

CASE STUDY

**Plantarflexion Resistance
of Selected Ankle-Foot Orthoses:**
*A Study of Commonly Prescribed
Prefabricated and Custom-Molded Alternatives*

William W. DeToro, CO



1399 E. Western Reserve Road, Poland, OH 44514-3250 • (800) 837-3888
www.anatomicalconceptsinc.com

Ankle-foot orthoses (AFOs) are one of the most commonly prescribed lower-limb orthoses, largely because of their versatility in providing selected biomechanical controls to augment impaired functions. They are also generally well tolerated by those who wear them and therefore are often associated with a positive clinical outcome.

Despite their widespread clinical application, only a limited amount of objective data have been published regarding the control offered by specific AFO designs. This study was conducted to objectively document, in a simple manner, the resistance to plantarflexion of a variety of commonly prescribed AFOs. These data may prove useful in verifying the effectiveness of such orthoses when they are used to facilitate ambulation for individuals with neuromuscular impairments and in developing prescription criteria for their clinical application.

Previous authors have shown that the sagittal plane stiffness of the tested AFOs varies widely^{1,2} and that the resistance provided significantly affects pathologic gait.³ It has also been shown that, for a given individual with a gait disorder, the range of acceptable resistances is fairly narrow.⁴ These recent objective data support the concept of providing functional orthoses based on the patient's biomechanical deficits, or "biotechnical matching," as originally termed by Sarno and Lehnis.⁵

No definitive method for measuring load deflection characteristics of orthoses has been established,⁴ although there is general consensus in the literature that sagittal plane stiffness is of particular clinical importance. The proliferation of custom designs and prefabricated variants in recent years has made it increasingly difficult to judge the effectiveness of a particular AFO before completion of the fitting. For this reason, we investigated the use of a simple digital tensiometer to objectively quantify the plantarflexion resistance supplied at the forefoot for a variety of common orthoses, as suggested by Rubin and Dixon⁶ in 1973.

Materials and Methods

To ensure that the orthoses tested were representative of actual devices that might be provided in a clinical setting, a healthy 59-year-old man, 6 feet 3 inches tall and weighing 245 pounds, was recruited as a volunteer subject. A negative impression was taken of his right lower leg in the conventional manner, using fiberglass casting tape wrapped over a single layer of cotton stockinet. His foot was placed in a neutral position on a casting plate to simulate the footbed of a conventional shoe with a 12-mm (half-inch) heel rise. Routine circumferential and linear measurements of the casted limb were also recorded.

The negative impression was filled with plaster to create a positive model that was then rectified in accordance with standard industry practice. All casting artifacts and stockinet marks were removed, and the malleolar regions were built up with plaster patches to create slight relief areas over these bony prominences.

The final model, which may be considered representative of a typical large adult male lower leg, was then duplicated in rigid polyurethane foam over a plywood core and reinforced externally with a polyester resin vacuum-bag lamination over a single layer of nylon stockinet. The laminated model was then used to create the test apparatus, which was similar to that described by Golay et al.²



Tested Orthoses

The original plaster positive model was then used to fabricate eight different custom-made AFOs of varying biomechanical designs, as described in Table 1. Three of the custom orthoses were the minimal-contact sidebar and band type, and five were total-contact thermoplastic orthoses. All were constructed in accordance with industry-standard fabrication methods using readily available components and materials that are routinely incorporated in typically prescribed orthoses.

Ten prefabricated AFOs were also fitted to the positive model. The prefabricated orthoses selected represent a range of commonly prescribed devices.

All orthoses were fitted and adjusted to the test model by an experienced orthotist who is certified by the American Board for Certification in Prosthetics and Orthotics. The prefabricated devices were adjusted in accordance with the manufacturer's written instructions and guidelines.

Test Apparatus

The test apparatus was constructed by placing a single-axis hinge in the transmalleolar region of the wood-reinforced foam model of the subject's lower leg. Sufficient material was removed so that the model could bend freely from 20° of dorsiflexion to 20° of plantarflexion to simulate normal sagittal plane ankle motion in this range.

The shank of the articulated leg model was mounted rigidly to the back of the test stand so that the foot segment could move freely, as shown in Figure 1. The various orthoses provided the resistance to plantarflexion motion. Plantarflexion motion was measured by placing a stainless steel band across the dorsum of the model in the region corresponding to the metatarsal heads, located 25.4 cm (10 inches) anterior to the ankle joint for this individual.

A MARK-10 digital force gauge with certified calibration (MARK-10 Corporation, Hicksville, NY) was connected from the steel band to a screw drive, which was used to apply a distraction force that would cause the model to plantarflex. A liquid-filled angle measurement device was attached solidly to the plantar surface of the foot segment so that it read 0.0° when the midline of the foot segment was at a right angle to the midline of the shank



Figure 1: The model, with a single-axis hinge at the malleolar level, permits free movement from 20° of plantarflexion to 20° of dorsiflexion and is anchored rigidly to a vertical stand. Test orthoses were attached rigidly to the calf segment; static dorsiflexion resistance was measured at the metatarsal heads, to simulate the clinical effect of walking with an AFO intended to compensate for dorsiflexion weakness. A digital goniometer measured the deflection angle, and a digital tensiometer recorded the static force applied at the metatarsal head region.

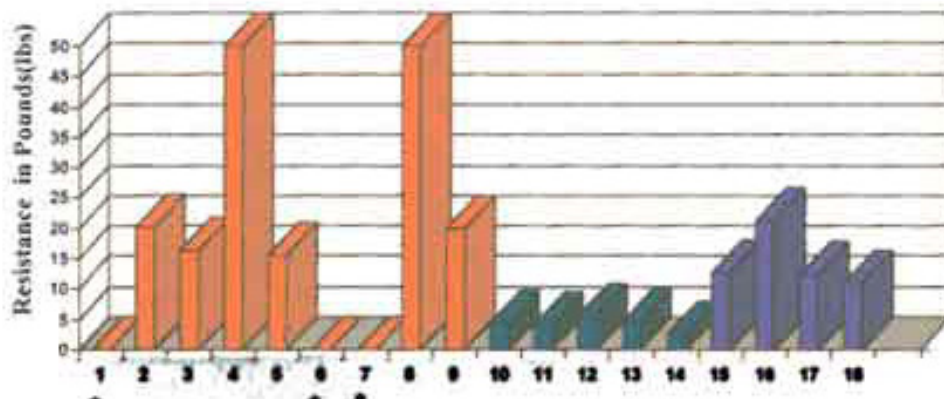
segment. As the foot segment was plantarflexed, the measured angle increased up to 20°, which was the maximum movement allowed by the test apparatus.

Testing

To eliminate any measurement artifacts due to compression of soft interfaces, all materials lining the orthoses were removed and the calf section was rigidly bolted to the shank portion of the model. The screw jack was then used to deflect the foot portion of the model in 1° increments, from 0° to 10° of plantarflexion, similar to the method of Sumiya et al.⁴ The resistance supplied by each orthosis was recorded for a total of 10 trials. Figure 2 illustrates the average resistance provided by each orthosis at 10° of plantarflexion.

A second set of measurements was performed on all orthoses that could be altered from a neutral position to one that held the foot segment in 10° of dorsiflexion. This change in position was accomplished by adjusting the stops, bending the uprights, or thermal remolding of the orthosis. Figure 3 shows the average resistance generated at 10° of plantarflexion for those orthoses that could be tested in this manner.

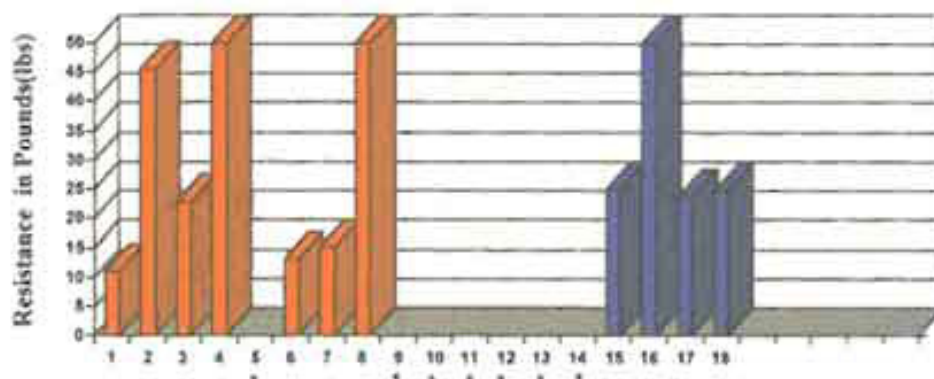
Figure 2



Orthosis Tested	Average Static Resistance	Standard Deviation
1. Klenzak, with spring	Not tested	*
2. Klenzak, with pin	20.1	0.866
3. Phelps	16.4	0.170
4. Custom thermoplastic AFO, with solid ankle	52.0	0.361
5. Custom thermoplastic AFO, with leaf spring trimline	16.6	0.169
6. Custom thermoplastic AFO, with Tamarack joint	Not tested	*
7. Custom thermoplastic AFO, with OK joints, DF assist	Not tested	*
8. Custom thermoplastic AFO, with OK joints, PF stop	50.4	0.347
9. Custom fitted prefab AFO, 3 mm polypropylene	20.0	0.200
10. Recumbent splint, Select Medical Products	4.3	0.157
11. Recumbent splint, Flexboot 01FB2	4.1	0.456
12. Recumbent splint, New Age Oscar	5.6	0.085
13. Recumbent splint, Multi-Podus 10MP	4.7	0.173
14. Recumbent splint, EZ-Boot #10320	2.6	0.084
15. PRAFO, standard	12.6	0.149
16. PRAFO, heavy duty	21.3	0.508
17. PRAFO, APU	12.4	0.092
18. PRAFO, stainless steel	11.0	0.483

* Note: Orthoses 1, 6 and 7 could not be tested from a position of 90 degrees since they are preset in 10 degrees of dorsiflexion.

Figure 3



Orthosis Tested	Average Static Resistance	Standard Deviation
1. Klenzak, with spring	11.1	0.165
2. Klenzak, with pin	45.4	1.055
3. Phelps	22.8	1.159
4. Custom thermoplastic AFO, with solid ankle	52.0	0.361
5. Custom thermoplastic AFO, with leaf spring trimline	Not tested	*
6. Custom thermoplastic AFO, with Tamarack joint	13.0	0.769
7. Custom thermoplastic AFO, with OK joints, DF assist	15.1	0.586
8. Custom thermoplastic AFO, with OK joints, PF stop	50.4	0.347
9. Custom fitted prefab AFO, 3 mm polypropylene	0.0	0.000
10. Recumbent splint, Select Medical Products	Not tested	*
11. Recumbent splint, Flexboot 01FB2	Not tested	*
12. Recumbent splint, New Age Oscar	Not tested	*
13. Recumbent splint, Multi-Podus 10MP	Not tested	*
14. Recumbent splint, EZ-Boot #10320	Not tested	*
15. PRAFO, standard	24.9	0.797
16. PRAFO, heavy duty	52.0	1.557
17. PRAFO, APU	23.8	0.955
18. PRAFO, stainless steel	24.7	1.086

* Note: Orthoses 5 and 9-14 could not be tested in this configuration as they were fabricated in a neutral attitude.

Results

As might be expected, the resistance provided by the orthoses increased as the foot segment was pulled in the direction of plantarflexion, whether the starting point was from 10° of dorsiflexion or from the neutral (0°) position. Furthermore, those orthoses that could be adjusted to a more acute dorsiflexion angle all demonstrated greater resistance to plantarflexion motion than they did when adjusted to a neutral position.

Although the range of resistances measured was large, the orthoses could be classified into three relatively distinct groups based on this parameter. Since the magnitude of the resistance to plantarflexion directly affects the amount of toe clearance in midswing for individuals with pretibial muscle group deficiencies, this grouping may have clinical significance.

Table 1: Characteristics of the tested orthoses

Sample #	Manufacturer	Material Used	Ankle Control	Source for Joints
1	Custom made	Aluminum & leather	Klenzak joint, spring loaded	USMC #211-140-802
2	Custom made	Aluminum & leather	Klenzak joint, with solid rod	USMC #211-140-802
3	Custom made	Aluminum & leather	Thrust bearing Phelps joint, 90 degree stop	USMC #19017
4	Custom made	4 mm thermoplastic polypropylene-polyethylene copolymer	Solid ankle trimline	Not applicable
5	Custom made	4 mm thermoplastic polypropylene-polyethylene copolymer	Posterior leaf spring trimline	Not applicable
6	Custom made	4 mm thermoplastic polypropylene-polyethylene copolymer	Tamarack #742-L-85 ankle joints [dorsiflexion assist]	Becker
7	Custom made	4 mm thermoplastic polypropylene-polyethylene copolymer	Becker 760-L Oklahoma Ankle Joints; Precision O&P Elite Line PSA-100-A adjustable dorsiflexion spring assist	Becker Precision O&P
8	Custom made	4 mm thermoplastic polypropylene-polyethylene copolymer	Becker 760-L Oklahoma Ankle Joints; Precision O&P Elite Line PAS-100-A adjustable plantar flexion stop	Becker Precision O&P
9	Orthomerica OA-3547-01	3 mm polypropylene homopolymer plastic	Trimline at ankle midline [limited motion]	Not applicable
10	Select Medical Products #001203	Low temperature Kydex thermoplastic	Not adjustable	Not applicable
11	Flexboot-01FB2 Flexboot Ortho	Low temperature Kydex thermoplastic	Not adjustable	Not applicable
12	New Age Oscar Orthosis Corrective Systems			
		Low temperature Kydex thermoplastic	Not adjustable	Not applicable
13	Multi-Podus-10MP Restorative Care of America	Low temperature Kydex thermoplastic	Not adjustable	Not applicable
14	EZ-Boot #10-320 Orthotic Rehab	Unspecified thermoplastic	Not adjustable	Not applicable
15	PRAFO®-650 Anatomical Concepts, Inc.	Injection-molded polypropylene, 1/8" anodized aluminum upright	Standard aluminum posterior bar	Not applicable
16	PRAFO®-650HD Anatomical Concepts, Inc.	Injection-molded polypropylene, 3/16" anodized aluminum upright	Heavy duty aluminum posterior bar	Not applicable
17	PRAFO®-650APU Anatomical Concepts, Inc.	Injection-molded polypropylene, 1/8" extruded aluminum upright	Adjustable aluminum posterior bar	Not applicable
18	PRAFO®-650SS Anatomical Concepts, Inc.	Injection-molded polypropylene, 1/8" spring steel upright	Spring steel posterior bar	Not applicable



Discussion

Paralysis or paresis of the pretibial muscles can result from traumatic injuries as well as a variety of neuromuscular disorders. The resulting inability to maintain toe clearance during swing phase is probably one of the most common reasons for the prescription of lower-limb orthoses. Impairment of the pretibial muscles also compromises the patient's ability to control the rate of descent of the foot during loading response and typically results in a characteristic "foot slap" gait.

A broad range of orthoses may be used to treat these biomechanical losses. The clinician must select a design that provides sufficient plantarflexion resistance to effectively decelerate the foot in early stance as well as maintain at least a neutral ankle-foot attitude in swing phase. However, if the orthosis provides too much resistance to plantarflexion, ankle motion will be inhibited throughout loading response and, as a result, the normal shock-absorbing mechanism is disrupted and knee stability is reduced. The challenge, therefore, is to apply sufficient resistance to eliminate the gait pathology without introducing additional aberrations.

It should be noted that the amount of plantarflexion resistance required clinically is also dependent on the gait characteristics of each individual. For example, a 200-pound individual will transfer twice as much body weight onto the affected limb in early stance as someone weighing 100 pounds. If one assumes that the two individuals are walking at the same velocity, this means that the lighter person will require less assistance from the orthosis. The range of available hip, knee, and ankle motion can also influence the resistance needed, as can such variables as stride length and weight of the footwear being worn. The multiple factors that affect optimal plantarflexion resistance are precisely the variables that the clinician must consider during the fitting process, and at present, iterative walking trials are the only practical method to determine the final resistance value for a specific individual.

In this study, the tested orthoses fell into three distinct groupings based on the average static resistance to plantarflexion. The first grouping can be termed the plantarflexion stop (PFS) orthoses because they all provided at least 20 pounds of resistance when set initially in a neutral position. Clinical observation has confirmed that these devices significantly limit plantarflexion motion, and their use is recommended only when such a reduction in the range of motion at the ankle is desired. The PFS orthoses were all custom-made devices, with the exception of the prefabricated solid ankle plastic AFO (sample 9) and the heavy-duty PRAFO® orthosis (sample 17; Anatomical Concepts, Inc., Boardman, OH). When adjusted to an initial alignment in 10° of dorsiflexion, these orthoses all provided at least 45 pounds of resistance in the measured position.

The second grouping of orthoses provided 10 to 16 pounds of resistance in neutral and 20 to 25 pounds of resistance from an initial angle of 10° of dorsiflexion. These devices would effectively dampen plantarflexion during loading response without unduly restricting ankle motion, thereby restoring some of the normal shock absorption that results from the interaction between knee and ankle movements in early stance. They would also maintain good toe clearance throughout the swing phase. This grouping could be termed the plantarflexion resistance (PFR) orthoses. They are preferable to the more rigid devices in the first group, unless there are



specific reasons to restrict this ankle motion. The PFR group included many of the custom-made orthoses and all but one of the PRAFO® orthoses.

The third group provided no more than 5 pounds of resistance from a neutral position and included all the tested prefabricated recumbent splints. This group may be termed the nonambulatory (NA) devices, because clinical experience has shown that such minimal resistance is insufficient to assure toe clearance or to significantly decelerate the foot in early stance. However, these devices may be quite suitable to maintain the foot in a neutral position when the patient is recumbent, because the force of gravity pulling the foot into plantarflexion is minimized in that posture.

It should be noted that this study looked only at one aspect of AFO function: resistance to sagittal plane movement toward plantarflexion. These data provide no information about resistance to dorsiflexion or resistances in other planes and should therefore be interpreted with caution. Klasson et al⁷ have documented clearly that plastic AFOs provide resistance in all planes even when prescribed primarily for single-plane control.

Furthermore, the test condition measures only static forces and the orthoses were much more solidly attached to the calf segment of the test fixture than if they were being worn by real patients. Because dynamic forces generated by human subjects wearing the AFOs were not measured in this study, caution must be used in drawing inferences about clinical prescription criteria.

Golay et al² used a measurement strategy similar to ours but looked exclusively at dorsiflexion resistance. They showed that for custom-made, polypropylene AFOs, such variables as the final wall thickness of the plastic and the degree of malleolar build-up significantly affected the amount of resistance to dorsiflexion. Sumiya et al⁴ reported that for the flexible plastic AFOs tested, the overall resistance to both plantarflexion and dorsiflexion increased almost in proportion to the width of the posterior portion of the device. Singerman et al⁸ recently published a more comprehensive look at four AFO types and noted that changes to the trimlines intended primarily to alter the resistance inevitably altered the effective axis of rotation of the device as well.

These studies all underscore the fact that the movement of plastic ankle-foot orthoses is multiplanar, and the resistance in each plane varies due to a variety of interactive factors. Although it may be useful to focus on single-plane forces to improve our conceptual understanding of how these devices function, such limited data cannot provide a complete picture of the clinical performance of the orthoses.



Conclusion

This study found that commonly prescribed contemporary AFO designs can be grouped according to the maximum static resistance provided when the AFO is deflected to 10° of plantarflexion. The designs that stopped plantarflexion movement all provided the greatest magnitude of measured resistance and were predominantly custom-made devices. The group that provided intermediate levels of resistance but also allowed a significant range of ankle motion included custom-made devices and all except the heavy-duty PRAFO® orthosis designs. The devices that provided only minimal resistance were all nonadjustable, prefabricated splints best suited for nonambulatory applications.

Prescribing physicians, orthotic clinicians, and reimbursement authorities may be able to use such objective data to help distinguish among orthoses that are superficially similar in appearance but offer distinctly different biomechanical advantages. In principle, it may be possible in the future to use the results from instrumented analysis of an individual's pathologic gait pattern to specify the desired amount of plantarflexion resistance for the orthosis so that it is no longer necessary to determine this by subjective means during iterative clinical walking trials. That is, if the missing ankle moments can be specified, then it should be possible to design an orthosis that offers an equivalent corrective force. Measurement and publication of a much broader array of data about the resistances and motion provided by lower-limb orthoses in all planes may be helpful in increasing objectivity in the prescription and design of such devices.

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About William W. DeToro, CO

William W. DeToro, CO, is Vice President of and an orthotic practitioner for ABI Orthotic & Prosthetic Labs, Ltd., Boardman, OH. The author has a proprietary interest in Anatomical Concepts, Inc., manufacturer of the PRAFO® orthosis. William W. DeToro, CO, ABI Orthotic & Prosthetic Labs, Ltd., 930 Trailwood Drive, Boardman, OH 44512-5007. Phone: (330) 758-1143; Fax: (330) 758-2361; E-mail: wwdco@aol.com.

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