

SELECTIVE LASER SINTERING OF PASSIVE DYNAMIC ANKLE-FOOT ORTHOSES

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Abstract

Passive dynamic ankle-foot orthoses (AFO's) are used to improve gait performance in those with various neuromuscular disorders. An important design characteristic of passive dynamic AFOs is the storage and release of elastic energy within its structure to help satisfy the energetic demands of walking. Thus, minimizing energy dissipation through internal friction is a fundamental criterion for selecting the appropriate AFO material. This study compared the mechanical damping of a carbon-fiber AFO to three geometrically identical AFO's fabricated using selective laser sintering with different materials. Mechanical damping characteristics ranked the materials as Nylon 11 (best), followed by DuraformTM PA and DuraformTM GF (worst).

Introduction

Passive dynamic ankle-foot orthoses (AFOs) are often prescribed to improve gait performance for those with various neuromuscular disorders. Recent biomechanical analyses of AFOs have shown an increase in walking speed and overall gait performance for patients with post-polio syndrome [1, 2]. Designs and materials used for AFOs range from simple polypropylene braces to advanced custom carbon fiber designs. AFO designs (Fig. 1) vary in the shape and length of the foot component as well as the stiffness and length of the tibial component depending on the desired functional outcomes. Selective Laser Sintering (SLS) is an ideal process for AFO fabrication because of its inherent customizing capabilities, which make fabrication of subject-specific AFOs feasible. In addition, SLS has already been successfully used in the fabrication of prosthetic sockets for lower limb amputees [3, 4].



Figure 1: Carbon fiber AFO design (Dynamic Brace, Advanced Prosthetics and Orthotics Inc.)

Patients with various neuromuscular disorders often lose control and strength in their lower-limbs, which impairs their mobility. The storage and release of elastic energy within the structure of passive dynamic AFOs is an important design characteristic that helps compensate for the neuromuscular disorders. During walking, the AFO deforms elastically under the influence of the body weight, thus storing elastic energy. During the second half of the stance when the AFO is unloaded, the AFO releases the stored energy to help satisfy the energetic demands of walking. Thus, a fundamental criterion for selecting the most appropriate SLS material for AFO fabrication is minimizing energy dissipation through internal friction during AFO deformation. The present study evaluated the energy dissipation characteristics in three nylon-based SLS materials: DuraformTM PA (Nylon 12), DuraformTM GF (glass-filled Nylon 12) and RilsanTM D80 (Nylon 11). The performance of AFOs's fabricated with these materials was compared directly to the carbon fiber Dynamic Brace.

Methods

A previously manufactured carbon fiber AFO design (Fig. 2A) was digitized and a corresponding point cloud file (Fig. 2B) was reconstructed using Raindrop GeomagicTM (Fig. 2C). The resulting model was used to generate a surface model with RhinocerosTM 3D (Fig. 2D). Three models were created from this file, one for each SLS material to be tested. All resulting solid models (Fig. 2E) preserved the overall geometric characteristics of the original carbon fiber AFO. However, the thicknesses of the struts in each model were altered such that the rotational stiffness of the resulting AFO in the sagittal plane (Fig. 3) matched that of the carbon fiber AFO. This was achieved through Finite Element Method (FEM) analyses using EDS I-ideasTM.

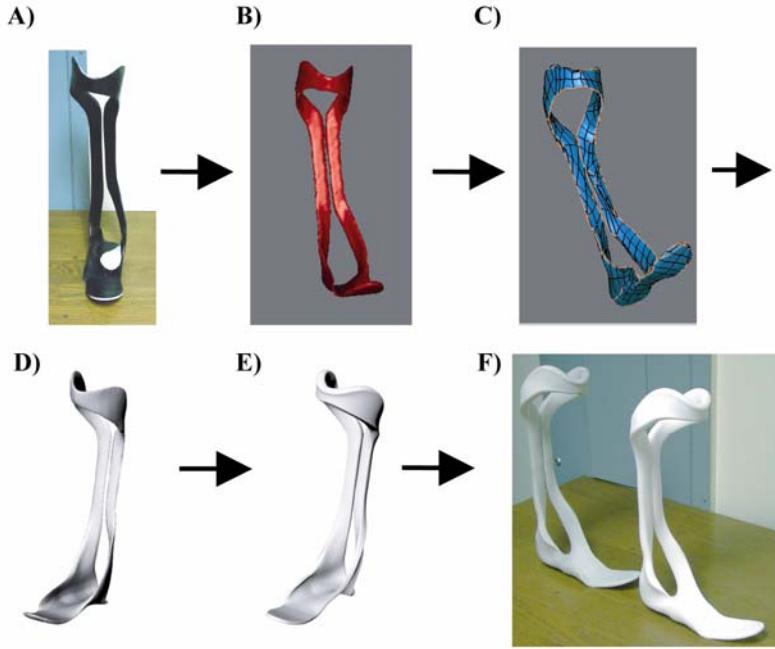


Figure 2: Steps for modeling and fabrication of SLS AFOs to be tested from original carbon fiber AFO. A) original carbon fiber AFO design (Advanced Prosthetics and Orthotics Inc.) to be replicated with SLS; B) corresponding triangulated point cloud from the digitized AFO; C) reconstructed point cloud; D) surface model; E) solid model; and F) SLS manufactured AFO prototypes.

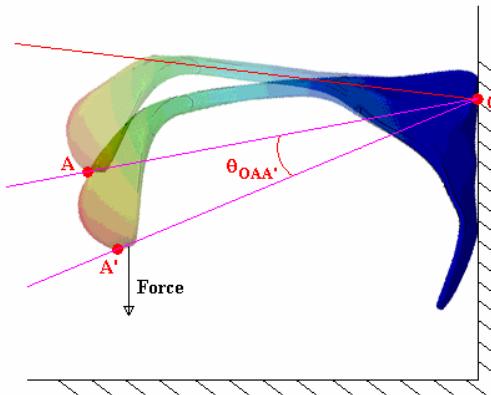


Figure 3: Rotational stiffness of the AFO in the sagittal (viewing) plane, defined as the ratio between the applied force and the resulting deformation angle $\theta_{OAA'}$.

The files were converted to STL format and AFOs were fabricated in a 3DSystems Vanguard HS Sinterstation using DuraformTM PA, DuraformTM GF and RilsanTM D80 material. Since the modeled AFOs were 17.7 inches tall, each part occupied the entire 18-inch build envelope during the SLS build (due to the build height space lost to warm up and cool down powder) and had to be positioned with the struts in the diagonal direction of the build volume to fit within the build boundaries of the SLS workstation. This part orientation was, however,

consistent across builds for all different evaluated materials. Each build took approximately 26 hours in total, including warming-up and cooling-down stages. An additional 6 hours were necessary, after finishing the build, for the part bed temperature to drop below 50 C and allow safe removal of the parts. The part bed was broken out and the AFO was then cleaned with a sandblaster. At this point, the AFO was ready to be tested (Fig. 2F).

The energy dissipation was measured in all four AFOs (i.e., the original carbon fiber AFO and three SLS prototypes). The dissipation measurement consisted of securely attaching to the cuff of each AFO a long iron rod, which had a lumped mass at the distal end (Figs. 4A and B) to produce a rigid AFO-rod system. The iron rod was 800 mm tall and weighed 9 kg (including the distal lumped mass). Each AFO was then fixed on a table and a calibrated accelerometer (Crossbow CXL04P3) was attached to the side of the cuff and connected to a data acquisition system (Fig. 4C). The AFO was rotated 20 degrees in the sagittal plane and released. Accelerometer data was collected during the following 30 seconds. Five measurement trials were performed for each AFO.

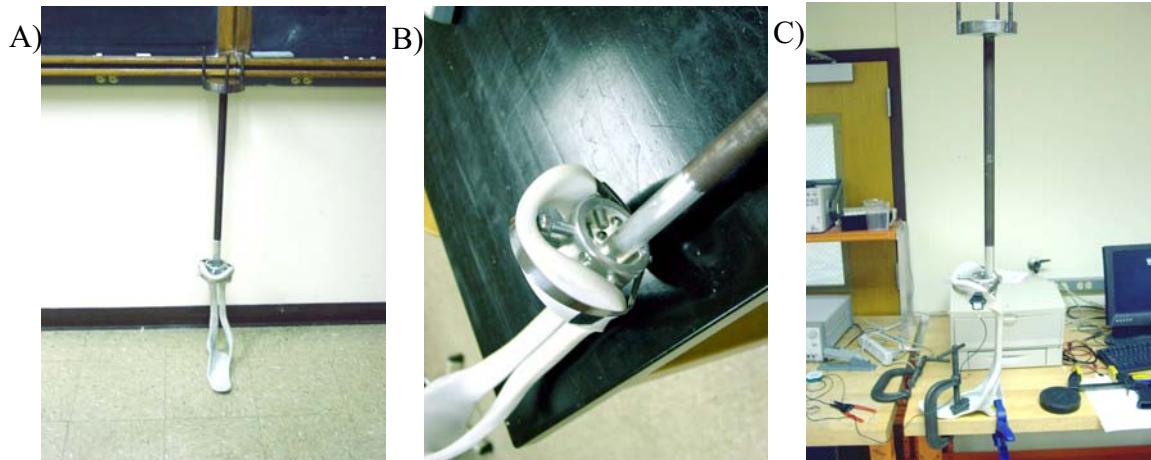


Figure 4: Energy dissipation measurement set up.

From the resulting acceleration-time trajectory, energy dissipation was quantified by the decay factor d of the best-fit envelope curve $A \cdot e^{-d \cdot t}$ where t is time. The initial 5 seconds of acquired data were discarded for the computation of the envelope curve to allow small out-of-plane vibrations to subside. For each AFO, d was averaged over the five measurement trials.

Results

The processed energy dissipation data (Fig. 5) for the carbon fiber AFO had a decay factor $d = 0.0488$, which was the lowest amount of energy dissipation of the four AFOs tested. DuraformTM GF had the largest energy dissipation, with $d = 0.3115$. DuraformTM PA had a decay factor $d = 0.08716$, while RilsanTM D80 had a $d = 0.06655$, the lowest amount of dissipation among the SLS fabricated AFOs. In all four cases, deviations in the averaged d values were similar and below 5%.

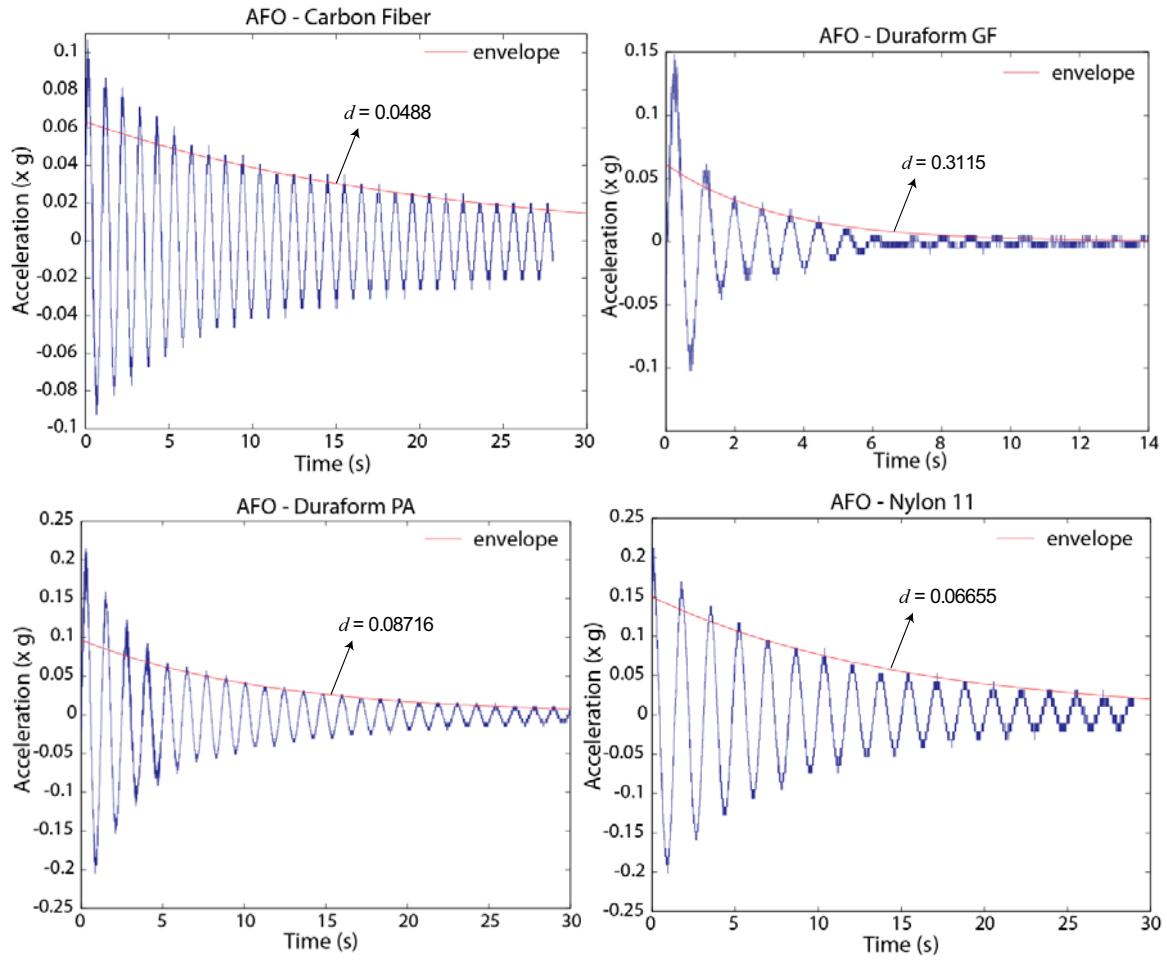


Figure 5: For each tested AFO, a graph showing acceleration (times gravity acceleration) vs. time (in seconds) and the best-fit envelope curve for which the decay factor d was calculated.

Discussion

Mechanical damping characteristics ranked the materials as Nylon 11 (best), followed by Duraform™ PA and Duraform™ GF (worst). However, the energy dissipated in the Rilsan™ D80 AFO was still significantly higher (36%) than the carbon fiber AFO, which may result in patient fatigue during long-term AFO use. Since minimizing damping in a dynamic AFO is an important characteristic, future work should be directed at optimizing the SLS-based AFO design itself (e.g., using topology optimization) to minimize energy losses. However, although different AFO designs could lead to a decrease in energy losses, the ranking of mechanical damping characteristics should remain similar among the materials analyzed. Another aspect that could alter the amount of energy loss is the part orientation during the SLS fabrication. Since the SLS process is a layer based process, and even within the layer plane the material is not expected to be perfectly uniform due to persistent laser scanning direction, a change in the part orientation alone could be enough to modify internal friction directions in the SLS AFOs. Thus, future work

should be directed towards studying the influence of build orientation on the mechanical damping characteristics of SLS AFO's.

Conclusion

A testing procedure to quantify mechanical energy dissipation in SLS fabricated AFOs was successfully implemented. This procedure allowed three SLS AFOs to be fabricated with different materials and evaluated relative to the dynamic performance of a conventionally fabricated carbon fiber AFO. Nylon 11 proved to be the SLS material that exhibited the least amount of energy dissipation through mechanical damping.

Future work should be dedicated to studying the effects of build orientation on the mechanical damping characteristics of SLS AFO's. Because of the limited build envelope of most commercially available SLS workstations compared to a full-sized AFO's usual dimensions, the AFO must be positioned with its struts aligned with the diagonal direction of the build volume. Thus, there are limited orientations possible for the AFO during SLS fabrication. Fabricating AFO's in various orientations and using the test method described in this study is a promising approach for this research.

Acknowledgements

This work was supported by the Physical Disabilities Branch of the National Institutes of Health.

References

- [1] Halstead, LS, Crittenden, DT, Nielsen, J-P, Loke, M. Dynamic bracing: a novel approach to lower extremity bracing for individuals with incomplete spinal cord injury and other neuromuscular disorders. *J Spinal Cord Med* 26 (1): 16, 2003.
- [2] Nelson KM, Kepple TM, Siegel KL, Halstead LS, Stanhope SJ: Ankle foot orthosis contribution to net ankle moments in gait. *Proceedings of American Society of Biomechanics*, 2003.
- [3] Faustini, M., Neptune, R.R., Crawford, R., Rogers, W., and Gordon Bosker. An experimental and theoretical framework for manufacturing prosthetic sockets for transtibial amputees. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* (in press).
- [4] Faustini, M., Crawford, R., Neptune, R.R., Rogers, W.E. and Bosker, G. Design and analysis of orthogonally compliant features for local contact pressure relief in transtibial prostheses. *Journal of Biomechanical Engineering* 127(6): 946-951, 2005.