Clinical Prosthetics and Orthotics, Vol. 11, No. 1, pp. 55–62


# Motion Analysis of SACH vs. Flex-Foot™ in Moderately Active Below-knee Amputees

by Judy Wagner, L.P.T.
Susan Sienko, B.Sc.
Terry Supan, C.P.O.
Daryl Barth, C.P.O.

## INTRODUCTION

Energy storing prosthetic feet have demonstrated clinical advantages for active below-knee amputees. However, indications for use of energy storage systems have not been established for less active, low velocity ambulators. The intent of this research was to determine if the general below-knee amputee population would benefit biomechanically from an energy storage prosthetic foot system.

A group of moderately active individuals, who could be referred to as non-vigorous ambulators, participated in the study. A physical exam (to rule out musculoskeletal causes of gait deviation) was followed by motion analysis. SACH and Flex Foot™ were analyzed. Statistically significant improvements were found in two important areas.

## CASE STUDIES

Six male below-knee amputees between the ages of 23 and 56 participated in the study. The gait of three subjects was studied with both SACH foot and Flex-Foot™ prostheses. Direct SACH/Flex-Foot™ comparisons were made from this group (n = 3). Three additional subjects were analyzed with the Flex-Foot™ (n = 6). They enabled the investigators to determine the consistency of trends.

All subjects were asked to walk at their own comfortable pace, therefore, the results represent the biomechanics of walking. They do not reflect the action of the Flex-Foot™ during intense activities or vigorous ambulation.

## RESULTS

### *Linear Gait Parameters*

There were no differences found in velocity (walking speed) and cadence (number of steps per minute) between the Flex-Foot™ and SACH foot trials. Both variables were below normal values. This finding is true for a broad population of below-knee amputees as evidenced by other researchers.[9] It indicates that our population is representative of low intensity ambulators.

Two periods of single limb stance exist within a gait cycle. (One period is on the prosthetic side; the other is on the sound limb side.) One hundred percent of a gait cycle is from heel contact to ipsilateral heel contact. The percentage of single limb stance spent on the prosthesis did not vary when the Flex-Foot™ or SACH foot prostheses were worn. Likewise, the percent of the gait cycle spent on the sound limb did not vary between the Flex-Foot™ and SACH trials. However, single limb stance time was more symmetrical between the prosthetic and sound limbs when the Flex-Foot™ was worn.

Double limb stance is a period of weight transfer when both feet are on the ground. The total percent of time spent in double limb stance did not change between Flex-Foot™ and SACH trials.

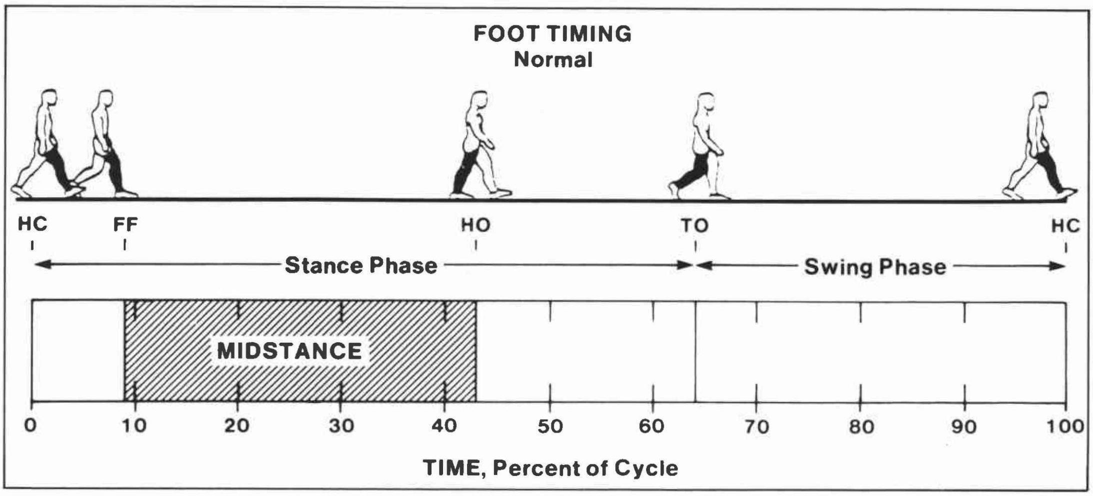


Figure 1. Normal foot timing. The periods of stance represented as a percent of the gait cycle: heel contact (HC), foot flat (FF), heel off (HO), and toe off (TO).[1,2,17]

Linear gait parameters did not change (between the Flex-Foot™ and SACH foot) for moderately active persons. The similarities found demonstrate the adaptability of the human body to the loss of a limb regardless of the type of prosthetic device. However, other variables such as joint rotations, foot timing, and force largely reflect the design and materials of the prosthesis.

## *Foot Timing and Joint Rotations*

Stance has been delineated by the periods of heel contact (HC), foot flat (FF), midstance (MS), heel off (HO), and toe off (TO). This sequence of events shall be referred to as foot timing. It is measured as a percentage of the gait cycle (Figure 1).

The results of joint rotations (ankle range of motion) will be reviewed concurrently. Even though neither the Flex-Foot™ nor SACH foot contains an ankle joint, the gait analysis cameras can perceive their action as ankle dorsiflexion and plantarflexion. Because the computer software does not specify a neutral position, "relative" motion is analyzed. For these two variables (foot timing and joint rotation), the periods of early and late stance will be evaluated separately.

In normal gait, the ankle is in a neutral position when the heel strikes the floor. The ankle then plantarflexes 12 to 15 degrees[6] in order for

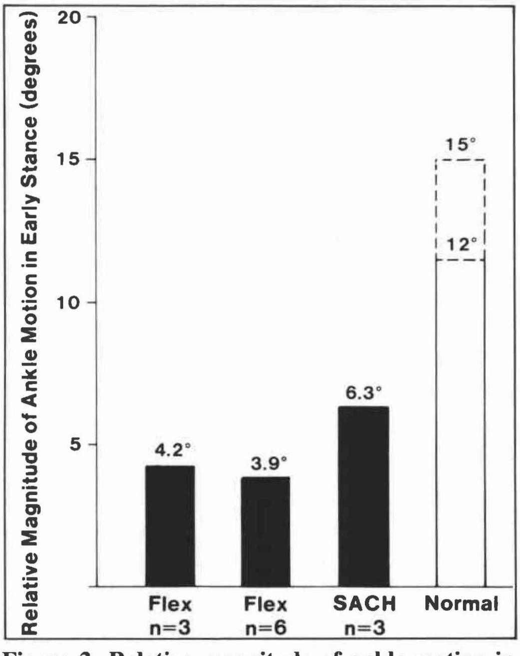


Figure 2. Relative magnitude of ankle motion in early stance.[6]

the foot to become flat on the ground. Thus, foot flat occurs at nine percent of the gait cycle[1,2] as a result of 12 to 15 degrees of relative motion in early stance.

Compression of the SACH heel simulates ankle plantarflexion in early stance. However, this provides considerably less than normal

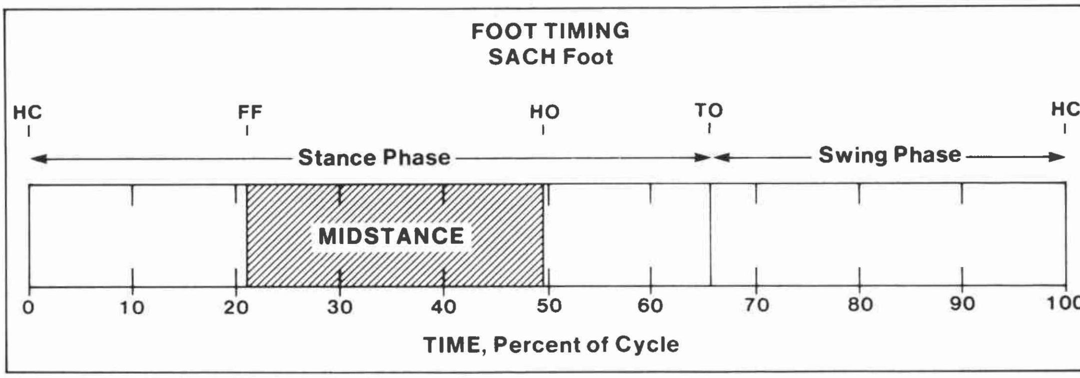


**Figure 3: Foot timing: mean of the SACH trials.**

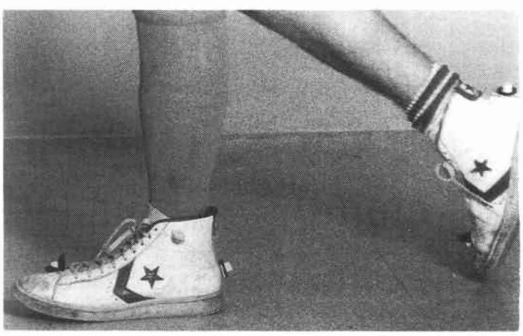

**Figure 4a. Ankle motion at foot flat: *Flex-Foot.*™**

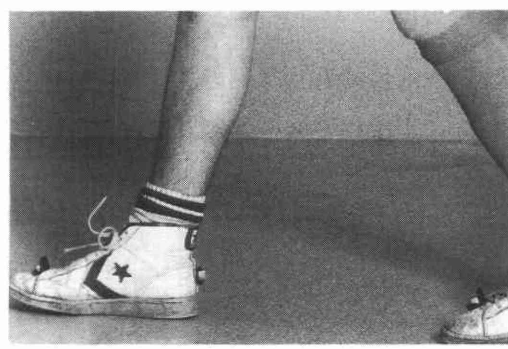

**Figure 4b. Ankle motion at foot flat: sound limb**

motion (Figure 2). As a result, the foot cannot become flat on the ground until much later in the gait cycle (21 percent) when the body weight is shifted farther forward (Figure 3).

Dynamic compression of the Flex-Foot™ heel pylon ("heel deflection") is another attempt to mimic ankle plantarflexion in early stance. Likewise, it results in less than normal motion (Figure 2). This finding was demonstrated even when the population was increased to six. A comparison of the Flex-Foot™ prosthesis to the sound limb is illustrated in Figure 4. Note the limited "ankle range of motion" of the prosthetic limb at foot flat. Lack of plantarflexion, likewise, delays the timing of foot flat (Figure 5), but to a lesser degree than with the SACH foot.

During the early stance phase of non-vigorous walking, joint rotations and foot timing did not significantly differ (between the Flex-Foot™ and SACH foot). Both deviate from normal values.

After the period of foot flat in normal gait, the leg rolls over the planted foot until it reaches a peak dorsiflexion of eight to ten degrees.[6] Shortly thereafter, the heel rises from the ground (HO). Immediately following, the ankle plantarflexes to a position of 18 to 23 degrees of plantarflexion[6]; a total of 30 degrees of relative motion usually occurs during this latter part of stance.

Peak dorsiflexion occurs on the SACH foot as the forefoot bends over the keel (followed by a return to the starting position). This resulted in only 11 degrees of relative motion, which is considerably less than normal (Figure 6).

The Flex-Foot™ provides a significantly more normal substitute of ankle motion in late stance. Dynamic compression of the anterior pylon ("toe deflection") and a plantarflexion rebound afforded 20 degrees of motion (Figure 6). Greater excursion potentially enables the amputee to lean farther forward and take more symmetrical steps. Figure 7 illustrates the similarities between the sound limb and Flex-Foot™ at the time of heel off.

Likewise, foot timing was positively affected in late stance. The period of toe off occurred

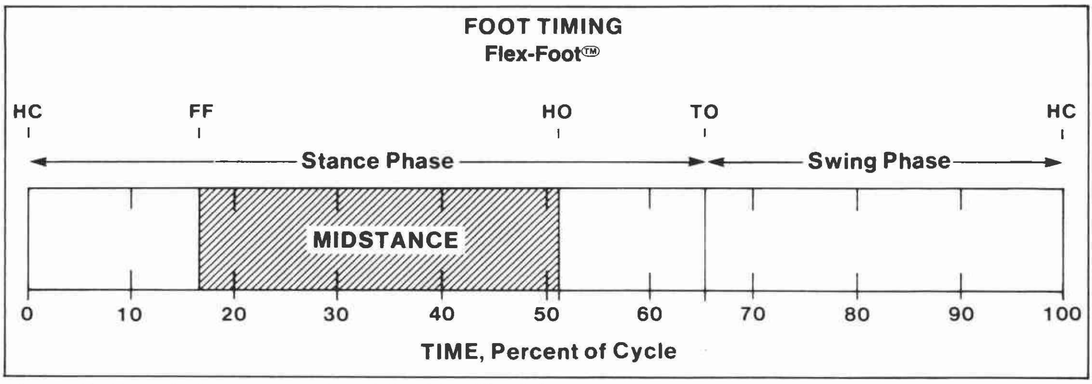


Figure 5. Foot timing: mean of the Flex-Foot™ trials.

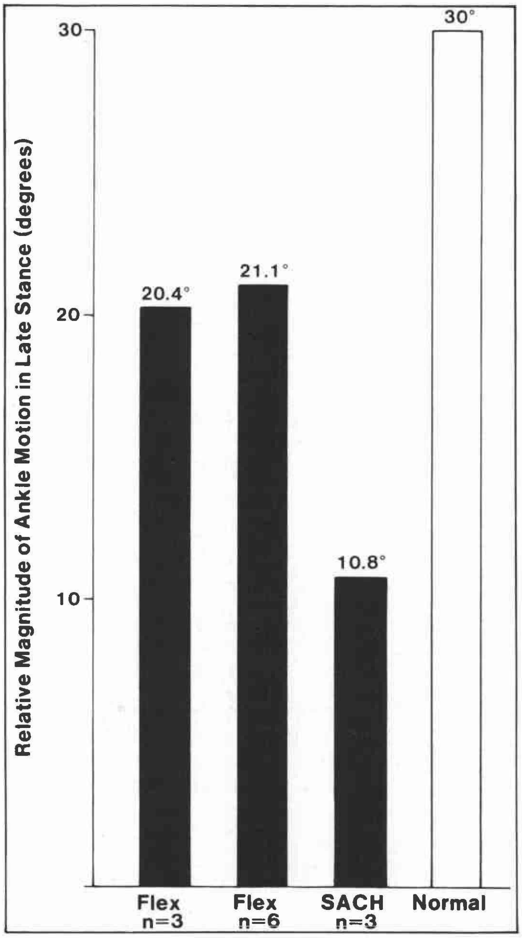


Figure 6. Relative magnitude of ankle motion in late stance.[6]

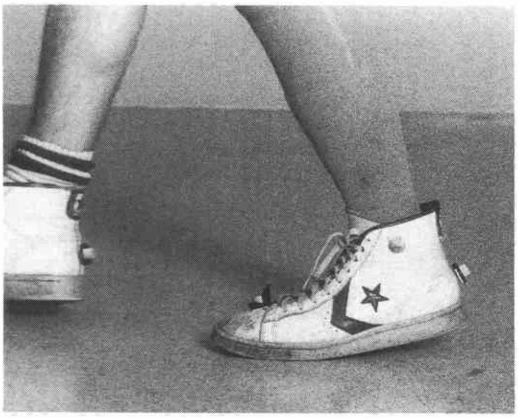

Figure 7a. Ankle motion of heel off: *Flex-Foot.*™

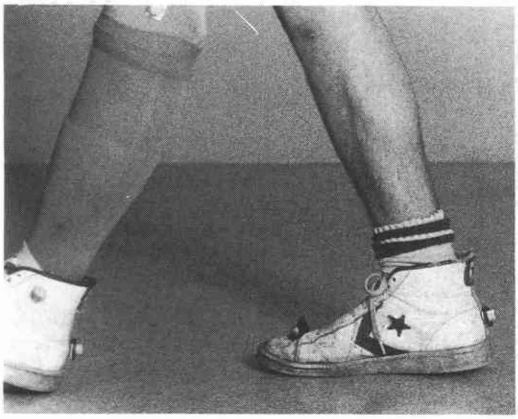

Figure 7b. Ankle motion of heel off: sound limb.

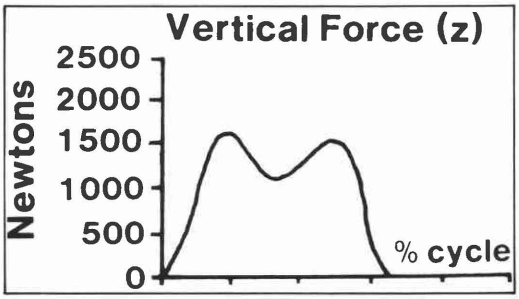


**Figure 8. Sample representation of vertical force data.**

more symmetrically between the sound and amputated limbs when the Flex-Foot™ was worn.

Therefore, even "average" walkers benefit from the ankle motion which the Flex-Foot™ provides in the late stance phase of gait.

## *Force*

"Vertical force is the floor reaction force in the vertical direction as measured by the force plate."[10] The magnitude of vertical force is typically measured in newtons. The value has been normalized to body mass (ie. newtons/kg) to enable comparison among subjects. Vertical force is graphically represented by two peaks separated by a valley (Figure 8). The first peak occurs as a result of the initial impact of the foot on the ground. In the past, the second peak has been thought to represent a push-off by the ankle plantarflexors (posterior calf muscles).

Many of the energy storing feet claim to provide a push-off similar in nature to the action of the ankle plantarflexor muscles. It seems reasonable, then, to investigate this claim by looking at the second peak of vertical force.

Before doing so, current theories regarding the precise function of the ankle plantarflexor muscles should be reviewed. Cavagna continues to support the idea that muscle action generates a forceful push-off.[3] Although Mann does not support a "push-off," he claims that the "plantarflexors control the forward propulsive momentum, making it possible for the body to move farther from its base of support."[5] Sutherland recognizes that there is a "knee-ankle-stability linkage, suggesting that the ankle plantarflexors decelerate the tibia as it rotates in a forward direction over the talus, controlling a selective rapid extension of the

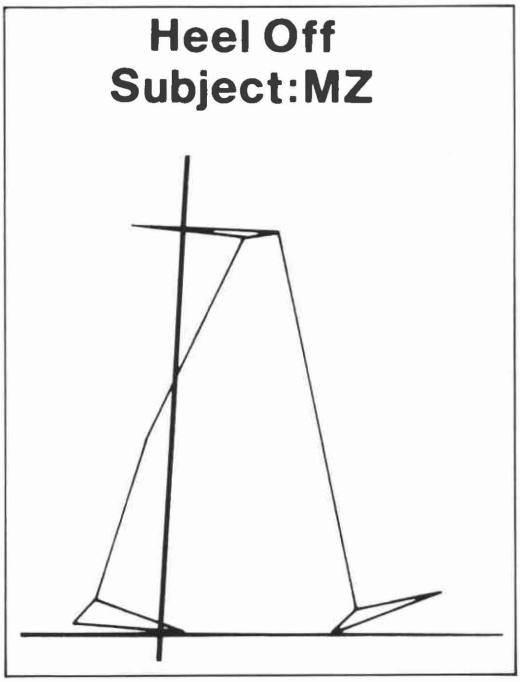


**Figure 9a. Subject MZ: lateral stick figure representation with superimposed floor reaction force vector.***

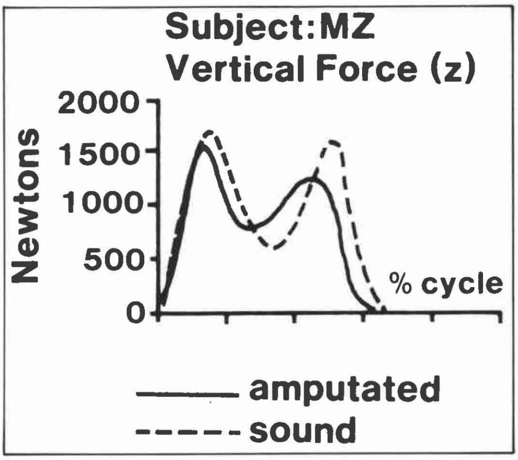


**Figure 9b. Subject MZ: vertical force comparison of sound and amputated limbs.**

*Hip joint marker placement on the ASIS rather than true joint center makes the knee appear more extended than what anatomically occurs.

knee."[12] Perry reviewed Cunningham's study (1950) which found force plate records for below-knee amputees with SACH feet to be similar to the data of normal muscled persons.[4] She then theorized that "the late floor reaction peak is the result of leverage by body alignment, rather than an active downward thrust."[7]

The results of this study support Perry's contentions. The magnitude of the second peak of vertical force (normalized to body mass) was the same for the SACH foot as it was for the corresponding sound limb. Assuming that the SACH foot is not an energy storage system, it can be concluded that the similarity was due to "leverage by body alignment"[7] (which for amputees is influenced by prosthetic alignment).

When the Flex-Foot™ population was three, the magnitude of the late floor reaction force was greater for the prosthesis than for the sound limb. This should not be interpreted as a superhuman "push-off." In fact, when the population was increased to six, the force generated by the Flex-Foot™ was less than that of the sound foot. This discrepancy made it necessary to investigate the cause.

A trend was found, suggesting that when a knee extension moment existed at heel off, the late stance peak (of vertical force) of the Flex-Foot™ was less than that of the sound limb (Figure 9). Those subjects who had a knee flexion moment at heel off generated forces which closely resembled the sound limb (Figure 10). The Flex-Foot™ responds with optimal vertical force in late stance when a knee flexion moment is created through prosthetic alignment.

Thus far, it has been determined that muscular action in the late stance of normal walking is not actually a "push-off." The data and literature suggest that the second peak of vertical force is primarily a product of alignment. Therefore, some other means of investigating the supposed energy storage and release of new prosthetic feet must be determined.

Simon studied subjects without posterior calf muscle activity and found ways in which they compensate. Because the plantarflexors were not available to provide a restraining force as the tibia rotated forward over the foot, his subjects spent less time in single limb stance (on the involved limb) and experienced premature opposite heel contact.[8] Neither of these occurrences were found in this study.

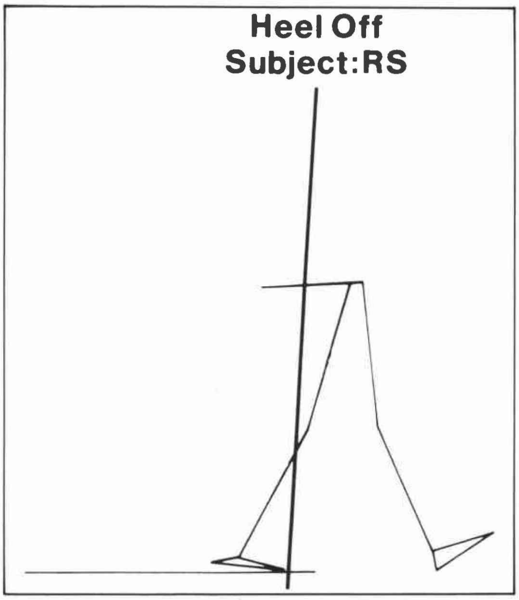


**Figure 10a. Subject RS: lateral stick figure representation with superimposed floor reaction force vector.***

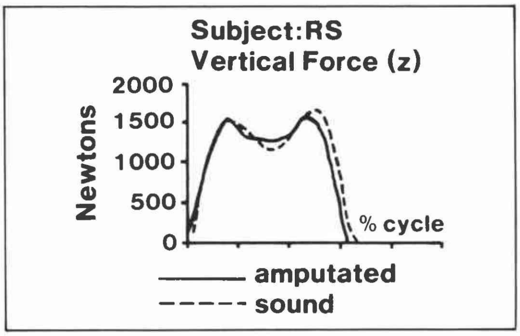


**Figure 10b. Subject RS: vertical force comparison of sound and amputated limbs.**

As a secondary means of compensation, Simon's subjects experienced excessive impact at opposite heel strike (HC on the sound limb).[8] This occurred because of an inadequate restraining force by the involved extremity.

In this study, the magnitude of vertical force at impact (i.e. the first peak, also normalized to

*Hip joint marker placement on the ASIS rather than true joint center makes the knee appear more extended than what anatomically occurs.

body mass) was the same when the Flex-Foot™ and SACH foot struck the ground. This finding is consistent with the fact that the sound limb provides a similar restraining action regardless of which prosthetic device is worn on the amputated limb.

However, the sound limb struck the floor with significantly greater force during the SACH trials (as compared to the Flex-Foot™ trials). Because SACH foot does not provide a controlled restraint, the corresponding sound extremity must hit the floor with an excessive amount of force. Such compensation was unnecessary during the Flex-Foot™ trials. Apparently, the restraining action which the Flex-Foot™ creates is a considerably better simulation of the function of the true ankle plantarflexor muscles.

After review of the data and current literature, the role of the calf muscles appears to be a restraining action rather than a "push-off" during normal gait. The Flex-Foot™ more effectively simulated the action of the calf muscles than the SACH foot in non-vigorous walking.

# SUMMARY

Non-vigorous walking was studied to determine if the general below-knee amputee population would biomechanically benefit from energy storing feet during walking. We found that linear gait parameters did not significantly differ between the Flex-Foot™ and SACH foot. However, symmetry did improve with the Flex-Foot™. Ankle joint rotation in early stance was considerably less than normal with both types of artificial limbs. However, the Flex-Foot™ allowed significantly greater range of motion during late stance. Although foot timing was minimally improved with the Flex-Foot™, the period of foot flat was delayed with both types of prostheses. Energy storage and release is apparent through an investigation of vertical forces. Forces which normally result from muscle activity must be mechanically created by the prosthesis. Force data suggests that the controlled restraining action of the Flex-Foot™ is a significantly better representation of the function of the posterior calf muscles.

# CONCLUSION

Moderately active below-knee amputees experience biomechanical benefits from an energy storage prosthetic foot system. Thus, indications for use of such systems should not be limited to athletes and vigorous ambulators. The Flex-Foot™ should also be considered for less active individuals. Biomechanical responsiveness should be considered along with other factors such as cost, fitting time, and cosmesis when selecting a prosthetic foot.

## ACKNOWLEDGMENTS

The authors would like to thank Steven Verhulst from the Statistics and Measurements Department and William Andrea of the Biomedical Communications Department at SIU.

A more detailed statistical analysis is being submitted for publication.

This research was supported by the Division of Orthopaedics of SIU School of Medicine and Flex-Foot™, Inc.

## AUTHORS

The authors are clinical faculty at Southern Illinois University School of Medicine. Address correspondence to Southern Illinois University, Orthotic and Prosthetic Services, Room 102, 707 North Rutledge Street, Springfield, IL 62702.

## REFERENCES

[1] Breakey, James, M.Sc., "Gait of Unilateral Below-Knee Amputees," *Orthotics and Prosthetics*, Vol. 30, 3, September, 1976, pp. 17–24.

[2] Breakey, J.W., "Gait of Unilateral Below-Knee Amputees: A Kinesiologic and Electromyographic Study," M.Sc. thesis, Queen's University at Kingston, Canada, 1975.

[3] Cavagna, Giovanni, "Force Platforms as Ergometers," *Journal of Applied Physiology*, 39:1, July, 1985, pp. 174–179.

[4] Cunningham, D.M., "Components of Floor Reaction During Walking," Prosthetic Devices Research Project, Technical Report, Series 11, No. 14, University of California, Berkeley, 1950.

[5] Mann, R.A.; J.L. Hagey; and S.R. Simon, "Push-Off Phase of Gait," *Abbott-Proc.*, 5:85, April, 1974.

[6] Murray, M.P., Ph.D.; A.B. Drought, Sc.D.; and R.C. Kory, M.D., "Walking Patterns of Normal Men," *The Journal of Bone and Joint Surgery*, 46-A:2, March, 1964, pp. 335–360.

[7] Perry, Jacuelin, M.D., "Kinesiology of Lower Extremity Bracing," *Clinical Orthopaedics and Related Research*, 102, July–August, 1974, pp. 18–31.

[8] Simon, Sheldon, M.D.; Roger Mann, M.D.; John Hagey; and Larsen, Loren, M.D., "Role of the Posterior

Calf Muscles in Normal Gait," *The Journal of Bone and Joint Surgery*, 60-A: June, 1978, pp. 465–472.

[9] Skinner, Harry, M.D., Ph.D., FACS, and David Effeney, M.B., B.S., FRACS, "Gait Analysis in Amputees," *American Journal of Physical Medicine*, 64:2, April, 1985, pp. 82–89.

[10] Sutherland, David, M.D., *Gait Disorders in Childhood and Adolescence*, Williams and Wilkins, 1984.

[11] Sutherland, David, M.D., "Interpretation of Gait Analysis Data," San Diego Children's Hospital, from a seminar presented March, 1986.

[12] Sutherland, David, M.D.; Les Cooper B.A.; and Dale Daniel, M.D., "The Role of the Ankle PlantarFlexors in Normal Walking," *The Journal of Bone and Joint Surgery*, 62-A:3, April, 1980, pp. 354–363.