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The S.A.F.E. Foot

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Inman and Eberhart (6), in their summary of research related to amputees and artificial limbs, postulated that the designer of artificial legs must have detailed knowledge of the normal leg and its role in locomotion in order to reproduce as nearly as possible the features of the normal leg. In our modest way, we have attempted to design an artificial foot that comforms to the shape and mechanism of the human foot.

The letters S.A.F.E. are the acronym for Stationary Attachment Flexible Endoskeleton or, in simpler language, a prosthetic foot bolted to the shin with a flexible keel as shown in Figure 1. To be consistent with other endoskeletal systems, the bolt block of this foot and the flexible keel are encased in a soft foam cover, as shown by section in Figure 2.

The first reaction of many prosthetists to the foot or any other structure so described is apt to be that it lacks stability and therefore is inherently unable to support super-incumbent weight. The foot must not only support weight, or compression, but moving, dynamic, forces that produce torque and shear forces as well. Yet, except for the lack of an ankle joint, we might be describing a human foot in very simplistic terms.

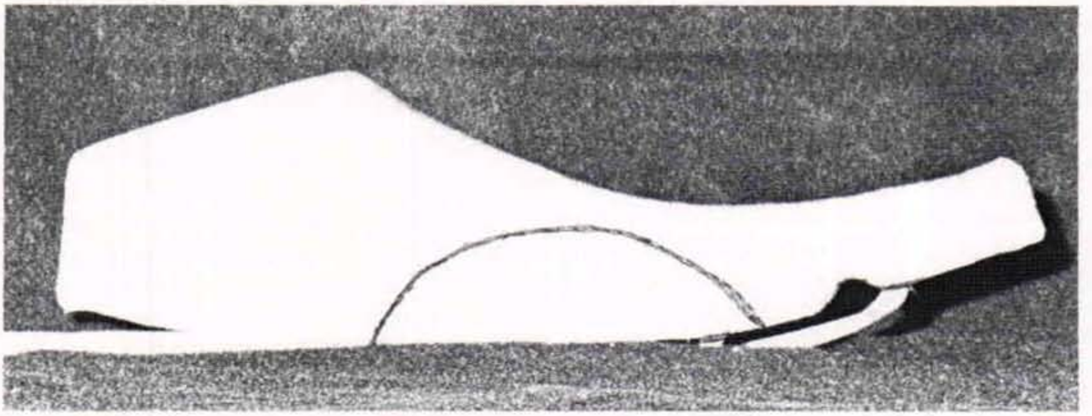

Fig. 1. The endoskeleton.

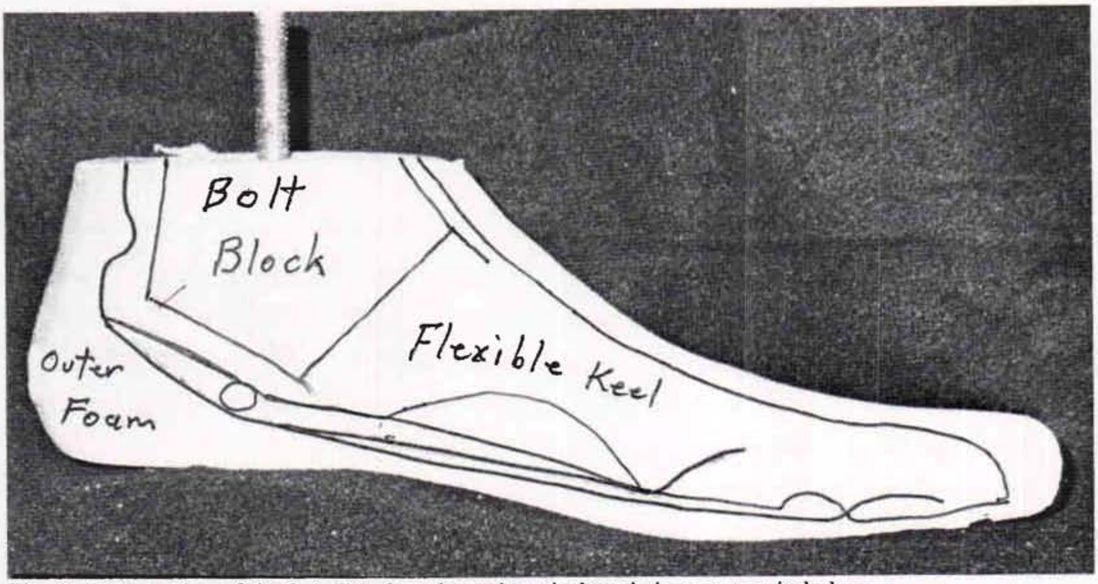

Fig. 2. Cross-section of the S.A.F.E. foot throughout its length in a parasagital plane.

The Arch

In the human foot there are twenty-six bones held together with ligaments, connective tissue, and capsules. Each bone is capable of some motion relative to its adjacent members. The motions of these bones are dictated by the shape of their articular surfaces and the restriction provided by the ligamentous structures. The seven tarsal bones and the five metatarsal bones are usually described as taking the shape of an arch, a very strong geometrical structure when it is rigid, but requiring additional support in the form of trusses or ties when flexible. In the cases of the foot, the trusses or ties are provided by the ligaments, particularly the long plantar ligament and the plantar aponeurosis (fascia).

Although the arch has stability along its long dimension, stability medially and laterally is lacking. A series of arches bound together with ligaments would increase mediolateral stability, and indeed this is the way the foot is seen by many people. However, the arches are arranged in a triangular shape having one common base proximally, the calcaneus, and five bases distally, the metatarsal heads (Fig. 3).

The Arch As a Half Dome

The anatomist, Frederick Wood Jones, (3) in his book *Structure and Function as Seen in the Foot,* described the foot in a way that would make it an even stronger structure. He said, in effect, that if we put our feet together geometrically, the arches form a dome, and, therefore, when we separate them, they each are a one-half dome, a form that is inherently more stable than a series of arches.

Perhaps the two theories are not really far apart, if we concede that most of the time the highest arch of the foot is the medial one as defined by its base, the first metatarsal head, and that each arch as we move laterally is lower at its apex until the fifth member as defined by the fifth metatarsal bone, cuboid, and calcaneus

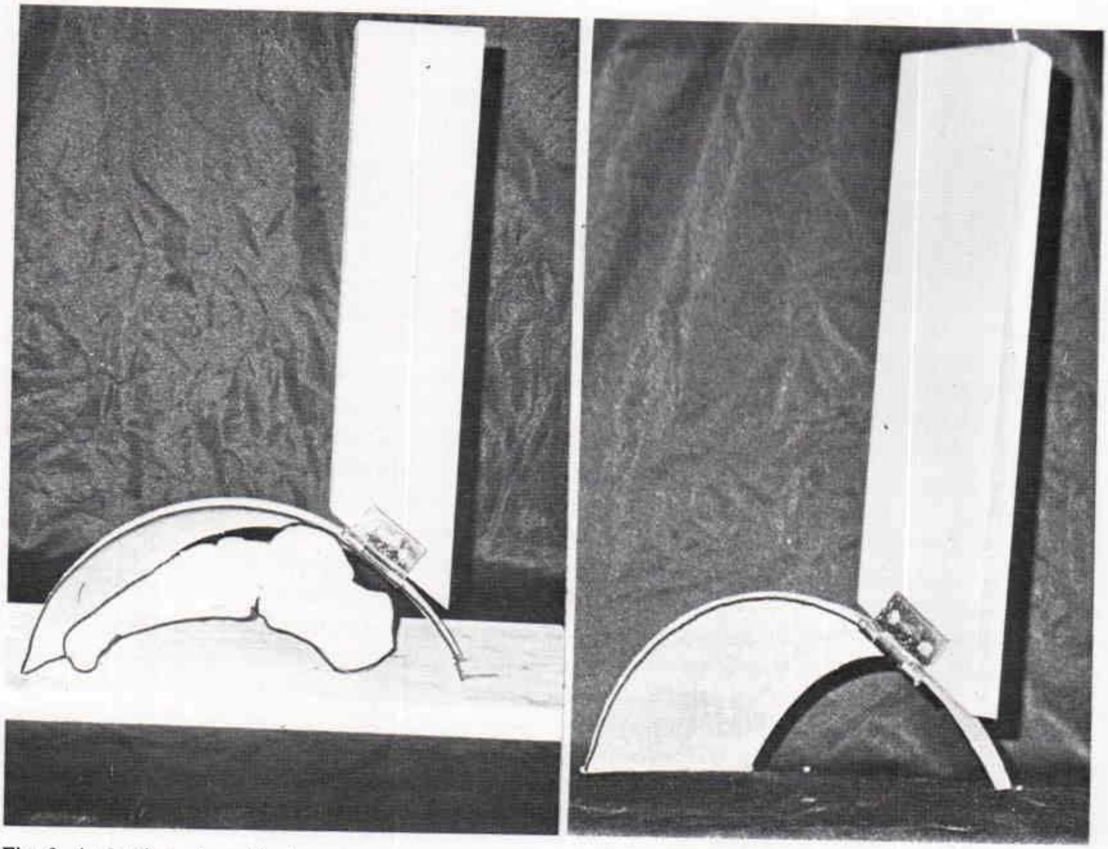

Fig. 3. A plastic arch and its bony counterpart.

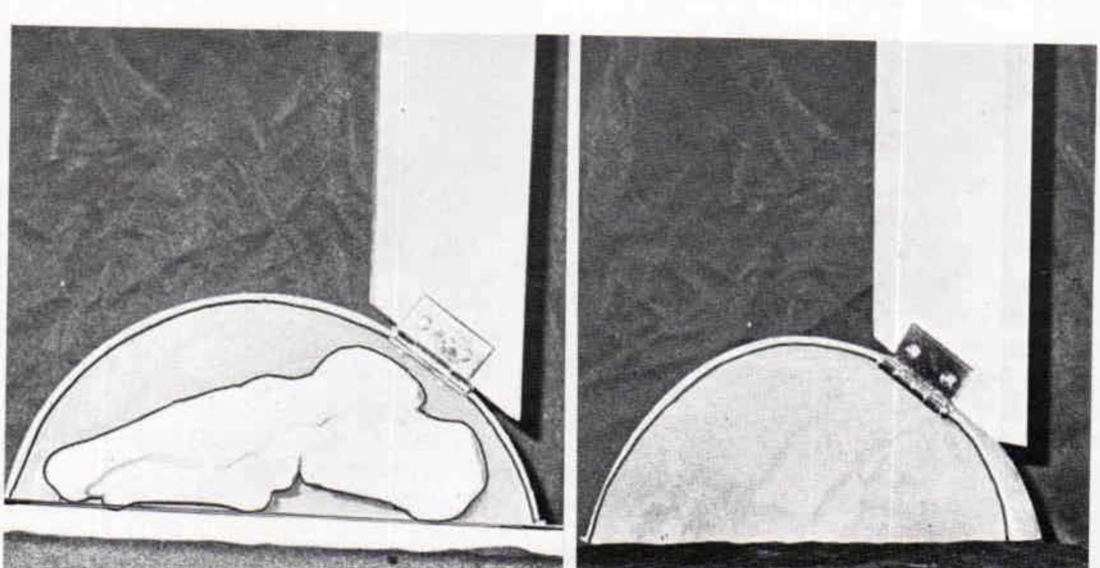

Fig. 4. A plastic dome and its bony counterpart.

has an apex so low that it usually cannot be described as an arch. Therefore, when these conditions are given, we have described a half dome, as shown in Figure 4. One only has to look at wet footprints or Elftman's (4) barograph studies of normal feet to realize that ground contact is primarily from the area of the calcaneus, cuboid, the length of the fifth metatarsal bone, and the metatarsal heads. Since the bones of the foot can articulate each with its adjacent member, the one-half dome is a flexible structure. Consider now the mechanical advantage of a long plantar ligament and a plantar aponeurosis tying the ends of the one-half dome together. The foot then becomes a very strong, stable member capable of resisting the forces of the super-incumbent weight but still retains flexibility when required.

The foot, without aid from muscles, can function when the ligaments remain intact. And thus many patients without muscle activity below the knee can apply weight to their foot without the loss of its integrity. Basmajian (1) made simultaneous electromyograms of six muscles of a group of twenty subjects. The muscles were under a static load of 45.4 to 181.4 Kg. The muscles tested were the tibialis anterior, and posterior peroneus longus, flexor hallucis longus, abductor hallucis, and flexor digitorum brevis. The contribution of these muscles was considered insignificant until the load reached 181.4 Kg, and even then some muscles remained inactive. Basmajian concluded that the passive structures (bone and ligament) are the only ones capable of sustaining an unremitting load.

Campbell and Inman (2) speculated that if the plantar aponeurosis were a major contributor to the stability of the arch, plantar fasciitis might be relieved when the arch was supported mechanically. The UC-BL Shoe Insert was used successfully on thirty of thirty-three patients to relax the plantar fascia and relieve the associated pain.

Mann and Inman (6) found that there was little, if any, significant activity of the intrinsic musculature during quiet standing. They also found that the intrinsic muscles did not elicit a significant response until 30 percent of the gait cycle (midstance) or just prior to heel rise. Therefore in the first half of stance phase, the entire weight of the body is borne by the passive structures (bones and ligaments).

Cunningham (3) found that during about 15 percent of the gait cycle (foot flat) the foot is subjected to 120 percent of body weight. The anterior shear force is as high as 30 percent of body weight and the torque was as high as sixty inch-pounds. These forces are all acting before the intrinsic muscles become active and before the extrinsic muscles fire, except for the anterior group that prevents foot slap. We submit that the one-half dome concept of the foot is the best geometrical shape that can remain flexible and still accept the aforementioned forces without muscular assistance. The one-half dome therefore provided the basic geometry for the design of our new artificial foot.

The Second Half of Stance Phase

Previously the emphasis has been on the first half of stance phase when the foot was required to have "flexible" stability. However, in the second half of stance phase, the function required of the foot changes significantly from a "flexible-stable" member to a semi-rigid lever. In the swing phase and in the first half of stance phase, the entire leg and foot are rotating internally around the long axis of the leg a total of 65 percent of the gait cycle (9). In the second half of stance phase these transverse rotations are reversed and the leg is externally rotated with the foot fixed with respect to the ground. These transverse rotations have been recorded to be as much as 29 degrees, in which case some individuals have to externally rotate their leg twenty-nine degrees in one third of a second when the gait cycle is one second. These transverse rotations of the leg are absorbed in the arch of the foot, but since the foot does not bend in the arch to absorb these rotations, they are converted to rotations in a vertical plane about the long axis of the foot by means of the subtalar joint.

The subtalar joint becomes a motion and torque converter much like two 45-degree beveled gears. Most investigators, including Isman and Inman (7), agree that the subtalar joint is usually close to a 45-degree angle with respect to the foot.

The forefoot must still maintain contact with the floor and therefore the arch mechanism, because of its flexibility, can twist about its axis and absorb these rotations.

Wright *et al* (11) produced a mechanical analogue of the ankle and subtalar joints which could be aligned to a subject so that the motions about these axes while walking could be recorded. They found that the significance of the subtalar joint was easily demonstrated by the fact that motion about the subtalar joint provided approximately half the motion attributed to the ankle joint.

When the foot is released from the ground it does not rotate violently to catch up with the leg, but merely untwists. This kind of motion cannot be attained when transverse rotations are absorbed in the shank of an artificial leg, a point that is demonstrated when an amputee first puts on a prosthesis provided with a rotator in the shank. For the first few steps the amputee usually tries to use all of his normal horizontal rotations. However, after a few steps he will automatically suppress some of these rotations, because if he does not the torque developed in the shank will externally rotate the prosthetic foot violently and strike the contralateral leg as it swings through. This will not occur when the rotation is placed where nature intended it, i.e., within the foot. Therefore our second design criterion for an artificial foot was provision of a subtalar joint, and the third criterion was provision of a flexible endoskeletal structure.

Immediately prior to heel rise, the intrinsic muscles fire off amost simultaneously to raise the arch to help produce conversion of the foot to the semi-rigid level attitude. For obvious reasons muscular control is one criterion that we eliminated. Fortunately, muscular activity is not the only mechanism for converting the foot to a semi-rigid lever.

In 1954 Hicks (5) described the contribution of the plantar fasciia in stabilizing the foot from heel rise to toe off. Since the attachment of the plantar fasciia is distal to the metatarsal phalangeal joints, extension of these joints such as occurs with dorsiflexion of the toes causes tension on the plantar fasciia and contributes to the conversion of the foot to a semi-rigid lever. Hicks called this the "Windlass effect" of the plantar aponeurosis (as shown by the model in Fig. 5). Therefore we now have the fourth criterion for the prosthetic foot. It must have a toe break which would flex somewhere near the toe break of the human foot and a plantar fasciial strap from the toe area spanning the toe break and the arch, and fixed to the posterior aspect of the foot.

Design of the S.A.F.E. Prosthetic Foot

To our design criterion we added one more condition: that the foot be made entirely from plastic materials and thus avoid mechanical joints. We then decided that we must start with the outside shape and work inward toward our endoskeleton configurations.

A model of a low profile, conventional, size 13 prosthetic foot was constructed out of plaster (Fig. 6), and a negative of this

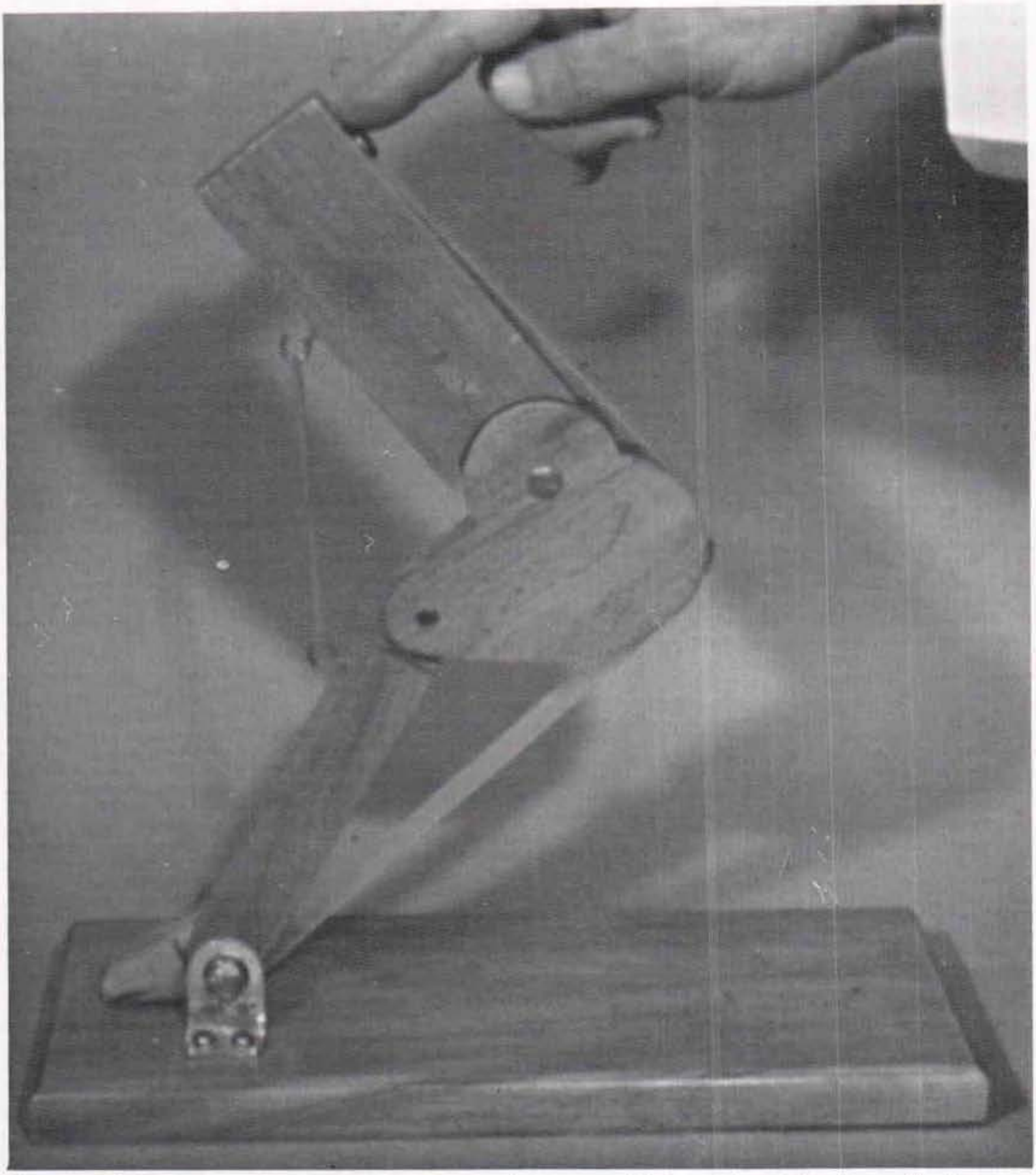

Fig. 5. Model that demonstrates how dorsiflexion of the toes, with shortening of the plantar fasciia, raises and stabilizes the arch.

model was constructed in two pieces, a top portion and a bottom portion. Bolts and wing nuts held the two sections together, and our outside mold was completed.

The Keel Mold

The plaster positive of the foot was modified into the keel shape in the following manner. The foot was divided

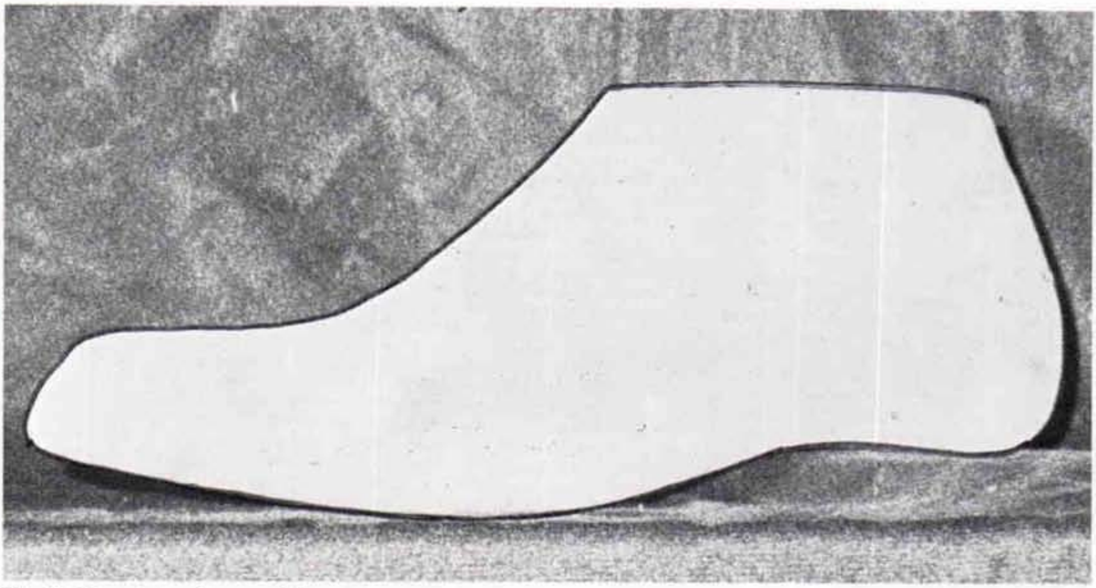

Fig. 6. The plaster foot model.

into three equal sections. The anterior third is the ball (metatarsal heads) toe break and toe section. The middle third is the arch (dome) and the posterior third is the soft heel and bolt block section. In the anterior section a toe break was carved on the plantar surface medial to lateral, one-half inch anterior to the ball line (Fig. 7). The toe-break groove is five-eighths of an inch from the top of the toe section. In the middle third, a dome was formed by carving the plantar and medial surfaces until they were thinned to seven-eighths of an inch from the top and the lateral border of the foot. One half-inch of plaster was cut off of the plantar surface of the foot. The plantar surface of the heel section was cut further from zero anteriorly to one inch at the heel. Three-eighths of an inch of material was then

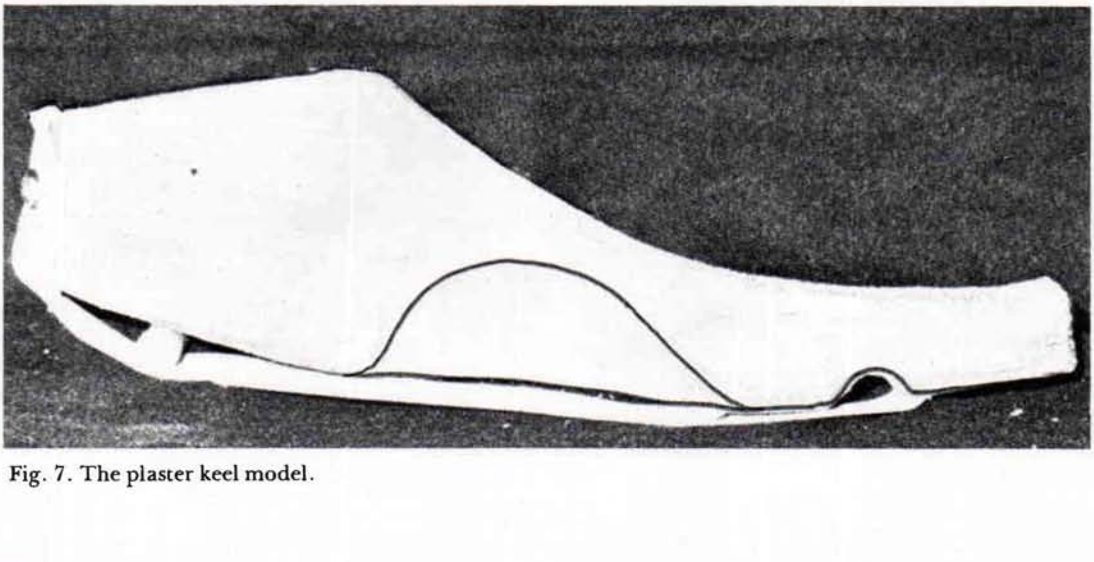

Fig. 7. The plaster keel model.

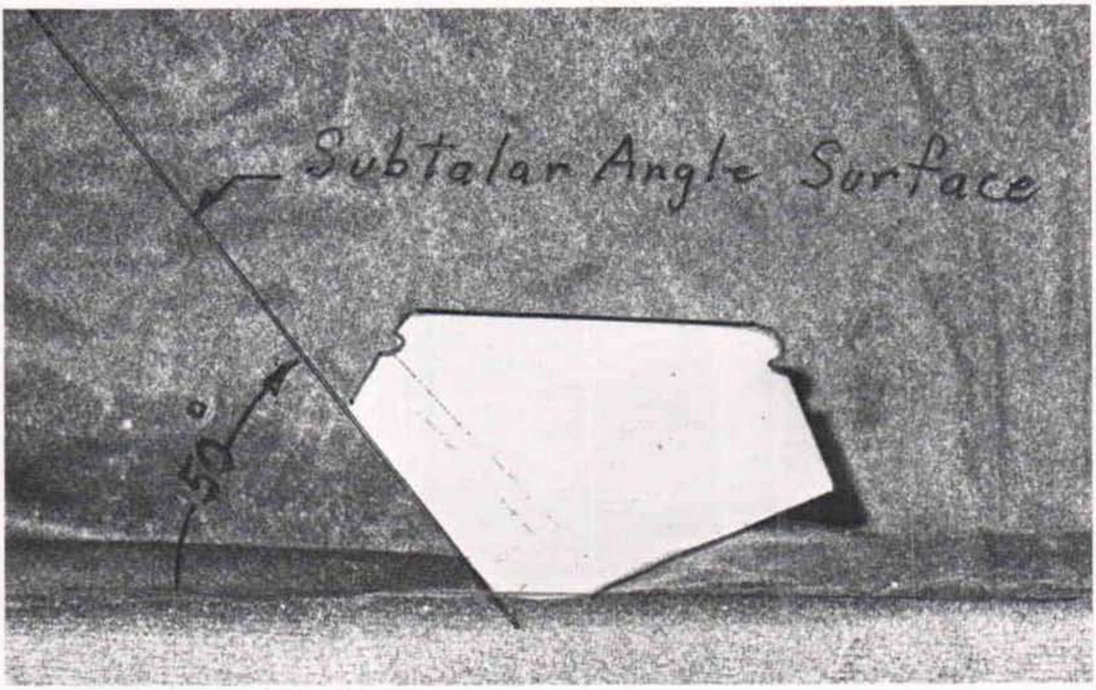

Fig. 8. The plaster bolt-block model.

removed from the top surfaces except where the foot bolts to the shank. We now had our positive keel mold shape (Fig. 7). Negative molds were made in the same manner as our foot molds.

The Bolt-Block Mold

The plaster keel mold was cut through in a mediolateral plane at a 50-degree angle just posterior to the arch area to obtain the posterior section, which became our bolt block. The anterior surface of this section, cut at a 50-degree angle, was rounded to become our subtalar angle surface. One-quarter inch of plaster was removed from the remaining surfaces as shown in Figure 8. A bolt hole was drilled through the center of the block and a clearance hole drilled on the plantar surface deep enough to clear the bolt head and washer. A negative mold of this shape was made with flexible polyurethane potting resin.

Fabrication of the S.A.F.E. Foot

A syntactic material is used to fill the bolt block negative mold. When this resin has set up the bolt block is sanded and bolted into position on the keel mold. The keel mold is bolted together around it's flanges. A polyurethane elastomer system is mixed and poured into the mold which encapsulates the ends of the bands and the bolt block. The keel is now removed from its mold and the two straps are pulled taut and secured with screws to the posterior surface of the keel into the bolt block. Holes are burned into the Dacron bands as they pass over the bolt-block access hole that the bolts may be inserted. The keel is now completed as shown in Figure 9.

The keel is bolted to the upper part of the foot mold; the two halves of the foot mold are bolted together; and a polyurethane integral skin elastomer foam is

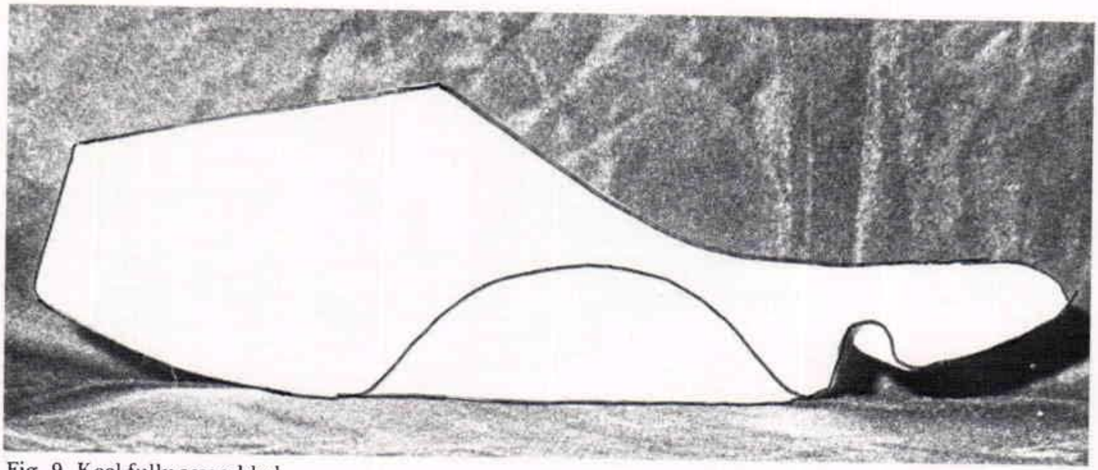

Fig. 9. Keel fully assembled.

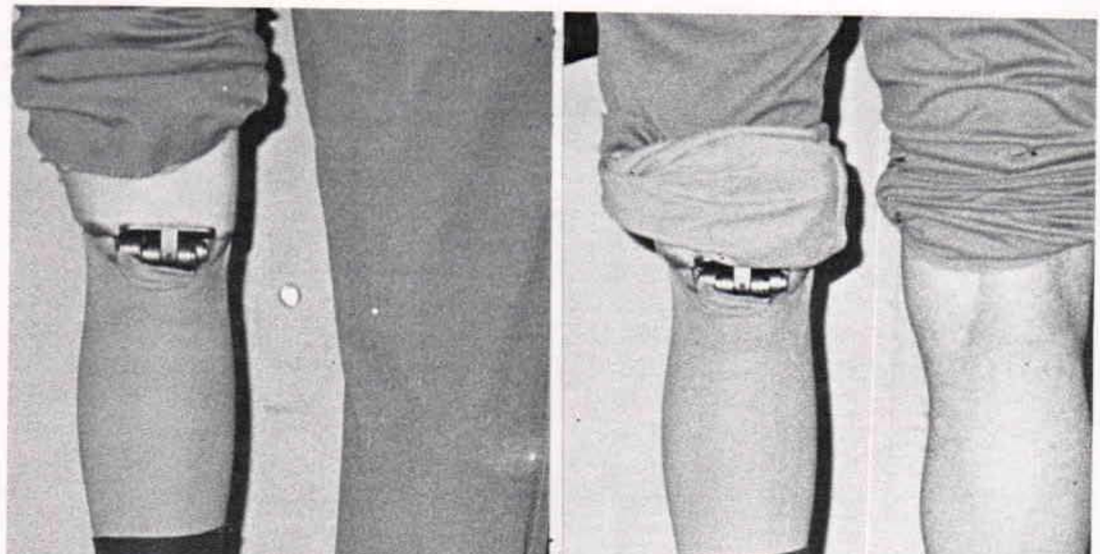

Fig. 10. Patient-subject with foot in eversion (left) and inversion (right).

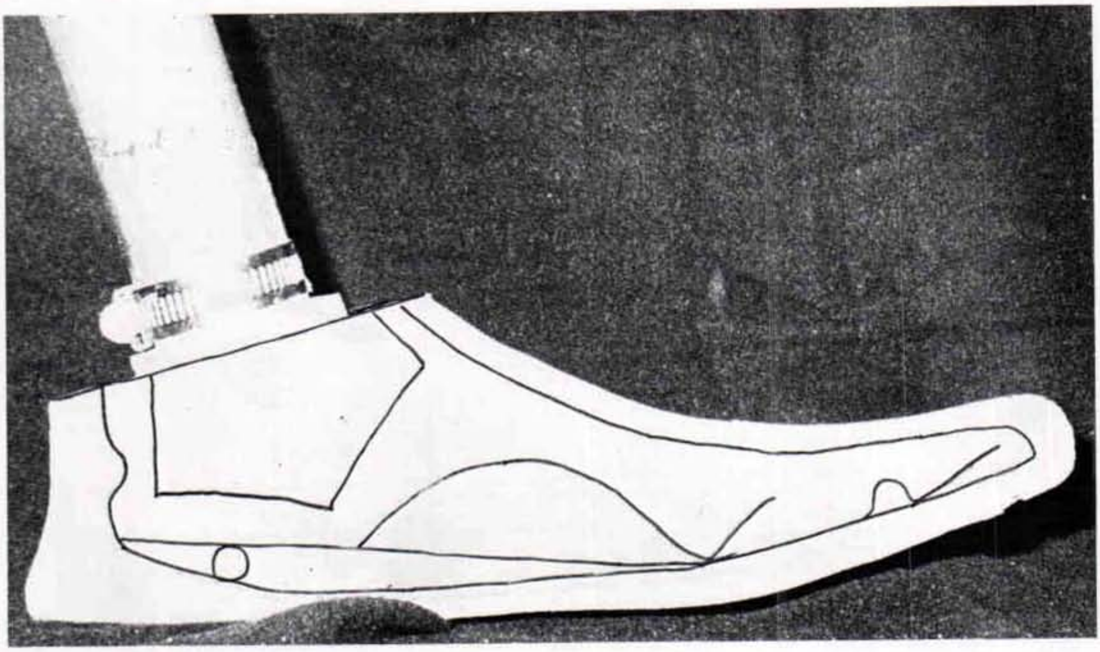

Fig. 11. The S.A.F.E. foot loaded at heel strike.

mixed and poured into the mold and left to foam and cure. The foam will form a skin against the sides of the mold, and bond to the keel, and it will completely encapsulate the keel except for the top of the bolt block. The foot is now completed and ready for use.

The foot is easily capable of sustaining the patient's weight due to the dome shape of the arch, and its tie, the long plantar ligament band. It can be inverted or everted by standing sideways on an incline as shown in figure 10. The equivalent of plantar flexion is provided by the soft heel and the fact that the bolt block has some motion relative to the rest of the foot (Fig. 11). Some degree of dorsiflexion can be simulated due to the movement of the bolt block relative to the rest of the foot (Fig. 12). It will permit transverse rotations due to the ability of the flexible keel twisting within the shoe (Fig. 13). When this twisting occurs from heel-off to toe-off the plantar faciial band is tightening as the windlass action takes effect, holding the forefoot tightly against the walking surface. The windlass action converts the foot to a semi-rigid lever (Fig. 12). The gradual tightening of the windlass band produces a smooth roll-over to toe-off without the usual snapping of the knee caused by a rigid keel or the toe bumper of an ankle-jointed foot.

Alignment

Static alignment of the S.A.F.E. prosthetic foot should be the same as used on ankle-jointed feet in the anterior-posterior plane. For static alignment of a below-knee prosthesis, we place the lateral reference line at brim point of the socket anteriorly one-half the distance normally used. All other alignment is done in the conventional manner in both below-knee and above-knee prostheses.

Results

As of this date we have tested the S.A.F.E. foot on thirty-five patient-

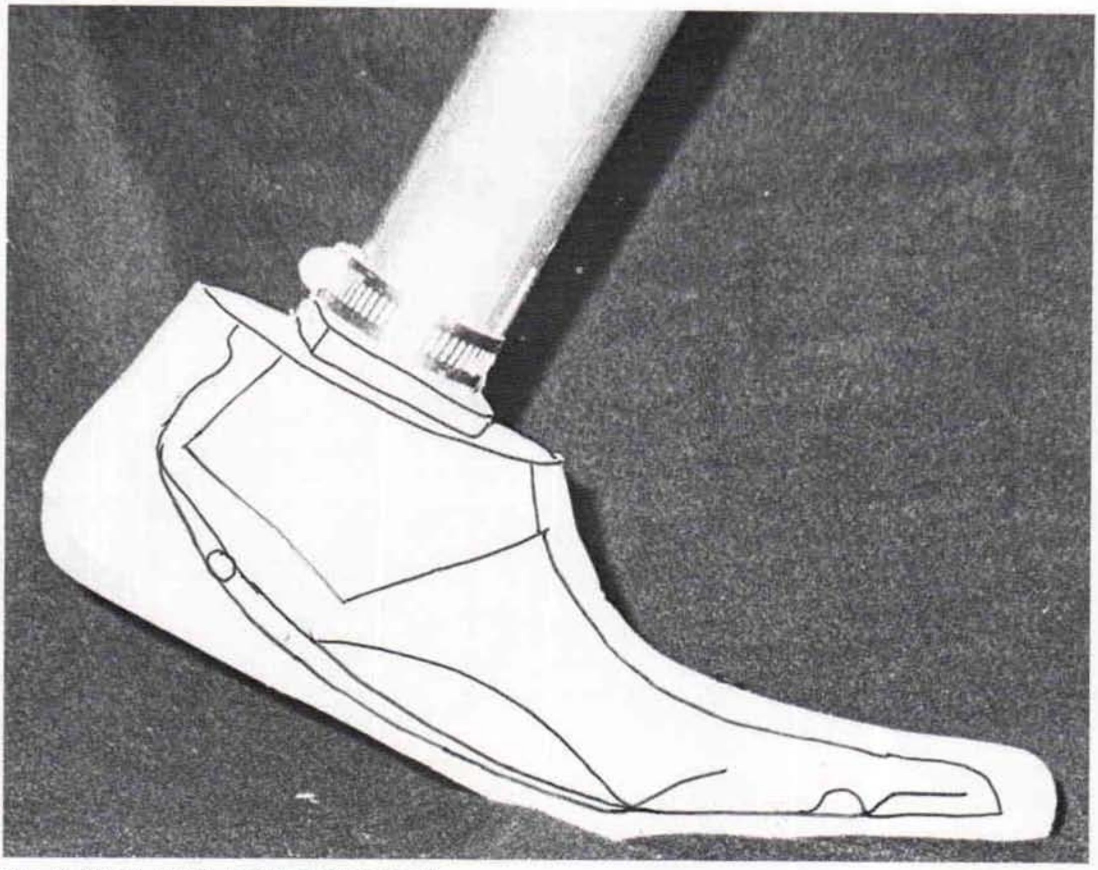

Fig. 12. The S.A.F.E. foot loaded at toe off.

subjects. Two of these feet were given to other facilities. One has not reported back as yet and the other put the foot on a BK patient-subject and reported that he was pleased with its performance. Of the remaining thirty-three feet, thirty-one are on patient-subjects.

Two subjects rejected the foot. The two rejections were BK patient-subjects with prostheses aligned for the SACH Foot. Of the remaining thirty-one patient-subjects nine are AK and eight are BK amputees.

The gait patterns of all of the thirty-one patient-subjects improved immediately. Their arm swings became equal and the lateral movement of the upper body and head was reduced to better than half of what it was previously. Their walking base was reduced to two inches or less in most cases. In the BK subjects the flexibility of the foot produced a smooth knee motion all through the stance phase and the acceleration of knee flexion through toe off was equal to the sound side.

Patient Reaction

One of the subjects is an AK amputee and a prosthetist. This subject has been an amputee for thirty two years. He has worn every type of prosthetic foot available. Three months ago he was wearing a Hydracadence leg. We removed the

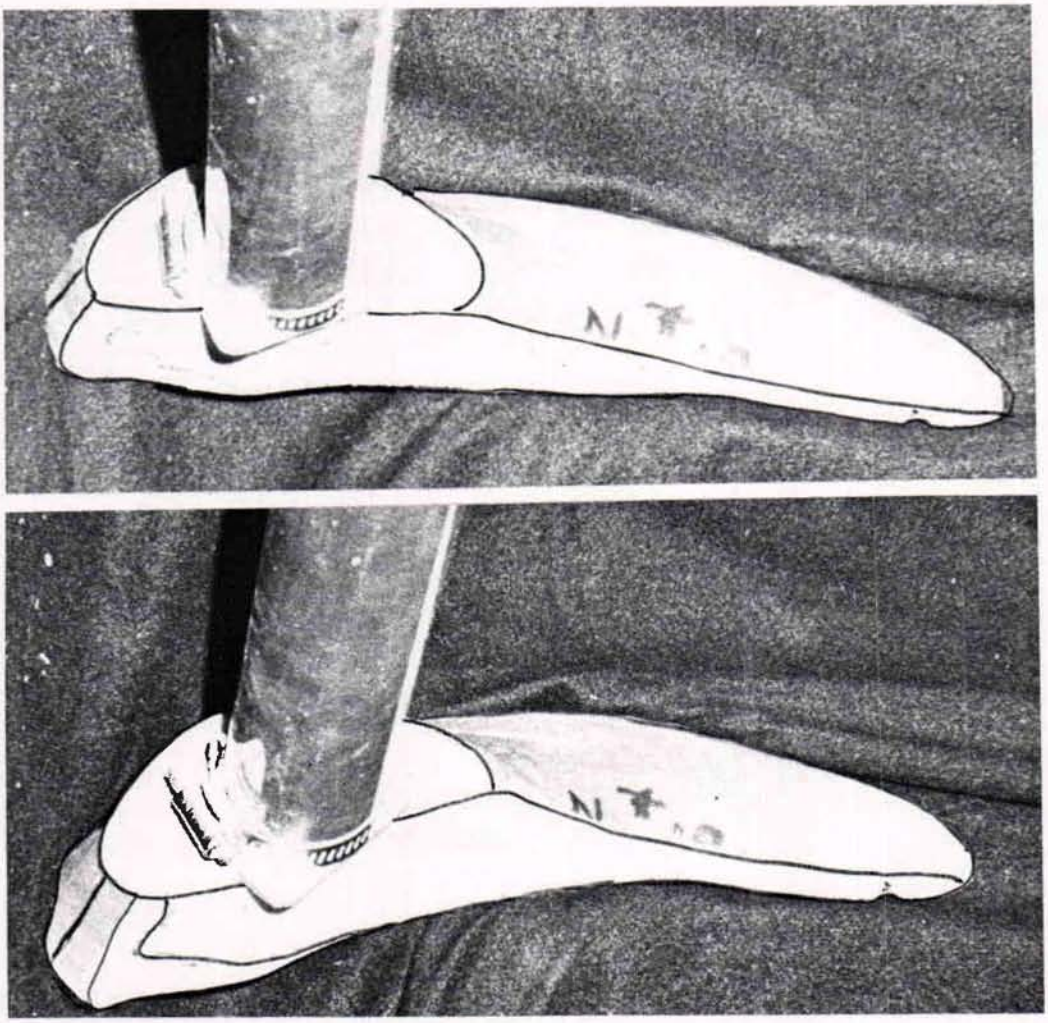

Fig. 13. The foot twisted to demonstrate transverse rotation.

Hydracadence unit from his socket and put on an O.H.C. Polycentric knee with a Dynaplex Swing Phase Control Unit and then bolted the S.A.F.E. Foot to the shank. We did not change the alignment and he was worn this combination for the last three months. He reports that energy expenditure has been reduced dramatically and that the torque and shear forces have diminished to a point that he cannot feel them. He states that the smooth transition of motion from heel strike through to toe off, caused by the flexibility of the arch and the gradual restraint of the toe, has solved almost all of his chronic socket problems. All other subjects expressed similar reactions but were not as explicit as the prosthetist-amputee subject.

One thirty-two-year-old woman, a left BK amputee, was a very athletic person before her amputation. When we used her as a subject she returned two weeks later to report that she was able to jog without injury to her stump for the first time. She returned one month later to report that her jogging endurance is up to one-half mile three times a week.

Summary and Conclusion

A prosthetic foot was designed and built to meet five predetermined criteria: a dome shaped arch and a long plantar ligament band, a flexible endoskeleton, a subtalar joint, a windlass mechanism (toe break and plantar fasciia), and made entirely out of plastic materials with no mechanical joints. All of the criteria were met in our design except the subtalar joint motion.

It was first thought that we could simulate both joints by merely thinning the plastic in appropriate areas. Although the toe break action was accomplished in this manner, the subtalar joint proved to be more elusive. At first, thinning of this area produced a definite instability in the foot. Elimination of this criterion produced a foot which worked so well for a while, we dropped this condition. However when we sectioned the foot lengthwise, we could see that to a certain extent the anterior edge of the bolt block dictated the motion of the flexible portion anteriorly in a coronal plane. Therefore we decided to cut the anterior surface as close to a 45 degree angle as possible, that is, 50 degrees. The cut was made approximately where the average subtalar joint of a human foot would be located. When we sectioned this model lengthwise the motion of the flexible keel was definitely influenced by this oblique angle. We designated this the subtalar angle of the bolt block (Figs. 2 and 7).

When the patient-subjects were given the new model, the rotation of the heel section inverted in a much slower and better controlled manner.

Summary

With the aid of research previously carried out on the mechanics of the foot by various research groups, particularly the investigators at the Bio-Mechanics Laboratory of the University of California, San Francisco and Berkeley, we have developed a new prosthetic foot. During the last six months, thirty-five patient-subjects have been fitted. Successful results have been obtained by thirty-two subjects. Two subjects rejected the new design, and report on the one remaining subject could not be obtained.

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Footnotes

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