLA Average	Cumulative BAS & COLA
Captain (O-3)	7 years	$20,160	$765	$9,184	$84,448
Major (O-4)	6 years	$17,280	$906	$10,872	$166,960
Lt Colonel (O-5)	6 years	$17,280	$997	$11,968	$256,048
Colonel (O-6)	1 year	$2,880	$1,069	$12,832	**$271,760**

Table 6.9. Basic Allowance for Subsistence (BAS) and Cost of Living Allowance (COLA) in Hawaii, Ranks E-4 to E-9 [136]

Rank	TIG	TIG BAS	COLA per Month	Annual COLA Average	Cumulative BAS & COLA
Specialist/ CPL (E-4)	1 year	$4,181	$515	$6,180	$10,361
Sergeant (E-5)	4 years	$16,724	$581	$6,972	$54,973
Staff Sergeant (E-6)	5 years	$20,905	$632	$7,584	$113,798
Sergeant 1st Class (E-7)	4 years	$16,724	$659	$7,908	$162,154
MSGT/ 1st SGT (E-8)	4 years	$16,724	$765	$9,180	$215,598
Sergeant Major (E-9)	2 years	$8,362	$848	$10,176	**$244,312**

Only hazardous duty pay and combat pay will be included in the analysis for soldiers with zero dependents and across all career fields. Combat pay includes either Imminent Danger Pay (IDP) or Hostile Fire Pay (HFP), and either one or the other are entitled to a soldier if ordered to an included area of this entitlement, such as deployments to Iraq and Afghanistan. The soldier can receive one or the other IDP/HFP, but not both. Each type of combat pay is added at a monthly rate of $225, or **$4,050** for an average of 18-month cumulative deployment time for each rank scenario. Hazardous duty incentive pay is provided to service members with orders including flight duty, parachute duty, flight deck duty, demolition, experimental stress duty, toxic fuels or propellants, toxic pesticides, dangerous viruses or bacteria lab duty, chemical munitions, maritime visit/board/search/seizure (VBSS) duty, and polar region flight operations duty [122]. Most if not all Army soldiers participate in one or more of these types of operations during deployment, and are entitled to a monthly additional pay average of $150. Because this additional entitlement is limited to the amount of time a soldier is on orders to complete a hazardous duty, this pay was also only included during the average deployment time of a soldier. This figure represents, on average, 18 months of their career or **$2,700** for each scenario [122].

Cost Analysis Tally and Conclusions

Tables 6.10 represents tallies for all costs included for each soldier ranks O-3 and E-4 over twenty years. Based on the totals, medical retirement of a military officer over 20 years costs the government a minimum of $1.23 million dollars more than returning that officer to duty, and retirement of an enlisted soldier costs $685K more than returning that soldier to duty. A majority of this cost was due to the replacement of an active duty service member to keep the Army end strength of 547,400 combat-ready service members at any given time.

Table 6.10. Cost Analysis Final Tally

	Return To Duty		Medical Retirement	
	Officer	Enlisted	Officer	Enlisted
Base Pay	**$1,657,968**	**$ 924,513**	NA	NA
BAH	**$ 604,132**	**$ 460,476**	NA	NA
BAS/COLA	**$ 271,760**	**$ 244,312**	NA	NA
HDIP/Combat Pay	**$ 6,750**	**$ 6,750**	NA	NA
MOS Reclassification	**$ 8,902**	**$ 8,902**	NA	NA
Reintegration Training	NA	NA	**$ 17,676**	**$ 17,676**
DoD Medical Retirement	NA	NA	**$ 905,540**	**$ 362,400**
VA Disability	NA	NA	**$ 305,280**	**$ 305,280**
Service Member Replacement	NA	NA	**$2,549,512**	**$1,644,953**
Total	**$2,549,512**	**$1,644,953**	**$3,778,008**	**$2,330,309**

Few studies have assessed the return to duty rate of a limb-salvage military population [138], though more studies from the civilian sector have identified factors involved in return to work rates [139]. Only 20% of a recently studied cohort of limb-salvaged soldiers (17 out of 83) were able to return to duty [138], while the remaining 66 were medically retired or separated (MRS). Some if not most of these 66 MRS soldiers may have had additional physical or psychological-related injuries contributing to the decision of the physical evaluation board (PEB), but soldiers found unfit for duty purely due to an orthopedic injury such as drop foot quite possibly may have been retained with a proper rehabilitation device. Ankle-foot orthotics of the highest technology currently cost between $10,000 and 15,000 [76], but if these devices allow soldiers to return to duty, the savings to the government would exceed $500K – $1.2M over twenty years. Supporting a motivated soldier's return to duty reaches far beyond saving the government money. Early treatment equates to increased self-efficacy, which transposes to a more valued member of society [140, 141]. Providing soldiers with the earliest and best physical and psychological support they need and deserve is not only paramount for their recovery; it is our duty to this dedicated profession of arms.

Cost-Utility Analysis: Traditional AFO versus Dynamic AFO

The ‘effectiveness’ or ‘utility’ of a medical device or treatment measured by currency alone does not paint an adequate or complete picture. Increases in quantity of life (reduced mortality) and the increase in quality-of-life (reduced morbidity) [44] of the patient using the device or treatment constitute the need for deeper analysis to justify an increased cost of medical treatments. Improvements in quality-of-life have been related to lower overall costs on a healthcare system [142]. Medical professionals often complete cost utility analyses (CUA) when quality-of-life is an important outcome, and the levels of improvement justify the increase in cost of a medical device [142, 143]. These types of analyses, in many ways like cost-effectiveness analyses, are only distinguished by how the results are expressed. An appraisal of quality of incremental health improvement attributable to the increased cost amount between a traditional ankle-foot orthotic (TAFO) and a dynamic ankle-foot orthotic (DAFO) was completed by means of a CUA in the present study.

METHODOLOGY

The CUA was accomplished by utilizing the short-form-36 (SF-36) quality-of-life (QoL) questionnaire, scored at the beginning of the protocol and then compared to the same service members’ final SF-36 QoL questionnaires obtained at the completion of the protocol. Each score contained eight categorized scales, five of which were used in the conversion to a Quality Well-Being (QWB) score established by Fryback et al. (1997) [144] (Equation 1), and multiplied by the number of years remaining in each service member’s life until they reached the average male life expectancy of 78.7 years [145], defined as their respective ‘mortality’. The result was a quality-adjusted-life-year

(QALY) measure, which accounted for the morbidity of a patient's health state, based on the assumption that a year with morbidity was not equal to a year without morbidity [142].

$$\text{Quality of Well Being (QWB)} = [0.59196 + (0.0012588 * PF) - (0.0011709 * MH) - (0.0014261 * BP) + (0.00000705 * (GH * RP)) + (0.00001140 * (PF * BP)) + (0.00001931 * (MH * BP))] \quad \text{(Equation 1)}$$

where PF = Physical Function, MH = Mental Health, BP = Bodily Pain, GH = General Health Perceptions, and RP = Role Function – Physical [144].

The QWB score originated from a specific Quality of Well-being Index, which along with the Short-Form-36 both measured health status, but through different methodologies [144]. Fryback et al. (1997) [144] developed and cross-validated a six-variable regression equation with a large data set to predict QWB scores from five of the eight health-related scales included in the SF-36 QoL questionnaire. The derived equation presented as Equation 1 above accounted for 57% of the variance in QWB scores provided by the original index [144]. Despite this percentage being lower than some statistical thresholds, authors reported that the reliabilities of the SF-36 (0.8-0.9 depending on the scale) and QWB (0.93) were strong enough that the residual variance unaccounted for in the model was likely due to limitations of measurement error by the two questionnaires and not the predictive relationships between the two [144].

After identifying average costs of each type of AFO, and converting a SF-36 score to a quality-adjusted-life-year, or (QALY), the last step in the analysis was to calculate the difference in cost between the intervention and alternative, and divide by the difference in QALYs, which yielded an incremental cost of the intervention per QALY, gained or lost (Equation 2).

$$\text{Cost Utility} = \frac{[(\text{Cost of average TAFO (\$)} - \text{Cost of average DAFO (\$)})}{(\text{TAFO QALY} - \text{DAFO QALY})]} \quad \text{(Equation 2)}$$

This yield was compared to benchmark thresholds set by the World Health Organization (WHO) [146] for what the U.S. or military healthcare considers “cost effective,” citing the U.S. range between $39,950 - $119,849, with anything less than $39,950 considered “very cost-effective”, within the range as “cost-effective”, and above $119,849 as “not cost-effective.” The lower end of the range was based on Gross National Product (GNP) per capita, with the upper range as three times the GNP per capita. An additional step in this appraisal was the inclusion of a sensitivity analysis, which assessed the effect of average AFO cost assumptions made in the equation [147]. Sensitivity analyses are important to any economic evaluation to account for extreme values of initial assumptions [142]. The average TAFO price reported by the service member group was $1,000; however the least expensive of these was $300. The average DAFO price reported by the orthotist was $12,000; however the most expensive brace was $15,000. The sensitivity analysis included the most conservative of the cost spectrum and recalculated the results to account for any bias in the calculation of averages.

RESULTS

The incremental cost of the intervention per QALY calculation is presented in Table 6.11. The CUA calculations based on the average costs of TAFO and DAFO in dollars determined all service members obtained a large enough increase in QALY yield to deem their respective DAFO "very cost-effective". The sensitivity calculations in Table 6.12 were based on conservative estimates of the TAFO (minimum reported cost $300) and DAFO (maximum reported cost $15,000) determined five of six service members' devices as very-cost effective, with the sixth device as cost-effective. Table 6.12 included calculated cost-effectiveness amounts based on the average brace cost used in Table 6.11, but with incremental decreases in the change in quality-of life expressed as a QALY, beginning below the minimum change in group, or 0.300. These calculations revealed one increment remaining in the very cost-effective threshold, the next four decreasing increments in the cost-effective threshold, and the lowest change in QALY (0.005), representing little-to-no change in quality-of-life in the not-cost-effective threshold.

Table 6.11. Cost-Utility Analysis Results

ID	TAFO QWB	DAFO QWB	Age	Mortality	TAFO QALY	DAFO QALY	CUA
1	0.5746	0.5839	29	49.7	28.5600	29.0198	$23,921
2	0.5794	0.5968	25	53.7	31.1142	32.0489	$11,769
3	0.5599	0.5786	42	36.7	20.5488	21.2335	$16,066
4	0.6087	0.6203	22	56.7	34.5140	35.1691	$16,790
5	0.5743	0.6053	33	45.7	26.2448	27.6637	$7,753
6	0.5859	0.5920	26	52.7	30.8791	31.1963	$34,682

Table 6.12. Sensitivity Analysis: Cost of AFO

Subject	CUA: TAFO $300 / DAFO $15,000
1	$31,968
2	$15,727
3	$21,470
4	$22,438
5	$10,360
6	$46,347

Table 6.13. Sensitivity Analysis: Decreasing Change in QALY

Δ QALY	CUA
0.300	$36,666.67
0.250	$44,000.00
0.200	$55,000.00
0.150	$73,333.33
0.100	$110,000.00
0.005	$2,200,000.00

DISCUSSION

Hippocrates said, "war is the only proper school for surgeons" [112]. Combat veteran amputees recuperating in military hospitals receive a great deal of press and public attention, denoting the cost of war on America's service members, and as a result the Department of Veteran's Affairs (VA) ensures prosthetics will be provided to these wounded warriors during their recovery [148]. The advancement of techniques used by military physicians such as primary vascular repair, management of extremity injuries and soft-tissue infections have resulted in decreased rates of primary amputations and increased the number of limb-salvage patients [7]. Just as developments in amputation surgery and prosthetic technologies have occurred as a result of armed conflicts, so too is the push for new techniques in limb reconstruction surgeries and thus, the need for improved ankle-foot orthotics (AFOs).

The evolution of the prosthetic began in World War II, when the VA provided limbs to amputees fabricated by the lowest contract bidder without standards of quality and accountability. The American public and wounded veterans demanded quality devices for the nation's war heroes, and voiced their outrage to Congress. Despite budget resistance, decades of high-ranking military, Congressional, Senate, and public support has resulted in a central prosthetic office that mandates prescription fill of any prosthetic regardless of lack of funding. [148] Veteran's Affairs Prosthetic and Sensory Aids service budgeted $1.4 billion to service devices for 1.9 million veterans in 2008 and $1.6 billion in 2009, with 5% used in the fabrication and repair of prosthetics [111, 149]. Cost analyses from 2010 appraised one current prosthetic system above $45,000, and 20-year and lifetime costs for unilateral lower limb amputees equates $855,000 and $1.45 million, respectively. The Government Accounting Office (GAO) recently released a report to congressional requesters on the increases in VA spending on prosthetic devices, evidence that the cost of assistive devices for combat veterans was at the forefront of congressional discussions [149].

The amount of attention paid to the cost of prosthetic devices places increased importance on a CUA of ankle-foot orthoses. The path towards including a DAFO as the standard of care for wounded service members will likely be easier than that of the prosthetic due to a differing mindset towards the service member of today compared to Vietnam. Cost-utility analyses have already been completed on limb-salvage versus amputation, finding limb-salvage as the cost-saving strategy based on surgery expense and QALY yield [142]. This CUA revealed the experimental DAFO was a very cost-effective device based on the change in quality-of-life of the six injured service members who participated in this study. The sensitivity analysis of conservative brace cost supported the CUA finding in five out of six service members, with the sixth remaining in the cost-effective threshold. This analysis indicated that even the most expensive DAFO used in the present study was very cost-effective 83% of the time based on the individual service members' improvements in quality-of-life. Issue of the DAFO with decreased incremental improvements in quality-of-life were very-cost effective 17% of the time, and cost-effective 83% of the time, based on the sensitivity analysis of incremental decreases in QALY yield. An $11,000 increase in brace cost was non-cost-effective for little-to-no improvement in quality-of-life.

The limitations of this CUA included a small sample of service member participation, and varying timeframes of protocol completion. Three of the service members were able to complete the six-month protocol; hence fewer SF-36 QoL scores and the subsequent CUA were based on six months of DAFO use. One service member was medically retired one month prior to protocol completion, and thus his fifth month SF-36 QoL score was used for comparison. The remaining soldier was only able to complete two months of the protocol prior to medical retirement; therefore his second month SF-36 QoL score was used for comparison. Overall, this CUA provides military leadership and health care professionals with fiscal justification to include the DAFO in the standard of care for wounded service members.

PART II

REVIEW OF LITERATURE

The most effective device and treatment for a limb-salvage patient depends on many factors. Lessons from the Vietnam War and the current Global War on Terror (GWOT), if nothing else, brought advancements in surgical treatments of military casualties, forward-thinking physical and mental rehabilitation, and the push for the most functional devices to provide wounded warriors with the best chance at a pre-injury quality-of-life. The definition of functional ambulation in ankle-foot orthoses is evolving with the active military population, and science and technology are responding by pushing boundaries in performance of these new devices which could translate to better rehabilitation for the civilian population in the long-term. Any new technology necessitates clinical, financial and quality-of-life outcomes to facilitate a paradigm shift to a new definition of 'standard care'. Physiological and biomechanical measures are valuable for defining clinical outcomes for refinement of device quality and justification of cost. The methods used to define these outcomes in laboratory settings are still evolving and little consensus exists on the emphasis of specific gait variables to aid in effective ambulation.

The assessment of cost-utility between the different outcomes of a soldier's career based on their standard of care is a key component to military and government leaders in the allocation of future funding. Funding in the fiscal year 2012 (FY12) Presidential Budget highlighted ongoing support of wounded warrior transition units and areas to continuously improve care provided to injured service members [133]. Evaluations of the cost associated with medical retirement versus continuing on active duty with the proper equipment deserves proper attention and focus to promote forward-thinking and improved care to service members while remembering the judicious fiscal decisions and constraints of the current administration. Therefore, the intent of this literature review is to provide a brief overview of the traumas which have resulted in a military drop foot population including lower limb pathology, the quality-of-life outcomes of current interventions, and the cost-utility analysis required to consider a change to standard care. The subsequent sections focus on the lower limb parameters outlined in normal walking and running gait, followed by the evolution of ankle-foot orthotic design and biomechanical comparisons of civilian drop foot populations. Finally, this literature sheds some light on the military importance of the present study, and the gaps in the literature needed to truly optimize the functional gait necessary to return a soldier to combat duty.

Military Combat Incidence of Injury

Current U.S. armed conflicts have experienced relatively equal percentages of upper and lower extremity injuries, mostly due to improvised explosive devices (IED), gunshot wounds, rocket-propelled grenade (RPG) with associated levels of bombs leading to secondary motor vehicle accidents [6]. Compared to previous conflicts, wound patterns are similar, but survival rates in present day are higher. Combat casualties are updated daily through the Department of Defense casualty status report [1] and totals have reached 50,532 as of 7 March 2013. Lessons from Vietnam in improving patient care led to the development of the Joint Theater Trauma Registry (JTTR) and Joint Theater Trauma System (JTTS) to organize and coordinate trauma care for military casualties sprawling three continents [6, 150]. Eastridge et al. (2009) [150] summarized the impact the JTTR has had on patient care by comparing patient outcomes 23,250 injured military members between 2003-2008. The registry was used to develop a pre-deployment medical training course, which 87% of trainees cited improved preparation for combat medical missions. More than 152 peer-reviewed manuscripts educated medical providers with data supplied in the JTTR, and paired with the military advancements of improved personal protective equipment and pre-hospital care, the JTTR has greatly contributed to the survival of soldiers after battlefield injury.

Efforts have been made on smaller scales to calculate casualty care incidents rates during different surge operations of the war. Belmont et al. (2009) [2] conducted a prospective, longitudinal analysis of 4,122 Army Brigade Combat Team soldiers deployed during "The Surge" portion of Operation Iraqi Freedom (OIF), using a casualty database and registry to discover combat casualty care statistics, distribution of wounds, and mechanism of injury during this portion of OIF, focused on soldiers killed in action (KIA) or wounded in action (WIA). Incidence rate was defined as the number of battle injuries within the study population divided by the number of combat-years at risk. Subjects were grouped according to military rank, and their wounds were categorized by body region and type of injury and then compared against previous U.S. conflicts. Statistical Analysis Software (SAS) with a significance level set at 0.05 was used to test the χ^2 test statistic to determine an association between nominally scaled values of rank group, combat care statistics, distribution of wounds, mechanisms of injury, body region, and combat casualty classification. Fisher's exact test was used in the absence of an adequate sample size. Tallied results of combat casualties showed the largest amount of combat wounds sustained were to the extremities (49.4%). There was a significant increase in head and neck wounds when compared to previous conflicts, but a significant decrease in wounds to the thorax and extremities. The greatest percentage of mechanisms of injury were due to explosions (87.4%), which were significantly greater than injury due to a gunshot wound (GSW) (9.0%) reflecting the lethality of modern weaponry. This study provided an epidemiological basis for future comparisons of combat injuries.

Established registries have provided the conveyance of medical information necessary in a broad military environment which has been useful for both prospective injury predictions and retrospective injury comparisons. New advances in treatment have been introduced because of the statistics from these cumulative reports such as surgical improvements of early limb-salvage in the military population.

Limb-salvage versus Amputation

Lack of personal protective gear on the extremities compared to the abdomen has been a contributing factor to the aforementioned casualty rates, and has made the lower limbs most vulnerable to devastating blast injuries. Spear (2009) [7] summarized evidence on the outcomes of lower extremity complex injuries experienced during OIF and OEF, citing improvements were due to the motivated soldiers and advance medical techniques including microvascular surgery, fracture repair, frequent debridement, and effective antibiotics. The current advancements of primary vascular repair in the battlefield and management of extremity injuries and soft-tissue infection have resulted in decreased rates of primary amputations and increased the number of limb-salvage patients. Military medicine used guidance gained from civilian studies for outcomes and proper care of the limb-salvage population.

Proper care including initiation of rehabilitation and complete physical therapy for limb salvage patients was reported by Johnny Owens (2010), the physical therapist at Center for the Intrepid in Brooks Army Medical Center (BAMC), San Antonio [8]. This editorial described the injuries of most recent conflicts as a result of blast injuries, classifying the injuries as primary, secondary, tertiary, or quaternary. Primary injuries stem from the increase in air pressure surrounding the explosion, resulting in a potential rupture of hollow organs, whereas secondary injuries are a result of flying debris. Tertiary injuries are sustained from being thrown by the blast, and lastly quaternary injuries are thermal. Service members are stabilized at BAMC and begin outpatient rehabilitation within 24 hours to establish early goals, increase ROM, and ambulation. Knowing long-term physical outcomes are similar for both amputee and limb salvage groups, the BAMC group continually refocuses service members' short-term frustrations back to their long-term established goals. Outcomes from previous literature have aided in prioritizing rehabilitation; the impairments of calf atrophy greater than three centimeters, hip flexion ROM less than 120°, ankle plantarflexion ROM less than 50°, and ankle dorsiflexion ROM less than 20° are addressed first. Aggressive ROM rehabilitation to the toes, ankles, and knees are possible in the patients with external fixation. Lower extremity strengthening and core training are also emphasized in their rehabilitation program, which was established through recommendations from the literature and successful rehabilitation of amputees.

The major study cited in the above rehabilitation program was the "Lower Extremity Assessment Project (LEAP)," which consisted of two and seven-year follow-ups of 601 patients who underwent lower extremity amputation or reconstruction following trauma. MacKenzie et al. (2006) [15] in part conducted a multi-center study of functional outcomes of severe lower extremity trauma reconstruction and amputee patients in the U.S. population. Conceptual framework for assessing outcomes was established using characteristics of the individuals (e.g. physical traits, state of self-efficacy, spirituality) combined with physical impairment and secondary medical conditions (post-traumatic stress disorder (PTSD) and pain), and characteristics of physical, educational/economic, and social/work environments to determine functional status and quality-of-life (QoL). The LEAP study group found the presence of anxiety and depression to be significant predictors of long-term poorer outcomes in the civilian population. Outcomes of the study strongly suggested the need for early post-acute recovery interventions that directly address each patient's psychological and self-management needs. Unfortunately some of the outcomes may not be applicable to a military population as they were excluded from participation in the LEAP study. Since the military population already contains established support networks, higher pre-injury physical conditions, and guaranteed health care, some of the study recommendations may be of minimal use. The concern that social support and self-efficacy may not remain high after discharge from active duty is noted for the military population. Results from this study may be used to identify interventions that could influence or improve functional outcomes and quality-of-life.

Quality-of-life in post-traumatic amputees compared to limb salvage patients was evaluated in a meta-analysis by Akula et al. (2011). Two authors searched three databases using the terms amputations, limb salvage, quality-of-life, SF-36, and Sickness Impact Profile (SIP) to thematically collate which treatment provided a better quality-of-life based on physical and psychological measures. Inclusion criteria for articles included those written in English, reported on post-traumatic unilateral amputation, a minimum of 20 cases in each study, minimum follow-up of two years, studies limited to a 27-year window (1990-2007), quality-of-life assessed by one or both of the aforementioned questionnaires, and a quality score greater than five based on questions related to inclusion criteria, subject withdrawal rate, follow-up, outcomes and pertinent characteristics. Eleven of 214 studies met the inclusion criteria and quality score, and presented mean SF-36 physical component scores of 39.76 for the amputee group and 38.5 for the limb reconstruction group. Mental component scores revealed higher tallies for the amputees (52.05) compared to limb reconstruction (50.76), with higher scores indicating better health. The SIP scores for physical and psychological were 16.2 and 15.6 for amputees, and 13.3 and 11.5 for limb reconstruction, respectively. Unpaired tests were used to compare the two treatments using scores from both questionnaires, and researchers discovered that physical scores were not significantly different between the two treatments; however psychological outcomes were more significantly more favorable for limb reconstruction.

Although quality-of-life is an important indicator in the decision between limb-salvage and amputation in the field, this judgment often must be made quickly, and currently no standardized protocol exists. The unique nature of these injuries and reasons demanding early triage include contaminating bacteria from another country, delayed definitive care and multiple concomitant injuries [151]. Hansen ST (1987) [16] reported on the results of limb-salvage decisions and severe open tibial fractures. Three of the eleven patients with Type-IIIA had a non-union, but no secondary amputations. A total of 42 patients presented with Type-IIIB tibial fractures, of which seven elected to complete a secondary amputation, but only two of 24 within this group who had been treated with early management of their soft tissues injuries had the secondary procedure. The Type-IIIC injury group, however, had secondary amputations in seven out of nine patients, and the remaining two patients had a poor result of limb-salvage. Hansen noted that there are times when salvage was not the best course of action, and a Type-IIIC injury leads this category. These surgeries may have been detrimental to the patient in terms of quality-of-life due to a longer recovery process filled with multiple surgeries and treatment programs. Statistics for this injury were similar to back injuries, and after two years of rehabilitation, almost 100% of patients never return to work. However, this study looked at the process 25 years ago, and may not reflect limb-salvage outcomes of today.

A more recent assessment by Dagum et al. (1999) [11] retrospectively examined the predictive value of amputation index scores as well as overall health status by administering the Short Form-36 (SF-36) questionnaire. The charts of 55 lower extremity injuries were referenced over a 12-year period, and provided outcomes back to the date of Hansen's report. These injuries were classified as an attempted-salvage group, and further subdivided between successful limb-salvage and secondary amputation. Relative indications of amputation determination consisted of severe ipsilateral foot injury, soft-tissue or bone loss, warm ischemia time greater than six hours, and concurrent injuries. The patients associated to each injury were contacted and asked to complete the SF-36 questionnaire, and results indicated a non-significant difference in the long-term physical outcomes between limb-salvage and the primary and secondary amputation patients. The amputees' worst scores were physical function outcomes but not mental or pain scores, which indicated the disability is a physical but not mental condition. Authors concluded that although amputation index scores did not make the decision easier between secondary amputation and limb-salvage, and early attempt at salvage should have been made if the patient did not meet the standard indications for early amputation. Successful salvage presented better functional outcomes than amputation.

A literature review regarding these decisions in severe lower-limb injuries was conducted by Busse et al. (2007) [9]. Reviewers extracted articles that met prospective or retrospective observational study criteria, where leg-threatening injuries were managed with limb-salvage or primary amputation and a report of outcome data useful for comparison. Nine studies met the inclusion criteria, and comparisons were made regarding length of hospital stay, total rehabilitation time, cost data, clinical outcomes, function and QoL. Limb-salvage and amputation had similar lengths of hospital stay, although limb-salvage required more surgeries with higher complication rates. Functional outcomes were similar after a seven-year follow-up with similar pain scales, and the rates of return-to-work were similar at approximately fifty percent.

A prospective view of functional outcomes between trauma patients with limb reconstruction versus amputation was provided by Bosse et al. (2002) [12]. Distal femur trauma patients (n=545), age 16-69 were included in the study. Some (n=149) underwent amputation during the initial hospitalization, with 25 more after the initial hospital release, twelve more amputees by three months, and 13 additional by six months. All patients were evaluated prior to initial hospital release, and at three, six, 12, and 24 months following their injury for presence or absence of surgery complications, extent of impairment, and patient assessment of functional outcomes. Although some study attrition was observed, the difference between reconstruction and amputation study attrition was non-significant. The Sickness Impact Profile (SIP) was used to assess 136 statements regarding limitations within 12 categories of function. Data were also collected on characteristics hypothesized to impact treatment assignment our outcome, including age, sex, race/ethnicity, education, income level, insurance status prior to the injury, personality characteristics as determined by the Neuroticism, Extroversion, and Openness (NEO) Personality Inventory, social support, directed guidance, emotional support, and self-efficacy. Measures of self-rated health were also examined, and longitudinal multivariate regression assessed associations between treatment and outcomes over a two-year period. The multiplicative model was chosen because although scores on the SIP continually improved, those improvements declined over time. Adjusted for severity of injury and various other confounders, patients who underwent amputations had functional outcomes similar to patients who underwent reconstruction. However, reconstruction was associated with a higher risk of complication, additional surgeries, and re-hospitalization.

The most recent editorial on the continued debate between limb-salvage and amputation was conducted by Shawen et al. (2010) [152], as severe trauma injuries to the lower extremity continued to rise above 50% in the Global War on Terror. Technology advancements now exist for surgical limb-salvage techniques as well as prosthetic designs. Few orthopedic surgeons were trained well enough during residency to deal with such cases in the field, but those experienced surgeons noted patients will opt for limb-salvage if possible. Yet in retrospect 75% of those cases would have elected for amputation due to the unsuccessful salvage. Centralized care facilities were established to provide soldiers the best support networks when working through rehabilitation for either treatment. Decisions to salvage versus amputate the limb revolved around the severity of the wound, and often vascular injuries precluded limb-salvage. Defects of the tibia and soft-tissue were well documented and required extensive surgeries to restore function at an optimal level. Other injuries such as wounds to the calcaneus and open foot traumas were less documented, and no current algorithm fit every decision for these patients. Following a limb-salvage of this type, emphasis was placed on the physiotherapy involved in moving the digits to reduce tendon scarring. Another important factor for treatment decision was the condition of the contralateral limb. Every effort was made to limb-salvage if the contralateral limb had been amputated. Regardless of the treatment options, the resultant level of disability was strongly influenced by the patient's coping mechanisms. Patients with borderline injuries were evaluated for pain tolerance and the need for free tissue transfer when making treatment decisions, and many were better served with amputation. Even as surgeons expand knowledge and techniques for limb-salvage, currently there is no way to determine whether the patient will thrive after the treatment decision is executed. A support network of mentors, peers, and educating patients will result in a more productive rehabilitation and life.

The fine line between limb salvage and amputation makes comparisons between the two appropriate for study. A case study amputee was added late in the present study to provide comparison against long-term DAFO use, and thus necessitated research on prosthetics. Harvey et al. (2012) provided an editorial on prosthetic advances of both the upper and lower extremity. Many of these advancements have been the result of injuries sustained in war, and noted the advanced surgical techniques in the field necessitate a continuation of prosthetic and orthotic technologies. The author provided a historical evolution of the prosthetic, and predictions that powered knee and ankle combinations are the next logical step in prosthetic development. Currently few above the knee amputees utilize the stance phase knee flexion powered prosthetic, instead using a locked knee with circumduction pattern because of the feeling that 'the knee is giving way". These compensations likely contribute to the high prevalence of low back pain in these patients.

Seroussi et al. (1996) also highlighted mechanical adaptations in above the knee amputations (AKA) by comparing AKA versus normal gait [119]. Eight AKA patients (ages 30-44) ranged in prosthetic use of 5-23 years. Subjects walked at a comfortable pace for four trials, and gait was measured using a Kistler force plate interfaced to a microcomputer. Reflective surface markers were applied on the lower extremities and recorded by video. A signal processing protocol was used to organize data for statistical comparison through factorial ANOVA. Researchers reported the prosthetic ankle only accomplishes 20% of the work seen in normal ambulators. A 270% increase in hip extensor work seen on the AKA patients' sound limb may be a potential compensation mechanism for this work reduction of the prosthetic. The sound limb also exhibited a 33% over normative levels for ankle push-off. This increase was coupled with a relative increase in concentric hip flexor work on the prosthetic limb, and increase eccentric work by the hip in mid to late stance. Authors concluded these increases in hip movement may necessitate a hip strengthening program for AKA patients in order to avoid secondary injuries due to compensation.

A dynamic analysis of AKA gait was also accomplished by Bae et al. (2007) [77]. Similar to Seroussi, the researchers in this study assessed eight AKA patients against a healthy control population for time-distance parameters, kinematics, EMG, and dynamic analysis using musculoskeletal modeling. A 7-camera Vicon motion capture system sampling at 60 Hz and two video cameras captured fifteen reflective markers assembled on the lower extremities. Clinical experimentation of stair ascent was analyzed as well as level walking. Common gait deviations of AKA patients were also found in this study, including significantly different pelvic obliquity, hip abduction and hip adduction compared to controls, due to musculoskeletal weakness on the involved limb and subsequent pelvic drop during stance. Results of the musculoskeletal modeling revealed lower muscle activity even in the sound limb for quadriceps and hamstrings compared to controls while climbing stairs. Instead, tibialis anterior and gastrocnemius muscle activity in the sound limb of AKA patients was significantly greater than controls. Conversely, the muscle activity of hamstrings and quadriceps of the AFA patients' sound limb was greater than controls during level walking. The results of the combined methods showed that weak hip abductors results in excessive adduction in the frontal plane, likely due to the dissected portions of the quadriceps and hamstrings on the prosthetic limb side. These muscles on the sound limb likely compensate for the insufficient muscle force on the amputated leg.

Czerniecki (1996) [113] remarked on the relationship between ankle and hip in a learning module chapter on rehabilitation of limb deficiency for practitioners in the field of Physical Medicine and Rehabilitation. This interrelationship has been explored in both normal and deficient gait, and this editorial provided adaptive strategies of amputees.

The marked ankle power generation loss seen in AKA patients is partially compensated for with hip extensor power. Despite the prosthetic limb having only 30% of the mass of a normal lower extremity, it must generate a normal level of hip flexor power. This compensation may result in secondary back pain mentioned in the several studies above. Despite the increased technology and emphasis placed on prosthetics, continual design is needed to reduce the pain associated with required compensatory movements in the above the knee amputee.

The studies outlined above further highlight the difficult decision between limb salvage and amputation, both from the perspective of the wounded service member and the military surgeon. Although QoL is an important factor in the decision, severity of injury and condition of the contralateral limb are first and foremost important. The increased incidence of limb salvage patients are living proof of the advancements of medical technology, and the evolution of AFOs to accompany these limbs must now follow suit.

Lower Limb Anatomy and Pathology

Damage to the tibial and common fibular (peroneal) nerves have been most associated with the prescription of an ankle-foot orthotic. Although the two nerves are enveloped in a common epineurium, the fascicles remain separate throughout the length of the sciatic nerve. Sciatic nerve division occurs at the level of the popliteal fossa and the common peroneal nerve passes inferolaterally under the biceps tendon, continuing posterior to the popliteal muscle tendon until curving around the neck of the fibula. The nerve is connected to the fibular periosteum under the intermuscular septum and peroneus longus muscle for ten centimeters. Exposure of the common fibular (peroneal) nerve while over the body prominence of the fibula, covered only by skin and superficial fascia, makes this nerve very susceptible to injury [153]. A fibro-osseous "peroneal tunnel" [154] is formed proximally by the heads of the peroneus longus and the attachments to the neck of the fibula and tibial condyle. The common fibular (peroneal) nerve divides into the superficial and deep fibular nerves deep to the fibularis (peroneus) longus muscle. Innervation of the fibularis longus, brevis and skin on the distal third of the anterior lower limb comes from the superficial branch, while innervation to the anterior muscles of the leg is provided by the deep branch [155]. The deep fibular nerve accompanies the anterior tibial artery into the compartment between the tibialis anterior muscle and the extensor digitorum longus muscle. Continuing outside the compartment across the ankle joint, the deep branch innervates intrinsics and a small area of skin on the foot. Damage to this nerve is the indicator of drop foot. Possible mechanism of injury for impairment of the peroneal nerve were reported by Meals (1977) [156]. These include sudden forcible supination may result in traction on the peroneal nerve as it rounds the fibular neck, hematomas pressed the nerve to the point of ischemia and paralysis, or the muscles lateral and posterior to the fibula may tense during forced inversion. The first hypothesis was supported by over half of the cases reviewed during this study, as witnessed by the patient's reported symptoms. This nerve injury resulted from an inversion sprain was more common than the literature suggested, as determined by one study of 22 of 133 ankle sprains found to have mild peroneal weakness. This study helped provide another etiology of peroneal neuropathy.

Understanding the ligamentous structure is helpful to relate secondary injuries to peroneal neuropathy. Damage to the peroneal nerve not only affects the related musculature, but can disrupt ligamentous control and increase the susceptibility to ankle instability. Chronic ankle instability has been attributed to loss of proprioception, possibly related to specialized mechanoreceptors stationed around the ankle which may be affected by peroneal nerve damage. Michelson and Hutchins (1995) [115] realized that knowledge on mechanoreceptors was lacking, and used cadaver tissue to categorize mechanoreceptors into four types. Five pairs of fresh frozen cadavers were dissected distally from the knee, half of which were exposed at the ankle ligaments. Specimens were cut into 5μm, tallied, and placed into placed into the categories based on morphology, parent nerve, and physiology, which allowed researchers to examine the frequency of mechanoreceptors in relation to bony insertions. Averages were determined by counting the number of receptors for each cross-section of ligament. Type-I receptors, characterized as thin encapsulated globular corpuscles, were see at low frequency in all ligaments. These receptors are slow-firing even at rest, and have been associated with postural sense with the central nervous system. Type-II receptors thought to have the most proprioceptive function and convey the beginnings of joint motion were the most common in all ligaments, and type-III receptors (thin encapsulated fusiform corpuscles) thought to provide sensation at the extremes of movement were also in high frequency. Researchers discovered the same types of mechanoreceptors previously found in other species and joints were also present in the ankle, which implied the initiation of movement at the ankle involved stress transmission through the ligaments. The authors strongly suggested the sensory output from ligaments controlled muscle stiffness and coordination around the joint to increase stability, and therefore damage of the nerves to this area may also have damaged the mechanoreceptors which would may help explain the ankle instability seen in drop foot patients.

The role of mechanoreceptors in proprioception was supported by Hertel (2000) in an editorial on functional instability following lateral ankle sprain [114]. He provides an overview of functional instability and how it is manifested clinically, as well as theoretical implications of the condition. Most pertinent to the present study is the theory that when lateral ankle sprains occur and ligamentous tissue is damaged, so too is nervous and musculotendinous tissue surrounding the ankle. Drop foot patients have nerve damage; therefore the likelihood that ligamentous and musculotendinous tissue is damage is also logical. This damage may manifest as impaired balance, a reduced sense of joint position, and slower firing of peroneal muscles, as well as slower nerve conduction velocities, strength deficits and reduced dorsiflexion. All of these impairments are seen with drop foot patients, and the patients in the present study complained they couldn't run partially due to the instability at the ankle. Mechanoreceptors, according to Hertel, are most active near the ends of ROM to sense joint movement. Little is known about these receptors following ligamentous injury. Hertel followed this with a review article in 2002 on chronic ankle instability (CAI) [94]. He reported that muscles of the anterior compartment may contribute to dynamic stability of the ankle, by slowing the plantarflexion component of supination, thus preventing injury to the ligaments. He also found the deep and superficial peroneal nerves along with the sural and saphenous nerves innervate the lateral ligaments, talocrural and subtalar joints. These joints are shown to be extensively innervated by mechanoreceptors that contribute to proprioception, and the muscles spindles in the peroneal group are most important to proprioception. Therefore it is reasonable to assume that damage to the peroneal nerve is also connected to service members' feeling of ankle instability and decreased balance.

Despite the mechanism of injury, some of the consequences of peroneal nerve damage are the same: the inability to dorsiflex and evert the foot. Understanding the anatomy and pathology associated with this condition assists physicians, physical therapists, and orthotists in the diagnosis, prognosis, and rehabilitation pathway for a drop foot patient.

Quality-of-life

The physical, mental, and social well-being facets of a defined quality-of-life (QoL) have been increasingly researched in the military since the inception of the Global War on Terror (GWOT). Understanding the patient's point of view of personal medical outcomes is one of the most important advances to health care. The increased emphasis on evaluating post OIF/OEF war veterans highlighted by the Department of Defense (DoD) and Veteran's Affairs (VA) identified and treated post-traumatic stress-related disorders that come with combat and sustained injuries. Identifying factors that influence maximal function of a drop foot patient was important for treatment options and prognosis [157]. Comparisons between limb salvage and amputation helped to formulate a complete picture of the aspects of QoL most important to those affected by lower extremity trauma.

One struggle in obtaining this point of view has been designing a survey tool that is psychometrically sound, and comprehensive yet brief [158] as to not create an additional burden on the patient. The Rand Corporation led a major Medical Outcomes

Study (MOS) to produce a valid and reliable general measure of QoL through health-related concepts [58, 159]. This questionnaire, entitled the MOS Short-Form (SF)-36 Quality-of-Life Questionnaire, proved manageable survey that was acceptable to patients with high internal validity and solid test-retest capabilities [58], which is currently used world-wide to assess physical and mental health [11]. Ware and Sherbourne (1992) [158] described the conceptual framework and item selection used for the SF-36. The goal of the first ten questions was to define the aspects of physical functioning and distinguish between patients able to perform physical tasks with or without difficulty. The second category, role functioning, was necessary to determine limitations in type or amount of work the patient is able to complete. Bodily pain questions captured both the frequency of pain and the amount of interference with normal activity, and social function questions assessed the effects of the patient's health on social or leisure activities. Mental health questions were chosen to capture four major areas, including anxiety, depression, loss of control, and psychological well-being, whereas four questions on vitality were targeted at assessing subjective well-being. The remainder of the questionnaire included general health perceptions. Scoring of the SF-36 used the Likert method of summated ratings, with scores in each section linearly related. One year after the form was established, McHorney et al. (1993) [159] reported on the SF-36 clinical tests of validity. Construct validity was used by establishing an internal structure for the observed variables, and verifying relationships between scores and other criteria. Previous validity tests were replicated to capture the two major dimensions of health represented by the questionnaire: physical and mental. Each scale was validated by matching patient groups to levels of severity as a measure of physical or mental health. Based on the validations, authors concluded the physical and mental health dimensions of the SF-36 are relatively pure; therefore when differences were observed between these dimensions, results were attributed to either physical or mental causes with a high degree of confidence.

Normative data for SF-36 from a large population was researched to be used for comparison with diverse population groups. Jenkinson et al. (1993) [58] sent the SF-36 to 13,042 randomly selected subjects between the ages of 18-64, non-stratified by age or gender. Data on 9,332 respondents, separated by age, gender, and health condition were reported. The sample was further separated into healthy individuals, those with long standing illnesses, and those who had received a health consultation within two weeks of completing the questionnaire. Women reported poorer health on all variables compared to men, exception for general health perception. Those with long-standing illnesses and those who had recently seen a doctor reported lower scores than the healthy group. Not only did the author provide normative data for comparison of QoL, their results also offered evidence for the construct validation of previous studies, and internal consistency on the domains presented in the SF-36.

Results from the SF-36 have also been compared between severe lower limb trauma patients who had transtibial amputation or limb-salvage surgery. De Brujin (2007) [56] reported on functional outcomes following peroneal nerve injuries, to include walking ability, strength, and QoL for a historic cohort of adults with unilateral peroneal nerve injuries. The health-related QoL measures in their population were significantly lower in physical functioning, mental health, vitality, and general health perception. Over two-thirds of the population reported limitations in stair ascension. Although muscular strength increased during five-year follow-up testing, 62% of patients still experienced paresis of the dorsiflexors to some degree. Despite this majority, only 11% still used their AFO. Seven patients had received a posterior tendon transfer, while others stopped using their assistive device due to improvement in dorsiflexion during swing, discomfort, or the use of orthopedic shoes. Authors concluded the lower mental health QoL might be related to the original reason for the injury, either trauma or tumor. They also reported patients may have adapt to their pathology five years post follow-up in a way that doesn't interfere as much with their daily lives, as evidenced by the normative reference QoL scores in role functioning.

Difficulty with stair ascension was a common theme in research that involved the limb salvage population. Bensoussan (2008) [55] noted this trend in an education review describing how to evaluate patients with gait abnormalities in physical medicine and rehabilitation. The step-by-step process included instructions on a clinical examination of health condition, body structure, patient activities and participation, and both qualitative and quantitative assessments of gait. This assessment was based on the International Classification of Functioning, Disability, and Health (ICF) model. The authors mentioned stair ascension was difficult for patients with leg muscle weakness, as the hip and ankle flexors are typically used to shorten the lower limb and the quadriceps and gluteus maximus push the body upwards. Also noted was the association between limb instability and knee collapse due to weakness of the quadriceps, and was cited as one reason for falls in populations with peripheral neuropathy. Other functional and postural assessments were described. This review was helpful in relating anthropometrics to gait assessment and quality-of-life in the drop foot population.

Patients with bone cancer have been studied in the literature as a population whose QoL was affected by limb salvage surgery (LSS) or amputation. Eiser et al. (2001) [18] focused solely on aspects related to QoL for LSS versus amputation, delayed amputation, comparisons to published norms, and an outline of factors that influenced QoL in that population. A retrospective analysis of patients treated for sarcomas over an 18-year span were the target of the study, and 34 patients volunteered to complete multiple questionnaires related to their amputation or LSS. Demographic, SF-36, body image, and everyday competence questionnaires combined with interviews revealed consistent findings with previous literature that LSS patients reported better everyday competence compared to their amputee counterparts. Patients with secondary amputations had lower SF-36 scores, however did not regret their initial LSS, as it gave them time to come to terms with the eventual amputation. Regression analysis indicated that QoL scores were related to everyday competence in the LSS population, possibly because LSS had higher expectations about what they should be able to accomplish compared to amputees.

Measurements of health outcomes are necessary for cost-effectiveness and cost-utility analyses. Quality-adjusted life years (QALYs) have been calculated as a measured descriptor of health outcomes [44, 147]. Texts by Drummond (1987) and Petitti (2000) offered guidance on cost-effectiveness and cost-utility analyses, and explained how weight indices were used to determine the Quality of Well-Being Scale (QWB), with a reliability of the weights at 0.90 [44, 147]. Tools other than the QWB to assess health-related quality-of-life, such as one widely by the military, is the aforementioned Short Form-36 (SF-36) based on health-related quality-of-life of eight domains: physical limitations, social limitations, physical work or role limitations, bodily pain, mental health, mental work or role limitations, vitality, and general health perceptions [53]. Extensive literature supported the use of the SF-36 questionnaire based on test-retest reliability, internal consistency, and criterion and construct validity [160-162]. Comparisons between the outcome scores of both tools on a large population and subsequent regression modeling was conducted by Fryback et al. (1997) [144]. The SF-36 and QWB were mathematically oriented, which higher scores considered better health. Both instruments had component similarities, and broad ordinal relationships were anticipated with the purpose to provide a tool for health researchers to use SF-36 tallies in lieu of QALYs. Participants in the study included 1,356 men and women over the age of 45 from Beaver Dam, Wisconsin. Researchers pursued the best regression polynomial using only the SF-36 scores, their squares, and the two-way cross products of the eight scores to find the best model with the fewest variables, considering all first and second-order terms for the prediction equation. Cross validation was conducted twice using separate sets of data. Low-to-moderate correlations between the variables determined the numerical scales of the SF-36 were unsuitable for use in cost-utility analyses; however the equation presented allows analysts with SF-36 data to translate to predicted QWB scores for cost-utility analysis use. The equation developed was $QWB = 0.59196 + 0.0012588 \times PF - 0.0011709 \times MH - 0.0014261 \times BP + 0.00000705 \times (GH \times RP) + 0.00001140 \times (PF \times BP) + 0.00001931 \times (MH \times BP)$, where PF = Physical Function, MH = Mental Health, BP = Bodily Pain, GH = General Health Perceptions, and RP = Role Function – Physical.

Studies comparing alternative treatments for other types of disabilities, e.g. diabetes, have used Fryback's equation to predict QWB scores from SF-36 assessments. Wu et al. (1998) [143] used a previously developed Markov cohort simulation model to estimate the associated total direct health care costs of a diabetes population. The SF-36 was assessed on a sample of 89 Type-1 diabetic patients who were cared for by a health maintenance organization for at least two years. Patients were divided into several groups according to age and health state for differing QALY calculation needs. A predicted QWB score from the SF-36 tally was used to calculate the total number of QALYs accumulated over time for the total population. These values were applied to a model to study cost-utility of an intensive versus conventional treatment strategy for retinopathy, an associated condition to diabetes. A 10-year simulation showed more QALYs with intensive therapy than conventional therapy, but intensive therapy is more expensive. Because of the delay in complications with the intensive therapy, it became less expensive after five years. Despite methodological limitations, Wu et al. concluded the model was reliable to estimate QALYs and useful to assist health planners in cost-effective analyses.

Quality-of-life measures specific to other populations with drop foot are important for comparison, especially researchers who considered a military population. Tekin et al. (2009) [57] reported QoL and functionality of daily activities, including energy expenditure on ten soldiers who underwent unilateral below knee amputation and nine soldiers with limb-salvage, all members of the Turkish Armed Forces. Evaluation tools included the Functional Ambulation Scale (FAS) which evaluated independent mobility of the soldier, the Visual Analogue Scale (VAS) which assessed local pain management, the Turkish validated version of the Short Form-36 (SF-36) tallied QoL scores, and Graves' radiological assessment method to evaluate ankle joint degeneration. Two walking tests (10-meter and 6-minute) assessed energy expenditure index (EEI) using Vmax 29C Sensor Medics, and limb-salvage patients performed all tests in the AFO they found most comfortable. Mann-Whitney U, Chi-Square, and Spearman correlation tests were completed using SPSS with a significance level set at $p<0.05$. Although gait parameters were relatively similar between the two groups, results of the SF-36 showed significantly higher scores for the amputee group in the areas of general health and vitality. This may have been due to the encouragement doctors provided amputee patients to become active in their lives, where limb-salvage patients fell somewhere between healthy and handicapped, and were often guided to protect their limbs, and were therefore less active. There was no significant difference found in the EEI for either of the walking tests between the two groups, perhaps attributed to the active military population and their ability to acclimate to altered ambulation patterns. Authors concluded that some areas of QoL measures were higher in the amputee population. Regardless of treatment, early rehabilitation was paramount for both groups.

The extensive construct validation of the SF-36 and the ability to calculate QALYs for cost-utility analyses throughout the summarized research validated the SF-36 questionnaire as an adequate tool for the present study. Researchers have been successful in using the SF-36 to identify aspects of QoL most affected by a population with limb salvage. Military populations have been assessed with this tool, although not as frequently and few with the drop foot pathology, and thus could benefit from further analysis of QoL. This questionnaire format combined with interviews and journaling should accurately assess the quality-of-life changes for a service member with drop foot.

Anthropometrics

Anthropometrics have also been useful to assist in clinical diagnosis of peroneal neuropathy, as well as a method to track improvements with rehabilitation over time. Muscular strength was measured using a Microfet 2 handheld dynamometer (Draper, Utah, USA) using procedures described by Kendall [46]. Kendall used muscle strength testing to determine the capability of muscle in terms of function, stability and support. The present study examined the return of muscular strength following trauma and repair, and the intervention of a DAFO. Objectivity and reliability were most important when using a dynamometer for manual muscle testing, and examiner technique was crucial for accurate measures. For example, the examiner's strength can affect the stability of a handheld dynamometer when testing hip abductor strength on stronger subjects. Two-joint and multi-joint muscles such as tibialis anterior, peroneus longus, brevis and tertius act like one-joint muscles and are tested for strength by actively shortening over the joints simultaneously at the end range with maximal shortening of the muscle.

The present study utilized one physical therapist (PT) for manual muscle testing of each injured service member, prior to and following completion of the six-month study protocol. The PT followed Kendall's [163] methods and was properly trained in this field to complete the testing. Intra-tester and inter-tester reliability of the inexpensive and practice manual muscle testing method was important to ensure consistent and accurate measures over time. This importance was highlighted by Kimura et al. (1996) [164] who investigated hand-held dynamometry (HHD) reliability between multiple testers and two specific devices: the Chatillon CSD500 and Microfet model dynamometers. Twelve certified athletic trainers (four females and eight males, mean age 27.7 years) practiced subject positioning and limb stabilization with the devices in two upper body and two lower body muscle groups until intraclass correlations (ICCs) of at least 0.90 were achieved. Testers then utilized random testing on 12 patients (three females and nine males, mean age 26.4 years) for strength in four muscle groups at the midpoint range of motion, counting aloud from one to three as subjects increased to a maximal muscle force. Test-retest sequences were performed twice with 30 seconds of rest in between to avoid fatigue. Moderate to high intra-tester ICCs were revealed for both dynamometers, and results were lowest when different strength groups were tested by different examiners with different HHD's. Intra-tester reliabilities were generally higher than inter-tester, especially with large muscle groups. Only five ICCs fell below 0.56 for the Microfet, and researchers concluded both HDD devices were found to be reliable when used by the same examiner; however the two devices used interchangeably on the same subject were not reliable. Kimura's results also referenced a study by Hosking that found hand-held devices suitable for testing in a pathologic population.

Hosking et al. (1976) [165] compared strength measures with a newly developed hand-held myometer on healthy school-aged children and children with neuromuscular disease. Strength measurements of the neck and hip flexors were taken on a healthy group of 300 boys and girls (ages 5-15) to identify faults with the myometer, and reproducibility studies were also accomplished. A second survey of 215 children (ages 5-14) underwent three trials of strength measures in six muscle groups using the Hammersmith myometer, using enough tester strength to overcome the children's maximal contractions. Measurements in a group of children with diseased muscle (n=32) were accomplished using the same procedures. Satisfactory correlations were found in the reproducibility studies for smaller muscle groups; however variability increased with larger muscle groups and subject age. Hip flexor and quadriceps muscle strength in adolescent boys exceeded the capability of the myometer in some cases. The children with neuromuscular disease performed to normative values in 8% of cases. A decreased ability to hold the leg up was noted in seven to ten year olds, possibly due to physical development. Authors concluded the hand-held device were suitable for identifying muscle weakness however not sensitive enough to monitor disease progression.

Proximal musculoskeletal weaknesses have been associated with gait deviations distally in the kinetic chain, which has proven important in rehabilitation strategy. Niemuth et al. (2005) [116] tested strength differences at the hip between healthy and injured recreational runners to determine any association between hip strength and running injuries. Thirty physician-referred recreational runners ages 18-55 participated in manual muscle testing of six hip muscle groups measured by a Microfet HHD. Muscle contractions were held for two seconds measured twice with adequate rest in between trials, and tested in accordance with prescribed methods. A group of 46 healthy control runners were also measured for comparison. Two-way mixed ANOVAs were utilized for muscle strength versus injury and leg between groups, and for duration of overuse symptoms and leg for the injured runners only. A relationship was determined between hip muscle imbalance and overuse injuries that did not exist in healthy runners. Lateral flexion of the trunk at heel strike was a shock absorption mechanism countered by balanced strength of the hip abductors. Injured group hip adductors were significantly stronger than the uninjured side, providing stiffness potentially as a compensation mechanism for the reduction in stabilizing hip abductor moment. Authors concluded that overuse injury treatment should include screening for hip muscle weakness, specifically the abductors.

The goniometer measurements used in passive range of motion follow the methods of Daniels and Worthingham (1986) [45]. The goniometer measures the angle of the joint range of motion which in a drop foot patient may be limited by injury and/or disuse. The measurement is based on a half circle (0 to 180°). Measurements for the present study included hip and knee flexion and extension, ankle dorsiflexion and plantarflexion, and calcaneal inversion and eversion. The range of motion for hip flexion is 0 to 120-130°, and 0 to 10-20° for extension beyond midline. Patient positioning for flexion was supine with hip and knee flexed, and prone with hips and knees straight for extension. Contralateral thigh stayed pressed against the table by the patient in both positions. The stationary arm was placed on the lateral longitudinal midline of the trunk for both flexion and extension. Flexion was measured as the thigh contacted the pelvis, and extension was measured beyond midline as the leg was lifted. Knee flexion range of motion is 0 to 135-145°, with extension the reverse. Patient positioning for flexion was supine with hip flexed to 90°, and the knee flexed completely. Extension positioning was supine with both hip and knee straight. The stationary arm was placed on the lateral longitudinal midline of the thigh, with the movable arm on the lateral longitudinal arm of the leg for both flexion and extension. Ankle plantarflexion and dorsiflexion were measured in the supine position, with the hip and knee straight in plantarflexion, and the knee partially flexed in dorsiflexion, with pillow support to lesson tension in the gastrocnemius. Range of motion for ankle plantarflexion is 0 to 45-55°, and dorsiflexion 0 to 15-25°. The stationary arm was placed on the lateral longitudinal midline of the leg for both plantarflexion and dorsiflexion. The movable arm for both movements was placed parallel to the fifth metatarsal with the fulcrum in line with the lateral malleolus. Calcaneal inversion and eversion were measured with the patient sitting on the side of the table with knees flexed, and the foot inverted or everted. Range of motion for inversion is 0 to 30-40°, and eversion 0 to 15-25 degrees. Both measurements started with the stationary arm placed on the lateral longitudinal midline of the leg, and the movable arm placed on the dorsum of the foot parallel to the second metatarsal. These techniques were considered standard care by the Tripler Army Medical Center (TAMC) Physical Medicine and Rehabilitation department, accomplished by the same TAMC physical therapist, and were part of the decision-making process for identifying soldiers with drop foot in need of a dynamic AFO to correct deviations from normal gait.

Published literature was useful to determine strength deviations from healthy populations, as well as compare levels of strength with other lower extremity trauma populations. Bohannon (1997) [59] established reference values for strength obtained through hand-held dynamometry. A convenience same of 106 men and 125 women were used to asses maximum voluntary force in Newtons using an Accuforce II device. Lower extremity measurements included ankle dorsiflexion, knee extension, and hip flexion and abduction. Average applicable forces in the dominant extremity for men age 30-39 included: ankle dorsiflexion: 372.6N (83.76 lbf), knee extension: 572.9N (128.79 lbf), hip flexion: 223.6N (50.27 lbf), and hip abduction: 329.1N (73.98 lbf). Measurements in parentheses were converted to pounds of force for comparison to the present study. The author then developed regression equations for sex, age, and weight with muscular strength. Conclusions from this study included confidence in reliable measurements were obtained, and were adequate for comparisons to impaired strength. The author reiterated the importance of tester strength and adherence to exact manual muscle testing methods for valid measures.

Archer (2006) [61] included strength measures from 24-month follow-up of lower extremity assessment project (LEAP) patients discussed previously, of which 381 limb salvage surgery (LSS) patients participated. Range of motion was tested using a universal goniometer, which was noted to have high intrarater reliability at the ankle and knee ranging from 0.82 – 0.99. Normative values from the American Academy of Orthopaedic Surgeons (AAOS) defined restrictive as 120°>hip flexion, 30°>hip extension, 135°>knee flexion, 20°>ankle dorsiflexion, and 50°>ankle plantarflexion. Sagittal strength measures at the hip, knee, and ankle as well as hip abduction were measured, but with a different technique than the present study: a strength apparatus with pivot clamps, force plates and a force transducer were used to assess patient strength. Researchers noted an association between decreased knee extension strength an abnormal spatial-temporal parameters in gait, but did not find this result significantly different in their study. Instead knee flexion strength and ankle plantarflexion ROM were low in their population. Wang (2007) [166] included ankle strength assessments in a study of gait and balance in patients with hemiparesis, and reported 26.72 lbf ankle dorsiflexion and 39.64 lbf ankle plantarflexion on the involved side, which provided another strength comparison to the injured service member group in the present study. Wang's study, however, did not identify the reason for hemiparesis in their population, and the mean age of subjects was 60.38 years. These differences were important to take into account when comparing strength values.

Understanding the methodology to successful strength measures is important for accuracy during manual muscle testing. Successful examiners have produced key studies to provide normative data of able-bodied persons for comparison to pathologic musculoskeletal deficits, which are in turn helpful to researchers in biomechanics as tools to justify gait deviations and the relationship to musculoskeletal deficits.

Lower Limb Parameters in Able-Bodied Gait

Knowledge of normal gait parameters is essential for understanding deviations in pathological gait. Differences in kinematics and kinetics also exist between walking and running conditions. Gait data from diverse, uninjured populations are essential for comparison with injured subjects. One of the most thorough works describing the gait cycle, ankle rockers, and techniques seen in the analysis of normal gait was provided by Gage et al. (1995) [107]. Deviations from normal gait may be caused by muscle weakness, joint position or muscle contracture. Kinematics in the sagittal and coronal plane were discussed, as they were slightly more repeatable than transverse plane data. Abnormalities throughout the kinetic chain were presented individually by joint, especially as they related to cerebral palsy. Drop foot was specifically mentioned as related to the interference in two out of four priorities of normal gait. Gage mentioned the text by Perry [167] when outlining the four priorities of gait, including stability of the weight-bearing foot through the stance phase, clearance of the non-weight-bearing foot during swing, appropriate position of the foot for the subsequent initial contact, and adequate step length. These four priorities were the baseline from which all gait abnormalities were discussed. This editorial was particularly useful when deciphering specific individual movements during abnormal gait and how they related to an overall change in the kinetic chain.

Mann (1980) [65] provided an additional reference for sagittal hip, knee, and ankle joint ROM in normal gait as well as electromyography of specific muscle groups during gait at varied velocities. Thirteen runners with different specialties (sprinting, jogging, and long-distance running) completed superficial electromyography and gait analysis by photometric methods at individualized velocities along a 150-foot runway. Spatial-temporal parameters increased with gait velocity in terms of step length, cadence, and cycle time. Range of motion at all joints also increased with gait velocity, likely due to a lower the center of gravity by increased flexion of the hip and knees, and dorsiflexion at the ankle. Hip and knee flexion increased but extension slightly decreased with faster velocities. Flexion at the knee increased from 10° in walking to 35° during running at initial contact. Quadriceps muscle contraction was active considerably longer during running then walking, from the last 20% of swing through 50% of stance, versus the last 10% of swing to 15% of the stance phase during walking. Shock absorption at initial contact was aided by knee flexion, followed by a period of extension. Shock absorption took place at the ankle during sprinting, however, due to insufficient time at the knee. The motion at the ankle changed considerably with an increase in velocity; initial contact foot position varied during running, but sprinters usually contacted with the toe. The authors concluded that hip flexion seemed extremely important during running; however the least amount of data were collected at that joint.

Normative walking values for angular motion at the pelvis, hip, knee, and ankle throughout three-dimensional trajectories of markers were presented by Kadaba et al. (1990) [168]. Forty healthy adults (age 18-40) with no history of musculoskeletal problems were assessed across a 9-meter walkway at self-selected natural walking velocities. Temporal-spatial parameters were also collected using foot switch data signals. Subjects were evaluated over three separate days for a total of nine data sets, which were averaged per individual and as a group for each point in the gait cycle. Sensitivity analysis was completed to quantify the effects of errors in estimation of the embedded axes. Temporal-spatial data were presented for men and women, and were found not significantly different. Reported walking values for males included cadence (112 steps/min), velocity (1.34 m/s), stride length (1.41 meters), and percent of stance phase (61%). No significant variables were found between men and women for kinematic measures as well. Mean and standard deviations for kinematic data were reported in graphical form, including pelvic tilt (15°), hip flexion at initial contact (IC) (37.5°), max hip extension (5°), knee flexion at IC (5°), max knee flexion in stance (17°), max knee flexion in swing (57°), ankle position at IC (2° of plantarflexion), max ankle dorsiflexion in stance (10°), and ankle position at toe-off (15° plantarflexion). Pelvic obliquity, hip adduction and abduction, knee varus and valgus and rotation at all lower extremity joints were also reported. Authors also included a table of joint ankle comparison data with previous work, making this a very useful study for walking gait comparisons of a pathologic population between ages 18-40.

Oberg et al. (1993) [64] provided normative spatial-temporal gait parameters for subjects ages 10-79. An even distribution of 116 men and 117 women were measured on a 10-meter walkway for slow, normal, and fast walking gait. Collection equipment included two photocells, self-aligning electrogoniometers, a computer, and a plotter. Ten trials without the goniometers were averaged for gait data, as well as three trials with goniometers. Two-way analysis of variance for age and sex were used to determine significant differences between sexes for all gait parameters, and age variability for gait speed and step length for normal and fast gait conditions. The authors noted that gait analysis is dependent on testing conditions, and comparisons with the resulting reference data should only be used for similar indoor testing conditions. This study provided several attachments with accurate normative gait data that is useful for comparison of spatial-temporal parameters in pathological gait.

Additions to the body of knowledge on walking and running gait was presented by Ounpuu (1994) [63], who also highlighting major differences between heel and foot flat striking styles. Although she used a sample of children which may have had smaller stride lengths and slower velocities, her sample was totally free of injury. Joint kinematics and kinetics were included in the sagittal, frontal and transverse planes for both walking and running conditions, as well as muscle activation, ankle rocker terminology, range of motion excursions, and same table comparisons of kinematic peak values for both velocities. Kinetics included forces, moments, and powers for the major joints of the kinetic chain, and all were represented graphically, in tables, and thoroughly described by plane. This analysis provided a reference point for consistent comparisons against the healthy service member controls used in the present study.

Novacheck (1998) [66] also provided a thorough report on gait differences between walking, running, and sprinting, although data was taken from previous research studies. The author delineated walking and running with two periods of double-float at the beginning and end of swing phase replacing the two periods of double-support in walking. Sagittal plane movements shifted into flexion with a lower center of mass as one changes from walking to running; however the pattern of pelvic tilt remained the same for all velocities to conserve energy and an efficient motion. The pelvis tilted further forward to assist in maximal propulsion. The hip reached max flexion during mid to terminal swing while walking, and max flexion just prior to toe-off. Running denoted similar degrees of flexion and extension, though they occur later in the gait cycle. Knee motion patterns between walking and running were similar, although max flexion and extension joint angles were more extreme in running. Maximal knee flexion in running stance (45°) occurred for absorptive purposes, followed by 25° during propulsion. The largest differences in knee flexion occurred during the swing phase, with a max knee flexion of 60° in walking, and an average of 90° in running. Eighty percent of runners were heel strikers, so the initial contacting during walking and running was the same for the ankle. However, during walking the ankle was technically in plantarflexion due to the relation to the tibia, where true dorsiflexion was exhibited during running. Both conditions exhibited approximately the same amount of plantarflexion at toe-off (20°). The author also combined kinematics with ground reaction forces to report calculated joint moments and powers. In the sagittal plane, ankle moment pattern was similar in walking and running, with an ankle plantarflexion moment occurring before 10% of the gait cycle. Power at the ankle for forward propulsion was directly related to the velocity of movement. There was very little knee moment in the stance phase of walking, and the magnitude of the peak knee extensor moment during running was related to the amount of knee flexion during loading. Hip moment pattern, like the ankle, was similar in both walking and running conditions. An internal flexion moment occurred at initial contact, but only the walking condition saw an extension moment prior to toe-off. Conclusions from the kinetic data found the main sources of power generation during forward propulsion occurred from the hip extensors during the transition from swing to stance, the hip flexors following toe-off, and the knee extensors, hip abductors, and ankle plantarflexors during stance phase.

Current ankle-foot orthoses research spans an entire spectrum of kinematic gait parameters, with very little consistency in testing. Determining the most important variables associated with normal walking and running gait is important not only to use as a reference for control, but also to make possible the comparison of one AFO to another. Information is limited on the cohesive reliability of kinematic gait measurements, but in response to this lack of information, McGinley et al. (2009) [169] conducted a systematic review of gait analysis studies to determine inter-session and inter-assessor reliability data for three dimensional kinematic gait analysis data. Extraction criteria included subject characteristics and recruitment, procedures, biomechanical models, and statistical analysis. The QUADAS tool, questionnaires, and expert panels were used to evaluate inclusive studies, and quality indicators were compiled into a standardized form to ensure structure during the review process. The systematic review yielded 23 studies (15 full papers and 8 published abstracts), with primary reference to gait laboratories collecting multi-joint lower body gait kinematic data. The kinematic measures with highest reliability were reported in the sagittal plane (>0.80) excluding pelvic tilt. Good reliability was considered to be less than two degrees of error, with 2-5 degrees considered widely acceptable with explicit consideration during data interpretation. Authors determined the studies with the highest within-assessor reliability included sagittal measurements of hip flexion, knee flexion, and ankle dorsiflexion.

Reliable sagittal gait measurements were validated by Diss et al. (2001) [170] after examination of 24 kinematic and kinetic variables from three different synchronized systems to determine the reliability of variables used to analyze gait. Five male runners (mean age 23.4 years) were tested on two separate occasions over the course of seven days. Ten acceptable running trials of consistent velocity (3.5-4.0 m/s) were completed over a Kistler force plate collecting at 500 Hz down a 15-meter runway. An amplifier and analogue-to-digital converter were used to capture the data from two Panasonic video cameras with 1/500 second shutter speed. Smoothing was accomplished using a generalized cross-validated quantic spline, and analysis of variance (ANOVA) was used to analyze the means of these trials. Authors concluded highest reliability found for time to maximal eversion, ankle and hip joint angle at toe-off, and total range of ankle motion with an overall 80% reliability rating for the 24 variables analyzed over the mean of three trials. Based on the reliability measures and conclusions of the aforementioned studies, biomechanics kinematic variables were mainly limited to the sagittal and coronal planes.

Specific gait parameters along various points in the proximal kinetic chain can explain changes distally. Some researchers also believe changes directed at the foot can be reflected proximally. Lewis and Ferris (2008) [98] examined the relationship between increased movement at the ankle and hip moments. The bipedal models of walking in powering gait can be manipulated by different methods: apply and impulsive push with the trailing limb at toe-off to redirect the center of mass forward and upward, or apply a torque between the limbs to pull the swing limb forward. Authors noted the latter is four times more energy consuming compared to the impulsive push, but investigative research is needed to determine the hip strategy. They hypothesized that increasing ankle push-off would be inversely related to peak moments at the hip. Ten healthy subjects (male and female, mean age 27) walked on a custom force-measuring treadmill with a motion capture system (120 Hz) recording reflective marker activity of the lower extremity. Subjects were instructed over three conditions to walk normally, with more push at the foot, and with less push at the foot. Subjects practiced each condition for at least one minute prior to data collection, and conditions were randomized. Visual 3D was used to calculate internal joint moments based on marker and force plate data. The results supported the initial hypothesis; when ankle plantarflexion impulse was higher, a decrease was seen in hip flexion and extension moments and impulses, as well as powers.

Relationships between the ankle and hip were also investigated by Bowsher and Vaughan (1995) [108] , who looked at the effect of foot progression angle on hip moments during normal gait, and hypothesized that toe-out angle had a significant effect on hip internal-external rotation moments during level walking. Six males and six females (ages 23-32) performed a total of nine trials, three of which were ‘foot straight’, three with the foot internally rotated by 30°, and three with 30° of external rotation. Anthropometric data were collected to calculate body segment procedures by Vaughan et al. (1992), and gait data were captured with a four-camera motion analysis system and reflective marker set described be Kadaba et al. (1990). Kistler force plates were used to collect ground reaction forces at 60 Hz. An inverse dynamics approached used all data collected to solve for resultant hip moments expressed about three axes, and were normalized by body weight and leg length, then multiplied by 100. Researchers determined that torsional load on the femur could be reduced by turning the foot outward, by a factor of 5 Nm. They noted uncertainty about whether this was a clinically significant reduction, and further research should explore this reduction for the purposes of improving total joint arthroplasty, especially at the hip.

Manipulating the foot in other planes is also useful in determining foot strike patterns, especially during running. Stackhouse et al. (2004) [92] analyzed the effects of orthotic intervention in both rearfoot strikers (RFS) and forefoot strikers (FFS) during running. Authors expected the foot orthotic devices (FODs) to decreases rearfoot eversion, eversion excursion and velocity with the RFS pattern but not the FFS pattern. Fifteen healthy runners (ages 18-45) exhibiting a RFS pattern were included in this study. Runners adjusted to custom FODs over a period of two weeks, after which gait was analyzed at 120 Hz using a Vicon 6-camera motion capture system, retro-reflexive markers and non-collinear tracking markers on each segment. Subjects work identical running shoes with holes cut in the heel to allow marker attachment directly to the calcaneus. A 10% reduction in heel counter stability was noted as a result of the holes, as measured by an Instron testing device. Five trials were completed in each of four conditions: RFS without FOD, RFS with FOD, FFS without FOD, and FFS with FOD. A 3.7 m/s ± 5% velocity was monitored with two photoelectric beams and a timer. Although the frontal plane was the focus of the study, authors also determined peak dorsiflexion velocities, dorsiflexion excursion, and sagittal motions at the knee. Two-way repeated measures ANOVA revealed the FODs had a similar effect on rearfoot mechanics regardless of running pattern. Dorsiflexion excursion increased with use of the FOD, which brought to light questions regarding mid-foot movement in the sagittal plane. This increase may have been a compensatory mechanism to allow forward progression of the tibia over the foot due to the reduction in rearfoot motion. The same sample was used for another article related to strike pattern and orthotics by Laughton et al. (2003) [93], who examined the variables related to loading rates and stiffness. These variables demonstrated an increase in tibial acceleration by increased vertical GRF, peak anteroposterior GRF, loading rates, and knee stiffness in the FFS compared to the RFS pattern. Authors revealed a FFS pattern allows for decreased overall stiffness in the joints and entire leg compared to RFS pattern, but with greater tibial acceleration. Further research is needed to determine if one strike pattern has a greater propensity for injury over the other.

Optimal function of an AFO also includes an understanding of how direct forces are applied to the leg to limit discomfort, as well as forces applied to the AFO to withstand vigorous activity and wear and tear. McHugh (1999) [88] explained how to determine the forces and moments acting on a leg in various stages of gait, demonstrated the process of determining force vectors and plantarflexion and dorsiflexion moments associated with these forces due to a lack of muscular power about the ankle, and the feasibility of predicting the magnitude of these forces to use in the design of an ankle-foot orthotic. During early stance, the muscular insufficiency of the dorsiflexion muscles caused the external plantarflexion moment from the ground reaction force to exceed the internal dorsiflexion moment, resulting in the indicative foot slap in the nerve palsy condition. A measurement of these associated forces can assist an orthotist in determining the compensating rigidity of the AFO to avoid deformation. The ground reaction forces in early and late stance phases were 10 Nm and 100 Nm, respectively. The ankle moments in swing phase are much less (approximately 1 Nm), and therefore an AFO which compensates for plantarflexion deficiencies in this stage of gait needed to be more rigid than a design which compensated for a lack of dorsiflexion in early stance or swing phase. Spasticity must also be considered when designing the rigidity of an AFO, since the force levels were different according to the degree of spasticity of the individual. These force measurements greatly changed the degree of rigidity needed for a custom AFO, therefore emphasizing the need for individual patients with conditions warranting an AFO to be individually assessed and measured for the greatest success in returning to a more normal gait.

Changes in ground reaction forces are detectable in pathologic populations, and therefore are important to consider in normal gait to understand deviations. Bellchamber and van den Bogert (2000) [171] examined the relationship between forces on the foot during walking and running gait, and the effect on internal and external tibial rotation. Twenty healthy subjects (10 male, 10 female) completed twelve acceptable walking and running trials at 4.0 ± 0.4 m/s and 2.0 ± 0.2 m/s, respectively. Kinematic and force plate data were collected using four motion analysis cameras collecting at 240 Hz, and a Kistler force plate sampling at 2400 Hz. Shank markers 12mm in diameter were secured to five sites at specific locations on the fibula, tibial tuberosity, anterior tibia, lateral malleolus, and lower shank. The shoes were fitted with markers in specified areas of the calcaneus and metatarsal, while lateral epicondyle and lateral malleolus were used during the static trial and removed prior to dynamic data collection. Angular velocity and moments were calculated with respect to the tibial axis, with proximal control identified as velocities in the opposite direction of moment of force, and distal control as velocities in the same direction as moment of force. During walking trials, internal tibial rotation drove pronation, and because of this, orthotics may have controlled eversion to the point of additional stress on the tibia and knee. During running trials, brief periods of axial tibial rotation seemed to be driven by the foot, measured by positive power flow patterns. Rapid oscillations in the angular velocity curves may have suggested errors in marker movement, however researchers were confident that results greater than 1.4 rad/s represented true tibial rotation. Researchers concluded that positive power flow patterns may benefit from foot orthotics, but notes the high variation of power flow patterns between one-quarter of subjects tested.

Examining the magnitudes and durations of ground reaction forces (GRF) enabled Munro et al. (1987) [81] to develop reference standards for GRF data as a function of running velocity. This investigation also provided the opportunity to reexamine questions related to GRF from the literature, specifically regarding the medial-lateral GRF pattern variability reported within and between subjects. Twenty adult male subjects with recorded weekly running mileage (22.8 ± 14.8 km) ran at various velocities (2.5 - 5.5 $m \cdot s^{-1}$) across a Kistler force platform for 20-30 trials. An equal number of right and left foot contacts were attempted. Velocity was determined using photoelectric cells placed at neck level, five meters apart on each side of the platform. Conclusions regarding medial-lateral GRF were notably variable within and between subjects, but most distinctly demonstrated bilateral asymmetries within a subject when compared with vertical or anterior-posterior GRF. Since GRF reflections acceleration of total body center of gravity, determining medial-lateral GRF changes resulting from shoe-type, orthotics, or other braces would be difficult discern even without the inherent variability within this axis. These results are useful for purposes of atypical running pattern assessment and rehabilitation progress charting.

Kinetic differences between velocity conditions and gait patterns have also been explained through energy exchange. Bertram et al. (2002) [172] explored the relationship between pace length and mechanical energy of the center of mass in level walking [172]. Three male volunteers with approximately equal pelvic lengths but varied body masses participated in the study. An electronic force plate was positioned on a constructed platform of raised planks at equal distances apart. Volunteers were instructed to walk at a comfortable pace that allowed a steady walking velocity across the platform. This steady-state was verified by computing the horizontal impulse for each pass. Individual platforms were constructed for various pace lengths, and volunteers were asked to continue this process until they reached a platform that required they switch to a running gait. Researchers collected three to five passes of each platform, with surface markers attached to the volunteers at the ankle, knee, and hip. Video recordings electronically shuttered at 0.002 seconds recorded the passes at 60 fields/second. Measurements from the calibrated video fields were made using an image analysis application. Vertical and horizontal ground reaction forces were collected from the force plate system and were converted to accelerations, and the integral was taken to determine change in vertical velocity. Gravitational potential energy (E_p), and translational kinetic energy (E_k), were then calculated from the change in vertical position. Apparent limb compliance was calculated from the change in length of the support limb from heel strike to mid-stance, and pace length was calculated into dimensionless terms. Results indicated the relationship between forward velocity and foot contact time remained consistent regardless of pace length variations. Changes in both vertical and horizontal acceleration were also associated with relative pace length. Translational kinetic energy increased systematically with pace length, due to the increase in average forward velocity at longer pace lengths. Shorter pace lengths caused kinetic energy oscillations due to a longer foot contact time at shorter velocities. Translational kinetic energy was the least when the center of mass was directly over the supported limb in all pace lengths.

Studying gait deviations of musculoskeletal injuries in otherwise healthy populations may help in observational gait analysis of the premorbidly fit military population. Establishing a knowledge base of injuries associated with running is especially important due to the injury pathology in one of the military members in my study. Willems et al. (2006) [50] provided insight into the gait-related risk factors associated with exercise-related lower leg pain of 400 physical education students. All students were biomechanically analyzed for three dimensional (3D) kinematics and plantar pressures prior to beginning a college level physical education semester. Using a foot scan pressure plate and Proreflex infrared cameras, subjects were markered with a set designed by McClay and Manal and ran barefoot at 3.3 ± 1.7 m/s. Standing alignment and three trials for each foot in a running gait were collected, determining five instants of foot rollover, and processed through Visual 3D for running stance phase data. Univariate Cox regression analysis revealed the effects of each gait variable on injury hazard, based on prospective tracking of those students who developed exercise-related lower leg pain over the academic year. First metatarsophalangeal joint extension range of motion was the only alignment variable different between the injured and uninjured group.

Biomechanically, three main gait deviations were apparent in the injured group: central heel strike at initial contact, higher eversion of the foot during forefoot contact and loading response (foot flat), and increased reinversion velocity with lateral roll-off. A follow-on study by Willems et al. (2007) [51] compared this cohort of subjects to the same group while running in shoes. The same methodology and analysis methods revealed an additional variable of increased pronation with prolonged eversion in addition to the second and third gait deviations previously described, although the results were more subtle shod than barefoot. These results provided a comparison of the differences between barefoot and shod gait patterns and values, and the intrinsic risk factors found in these studies may help clinicians in rehabilitation exercise-related lower leg pain.

Crossley et al. (1999) [52] also addressed gait related lower extremity injuries by comparing runners with and without history of tibial stress fractures to tibial bone mass and geometry, and ground reaction force (GRF) parameters. Forty six male runners evenly split between those with tibial stress injury and healthy controls completed ten successful running trials over a force plate at $4.0 \pm 10\%$ m/s. Peak vertical and horizontal GRF parameters as well as GRF timing variables were normalized for body mass prior to analysis. Runners also completed dual energy x-ray absorptiometry (DXA) and computerized tomography (CT) scans measured bone area, bone mineral composition (BMC), and bone mineral density (BMD). Independent t-tests were used to compare the injured and control groups, finding the injured runners had a smaller tibial cross-sectional area over the control group. No significant differences in GRF, BMC, or BMD were reported.

Based on current research of normal lower limb gait and the differences between walking and running conditions, the most important variables to consider for comparison of gait in an AFO to normal are spatial-temporal changes, sagittal and frontal views at the pelvis and hip, knee, and ankle joints kinematically, and ground reaction forces and corresponding moments at each joint kinetically. Additionally, optimal energy exchange with a dynamic AFO may be related to pace length, possibly making gait re-training an essential part of rehabilitation to return a soldier to optimal function for combat.

Ankle-Foot Orthotic Design Categories

The main focus of ankle-foot orthotic research has been engineering and design. The competitive nature of research patents makes AFO designs are difficult to compare on established name alone. Ankle-foot orthoses have been characterized based on materials, assessment of static and dynamic gait pattern, functionality, and comfort. Some AFO designs incorporate the inclusion of a shoe, while others are designed without shod conditions. The methods of testing and comparing the AFO designs are as numerous as the number of orthotics in development. A look into the evolution of AFO design provides a foundation of knowledge necessary to understand AFO comparison studies.

The main characteristic distinguishing between major AFO designs: active dynamic (operation of the device using motors, pumps, and actuators), or passive dynamic (the device relies on material composition to regulate the storage and return of mechanical energy) was coined by Faustini et al. (2008) [34]. To duplicate a passive dynamic AFO (PD-AFO) from a Dynamic Bracing Solutions carbon fiber AFO (CF-AFO), the feasibility of selective laser sintering (SLS) in the manufacturing process was determined, as well as the optimal SLS material needed for a PD-AFO to store and release energy. Three different materials were compared in the damping of CF-AFO to PD-AFO. The SLS method was used, fabricating the AFO in sequential cross-sectional layers from a unique digital image of the patient's limb. Researchers developed a manufacturing framework which included: 1) determine the bending stiffness of the CF-AFO, 2) develop a surface model of the CF-AFO, 3) create a computer-aided design from the surface model, 4) perform FEM analysis using different load conditions, 5) modify the strut dimensions to mimic that of the CF-AFO, and 6) use SLS to fabricate the prototypes. The CF-AFO was scanned through computed tomography (CT), delineating the surface of the AFO through NIH software. The FEM analysis continued until the desired rotational stiffness was achieved, and then files were converted to STL format and the AFO was manufactured in a 3DSystems Vanguard HS Sinterstation using three separate material prototypes. The three prototypes were then tested using a nondestructive technique, but clamping the footplate of the AFO to a vertical base, and applying an increasing vertical force to the cuff through a cable system. Energy dissipation for each prototype was also measured by a calibrated accelerometer. Each prototype was rotated 20 degrees in the sagittal plane, released, and measured by the accelerometer for 30 seconds. Destructive tests were also used to determine flexural strength using a hydraulic axial load cell. The mechanical damping characteristics rank ordered the SLS materials, and Rilsan D80 proved to be the material that had the least energy dissipation and held up to the destructive testing. However, this material still had higher mechanical damping when compared to the DBS-AFO, which may contribute to patient fatigue during AFO use. One potential limitation to SLS fabrication is a height limitation due to the layering process, but authors concluded this manufacturing process holds promise for fabrication and customization of future ankle-foot orthotics. This study demonstrated the difficulty of manufacturing a dynamic AFO, and the highlighted the advanced design of the DBS-AFO used in this study.

Variable joint impedance designed to reduce the occurrence of slap foot and improve gait symmetry in patients with peroneal nerve damage were the design approach of Blaya et al. (2004) [22]. This author presented an active ankle-foot orthotic (AAFO) design with actuators and sensors attached to a modified conventional AFO to measure the angle between the shank and foot, and ultimately determine ankle velocity. An Ultraflex system was used to measure ground reaction forces through sensors placed on the bottom of the AFO. The adaptive controller connected to the AFO measured three different states, aiming at controlling foot slap, allowing plantar flexion movements, and prevent toe drag. At each velocity, orthotic joint stiffness was adjusted manually to mirror swing dorsiflexion velocity of the unaffected foot. An eight-camera Vicon system (120 Hz) with two AMTI force plates collected kinetic and kinematic data on two drop foot patients measured at slow, self-selected, and fast velocities. The AAFO was donned in three control conditions: zero, variable, and constant impedance. Multiple comparisons using a one-way ANOVA at an alpha level set at 0.05 determined different means in gait symmetry. The author determined that by actively adjusting joint impedance, slap foot occurrences can be reduced, and swing phase ankle kinematics more closely resemble normative values. Joint impedance did not affect gait symmetry when evaluating spatial-temporal variables such as step length. Controlled plantar flexion stiffness increased two-fold from slow to fast velocities. Constant swing phase impedance caused both subjects to catch their toe during swing phase and thus lose balance. Anecdotally, both patients found the AAFO to subconsciously improve their walking. Researchers concluded that joint impedance is an important characteristic to include in future ankle-foot orthotic designs.

Evidence exists between AFO ankle stiffness and patient gait related problems, prompting Bregman et al. (2009) [30] to design a device to quantify this stiffness and neutral angle about the ankle and metatarsal-phalangeal (MTP) joints. The Bi-articular Reciprocating Universal Compliance Estimator (BRUCE), measures the mechanical characteristics of the AFO, shoe, and AFO-shoe combination, and accommodates a large variety of AFO types. The linear model was displayed real-time by Matlab, with user feedback plotted as measured angle versus net moment. Reliability was measured using a G-study to determine the influence of operator, occasion, and repetition on error variance. Four differing AFOs (two custom carbon fiber posterior leaf springs of differing stiffness, one rigid polypropylene AFO, and a Dyanfo posterior leaf spring confection AFO) were tested by three separate testers, with three different measures. The procedure was repeated two days later for a differing occasion. The AFO casts were based on a healthy subject with a 25 cm foot length. Based on the G-study, most of the variance could be attributed to the different AFOs, with the largest error variance found in the neutral ankle and MTP joints. Clinically, researchers concluded the device is fast in donning and doffing, and the measurement of MTP joints is reliable with different testers. This device assists in gaining insight in the match between patient gait characteristics and AFO characteristics.

Other technologies in AFO design include power harvesting capacity of a bellow pump to generate the required pneumatic pressure for toe clearance. Chin et al. (2009) [25] investigated the effects of a power harvested ankle-foot orthosis (PhAFO) on lower limb joint behavior. The stand-alone device does not require a shoe; fitted to a subject by two major components: a posterior tibial section and foot piece. The toe section of the foot plate was oriented five degrees to the ground to emulate late stance rollover, and connected by a free motion ankle hinge secured by Velcro. The lateral aspect of the AFO held the actuator and cam-lock control mechanism, designed to influence gait parameters through specific motion-control tasks. Fluid power was harvested through a circuit between the foot piece and foam sole. The bellow pump used for this power is made from a hypalon molded accordion bellow epoxied to a conical compression spring and two polycarbonate plates. During stance phase, the foot compressed the bellow, acting as the power source for the pneumatic circuit. This compression allows air to be discharged from a linear cylinder, with a spring inside that retracts an actuator rod to disengage the cam lock for free ankle movement. This allows dorsiflexion in mid-stance to roll over the cam surface. When the locking mechanism is engaged, the ankle joint should not be able to plantar flex or dorsiflex from neutral. Researchers conducted a pilot study on a healthy control subject (age: 22, weight: 85kg, height: 174 cm). The healthy control subject (male, age 22) walked on a split-belt instrumented treadmill with two imbedded force plates at a self-selected pace. Two 2-minute walking trials were performed, one in regular running shoes, and the second in the PhAFO on the right foot, and the same running shoe on the left foot. Kinematic data were collected using a six-camera Vicon motion analysis system sampling at 100 fps, with markers attached to the head, torso, arms, and legs. Ground reaction force data were collected at 1000 Hz, and both sets of data were low-pass filtered at 14 Hz and 39 Hz, using a fourth-order, zero-lag Butterworth filter. Kinetic and kinematic measurements indicated successful control of plantarflexion by the PhAFO during swing phase, as well as permitting free ankle moment during stance. The unlocked ankle joint following initial contact allowed for full sagittal range of motion (ROM). The subject displayed excessive dorsiflexion in swing phase, although researchers determined this should not occur for drop foot patients due to dorsiflexion weakness, but noted the observation as a design parameter that needs refinement. The PhAFO was also able to repeatedly harvest fluid power, consistently generating pressures over 150 kPa, always exceeding the minimum pressure needed to activate the linear cylinder and thus actuator. Use of the PhAFO did change timing of peak dorsiflexion, as well as stance-swing phase timing. Stance phase was shorter for the PhAFO limb (63%) when compared to the control (67%), and was noted as a limitation to conventional mechanical ankle joints and this study, along with a small sample size. Miniaturization of components is a future research emphasis, but authors concluded this study a successful look into the harvesting of fluid power through means of a bellow pump and pneumatic circuit.

Fatone et al. (2009) [173] investigated the effect of ankle-foot orthosis alignment and foot-plate length on sagittal plane ankle and knee kinematics. Sixteen adults (mean age 55 years) with post-stroke hemiplegia and no major involvement of the contralateral limb volunteered for gait analysis in four conditions: shoes only, an AFO with PF stop set to 90° with a flat full-length footplate (CAFO), an AFO with an alignment change of the PF stop by 5-7° and unaltered flat full-length footplate (HHCAFO), and an AFO with ankle alignment unaltered from the HHCAFO, but the footplate was trimmed to three-quarter length (3/4 AFO). Each subject was given two weeks to familiarize to each condition, and then tested in three separate trials at self-selected velocity. An eight-camera motion analysis system and six force plates were used to record data for each subject affixed with a modified Helen Hayes marker set. Normal reference gait data were collected from 12 control subjects at five different self-selected velocities ranging from very slow to very fast. Orthotrak software calculated temporal spatial parameters, kinetics, and kinematics. Nonparametric Friedman tests determined differences between repeated measures at an alpha of 0.05. Significant results were followed by a Wilcoxon signed-rank test with a Bonferroni correction as used with an alpha level of 0.008. Mann-Whitney and Kruskal-Wallis tests determined if there were differences in the independent groups. Researchers found that compared to no AFO, all conditions decreased PF of the ankle at IC and MS, increased early stance peak knee extensor moment and increased CoP excursion. Articulated AFOs with a PF stop and a full-length footplate demonstrated improvement in sagittal plane stance and swing phase ankle joint orientation, as well as improved knee moments during early stance.

Normalizing the ankle rockers is an important design outcome for many orthotists. Ounpuu et al. (1996) [174] evaluated the effects of a posterior leaf spring (PLS) orthotic on the gait of children with cerebral palsy to determine ankle rocker normalization using computerized gait analysis techniques. Children diagnosed with cerebral palsy (mean age 10.6, n=31) were fitted with reflective markers on the pelvis, bilateral thigh, shank, and foot, and were asked to walk with their normal gait down a 30m walkway while kinematics and kinetics were collected using an infrared camera system and force plates. Three trials per foot were collected for both barefoot and PLS conditions, and paired *t*-tests were used to examine the differences between the two conditions, as well as Pearson correlations to evaluate relationships between different variables. Results from this study indicated the PLS is successful in improving ankle rocker modulation during stance, indicative of more normative values in each rocker condition. The PLS improved dorsiflexion in terminal swing which allowed for sufficient ground clearance that phase of gait. The device was flexible enough to allow for dorsiflexion in mid-stance, but not too flexible as it reduced peak plantar flexion in pre-swing to eliminate the condition of drop-foot. One limitation of the PLS is the energy dissipation during mid-stance which is apparent by the reduced push-off capability, although power generation is not totally eliminated in the orthotic.

Another device to measure stiffness by the dorsi- and plantarflexion moments of plastic ankle-foot orthoses was developed by Sumiya et al. (1996) [175]. A molded plastic foot and leg model was attached to a plastic AFO for use in deformation testing of the device. Tensiometers were attached to two metal bars acting as a lever at the ankle axis. The ankle moment applied to the orthosis was measured by the tension force times the lever arm, with a protractor setting the deflection angle in 2.5° increments. Reproducibility tests confirmed accuracy of the device at four force levels, ranging from 0-30 Nm. The maximum moment this device measured was 40 Nm. Despite measurement errors from the friction between the pipe and the leg model, gravity, and the manual application of force and control of velocity, these errors can be controlled with same-tester reliability. Methods are in place to recognize when incorrect measurements are made. Results indicated AFOs produce a dynamic moment accompanied by material fatigue. Limitations of this device make it appropriate for plastic AFOs made of low-viscosity materials, and only static measurements should be made. Despite the limitations, this device provides a simple mechanism for testing AFO stiffness with high reproducibility. Sumiya et al. (1996) [176] continued research on AFO stiffness measurement by quantitatively determining the change in orthotic stiffness corresponding with regulated ankle trim lines on human subjects, with the intention of advancing prescription criteria of the AFO. Thirty posterior-type plastic AFOs were fabricated for 24 patients and six healthy control subjects. These AFOs had a proximal trim line 3 cm below the fibular head, and a distal trim line extended to the end of the toes. The AFOs were placed horizontal to eliminate the effects of gravity, and tested using the previously developed stiffness device for ten trials at each of the nine-stage ankle trim lines. One 55-year-old patient with left-side hemiplegia was simultaneously clinically tested using the nine-stage trimmed AFO, with observations and interview questions completed at each of the nine stages. Based on the averages of the ten trials and the clinical assessment, each trim line stage displayed 5-15° of dorsi- and plantarflexion corresponding to each trim line stage. The requirements for toe clearance from the dorsiflexion assist are related to knee stabilization at heel strike. For these reasons, dorsiflexion assist should be minimized during swing phase for safety to the knee at the next heel strike. Researchers also found the AFO with anterior stop to successfully substitute for the triceps surae during flaccid paralysis, though this does not seem to hold true in the literature for all populations. Therefore muscle tone, gait pattern, and energy consumption should be taken into account when adjusting for AFO plantar flexion assist. Due to the vast differences in gait pattern needs for hemiplegic patients, researchers concluded plastic AFOs are most appropriate for those requiring dorsiflexion assist, and the ankle stiffness must be individually adjusted through trimming to ensure maximum support.

Yamamoto et al. (1997) [177] determined the characteristics needed for the development of a new AFO based on the magnitude of the dorsiflexion assist moment and the initial ankle angle. Prior to the current study analysis, a pilot study was completed on four hemiplegic patients to proper fitting of an experimental AFO. Various dorsiflexion assist spring combinations for various initial ankle angles were measured to determine the best possible combination for each patient. Angular displacements and moments were measured by goniometers and capacitive transducers to ensure accurate temporal factors. Results of this pilot study verified the current literature, that plantarflexion assist moments are not necessary and may impede a successful gait for hemiplegic patients. Criteria were determined to adjust mechanical characteristics of the AFO for each subject, and this fitting methodology was then carried over into the procedures for the current study. Thirty-three hemiplegic patients comfortable with plastic AFO use were fitted during the methods previously determined in the pilot study. Authors summarized the characteristics of the new AFO as including articulated ankle joint and corrective inversion/eversion ability, a 0-10 degree initial ankle angle of dorsiflexion, range of dorsiflexion should be within 30 degrees of initial dorsiflexion angle based on outside factors such as stair use and squatting, the AFO should not generate a PF assist moment during DF, range of PF should not exceed 10 degrees from initial ankle angle, adjusted for downward stair use, and the AFO should generate a DF assist moment during PF in the range of 5-20 degrees of DF for every 10 degrees of PF.

Two years later, Yamamoto et al. (1999) [178] evaluated five hemiplegic patients while wearing a newly developed dorsiflexion assist AFO controlled by a spring (DACS-AFO). The DACS-AFO was design based on desired characteristics from a previous study. Subject gait was evaluated using a three-dimensional Vicon position sensor system. Five sets of gait trial data were averaged for each subject and ankle-joint angle, knee-joint angle, forward progression of the hip joint, and rotation of the foot in the horizontal lane were determined. Walking velocity was also measured when possible. Patients evaluated the comfort ability of the DACS-AFO using a written questionnaire. Preliminary test results indicated a smoother gait could be achieved with the DACS-AFO compared to other conventional AFOs, but the DACS-AFO could not reach 20 Nm of dorsiflexion assist moment per 10° of rotation as previously hypothesized in the developmental stages. Patients didn't indicate issues with the weight or donning the DACS-AFO. Mechanical endurance was at the time still being completed to check for durability. Authors concluded the current DACS-AFO design with a moderate dorsiflexion assist moment and initial ankle-angle adjustments help a hemiplegic patient increase knee extension forces and therefore thrust during initial stance of gait.

The closest DAFO design to the brace used in this study was created by the Center For the Intrepid at Brooks Army Medical Center in San Antonio, Texas. Patzkowski et al. (2011) [42] reported on a case study of a military limb salvage patient outfitted with the new dynamic ankle-foot orthosis (DAFO) called the Intrepid Dynamic Exoskeletal Orthosis (IDEO). Also described was a tailored rehabilitation program by the military specifically for wounded limb salvage service members to assist in the return to vigorous activity. This case study subject was injured in a Humvee rollover, sustaining a severe open ankle fracture. After several surgeries and prolonged hospitalization for six months, this soldier was fitted with a DAFO made by Dynamic Bracing Solutions, but reported the brace was uncomfortable despite successful return to vigorous activity. Six months later he was refitted for an IDEO and noted improvements in comfort and function, returning to combat readiness after another six months of rehabilitation. The orthopedic physicians working in concert with the orthotist developers of the IDEO often will complete ankle fusion surgeries if the injury precludes salvage of the tibiotalar joint, noting the IDEO performs better with an ankle at neutral dorsiflexion. The IDEO as a DAFO is designed with the goal of energy storage with tibial advancement during mid and terminal stance as the ankle dorsiflexes, and returning the stored energy in the form of ankle power at the initial swing phase. The brace is made of mostly carbon fiber paired with a modified Ottobock Carbon 7 posterior-mounted strut, proximal ground reaction cuff and distal supra-malleolar ankle-foot orthosis (AFO). Multiple components of the IDEO can be fine-tuned for a given wounded soldier's injury severity and type. Despite the anecdotal success of the IDEO in returning service members to combat duty, currently rehabilitation and structural research for future brace iterations supersede published quantitative gait measures, and as such this brace cannot be retrospectively compared to data from the DAFO of the current study.

The design and engineering phase of ankle-foot orthotic evolution has been extremely important in establishing choices for clinicians and patients to facilitate rehabilitation of a more normal gait. Just like any new product to the market, extensive testing and comparison is needed to guide these professionals in the best options for individual needs of drop foot patients. The next sections attempt to summarize the clinical findings of AFO designs to date, as well as highlight unanswered questions and AFO characteristics necessary for a military member with a drop foot condition.

Comparison of AFOs to Barefoot and Shod Conditions

Ankle-foot orthoses have been prescribed to treat patients with hemiparesis and weak ankle dorsiflexion for decades. A large number of studies have shown beneficial effects in terms of temporal-spatial parameters, but the methods to determine this effectiveness varies as greatly as the types of AFOs prescribed. Some studies use barefoot gait as the control condition within the patient with the altered gait pattern, while other studies use the gait of healthy controls for comparison of fluidity of gait patterns. Concerns exist that comparisons with donned AFOs in shoes should be made to shod conditions without the AFO, due to the relative contribution of footwear. Churchill et al. (2003) [26] conducted a pilot study to evaluate the effects of a custom molded plastic AFO and footwear on gait performance, predominately interested in the proportion of benefit associated footwear has with a donned AFO. Five stroke patients, ages 25-60, were instructed to walk along a straight line with white markers on the affected leg. A RIVCAM two-dimensional kinematic system recorded six gait trials of each condition (barefoot in socks, with footwear only, and with footwear and AFO), converted the footage to an image sequence to analyze for XY location of markers, and smoothed using a 12 Hz, 4^{th} order dual pass Butterworth filter. Spatio-temporal characteristics including stride length, cycle time, velocity, cadence, swing time, and swing velocity were analyzed using repeated measures ANOVA with two planned contrasts (barefoot versus with footwear, and footwear versus AFO and footwear) for each parameter. Only stride length ($p<0.001$) and swing velocity ($p<0.05$) were significantly different across conditions. Despite data limitations, the authors concluded that the appropriate baseline for assessing AFO effectiveness is to compare a patient's gait using an AFO against gait with footwear alone.

Supporting this finding of footwear and AFO research was Desloovere et al. (2006) [41], who found the effects of AFOs should always be weighed against the influence of shoes, as shoes significantly alter gait patterns compared to barefoot conditions. Contrasting two AFO designs against each other, the common posterior leaf-spring (PLS) AFO and the dual carbon fiber spring AFO (CFO), fifteen children diagnosed with hemiplegia (mean age 5.86 ± 1.76 years) were studied using three-dimensional gait analysis including kinetics and kinematics. The children adjusted to the PLS and CFO designs by alternating wear for three weeks prior to evaluation. Barefoot and shod without AFO conditions were used as controls, with PLS and CFO conditions randomized for each subject. Following collection of anthropometric data, subjects were assessed using an eight-camera VICON system collecting at 120 Hz, fitted with a lower PlugInGait marker set, at a self-selected velocity down a 10-meter walkway. Three embedded force plates registered force data, and surface EMG data were collected using a 16-channel K-Lab EMG system. Investigators further analyzed kinetic and kinematic data for three trials on the involved side, selecting 52 gait parameters for comparison against a group of healthy age-matched controls. Non-parametric statistical analysis performed on SAS ($p=0.01$) included Wilcoxon signed rank tests for comparisons between barefoot and shoes, PLS and CFO versus control conditions, and PLS versus CFO. Both AFOs improved first and second ankle rocker when compared to barefoot conditions, but ankle ROM during push-off, power generation in pre-swing, and ankle velocity at toe-off worsened. In comparison, the CFO deteriorated less in the above characteristics versus the PLS. In terms of the third rocker, the CFO produced a greater push-off, with significantly greater ankle ROM, angular velocity and power at pre-swing. This study supported previous findings that AFOs improve spatial-temporal gait parameters, including improved walking velocity with a decrease in cadence and an increased step length. Authors concluded that the CFO displayed more significant improvements during push-off at the ankle compared to the PLS.

Additional support for shod research presented in a case report for one post-stroke subject was especially helpful in compiling the case studies. Nolan et al. (2010) [73] utilized a single case design to evaluate the effects of a dynamic AFO on hemiplegia between no brace and the DAFO with shoes over level ground. The subject in this study was a post-stroke survivor, and was the same age and gender as the service member included in the present study who had suffered a stroke. Hemiplegia was also consistent between Nolan's subject and the service member, with partial nerve paralysis on the right side. Spatial-temporal and kinematic measures were collected by five trials for each condition along a five-meter runway for both DAFO and without DAFO conditions. A 7-camera Vicon system collected gait data at 60 Hz using a plug-in-gait marker set. Matlab was used for custom analysis of all data, which was normalized to 100% of the gait cycle from foot strike to ipsilateral foot strike. Independent samples *t*-tests were used to determine differences between the two conditions (however it should be noted that these two conditions were not independent). Velocity, cadence, step and stride lengths increased with use of the DAFO. Kinematically, there was an increase in dorsiflexion at initial contact without a significant change in toe position at push-off. The affected limb presented with a stiff knee gait regardless of condition, though knee ROM increased slightly during stance with use of the DAFO. Sagittal hip velocity increased on the involved side with the DAFO, and hip abduction decreased with use of the device. Nolan noted that parameters of gait are interdependent, and changes in one parameter will likely cascade into changes of others. The most notable changes in the DAFO were at the hip with an increase in angular velocity and decrease in abduction related to compensatory circumduction.

De Wit et al. (2000) [179] provided a comprehensive statistical data set to thoroughly describe the barefoot condition, which supports the argument that gait is altered with the use of shoes. Nine trained male distance runners (age 27.3 ± 9 years, running 30-40 km/week) free from injury were tested running in both barefoot and shod conditions at 3.5, 4.5, and 5.5 m/s. Kinetic measurements were taken by a two-dimensional approach using a Kistler force plate while two high-speed digital cameras recorded video of the entire body and a close view of the foot and shank. Infrared photocells ensure consistent running velocity was maintained, and five LED lights were used to time synchronize the cameras and force plate to 0.001 of a second. Following familiarization trials, runners continued attempts until ten successful contacts with the force plate were obtained on the right side without altering technique. Additional tests of local pressures were measured under the tuber calcaneum on seven of the runners using a Footscan system sampling at 200 Hz. Step length, frequency, ground reaction forces, rearfoot and sagittal plane kinematics were collected. Researchers analyzed the results using the general linear method, factorial model, with significance set at $p \leq 0.05$. Pearson product moment correlations and linear regression provided single correlations and best predicted values of one dependent variable, respectively. Barefoot runners were found to have a shorter contact time and take significantly smaller steps than in shod conditions. Barefoot runners also demonstrated a larger loading rate, as well as more than one impact peak. Local pressures underneath the heel were also found to be highest in the barefoot condition, driving a more horizontal foot placement, which seems to be predetermined well before initial contact of the foot for the intention of reducing the local pressures underneath the heel. This flatter foot placement coincides with larger plantar flexion and greater knee flexion which has been seen with more compliant surface conditions in leg stiffness studies, but this contradicts the finding that leg stiffness during stance phase was higher for barefoot running than the shod condition. A change to the barefoot running condition is mostly evident in external loading rate and flatter foot placement at initial contact, assumed that this placement will minimize pressure to the heel. For a drop foot patient fitted with an AFO to assist dorsiflexion, it seems most appropriate to compare this assessment to a shod condition.

Though the aforementioned studies present convincing arguments that shoes alter the gait pattern, many studies prior to this point compared AFOs to barefoot gait. Despite the difference in control conditions, much can still be learned from the barefoot comparisons in terms of kinetic and kinematic variations. Brunner et al. (1998) [39] expanded the contrast of AFO versus barefoot conditions to include a comparisons between a conventional stiff and a spring-type orthotic, and then compared both to a barefoot pattern. The same orthotist was used to mold a conventional stiff orthotic to each subject's foot, and then cut a gap to allow 10-15° of dorsiflexion, while blocking plantarflexion. This modification was designated as the spring-type orthotic. The gap was then bridged on the same orthotic by an aluminum bar to stiffen the orthotic for the second design. Fourteen subjects (eight male, 6 female, mean age 11.42) diagnosed with spastic hemiplegia walked barefoot and with each orthotic design at a self-selected pace. Kinetic analysis using a six-camera Vicon motion analysis was collected over six trials (six steps each) for each subject, and Kistler force plates with self-made software collected with a resolution of 400 Hz, and an oversampling rate of 2000 Hz. Kinetics parameters were analyzed on the basis of ground reaction forces due to a defective hard and software link between the Vicon and the force plates. The gait was also recorded front to back and each side for visual control. Multiple *t*-tests for paired samples were used to compare the AFO types, followed by AFO versus barefoot. Compared with barefoot conditions, both orthotic designs offered better stability in stance, measured by an increase in double-support time, and both restored a heel-to-toe gait. In comparison to the conventional stiff orthotic, the spring-type orthotic increased subject velocity, stride, and step-length. In general gait parameters and kinetics, the spring-type orthotic was superior to the conventional stiff orthotic with a faster and more symmetric gait, correcting the pathological gait closer to normal.

Symmetry in gait has also been studied, and though a somewhat controversial topic, has been found reproducible in kinematic angular displacement measures [180]. Esquenazi et al. (2009) [40] investigated the effect of an AFO on patients with hemiplegia to assess specific gait parameters and a reduction of asymmetry. Over a thousand records were reviewed and narrowed down to 42 patients (male and female, mean age: 54.5) whose files contained temporal spatial analysis data for both walking barefoot and with AFO conditions. An electronic 3.8-meter gait mat was used to assess these parameters at 100 Hz by 10,000 electronic switches, displaying an electronic foot print. Patients were asked to walk at a self-selected pace across this mat for ten steady-state steps in both conditions. Calculations of stance and step length asymmetry ratio were completed using an adapted formula, and then run against Wilcoxon signed rank tests to assess the difference between barefoot and AFO conditions. Spearman rank correlation coefficients were determined using SPSS. Velocity, cadence, percent stance, and double support significantly increased when wearing an AFO ($p = 0.0001$), as well as a reduction in the width of the base of support and percent swing. Both the stance phase asymmetry ratio and the step length asymmetry ratio significantly improved with the use of an AFO, and there was a positive correlation between velocity and cadence in both groups. Authors concluded the use of an AFO following hemiparetic stroke improves walking velocity and symmetry as well as other temporal spatial parameters.

Roll-over shape, or the effective geometry to which the ankle-foot complex conforms between initial contact (IC) and opposite IC, is related to the three ankle rockers in normal gait. The three ankle rockers, occurring at initial contact, mid-stance, and terminal stance, advance the body smoothly over the supporting foot as a pivot system[62]. Fatone et al. (2007) [33] investigated the effect of a custom, articulated AFO on roll-over shape (ROS) in adults with hemiplegia. An eight-camera motion analysis system (collecting at 120 Hz) with six force plates (recording at 960 Hz) embedded in a 10m walkway collected gait data from 13 hemiplegic post stroke patients (mean 51.5 years) and 12 controls. Patients were fitted with a Helen Hayes marker set, and walked in identical shoes with and without the custom AFO. Three successful trials were recorded for each foot, and EVa RealTime software determined 3-D positioning. A

Butterworth second-order bidirectional low-pass filter with a cut-off frequency of 6 Hz was used to smooth the raw data. Roll-over shape was determined by transforming the center of pressure data of the GRF into a shank-based coordinate system (using the lateral ankle marker, a virtual ankle joint center marker, and a virtual marker at knee center) which included movements in all planes to account for the variability in the gait of disabled persons. Non-parametric tests were used as some variables lacked homogeneity of variance and were not normally distributed. Mann-Whitney tests compared independent groups with a Bonferroni-adjusted significance level set at $\alpha<0.025$. Wilcoxon signed-rank tests compared dependent groups with $\alpha<0.05$. Normal walking velocity for hemiplegic patients was matched with very-slow walking velocities of controls for analysis. Results indicated the AFO improved ROS, moving the CoP posterior to the ankle joint compared to a more anterior position when not wearing the AFO. This was accomplished in the AFO by mechanically blocking plantar flexion. Despite a significant improvement in the CoP progression, a perturbation was still seen in hemiplegic patients during mid-stance, though less pronounced. The improved ROS in the AFO-donned hemiplegic patients was not significantly different than the ROS of control subjects, indicating the AFO assisted in a more uniform forward progression of the CoP compared to the subjects walking without an AFO. Literature suggests a normal ROS may also be associated with stability, which affects step width, a parameter that is abnormal in disabled patients.

Additional studies associated with the ankle rockers measures ankle moments and muscle activity affecting ankle movement. Miyazaki et al. (1997) [181] measured the dynamic change of active ankle moments (AAM) generated by musculature while walking with or without an experimental AFO with varying degrees of rigidity. The effect of rigidity in this study was focused in the dorsiflexion and plantarflexion directions, as well as the initial angle of the AFO during the AAM. Twenty post-stroke hemiparetic subjects, ages 38-76, voluntarily walked at a self-selected pace for 10-30 steps, barefoot and while wearing an experimental AFO. The AFO rigidity was altered by changing the diameter of the coil springs that were anterior to the AFO uprights. Researchers attached a capacitive floor reaction force transducer to the sole of the AFO to measure floor reaction moment (FRM), and the moment generated by the orthotic (OM) was measured by a potentiometer attached to the center of one of the joints to measure joint angle, and then calculating moment based on elongation of the springs. The AAM was calculated from the equation AAM = - (FRM +OM). One-way analysis of variance was used to detect significance of maximal AAM in the plantarflexion direction. The researchers found the AFO to assist weak dorsiflexors, but only played a small role in assisting plantarflexion, which was contrary to literature findings. The AAM was much larger than the OM in the direction of plantarflexion, and changed significantly in over half of subjects when adjusting rigidity and initial angle. Researchers also noted the maximal AAM measurement coincided with best combination of rigidity and initial angle based on subjective gait performance. They concluded that AFOs should not just be regarded as "mechanical substitutes for insufficient muscles forces," but that each AFO should be tailored to the entire dynamic system of the patient, and customized to best suit the gait needs of each condition.

Geboers et al. (2002) [27] conducted a randomized study of the effect of AFOs on dorsiflexion and antagonist muscle activity during walking. Twenty-nine patients (mean age 33.2) with recent unilateral peripheral paresis of the ankle dorsiflexors were randomly assigned to an AFO group and a non-AFO group. Patients and controls walked on a treadmill at a self-selected velocity. Following a two minute warm-up, electromyography (EMG) sampling at 1000 Hz recorded muscle activity for five trials at 20 seconds each. Activity was recorded from the tibialis anterior (TA) and extensor digitorum longus (EDL) because they are muscles innervated by the peroneal nerve, specifically used for dorsiflexion. The peroneus longus was tested due to the potential for it to compensate for drop foot through eversion of the leg. The soleus and lateral gastrocnemius were also tested to determine the use of antagonist muscle groups. Reproducibility tests of the EMG were conducted twice on 14 controls prior to the clinical study, and patients completed three trials with three-week intervals in between each trial. Mann-Whitney *U* and Wilcoxon tests tested between group and within group values. Patients wearing an AFO had a 7% decrease of TA muscle activity, and a 20% decrease in TA muscle activity in healthy patients wearing an AFO, but at different times in the gait cycle. When healthy patients donned an AFO, TA and EDL muscle activity decreased in the first 15% of the step cycle, but not overall. In paretic patients wearing the AFO, TA and EDL muscle activity decreased over the entire step cycle, but not specifically in the first 15%. This reduction in muscle activity did not accumulate over the six weeks, and researchers concluded that any decrease in muscle activity by paretic patients was made up for in the total walking time they were able to complete with a donned AFO.

Muscle activity and affected gait of a paretic lower limb while wearing a 1-bar rigid AFO was the focus of Hesse et al. (1999) [182]. Twenty-one hemiparetic patients (mean age 58.2) suffering from marked plantar flexion spasticity walked at a maximum velocity for 10 meters wearing overshoe slippers embedded with eight insole force sensors. Three trials were recorded, and ankle dorsiflexion, onset/duration of stance, swing, and double-support were measured by biaxial goniometers. The Force sensors (collecting at 100 Hz) measured vertical ground reaction force at heel-strike and toe-off. Surface EMG data on the tibialis anterior, medial head of gastrocnemius, vastis lateralis, and gluteus medius were recorded. Symmetry ratios were conducted on the stance, swing, and double-support phases. Trials were conducted over two conditions: barefoot, and with the Valens caliper 1-bar rigid AFO with a firm shoe, and self-selected velocity was held constant by the use of a metronome. Multivariate profile analysis was used to analyze data at $\alpha=0.05$, and univariate tests were used to follow-up significant multivariate results. No significant differences were found in gait velocity, stride length, or cadence between the two conditions. Use of the Valens caliper resulted in a significant increase in single-stance of the affected limb, and improved swing phase symmetry. Ankle excursion data revealed a significant increase in dorsiflexion and a significant decrease in plantarflexion with use of the caliper. Caution was noted with use if the caliper in terms of muscle activity due to a decrease in tibialis anterior by 48%, but an increase in vastis lateralis (35%) muscle activity was also observed. Researchers concluded the use of the Valens caliper 1-bar rigid AFO was characterized by a more dynamic and balanced gait.

Kinetics and kinematics can highlight changes in gait, but just as important are the concerns of the patient in terms of comfort and stability in an AFO. Tyson and Thornton (2001) [79] examined the effect of a hinged AFO on hemiplegic gait, with a major focus on patient opinion. Twenty-five patients over 18 years old who had suffered a stroke and subsequent unilateral hemiplegia were included in the study, as long as they were able to weight-bear on the weak leg. Patients were assessed using the Functional Ambulation Categories as they walked in ordinary shoes with and without the hinged AFO. Paper walkways set with a 5-m start-to-finish line marked inked footprints which were used to measure temporal-spatial gait characteristics. At least ten strides were collected for averaging stride and step length, and symmetry was calculated as a ratio of unaffected to affected leg step length. Patients were given a questionnaire regarding the hinged AFO focusing on key aspects of function, effect on their confidence, perceived safety, velocity and distance they were capable of walking, comfort and weight, ease of donning the device, and appearance. Data were analyzed using paired *t*-tests to compare parameters with and without the hinged AFO; Functional Ambulation Categories were analyzed using Wilcoxon signed ranks test, and the questionnaire was analyzed via descriptive statistics. Results revealed FAC scores improved using the hinged AFO, and gait parameters showing significant improvement included stride length of both affected and unaffected legs, cadence and velocity. Subject views of the hinged AFO were very positive in terms of function and comfort, although 24% felt the orthotic was too heavy. When asked to compare function to quality of movement, 96% stated they would rather walk faster with a limb than slower without a limp, even at a cosmetic cost.

The most important outcome in the De Wit et al. (2004) [32] examination of AFO walking ability of stroke patients was also the opinions of the patient. Judged clinically relevant, subject impressions indicated 65% of patients experienced less difficulty while wearing the AFO and 70% felt more confident. Twenty stroke patients (mean age 61.2

years) were assessed using walking velocity, timed up and go (TUG), and stair tests. Walking velocity was assessed on the modified Hoffer Functional Ambulation Scale. The ankle-foot orthoses were labeled as plastic non-articulating, with differences in the posterior steel and heel. Sequencing of AFO versus barefoot were randomized, and the average of three measurements per test were analyzed using *t*-tests with a significance value $p \leq 0.05$. Subjective perspectives of self-confidence and difficulty were also collected, and analyzed using Wilcoxon sighed ranks test. Results from the functional tests indicated an increase in walking ability in favor of the AFO, with statistical significance but not clinical significance (20 cm/s), found by all tests.

Research studying the effects of walking gait with and without an AFO created the foundation for comparisons between AFO designs. A focus on the ankle movements in terms of symmetry, stability, and function of muscle activity and related joint moments provides a focused attention to drop foot related to peroneal neuropathy, but may not include other important considerations up the kinetic chain. Some soldiers may have developed drop foot due to damage further up the kinetic chain, such as the L4 and L5 nerve roots that form the lumbosacral trunk. A thorough examination of motions up the kinetic chain may be equally important to determine which gait parameters need the most focus to assist in the rehabilitation of more normal gait.

Comparison between AFO Designs

After early research favored a donned AFO for gait improvement in comparison to barefoot conditions or shoes alone, more recent studies have focused on comparing one or more AFO design to another in search of the most optimal orthotic for their subject base. Despite difference in name, the AFOs correspond to a few categories: solid, hinged, posterior leaf spring, or dynamic carbon fiber. One of the earliest and longest running designs, the solid AFO, has been used for comparison in many studies. Radtka et al. (1997) [183] compared a solid AFO (SAFO) to a dynamic AFO (DAFO) against barefoot conditions on children with cerebral palsy (CP). Ten children (mean age = 6.5 years) with spastic CP (four subjects has hemiplegia, and six had diplegia) wore no orthosis for two weeks, SAFO for one month, another period of no AFO for an additional two weeks, and then a DAFO for a one-month period. The subjects with hemiplegia wore one orthotic on the affected side, and the subjects with diplegia wore two orthoses. Gait measures included electromyography (EMG) of lower-extremity muscle groups during stance phase, contact-closing footswitches for temporal-distance characteristics, and three dimensional motion analyses to determine joint motion at initial contact and mid-stance. EMG and footswitch data were collected with CODAS data-collection software while subjects walked at a self-selected pace down a 10-meter runway. At least two successful trials determined average walking velocity, cadence, and stride length. Twenty-one retroreflective markers were applied using the Helen Hayes Hospital marker set, and six video cameras recorded sagittal, coronal, and transverse motions at 60 frames per second. Two-way repeated-measures ANOVA with significance set at 0.02 for temporal-distance gait characteristics, 0.002 for joint motions, and 0.01 for timing examined the differences among subjects with clinical recommendations of each orthosis type. Authors concluded that although both orthotics functioned as restraints of excessive plantarflexion, no change in prolonged activity of the triceps surae muscle group were found, and no differences in temporal-distance gait characteristics, joint motions, and muscle timing between the two orthoses were shown. However various limitations to this study were cited: subjects were tested barefoot in no orthoses conditions but wore shoes when tested in the AFOs, and the small sample size and carryover effects between the AFO tests were noted. Eight years later, Radtka et al. (2005) [184] improved upon the 1997 study with six cameras compared with the original two, and the addition of two Kistler force plates. Twelve subjects, aged 4-10 (mean age = 7.5) were tested against the same protocol used in the 1997 study, but compared a solid AFO to a hinged AFO versus barefoot conditions. The same analysis was conducted on the trials, and authors concluded the hinged AFO produced more normal dorsiflexion at both mid-stance and terminal stance phases when compared with the solid AFO, as well as greater power generation at pre-swing phase for a greater contraction at push-off. All other temporal-distance gait parameter results determined the same as the previous study.

An additional study evaluating the solid AFO was completed by Bartonek et al. (2006) [29]. The subject's regularly used orthosis (RO – a solid AFO) and a new carbon-fiber design (designated "SO") were the independent variables included for comparison. The SO AFO consisted of a lightweight, L-shaped carbon fiber spring with a corresponding cast in 5° plantarflexion that is ordered based on patient ambulation and body weight. According to the manufacturer, the carbon fiber (SO) aims to enable progressive dorsiflexion, tibial stabilization, knee extending effect equivalent to the material properties of the forefoot, and active kinetic support at push-off. The testing environment consisted of a VICON six camera motion analysis system with two embedded Kistler force plates. Thirty-four reflective markers were placed on the subjects according to Newington model for lower extremity, and the Plug-In-Gait model was used for the upper-body. Seventeen children (mean age 11.3 years) with various motor disorders contributing to plantar flexor weakness completed three walking trials at a self-selected pace along a ten-meter walkway. Authors chose to report on three subjects with varying disorders. Ankle plantarflexion, ankle sagittal moment, and ankle power absorption and generation increased when subject wore the SO. Despite limitations of not including data from the other 14 subjects tested, authors preliminarily report that subjects with plantar flexor weakness benefit from a carbon spring orthosis. One year later, Bartonek et al. (2007) [37] followed-up on the previously summarized study with further statistical analysis. Carbon fiber spring orthoses (CFSO) were compared to the subjects' regular orthoses (RO) using a two-way repeated ANOVA with within-participant factors of both affected side and orthosis type. The entire group displayed increased ankle dorsiflexion and plantarflexion moments, as well as power at the ankle. Positive and negative joint work, stride length, and walking velocity also increased by the entire group.

The solid AFO is indicated when the patient has high spasticity in the plantarflexors or significant mediolateral stability [185]. The hinged AFO has mechanical ankle joints which can be incorporated to assist motion in a certain direction, making it more customizable to the needs of a patient. Rethlefsen et al. (1999) [186] compared the fixed and articulated AFOs with shoes alone by gait analysis in children with diplegic cerebral palsy. Following a six-week accommodation period with an individualized wear schedule of fixed (FAFO) and articulated (AAFO) orthotics, 21 children (age 9.1 ± 1.2 years) underwent kinematic and kinetic gait analysis testing using a 7-camera Vicon motion analysis system and three AMTI force plates. Three successful trials per side at a self-selected velocity were analyzed for ankle and knee position, ankle and knee kinetics at key points in the gait cycle, and the timing of peak calf muscle activity. Surface electrodes collected EMG data for five muscle groups: hip and knee extensors, knee flexors, ankle dorsiflexors and plantarflexors. A same-day clinical examination of range of motion manual muscle testing was completed, with spasticity rated using a modified Ashworth scale. Results of the gait analysis indicated both AFOs achieved greater dorsiflexion at initial contact when compared to shoes alone, and the AAFO provided greater dorsiflexion at terminal stance, much closer to normal values versus the other two conditions. This may allow the triceps surae to stretch with each step, and generate power at pre-swing which was adversely affected in the FAFO. Correlation analysis indicated that AAFOs are appropriate for subjects with varying degrees of calf spasticity granted their dorsiflexion ROM is at least present to neutral.

Neville and Houck (2009) [187] examined kinematics of flatfoot deformity in patients with stage II posterior tibial tendon dysfunction (PTTD) to select the most appropriate of three ankle-foot orthotic (AFO) designs. A case report of one patient, age 77, diagnosed with stage II PTTD was given three different fabricated AFOs: a custom solid AFO, a custom articulated AFO, and off-the-shelf AFO with neoprene sleeves and plastic clamshell supports. The three AFOs were modified to include a “window” for calcaneus examination during gait analysis. Windows were also added to the testing shoe for the heel marker and dorsal surface of the first metatarsal. The patient followed a wearing schedule for each AFO with included an average of 15 hours use for each design, and was asked to keep a journal of AFO use. Infrared diodes were attached to the skin through the shoe and AFO windows, and the patient walked down a 5-m walkway at 1.3 m/s, which was maintained within 5% using an infrared timing system. A custom heel counter was attached to the back of the testing shoe. Five successful trials of foot landing on a force plate (collecting at 1000 Hz) with diodes visible completed the procedure for each AFO design. Kinematic data were collected using a 3-segment foot model with tibia, calcaneus, and first metatarsal locations, and three IREDs were attached using double-sided tape. Two banks of infrared cameras sampling at 60 Hz and motion analysis software determined foot model rotation using the Euler rotation sequence. The greatest change in foot kinematics between the three designs was determined using an intraclass correlation coefficient (ICC) for standard error in the measurements. Results indicated the articulated AFO was the best fit for this patient because it provided the greatest correction of flatfoot kinematics when compared to a shoe (first, second, and third rocker in the areas of hind foot in- and eversion, forefoot plantar- and dorsiflexion, and forefoot abduction and adduction), and the articulated ankle allows for dorsi- and plantarflexion moments which may help prevent muscle weakness. In the case of this patient, the most important kinematics needing corrections were hind foot eversion, forefoot abduction, and medial longitudinal arch (MLA).

An additional comparison of conventional versus dynamic AFO comparison was completed by Lam et al. (2005) [99]. Thirteen children with spastic diplegic cerebral palsy (CP) were assessed barefoot, with a conventional AFO, and with a dynamic AFO and then compared to an age-match control group tested barefoot. A 6-camera Vicon system sampling at 60 Hz and a Helen Hayes marker set were used in conjunction with EMG surface electrodes. Patients walked at a self-selected pace along a 10-meter runway until a minimum of six trials with successful force plate strike were collected. Values were normalized for body weight, and ROM, forces, internal moments, and powers were reported. Both orthoses provided better foot positioning, though limited plantarflexion at push-off. An increase in hip flexion at initial contact was reported for the DAFO group without explanation. No significant differences existed for moments and powers between the conventional AFO and the dynamic AFO, although plantarflexion moment returned to normative values compared to barefoot of the CP group. Authors noted future studies on long-term effects between AFO types could be useful.

Romkes et al. (2002) [28] compared the effect of a hinged AFO (h-AFO) with a dynamic AFO (d-AFO) on gait of patients diagnosed with spastic hemiplegic cerebral palsy. The h-AFO design extended posterior just below the knee with a flat foot-plate extended to the tips of the toes. The design blocked ankle plantarflexion, but dorsiflexion was free through the hinge. The d-AFO was trimmed above the malleoli and included a foot-plate with medial longitudinal arch, peroneal notch, metatarsal arch and space underneath and between the toes. Three females and nine males (n=12), ages 11.9 ± 4.9 years who previously worn the h-AFO on the involved side were enrolled in the study. Each subject had a minimum of four weeks to adapt to each orthoses. A control group consisting of seven females and three males (n=10), ages 26.9 ± 6.3 years were established to obtain normal reference data during barefoot conditions. Gait was assessed using a three-dimensional, six-camera, VICON motion measurement system measuring at 50 Hz, and two Kistler force plates. Davis et al [188] marker protocol was used to configure subjects for a three-dimensional, bilateral analysis of gait. Six trials for each subject under each condition were collected during a self-selected velocity along a 10-meter walkway. Six trials were collected for each condition of barefoot, h-AFO, and d-AFO. Trials for each condition were averaged, and kinetic data was separated into 0-

30%, and 30-65% of the gait cycle for analysis. Repeated measures ANOVA tested the means of the conditions within the patient group with a significance set at *($p<0.05$)*. An improvement indicated the measures approached normal reference values. None of the patients held a heel-toe gait pattern in barefoot conditions, but all patients produced a heel-toe gait in the h-AFO. The d-AFO only produced a heel-toe gait in four patients. Both the d-AFO and the h-AFO significantly improved ankle plantarflexion at initial foot contact, but the h-AFO produced a more normal reference value. Both AFO's increased peak dorsiflexion to values close to the normal reference value. These values improved foot shank position throughout the gait cycle, as well as increased stride and step lengths. Overall, the authors concluded that the h-AFO was superior to the d-AFO in terms of gait correction when examining heel-to gait pattern, and ankle plantarflexion angle at initial foot contact.

A comparison of the same types of AFOs, a carbon fiber spring and a joint variety, yielded different results in a study by Alimusaj et al. [36]. Five subjects (ages 10-25) with bilateral Spina bifida and polio paralysis, as well as one subject (age 34) with temporary tibial nerve block were evaluated using a Vicon infrared camera system and two Kisler force plates to analyze kinetic and kinematic gait parameters. Subjects were fitted with the Helen Hayes marker set, and kinematics of the stance phase was determined by ROM and maximal plantar and dorsiflexion of three sub-phases: presence, roll, and push-off. Data analysis included student *t*-tests for paired samples with a significance level set at $p<0.05$. The carbon fiber spring demonstrated an increased stride length and walking velocity, but not to a level that was significant. Authors concluded the carbon fiber spring orthotic provided patients with calf paresis a significant advantage in the presence and kick phase, as well as a more normal physiological gait pattern and momentum when compared to the joint orthotic.

A pilot study supporting the carbon fiber spring orthotic was published Wolf et al. (2008) [189], who compared a carbon fiber spring orthotic (CFO) to a conventional hinge device (HJO) in terms of ankle kinematics, an initial look into kinetics of the CFO, and the effectiveness of the spring mechanism on patient gait. Five patients (mean age 19.1 years) with myelomeningocele were fitted with a modified Helen-Hayes marker set and analyzed using a nine-camera Vicon motion capture system. Two dorsal markers were attached to the vertical portion of the AFO spring and manually dorsiflexed on a Kistler force plate to measure ground reaction forces. The spring constant was then calculated according to Hook's Law, and moment and power of the carbon spring were then estimated from the calculation. These quantities were expressed as a percentage of the total moment and power of the orthotic. Five dynamic trials were collected for the three rockers in stance phase, and analyzed using the Wilcoxon signed rank test at a significance level of $p<0.05$. Results showed the CFO normalized kinematics in the first rocker, and enhanced range of motion. However, slightly excessive dorsiflexion was noted in the second rocker when compared with the HJO. Both devices exhibited a moderate ankle moment at push-off, with 62% of the total ankle power at push-off attributed to the CFO. Authors concluded the CFO materials offer a higher level of elasticity which may help compensate for gastrocnemius muscle weakness. Carbon fiber spring loading during mid-stance and loading response may contribute to forward progression of the tibia during in early mid-stance to assist patient movement.

The third ankle rocker was also the focus of an AFO comparison was accomplished by Van Gestel et al. (2008) [190]. Thirty-seven children (mean age 8.4 years), were compared against three differing AFO designs; Orteams®, Posterior Leaf Spring (PLS), and Dual Carbon Fiber Spring AFO (named CFO®), all common AFO's designed to correct ankle position and influence plantarflexion. All designs in this study combine a built-in plantarflexion stop with free dorsiflexion to allow for more range of motion. The children were separated into three groups based on Gage's classification of hemiplegic gait and age at the time of the study. Anthropometric measurements were taken for estimation of joint center, and lower limb PlugInGait marker set was attached. Subjects walked at self-selected velocity along a ten-meter walkway while kinematic measurements were collected at 120 Hz using an eight-camera Vicon motion analysis system. Forces were registered by three force plates embedded into the walkway, and electromyographic (EMG) data were collected on eight lower extremity muscle groups. Workstation and Polygon analyzed three trials of the involved side for kinetic and kinematic data. Authors concluded that despite all three orthoses increasing maximal ankle dorsiflexion in stance, only the CFO could significantly improve ankle dorsiflexion at loading response when compared with barefoot conditions. Compared with the other two orthoses, the CFO produced the greatest ankle ROM during push-off. The CFO also surpassed the PLS at push-off ankle plantarflexion moment with the closest to normal gait values. The CFO allowed for the most physiological third rocker, the most flexible and greatest ROM in the second rocker, while the PLS was the only orthosis that normalized the first ankle rocker; in that case the CFO had the smallest correction. Based on the combined results, authors seemed to favor the CFO design in terms of gait parameters.

A different perspective used a normal, healthy population to examine the effects of the rigid (solid) AFO and dynamic AFO on walking gait. Guillebastre et al. (2009) [35], tested eleven healthy subjects (median age 25.8) in five randomly executed experimental conditions on their dominant right side: walking without an orthosis (REF), walking with a rigid orthosis (R-AFO), and three conditions with a dynamic orthosis (D-AFO) in different stiffness configurations (30%, 70%, and 100% of maximum strain). Subjects walked at a self-selected pace over a 12m distance in which an electric carpet mat (8.3m) was centered. Pressure sensors were embedded into the carpet to collect spatiotemporal data. Two successful trials were collected in all five conditions, and a nonparametric one-way repeated measures ANOVA followed by Dunn's multiple comparison tests were completed. Wilcoxon signed-ranks tests were used to address asymmetry with a significance set at $p<0.05$. Results indicated a decrease in velocity while wearing the R-AFO, both only statistically significantly different than the D-AFO without elasticity. Of all conditions, only the R-AFO showed step time and length asymmetries, possibly accounted for by the prevention of dorsiflexion at push-off. The R-AFO caused a backward shift of the foot during stance, seemingly decreasing opposite step length, caused by a decreased plantarflexion capacity. The D-AFO only affected midline lengths of the foot, influenced by the stiffness of the elastic band. With a low level of stiffness (30% configuration), a more normal gait was seen. Authors concluded the R-AFO changes the gait of a normal individual, whereas closest to normal values were seen in the least stiff D-AFO.

Depending on the etiology of the peroneal neuropathy, additional pathologies proximal in the lower extremity may require other assistive aids for ambulation. Some patients using unilateral or bilateral AFOs may also use forearm crutches to counter the effects of musculoskeletal weakness and atrophy, as in the case study of the veteran with long-term outcomes following viral meningitis. Vankoski et al. (1997) [105] examined the influence of forearm crutches on pelvic and hip kinematics on 16 myelomeningocele patients (mean age nine years, nine months). Manual muscle tests were used to assess muscle strength and joint range of motion, followed by gait testing with forearm crutches using a four-point reciprocal crutch-walking gait pattern. A 3D Vicon motion capture system was used to assess a minimum of three trials in crutch and no-crutch conditions, and the kinematic data were compared to normal pediatric kinematics collected in the same laboratory. Stride length and cadence were significantly different with and without forearm crutches, although gait velocity did not differ statistically. Patient gait compensations were primarily due to musculoskeletal weaknesses about the hip. Increased anterior pelvic tilt was evident when bearing weight on the forearm crutches, although not significantly different than forearm crutch use. The authors concluded pelvic and hip kinematics improved with the use of forearm crutches and were highly beneficial for patients with high sacral-level pathologies, despite the limitations crutches put on the upper extremities. The use of crutches may improve ambulatory status and preserve joint integrity, but may not have been needed in some patients once musculoskeletal strength was improved.

The evolution of the AFO has demonstrated dynamic movement can be paired with stability and control. The materials and technologies available today, when combined in the right configuration, can take ambulation to the next level. The carbon fiber spring orthotic has demonstrated the ability to improve ankle dorsiflexion throughout the three rocker phases, during loading response, mid-stance, and at push-off. The next step in testing the performance of a dynamic AFO is with a premorbidly fit military population in vigorous activities surpassing functional walking ambulation.

Ankle-Foot Orthotic Running Studies

Very few studies have attempted running gait with a donned AFO. Owens et al. (2011) [10] reported the different types of athletic endeavors limb-salvage patients were able to complete after a minimum 12-week rehabilitation program in a dynamic ankle-foot orthotic, defined as the Intrepid Dynamic Exoskeletal Orthotic (IDEO). Ten patients were rehabilitated to their highest functional level using a sports-medicine approach to rehabilitation, centered on a strength training program for normal force-generating capacity required for running gait. Strength training exercises such as the box squat, bridge, and wall sit were completed in three sets of 10 repetitions prior to advancing to an applied external weight. Plyometric exercises were added once the patients could perform a leg press or squat with at least 80% body weight, consisting of exercises such as shuttle jump, box hop, vertical jump, etc. Running gait re-training was the next progression in the rehab, with a mid-foot strike emphasis to maximize energy return from the IDEO. Running progression advanced in quarter-mile-per-week increments until patients could run on a treadmill without stopping for two miles. Pre-deployment training for soldiers preparing for combat consisted of running and walking program with a rucksack progressively increasing in weigh up to 80 pounds. Eight of out the ten soldiers successfully completed this program, with one completing marathon distance running. Other sports returned to were triathlon, cycling, basketball, softball, golf, and weight-lifting. Finally, three of the ten patients deployed with Special Forces and Army Ranger units.

Bishop et al. (2009) [23] developed a new ankle-foot orthotic with a design more compatible in terms of shoe fit, pressure, and ambient temperature. Major goals for this AFO included no cramping or blisters to the heel, comfort even when worn for a long period of time, and streamlined enough to fit into several types of shoes. An articulating rotator in-line with the ankle joint was determined as the best design to alleviate discomfort caused by rubbing of the foot skin against the orthosis. The dorsiflexion moment needed to maintain 90 degrees of dorsiflexion was determined using a goniometer and assisted force, and calculated as 3.1 Nm. The hard heel cup was eliminated from a traditional AFO design and the foot was supported from the top to allow for circumferential tightening of the shoe to hold the AFO in place. A semi-flexible wrap was used in place of the anterior tibial shell to decrease discomfort caused by the difference in height between the toe and the footpad of the original design. Struts down the sides of the leg as opposed to front and back eliminated rubbing through the range of motion, and a soft 4 mm layer covering the entire inside of the brace eliminated points of pressure of the previous design. The AFO was worn by one subject (age: 25 years, height: 188 cm, weight: 190 pounds) 8-10 hours per day, almost every day for one year. The subject continued regular activity such as walking, biking, and jogging. The redesigned AFO properly fit into an athletic shoe, and was determined to successfully support the foot at 90° during the swing phase of gait. Successful foot clearance was obtained, and no blisters or foot cramps were reported during the study period. Authors concluded the new design eliminated many of the shortcomings of traditional AFOs, although the space required for the AFO did not accommodate all types of shoes. Future studies with this design should test more than one subject, and kinematic /kinetic analysis should be included.

Ankle-foot orthotic functionality of walking gait has been established, but gaps in the literature suggest the need for a standardized set of gait parameters deemed necessary for a specific population, as well as the need to determine functional clinical outcomes of all AFO designs. To sufficiently determine the most optimal AFO needed to return active duty service members back to duty status and physically active lifestyles, clinical outcomes of a running gait in the AFO must also be determined.

Cost Analysis and Cost-Utility Analysis

The survival rate of soldiers in the current GWOT conflict is higher than previous wars, yet similarly those returning home with devastating injuries at a young age are faced with an uncertain future in addition to an arduous recovery. Steps have been taken within the military chain of command to change the culture regarding the capabilities of a wounded veteran. Centralized centers for amputee care combined with a high level of motivation have helped these men and women rehabilitate to the point of returning to duty, and even redeployment. The cost associated with medically retiring service members at a young age versus returning to active duty with adequate ambulation devices has not been well documented, and is needed to assess this difference. Costs of associated traumas such as post-traumatic stress disorder (PTSD) were originally considered in the analysis, but determined to inconclusive to include. A disparity within the literature existed on the true percentage of wounded service members who have documented PTSD. Litz and Schlenger (2009) reported in a PTSD Research Quarterly report on the extensive mental impact of ongoing conflicts [124]. A summary of researchers suggested that 10-18% of service members returning from recent conflicts have “probable” PTSD following deployments, however noted the urgent need for empirical studies to assess self-report statistics and the trajectories to response to war stressors. Current research has been mostly cross-sectional, full of causal inference and lacking prevention strategies.

Due to a lack of support network that one might find after medical retirement, some soldiers returned home to rural areas often hours from a VA facility where communication between patients and advocates became lost [191]. Documents have been published to assist soldiers in determining medical benefits. The Department of Defense (DoD) Wounded, Ill, and Injured Compensation and Benefits Handbook thoroughly explained the different types of compensation available to veterans, how to apply for these compensations, and the mathematic derivatives of their origin [135, 192]. Pay and compensation entitled to a wounded soldier was outlined from deployed location through the soldier’s path of retirement or return to duty. The Veteran’s Affairs disability compensation tables were referenced for every familial combination, and included benefits for the VA’s medical package. This handbook provided a preliminary glimpse at the list of compensation needed to determine the cost of medically retiring or discharging a wounded warrior, and well as variations based on time in service, rank, and severity of disability.

Medical retirement or return to active service is mostly based on a determination of “fitness” for duty, and the associated variations outlined above in the handbook. The return rate of wounded soldiers also varies, but trends were recently reported by Stinner et al. (2010) [20] and Cross et al. (2012) [138]. Current military regulations consider a major limb amputation as "unfit for military service", but motivated soldiers with the support of medical professionals have been known to demonstrate high levels of function with their device and return to duty. Higher military rank was positively related to return to duty, as they were closer to retirement and job profiles changed to more managerial roles and less physically demanding duties. The overall return to duty rate of motivated service members for isolated open fractures were 22%, salvaged extremities were 20.5%, and amputees were 12.5%. Level of motivation for return to work was echoed by Arnstein et al. (1999) [141], who added that self-efficacy was most important in patients with chronic pain. Motivation and an internal belief that one can cope with pain and remain functional were critical, as self-efficacy "mediated the relationship between pain intensity and related disability."

Motivation and performance of amputees returning to duty was also noted in a study by Gambel et al. (2009) [19], who provided an overview of the Army Physical Disability Evaluation System, factors involved in the return to duty, military retirement and disability compensation outlined by Veterans Affairs (VA). Return to duty after a major limb loss involved additional factors to the aforementioned, and included full support by their family and the military unit and service command they hoped to return to, and the possession of a valued specific military skill set. These soldiers also needed to be in good standing prior to their injuries, as outlined in their service records, and it was wise for the member to be flexible if returning to duty meant retraining into a different Military Occupational Security (MOS) code. During the year following injury, a Medical

Evaluation Board (MEB) determined retention capability, regardless of MOS, as outlined in Army Regulation 40-501, Chapter 3. If deemed unfit for retention, the MEB recommended a Physical Evaluation Board (PEB) to determine "fitness" based on the soldier's MOS. The soldier could also appeal the decision of the MEB prior to this point, for either fitness or unfitness by an unbiased physician. If the soldier was found unfit by the PEB, a disability rating was generated by the PEB using the Veterans Affairs Schedule for Rating Disabilities (VASRD). A rating of at least 30% approved either permanent (PDR) disability or a temporary disability retirement list (TDRL), whereas anything less than 30% included a military separation with lump sum compensation entitlement. Under the pilot system, a member must restart the medical evaluation process with a new set of clinical examinations in order to also receive VA disability compensation, but this will no longer be necessary by 2014. The soldier may also be eligible for concurrent receipt (CR) in which they receive both military retirement and disability compensation, or combat-related special compensation (CRSC). Should a soldier meet retention standards during the MEB, he or she will be referred to a MOS/Medical Retention Board (MMRB). This board then has the option of returning the soldier to their unit, placing them on a medical probationary status, reclassifying their MOS, or referring them to an MEB if determined necessary.

Military members retired for medical reasons draw pay that is based on a combination of disability severity, years of military service, and military retired pay if the veteran was on active-duty for twenty years or more. The DoD evaluates compensation based on conditions that make military members unable to perform military duties, and payment is intended to offset the disruption of their career. Depending upon time in service (i.e. if time on active duty was deemed short), the member may be given a lump sum VA severance instead of a military retirement. Levels of VA disability compensation depend on disability severity and the performance of tasks that directly affect civilian earnings. Buddin and Kapur (2005) [121] compared the differences in definitions between Department of Defense (DoD) and Veteran Affairs (VA) levels of disability compensation. Both systems drew from 1940's medical and legal decisions on ability to perform in the workplace, which may not accurately assess the abilities in the workplace today. Earning losses were not differentiated between the potential of a high-ranking officer and a low-ranking enlisted member, and allowances were not made for pain and suffering unless it directly affected civilian earnings. The labor force for nondisabled and disable retirees was 84% and 74%, respectively. In 2005, 2.5 million veterans received benefits from the VA, through a budget of 200 billion dollars. Average annual compensation for a veteran received 100% disability was $30,338, and adjusted up slightly based on marital and dependent status. Unlike civilian workers' compensation which had weekly limits and length of eligibility, there was no dollar or time limit for military benefit eligibility. The researcher concluded the DoD and VA need to investigate the current rating system to determine if disabilities affecting civilian wages were still accurate after 60 years.

Linda J. Bilmes, a Harvard University professor and writer, reported similar compensations to Buddin and Kapur in her summary of projected costs for Iraq and Afghanistan war veterans in 2007; and updated in an editorial in 2011[123, 193]. Estimates were calculated based on budgetary costs already incurred between 2001 and 2011, long-term incurred costs for soldiers who served in these conflicts that have not yet been paid, total long-term estimates for veterans from the federal government including medical care and compensation, and economic costs for veterans in the social realm. Cost projections were based on two factors associated with wounded soldiers: the medical costs of care over their average life span, and the cash compensation and other benefits eligible to veterans and their families. Annual benefits based on the 10% increment VA disability system ranged from $1474 per year (10% disabled) to approximately $35,366 per year (100% disabled) (these numbers have been updated from the FY11 VA Annual Benefits Report) [129].

The VA also considers changes to the home and car for disabled living, as well as maintenance of assistive devices is an additional compensation. Lifetime projected health-care costs between amputation and limb salvage treatment methods was measured by MacKenzie et al. (2007) [13], who found costs five-times greater for amputation versus reconstruction mostly due to maintenance and replacement of prosthetics, and alterations to the home, vehicle, and workplace. Researchers examined 545 patient files of subjects enrolled in the Lower Extremity Assessment Project (LEAP) study. A majority of these patients were male, ages 20-45 with lower-extremity injuries associated with motor-vehicle, motorcycle, or pedestrian-vehicular accidents. Within 84 months of the injury, 203 of the 545 underwent amputation, and the remaining 342 had limb reconstruction surgery. Estimations of two-year health-care costs included the initial hospitalization, all re-hospitalizations related to the limb injury, in-patient rehabilitation, out-patient doctor visits, out-patient physical and occupational therapy, and prosthetic devices and related services. Multivariate regression analysis techniques revealed non-significant differences between two-year health-care costs of the initial, post-acute, and re-hospitalization totals. When prosthetic device costs were included, the difference became greater ($81,316 for reconstruction and $91,106 for amputation). Authors concluded the addition costs associated with daily living as an amputee merits the efforts to improve successful limb reconstruction procedures.

Downs (2008) [148] published an article in the U.S. Naval Institute magazine on past, present, and future prosthetics in the V.A., including the history behind the support of how present day prosthetics came to be standard issue through the voice of service members and their families, up through senate and congressional hearings, and even an overhaul of the V.A. system by President Eisenhower and his theater surgeon. Based on the voice of the nation, cost is not a factor in these issued items. Blough (2010) [111] provided prosthetic cost projections for service members with limb loss from Vietnam and current conflicts, separated into 5-year, 10-year, 20-year, and lifetime cost increments based on eight Marcov models. Five-year projected unilateral lower limb prosthetic costs for service members returning from recent conflicts averaged $228,665. Not only did these figures provide the VA a funding estimate for future technologies, they also served as a basis to compare the ankle-foot orthotic for increasing numbers of limb salvage patients. Budget figures for the VA have been monitored by the Government Accounting Office (GAO), an investigative arm of Congress developed to audit and evaluate funded programs to ensure accountability of the government for the American public. The GAO recently complete a study to highlight how the VA's spending for prosthetics compared to budget estimates from 2005 to 2009, how the VA is monitored for their performance in supplying these devices to wounded service members, and efforts to improve based on performance feedback [149]. These reports ensured wounded service members continue to receive the health care and rehabilitation they deserve.

A cost-utility analysis of the ankle-foot orthotic was also warranted due to the difference in cost between the issued AFO and the dynamic AFO, and the incremental quality-of-life improvement that has been anecdotally justified. An example of cost-utility methods for a similar population to limb-salvage patients was performed by Gerzeli et al. (2008) [194], who assessed an electronic knee prosthesis (C-leg) compared to mechanical alternatives in trans-femoral amputees. This study focused on an economic comparison between two prosthetic devices, but included a QALY score from responses to the European Quality-of-life (EuroQol) questionnaire for each group. Economic data related to client demographics was procured retrospectively over 12 months from the Italian Workers' Compensation Authority (INAIL), and costs related to the prosthetic device were totaled using a phone questionnaire including healthcare received outside of INAIL, transportation costs, informal caregiving provided to patients, and patients' loss of productivity from the amputation. Cost analyses were performed from two perspectives: strictly costs associated with the device, and a cost-utility ratio based on QALY scores from the questionnaires of both groups. QALY measurements revealed an incremental effectiveness increase of 9% in the C-leg over a period of one year, based on a 0.09 point difference in scores. The two groups appeared similar if non-healthcare costs and productivity losses were not taken into account, but from a utility perspective, the C-leg had an incremental cost-utility ratio of €35,971 per QALY gained. Ratio benchmarks for Europe and North America range from $30,000 - $100,000 per

QALY, in which case the C-leg technology would be considered cost effective for the Italian amputee population.

Several texts have been written on the proper methods for cost analysis with subsets of utility, effectiveness, benefits, and minimization. These subsets had degrees of overlap, especially between cost-utility and cost-effectiveness, and therefore the refining of methodology was used interchangeably between these two subsets for this study. Petitti (2000) provided a text on meta-analysis, decision analysis, and cost-effectiveness analysis that outlined all necessary steps for conducting a successful cost analysis with utility inclusion [147]. After identifying all costs associated with different intervention and alternative, and calculating the difference in QALYs for each health state of the intervention and alternative, the last step in the analysis was to calculate the difference in cost between the intervention and alternative, and divide by the difference in QALYs, yielding an incremental cost of the intervention per QALY gained or lost. This yield was then compared to benchmark threshold for what U.S. or military healthcare considers "cost effective." The World Health Organization developed a program on choosing interventions that are cost-effective (WHO-CHOICE) [146] and cited the U.S. cost-effective range is between \$39,950 - \$119,849, with anything less than \$39,950 considered "very cost-effective", within the range as "cost-effective", and above \$119,849 as "not cost-effective." The lower end of the range was based on Gross National Product (GNP) per capita, with the upper range as three times the GNP per capita [146]. Drummond et al. (1987) [44] produced another excellent text on different types of cost analysis, but specifically from the economic evaluation of healthcare programs. A checklist for assessing economic evaluations was outlined, which captured questions regarding validity and reliability of economic data collected, as well as the proper adjustments, incremental and sensitivity analyses, and insight on how to present results.

There is a vast body of knowledge reporting the compensation benefits for wounded soldiers. Costs included for the analysis in this study required a military financial viewpoint, and costs associated with both pathways of medical retirement and return to duty were omitted. All numbers included in the cost analysis and cost-utility analysis were based on the most up-to-date costs from DoD websites [122, 130, 131, 136, 192]. Retention rates of higher ranking military personnel warrant an analysis for both a junior enlisted and a junior officer perspective, since the average age of a wounded soldier was 27.0 (range, 18-52 years) and the median military rank was enlisted grade E-4. Therefore, the cost analyses were based on enlisted E-4 and officer O-3 ranks, and included unique costs associated with medical retirement and return to duty over a 20-year period.

Data Analysis

An intraclass correlation (ICC) was completed to determine the intra-tester reliability of strength scores by use of a hand-held dynamometer (HHD) in the injured service member group. This importance was highlighted in a previously summarized article by Kimura et al. (1996) [164] who also referenced an article summarized by Hosking et al. (1976) [165] regarding higher reliability of strength measures tested in pathological populations. Insight into the different ICC's utilized in the literature was important when determining HHD reliability. Shrout and Fleiss (1979) [195] reported on six different ICC's that provided different results when applied to the same data, discussed with regard to several testers however applicable to one-facet studies. Three questions were proposed to assist researchers in choosing the correct ICC: is an ANOVA appropriate for the reliability study, are differences between judges' mean ratings relevant to the reliability, and is the unit of analysis an individual rating or a mean rating of several observations? The first analysis of one-way ANOVA yielded a between-subject mean (BM) and a within-subject mean (WM). The expected means square from this analysis allowed an unbiased estimate of target variance by subtracting the within from the between and dividing by the number of testers, and allowed an ICC to be written as: ICC = (BM – WM) / (BM (k+1) WM), where k is the number of judges. Two-way ANOVA partitioned the sum of squares to a between, within, and residual. Differences in two-way fixed effects versus two-way random effects are found in potential bias, and may result in different ICC values. The results of this study provided insight into different ICC's for multiple testers, but did not discuss the decision process for ICC of a single tester.

Non-parametric statistical tests were required in the present study due to the small sample size and subsequent non-normative data set. These tests worked on the principle of ranking data, with high scores represented by high ranks, and low scores represented by low ranks. The Field [196] and Polit [197]texts were used extensively to understand the different tests necessary for each research question, the requirements for selecting the appropriate test, and insight into interpretation procedures and data inclusion when reporting results.

Field (2009) [196] described the Mann Whitney *U* test as the non-parametric equivalent of the independent-samples t-test, and used to compare two conditions when different groups took part in each condition. This test was appropriate for comparing each of the injured service member conditions (No Brace, TAFO, DAFO Time 0, and DAFO Time 6) with the service member control group. The "exact" test option was also utilized due to the small sample size, which calculated the significance of the Mann-Whitney *U* test exactly, versus the asymptomatic method which was used for larger samples. Medians and quartiles were displayed for each output due to the inclusion of the median and ranges in each results table. The data output summarized the data after they were ranked, including the average and total ranks for each condition. Results included the *U* statistic and corresponding *Z* score, where

$U = n_1 n_2 + (N_1 (N_1+1)/2) - R_1$, with which $n_1 n_2$ represented the sample sizes of the two groups, and *R* was the sum of ranks for group 1. The *Z*-score was the value of an observation expressed in standard deviation units, which when converted created a new distribution that had a mean of zero and a standard deviation of one. This score was also used to calculate the effect size for inclusion in the results tables. Although some researchers think some information regarding the magnitude of scores is lost when ranking data, the power of the parametric test is only genuine if the assumptions of the parametric test were met. [196] Wilcoxon Signed-Ranks test and Friedman's tests were used to answer research questions regarding the same group of injured service members over different conditions. Wilcoxon also ranks the scores but then assigns a plus or minus sign to the rank to indicate an increase or decrease in score over conditions or time. This test was also used as follow-up testing to Friedman's test if a significant main effect for time was discovered, but involved a Bonferroni correction when multiple post hoc tests were used. Output from the Friedman's test included a Chi-Square statistic as opposed to a frequency distribution. This test was used to compare several conditions or effect of time and was related to the repeated measures one-way ANOVA. Polit (2010) [197] provided a decision matrix for these non-parametric tests, further insight into Mann Whitney *U*, Wilcoxon Signed-Ranks and Friedman's Tests, and research applications as well as the presentation of these results in research. The non-parametric tests were useful to answer inferential research questions, test biases, and select variables for follow-on multivariate statistics.

Based on the current literature associated with this population, this study evaluated the service members with shoes, both with and without DAFO as well as a comparison with their traditional issued AFO, as indications have been made to the relative contribution of footwear to gait and energy cost. Kinematic focus included sagittal and frontal displacements at the hip, knee, and ankle, kinetic considerations including forces, moments, and levels of stiffness based on reliability and validity measures discovered in analysis of normal gait. Spatial-temporal gait parameters have been consistent shown to improve with the use of an AFO, and therefore will be quantified and compared between the tradition and dynamic AFOs. All biomechanical variables were compared to a healthy age and anthropometrically-matched service member population. Considerations will also be made during analysis of the level of trauma sustained by each subject, which may display variations in the results between conditions. Finally, quality-of-life and cost-utility considerations were determined both between the two braces and over time. All variables summarized in this review are important determinants for the proper AFO device to return a service member to active duty.

APPENDIX I

INFORMED CONSENT and DATA COLLECTION FORMS

Data Collection Forms Index

1. Informed Consent

2. EMG/NCS Data Collection Sheet
   -To be used prior to DAFO issue and six months after brace issue

3. Anthropometric Data Collection Sheet
   -To be used prior to DAFO issue and six months after brace issue

4. Traditional AFO Use Questionnaire
   -To be used prior to DAFO issue, at the first metabolic and biomechanics data collection session

5. Metabolic Data Collection Sheet 1
   -To be used at the first metabolic data collection session

6. Metabolic Data Collection Sheet 2
   -To be used at the first metabolic data collection session after DAFO issue, and at the one, two, four, and five month follow-up data collection sessions

7. Metabolic data Collection Sheet 3
   -To be used at the three and six month follow-up data collection sessions

8. Borg's Ratings of Perceived Exertion (RPE)

9. List and Photograph of Reflective Marker Placement

10. Biomechanics Data Collection Sheet 1
    -To be used at the first biomechanics data collection session

11. Biomechanics Data Collection Sheet 2
    -To be used at the first biomechanics data collection session after DAFO issue, and at the one, two, four, and five month follow-up data collection sessions

12. Biomechanics data Collection Sheet 3
    -To be used at the three and six month follow-up data collection sessions

Revised 22 Nov 11 *Revised 09 Feb 12*

VOLUNTEER AGREEMENT AFFIDAVIT

For use of this form, see AR 70-25 or AR 40-38, the proponent agency is OTSG

PRIVACY ACT OF 1974

Authority: 10 USC 3013, 44 USC 3101, and 10 USC 1071-1087.

Principle Purpose: To document voluntary participation in the Clinical Investigation and Research Program. Home address will be used for identification and locating purposes.

Routine Uses: The home address will be used for identification and locating purposes. Information derived from the study will be used to document the study; implementation of medical programs; adjudication of claims; and for the mandatory reporting of medical conditions as required by law. Information may be furnished to Federal, State and local agencies.

Disclosure: The furnishing of your home address is mandatory and necessary to provide identification and to contact you if future information indicates that your health may be adversely affected. Failure to provide the information may preclude your voluntary participation in this investigational study.

PART A(1) - VOLUNTEER AFFIDAVIT

Volunteer Subjects in Approved Department of the Army Research Studies

Volunteers under the provisions of AR 40-38 and AR 70-25 are authorized all necessary medical care for injury or disease which is the proximate result of their participation in such studies.

I, ______________________________,

having full capacity to consent and having attained my ______________ birthday, do hereby volunteer/give consent as legal representative for ______________________________ to participate in an investigational study entitled Functional Analysis of Dynamic Ankle-Foot Orthoses to Improve Outcomes in Partial Lower Extremity Paralysis under the direction of Dr. Gerard M. Antoine conducted at the Tripler AMC Physical Medicine and Rehabilitation Services Department and the University of Hawaii Human Performance and Biomechanics Lab.

The implications of my voluntary participation/consent as legal representative; duration and purpose of the research study; the methods and means by which it is to be conducted; and the inconveniences and hazards that may reasonably be expected have been explained to me by Richard H. Todd, MPT, DSc, or David P. Newman, PT, DPT, OCS

I have been given an opportunity to ask questions concerning this investigational study. Any such questions were answered to my full and complete satisfaction. Should any further questions arise concerning my rights/the rights of the person I represent on study-related injury, I may contact

the Center Judge Advocate

at Tripler Army Medical Center, Tripler AMC, HI 96859-5000 (808) 433-5311

(Name, Address and Phone Number of Hospital (Include Area Code))

I understand that I may at any time during the course of this study revoke my consent and withdraw/have the person I represent withdrawn from the study without further penalty or loss of benefits; however, I/the person I represent may be required (military volunteer) or requested (civilian volunteer) to undergo certain examinations if, in the opinion of the attending physician, such examinations are necessary for my/the person I represent's health and well-being. My/the person I represent's refusal to participate will involve no penalty or loss of benefits to which I am/the person I represent is otherwise entitled.

PART A(2) - ASSENT VOLUNTEER AFFIDAVIT *(MINOR CHILD)*

I, ______________________________,

having full capacity to assent and having attained my ______________ birthday, do hereby volunteer for ______________________________ an investigational study entitled ______________________________

____

under the direction of ______________________________

conducted at ______________________________

(Name of Institution)

NOT APPLICABLE

(Continue on Reverse)

DA FORM 5303-R, MAY 89 PREVIOUS EDITIONS ARE OBSOLETE

TAMC HUC/IRB

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jun 12 for TAMC # 31H10.
This version of the consent form expires on 26 Jun 13.

PART A(2) - ASSENT VOLUNTEER AFFIDAVIT *(MINOR CHILD) (Cont'd.)*

The implications of my voluntary participation; the nature, duration and purpose of the research study; the methods and means by which it is to be conducted; and the inconveniences and hazards that may reasonably be expected have been explained to me by

____

I have been given an opportunity to ask questions concerning this [illegible] questions were answered to my full and complete satisfaction. Should any further questions [illegible] may contact

____

at ____

(Name, Address, and Phone Number of Hospital (Include Area Code))

NOT APPLICABLE

I understand that I may at any time during the course of this study revoke my assent and withdraw from the study without further penalty or loss of benefits; however, I may be requested to undergo certain examinations if, in the opinion of the attending physician, such examinations are necessary for my health and well-being. My refusal to participate will involve no penalty or loss of benefits to which I am otherwise entitled.

PART B - TO BE COMPLETED BY INVESTIGATOR

PARTICIPATION INFORMATION: You have been invited to participate in a clinical investigational/research study conducted at Tripler Army Medical Center and the University of Hawaii's Human Performance and Biomechanics Lab. It is very important that you read and understand the following general principles that apply to all participants in our studies: (a) your participation is entirely voluntary; (b) you may withdraw from participation in this study or any part of the study at any time; refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled; (c) after you read the explanation, please feel free to ask any questions that will allow you to clearly understand the nature of the study.

NATURE OF STUDY: The purpose of this study is to measure the changes in level of function and quality of life of soldiers with partial lower extremity paralysis after the issue of a custom dymanic ankle foot orthosis (AFO). The device being tested is commercially available, however no study has examined its use with a physically active population.

EXPECTED DURATION OF SUBJECT'S PARTICIPATION: This study will last approximately seven months from the date of enrollment. This includes one month before being fitted for the custom AFO, and then six months of follow-up after being issued the brace. Participants will be expected to follow-up monthly after brace fitting for a 1.5 hour gait and metabolic testing session at the University of Hawaii, and to follow their regular physical therapy and medical appointments as indicated by their physician.

WHAT WILL BE DONE: After study enrollment, your physician will perform an electromyographic (EMG) and nerve conductive study (NCS) as part of the standard follow-up for your condition. At this time you will be scheduled for an appointment with the orthotist who will take measurements and a cast of your leg in order to fabricate the custom dynamic AFO. You will also be scheduled with an appointment for evaluation by a physical therapist who will measure the range of motion in the joints, strength of the muscles, and muscle mass in your lower extremity.

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jun 12 for TAMC # 31H10
This version of the consent form expires on 26 Jun 13

Volunteer Agreement Affidavit

You will also be asked to report to the Human Performance and Biomechanics Lab at the University of Hawaii for metabolic testing and gait analysis. You will be asked to wear your physical training uniform and your normal running shoes. First, you will fill out a brief survey about your current AFO and your current quality of life, and then your height and weight will be measured by a member of the research team who is a certified athletic trainer. Next, you will be given 10 minutes for a self-selected warm-up (i.e. riding a stationary bike or stretching). Before beginning the treadmill metabolic test, you will be asked to select a comfortable walking speed on the treadmill and instructed that you will need to maintain this speed for eight minutes (the duration of the test). After selecting your speed, the treadmill will be stopped and you will be fitted with a heart rate monitor, headgear and breathing mask. Throughout the duration of the test you will breathe through the mask that is connected via ventilation tubes to a metabolic cart that will determine your oxygen consumption and energy expenditure during the walking test. At the time of test initiation the treadmill will be started and the speed of the treadmill will be set to your previously determined comfortable walking speed. You will be instructed to maintain this speed for eight minutes. Upon completion of the test, the treadmill speed will be immediately decreased to allow you to walk at an easy pace to cool down, the headgear and breathing mask will be removed at this time as well. You will complete this procedure without an AFO and wearing your current AFO with a 10 minute rest period between trials.

After you complete the metabolic testing, you will be given 10 minutes to rest. During this time, reflective markers will be placed on several bilateral landmarks on your body (ex: shoulders, low back, hips, thighs, shins, ankles and feet) for the gait analysis. Measurements and marker application for female volunteers will be performed by a female member for the research team. You will be asked to walk at a self-selected pace down an 18-meter runway. You will perform three successful trials for each leg without an AFO and while wearing your current AFO, and then in both conditions while jogging. The entire procedure will take approximately two and a half hours.

The metabolic and gait analysis will be repeated while wearing your new custom AFO within one week after being issued the new brace, this session should last approximately 1 hour. You will be asked to keep a journal on your use of the new custom AFO and return this journal to study personnel monthly. You will also be asked to return to the Human Performance and Biomechanics lab monthly (+/- one week) to repeat the metabolic and gait analysis while wearing your new custom AFO monthly for six months after you are issued the brace. You will also repeat the anthropometric measures and the EMG/NCS at six months as part of the standard care for your condition.

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jan 12 for TAMC # 31H10
This version of the consent form expires on 26 Jan 13

Volunteer Agreement Affidavit

If you are unable to continue participation in the study due to a change of station, military discharge, or if you withdraw for any reason, you will be asked to complete an exit study visit. Study procedures that will be done at the early exit visit include leg measurements, and EMG, NCS and anthropometric testing in the Physical Medicine and Rehabilitative Services and Physical Therapy Departments at Tripler as well as gait and metabolic analysis (shoes only and DBS-AFO) at the Human Performance and Biomechanics Lab at UH. AFO use journal and quality of life data will also be collected at the exit study visit.

All testing procedures will be explained to you again at each data collection session. The table below explains the data collection time line:

Procedure	Brace Fitting and Issue		Follow-up (months)					
	1 to 28 days before brace issue	1 to 14 days after brace issue	1	2	3	4	5	6 (or exit visit)
EMG / NCS	X							X
Anthropometrics	X							X
Gait Analysis: shoes only	X				X			X
Gait Analysis: current AFO	X							
Gait Analysis: custom AFO		X	X	X	X	X	X	X
Metabolic Analysis: shoes only	X				X			X
Metabolic Analysis: traditional AFO	X							
Metabolic Analysis: custom AFO		X	X	X	X	X	X	X
Quality of Life Survey	X		X	X	X	X	X	X
AFO use journal			X	X	X	X	X	X

INCLUSION AND EXCLUSION CRITERIA: All service members with drop-foot will be eligible for study participation. Service members with transient lower extremity nerve palsy, those who have used a traditional AFO for more than two years or less than one month, and those unable to continuously walk for 10 minutes will be excluded from study participation.

REASONABLY FORESEEABLE RISKS OR DISCOMFORTS: Due to the level of physical activity involved in the gait and metabolic analyses, there is a risk of injury. Subjects may also have some discomfort, muscle cramping or soreness during or after test sessions. The Human Performance and Biomechanics lab is equipped with a fall prevention system; however there is chance of falling during the walking and running tests. There is a very remote chance of cardiac arrest and/or death. These risks are comparable to routine rehabilitation and activities of daily living, and will not affect subjects' rehabilitation and recovery. The investigators are BOC certified athletic trainers and First Aid/CPR/AED trained. In the event of any physical injury from the gait and metabolic analyses, only immediate and essential medical treatment is available including an AED. First Aid/CPR and a referral to a medical emergency room will be provided. All data will be collected by medically certified personnel associated with the research study.

COMPENSATION FOR INJURY: Should you be injured as a direct result of participating in this research project, you will be provided medical care at TAMC, at no cost to you, for that injury. You will not receive any injury compensation, only medical care. This is not a waiver or release of your legal rights. You should discuss this issue thoroughly with the principal investigator before you enroll in this study.

BENEFIT(S) TO THE SUBJECT OR TO OTHERS: Participants will receive one custom Dynamic Bracing Solutions AFO with the goal of improving your physical function and improving your quality of life. Subjects may not receive direct or immediate benefits from participation in this study. However, you will obtain information regarding their neuromuscular function and walking and running characteristics. Results of this study may assist military physicians, physical therapists and athletic trainers improve the standard care for patients drop foot. You will not be paid for participating in this study.

ALTERNATIVE PROCEDURES OR COURSES OF TREATMENT: If you do not wish to participate in this study, you will receive standard therapy as deemed appropriate by your physician.

CONFIDENTIALITY: Information gained because of your participation in this study may be publicized in the medical literature, discussed as an educational model, and used generally in the furtherance of medical science. Information from this study may be used as part of a scientific publication in medical or professional journals, but you will in no way be personally identified. Complete confidentiality cannot be promised because information bearing on your health may be required to be reported to appropriate medical or command authorities.

Your medical records relating to this study may be reviewed by the Institutional Review Board (IRB) at Tripler Army Medical Center, Clinical Investigation Regulatory Office at Fort Sam Houston, Texas and other government agencies as part of their normal duties in protecting human research subjects, and results of the study will be reported to them. The recipients will treat this information confidentially, and in the event of publication regarding this study, your identity will not be disclosed.

This research study meets the confidentiality requirements of the Health Insurance Portability and Accountability Act (HIPAA). A HIPAA Authorization form for this study will be provided to you separately, and you will be asked to sign that form.

PRECAUTIONS TO BE OBSERVED BY SUBJECT BEFORE AND FOLLOWING THE STUDY: As with any physical activity, subjects may have some discomfort, muscle cramping or soreness during or after test sessions. Report any discomfort to the study investigators.

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jun 12 for TAMC # 31H10
This version of the consent form expires on 26 Jun 13

CIRCUMSTANCES UNDER WHICH YOUR PARTICIPATION MAY BE TERMINATED WITHOUT YOUR CONSENT: (a) Health conditions or other conditions that might occur which may be dangerous or detrimental to you or your health; (b) if military contingency requires it; (c) if you become ineligible for military care as authorized by Army regulation; (d) if the safety monitor determines that continued treatment under this study may be harmful to you.

COSTS TO SUBJECT THAT MAY RESULT FROM PARTICIPATION IN STUDY: Participants are responsible for their own transportation to and from Tripler Army Medical Center and the University of Hawaii. Participants will be reimbursed for parking fees at the University of Hawaii. In accordance with AR 40-38, paragraph 3-3(j)(2), daily charges for inpatient care will be waived while the volunteer is in the hospital if the volunteer would not normally enter the hospital for treatment but is requested to do so as part of a research study or as a result of adverse reaction to the drug(s) or procedure(s) used in this study. This also applies to the volunteer's extension of time in a hospital for a research study when the volunteer is already in the hospital.

SIGNIFICANT NEW FINDINGS: Any significant new findings developed during the course of this study that could affect your willingness to continue participation will be made available to you. The results of the research will be made available to you if you so desire. In some cases complete results may not be known for several years.

APPROXIMATE NUMBER OF SUBJECTS INVOLVED IN THE STUDY: Approximately 15 service members will be involved in this study.

DOMICILIARY CARE STATEMENT: The extent of medical care provided, should it become necessary, is limited and will be within the scope authorized for Department of Defense (DOD) health care beneficiaries. Necessary medical care does not include domiciliary (home or nursing home) care.

FOR FURTHER INFORMATION: For questions about the study, contact the principal investigator:

Dr. Gerard M. Antoine
Chief, Physical Medicine and Rehabilitation Services
(808) 433-6958

For questions about your rights as a research participant, contact the Tripler Army Medical Center's Institutional Review Board (which is a group of people who review the research to protect your rights) at (808) 433-6709 or the University of Hawaii's Committee on Human Subjects at 808-956-5007.

For questions about research related injury, contact the Center Judge Advocate at Tripler Army Medical Center at (808) 433-5311.

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jan 12 for TAMC # 31H10
This version of the consent form expires on 26 Jan 13

Volunteer Agreement Affidavit

IF THERE IS ANY PORTION OF THIS EXPLANATION THAT YOU DO NOT UNDERSTAND, ASK THE INVESTIGATOR BEFORE SIGNING. A COPY OF THE VOLUNTEER AGREEMENT AFFIDAVIT WILL BE PROVIDED TO YOU.

******

I have read the above explanation and agree to participate in the investigational study described.

I do [ ] do not [ ] (check one & initial) consent to the inclusion of this form in my outpatient medical treatment record

SIGNATURE OF VOLUNTEER	DATE	SIGNATURE OF LEGAL GUARDIAN (if volunteer is a minor)
PERMANENT ADDRESS OF VOLUNTEER	TYPED NAME OF WITNESS	
	SIGNATURE OF WITNESS	DATE

NOT APPLICABLE

PERSON OBTAINING CONSENT:

Print Name and Title: ______________________________

Signature: ______________________________

Date: ____________________

A PHOTOCOPY OF THIS FORM MUST BE SIGNED BY ALL VOLUNTEERS.
Approved by the TAMC HUC/IRB on 18 Jun 13 for TAMC # 31H10
This version of the consent form expires on 26 Jun 15

TAMC

Date: __________ Subject ID: __________

Data Collection Session (circle one): Pre Post

EMG/NCS Data Collection Sheet

Nerve Conductive Study:

Nerve	Conduction Velocity	Conduction Amplitude	Latency
Sural nerve			
Superficial Peroneal Nerves			
Tibial Nerve			
Peroneal Nerve			

Electromyographic Data:

Muscle	Motor Unit Analysis
Tibialis Anterior	
Extensor Digitorum Brevis	
Medial Gastroc	
Gluteus Medius	

Date: ___________ Subject ID: ___________

Data Collection Session (circle one): Pre Post

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				

Date: __________ Subject ID: __________
Data Collection Session: 0-0 1 2 3 4 5 6

Traditional AFO Use Questionnaire

1. When did you receive your current AFO?

2. On average, how many hours per day do you wear your AFO?

3. What kind of activities are you ABLE to perform in this AFO?

4. What kind of activities are you UNABLE to perform in this AFO?

Date: __________ Subject ID: __________

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

Metabolic Data Collection Sheet 1

Age: _________Gender: _________Height (cm): __________

Resting Heart Rate (bpm):__________

Wt Shoes Only (kg): __________Wt Shoes + TAFO (kg):_________

Pre-Test %O_2(From Computer):_____________

Self-Selected Treadmill Speed (mph): __________

Time from Calibration to Test Start: _________

<u>No AFO (shoes only)</u>

Post-Test RPE: Chest & Breathing: __________Legs & Joints: _________Overall: __________

Post-Test O_2% (From Computer): __________

Peak VO_2 (ml/kg/min):___________ Max Heart Rate (bpm):__________

Total O_2 Drift Time: Calibration Time + 8 minutes= ______________

Pre-Test %O_2(From Computer):_____________

Time from Calibration to Test Start: _________

<u>Traditional AFO</u>

Post-Test RPE: Chest & Breathing: _________ Legs & Joints: _________Overall: __________

Post-Test O_2% (from computer): __________

Peak VO_2 (ml/kg/min):____________ Max Heart Rate (bpm):- ___________

Total O_2 Drift Time: Calibration Time + 8 minutes= ______________

<u>Additional Test Notes:</u>

Date: __________ Subject ID: __________

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

Metabolic Data Collection Sheet 2 (DBS ONLY)

Age: _______Gender: _______Height (cm): _________

Resting Heart Rate (bpm):__________

Wt Shoes + DAFO (kg):___________

Self-Selected Treadmill Speed From Sheet 1 (mph): __________

Pre-Test %O_2(From Computer):_____________

Time from Calibration to Test Start: _________

DAFO

Post-Test RPE: Chest & Breathing: _________Legs & Joints: __________Overall: __________

Post-Test O_2% (from computer): __________

Peak VO_2 (ml/kg/min):___________ Max Heart Rate (bpm):__________

Total O_2 Drift Time: Calibration Time + 8 minutes= ______________

Additional Test Notes:

Date: ________ Subject ID: __________

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

Metabolic Data Collection Sheet 3

Age: _______Gender: _______Height (cm): __________ Resting Hear Rate (bpm):_________

Wt Shoes Only (kg): __________Wt Shoes + DAFO (kg):___________

Self-Selected Treadmill Speed From Sheet 1 (mph): __________

Pre-Test %O_2(From Computer):_____________

Time from Calibration to Test Start: _________

No AFO (shoes only)

Post-Test RPE: Chest & Breathing: __________ Legs & Joints: _________Overall: __________

Post-Test O_2% (From computer): __________

Peak VO_2 (ml/kg/min):___________ Max Heart Rate (bpm):__________

Total O_2 Drift Time: Calibration Time + 8 minutes= ______________

Pre-Test %O_2 (From Computer):_____________

Time from Calibration to test start: _________

DBS – AFO

Post-Test RPE: Chest & Breathing: _________Legs & Joints: __________Overall: __________

Post-Test O_2% (from computer): __________

Peak VO_2 (ml/kg/min):____________ Max Heart Rate (bpm):___________

Total O_2 Drift Time: Calibration Time + 8 minutes= ______________

Additional Test Notes:

Borg’s Ratings of Perceived Exertion

rating	description
6	NO EXERTION AT ALL
7	
	EXTREMELY LIGHT
8	
9	VERY LIGHT
10	
11	LIGHT
12	
13	SOMEWHAT HARD
14	
15	HARD (HEAVY)
16	
17	VERY HARD
18	
19	EXTREMELY HARD
20	MAXIMAL EXERTION

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

Date: __________ Subject ID: __________

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

Biomechanics Data Collection Sheet 1

Resting HR: ___________

	Shoes Only	Traditional 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

TRADITIONAL AFO

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

Self-Selected Running 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

TRADITIONAL AFO

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

<u>Additional Test Notes:</u>

Date: ___________ Subject ID: ___________

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

Biomechanics Data Collection Sheet 2

Resting HR: ____________

	DAFO
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

DAFO Walking

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

Self-Selected Running 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

DAFO Running

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

Additional Test Notes:

Date: __________ Subject ID: __________

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

Biomechanics Data Collection Sheet 3

Resting HR: ___________

	Shoes Only	DAFO
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

DAFO

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

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

Total Trials: SHOES

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

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

Total Trials: DAFO

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

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

<u>Additional Test Notes:</u>

Subject ID: __________

SF-36 Version 2.0™ Health Survey

(SF-36 v2 Standard, US Version 2.0)

Directions: Answer every question by filling in the correct circle or writing in the information. If you need to change an answer, completely erase the incorrect mark and fill in the correct circle. If you are unsure about how to answer a question, please give the best answer you can. Mark only one answer for each question unless directed otherwise.

---

1. In general, would you say your health is:
   ○ Excellent ○ Very Good ○ Good ○ Fair ○ Poor

2. Compared to one year ago, how would you rate your health in general now?
   ○ Much better ○ Somewhat better ○ About the same ○ Somewhat worse ○ Much worse

---

The following questions are about activities that you might do during a typical day. Does your health now limit you in these activities?	Yes, limited a lot	Yes, limited a little	No, not limited at all
3. Vigorous activities, such as running, lifting heavy objects, participating in strenuous sports	○	○	○
4. Moderate activities, such as moving a table, pushing a vacuum cleaner, bowling, or playing golf	○	○	○
5. Lifting or carrying groceries	○	○	○
6. Climbing several flights of stairs	○	○	○
7. Climbing one flight of stairs	○	○	○
8. Bending, kneeling, or stooping	○	○	○
9. Walking more than a mile	○	○	○
10. Walking several blocks	○	○	○
11. Walking one block	○	○	○
12. Bathing or dressing yourself	○	○	○

During the past 4 weeks, how much of the time have you had any of the following problems with your work or other regular daily activities as a result of your physical health?	All of the time	Most of the time	Some of the time	A little of the time	None of the time
13. Cut down on the amount of time you spent on work or other activities	○	○	○	○	○
14. Accomplished less than you would like	○	○	○	○	○
15. Were limited in the kind of work or other activities	○	○	○	○	○
16. Had difficulty performing the work or other activities (for example, it took extra effort)	○	○	○	○	○

During the past 4 weeks, how much of the time have you had any of the following problems with your work or other regular daily activities as a result of any emotional problems (such as feeling depressed or anxious)?	All of the time	Most of the time	Some of the time	A little of the time	None of the time
17. Cut down on the amount of time you spent on work or other activities	○	○	○	○	○
18. Accomplished less than you would like	○	○	○	○	○
19. Did work or activities less carefully than usual	○	○	○	○	○

20. During the past 4 weeks, to what extent has your physical health or emotional problems interfered with your normal social activities with family, friends, neighbors, or groups?

○ Not at all ○ Slightly ○ Moderately ○ Quite a bit ○ Extremely

21. How much bodily pain have you had during the past 4 weeks?

○ None ○ Very mild ○ Mild ○ Moderate ○ Severe ○ Very severe

22. During the past 4 weeks, how much did pain interfere with your normal work (including both work outside the home and housework)?

○ Not at all ○ A little bit ○ Moderately ○ Quite a bit ○ Extremely

These questions are about how you feel and how things have been with you during the past 4 weeks. For each question, please give the one answer that comes closest to the way you have been feeling. How much of the time during the past 4 weeks.	All of the time	Most of the time	Some of the time	A little of the time	None of the time
23. Did you feel full of life?	○	○	○	○	○
24. Have you been very nervous?	○	○	○	○	○
25. Have you felt so down in the dumps that nothing could cheer you up?	○	○	○	○	○
26. Have you felt calm and peaceful?	○	○	○	○	○
27. Did you have a lot of energy?	○	○	○	○	○
28. have you felt downhearted and depressed?	○	○	○	○	○
29. Did you feel worn out?	○	○	○	○	○
30. Have you been happy?	○	○	○	○	○
31. Did you feel tired?	○	○	○	○	○
32. During the past 4 weeks, how much of the time has your physical health or emotional problems interfered with your social activities (like visiting friends, relatives, etc.)?	○	○	○	○	○

How TRUE or FALSE is each of the following statements for you?	Definitely true	Mostly true	Don't know	Mostly False	Definitely false
33. I see to get sick a little easier than other people	○	○	○	○	○
34. I am as healthy as anybody I know	○	○	○	○	○
35. I expect my health to get worse	○	○	○	○	○
36. My health is excellent	○	○	○	○	○

Subject ID: ___________

DAFO Use Journal

This journal is intended to track your use of the DAFO. Please complete this journal weekly and return it at you next visit to the University of Hawaii Human Performance and Biomechanics Lab. If you have any questions about this journal, please contact Melanie Presuto (Associate Investigator) at mpresuto@hawaii.edu.

Circle the answer(s) that best describes your experience with the DAFO each week.

Week: _____, Dates: __________

On average, how many hours per day did you wear the DAFO? 0-2 2-4 4-6 6-8 8-10 10-12 >12
How many days of physical therapy did you attend this week? 1 2 3 4 5
What kind of activities were you ABLE to perform in the DAFO this week?
What kind of activities were you UNABLE to perform in the DAFO this week that you want to do?

Week: _____, Dates: ___________

On average, how many <u>hours</u> per day did you wear the DAFO? 0-2 2-4 4-6 6-8 8-10 10-12 >12
How many days of physical therapy did you attend this week? 1 2 3 4 5
What kind of activities were you ABLE to perform in the DAFO this week?
What kind of activities were you UNABLE to perform in the DAFO this week that you want to do?

Week: _____, Dates: ___________

On average, how many <u>hours</u> per day did you wear the DAFO? 0-2 2-4 4-6 6-8 8-10 10-12 >12
How many days of physical therapy did you attend this week? 1 2 3 4 5
What kind of activities were you ABLE to perform in the DAFO this week?
What kind of activities were you UNABLE to perform in the DAFO this week that you want to do?

Week: _____, Dates: ___________

On average, how many <u>hours</u> per day did you wear the DAFO? 0-2 2-4 4-6 6-8 8-10 10-12 >12
How many days of physical therapy did you attend this week? 1 2 3 4 5
What kind of activities were you ABLE to perform in the DAFO this week?
What kind of activities were you UNABLE to perform in the DAFO this week that you want to do?

APPENDIX II

Institutional Review Board Approvals

U N I V E R S I T Y   O F   H A W A I ' I

Committee on Human Studies

MEMORANDUM

October 28, 2011

TO: Iris Kimura, Ph.D.
Melanie M. Presuto, M.S.
Principal Investigators
Kinesiology & Rehabilitation Science

FROM: Nancy R. King
Director

SUBJECT: CHS #18147- "Functional Analysis of Dynamic Ankle-Foot Orthoses to Improve Outcomes in Partial Lower Extremity Analysis"

Your research project identified above, including the protocol, informed consent/privacy authorization form, traditional AFO Use Questionnaire, DBS-AFO Use Journal, SF-36 Health Related Quality of Life Questionnaire, was approved for one year by the University of Hawaii (UH) Committee on Human Studies (CHS) at its IRB meeting on October 19, 2011.

This memorandum is your record of CHS approval of this study. Please maintain it with your study records.

CHS approval for this project will expire on October 18, 2012. If you expect your project to continue beyond this date, you must submit an application for renewal of this CHS approval. CHS approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes to this study, you must obtain CHS approval prior to implementing them. Unanticipated problems that are likely to affect study participants must be promptly reported to the CHS.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by CHS and other authorized agencies.

Please notify this office when your project is completed. Upon notification, we will close our files pertaining to your project. Reactivation of CHS approval will require a new CHS application.

Please contact this office if you have any questions or require assistance. We appreciate your cooperation, and wish you success with your research.

1960 East-West Road, Biomedical B104, Honolulu, HI 96822-2303
Telephone: (808) 956-5007, Facsimile: (808) 956-8683, Website: www.hawaii.edu/irb

UNIVERSITY of HAWAI'I®
MĀNOA

Office of Research Compliance
Human Studies Program

MEMORANDUM

April 30, 2012

TO: Iris Kimura, Ph.D.
Melanie M. Presuto, M.S.
Principal Investigators
Kinesiology & Rehabilitation Science

FROM: Ching Yuan Hu, Ph.D.
Interim Director
Human Studies Program
Office of Research Compliance
University of Hawaii, Manoa

SUBJECT: CHS #18147 – "Functional Analysis of Dynamic Ankle-Foot Orthoses to Improve Outcomes in Partial Lower Extremity Paralysis"

Your application for the Human Studies Program approval of a proposed change for the study identified above was approved by the Human Studies Program on April 28, 2012. The approved changes were for the addition of the age and anthropometrically-matched healthy, uninjured service member control population and revised consent form. This application qualified for Expedited Review under CFR 46.110 and 21 CFR 56.110, Category (b).

If future revisions to your study are required, please seek the Human Studies Program approval prior to their implementation. If a change is necessary to protect the safety or welfare of study participants, it is permissible to make the change without prior approval. However, you must notify the Human Studies Program as soon as possible, requesting approval for the change.

When seeking approval to modify a Human Studies Program-approved document, please submit the document using "Track Changes" to identify the proposed modifications. Clearly explain the reason for the change on the Human Studies Modification form.

Please contact the Human Studies Program office at 956-5007 if you have any questions or require assistance.

1960 East-West Road
Biomedical Sciences Building B104
Honolulu, Hawai'i 96822
Telephone: (808) 956-5007
Fax: (808) 956-8683
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Office of Research Compliance
Human Studies Program

UNIVERSITY of HAWAI'I
MĀNOA

MEMORANDUM

September 24, 2012

TO: Iris Kimura, Ph.D.
Melanie M. Presuto, M.S.
Principal Investigators
Kinesiology & Rehabilitation Science

FROM: Denise A. Lin-DeShetler, MPH, MA
Director

SUBJECT: CHS #18147- "Functional Analysis of Dynamic Ankle-Foot Orthoses to Improve Outcomes in Partial Lower Extremity Paralysis"

Your research project identified above, including the informed consent/privacy authorization form, was approved for one year by the University of Hawaii (UH) Human Studies Program at its IRB meeting on September 19, 2012.

The application states that the enrollment target is 15, yet the actual enrollment target is 30. Report the correct enrollment target in future applications.

This memorandum is your record of the Human Studies Program approval of this study. Please maintain it with your study records.

The Human Studies Program approval for this project will expire on September 18, 2013. If you expect your project to continue beyond this date, you must submit an application for renewal of this Human Studies Program approval. Human Studies Program approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes to this study, you must obtain approval from the Human Studies Program prior to implementing any changes. If an Unanticipated Problem occurs during the course of the study, you must notify the Human Studies Program within 24 hours of knowledge of the problem. A formal report must be submitted to the Human Studies Program within 10 days. The definition of "Unanticipated Problem" may be found at:
http://hawaii.edu/irb/download/documents/SOPP_101_UP_Reporting.pdf, and the report form may be downloaded here: http://hawaii.edu/irb/download/forms/App_UP_Report.doc.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by the Human Studies Program and other authorized agencies.

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Biomedical Sciences Building B104
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Telephone: (808) 956-5007
Fax: (808) 956-8683

UNIVERSITY OF HAWAI'I

Committee on Human Studies

MEMORANDUM

September 9, 2011

TO: Melanie Presuto, M.S.
Principal Investigator
Kinesiology & Rehabilitation Science

FROM: Nancy R. King
Director

SUBJECT: CHS #19463- "Case Study: Biomechanical Gait Analysis of Dynamic Ankle-Foot Orthosis"

Under an expedited review procedure, the research project identified above was approved for one year on September 9, 2011 by the University of Hawaii (UH) Committee on Human Studies (CHS). The application qualified for expedited review under CFR 46.110 and 21 CFR 56.110, Category (4).

This memorandum is your record of CHS approval of this study. Please maintain it with your study records.

CHS approval for this project will expire on September 8, 2012. If you expect your project to continue beyond this date, you must submit an application for renewal of this CHS approval. CHS approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes to this study, you must obtain CHS approval prior to implementing them. Unanticipated problems that are likely to affect study participants must be promptly reported to the CHS.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by CHS and other authorized agencies.

Please notify this office when your project is complete. Upon notification, we will close our files pertaining to your project. Reactivation of CHS approval will require a new CHS application.

Please contact this office if you have any questions or require assistance. We appreciate your cooperation, and wish you success with your research.

1960 East-West Road, Biomedical B104, Honolulu, Hawaii 96822-2303
Telephone: (808) 956-5007, Facsimile: (808) 956-8683, Website: www.hawaii.edu/irb

Office of Research Compliance
Human Studies Program

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MEMORANDUM

July 19, 2012

TO: Melanie M. Presuto, MS
Principal Investigator
Kinesiology & Rehabilitation Sciences

FROM: Ching Yuan Hu, Ph.D.
Interim Director
Human Studies Program
Office of Research Compliance
University of Hawaii, Manoa

SUBJECT: CHS #19463- "Case Study: Biomechanical Gait Analysis of Dynamic Ankle-Foot Orthosis"

Under an expedited review procedure, the research project identified above was approved for one year on July 18, 2012 by the University of Hawaii (UH) Human Studies Program. The application qualified for expedited review under CFR 46.110 and 21 CFR 56.110, Category (4).

This memorandum is your record of the Human Studies Program approval of this study. Please maintain it with your study records.

<u>The Human Studies Program approval for this project will expire on July 17, 2013</u>. If you expect your project to continue beyond this date, you must submit an application for renewal of this Human Studies Program approval. The Human Studies Program approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes to this study, you must obtain the Human Studies Program approval prior to implementing them. Unanticipated problems that are likely to affect study participants must be promptly reported to the Human Studies Program.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by the Human Studies Program and other authorized agencies.

1960 East-West Road
Biomedical Sciences Building B104
Honolulu, Hawai'i 96822
Telephone: (808) 956-5007
Fax: (808) 956-8683

Office of Research Compliance
Human Studies Program

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MEMORANDUM

November 8, 2012

TO: Melanie Presuto, M.S.
Principal Investigator
Kinesiology & Rehabilitation Sciences

FROM: Denise A. Lin-DeShetler, MPH, MA
Director

SUBJECT: CHS #19463 – "Case Study: Biomechanical Gait Analysis of Dynamic Ankle-Foot Orthosis or Prosthetic"

Your application for the Human Studies Program approval of a proposed change for the study identified above was approved by the Human Studies Program on November 2, 2012. The approved changes were for the title change and the change of the third case study from a limb-salvage patient to an amputee patient for comparison of prosthetic and dynamic ankle-foot orthotic (DAFO) gait characteristics and quality of life. This application qualified for Expedited Review under CFR 46.110 and 21 CFR 56.110, Category (b). Note that this approval date is for the proposed revision, and does not reset the annual study expiration date. Please refer back to your most recent IRB approval letter (initial application or continuing review) for the study's expiration date. Regulations require that continuing review be conducted on or before the one-year anniversary date of IRB approval.

If future revisions to your study are required, please seek the Human Studies Program approval prior to their implementation. If a change is necessary to protect the safety or welfare of study participants, it is permissible to make the change without prior approval. However, you must notify the Human Studies Program as soon as possible, requesting approval for the change.

When seeking approval to modify a Human Studies Program-approved document, please submit the document using "Track Changes" to identify the proposed modifications. Clearly explain the reason for the change on the Human Studies Modification form.

Please contact the Human Studies Program office at 956-5007 if you have any questions or require assistance.

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Honolulu, Hawai'i 96822
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APPENDIX III

Descriptive Data

Table A.1. Control Walking Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	6	2.805	12.438	7.462	1.530	3.748
Ank_Max_DF	6	10.557	17.688	15.059	1.161	2.844
Ank_Time_Max_ DF	6	22.308	76.278	64.603	8.496	20.810
Ank_Max_PF	6	-26.064	-19.504	-24.230	1.016	2.488
Ank_Time_Max_PF	6	98.396	100.000	99.696	0.263	0.643
DF_exc	6	0.975	11.590	7.596	1.476	3.614
Ank_Po_TO	6	-25.874	-19.504	-24.118	0.981	2.402
Ank_MaxDF_Vel	6	88.171	151.191	116.160	10.033	24.575
Ank_Time_MaxDF_Vel	6	13.818	70.972	26.261	9.083	22.248
Ank_MeanDF_Vel	6	-26.069	28.331	11.206	7.725	18.922
Ank_MaxPF_Vel	6	-523.389	-424.468	-464.896	16.748	41.025
Ank_Time_MaxPF_Vel	6	90.374	93.251	92.382	0.440	1.078
Ank_IVEV_Po_IC	6	-9.820	31.563	12.252	5.911	14.480
Ank_MaxEV	6	7.219	51.582	26.070	6.566	16.085
Ank_Time_MaxEV	6	28.017	73.329	50.612	6.928	16.970
Ank_MaxIV	6	-25.265	30.210	2.464	7.423	18.183
Ank_Time_MaxIV	6	3.079	100.000	67.370	19.346	47.387
Ank_EV_exc	6	14.816	32.483	23.605	2.338	5.726
Ank_Po_TO	6	-25.265	40.975	7.401	9.373	22.960
Ank_MaxEV_Vel	6	119.902	234.467	182.893	18.569	45.485
Ank_Time_MaxEV_Vel	6	2.667	95.909	31.266	14.023	34.349
Ank_IRER_Po_IC	6	-24.711	12.707	-7.661	5.487	13.440
Ank_MaxER	6	-40.018	-13.105	-26.543	4.643	11.373
Ank_Time_MaxER	6	14.216	23.040	18.469	1.251	3.065
Ank_MaxIR	6	-23.543	13.065	-1.845	5.335	13.069
Ank_Time_MaxIR	6	2.199	93.800	71.601	14.187	34.751
IRER_Exc	6	14.978	32.007	24.699	2.545	6.235
Ank_IRER_Po_TO	6	-34.187	2.469	-14.152	6.473	15.855
FootProgAng_30percstance	6	-20.234	-2.960	-10.929	2.690	6.589
Mean_footprogang_stance	6	1.128	5.968	4.059	0.843	2.064
Ank_MaxER_Vel	6	-534.062	-213.616	-388.198	46.685	114.355
Ank_Time_MaxER	6	7.710	100.000	40.458	18.566	45.478
Knee_Po_IC	6	1.509	11.040	5.912	1.354	3.316
Knee_Max_FLX	6	40.986	47.353	44.630	1.024	2.508
Knee_Time_Max_FLX	6	100.000	100.000	100.000	0.000	0.000

Knee_FLX_Exc	6	31.064	42.950	38.718	1.628	3.988
Knee_Po_TO	6	40.986	47.353	44.630	1.024	2.508
Knee_Max_FLX_Vel	6	317.113	487.139	392.934	26.853	65.777
Knee_Time_Max_FLX_Vel	6	69.807	100.000	90.967	4.391	10.757
Knee_Mean_FLX_Vel	6	54.953	78.535	67.163	3.590	8.793
Knee_Varval_Po_IC	6	-4.502	10.641	4.728	2.046	5.013
Knee_Max_VAR	6	3.812	19.837	10.789	2.250	5.511
Knee_Time_Max_VAR	6	19.283	100.000	81.157	13.455	32.957
Knee_VAR_Exc	6	1.124	9.195	6.061	1.276	3.125
Knee_Po_TO	6	-0.875	19.837	7.947	3.297	8.076
Knee_Max_VAR_Vel	6	66.141	231.497	161.697	28.063	68.740
Knee_Time_Max_VAR_Vel	6	24.603	98.025	66.578	13.997	34.285
Knee_Mean_VAR_Vel	6	3.409	32.333	14.895	3.973	9.731
Knee_IRER_Po_IC	6	-17.028	5.722	-3.264	3.362	8.235
Knee_MaxIR	6	6.139	25.954	14.192	2.956	7.240
Knee_Time_MaxIR	6	18.556	100.000	60.265	15.112	37.017
Knee_IR_Exc	6	9.468	29.919	17.456	2.863	7.013
KneeIR_Po_TO	6	-5.039	18.881	8.163	3.181	7.793
Knee_MaxIRER_Vel	6	223.552	709.599	413.861	66.978	164.063
Knee_Time_Max_Rot_Vel	6	6.265	11.387	8.786	0.728	1.783
Knee_MeanIRER_Vel	6	22.690	189.533	75.285	24.602	60.263
Hip_FLXEXT_Po_IC	6	26.679	44.288	35.563	2.612	6.398
Hip_Max_FLX	6	26.679	44.288	35.629	2.627	6.435
Hip_Time_MaxFLX	6	0.614	5.778	2.141	0.917	2.246
Hip_Max_EXT	6	-25.741	-1.292	-13.997	3.399	8.325
Hip_Time_MaxEXT	6	78.345	85.236	82.066	1.067	2.612
Hip_ Exc_ICtomaxEXT	6	-54.956	-43.006	-49.560	1.884	4.614
Hip_FLXEXT_Po_TO	6	-15.495	10.344	-2.564	3.774	9.245
Hip_MaxEXT_Vel	6	-213.703	212.429	-48.133	82.656	202.464
Hip_Time_MaxEXT_Vel	6	24.588	100.000	57.348	13.809	33.825
Hip_MeanEXT_Vel_ICtoMaxE XT	6	-129.034	-81.702	-105.181	7.667	18.781
HipABAD_Po_IC	6	-11.268	1.098	-6.353	1.897	4.648
Hip_MaxADD	6	2.419	8.212	4.132	0.863	2.115
Hip_Time_MaxADD	6	29.397	52.865	40.097	3.863	9.461
Hip_MaxABD	6	-14.531	-4.355	-10.428	1.362	3.335
Hip_Time_MaxABD	6	33.775	100.000	72.421	13.287	32.547
Hip_Exc_ICtoMaxABD	6	-11.496	-0.236	-4.075	1.710	4.188
Hip_ABAD_Po_TO	6	-14.531	-4.355	-10.173	1.345	3.294

Hip_MaxADD_Vel	6	100.276	128.615	114.220	4.875	11.942
Hip_Time_MaxADD_Vel	6	12.835	17.356	15.541	0.630	1.542
Hip_Mean_Vel_ICtoMaxABD	6	-21.806	29.304	-0.114	7.728	18.929
HipIRER_Po_IC	6	-21.040	10.541	-7.241	4.323	10.588
Hip_MaxIR	6	-9.602	18.963	4.283	3.731	9.138
Hip_Time_MaxIR	6	5.346	100.000	45.501	14.939	36.593
Hip_MaxER	6	-21.104	9.356	-12.012	4.485	10.986
Hip_Time_MaxER	6	1.931	93.595	58.181	16.006	39.207
Hip_Exc_ICtoMaxIR	6	0.457	27.556	11.523	3.818	9.353
Hip_Exc_MaxIRtoMaxER	6	-27.620	-9.126	-16.295	2.777	6.803
HipIRER_Po_TO	6	-18.249	12.579	-0.928	4.796	11.748
Hip_MaxIR_Vel	6	101.883	403.773	271.747	51.296	125.648
Hip_Time_MaxIR_Vel	6	0.715	48.810	21.948	7.860	19.254
Hip_Mean_Vel_ICtoMaxIR	6	17.887	87.000	51.273	10.042	24.598
Hip_MaxER_Vel	6	-376.507	-133.524	-211.493	36.335	89.002
Hip_Time_MaxER_Vel	6	8.385	51.242	25.742	6.430	15.750
Max_Pelv_Ang	6	3.636	7.386	4.974	0.564	1.381
Pelv_Ang_Exc	6	7.672	11.844	9.669	0.655	1.604
Max_thor_angle	6	1.072	3.557	2.603	0.382	0.936
Thor_ang_exc	6	2.182	7.054	4.565	0.740	1.813
Max_spine_ang	6	2.053	9.841	5.885	1.333	3.265
Spine_ang_exc	6	9.838	17.907	13.514	1.505	3.687
MaxGRFz	6	11.254	13.786	12.576	0.374	0.916
Time_MaxGRFz	6	19.253	78.465	43.842	9.230	22.608
Load_Rate	6	1990.085	11947.782	6586.068	1455.568	3565.398
GRFimp	6	326.895	525.050	401.830	30.083	73.687
GRFimp_norm	6	0.422	0.549	0.476	0.019	0.047
Max_brake_GRFy	6	-3.596	-2.092	-2.928	0.239	0.584
Time_max_brake_GRFy	6	16.313	18.607	17.362	0.446	1.093
Max_prop_GRFy	6	2.425	3.245	2.896	0.144	0.354
Time_max_prop_GRFy	6	84.490	88.104	86.491	0.513	1.257
Max_Abs_FM	6	0.002	0.006	0.004	0.001	0.001
Max_ADD_FM	6	0.002	0.006	0.004	0.001	0.001
Time_max_ADD_FM	6	36.175	76.248	62.446	7.108	17.410
Max_ABD_FM	6	-0.002	0.000	-0.001	0.000	0.001
Time_Max_ABD_FM	6	0.802	100.000	33.465	16.223	39.739
FM_Peak_Brake_GRFy	6	-0.002	0.005	0.001	0.001	0.002
ADD_FM_imp	6	0.001	0.002	0.001	0.000	0.000
Max_DF_Mom	6	1475.033	1862.033	1611.257	65.481	160.396
Time_Max_DF_Mom	6	77.543	80.678	79.037	0.479	1.174

Max_PF_Mom	6	-653.170	-299.850	-445.198	51.578	126.340
Time_Max_PF_Mom	6	8.675	12.270	10.852	0.534	1.308
MaxIV_Mom	6	6.489	307.981	167.758	51.393	125.886
Time_MaxIV_Mom	6	79.143	100.000	84.875	3.206	7.853
MaxEV_Mom	6	-108.764	-13.881	-61.462	16.901	41.398
Time_MaxEV_Mom	6	7.377	64.034	31.086	10.790	26.429
Ank_MaxIR_Mom	6	145.717	277.841	230.823	18.845	46.160
Time_MaxIR_Mom	6	73.493	88.956	78.375	2.437	5.969
MaxER_Mom	6	-31.093	-11.823	-22.689	2.740	6.711
Time_MaxER_Mom	6	9.888	100.000	61.976	16.107	39.455
Max_FLX_Mom_LR	6	612.593	1547.060	1108.007	137.133	335.906
Time_Max_FLX_Mom_LR	6	16.385	20.657	18.911	0.655	1.605
Max_FLX_Mom_TO	6	612.593	1547.060	1114.296	135.380	331.613
Time_Max_FLX_Mom_TO	6	16.385	65.722	26.422	7.879	19.301
Max_EXT_Mom	6	-569.373	-197.486	-351.028	51.404	125.914
Time_Max_EXT_Mom	6	3.373	43.736	10.921	6.571	16.097
Knee_ Stiff	6	5.168	11.705	6.980	0.995	2.438
Knee_MaxADD_Mom	6	550.815	1529.670	1103.481	138.042	338.132
Knee_Timing_MaxADD_Mom	6	18.314	23.797	21.169	0.758	1.857
Knee_MaxABD_Mom	6	-124.032	-15.124	-76.021	18.598	45.556
Knee_Time_MaxABD_Mom	6	0.794	97.605	55.051	18.657	45.700
Knee_MaxIR_Mom	6	132.598	276.141	225.537	20.197	49.473
Knee_Time_MaxIR_Mom	6	74.841	78.606	76.298	0.540	1.323
Knee_MaxER_Mom	6	-48.082	-5.038	-25.712	6.108	14.961
Time_MaxER_Mom	6	11.653	100.000	52.634	17.288	42.346
Hip_Max_FLX_Mom	6	549.423	1151.193	969.282	102.341	250.683
Hip_Time_MaxFLX_Mom	6	3.469	12.561	9.316	1.452	3.557
Hip_MaxEXT_Mom	6	-2271.163	-1583.557	-1952.735	119.560	292.860
Hip_Time_MaxEXT_Mom	6	80.483	84.396	82.337	0.529	1.295
Hip_MaxADD_Mom	6	816.080	1242.267	1042.118	58.633	143.621
Hip_Time_MaxADD_Mom	6	22.826	61.885	30.925	6.219	15.235
Hip_MaxABD_Mom	6	-564.826	-186.262	-350.541	60.854	149.060
Hip_Time_MaxABD_Mom	6	6.673	95.097	46.783	17.478	42.812
Hip_MaxIR_Mom	6	127.388	231.153	194.537	15.087	36.956
Hip_Time_MaxIR_Mom	6	68.503	77.350	74.333	1.324	3.243
Hip_MaxER_Mom	6	-236.885	-66.268	-168.192	27.055	66.271
Hip_Time_MaxER_Mom	6	15.271	23.716	20.941	1.243	3.045
Inv_A1_Joint_Power	6	-1.054	-0.123	-0.716	0.140	0.344
Inv_A2_Joint_Power	6	5.615	7.656	6.629	0.342	0.837

Inv_K1_Joint_Power	6	-4.650	-1.559	-3.354	0.525	1.286
Inv_K2_Joint_Power	6	0.999	3.383	2.251	0.423	1.035
Inv_K3_Joint_Power	6	-0.298	-0.074	-0.179	0.038	0.093
Inv_K4_Joint_Power	6	-5.609	-1.985	-3.681	0.587	1.438
Inv_H1_Joint_Power	6	0.340	0.769	0.605	0.067	0.165
Inv_H2_Joint_Power	6	-0.901	-0.190	-0.408	0.102	0.251
Inv_Swing_Time	6	0.367	0.417	0.392	0.009	0.021
Uninv_Swing_Time	6	0.358	0.418	0.389	0.010	0.025
Inv_Stance_Time	6	0.513	0.666	0.579	0.025	0.061
Uninv_Stance_Time	6	0.519	0.661	0.584	0.024	0.058
Inv_Step_Length	6	0.828	0.956	0.896	0.023	0.056
Uninv_Step_Length	6	0.819	0.943	0.885	0.022	0.054
Stride_L	6	1.648	1.893	1.781	0.045	0.110
Velocity	6	1.596	2.125	1.860	0.088	0.215

Table A.2. No Brace (NB) Walking Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	6	-24.927	-1.880	-10.390	3.420	8.378
Ank_Max_DF	6	13.483	26.028	19.439	1.774	4.345
Ank_Time_Max_ DF	6	72.815	85.462	77.748	2.027	4.966
Ank_Max_PF	6	-30.760	-12.029	-20.849	2.761	6.762
Ank_Time_Max_PF	6	0.594	100.000	68.218	20.128	49.304
DF_exc	6	15.363	45.915	29.829	4.231	10.364
Ank_Po_TO	6	-30.760	2.559	-15.549	4.636	11.355
Ank_MaxDF_Vel	6	121.209	314.730	200.739	36.547	89.522
Ank_Time_MaxDF_Vel	6	4.754	75.615	23.645	10.638	26.057
Ank_MeanDF_Vel	6	33.476	83.634	61.352	7.918	19.395
Ank_MaxPF_Vel	6	-1726.223	-325.480	-699.350	210.190	514.857
Ank_Time_MaxPF_Vel	6	93.451	100.000	97.660	1.307	3.201
Ank_IVEV_Po_IC	6	-22.559	35.494	1.275	7.766	19.023
Ank_MaxEV	6	-0.952	58.515	23.333	8.571	20.994
Ank_Time_MaxEV	6	10.209	87.699	48.753	11.934	29.231
Ank_MaxIV	6	-26.088	35.457	-5.063	8.992	22.026
Ank_Time_MaxIV	6	0.594	91.101	44.650	19.631	48.087
Ank_EV_exc	6	15.405	52.440	28.396	5.286	12.949
Ank_Po_TO	6	-16.200	51.944	8.959	10.351	25.355
Ank_MaxEV_Vel	6	179.979	937.935	428.682	119.074	291.670
Ank_Time_MaxEV_Vel	6	2.627	94.668	36.076	14.984	36.703
Ank_IRER_Po_IC	6	-11.297	15.637	4.883	3.885	9.517

Ank_MaxER	6	-24.119	12.553	-12.134	5.543	13.578
Ank_Time_MaxER	6	9.786	42.720	23.340	4.786	11.723
Ank_MaxIR	6	-4.208	29.002	15.824	5.048	12.364
Ank_Time_MaxIR	6	49.743	100.000	77.165	8.317	20.372
IRER_Exc	6	10.425	48.368	27.958	6.577	16.111
Ank_IRER_Po_TO	6	-19.785	29.002	3.707	7.466	18.287
FootProgAng_30percstance	6	-20.320	-4.426	-11.349	2.346	5.748
Mean_footprogang_stance	6	-0.023	6.751	3.668	0.993	2.433
Ank_MaxER_Vel	6	-797.124	-119.628	-345.103	96.751	236.990
Ank_Time_MaxER	6	2.671	99.037	36.261	15.289	37.450
Knee_Po_IC	6	0.140	13.234	6.898	2.179	5.337
Knee_Max_FLX	6	26.763	59.486	44.468	4.881	11.955
Knee_Time_Max_FLX	6	94.177	100.000	99.029	0.971	2.377
Knee_FLX_Exc	6	22.209	54.061	37.570	5.032	12.325
Knee_Po_TO	6	9.434	59.486	41.580	7.254	17.768
Knee_Max_FLX_Vel	6	241.952	579.560	406.506	48.323	118.367
Knee_Time_Max_FLX_Vel	6	87.702	97.826	94.427	1.479	3.622
Knee_Mean_FLX_Vel	6	36.634	86.183	62.651	8.047	19.711
Knee_Varval_Po_IC	6	-2.290	9.612	3.112	2.036	4.987
Knee_Max_VAR	6	6.149	32.523	15.876	3.848	9.427
Knee_Time_Max_VAR	6	11.689	100.000	59.870	18.110	44.359
Knee_VAR_Exc	6	3.021	34.813	12.764	4.801	11.760
Knee_Po_TO	6	-6.110	32.523	12.610	5.377	13.170
Knee_Max_VAR_Vel	6	120.621	559.181	249.207	64.823	158.783
Knee_Time_Max_VAR_Vel	6	63.060	96.766	85.106	5.858	14.349
Knee_Mean_VAR_Vel	6	22.216	54.065	37.591	5.341	13.083
Knee_IRER_Po_IC	6	-22.237	0.996	-8.130	3.743	9.168
Knee_MaxIR	6	-9.550	20.324	6.261	4.240	10.385
Knee_Time_MaxIR	6	21.690	100.000	71.665	13.797	33.795
Knee_IR_Exc	6	6.314	21.847	14.391	2.179	5.337
KneeIR_Po_TO	6	-96.645	20.324	-15.411	17.187	42.099
Knee_MaxIRER_Vel	6	84.334	882.700	331.616	117.962	288.948
Knee_Time_Max_Rot_Vel	6	5.178	71.705	24.960	11.077	27.132
Knee_MeanIRER_Vel	6	17.062	116.923	46.028	16.042	39.294
Hip_FLXEXT_Po_IC	6	23.735	53.343	34.774	4.160	10.190
Hip_Max_FLX	6	23.735	53.343	34.774	4.160	10.190
Hip_Time_MaxFLX	6	0.594	0.784	0.693	0.028	0.069
Hip_Max_EXT	6	-22.494	0.333	-10.358	3.377	8.273

Hip_Time_MaxEXT	6	77.046	85.931	81.648	1.406	3.444
Hip_ Exc_ICtomaxEXT	6	-53.873	-38.326	-45.133	2.821	6.911
Hip_FLXEXT_Po_TO	6	-12.418	17.002	1.579	4.374	10.715
Hip_MaxEXT_Vel	6	-211.984	198.213	-46.236	71.323	174.705
Hip_Time_MaxEXT_Vel	6	0.671	100.000	49.003	15.954	39.079
Hip_MeanEXT_Vel_ICtoMaxEXT	6	-121.746	-72.967	-93.380	8.029	19.668
HipABAD_Po_IC	6	-10.983	-1.415	-6.263	1.380	3.379
Hip_MaxADD	6	-0.534	9.125	4.921	1.384	3.390
Hip_Time_MaxADD	6	19.597	87.343	35.287	10.506	25.735
Hip_MaxABD	6	-16.074	-8.452	-11.758	1.295	3.173
Hip_Time_MaxABD	6	99.114	100.000	99.852	0.148	0.362
Hip_Exc_ICtoMaxABD	6	-10.261	-2.113	-5.495	1.261	3.090
Hip_ABAD_Po_TO	6	-16.050	-8.452	-11.754	1.293	3.167
Hip_MaxADD_Vel	6	51.822	197.139	145.960	22.982	56.295
Hip_Time_MaxADD_Vel	6	4.693	82.599	21.024	12.403	30.380
Hip_Mean_Vel_ICtoMaxABD	6	-16.369	-3.337	-8.529	2.028	4.967
HipIRER_Po_IC	6	-32.883	3.417	-15.293	5.427	13.293
Hip_MaxIR	6	-5.528	64.761	18.814	9.807	24.022
Hip_Time_MaxIR	6	18.930	100.000	70.048	15.106	37.002
Hip_MaxER	6	-32.883	-5.798	-17.491	4.362	10.684
Hip_Time_MaxER	6	0.671	91.267	29.798	17.986	44.056
Hip_Exc_ICtoMaxIR	6	4.495	71.047	34.107	10.386	25.440
Hip_Exc_MaxIRtoMaxER	6	-71.518	-9.292	-36.304	9.301	22.782
HipIRER_Po_TO	6	-8.308	64.761	14.397	11.046	27.057
Hip_MaxIR_Vel	6	107.658	1871.085	550.074	269.665	660.542
Hip_Time_MaxIR_Vel	6	24.944	100.000	58.651	12.372	30.305
Hip_Mean_Vel_ICtoMaxIR	6	28.909	257.658	96.725	34.426	84.326
Hip_MaxER_Vel	6	-349.637	-66.591	-186.531	52.740	129.186
Hip_Time_MaxER_Vel	6	10.844	55.946	29.733	6.662	16.319
Max_Pelv_Ang	6	2.464	7.245	4.885	0.773	1.893
Pelv_Ang_Exc	6	4.747	15.446	10.451	1.424	3.487
Max_thor_angle	6	1.454	7.044	4.199	0.808	1.980
Thor_ang_exc	6	5.042	8.252	6.542	0.437	1.071
Max_spine_ang	6	2.814	10.285	6.727	1.193	2.923
Spine_ang_exc	6	10.328	19.489	15.642	1.412	3.458
MaxGRFz	6	9.975	12.329	10.882	0.370	0.907
Time_MaxGRFz	6	15.664	76.374	32.628	9.570	23.443
Load_Rate	6	1573.580	10587.102	6900.902	1271.047	3113.417
GRFimp	6	292.515	588.509	381.563	44.419	108.803

GRFimp_norm	6	0.379	0.530	0.446	0.022	0.053
Max_brake_GRFy	6	-3.277	-1.524	-2.342	0.241	0.590
Time_max_brake_GRFy	6	13.049	18.770	16.284	0.897	2.197
Max_prop_GRFy	6	1.438	2.827	2.116	0.194	0.476
Time_max_prop_GRFy	6	74.478	87.995	84.496	2.046	5.012
Max_Abs_FM	6	0.002	0.004	0.003	0.000	0.001
Max_ADD_FM	6	0.002	0.004	0.003	0.000	0.001
Time_max_ADD_FM	6	29.730	86.782	63.044	8.180	20.038
Max_ABD_FM	6	-0.003	-0.001	-0.001	0.000	0.001
Time_Max_ABD_FM	6	9.900	80.697	29.895	10.866	26.617
FM_Peak_Brake_GRFy	6	-0.001	0.003	0.001	0.000	0.001
ADD_FM_imp	6	0.000	0.001	0.001	0.000	0.000
Max_DF_Mom	6	454.423	1566.217	1243.443	168.931	413.794
Time_Max_DF_Mom	6	77.655	84.562	79.877	1.063	2.604
Max_PF_Mom	6	-480.566	-38.400	-218.656	79.598	194.976
Time_Max_PF_Mom	6	6.028	100.000	69.168	19.504	47.774
MaxIV_Mom	6	-1.194	332.250	127.065	50.286	123.175
Time_MaxIV_Mom	6	36.181	86.409	68.738	9.288	22.751
MaxEV_Mom	6	-159.865	-9.843	-67.705	21.992	53.869
Time_MaxEV_Mom	6	9.673	51.201	18.661	6.565	16.081
Ank_MaxIR_Mom	6	36.532	231.590	145.825	25.752	63.080
Time_MaxIR_Mom	6	35.106	93.852	74.922	8.785	21.518
MaxER_Mom	6	-98.059	2.686	-44.011	15.083	36.946
Time_MaxER_Mom	6	0.671	100.000	58.704	15.487	37.936
Max_FLX_Mom_LR	6	306.459	6560.360	1467.124	1021.611	2502.426
Time_Max_FLX_Mom_LR	6	-67.862	20.879	-30.627	14.263	34.937
Max_FLX_Mom_TO	6	229.936	2837.110	1029.437	399.637	978.908
Time_Max_FLX_Mom_TO	6	16.744	99.677	62.091	12.293	30.110
Max_EXT_Mom	6	-767.180	-157.332	-366.049	96.537	236.466
Time_Max_EXT_Mom	6	7.565	99.026	43.463	15.553	38.096
Knee_ Stiff	6	0.486	52.730	11.651	8.286	20.296
Knee_MaxADD_Mom	6	579.121	1064.439	744.108	72.172	176.785
Knee_Timing_MaxADD_Mom	6	23.362	98.383	41.897	12.358	30.270
Knee_MaxABD_Mom	6	-5098.735	-38.137	-1000.128	823.425	2016.970
Knee_Time_MaxABD_Mom	6	64.770	100.000	90.433	5.435	13.313
Knee_MaxIR_Mom	6	39.373	208.129	144.016	23.955	58.678
Knee_Time_MaxIR_Mom	6	59.324	88.636	75.650	4.116	10.083
Knee_MaxER_Mom	6	-82.176	5.251	-33.919	14.981	36.695
Time_MaxER_Mom	6	0.671	100.000	64.368	16.821	41.202

Hip_Max_FLX_Mom	6	297.522	4474.275	1399.183	641.243	1570.719
Hip_Time_MaxFLX_Mom	6	4.592	97.736	24.784	14.666	35.924
Hip_MaxEXT_Mom	6	-30805.350	-803.452	-6414.703	4881.226	11956.514
Hip_Time_MaxEXT_Mom	6	73.602	99.677	83.308	3.608	8.837
Hip_MaxADD_Mom	6	710.383	15785.050	3448.295	2468.696	6047.045
Hip_Time_MaxADD_Mom	6	18.542	99.677	35.909	12.831	31.430
Hip_MaxABD_Mom	6	-1090.486	-65.590	-352.337	163.021	399.319
Hip_Time_MaxABD_Mom	6	35.676	97.888	76.548	10.883	26.657
Hip_MaxIR_Mom	6	63.499	2654.695	572.277	417.508	1022.681
Hip_Time_MaxIR_Mom	6	65.193	100.000	82.507	5.863	14.361
Hip_MaxER_Mom	6	-628.243	0.977	-183.877	93.701	229.520
Hip_Time_MaxER_Mom	6	2.014	98.383	33.513	13.738	33.652
Inv_A1_Joint_Power	6	-1.976	-0.163	-1.058	0.252	0.616
Inv_A2_Joint_Power	6	1.183	7.242	4.342	0.816	1.999
Inv_K1_Joint_Power	6	-2.977	-0.134	-1.045	0.477	1.168
Inv_K2_Joint_Power	6	0.191	2.535	1.026	0.384	0.940
Inv_K3_Joint_Power	6	-0.100	0.301	0.007	0.061	0.150
Inv_K4_Joint_Power	6	-5.219	-1.421	-2.552	0.554	1.358
Inv_H1_Joint_Power	6	0.398	2.159	1.247	0.258	0.631
Inv_H2_Joint_Power	6	-1.112	-0.079	-0.527	0.181	0.443
Inv_Swing_Time	6	0.380	0.449	0.422	0.010	0.025
Uninv_Swing_Time	6	0.344	0.410	0.376	0.012	0.029
Inv_Stance_Time	6	0.541	0.663	0.611	0.021	0.052
Uninv_Stance_Time	6	0.568	0.743	0.663	0.029	0.071
Inv_Step_Length	6	0.766	0.823	0.797	0.009	0.022
Uninv_Step_Length	6	0.661	0.807	0.729	0.022	0.054
Stride_L	6	1.467	1.631	1.526	0.028	0.068
Velocity	6	1.319	1.717	1.487	0.064	0.157

Table A.3. Traditional AFO (TAFO) Walking Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	5	-6.254	7.909	1.705	2.500	5.589
Ank_Max_DF	5	17.151	27.325	21.992	1.636	3.659
Ank_Time_Max_ DF	5	77.069	87.276	82.277	1.665	3.723
Ank_Max_PF	5	-13.786	-1.601	-8.502	2.680	5.993
Ank_Time_Max_PF	5	8.042	14.815	12.505	1.225	2.740
DF_exc	5	11.027	27.692	20.288	3.282	7.340
Ank_Po_TO	5	-12.924	5.164	-2.181	3.573	7.990

Ank_MaxDF_Vel	5	71.569	1339.465	376.640	242.255	541.698
Ank_Time_MaxDF_Vel	5	17.559	94.298	50.860	14.622	32.697
Ank_MeanDF_Vel	5	17.916	60.338	37.247	8.597	19.223
Ank_MaxPF_Vel	5	-4749.390	-136.086	-1192.699	891.475	1993.400
Ank_Time_MaxPF_Vel	5	91.903	97.395	94.734	0.924	2.066
Ank_IVEV_Po_IC	5	-6.922	29.403	11.068	7.184	16.063
Ank_MaxEV	5	-2.422	37.782	22.698	7.505	16.783
Ank_Time_MaxEV	5	24.790	94.898	63.687	13.510	30.210
Ank_MaxIV	5	-43.559	24.169	-0.856	11.907	26.625
Ank_Time_MaxIV	5	26.515	80.853	49.128	12.037	26.916
Ank_EV_exc	5	4.909	58.783	23.554	9.284	20.760
Ank_Po_TO	5	-4.296	31.909	17.086	6.845	15.305
Ank_MaxEV_Vel	5	79.765	1451.282	489.125	251.974	563.431
Ank_Time_MaxEV_Vel	5	2.444	95.508	58.194	17.152	38.354
Ank_IRER_Po_IC	5	-181.188	19.634	-26.159	38.909	87.003
Ank_MaxER	5	-247.365	11.945	-51.538	49.204	110.023
Ank_Time_MaxER	5	42.238	100.000	82.737	11.006	24.609
Ank_MaxIR	5	-168.211	31.481	-16.154	38.219	85.461
Ank_Time_MaxIR	5	15.314	79.911	58.916	12.509	27.970
IRER_Exc	5	11.064	79.154	35.384	12.331	27.572
Ank_IRER_Po_TO	5	-214.813	20.854	-42.984	43.429	97.111
FootProgAng_30percstance	5	-13.330	18.900	-3.827	5.776	12.916
Mean_footprogang_stance	5	1.283	183.007	39.412	35.904	80.283
Ank_MaxER_Vel	5	-2730.900	-119.205	-809.736	490.650	1097.126
Ank_Time_MaxER	5	20.923	94.898	66.777	15.939	35.640
Knee_Po_IC	5	0.407	10.777	3.911	1.859	4.156
Knee_Max_FLX	5	36.820	58.560	48.261	3.549	7.936
Knee_Time_Max_FLX	5	94.898	100.000	98.980	1.020	2.282
Knee_FLX_Exc	5	32.044	58.153	44.350	4.582	10.246
Knee_Po_TO	5	34.001	58.560	47.697	4.017	8.983
Knee_Max_FLX_Vel	5	315.581	524.337	404.662	36.742	82.158
Knee_Time_Max_FLX_Vel	5	91.020	94.903	93.707	0.716	1.601
Knee_Mean_FLX_Vel	5	44.560	98.878	67.538	9.921	22.185
Knee_Varval_Po_IC	5	-2.458	5.482	1.363	1.699	3.799
Knee_Max_VAR	5	3.672	28.140	16.290	4.405	9.851
Knee_Time_Max_VAR	5	45.506	100.000	87.206	10.585	23.670
Knee_VAR_Exc	5	3.876	30.353	14.927	4.770	10.667
Knee_Po_TO	5	3.499	28.140	16.081	4.506	10.075

Knee_Max_VAR_Vel	5	120.177	595.483	286.322	83.539	186.799
Knee_Time_Max_VAR_Vel	5	73.029	98.225	90.487	4.543	10.159
Knee_Mean_VAR_Vel	5	7.691	43.714	26.104	6.109	13.659
Knee_IRER_Po_IC	5	-16.316	9.049	-4.187	4.184	9.357
Knee_MaxIR	5	-1.428	40.379	12.633	7.246	16.202
Knee_Time_MaxIR	5	10.062	100.000	74.721	17.064	38.157
Knee_IR_Exc	5	2.515	42.127	16.821	6.696	14.973
KneeIR_Po_TO	5	-23.754	7.176	-4.035	5.561	12.434
Knee_MaxIRER_Vel	5	114.540	2456.570	696.965	446.356	998.083
Knee_Time_Max_Rot_Vel	5	6.340	94.898	43.909	19.834	44.350
Knee_MeanIRER_Vel	5	13.952	63.544	34.811	8.325	18.614
Hip_FLXEXT_Po_IC	5	27.843	48.439	37.846	3.342	7.473
Hip_Max_FLX	5	27.843	48.448	37.848	3.344	7.477
Hip_Time_MaxFLX	5	0.565	0.984	0.718	0.073	0.164
Hip_Max_EXT	5	-14.293	5.181	-6.341	3.721	8.320
Hip_Time_MaxEXT	5	79.239	83.905	81.921	0.927	2.072
Hip_ Exc_ICtomaxEXT	5	-49.692	-41.803	-44.187	1.433	3.205
Hip_FLXEXT_Po_TO	5	-5.889	17.662	5.811	3.943	8.818
Hip_MaxEXT_Vel	5	-168.226	160.818	-75.107	59.769	133.648
Hip_Time_MaxEXT_Vel	5	21.518	93.071	42.740	13.115	29.325
Hip_MeanEXT_Vel_ICtoMaxEXT	5	-91.568	-64.952	-80.518	4.977	11.129
HipABAD_Po_IC	5	-8.198	-2.156	-4.813	1.081	2.417
Hip_MaxADD	5	0.214	9.452	5.380	1.555	3.477
Hip_Time_MaxADD	5	23.142	68.840	39.239	8.130	18.180
Hip_MaxABD	5	-12.740	-4.304	-8.530	1.674	3.743
Hip_Time_MaxABD	5	35.119	100.000	80.109	12.907	28.861
Hip_Exc_ICtoMaxABD	5	-7.590	-0.582	-3.716	1.299	2.904
Hip_ABAD_Po_TO	5	-12.699	-2.158	-8.026	1.963	4.391
Hip_MaxADD_Vel	5	54.866	139.596	107.651	15.581	34.840
Hip_Time_MaxADD_Vel	5	1.196	62.822	22.830	10.685	23.892
Hip_Mean_Vel_ICtoMaxABD	5	-16.800	3.749	-7.722	3.613	8.079
HipIRER_Po_IC	5	-37.728	-14.986	-22.637	3.923	8.772
Hip_MaxIR	5	-4.708	40.375	15.614	7.946	17.768
Hip_Time_MaxIR	5	9.796	100.000	74.136	17.779	39.756
Hip_MaxER	5	-38.300	-15.847	-23.341	3.893	8.704
Hip_Time_MaxER	5	1.662	55.621	14.149	10.392	23.237
Hip_Exc_ICtoMaxIR	5	10.278	60.313	38.251	8.096	18.103
Hip_Exc_MaxIRtoMaxER	5	-61.091	-11.139	-38.955	8.091	18.091

HipIRER_Po_TO	5	-5.890	40.375	14.938	8.289	18.534
Hip_MaxIR_Vel	5	219.233	633.226	423.814	72.950	163.122
Hip_Time_MaxIR_Vel	5	0.710	94.602	41.637	21.221	47.451
Hip_Mean_Vel_ICtoMaxIR	5	65.889	150.831	95.180	14.616	32.681
Hip_MaxER_Vel	5	-194.747	-110.132	-145.433	15.508	34.678
Hip_Time_MaxER_Vel	5	15.143	30.830	21.024	2.922	6.534
Max_Pelv_Ang	5	2.012	5.938	3.909	0.829	1.853
Pelv_Ang_Exc	5	5.317	13.675	9.504	1.582	3.536
Max_thor_angle	5	2.514	6.126	4.308	0.604	1.350
Thor_ang_exc	5	4.751	10.233	7.529	0.890	1.989
Max_spine_ang	5	4.901	9.631	7.480	0.870	1.946
Spine_ang_exc	5	12.934	18.844	15.122	1.310	2.930
MaxGRFz	5	10.033	11.721	10.862	0.326	0.729
Time_MaxGRFz	5	23.880	75.666	40.488	9.161	20.485
Load_Rate	5	1457.578	7246.862	4907.420	1082.617	2420.806
GRFimp	5	282.423	609.496	426.126	61.900	138.413
GRFimp_norm	5	0.364	0.547	0.476	0.035	0.079
Max_brake_GRFy	5	-2.397	-1.236	-1.848	0.193	0.431
Time_max_brake_GRFy	5	17.051	30.699	23.492	2.248	5.026
Max_prop_GRFy	5	0.736	2.093	1.593	0.239	0.535
Time_max_prop_GRFy	5	31.639	87.985	73.709	10.789	24.125
Max_Abs_FM	5	0.002	0.003	0.003	0.000	0.001
Max_ADD_FM	5	0.002	0.003	0.003	0.000	0.001
Time_max_ADD_FM	5	35.131	86.540	65.235	8.998	20.119
Max_ABD_FM	5	-0.002	0.000	-0.001	0.000	0.001
Time_Max_ABD_FM	5	6.615	56.616	28.792	10.060	22.495
FM_Peak_Brake_GRFy	5	0.000	0.002	0.001	0.000	0.001
ADD_FM_imp	5	0.000	0.001	0.001	0.000	0.000
Max_DF_Mom	5	584.059	1524.237	1179.220	170.135	380.433
Time_Max_DF_Mom	5	76.102	90.828	82.651	2.415	5.399
Max_PF_Mom	5	-418.206	-106.830	-221.361	66.533	148.773
Time_Max_PF_Mom	5	13.014	100.000	42.698	18.124	40.527
MaxIV_Mom	5	6.690	352.620	120.689	61.985	138.602
Time_MaxIV_Mom	5	15.009	82.679	54.998	11.952	26.725
MaxEV_Mom	5	-576.483	-10.717	-150.256	107.291	239.911
Time_MaxEV_Mom	5	14.135	79.871	43.690	10.485	23.444
Ank_MaxIR_Mom	5	47.149	1928.808	485.855	361.686	808.754
Time_MaxIR_Mom	5	58.577	97.337	77.055	7.717	17.256
MaxER_Mom	5	-839.576	-14.152	-193.844	161.547	361.231
Time_MaxER_Mom	5	9.654	98.695	71.979	16.601	37.121

Max_FLX_Mom_LR	5	328.327	11950.700	2834.092	2280.030	5098.302
Time_Max_FLX_Mom_LR	5	-62.218	31.968	-3.221	18.703	41.821
Max_FLX_Mom_TO	5	520.458	8493.245	2193.268	1575.517	3522.963
Time_Max_FLX_Mom_TO	5	25.984	94.898	57.197	14.104	31.538
Max_EXT_Mom	5	-12419.015	-252.978	-2772.495	2411.803	5392.956
Time_Max_EXT_Mom	5	3.387	97.929	24.490	18.381	41.102
Knee_ Stiff	5	0.639	179.681	39.115	35.157	78.612
Knee_MaxADD_Mom	5	474.589	11533.950	2772.534	2190.773	4898.717
Knee_Timing_MaxADD_Mom	5	23.727	97.337	43.238	13.777	30.806
Knee_MaxABD_Mom	5	-1685.558	-28.363	-429.820	315.039	704.449
Knee_Time_MaxABD_Mom	5	85.121	99.259	93.648	2.799	6.258
Knee_MaxIR_Mom	5	104.475	1912.997	491.406	355.761	795.506
Knee_Time_MaxIR_Mom	5	69.396	97.337	82.435	4.571	10.221
Knee_MaxER_Mom	5	-860.440	-5.923	-192.481	167.138	373.731
Time_MaxER_Mom	5	66.670	98.366	83.990	7.091	15.856
Hip_Max_FLX_Mom	5	651.136	28152.000	6440.447	5428.904	12139.399
Hip_Time_MaxFLX_Mom	5	9.753	97.929	30.802	16.835	37.645
Hip_MaxEXT_Mom	5	-29523.750	-1202.567	-7059.193	5616.881	12559.727
Hip_Time_MaxEXT_Mom	5	80.033	94.898	85.506	2.526	5.649
Hip_MaxADD_Mom	5	649.998	11479.350	3032.287	2113.374	4725.648
Hip_Time_MaxADD_Mom	5	27.292	97.337	43.608	13.543	30.283
Hip_MaxABD_Mom	5	-10641.690	13.175	-2282.500	2090.867	4675.322
Hip_Time_MaxABD_Mom	5	11.865	95.795	74.572	15.877	35.502
Hip_MaxIR_Mom	5	84.148	1120.195	324.706	199.665	446.465
Hip_Time_MaxIR_Mom	5	71.822	96.728	82.629	4.458	9.969
Hip_MaxER_Mom	5	-5319.672	-13.797	-1120.218	1049.988	2347.844
Hip_Time_MaxER_Mom	5	11.919	100.000	35.923	16.410	36.693
Inv_A1_Joint_Power	5	-1.786	-0.701	-1.175	0.187	0.418
Inv_A2_Joint_Power	5	1.177	6.053	3.080	0.821	1.836
Inv_K1_Joint_Power	5	-0.922	-0.172	-0.484	0.151	0.338
Inv_K2_Joint_Power	5	0.321	1.001	0.592	0.127	0.284
Inv_K3_Joint_Power	5	-0.217	0.091	-0.066	0.050	0.111
Inv_K4_Joint_Power	5	-4.499	-1.069	-2.433	0.580	1.297
Inv_H1_Joint_Power	5	0.720	1.734	1.181	0.195	0.437
Inv_H2_Joint_Power	5	-0.643	-0.050	-0.454	0.104	0.233
Inv_Swing_Time	5	0.362	0.418	0.398	0.012	0.026
Uninv_Swing_Time	5	0.334	0.408	0.370	0.015	0.033
Inv_Stance_Time	5	0.539	0.818	0.665	0.048	0.108
Uninv_Stance_Time	5	0.570	0.844	0.695	0.046	0.103

Inv_Step_Length	5	0.734	0.926	0.794	0.035	0.077
Uninv_Step_Length	5	0.728	0.825	0.775	0.019	0.041
Stride_L	5	1.462	1.751	1.569	0.051	0.114
Velocity	5	1.220	1.810	1.494	0.101	0.225

One service member had discarded TAFO due to medical complications

Table A.4. Dynamic AFO (DAFO) Walking Descriptive Data at Time 0

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	6	1.970	11.699	6.779	1.296	3.175
Ank_Max_DF	6	6.403	17.100	12.625	1.507	3.690
Ank_Time_Max_ DF	6	72.920	84.528	77.804	1.754	4.297
Ank_Max_PF	6	0.105	7.068	3.238	1.017	2.491
Ank_Time_Max_PF	6	6.856	100.000	26.170	14.835	36.338
DF_exc	6	3.548	7.508	5.846	0.668	1.636
Ank_Po_TO	6	2.704	9.367	7.126	1.183	2.898
Ank_MaxDF_Vel	6	35.080	75.705	53.253	7.005	17.158
Ank_Time_MaxDF_Vel	6	17.274	95.430	40.287	12.385	30.337
Ank_MeanDF_Vel	6	7.060	15.765	11.837	1.294	3.170
Ank_MaxPF_Vel	6	-168.294	-85.081	-109.872	12.674	31.044
Ank_Time_MaxPF_Vel	6	0.734	95.836	42.566	18.918	46.339
Ank_IVEV_Po_IC	6	-4.960	29.116	6.271	5.468	13.393
Ank_MaxEV	6	-4.789	29.150	8.321	5.517	13.513
Ank_Time_MaxEV	6	33.573	100.000	59.571	10.548	25.838
Ank_MaxIV	6	-31.438	24.532	-2.570	8.218	20.130
Ank_Time_MaxIV	6	34.252	83.897	62.904	8.849	21.676
Ank_EV_exc	6	4.310	28.880	10.891	3.836	9.396
Ank_Po_TO	6	-11.813	28.644	5.541	6.348	15.550
Ank_MaxEV_Vel	6	43.531	324.069	132.668	43.251	105.944
Ank_Time_MaxEV_Vel	6	30.200	98.562	69.525	12.203	29.891
Ank_IRER_Po_IC	6	-15.035	8.887	-4.390	4.170	10.215
Ank_MaxER	6	-28.701	5.044	-12.205	4.748	11.630
Ank_Time_MaxER	6	7.677	100.000	58.772	14.361	35.178
Ank_MaxIR	6	-3.641	26.929	11.210	4.625	11.328
Ank_Time_MaxIR	6	50.713	85.351	73.068	5.770	14.135
IRER_Exc	6	4.441	55.630	23.415	7.163	17.546
Ank_IRER_Po_TO	6	-11.267	6.534	-5.068	2.575	6.308
FootProgAng_30percstance	6	-8.909	33.905	8.510	6.799	16.654

Mean_footprogang_stance	6	0.617	6.784	3.542	0.875	2.144
Ank_MaxER_Vel	6	-1003.317	-70.482	-354.153	136.491	334.332
Ank_Time_MaxER	6	65.009	100.000	82.365	6.871	16.831
Knee_Po_IC	6	-0.661	18.750	7.786	3.688	9.035
Knee_Max_FLX	6	28.002	48.669	39.652	2.772	6.789
Knee_Time_Max_FLX	6	76.101	100.000	96.017	3.983	9.757
Knee_FLX_Exc	6	19.706	42.476	31.865	4.051	9.924
Knee_Po_TO	6	19.874	48.669	38.297	3.979	9.746
Knee_Max_FLX_Vel	6	229.371	488.639	402.778	38.045	93.190
Knee_Time_Max_FLX_Vel	6	64.926	96.423	90.115	5.054	12.379
Knee_Mean_FLX_Vel	6	29.907	91.067	60.006	9.216	22.573
Knee_Varval_Po_IC	6	-5.999	6.604	0.088	1.806	4.423
Knee_Max_VAR	6	-1.125	15.418	4.831	2.681	6.568
Knee_Time_Max_VAR	6	14.798	100.000	53.256	13.429	32.895
Knee_VAR_Exc	6	0.347	13.951	4.743	1.975	4.839
Knee_Po_TO	6	-5.997	13.437	1.305	2.688	6.584
Knee_Max_VAR_Vel	6	69.309	470.170	226.512	58.812	144.059
Knee_Time_Max_VAR_Vel	6	72.327	100.000	95.084	4.555	11.158
Knee_Mean_VAR_Vel	6	-19.348	35.592	14.075	8.363	20.484
Knee_IRER_Po_IC	6	-0.568	10.620	4.187	1.807	4.426
Knee_MaxIR	6	2.466	14.818	8.775	1.672	4.096
Knee_Time_MaxIR	6	4.626	90.490	52.448	14.893	36.480
Knee_IR_Exc	6	1.165	7.540	4.588	0.932	2.284
KneeIR_Po_TO	6	-4.727	14.295	2.327	2.607	6.387
Knee_MaxIRER_Vel	6	23.353	434.738	166.824	62.907	154.089
Knee_Time_Max_Rot_Vel	6	9.543	65.053	38.411	9.013	22.078
Knee_MeanIRER_Vel	6	4.568	84.779	30.187	11.753	28.789
Hip_FLXEXT_Po_IC	6	34.813	50.155	41.142	2.384	5.839
Hip_Max_FLX	6	34.813	50.155	41.293	2.364	5.791
Hip_Time_MaxFLX	6	0.606	9.740	2.204	1.507	3.693
Hip_Max_EXT	6	-18.446	-2.963	-10.841	2.819	6.906
Hip_Time_MaxEXT	6	82.634	91.341	86.033	1.271	3.112
Hip_ Exc_ICtomaxEXT	6	-57.644	-43.642	-51.982	2.064	5.055
Hip_FLXEXT_Po_TO	6	-12.531	5.637	-3.790	2.933	7.185
Hip_MaxEXT_Vel	6	-196.390	185.106	-61.574	71.114	174.193
Hip_Time_MaxEXT_Vel	6	28.930	100.000	56.652	13.067	32.008
Hip_MeanEXT_Vel_ICtoMaxEXT	6	-117.121	-75.004	-102.323	6.502	15.926
HipABAD_Po_IC	6	-5.896	2.587	-1.601	1.461	3.579

Hip_MaxADD	6	4.095	12.934	6.931	1.326	3.248
Hip_Time_MaxADD	6	26.306	34.646	30.291	1.383	3.388
Hip_MaxABD	6	-9.957	-1.579	-5.971	1.091	2.673
Hip_Time_MaxABD	6	0.713	100.000	66.056	20.676	50.645
Hip_Exc_ICtoMaxABD	6	-12.543	0.000	-4.370	1.891	4.631
Hip_ABAD_Po_TO	6	-9.957	-1.579	-5.200	1.168	2.861
Hip_MaxADD_Vel	6	45.583	158.370	97.153	16.593	40.645
Hip_Time_MaxADD_Vel	6	1.966	66.761	20.751	9.587	23.484
Hip_Mean_Vel_ICtoMaxABD	6	-21.927	151.616	28.638	27.537	67.451
HipIRER_Po_IC	6	-35.288	3.791	-14.413	5.588	13.687
Hip_MaxIR	6	-12.844	15.584	2.414	3.901	9.556
Hip_Time_MaxIR	6	7.888	100.000	70.009	15.519	38.015
Hip_MaxER	6	-35.345	-8.190	-20.974	4.103	10.050
Hip_Time_MaxER	6	3.802	91.421	47.091	17.534	42.950
Hip_Exc_ICtoMaxIR	6	3.162	26.341	16.827	3.857	9.448
Hip_Exc_MaxIRtoMaxER	6	-33.181	-15.143	-23.388	2.955	7.237
HipIRER_Po_TO	6	-12.844	11.201	-1.008	3.849	9.429
Hip_MaxIR_Vel	6	223.618	669.674	362.405	71.216	174.443
Hip_Time_MaxIR_Vel	6	2.023	100.000	63.246	18.585	45.523
Hip_Mean_Vel_ICtoMaxIR	6	11.240	146.616	68.251	23.393	57.302
Hip_MaxER_Vel	6	-345.537	-130.471	-228.673	35.567	87.120
Hip_Time_MaxER_Vel	6	5.934	60.953	22.744	8.149	19.960
Max_Pelv_Ang	6	1.339	6.725	3.984	0.710	1.739
Pelv_Ang_Exc	6	6.126	18.799	11.038	1.774	4.346
Max_thor_angle	6	1.371	6.302	3.573	0.814	1.994
Thor_ang_exc	6	4.966	6.837	5.967	0.334	0.817
Max_spine_ang	6	2.054	16.901	8.172	2.302	5.638
Spine_ang_exc	6	11.265	26.236	16.379	2.172	5.320
MaxGRFz	6	9.847	13.469	11.636	0.482	1.182
Time_MaxGRFz	6	17.360	38.154	24.637	3.097	7.585
Load_Rate	6	4306.128	11852.459	8378.409	1050.434	2573.027
GRFimp	6	304.483	596.836	409.378	42.863	104.991
GRFimp_norm	6	0.388	0.532	0.465	0.019	0.047
Max_brake_GRFy	6	-2.985	-1.190	-2.034	0.273	0.668
Time_max_brake_GRFy	6	15.135	24.350	19.035	1.476	3.617
Max_prop_GRFy	6	1.310	2.389	1.864	0.182	0.446
Time_max_prop_GRFy	6	60.230	89.730	83.626	4.704	11.523
Max_Abs_FM	6	0.003	0.005	0.004	0.000	0.001
Max_ADD_FM	6	0.003	0.005	0.004	0.000	0.001
Time_max_ADD_FM	6	20.605	83.107	46.877	11.218	27.480

Max_ABD_FM	6	-0.003	0.000	-0.002	0.000	0.001
Time_Max_ABD_FM	6	7.706	61.665	23.788	8.747	21.426
FM_Peak_Brake_GRFy	6	-0.001	0.003	0.000	0.001	0.001
ADD_FM_imp	6	0.000	0.001	0.001	0.000	0.000
Max_DF_Mom	6	1079.500	1730.143	1359.940	109.593	268.448
Time_Max_DF_Mom	6	78.360	83.823	81.893	0.820	2.009
Max_PF_Mom	6	-450.152	-216.233	-378.618	36.755	90.031
Time_Max_PF_Mom	6	12.097	19.057	14.703	1.054	2.582
MaxIV_Mom	6	52.702	418.142	234.441	55.021	134.772
Time_MaxIV_Mom	6	12.496	62.188	25.619	7.447	18.242
MaxEV_Mom	6	-744.715	-0.624	-286.703	132.507	324.575
Time_MaxEV_Mom	6	2.243	81.414	56.601	12.938	31.693
Ank_MaxIR_Mom	6	78.226	318.494	169.300	34.096	83.517
Time_MaxIR_Mom	6	65.068	81.879	74.261	2.812	6.887
MaxER_Mom	6	-90.860	-4.108	-34.373	13.141	32.189
Time_MaxER_Mom	6	0.797	100.000	38.086	19.574	47.946
Max_FLX_Mom_LR	6	313.019	1165.323	646.296	119.057	291.628
Time_Max_FLX_Mom_LR	6	-38.276	21.439	0.476	10.211	25.012
Max_FLX_Mom_TO	6	506.337	1222.487	826.013	125.374	307.102
Time_Max_FLX_Mom_TO	6	17.505	91.197	47.058	14.252	34.909
Max_EXT_Mom	6	-419.563	-337.991	-365.202	12.845	31.463
Time_Max_EXT_Mom	6	4.826	76.460	41.521	12.335	30.216
Knee_ Stiff	6	2.259	14.137	5.734	1.769	4.334
Knee_MaxADD_Mom	6	206.143	899.058	663.619	98.795	241.998
Knee_Timing_MaxADD_Mom	6	19.895	28.137	23.624	1.210	2.965
Knee_MaxABD_Mom	6	-126.970	-28.114	-74.768	18.261	44.730
Knee_Time_MaxABD_Mom	6	6.405	98.519	55.445	16.395	40.159
Knee_MaxIR_Mom	6	68.229	227.795	137.190	25.909	63.465
Knee_Time_MaxIR_Mom	6	65.491	82.604	74.990	2.711	6.642
Knee_MaxER_Mom	6	-49.508	-7.302	-25.853	6.092	14.922
Time_MaxER_Mom	6	6.048	69.960	29.606	10.494	25.705
Hip_Max_FLX_Mom	6	655.056	2155.337	1403.722	202.123	495.098
Hip_Time_MaxFLX_Mom	6	8.724	14.564	11.932	0.887	2.172
Hip_MaxEXT_Mom	6	-2416.280	-1337.303	-1830.227	155.405	380.663
Hip_Time_MaxEXT_Mom	6	56.876	86.686	80.228	4.692	11.493
Hip_MaxADD_Mom	6	683.585	1294.343	1004.301	85.519	209.477
Hip_Time_MaxADD_Mom	6	25.665	42.670	29.116	2.733	6.694
Hip_MaxABD_Mom	6	-718.208	29.608	-249.393	107.669	263.734
Hip_Time_MaxABD_Mom	6	2.283	94.869	54.287	16.180	39.632

Hip_MaxIR_Mom	6	57.136	181.483	125.491	20.127	49.302
Hip_Time_MaxIR_Mom	6	5.817	82.838	64.736	11.948	29.267
Hip_MaxER_Mom	6	-265.994	-73.617	-137.127	30.932	75.768
Hip_Time_MaxER_Mom	6	20.590	74.905	31.389	8.740	21.409
Inv_A1_Joint_Power	6	-1.213	-0.109	-0.638	0.175	0.428
Inv_A2_Joint_Power	6	0.768	3.124	1.768	0.330	0.808
Inv_K1_Joint_Power	6	-1.613	-0.075	-0.845	0.257	0.630
Inv_K2_Joint_Power	6	0.390	2.763	1.383	0.341	0.835
Inv_K3_Joint_Power	6	-0.427	0.015	-0.200	0.063	0.154
Inv_K4_Joint_Power	6	-6.663	-1.516	-3.106	0.813	1.990
Inv_H1_Joint_Power	6	0.550	2.744	1.858	0.332	0.813
Inv_H2_Joint_Power	6	-0.897	0.394	-0.473	0.195	0.476
Inv_Swing_Time	6	0.383	0.458	0.423	0.012	0.028
Uninv_Swing_Time	6	0.368	0.424	0.392	0.009	0.021
Inv_Stance_Time	6	0.557	0.673	0.602	0.022	0.053
Uninv_Stance_Time	6	0.580	0.738	0.634	0.027	0.066
Inv_Step_Length	6	0.684	1.048	0.909	0.054	0.132
Uninv_Step_Length	6	0.665	1.002	0.884	0.050	0.121
Stride_L	6	1.348	2.050	1.794	0.103	0.253
Velocity	6	1.190	2.101	1.775	0.137	0.334

Table A.5. Dynamic AFO (DAFO) Walking Descriptive Data at Last Visit

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	4	2.749	6.632	4.982	0.904	1.807
Ank_Max_DF	4	11.016	14.712	12.679	0.762	1.524
Ank_Time_Max_ DF	4	73.568	79.556	77.292	1.421	2.842
Ank_Max_PF	4	-1.918	3.900	1.040	1.352	2.704
Ank_Time_Max_PF	4	9.129	70.602	25.723	14.973	29.947
DF_exc	4	4.763	9.788	7.696	1.054	2.108
Ank_Po_TO	4	3.948	5.581	4.798	0.361	0.722
Ank_MaxDF_Vel	4	61.770	88.706	69.771	6.371	12.743
Ank_Time_MaxDF_Vel	4	19.743	70.150	33.673	12.210	24.420
Ank_MeanDF_Vel	4	9.976	18.787	16.289	2.112	4.224
Ank_MaxPF_Vel	4	-177.230	-90.348	-132.658	19.336	38.672
Ank_Time_MaxPF_Vel	4	0.988	93.058	47.311	26.389	52.778
Ank_IVEV_Po_IC	4	-2.472	15.708	10.261	4.283	8.566
Ank_MaxEV	4	1.187	21.674	14.097	4.661	9.322
Ank_Time_MaxEV	4	24.690	100.000	55.418	15.971	31.942

Ank_MaxIV	4	-11.978	14.700	3.151	5.675	11.349
Ank_Time_MaxIV	4	31.219	79.465	64.957	11.309	22.619
Ank_EV_exc	4	5.408	19.709	10.946	3.440	6.881
Ank_Po_TO	4	-1.119	21.674	12.835	5.000	10.000
Ank_MaxEV_Vel	4	44.740	331.147	137.843	66.516	133.032
Ank_Time_MaxEV_Vel	4	11.475	94.928	71.335	20.013	40.026
Ank_IRER_Po_IC	4	-15.731	10.528	-4.844	5.534	11.069
Ank_MaxER	4	-26.507	-2.938	-16.132	4.938	9.875
Ank_Time_MaxER	4	21.980	100.000	66.160	19.988	39.977
Ank_MaxIR	4	-5.898	23.600	7.576	6.066	12.133
Ank_Time_MaxIR	4	2.581	79.736	58.057	18.527	37.053
IRER_Exc	4	13.498	32.944	23.708	4.093	8.185
Ank_IRER_Po_TO	4	-21.538	-2.938	-14.398	4.161	8.322
FootProgAng_30percstance	4	-14.128	16.429	1.923	6.792	13.585
Mean_footprogang_stance	4	-2.405	5.350	1.985	1.614	3.228
Ank_MaxER_Vel	4	-591.204	-223.404	-358.474	84.191	168.383
Ank_Time_MaxER	4	95.336	98.829	96.695	0.808	1.616
Knee_Po_IC	4	0.842	22.607	9.213	4.819	9.637
Knee_Max_FLX	4	37.055	51.885	42.536	3.288	6.575
Knee_Time_Max_FLX	4	100.000	100.000	100.000	0.000	0.000
Knee_FLX_Exc	4	14.449	48.060	33.323	7.370	14.741
Knee_Po_TO	4	37.055	51.885	42.536	3.288	6.575
Knee_Max_FLX_Vel	4	384.723	608.698	468.321	50.546	101.092
Knee_Time_Max_FLX_Vel	4	95.115	99.259	96.520	0.943	1.886
Knee_Mean_FLX_Vel	4	25.613	93.433	58.900	14.222	28.445
Knee_Varval_Po_IC	4	-3.752	12.330	4.560	3.583	7.165
Knee_Max_VAR	4	-3.070	19.557	8.669	5.262	10.524
Knee_Time_Max_VAR	4	15.332	91.204	47.763	18.185	36.371
Knee_VAR_Exc	4	0.682	7.227	4.109	1.722	3.444
Knee_Po_TO	4	-3.419	4.791	0.926	2.131	4.263
Knee_Max_VAR_Vel	4	123.136	518.921	267.924	88.933	177.867
Knee_Time_Max_VAR_Vel	4	14.073	100.000	70.500	20.201	40.403
Knee_Mean_VAR_Vel	4	0.893	75.445	34.995	19.114	38.227
Knee_IRER_Po_IC	4	-14.435	8.564	-0.794	4.877	9.753
Knee_MaxIR	4	6.730	14.799	10.803	2.141	4.282
Knee_Time_MaxIR	4	21.869	99.502	69.207	18.544	37.088
Knee_IR_Exc	4	4.252	21.926	11.597	4.135	8.270
KneeIR_Po_TO	4	1.669	14.731	7.089	2.793	5.585

Knee_MaxIRER_Vel	4	91.908	233.781	148.421	31.275	62.550
Knee_Time_Max_Rot_Vel	4	3.528	33.363	14.016	6.801	13.603
Knee_MeanIRER_Vel	4	19.891	47.593	32.406	5.957	11.915
Hip_FLXEXT_Po_IC	4	28.811	52.681	41.732	5.010	10.021
Hip_Max_FLX	4	28.886	52.681	41.764	4.993	9.986
Hip_Time_MaxFLX	4	0.667	3.665	1.792	0.705	1.411
Hip_Max_EXT	4	-21.574	-11.019	-16.091	2.384	4.769
Hip_Time_MaxEXT	4	81.118	86.677	83.178	1.245	2.490
Hip_ Exc_ICtomaxEXT	4	-74.254	-47.182	-57.823	5.798	11.596
Hip_FLXEXT_Po_TO	4	-14.135	-2.194	-6.816	2.739	5.477
Hip_MaxEXT_Vel	4	-223.550	187.299	-14.536	105.320	210.641
Hip_Time_MaxEXT_Vel	4	31.442	100.000	67.404	18.870	37.739
Hip_MeanEXT_Vel_ICtoMaxEXT	4	-154.302	-97.362	-122.061	12.104	24.207
HipABAD_Po_IC	4	-16.952	-1.210	-7.301	3.476	6.953
Hip_MaxADD	4	3.126	9.572	5.638	1.547	3.094
Hip_Time_MaxADD	4	33.071	61.312	40.218	7.031	14.063
Hip_MaxABD	4	-16.952	-3.953	-9.828	2.681	5.362
Hip_Time_MaxABD	4	0.741	100.000	51.634	27.943	55.887
Hip_Exc_ICtoMaxABD	4	-6.052	0.000	-2.526	1.302	2.604
Hip_ABAD_Po_TO	4	-9.516	-1.483	-5.261	1.703	3.405
Hip_MaxADD_Vel	4	72.062	276.426	157.121	45.226	90.453
Hip_Time_MaxADD_Vel	4	4.690	72.212	22.645	16.548	33.097
Hip_Mean_Vel_ICtoMaxABD	4	-11.239	166.526	35.523	43.695	87.390
HipIRER_Po_IC	4	-33.905	11.903	-7.826	10.242	20.483
Hip_MaxIR	4	-11.676	21.255	4.921	7.349	14.697
Hip_Time_MaxIR	4	8.452	100.000	50.307	21.589	43.178
Hip_MaxER	4	-43.332	-5.006	-18.973	8.400	16.799
Hip_Time_MaxER	4	3.982	91.612	50.273	21.806	43.613
Hip_Exc_ICtoMaxIR	4	7.776	22.229	12.747	3.258	6.516
Hip_Exc_MaxIRtoMaxER	4	-31.655	-13.015	-23.894	3.924	7.848
HipIRER_Po_TO	4	-11.676	-0.509	-5.204	2.387	4.773
Hip_MaxIR_Vel	4	193.630	778.861	415.943	134.520	269.040
Hip_Time_MaxIR_Vel	4	2.453	98.660	39.173	21.479	42.958
Hip_Mean_Vel_ICtoMaxIR	4	32.128	152.885	75.398	27.227	54.455
Hip_MaxER_Vel	4	-405.903	-96.246	-256.091	64.495	128.990
Hip_Time_MaxER_Vel	4	17.334	38.624	25.275	4.984	9.967
Max_Pelv_Ang	4	2.333	7.633	4.999	1.098	2.196
Pelv_Ang_Exc	4	8.259	12.971	11.037	0.994	1.988
Max_thor_angle	4	1.338	6.355	3.840	1.080	2.160

Thor_ang_exc	4	4.620	7.526	6.298	0.623	1.247
Max_spine_ang	4	3.931	12.460	6.504	2.002	4.004
Spine_ang_exc	4	11.413	20.365	15.278	1.873	3.746
MaxGRFz	4	11.461	14.632	13.078	0.715	1.429
Time_MaxGRFz	4	14.894	29.980	19.108	3.632	7.264
Load_Rate	4	5198.543	16273.993	12248.454	2479.007	4958.013
GRFimp	4	327.105	588.893	400.369	63.170	126.340
GRFimp_norm	4	0.404	0.492	0.449	0.018	0.036
Max_brake_GRFy	4	-3.389	-2.013	-2.761	0.317	0.633
Time_max_brake_GRFy	4	14.002	27.632	19.088	3.116	6.232
Max_prop_GRFy	4	1.478	2.655	2.131	0.252	0.503
Time_max_prop_GRFy	4	85.677	89.868	88.140	0.964	1.927
Max_Abs_FM	4	0.003	0.006	0.004	0.001	0.002
Max_ADD_FM	4	0.003	0.006	0.004	0.001	0.002
Time_max_ADD_FM	4	16.891	76.647	43.929	15.257	30.513
Max_ABD_FM	4	-0.003	-0.001	-0.001	0.000	0.001
Time_Max_ABD_FM	4	6.668	44.860	28.252	8.502	17.004
FM_Peak_Brake_GRFy	4	-0.001	0.006	0.002	0.002	0.003
ADD_FM_imp	4	0.001	0.001	0.001	0.000	0.000
Max_DF_Mom	4	1357.310	1746.920	1561.772	83.565	167.131
Time_Max_DF_Mom	4	81.034	82.888	81.938	0.399	0.799
Max_PF_Mom	4	-595.081	-333.367	-485.736	55.193	110.386
Time_Max_PF_Mom	4	11.110	14.415	12.768	0.676	1.352
MaxIV_Mom	4	13.847	386.655	222.859	77.593	155.186
Time_MaxIV_Mom	4	16.389	83.334	34.367	16.335	32.670
MaxEV_Mom	4	-502.788	-16.955	-207.618	106.234	212.468
Time_MaxEV_Mom	4	4.002	80.538	55.602	17.979	35.957
Ank_MaxIR_Mom	4	37.242	249.233	162.705	47.872	95.743
Time_MaxIR_Mom	4	35.909	86.127	69.568	11.377	22.755
MaxER_Mom	4	-113.781	-7.411	-45.247	24.516	49.032
Time_MaxER_Mom	4	16.005	100.000	66.923	20.200	40.400
Max_FLX_Mom_LR	4	605.452	1591.120	973.001	230.052	460.104
Time_Max_FLX_Mom_LR	4	17.788	25.285	20.494	1.674	3.348
Max_FLX_Mom_TO	4	647.749	1591.120	995.996	218.521	437.043
Time_Max_FLX_Mom_TO	4	17.788	88.055	48.147	17.745	35.491
Max_EXT_Mom	4	-422.491	-287.538	-347.588	30.807	61.614
Time_Max_EXT_Mom	4	3.999	75.312	22.492	17.611	35.221
Knee_ Stiff	4	4.038	18.629	10.177	3.053	6.105
Knee_MaxADD_Mom	4	333.669	1349.417	847.819	240.078	480.156
Knee_Timing_MaxADD_Mom	4	19.559	24.814	22.123	1.074	2.148

Knee_MaxABD_Mom	4	-382.577	-1.098	-132.166	89.799	179.599
Knee_Time_MaxABD_Mom	4	9.605	100.000	54.480	23.718	47.436
Knee_MaxIR_Mom	4	53.164	242.400	165.573	39.975	79.950
Knee_Time_MaxIR_Mom	4	68.002	83.726	77.518	3.349	6.699
Knee_MaxER_Mom	4	-47.485	-3.590	-27.082	10.620	21.241
Time_MaxER_Mom	4	13.112	100.000	48.557	18.332	36.664
Hip_Max_FLX_Mom	4	715.758	2198.763	1323.318	312.589	625.178
Hip_Time_MaxFLX_Mom	4	7.202	10.894	9.575	0.820	1.640
Hip_MaxEXT_Mom	4	-2389.490	-1673.190	-2013.348	153.476	306.952
Hip_Time_MaxEXT_Mom	4	83.735	86.178	85.010	0.500	1.000
Hip_MaxADD_Mom	4	880.045	1062.331	996.965	41.761	83.522
Hip_Time_MaxADD_Mom	4	23.334	28.090	25.032	1.074	2.148
Hip_MaxABD_Mom	4	-1417.580	11.229	-524.261	309.568	619.136
Hip_Time_MaxABD_Mom	4	9.334	87.188	42.828	19.317	38.635
Hip_MaxIR_Mom	4	74.018	244.342	166.954	35.544	71.089
Hip_Time_MaxIR_Mom	4	7.030	88.797	62.631	18.731	37.462
Hip_MaxER_Mom	4	-227.611	-100.139	-169.958	32.062	64.124
Hip_Time_MaxER_Mom	4	20.240	25.756	22.278	1.204	2.407
Inv_A1_Joint_Power	4	-1.440	-0.344	-0.833	0.230	0.461
Inv_A2_Joint_Power	4	2.370	3.607	2.700	0.303	0.605
Inv_K1_Joint_Power	4	-3.918	-0.508	-2.095	0.784	1.568
Inv_K2_Joint_Power	4	0.706	4.857	2.500	0.915	1.829
Inv_K3_Joint_Power	4	-0.169	0.032	-0.077	0.047	0.093
Inv_K4_Joint_Power	4	-6.975	-2.239	-4.292	1.013	2.026
Inv_H1_Joint_Power	4	1.326	2.468	1.694	0.262	0.525
Inv_H2_Joint_Power	4	-1.049	0.204	-0.414	0.256	0.512
Inv_Swing_Time	4	0.320	0.425	0.381	0.024	0.048
Uninv_Swing_Time	4	0.344	0.399	0.372	0.012	0.024
Inv_Stance_Time	4	0.500	0.606	0.564	0.023	0.046
Uninv_Stance_Time	4	0.558	0.673	0.617	0.025	0.050
Inv_Step_Length	4	0.860	1.012	0.945	0.037	0.073
Uninv_Step_Length	4	0.842	0.998	0.917	0.035	0.070
Stride_L	4	1.701	2.011	1.862	0.071	0.143
Velocity	4	1.719	2.297	1.955	0.134	0.267

Three service members completed at time 6, one at time 5 due to medical retirement, one not included due to disenrollment at month two, and one pending completion April 2013.

Table A.6. Control Running Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	6	-11.248	15.611	6.944	3.854	9.439
Ank_Max_DF	6	24.142	36.140	30.506	2.155	5.278
Ank_Time_Max_ DF	6	44.371	50.221	48.092	0.961	2.354
Ank_Max_PF	6	-34.660	-18.825	-27.976	2.237	5.479
Ank_Time_Max_PF	6	100.000	100.000	100.000	0.000	0.000
DF_exc	6	16.744	36.991	23.562	2.908	7.124
Ank_Po_TO	6	-34.660	-18.825	-27.976	2.237	5.479
Ank_MaxDF_Vel	6	326.323	613.223	405.593	43.909	107.555
Ank_Time_MaxDF_Vel	6	10.046	24.854	20.656	2.412	5.909
Ank_MeanDF_Vel	6	130.494	369.660	200.003	36.757	90.035
Ank_MaxPF_Vel	6	-859.601	-617.677	-759.906	35.839	87.787
Ank_Time_MaxPF_Vel	6	78.644	82.765	80.768	0.691	1.693
Ank_IVEV_Po_IC	6	-4.745	22.917	7.109	4.199	10.286
Ank_MaxEV	6	18.889	62.420	37.632	6.304	15.441
Ank_Time_MaxEV	6	36.685	49.966	44.067	1.804	4.419
Ank_MaxIV	6	-22.307	22.917	-0.059	6.172	15.119
Ank_Time_MaxIV	6	1.686	100.000	59.331	18.795	46.038
Ank_EV_exc	6	23.442	49.939	37.691	3.456	8.466
Ank_Po_TO	6	-22.307	39.891	5.636	8.989	22.019
Ank_MaxEV_Vel	6	271.753	816.205	507.742	76.157	186.546
Ank_Time_MaxEV_Vel	6	9.461	21.356	14.605	1.699	4.161
Ank_IRER_Po_IC	6	-33.091	7.570	-10.007	6.494	15.906
Ank_MaxER	6	-45.803	-20.280	-28.063	3.894	9.538
Ank_Time_MaxER	6	10.046	40.678	29.695	4.465	10.938
Ank_MaxIR	6	-23.000	7.570	-2.133	4.701	11.515
Ank_Time_MaxIR	6	1.775	96.679	76.164	14.962	36.649
IRER_Exc	6	21.146	30.836	25.930	1.548	3.791
Ank_IRER_Po_TO	6	-28.136	5.745	-5.204	5.186	12.703
FootProgAng_30percstance	6	-14.902	-4.496	-10.663	1.875	4.594
Mean_footprogang_stance	6	4.030	11.610	8.118	1.143	2.800
Ank_MaxER_Vel	6	-616.135	-340.242	-463.796	46.844	114.743
Ank_Time_MaxER	6	5.546	100.000	32.349	14.276	34.969
Knee_Po_IC	6	10.088	22.268	15.616	1.696	4.154
Knee_Max_FLX	6	39.759	52.670	44.231	1.967	4.818
Knee_Time_Max_FLX	6	31.489	37.865	35.281	0.934	2.288

Knee_FLX_Exc	6	24.392	33.790	28.615	1.488	3.645
Knee_Po_TO	6	2.263	16.882	11.248	2.520	6.173
Knee_Max_FLX_Vel	6	496.739	616.101	567.921	21.223	51.986
Knee_Time_Max_FLX_Vel	6	12.416	19.661	15.134	1.051	2.575
Knee_Mean_FLX_Vel	6	293.758	376.579	341.411	11.908	29.170
Knee_Varval_Po_IC	6	-5.352	15.299	5.028	2.690	6.589
Knee_Max_VAR	6	5.486	19.873	9.309	2.169	5.313
Knee_Time_Max_VAR	6	4.114	73.451	32.812	9.606	23.530
Knee_VAR_Exc	6	0.443	10.838	4.280	1.426	3.493
Knee_Po_TO	6	-5.088	12.261	3.973	2.408	5.898
Knee_Max_VAR_Vel	6	120.585	266.937	160.594	22.779	55.798
Knee_Time_Max_VAR_Vel	6	5.066	49.897	27.740	5.858	14.348
Knee_Mean_VAR_Vel	6	20.825	129.388	60.797	17.799	43.598
Knee_IRER_Po_IC	6	-6.585	11.905	2.284	3.052	7.475
Knee_MaxIR	6	18.840	33.699	25.426	2.486	6.089
Knee_Time_MaxIR	6	38.269	46.430	43.138	1.165	2.853
Knee_IR_Exc	6	12.309	34.494	23.141	3.210	7.863
KneeIR_Po_TO	6	-3.108	18.132	3.404	3.140	7.691
Knee_MaxIRER_Vel	6	243.139	684.285	463.761	80.635	197.516
Knee_Time_Max_Rot_Vel	6	8.271	21.723	14.076	1.981	4.853
Knee_MeanIRER_Vel	6	120.234	349.640	226.429	35.147	86.093
Hip_FLXEXT_Po_IC	6	32.124	51.483	42.740	3.088	7.563
Hip_Max_FLX	6	32.124	52.913	43.684	3.193	7.821
Hip_Time_MaxFLX	6	1.685	25.342	14.819	4.271	10.461
Hip_Max_EXT	6	-17.747	-2.467	-8.968	2.406	5.893
Hip_Time_MaxEXT	6	93.831	100.000	97.274	1.231	3.015
Hip_ Exc_ICtomaxEXT	6	-54.490	-49.871	-51.708	0.764	1.872
Hip_FLXEXT_Po_TO	6	-17.371	-1.845	-8.668	2.373	5.814
Hip_MaxEXT_Vel	6	-483.913	40.133	-295.867	93.697	229.509
Hip_Time_MaxEXT_Vel	6	58.832	100.000	74.291	8.212	20.116
Hip_MeanEXT_Vel_ICtoMaxEXT	6	-226.741	-195.857	-218.677	4.655	11.402
HipABAD_Po_IC	6	-9.087	8.053	0.529	2.462	6.032
Hip_MaxADD	6	5.171	12.802	8.129	1.104	2.703
Hip_Time_MaxADD	6	26.023	37.646	32.396	1.556	3.812
Hip_MaxABD	6	-9.087	-0.890	-5.292	1.160	2.841
Hip_Time_MaxABD	6	1.685	100.000	69.503	15.896	38.937
Hip_Exc_ICtoMaxABD	6	-10.845	0.000	-5.822	1.920	4.702
Hip_ABAD_Po_TO	6	-8.013	0.873	-4.267	1.417	3.470

Hip_MaxADD_Vel	6	105.616	281.666	163.647	27.391	67.095
Hip_Time_MaxADD_Vel	6	4.731	22.034	16.552	2.558	6.267
Hip_Mean_Vel_ICtoMaxABD	6	-44.984	26.736	-16.961	13.652	33.439
HipIRER_Po_IC	6	-13.022	26.575	0.131	5.723	14.018
Hip_MaxIR	6	-5.754	30.457	7.068	5.054	12.379
Hip_Time_MaxIR	6	3.530	72.776	32.001	9.842	24.108
Hip_MaxER	6	-21.956	9.105	-10.030	4.187	10.255
Hip_Time_MaxER	6	4.377	100.000	69.416	18.513	45.348
Hip_Exc_ICtoMaxIR	6	0.249	16.437	6.937	2.282	5.591
Hip_Exc_MaxIRtoMaxER	6	-21.353	-12.268	-17.098	1.392	3.409
HipIRER_Po_TO	6	-21.789	9.105	-7.454	4.215	10.326
Hip_MaxIR_Vel	6	92.743	339.943	242.636	35.565	87.116
Hip_Time_MaxIR_Vel	6	1.686	26.775	14.070	5.432	13.305
Hip_Mean_Vel_ICtoMaxIR	6	-37.529	157.225	81.888	28.772	70.476
Hip_MaxER_Vel	6	-314.392	-182.687	-244.466	22.273	54.558
Hip_Time_MaxER_Vel	6	22.389	59.157	40.052	5.038	12.340
Max_Pelv_Ang	6	3.529	7.682	5.806	0.623	1.525
Pelv_Ang_Exc	6	4.966	10.461	8.612	0.786	1.924
Max_thor_angle	6	2.411	5.527	3.669	0.519	1.272
Thor_ang_exc	6	4.988	8.891	6.713	0.623	1.526
Max_spine_ang	6	5.940	8.688	7.027	0.379	0.929
Spine_ang_exc	6	13.017	23.463	17.506	1.688	4.135
MaxGRFz	6	21.517	25.372	24.113	0.604	1.479
Time_MaxGRFz	6	42.648	50.000	47.411	1.094	2.679
Load_Rate	6	14848.264	23051.777	17859.295	1219.029	2986.000
GRFimp	6	230.348	371.959	297.880	22.810	55.872
GRFimp_norm	6	0.331	0.389	0.352	0.008	0.020
Max_brake_GRFy	6	-3.656	-3.156	-3.427	0.075	0.184
Time_max_brake_GRFy	6	24.269	32.589	29.930	1.271	3.114
Max_prop_GRFy	6	2.863	4.303	3.434	0.216	0.529
Time_max_prop_GRFy	6	75.741	79.292	77.668	0.502	1.231
Max_Abs_FM	6	0.003	0.008	0.005	0.001	0.002
Max_ADD_FM	6	0.003	0.008	0.005	0.001	0.002
Time_max_ADD_FM	6	24.259	55.057	40.676	4.697	11.506
Max_ABD_FM	6	-0.004	0.000	-0.001	0.001	0.001
Time_Max_ABD_FM	6	5.704	100.000	43.741	13.952	34.175
FM_Peak_Brake_GRFy	6	0.000	0.007	0.003	0.001	0.002
ADD_FM_imp	6	0.000	0.001	0.000	0.000	0.000
Max_DF_Mom	6	2339.423	2628.230	2507.141	47.280	115.812
Time_Max_DF_Mom	6	57.989	59.785	58.874	0.279	0.684

Max_PF_Mom	6	-934.952	-43.637	-410.393	125.949	308.512
Time_Max_PF_Mom	6	1.775	15.172	11.361	2.108	5.164
MaxIV_Mom	6	17.632	370.814	185.096	69.838	171.067
Time_MaxIV_Mom	6	54.438	97.185	72.697	6.716	16.450
MaxEV_Mom	6	-356.917	-77.955	-177.783	46.270	113.339
Time_MaxEV_Mom	6	18.544	30.348	22.302	1.856	4.547
Ank_MaxIR_Mom	6	417.913	640.544	518.618	33.477	82.002
Time_MaxIR_Mom	6	49.708	57.435	52.305	1.155	2.829
MaxER_Mom	6	-66.788	-6.140	-34.836	7.982	19.551
Time_MaxER_Mom	6	39.275	100.000	72.010	9.980	24.445
Max_FLX_Mom_LR	6	2618.670	3808.813	3138.151	211.447	517.938
Time_Max_FLX_Mom_LR	6	30.196	41.416	36.099	2.076	5.084
Max_FLX_Mom_TO	6	2618.670	3815.200	3139.216	212.124	519.596
Time_Max_FLX_Mom_TO	6	30.196	42.011	36.198	2.128	5.213
Max_EXT_Mom	6	-689.682	-395.033	-578.623	47.964	117.488
Time_Max_EXT_Mom	6	1.436	59.002	16.147	9.828	24.073
Knee_ Stiff	6	8.879	20.894	12.185	1.794	4.394
Knee_MaxADD_Mom	6	2102.190	3250.700	2458.666	172.055	421.446
Knee_Timing_MaxADD_Mom	6	36.685	45.562	39.478	1.414	3.463
Knee_MaxABD_Mom	6	-291.516	-150.484	-229.752	23.552	57.689
Knee_Time_MaxABD_Mom	6	1.685	97.759	70.241	17.102	41.890
Knee_MaxIR_Mom	6	113.306	444.090	272.475	49.310	120.784
Knee_Time_MaxIR_Mom	6	51.222	87.072	62.710	5.287	12.951
Knee_MaxER_Mom	6	-223.482	-13.786	-75.770	30.771	75.373
Time_MaxER_Mom	6	22.262	97.759	65.904	13.984	34.254
Hip_Max_FLX_Mom	6	1156.853	1502.397	1339.454	60.394	147.935
Hip_Time_MaxFLX_Mom	6	1.775	19.129	13.331	2.719	6.661
Hip_MaxEXT_Mom	6	-4212.770	-2203.453	-2829.648	305.332	747.907
Hip_Time_MaxEXT_Mom	6	58.446	62.375	60.647	0.695	1.702
Hip_MaxADD_Mom	6	1174.635	1933.710	1618.070	125.650	307.779
Hip_Time_MaxADD_Mom	6	20.698	55.631	41.809	6.236	15.275
Hip_MaxABD_Mom	6	-716.942	-459.267	-617.662	40.960	100.331
Hip_Time_MaxABD_Mom	6	1.685	96.054	77.782	15.309	37.499
Hip_MaxIR_Mom	6	35.649	259.958	94.377	34.346	84.131
Hip_Time_MaxIR_Mom	6	28.006	76.403	64.690	7.460	18.272
Hip_MaxER_Mom	6	-550.904	-232.830	-351.696	49.202	120.520
Hip_Time_MaxER_Mom	6	20.113	36.405	29.611	2.257	5.528
Inv_A1_Joint_Power	6	-8.189	-2.279	-4.578	0.816	1.999
Inv_A2_Joint_Power	6	14.960	21.250	19.273	1.055	2.583

Inv_K1_Joint_Power	6	-22.199	-10.715	-16.493	2.018	4.942
Inv_K2_Joint_Power	6	11.952	21.840	16.555	1.530	3.747
Inv_H1_Joint_Power	6	-3.603	1.252	-0.352	0.732	1.792
Inv_H2_Joint_Power	6	-26.697	-12.630	-17.825	2.293	5.617
Inv_Swing_Time	6	0.463	0.498	0.483	0.007	0.015
Uninv_Swing_Time	6	0.467	0.525	0.497	0.010	0.025
Inv_Stance_Time	6	0.221	0.292	0.245	0.010	0.025
Uninv_Stance_Time	6	0.227	0.284	0.253	0.008	0.019
Inv_Step_Length	6	1.417	1.489	1.443	0.011	0.028
Uninv_Step_Length	6	1.334	1.552	1.427	0.030	0.074
Stride_L	6	2.774	3.041	2.870	0.040	0.097
Velocity	6	3.851	4.124	3.997	0.045	0.110

Table A.7. No Brace (NB) Running Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	5	-15.550	0.957	-5.481	2.742	6.132
Ank_Max_DF	5	16.683	36.073	25.343	3.375	7.546
Ank_Time_Max_ DF	5	45.270	71.127	52.075	4.842	10.827
Ank_Max_PF	5	-32.460	-12.851	-21.131	3.542	7.919
Ank_Time_Max_PF	5	98.592	100.000	99.718	0.282	0.630
DF_exc	5	21.015	44.840	30.824	4.713	10.539
Ank_Po_TO	5	-25.938	-12.851	-18.640	2.153	4.815
Ank_MaxDF_Vel	5	275.224	3338.990	1029.462	579.587	1295.996
Ank_Time_MaxDF_Vel	5	6.983	98.592	30.881	17.180	38.417
Ank_MeanDF_Vel	5	47.839	363.147	235.109	57.503	128.580
Ank_MaxPF_Vel	5	-1723.760	-443.807	-763.416	241.620	540.279
Ank_Time_MaxPF_Vel	5	79.255	98.592	85.026	3.503	7.833
Ank_IVEV_Po_IC	5	-25.571	19.286	-3.354	7.334	16.400
Ank_MaxEV	5	1.772	41.111	24.973	7.320	16.368
Ank_Time_MaxEV	5	33.265	98.592	49.942	12.294	27.490
Ank_MaxIV	5	-25.571	4.226	-7.651	5.539	12.386
Ank_Time_MaxIV	5	0.704	95.769	54.166	21.730	48.589
Ank_EV_exc	5	16.212	56.368	32.624	6.623	14.809
Ank_Po_TO	5	-24.441	17.353	-2.780	7.259	16.232
Ank_MaxEV_Vel	5	226.851	966.156	589.937	138.621	309.966
Ank_Time_MaxEV_Vel	5	9.171	98.592	28.685	17.491	39.111
Ank_IRER_Po_IC	5	-14.616	10.513	1.926	4.493	10.047
Ank_MaxER	5	-32.342	-4.302	-15.742	5.393	12.059

Ank_Time_MaxER	5	27.239	92.958	45.032	12.260	27.414
Ank_MaxIR	5	-7.354	33.688	12.105	6.839	15.293
Ank_Time_MaxIR	5	67.320	100.000	88.015	6.005	13.428
IRER_Exc	5	13.694	51.205	27.848	7.990	17.866
Ank_IRER_Po_TO	5	-7.354	7.938	3.145	2.733	6.112
FootProgAng_30percstance	5	-16.463	-3.351	-11.916	2.372	5.303
Mean_footprogang_stance	5	1.950	6.178	3.515	0.705	1.577
Ank_MaxER_Vel	5	-6137.430	-161.744	-1479.708	1167.502	2610.614
Ank_Time_MaxER	5	10.615	99.296	31.924	16.883	37.752
Knee_Po_IC	5	5.644	22.704	14.092	2.838	6.347
Knee_Max_FLX	5	26.092	56.075	40.787	5.009	11.200
Knee_Time_Max_FLX	5	37.384	100.000	55.501	11.990	26.810
Knee_FLX_Exc	5	13.144	50.431	26.695	6.576	14.704
Knee_Po_TO	5	17.421	56.075	28.523	7.123	15.927
Knee_Max_FLX_Vel	5	292.212	11027.900	2506.457	2130.513	4763.972
Knee_Time_Max_FLX_Vel	5	13.646	99.296	56.828	16.065	35.923
Knee_Mean_FLX_Vel	5	85.029	299.390	180.031	37.095	82.947
Knee_Varval_Po_IC	5	-7.397	9.365	0.653	3.101	6.934
Knee_Max_VAR	5	-0.109	31.898	12.445	5.724	12.799
Knee_Time_Max_VAR	5	29.768	100.000	59.953	16.206	36.238
Knee_VAR_Exc	5	2.499	34.284	11.793	5.729	12.810
Knee_Po_TO	5	-3.305	5.666	1.447	1.518	3.394
Knee_Max_VAR_Vel	5	96.032	554.565	254.824	82.679	184.877
Knee_Time_Max_VAR_Vel	5	9.613	100.000	57.281	18.438	41.229
Knee_Mean_VAR_Vel	5	10.827	116.277	69.665	17.368	38.837
Knee_IRER_Po_IC	5	-17.337	13.477	-0.716	6.133	13.714
Knee_MaxIR	5	-6.222	36.373	15.115	7.669	17.149
Knee_Time_MaxIR	5	35.294	61.268	45.604	4.533	10.136
Knee_IR_Exc	5	11.115	26.058	15.832	2.687	6.008
KneeIR_Po_TO	5	-16.514	11.387	0.099	6.084	13.604
Knee_MaxIRER_Vel	5	180.562	21272.100	4530.217	4186.290	9360.829
Knee_Time_Max_Rot_Vel	5	6.067	99.296	33.673	16.793	37.550
Knee_MeanIRER_Vel	5	41.348	254.169	133.246	37.043	82.830
Hip_FLXEXT_Po_IC	5	23.658	43.952	34.309	3.953	8.839
Hip_Max_FLX	5	23.658	44.124	34.477	3.915	8.754
Hip_Time_MaxFLX	5	0.704	13.729	5.440	2.593	5.798
Hip_Max_EXT	5	-13.627	4.158	-3.900	3.007	6.724
Hip_Time_MaxEXT	5	81.690	100.000	92.787	3.456	7.728

Hip_ Exc_ICtomaxEXT	5	-47.905	-30.999	-38.209	3.015	6.742
Hip_FLXEXT_Po_TO	5	-5.653	4.916	0.004	2.155	4.820
Hip_MaxEXT_Vel	5	-472.617	57.067	-164.317	95.847	214.321
Hip_Time_MaxEXT_Vel	5	28.279	72.059	53.648	7.606	17.007
Hip_MeanEXT_Vel_ICtoMaxEXT	5	-187.547	-78.039	-142.538	18.381	41.100
HipABAD_Po_IC	5	-4.311	1.901	-1.970	1.127	2.520
Hip_MaxADD	5	0.928	8.704	5.067	1.260	2.817
Hip_Time_MaxADD	5	20.423	68.359	36.700	8.419	18.825
Hip_MaxABD	5	-12.606	-5.842	-8.672	1.116	2.495
Hip_Time_MaxABD	5	71.212	98.592	90.163	5.041	11.273
Hip_Exc_ICtoMaxABD	5	-11.068	-1.703	-6.702	1.880	4.205
Hip_ABAD_Po_TO	5	-12.539	-2.812	-6.928	1.687	3.772
Hip_MaxADD_Vel	5	92.869	1318.960	373.252	237.251	530.510
Hip_Time_MaxADD_Vel	5	14.149	99.296	47.018	16.291	36.429
Hip_Mean_Vel_ICtoMaxABD	5	-55.405	-6.539	-30.435	8.717	19.492
HipIRER_Po_IC	5	-28.097	7.853	-9.823	6.204	13.873
Hip_MaxIR	5	0.836	64.754	17.392	12.064	26.976
Hip_Time_MaxIR	5	28.251	100.000	63.070	15.427	34.496
Hip_MaxER	5	-36.560	-2.445	-15.443	5.883	13.156
Hip_Time_MaxER	5	0.704	97.030	56.813	20.240	45.259
Hip_Exc_ICtoMaxIR	5	6.428	72.877	27.216	12.302	27.508
Hip_Exc_MaxIRtoMaxER	5	-72.877	-15.578	-32.836	10.933	24.447
HipIRER_Po_TO	5	-27.068	6.111	-6.457	5.782	12.929
Hip_MaxIR_Vel	5	125.398	1676.260	618.219	278.070	621.784
Hip_Time_MaxIR_Vel	5	10.096	100.000	48.606	20.757	46.414
Hip_Mean_Vel_ICtoMaxIR	5	71.398	232.636	119.914	29.680	66.366
Hip_MaxER_Vel	5	-14151.800	-201.206	-3151.183	2753.429	6156.854
Hip_Time_MaxER_Vel	5	38.908	99.296	64.406	10.172	22.746
Max_Pelv_Ang	5	1.189	8.793	4.908	1.576	3.525
Pelv_Ang_Exc	5	5.109	11.173	7.975	1.154	2.581
Max_thor_angle	5	-0.606	9.869	4.372	1.864	4.169
Thor_ang_exc	5	6.302	8.011	7.088	0.282	0.631
Max_spine_ang	5	2.105	10.062	5.710	1.374	3.072
Spine_ang_exc	5	11.061	21.112	16.130	1.633	3.650
MaxGRFz	4	20.574	26.750	23.285	1.375	2.749
Time_MaxGRFz	4	41.624	48.055	46.151	1.530	3.059
Load_Rate	4	11559.111	22575.180	18118.982	2329.809	4659.618
GRFimp	4	237.847	394.514	302.025	35.611	71.222
GRFimp_norm	4	0.315	0.355	0.338	0.010	0.019

Max_brake_GRFy	4	-2.667	-1.735	-2.262	0.210	0.420
Time_max_brake_GRFy	4	20.227	31.310	25.722	2.508	5.017
Max_prop_GRFy	4	2.461	3.081	2.715	0.135	0.271
Time_max_prop_GRFy	4	75.256	78.796	76.697	0.773	1.545
Max_Abs_FM	4	0.003	0.010	0.006	0.001	0.003
Max_ADD_FM	4	0.001	0.010	0.005	0.002	0.004
Time_max_ADD_FM	4	37.840	100.000	58.202	14.117	28.234
Max_ABD_FM	4	-0.006	0.000	-0.003	0.001	0.002
Time_Max_ABD_FM	4	13.964	68.783	35.968	13.214	26.429
FM_Peak_Brake_GRFy	4	-0.002	0.002	0.000	0.001	0.002
ADD_FM_imp	4	0.000	0.001	0.000	0.000	0.000
Max_DF_Mom	4	1977.413	3099.243	2627.290	275.468	550.937
Time_Max_DF_Mom	4	50.361	58.097	54.645	1.611	3.223
Max_PF_Mom	4	-164.938	-24.636	-77.492	31.538	63.076
Time_Max_PF_Mom	4	1.515	34.641	11.214	7.917	15.834
MaxIV_Mom	4	134.695	349.398	245.324	48.936	97.872
Time_MaxIV_Mom	4	50.788	76.260	60.505	5.571	11.142
MaxEV_Mom	4	-209.396	-7.358	-101.416	43.279	86.557
Time_MaxEV_Mom	4	16.653	100.000	39.702	20.127	40.254
Ank_MaxIR_Mom	4	228.486	825.314	442.887	132.930	265.860
Time_MaxIR_Mom	4	45.956	50.946	49.012	1.083	2.165
MaxER_Mom	4	-60.306	16.856	-33.195	17.322	34.643
Time_MaxER_Mom	4	64.192	99.415	82.008	8.872	17.744
Max_FLX_Mom_LR	4	-59.525	1512.676	667.793	351.944	703.889
Time_Max_FLX_Mom_LR	4	-148.778	30.696	-50.006	44.989	89.978
Max_FLX_Mom_TO	4	1217.830	2225.803	1945.062	243.273	486.545
Time_Max_FLX_Mom_TO	4	27.278	40.316	36.283	3.077	6.154
Max_EXT_Mom	4	-773.409	-333.196	-521.501	106.294	212.588
Time_Max_EXT_Mom	4	1.608	87.858	30.654	20.297	40.594
Knee_ Stiff	4	0.682	10.151	4.978	2.022	4.044
Knee_MaxADD_Mom	4	708.756	2027.763	1387.805	302.974	605.947
Knee_Timing_MaxADD_Mom	4	37.352	52.253	44.233	3.103	6.205
Knee_MaxABD_Mom	4	-363.456	4.325	-172.240	78.458	156.916
Knee_Time_MaxABD_Mom	4	65.359	95.454	83.558	6.412	12.825
Knee_MaxIR_Mom	4	172.358	354.117	274.410	45.550	91.100
Knee_Time_MaxIR_Mom	4	46.982	57.876	53.391	2.294	4.587
Knee_MaxER_Mom	4	-69.713	-2.643	-39.484	13.892	27.784
Time_MaxER_Mom	4	21.165	96.938	69.797	17.604	35.208
Hip_Max_FLX_Mom	4	781.779	2042.253	1332.396	261.409	522.818

Hip_Time_MaxFLX_Mom	4	1.515	15.666	5.170	3.499	6.999
Hip_MaxEXT_Mom	4	-2776.963	-1591.447	-2136.386	277.507	555.015
Hip_Time_MaxEXT_Mom	4	59.391	64.655	62.087	1.096	2.193
Hip_MaxADD_Mom	4	1194.117	2139.747	1556.481	203.326	406.651
Hip_Time_MaxADD_Mom	4	18.335	56.581	43.491	8.797	17.594
Hip_MaxABD_Mom	4	-1283.087	-302.137	-591.280	231.844	463.688
Hip_Time_MaxABD_Mom	4	86.546	95.454	89.667	2.013	4.027
Hip_MaxIR_Mom	4	-9.305	121.449	50.972	27.882	55.764
Hip_Time_MaxIR_Mom	4	75.763	100.000	83.659	5.517	11.034
Hip_MaxER_Mom	4	-352.237	-100.963	-231.496	57.254	114.508
Hip_Time_MaxER_Mom	4	19.573	37.352	28.305	3.720	7.441
Inv_A1_Joint_Power	4	-8.730	-3.545	-6.669	1.145	2.290
Inv_A2_Joint_Power	4	11.163	13.378	12.490	0.518	1.035
Inv_K1_Joint_Power	4	-9.198	-2.663	-7.047	1.493	2.986
Inv_K2_Joint_Power	4	4.026	14.277	8.965	2.245	4.489
Inv_H1_Joint_Power	4	-0.276	2.591	0.681	0.653	1.307
Inv_H2_Joint_Power	4	-13.697	-5.210	-8.464	1.824	3.648
Inv_Swing_Time	4	0.392	0.531	0.478	0.044	0.076
Uninv_Swing_Time	4	0.392	0.467	0.444	0.018	0.035
Inv_Stance_Time	5	0.224	0.588	0.321	0.067	0.150
Uninv_Stance_Time	5	0.216	0.579	0.334	0.063	0.141
Inv_Step_Length	4	0.929	1.445	1.182	0.131	0.262
Uninv_Step_Length	4	0.827	1.288	1.077	0.101	0.202
Stride_L	4	1.756	2.657	2.259	0.227	0.453
Velocity	4	2.778	3.636	3.183	0.201	0.402

Two of six service member had difficulty running without a brace. One did not attempt running condition, and one never hit the force plate, therefore n=5 for kinematics, and n=4 for kinetics.

Table A.8. Traditional AFO (TAFO) Running Descriptive Data

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	4	-3.106	3.507	-0.226	1.420	2.840
Ank_Max_DF	4	14.401	25.132	21.183	2.393	4.787
Ank_Time_Max_ DF	4	45.802	64.557	53.760	4.031	8.062
Ank_Max_PF	4	-14.420	-0.053	-4.339	3.376	6.753
Ank_Time_Max_PF	4	1.563	100.000	44.212	24.708	49.417
DF_exc	4	10.894	26.679	21.409	3.564	7.129
Ank_Po_TO	4	-14.420	2.784	-2.720	4.001	8.002
Ank_MaxDF_Vel	4	182.832	444.408	299.917	56.481	112.962
Ank_Time_MaxDF_Vel	4	5.807	25.274	18.035	4.263	8.525
Ank_MeanDF_Vel	4	39.516	263.982	157.366	48.403	96.806

Ank_MaxPF_Vel	4	-464.468	-231.733	-318.347	51.007	102.013
Ank_Time_MaxPF_Vel	4	75.763	98.745	87.177	5.809	11.619
Ank_IVEV_Po_IC	4	-11.121	25.597	7.423	8.655	17.309
Ank_MaxEV	4	1.805	38.354	22.863	7.641	15.282
Ank_Time_MaxEV	4	28.381	100.000	51.377	16.614	33.227
Ank_MaxIV	4	-11.121	21.706	4.408	7.227	14.454
Ank_Time_MaxIV	4	1.226	79.349	35.509	20.092	40.185
Ank_EV_exc	4	12.925	27.722	18.455	3.207	6.414
Ank_Po_TO	4	-6.432	26.559	13.442	8.040	16.079
Ank_MaxEV_Vel	4	159.611	558.060	312.582	93.772	187.544
Ank_Time_MaxEV_Vel	4	2.029	53.740	20.779	11.378	22.757
Ank_IRER_Po_IC	4	-4.436	26.168	15.417	6.828	13.656
Ank_MaxER	4	-15.849	9.163	-3.371	6.852	13.705
Ank_Time_MaxER	4	29.022	100.000	65.796	18.084	36.168
Ank_MaxIR	4	0.217	28.894	17.894	6.309	12.618
Ank_Time_MaxIR	4	1.226	100.000	48.829	25.288	50.576
IRER_Exc	4	14.813	33.777	21.266	4.291	8.583
Ank_IRER_Po_TO	4	-15.849	24.538	4.525	8.778	17.557
FootProgAng_30percstance	4	-12.948	-4.989	-7.948	1.806	3.612
Mean_footprogang_stance	4	0.184	6.188	3.355	1.274	2.548
Ank_MaxER_Vel	4	-512.821	-146.234	-302.688	80.284	160.569
Ank_Time_MaxER	4	4.069	84.145	40.346	19.813	39.626
Knee_Po_IC	4	12.903	18.906	17.116	1.412	2.823
Knee_Max_FLX	4	33.490	55.225	42.480	4.589	9.177
Knee_Time_Max_FLX	4	39.329	100.000	70.760	16.905	33.811
Knee_FLX_Exc	4	14.584	36.984	25.364	4.630	9.259
Knee_Po_TO	4	16.352	55.225	34.343	8.748	17.496
Knee_Max_FLX_Vel	4	309.497	520.090	421.914	55.474	110.948
Knee_Time_Max_FLX_Vel	4	14.090	100.000	61.353	20.631	41.263
Knee_Mean_FLX_Vel	4	69.338	238.150	144.001	40.232	80.464
Knee_Varval_Po_IC	4	-6.623	4.576	-0.772	2.660	5.321
Knee_Max_VAR	4	1.487	21.379	11.357	4.376	8.752
Knee_Time_Max_VAR	4	38.278	100.000	79.715	14.551	29.102
Knee_VAR_Exc	4	5.322	18.584	12.129	2.787	5.574
Knee_Po_TO	4	1.024	21.379	7.787	4.747	9.494
Knee_Max_VAR_Vel	4	115.686	314.059	217.548	44.446	88.892
Knee_Time_Max_VAR_Vel	4	25.775	96.705	72.286	15.990	31.980
Knee_Mean_VAR_Vel	4	23.538	105.346	57.050	17.404	34.809

Knee_IRER_Po_IC	4	-6.316	3.005	-1.457	1.937	3.874
Knee_MaxIR	4	3.135	9.491	7.562	1.484	2.968
Knee_Time_MaxIR	4	30.602	75.411	53.535	11.718	23.436
Knee_IR_Exc	4	5.237	15.808	9.019	2.431	4.862
KneeIR_Po_TO	4	-23.749	5.904	-8.140	6.781	13.562
Knee_MaxIRER_Vel	4	136.747	314.960	236.408	38.167	76.334
Knee_Time_Max_Rot_Vel	4	10.269	55.889	35.644	9.435	18.871
Knee_MeanIRER_Vel	4	18.524	159.767	85.141	35.410	70.820
Hip_FLXEXT_Po_IC	4	39.797	47.679	42.234	1.830	3.659
Hip_Max_FLX	4	39.797	47.679	42.234	1.830	3.659
Hip_Time_MaxFLX	4	0.971	1.936	1.424	0.209	0.419
Hip_Max_EXT	4	-6.264	2.552	-1.251	2.150	4.299
Hip_Time_MaxEXT	4	70.904	100.000	87.160	6.713	13.426
Hip_ Exc_ICtomaxEXT	4	-47.155	-37.244	-43.484	2.180	4.359
Hip_FLXEXT_Po_TO	4	-6.264	17.274	3.985	4.885	9.770
Hip_MaxEXT_Vel	4	-378.782	-7.057	-267.206	87.308	174.616
Hip_Time_MaxEXT_Vel	4	30.415	65.787	52.865	7.766	15.532
Hip_MeanEXT_Vel_ICtoMaxEXT	4	-182.860	-151.995	-168.639	7.417	14.833
HipABAD_Po_IC	4	-4.786	4.577	-0.455	2.155	4.310
Hip_MaxADD	4	-0.424	8.150	3.993	1.842	3.684
Hip_Time_MaxADD	4	26.471	62.967	38.671	8.362	16.724
Hip_MaxABD	4	-9.482	-5.686	-7.146	0.823	1.646
Hip_Time_MaxABD	4	56.154	92.911	75.287	7.592	15.183
Hip_Exc_ICtoMaxABD	4	-11.001	-2.490	-6.691	1.909	3.819
Hip_ABAD_Po_TO	4	-8.027	-2.608	-5.121	1.184	2.367
Hip_MaxADD_Vel	4	81.961	115.791	92.947	7.812	15.624
Hip_Time_MaxADD_Vel	4	11.614	50.765	31.694	8.636	17.272
Hip_Mean_Vel_ICtoMaxABD	4	-53.652	-18.807	-33.144	7.536	15.073
HipIRER_Po_IC	4	-34.664	-13.809	-20.289	4.847	9.694
Hip_MaxIR	4	4.368	17.529	8.055	3.176	6.353
Hip_Time_MaxIR	4	38.278	100.000	79.715	14.551	29.102
Hip_MaxER	4	-41.270	-15.313	-23.076	6.098	12.195
Hip_Time_MaxER	4	1.226	86.679	35.649	20.853	41.707
Hip_Exc_ICtoMaxIR	4	19.668	39.032	28.344	4.584	9.167
Hip_Exc_MaxIRtoMaxER	4	-45.638	-21.835	-31.131	5.383	10.766
HipIRER_Po_TO	4	-15.888	17.529	1.987	6.875	13.751
Hip_MaxIR_Vel	4	282.455	636.684	379.093	86.112	172.224
Hip_Time_MaxIR_Vel	4	22.650	71.213	50.314	11.929	23.858
Hip_Mean_Vel_ICtoMaxIR	4	99.578	194.206	131.578	21.913	43.826

Hip_MaxER_Vel	4	-532.516	-76.605	-257.333	108.521	217.043
Hip_Time_MaxER_Vel	4	45.802	61.945	53.789	3.705	7.410
Max_Pelv_Ang	4	3.209	6.375	4.670	0.669	1.338
Pelv_Ang_Exc	4	7.550	11.204	8.767	0.850	1.700
Max_thor_angle	4	0.822	5.697	3.010	1.085	2.170
Thor_ang_exc	4	3.959	9.639	7.373	1.288	2.576
Max_spine_ang	4	7.641	11.647	9.437	0.934	1.869
Spine_ang_exc	4	17.252	19.072	18.127	0.375	0.751
MaxGRFz	4	15.739	25.567	19.842	2.422	4.843
Time_MaxGRFz	4	40.649	46.789	43.015	1.336	2.671
Load_Rate	4	8650.633	23495.047	15356.075	3885.863	7771.726
GRFimp	4	207.714	387.935	282.577	38.684	77.368
GRFimp_norm	4	0.268	0.348	0.306	0.017	0.034
Max_brake_GRFy	4	-3.506	-1.079	-2.093	0.556	1.112
Time_max_brake_GRFy	4	22.464	32.796	27.881	2.577	5.154
Max_prop_GRFy	4	0.954	2.308	1.653	0.364	0.728
Time_max_prop_GRFy	4	57.114	78.120	69.926	4.614	9.228
Max_Abs_FM	4	0.002	0.007	0.004	0.001	0.002
Max_ADD_FM	4	0.002	0.004	0.003	0.001	0.001
Time_max_ADD_FM	4	12.503	58.750	39.691	11.014	22.029
Max_ABD_FM	4	-0.007	0.000	-0.003	0.001	0.003
Time_Max_ABD_FM	4	12.267	81.273	52.050	15.023	30.047
FM_Peak_Brake_GRFy	4	-0.002	0.004	0.001	0.001	0.003
ADD_FM_imp	4	0.000	0.001	0.000	0.000	0.000
Max_DF_Mom	4	536.651	2654.190	1646.814	474.688	949.375
Time_Max_DF_Mom	4	49.673	59.341	53.135	2.220	4.440
Max_PF_Mom	4	-233.439	-15.813	-121.821	44.630	89.260
Time_Max_PF_Mom	4	1.936	70.297	22.320	16.098	32.195
MaxIV_Mom	4	22.404	548.058	280.355	134.651	269.302
Time_MaxIV_Mom	4	52.681	65.836	57.197	2.940	5.881
MaxEV_Mom	4	-136.667	7.930	-48.289	30.965	61.929
Time_MaxEV_Mom	4	1.936	56.798	24.701	11.754	23.508
Ank_MaxIR_Mom	4	43.331	333.851	204.993	61.777	123.554
Time_MaxIR_Mom	4	40.396	55.421	48.607	3.840	7.680
MaxER_Mom	4	-95.737	-13.506	-39.507	19.201	38.401
Time_MaxER_Mom	4	1.563	100.000	65.209	23.017	46.033
Max_FLX_Mom_LR	4	-69.402	1268.150	636.763	308.835	617.670
Time_Max_FLX_Mom_LR	4	-109.075	35.589	-36.720	37.998	75.997
Max_FLX_Mom_TO	4	1073.374	2640.680	1954.918	341.762	683.524
Time_Max_FLX_Mom_TO	4	37.239	41.622	38.598	1.023	2.046

Max_EXT_Mom	4	-587.494	-294.201	-432.370	62.959	125.918
Time_Max_EXT_Mom	4	0.971	4.902	2.343	0.876	1.752
Knee_ Stiff	4	1.122	8.248	4.048	1.502	3.005
Knee_MaxADD_Mom	4	444.447	2116.410	1030.442	370.795	741.590
Knee_Timing_MaxADD_Mom	4	33.072	51.621	43.669	3.864	7.729
Knee_MaxABD_Mom	4	-258.083	-22.131	-105.624	52.862	105.723
Knee_Time_MaxABD_Mom	4	50.769	98.064	81.066	11.073	22.146
Knee_MaxIR_Mom	4	54.902	487.547	269.544	100.510	201.021
Knee_Time_MaxIR_Mom	4	43.395	57.057	52.304	3.095	6.190
Knee_MaxER_Mom	4	-88.994	-16.945	-40.273	16.471	32.941
Time_MaxER_Mom	4	1.563	98.064	71.907	23.509	47.018
Hip_Max_FLX_Mom	4	609.105	1473.900	994.525	180.730	361.461
Hip_Time_MaxFLX_Mom	4	1.563	11.029	4.198	2.281	4.563
Hip_MaxEXT_Mom	4	-1995.877	-1259.917	-1647.187	154.144	308.289
Hip_Time_MaxEXT_Mom	4	63.211	84.147	70.789	4.627	9.254
Hip_MaxADD_Mom	4	603.047	2144.315	1495.198	322.507	645.013
Hip_Time_MaxADD_Mom	4	16.780	63.829	41.573	9.769	19.538
Hip_MaxABD_Mom	4	-717.106	69.287	-237.564	168.101	336.202
Hip_Time_MaxABD_Mom	4	25.505	94.469	71.171	15.942	31.884
Hip_MaxIR_Mom	4	10.110	60.207	34.533	10.255	20.509
Hip_Time_MaxIR_Mom	4	62.670	96.923	73.713	7.961	15.922
Hip_MaxER_Mom	4	-384.689	-118.796	-214.307	61.183	122.366
Hip_Time_MaxER_Mom	4	16.780	44.530	30.384	6.253	12.507
Inv_A1_Joint_Power	4	-6.807	-1.303	-3.610	1.364	2.729
Inv_A2_Joint_Power	4	1.089	12.106	6.201	2.771	5.543
Inv_K1_Joint_Power	4	-10.977	-0.954	-5.149	2.121	4.242
Inv_K2_Joint_Power	4	1.182	11.610	6.733	2.138	4.276
Inv_H1_Joint_Power	4	-1.228	2.046	0.390	0.671	1.342
Inv_H2_Joint_Power	4	-8.329	-3.606	-6.497	1.016	2.032
Inv_Swing_Time	4	0.371	0.492	0.420	0.037	0.064
Uninv_Swing_Time	4	0.323	0.381	0.352	0.029	0.041
Inv_Stance_Time	4	0.214	0.454	0.320	0.052	0.104
Uninv_Stance_Time	4	0.210	0.446	0.324	0.050	0.099
Inv_Step_Length	4	0.985	1.674	1.241	0.159	0.317
Uninv_Step_Length	4	0.789	1.365	1.000	0.126	0.251
Stride_L	4	1.902	2.646	2.241	0.185	0.370
Velocity	4	2.103	3.561	2.926	0.317	0.634

One service member was unable to run in the TAFO due to fatigue, and one service member discarded his TAFO due to medical complications.

Table A.9. Dynamic AFO (DAFO) Running Descriptive Data at Time 0

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	5	5.169	11.626	7.882	1.135	2.538
Ank_Max_DF	5	10.670	22.527	15.713	2.261	5.055
Ank_Time_Max_ DF	5	46.176	72.314	53.969	5.061	11.317
Ank_Max_PF	5	4.462	10.537	7.125	1.140	2.550
Ank_Time_Max_PF	5	4.127	67.347	32.735	13.504	30.196
DF_exc	5	5.198	13.444	7.831	1.489	3.330
Ank_Po_TO	5	4.463	11.415	8.046	1.315	2.940
Ank_MaxDF_Vel	5	93.971	227.927	136.976	24.210	54.134
Ank_Time_MaxDF_Vel	5	13.081	38.188	25.451	4.083	9.130
Ank_MeanDF_Vel	5	26.396	123.101	65.357	15.783	35.293
Ank_MaxPF_Vel	5	-142.597	-92.009	-118.643	10.063	22.502
Ank_Time_MaxPF_Vel	5	69.828	93.701	77.944	4.132	9.240
Ank_IVEV_Po_IC	5	-8.358	27.688	3.579	6.650	14.871
Ank_MaxEV	5	-5.767	28.693	6.687	6.167	13.790
Ank_Time_MaxEV	5	27.781	98.387	61.283	14.266	31.899
Ank_MaxIV	5	-29.154	21.630	-5.039	8.236	18.415
Ank_Time_MaxIV	5	51.686	92.208	70.661	7.413	16.576
Ank_EV_exc	5	7.064	23.388	11.726	2.982	6.667
Ank_Po_TO	5	-11.104	28.358	2.778	7.361	16.460
Ank_MaxEV_Vel	5	81.878	585.433	225.789	91.924	205.549
Ank_Time_MaxEV_Vel	5	12.477	99.454	58.000	16.449	36.781
Ank_IRER_Po_IC	5	-20.498	7.856	-4.650	5.742	12.840
Ank_MaxER	5	-27.144	4.788	-10.244	6.286	14.055
Ank_Time_MaxER	5	23.785	68.553	42.745	7.399	16.544
Ank_MaxIR	5	-10.025	25.022	10.728	6.650	14.869
Ank_Time_MaxIR	5	61.700	92.208	74.069	5.698	12.742
IRER_Exc	5	13.475	38.248	20.972	4.427	9.898
Ank_IRER_Po_TO	5	-11.645	8.459	-0.265	3.394	7.589
FootProgAng_30percstance	5	-13.732	37.087	7.797	8.456	18.908
Mean_footprogang_stance	5	0.434	7.392	3.030	1.308	2.925
Ank_MaxER_Vel	5	-1151.130	-117.704	-397.896	190.778	426.593
Ank_Time_MaxER	5	11.364	100.000	65.376	15.825	35.386
Knee_Po_IC	5	9.627	28.355	21.185	3.192	7.137
Knee_Max_FLX	5	27.652	45.917	37.039	3.612	8.077
Knee_Time_Max_FLX	5	47.982	67.275	54.352	3.344	7.477

Knee_FLX_Exc	5	6.948	26.081	15.854	3.461	7.738
Knee_Po_TO	5	20.011	31.187	26.391	2.348	5.249
Knee_Max_FLX_Vel	5	180.093	330.711	272.011	28.271	63.216
Knee_Time_Max_FLX_Vel	5	10.666	100.000	47.444	17.597	39.348
Knee_Mean_FLX_Vel	5	51.549	216.815	132.877	29.483	65.926
Knee_Varval_Po_IC	5	-10.504	5.734	-1.921	3.029	6.774
Knee_Max_VAR	5	-9.595	13.530	4.256	4.527	10.123
Knee_Time_Max_VAR	5	27.529	43.178	34.837	3.124	6.986
Knee_VAR_Exc	5	0.909	12.900	6.177	1.966	4.396
Knee_Po_TO	5	-15.058	2.060	-5.157	2.973	6.647
Knee_Max_VAR_Vel	5	125.890	308.128	189.369	34.023	76.077
Knee_Time_Max_VAR_Vel	5	23.624	83.668	49.027	9.704	21.698
Knee_Mean_VAR_Vel	5	-29.885	136.145	65.379	26.968	60.303
Knee_IRER_Po_IC	5	2.544	17.571	9.394	2.553	5.709
Knee_MaxIR	5	3.685	22.331	13.533	3.074	6.873
Knee_Time_MaxIR	5	5.149	94.145	60.629	16.658	37.249
Knee_IR_Exc	5	1.141	7.313	4.140	1.008	2.254
KneeIR_Po_TO	5	-0.536	20.804	10.462	3.621	8.096
Knee_MaxIRER_Vel	5	77.993	491.643	257.655	81.704	182.695
Knee_Time_Max_Rot_Vel	5	11.405	82.183	48.597	12.162	27.195
Knee_MeanIRER_Vel	5	15.075	77.484	41.730	11.227	25.104
Hip_FLXEXT_Po_IC	5	35.597	52.085	44.114	2.880	6.441
Hip_Max_FLX	5	35.597	52.085	44.114	2.880	6.441
Hip_Time_MaxFLX	5	1.564	2.070	1.776	0.083	0.185
Hip_Max_EXT	5	-4.400	6.964	3.781	2.134	4.772
Hip_Time_MaxEXT	5	97.959	100.000	99.592	0.408	0.913
Hip_ Exc_ICtomaxEXT	5	-46.241	-32.001	-40.332	2.793	6.246
Hip_FLXEXT_Po_TO	5	-4.400	7.020	3.805	2.143	4.792
Hip_MaxEXT_Vel	5	-463.786	30.198	-224.846	92.971	207.890
Hip_Time_MaxEXT_Vel	5	30.570	82.834	62.036	10.904	24.383
Hip_MeanEXT_Vel_ICtoMaxEXT	5	-198.325	-133.420	-175.017	12.537	28.034
HipABAD_Po_IC	5	-1.863	6.444	2.760	1.527	3.414
Hip_MaxADD	5	-0.860	15.704	7.106	2.659	5.945
Hip_Time_MaxADD	5	28.029	87.733	45.137	11.175	24.988
Hip_MaxABD	5	-6.983	-1.311	-4.266	1.147	2.566
Hip_Time_MaxABD	5	41.710	100.000	83.667	10.847	24.255
Hip_Exc_ICtoMaxABD	5	-12.692	-2.635	-7.026	1.857	4.153
Hip_ABAD_Po_TO	5	-6.900	2.751	-3.097	1.675	3.745

Hip_MaxADD_Vel	5	46.497	207.477	99.865	29.285	65.484
Hip_Time_MaxADD_Vel	5	22.692	70.096	42.711	8.005	17.899
Hip_Mean_Vel_ICtoMaxABD	5	-55.451	-19.601	-40.133	7.692	17.199
HipIRER_Po_IC	5	-31.077	11.488	-7.774	7.642	17.088
Hip_MaxIR	5	-9.154	26.268	6.970	6.341	14.179
Hip_Time_MaxIR	5	20.926	49.949	35.320	5.592	12.505
Hip_MaxER	5	-35.774	-3.583	-17.096	5.658	12.653
Hip_Time_MaxER	5	59.575	86.398	70.688	5.025	11.236
Hip_Exc_ICtoMaxIR	5	2.601	28.880	14.744	4.935	11.036
Hip_Exc_MaxIRtoMaxER	5	-33.800	-17.249	-24.065	3.078	6.882
HipIRER_Po_TO	5	-32.537	5.501	-10.004	6.573	14.698
Hip_MaxIR_Vel	5	154.990	751.291	403.277	99.074	221.537
Hip_Time_MaxIR_Vel	5	27.505	100.000	62.830	13.315	29.774
Hip_Mean_Vel_ICtoMaxIR	5	-64.058	302.826	152.139	68.445	153.047
Hip_MaxER_Vel	5	-537.030	-296.399	-406.226	46.643	104.297
Hip_Time_MaxER_Vel	5	45.064	73.399	54.963	5.355	11.974
Max_Pelv_Ang	5	2.265	6.605	3.920	0.726	1.624
Pelv_Ang_Exc	5	6.292	10.777	8.433	0.872	1.949
Max_thor_angle	5	-1.108	12.571	4.942	2.398	5.362
Thor_ang_exc	5	6.630	9.688	7.827	0.616	1.376
Max_spine_ang	5	2.249	13.827	7.947	2.086	4.665
Spine_ang_exc	5	14.083	22.938	17.922	1.571	3.513
MaxGRFz	5	17.548	25.959	22.564	1.571	3.512
Time_MaxGRFz	5	44.174	59.908	51.474	2.983	6.670
Load_Rate	5	8794.828	24411.031	18133.406	2956.895	6611.818
GRFimp	5	209.990	349.091	269.820	26.555	59.380
GRFimp_norm	5	0.268	0.327	0.301	0.011	0.024
Max_brake_GRFy	5	-2.560	-1.712	-2.111	0.173	0.387
Time_max_brake_GRFy	5	33.087	45.788	38.500	2.265	5.065
Max_prop_GRFy	5	0.869	2.492	1.663	0.277	0.620
Time_max_prop_GRFy	5	75.163	85.913	80.453	1.853	4.144
Max_Abs_FM	5	0.003	0.009	0.006	0.001	0.002
Max_ADD_FM	5	0.001	0.009	0.003	0.001	0.003
Time_max_ADD_FM	5	10.976	56.984	34.930	9.515	21.276
Max_ABD_FM	5	-0.007	-0.001	-0.005	0.001	0.003
Time_Max_ABD_FM	5	19.326	80.738	59.471	10.478	23.429
FM_Peak_Brake_GRFy	5	-0.006	0.005	0.000	0.002	0.004
ADD_FM_imp	5	0.000	0.001	0.000	0.000	0.000
Max_DF_Mom	5	1912.143	2740.433	2280.920	133.713	298.991
Time_Max_DF_Mom	5	44.941	68.549	56.951	4.520	10.107

Max_PF_Mom	5	-266.719	-66.421	-164.093	31.716	70.920
Time_Max_PF_Mom	5	2.070	19.733	11.251	3.137	7.015
MaxIV_Mom	5	26.328	270.751	134.323	44.133	98.685
Time_MaxIV_Mom	5	4.140	73.413	40.812	11.269	25.198
MaxEV_Mom	5	-1239.397	-10.445	-343.545	233.958	523.147
Time_MaxEV_Mom	5	2.624	71.106	44.576	12.710	28.419
Ank_MaxIR_Mom	5	168.448	534.615	346.452	68.541	153.262
Time_MaxIR_Mom	5	43.771	68.527	57.972	4.236	9.472
MaxER_Mom	5	-67.171	-18.874	-43.486	8.393	18.768
Time_MaxER_Mom	5	12.604	72.918	51.357	10.867	24.299
Max_FLX_Mom_LR	5	106.900	3755.903	978.048	697.290	1559.187
Time_Max_FLX_Mom_LR	5	-133.398	55.078	-53.527	35.469	79.311
Max_FLX_Mom_TO	5	917.035	3869.643	2240.872	480.775	1075.046
Time_Max_FLX_Mom_TO	5	47.576	58.807	52.925	2.169	4.850
Max_EXT_Mom	5	-778.997	-261.351	-485.142	101.587	227.155
Time_Max_EXT_Mom	5	1.773	98.949	33.593	17.746	39.681
Knee_ Stiff	5	0.643	16.131	9.456	2.942	6.578
Knee_MaxADD_Mom	5	271.324	1560.020	855.554	248.946	556.660
Knee_Timing_MaxADD_Mom	5	17.029	60.125	46.326	7.907	17.680
Knee_MaxABD_Mom	5	-776.966	-44.875	-232.545	138.103	308.807
Knee_Time_MaxABD_Mom	5	5.734	97.221	57.823	14.827	33.154
Knee_MaxIR_Mom	5	99.165	415.215	273.038	63.572	142.151
Knee_Time_MaxIR_Mom	5	52.600	60.465	56.293	1.734	3.878
Knee_MaxER_Mom	5	-63.319	-16.577	-37.278	8.123	18.164
Time_MaxER_Mom	5	34.351	98.342	53.251	12.461	27.864
Hip_Max_FLX_Mom	5	492.130	1644.583	1031.091	234.004	523.250
Hip_Time_MaxFLX_Mom	5	2.070	46.460	16.244	8.104	18.121
Hip_MaxEXT_Mom	5	-2271.880	-887.842	-1683.832	275.374	615.755
Hip_Time_MaxEXT_Mom	5	46.468	81.811	63.154	5.918	13.233
Hip_MaxADD_Mom	5	888.138	1347.880	1194.117	88.154	197.119
Hip_Time_MaxADD_Mom	5	17.948	100.000	47.616	13.802	30.861
Hip_MaxABD_Mom	5	-910.610	50.005	-327.153	161.017	360.045
Hip_Time_MaxABD_Mom	5	61.445	98.324	80.336	6.014	13.448
Hip_MaxIR_Mom	5	19.083	39.066	30.833	3.870	8.653
Hip_Time_MaxIR_Mom	5	63.729	94.015	79.392	6.171	13.799
Hip_MaxER_Mom	5	-327.495	-110.942	-217.390	34.331	76.766
Hip_Time_MaxER_Mom	5	31.524	48.586	40.487	2.994	6.695
Inv_A1_Joint_Power	5	-3.906	-1.373	-2.578	0.404	0.903
Inv_A2_Joint_Power	5	1.860	5.653	3.887	0.621	1.388

Inv_K1_Joint_Power	5	-6.109	-0.754	-4.241	0.963	2.154
Inv_K2_Joint_Power	5	1.569	8.779	5.960	1.442	3.225
Inv_H1_Joint_Power	5	-4.087	3.254	0.250	1.212	2.710
Inv_H2_Joint_Power	5	-8.231	3.274	-3.740	1.937	4.332
Inv_Swing_Time	5	0.404	0.524	0.463	0.024	0.049
Uninv_Swing_Time	5	0.360	0.494	0.414	0.023	0.052
Inv_Stance_Time	5	0.208	0.286	0.242	0.013	0.028
Uninv_Stance_Time	5	0.264	0.338	0.294	0.012	0.028
Inv_Step_Length	5	0.966	1.496	1.178	0.106	0.237
Uninv_Step_Length	5	0.842	1.433	1.129	0.131	0.293
Stride_L	5	1.809	2.926	2.307	0.235	0.525
Velocity	5	2.548	4.237	3.284	0.310	0.693

One service member was unable to complete this condition due to fatigue

Table A.10. Dynamic AFO (DAFO) Running Descriptive Data at Last Visit

	N	Minimum	Maximum	Mean		Std. Deviation
	Statistic	Statistic	Statistic	Statistic	Std. Error	Statistic
Ank_Po_IC	4	6.193	8.804	7.758	0.555	1.110
Ank_Max_DF	4	13.143	20.583	16.680	1.578	3.155
Ank_Time_Max_ DF	4	30.851	49.020	43.054	4.123	8.245
Ank_Max_PF	4	4.622	6.954	5.553	0.517	1.035
Ank_Time_Max_PF	4	67.273	100.000	91.286	8.020	16.040
DF_exc	4	5.022	14.390	8.922	1.979	3.957
Ank_Po_TO	4	4.639	6.954	5.614	0.523	1.045
Ank_MaxDF_Vel	4	77.091	230.774	170.066	33.826	67.651
Ank_Time_MaxDF_Vel	4	11.702	28.630	19.183	3.812	7.625
Ank_MeanDF_Vel	4	45.220	132.477	97.109	21.133	42.266
Ank_MaxPF_Vel	4	-183.659	-91.939	-140.109	18.813	37.626
Ank_Time_MaxPF_Vel	4	70.213	83.420	75.885	2.971	5.941
Ank_IVEV_Po_IC	4	-0.379	15.314	9.006	3.357	6.714
Ank_MaxEV	4	4.498	22.934	15.561	3.928	7.857
Ank_Time_MaxEV	4	25.532	100.000	53.143	16.208	32.416
Ank_MaxIV	4	-5.377	14.778	4.145	5.273	10.546
Ank_Time_MaxIV	4	33.217	85.106	50.461	12.091	24.183
Ank_EV_exc	4	4.872	22.759	11.416	3.921	7.842
Ank_Po_TO	4	-4.540	18.307	10.583	5.205	10.410
Ank_MaxEV_Vel	4	76.005	313.497	161.382	52.741	105.481
Ank_Time_MaxEV_Vel	4	9.433	84.741	31.373	17.967	35.934
Ank_IRER_Po_IC	4	-23.054	19.108	-4.715	8.778	17.557

Ank_MaxER	4	-28.014	3.846	-14.476	6.699	13.399
Ank_Time_MaxER	4	20.213	100.000	57.334	16.552	33.104
Ank_MaxIR	4	-8.119	28.255	1.662	8.880	17.759
Ank_Time_MaxIR	4	1.702	84.043	40.665	17.535	35.069
IRER_Exc	4	7.110	24.410	16.138	3.798	7.595
Ank_IRER_Po_TO	4	-12.611	3.846	-7.720	3.916	7.832
FootProgAng_30percstance	4	-9.581	17.265	4.918	6.944	13.888
Mean_footprogang_stance	4	-4.619	5.846	2.249	2.365	4.729
Ank_MaxER_Vel	4	-451.629	-136.691	-247.140	72.079	144.159
Ank_Time_MaxER	4	4.255	92.352	37.460	19.918	39.836
Knee_Po_IC	4	13.457	26.233	21.134	2.808	5.616
Knee_Max_FLX	4	28.719	43.410	38.994	3.467	6.934
Knee_Time_Max_FLX	4	43.617	58.642	51.142	3.090	6.180
Knee_FLX_Exc	4	8.078	29.953	17.860	4.589	9.177
Knee_Po_TO	4	25.980	32.369	28.070	1.452	2.903
Knee_Max_FLX_Vel	4	263.467	335.431	303.542	17.574	35.149
Knee_Time_Max_FLX_Vel	4	19.149	100.000	46.462	19.013	38.026
Knee_Mean_FLX_Vel	4	67.064	255.283	167.473	38.595	77.190
Knee_Varval_Po_IC	4	-14.196	10.178	-1.282	6.375	12.750
Knee_Max_VAR	4	-6.153	16.246	3.917	5.678	11.355
Knee_Time_Max_VAR	4	7.447	100.000	59.695	23.775	47.551
Knee_VAR_Exc	4	0.697	8.043	5.199	1.633	3.266
Knee_Po_TO	4	-6.153	0.016	-3.651	1.374	2.749
Knee_Max_VAR_Vel	4	78.317	264.314	148.191	41.454	82.908
Knee_Time_Max_VAR_Vel	4	13.620	95.523	49.689	19.276	38.552
Knee_Mean_VAR_Vel	4	20.407	105.684	48.683	19.294	38.587
Knee_IRER_Po_IC	4	-9.582	13.843	5.697	5.214	10.428
Knee_MaxIR	4	9.984	21.029	14.906	2.289	4.578
Knee_Time_MaxIR	4	1.912	82.828	29.318	18.976	37.951
Knee_IR_Exc	4	0.000	24.351	9.209	5.839	11.679
KneeIR_Po_TO	4	1.100	17.842	9.193	3.863	7.726
Knee_MaxIRER_Vel	4	43.047	333.561	215.044	64.880	129.759
Knee_Time_Max_Rot_Vel	4	12.766	70.668	45.021	12.356	24.712
Knee_MeanIRER_Vel	4	-126.074	228.480	51.439	78.572	157.144
Hip_FLXEXT_Po_IC	4	32.649	49.690	41.160	4.481	8.962
Hip_Max_FLX	4	32.649	49.690	41.160	4.481	8.962
Hip_Time_MaxFLX	4	1.702	2.128	1.879	0.094	0.187
Hip_Max_EXT	4	-3.530	5.886	1.176	2.345	4.690

Hip_Time_MaxEXT	4	90.473	100.000	97.618	2.382	4.764
Hip_ Exc_ICtomaxEXT	4	-45.226	-36.179	-39.983	2.237	4.473
Hip_FLXEXT_Po_TO	4	-3.530	5.886	1.518	2.206	4.411
Hip_MaxEXT_Vel	4	-381.328	19.914	-167.007	88.521	177.042
Hip_Time_MaxEXT_Vel	4	29.787	78.328	52.243	11.576	23.152
Hip_MeanEXT_Vel_ICtoMaxEXT	4	-211.035	-166.940	-188.210	9.130	18.259
HipABAD_Po_IC	4	-4.625	-1.086	-2.440	0.768	1.536
Hip_MaxADD	4	-0.607	10.428	6.139	2.375	4.751
Hip_Time_MaxADD	4	29.405	93.982	50.309	14.772	29.545
Hip_MaxABD	4	-6.172	-2.121	-4.108	0.911	1.821
Hip_Time_MaxABD	4	6.539	98.958	58.725	19.970	39.940
Hip_Exc_ICtoMaxABD	4	-3.898	-0.345	-1.668	0.843	1.687
Hip_ABAD_Po_TO	4	-4.437	9.753	-0.395	3.394	6.787
Hip_MaxADD_Vel	4	56.561	295.119	164.788	49.182	98.365
Hip_Time_MaxADD_Vel	4	19.149	44.221	33.153	6.305	12.610
Hip_Mean_Vel_ICtoMaxABD	4	-22.618	25.967	-5.650	10.946	21.892
HipIRER_Po_IC	4	-42.116	16.132	-7.266	13.910	27.819
Hip_MaxIR	4	-10.300	16.929	5.010	6.671	13.342
Hip_Time_MaxIR	4	6.383	100.000	45.563	22.663	45.325
Hip_MaxER	4	-42.116	-3.791	-18.925	8.218	16.437
Hip_Time_MaxER	4	1.702	90.426	43.616	24.225	48.450
Hip_Exc_ICtoMaxIR	4	0.797	31.816	12.276	7.327	14.655
Hip_Exc_MaxIRtoMaxER	4	-31.816	-15.253	-23.935	3.696	7.393
HipIRER_Po_TO	4	-10.300	-2.260	-5.508	1.903	3.806
Hip_MaxIR_Vel	4	161.657	533.721	277.606	86.169	172.338
Hip_Time_MaxIR_Vel	4	37.388	69.241	56.084	7.413	14.826
Hip_Mean_Vel_ICtoMaxIR	4	3.465	148.553	73.294	32.715	65.430
Hip_MaxER_Vel	4	-346.386	-77.954	-238.514	59.329	118.658
Hip_Time_MaxER_Vel	4	29.787	60.948	44.315	7.108	14.216
Max_Pelv_Ang	4	2.779	6.794	5.578	0.940	1.880
Pelv_Ang_Exc	4	7.574	9.829	8.851	0.468	0.937
Max_thor_angle	4	2.087	10.554	5.420	2.605	4.511
Thor_ang_exc	4	5.353	9.066	7.027	1.087	1.883
Max_spine_ang	4	3.600	5.646	4.703	0.596	1.032
Spine_ang_exc	4	14.292	18.496	16.517	1.220	2.113
MaxGRFz	4	18.018	23.900	21.284	1.406	2.811
Time_MaxGRFz	4	36.170	57.406	49.144	4.576	9.152
Load_Rate	4	11964.880	25391.498	18189.559	3092.088	6184.176
GRFimp	4	200.349	383.935	253.283	43.718	87.437

GRFimp_norm	4	0.253	0.321	0.283	0.016	0.032
Max_brake_GRFy	4	-2.398	-1.457	-2.022	0.227	0.453
Time_max_brake_GRFy	4	22.340	41.372	34.075	4.167	8.335
Max_prop_GRFy	4	0.947	2.403	1.506	0.344	0.687
Time_max_prop_GRFy	4	75.747	82.324	80.223	1.521	3.043
Max_Abs_FM	4	0.003	0.008	0.005	0.001	0.002
Max_ADD_FM	4	0.000	0.004	0.002	0.001	0.002
Time_max_ADD_FM	4	6.402	44.681	24.485	8.594	17.188
Max_ABD_FM	4	-0.008	-0.001	-0.004	0.001	0.003
Time_Max_ABD_FM	4	53.505	82.081	69.825	6.992	13.984
FM_Peak_Brake_GRFy	4	-0.007	0.003	-0.001	0.002	0.004
ADD_FM_imp	4	0.000	0.000	0.000	0.000	0.000
Max_DF_Mom	4	2043.837	2835.130	2391.745	164.428	328.857
Time_Max_DF_Mom	4	44.912	59.256	53.155	3.022	6.044
Max_PF_Mom	4	-212.156	-97.427	-145.797	25.375	50.751
Time_Max_PF_Mom	4	3.191	13.952	8.898	2.450	4.901
MaxIV_Mom	4	-13.801	463.664	154.141	108.856	217.712
Time_MaxIV_Mom	4	2.128	62.669	21.746	14.070	28.141
MaxEV_Mom	4	-787.375	-6.105	-398.126	173.610	347.220
Time_MaxEV_Mom	4	28.723	99.415	60.984	14.600	29.199
Ank_MaxIR_Mom	4	36.139	281.462	193.223	57.896	115.793
Time_MaxIR_Mom	4	6.605	65.586	45.570	13.299	26.599
MaxER_Mom	4	-134.517	-40.023	-73.648	22.087	44.175
Time_MaxER_Mom	4	9.484	98.936	44.669	21.166	42.332
Max_FLX_Mom_LR	4	18.743	3110.223	1071.297	711.058	1422.117
Time_Max_FLX_Mom_LR	4	-89.206	50.388	-33.061	29.817	59.633
Max_FLX_Mom_TO	4	1375.543	3110.223	2305.678	377.376	754.753
Time_Max_FLX_Mom_TO	4	41.489	52.202	47.578	2.383	4.767
Max_EXT_Mom	4	-384.968	-231.876	-332.590	34.842	69.685
Time_Max_EXT_Mom	4	1.702	100.000	34.534	23.144	46.289
Knee_ Stiff	4	0.321	15.100	5.513	3.376	6.752
Knee_MaxADD_Mom	4	181.476	1656.040	919.920	374.682	749.364
Knee_Timing_MaxADD_Mom	4	12.730	52.663	40.772	9.476	18.951
Knee_MaxABD_Mom	4	-552.441	-180.320	-308.475	83.515	167.029
Knee_Time_MaxABD_Mom	4	40.769	100.000	71.980	16.072	32.144
Knee_MaxIR_Mom	4	33.296	349.022	231.109	75.266	150.533
Knee_Time_MaxIR_Mom	4	6.605	63.860	46.096	13.279	26.558
Knee_MaxER_Mom	4	-56.967	-27.252	-41.978	6.562	13.123
Time_MaxER_Mom	4	3.764	98.936	49.640	25.303	50.606

Hip_Max_FLX_Mom	4	268.890	745.159	588.085	110.808	221.617
Hip_Time_MaxFLX_Mom	4	5.046	25.532	12.416	4.524	9.048
Hip_MaxEXT_Mom	4	-2467.157	-1429.363	-1876.138	222.313	444.626
Hip_Time_MaxEXT_Mom	4	66.784	100.000	78.155	7.506	15.011
Hip_MaxADD_Mom	4	980.538	1681.663	1214.435	162.566	325.131
Hip_Time_MaxADD_Mom	4	11.702	49.006	30.278	9.147	18.294
Hip_MaxABD_Mom	4	-860.539	-24.175	-362.956	176.776	353.552
Hip_Time_MaxABD_Mom	4	27.836	97.872	67.333	14.649	29.298
Hip_MaxIR_Mom	4	-5.605	91.254	46.034	20.741	41.481
Hip_Time_MaxIR_Mom	4	41.981	100.000	82.458	13.793	27.586
Hip_MaxER_Mom	4	-305.768	-91.389	-218.524	52.143	104.286
Hip_Time_MaxER_Mom	4	20.935	50.218	37.216	6.056	12.113
Inv_A1_Joint_Power	4	-3.370	-1.018	-2.299	0.507	1.014
Inv_A2_Joint_Power	4	4.022	6.317	5.033	0.512	1.023
Inv_K1_Joint_Power	4	-9.064	-0.799	-5.661	1.753	3.506
Inv_K2_Joint_Power	4	2.629	12.302	7.684	2.046	4.092
Inv_H1_Joint_Power	4	0.132	1.266	0.751	0.270	0.540
Inv_H2_Joint_Power	4	-9.744	-3.673	-7.554	1.361	2.722
Inv_Swing_Time	4	0.458	0.500	0.481	0.012	0.021
Uninv_Swing_Time	4	0.454	0.469	0.461	0.007	0.010
Inv_Stance_Time	4	0.217	0.258	0.235	0.010	0.020
Uninv_Stance_Time	4	0.269	0.301	0.285	0.008	0.016
Inv_Step_Length	4	1.072	1.479	1.288	0.086	0.172
Uninv_Step_Length	4	1.177	1.422	1.285	0.054	0.107
Stride_L	4	2.249	2.791	2.572	0.130	0.259
Velocity	4	3.004	4.000	3.614	0.216	0.433

Three service members completed at time 6, one at time 5 due to medical retirement, one not included due to disenrollment at month two, and one pending completion April 2013.

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