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The foot and footwear*

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Coming from a country where the majority of people walk in unshod feet and where shoes are an expensive luxury, I would like to offer some unconventional thoughts which might provide a global overview of the problems involved in the design of footwear and foot and ankle orthoses.

I was brought up with the traditional concept of the foot being merely a static tripod or a semi-rigid support for superincumbent body weight. The arches of the foot were held to be very important and the height of the longitudinal arch was correlated with a superior kind of foot.

We owe it to Hicks (1961) of Birmingham who taught us to clearly differentiate between a beam, an arch and a truss, and the role of muscles and the plantar aponeurosis in the weight bearing and balancing mechanism of the foot.

But the foot can no longer be viewed as a mere weight-bearing device meant only for standing. It has evolved, as the bioengineering team at the University of California (1947) has been telling us over the last two decades, as a dynamic mechanism functioning as an integral part of the locomotor system. The human foot is primarily meant for walking and it is important to understand its behaviour in the walking cycle.

The foot is a remarkably mobile mechanism, adapting itself to the contours of the ground as well as responding to the forces transmitted via the suprapedal segments from above. By supinating and pronating, it causes the weight line to follow a curved pathway leading to an even load distribution on the metatarsal heads. It becomes supple as it is grounded from heel strike to foot flat, and rigid as it prepares itself for heel rise and the final take-off. There is a locking mechanism in the foot, operated through the medium of the subtalar and transverse tarsal articulations, using the musculature as well as the windlass effect of the plantar aponeurosis (Figure 1). Inversion of the

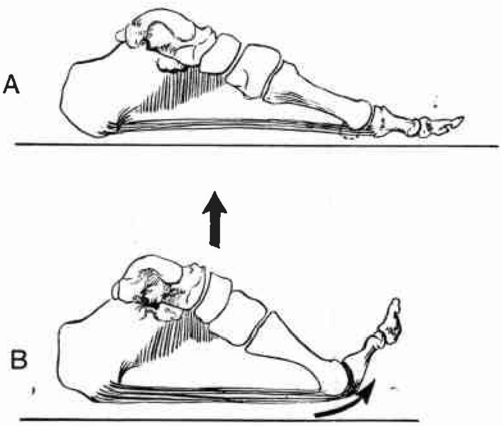

Fig. 1. Diagrammatic representation of "windlass action". A. Foot flat. B. Increased tension of plantar aponeurosis caused by dorsiflexion of toes, with resultant elevation of longitudinal arch.

heel against a fixed forefoot locks the foot; eversion of the heel unlocks it. Curiously, the traditional designs of orthoses have been preoccupied essentially with the ankle joint and have more or less neglected the subtalar joint. Lehneis (1965), in an attempt to refine the congruence of the orthotic ankle joint to the anatomic ankle axis, which is directed laterally and backwards, designed his torsion measuring device. But the problem is not nearly so simple. The ankle axis is not truly horizontal but also inclines downwards and laterally, having a range of variation from 68 degrees to 88 degrees, with a mean of 79 degrees (Figure 2). The implication of this obliquity is that when the leg is

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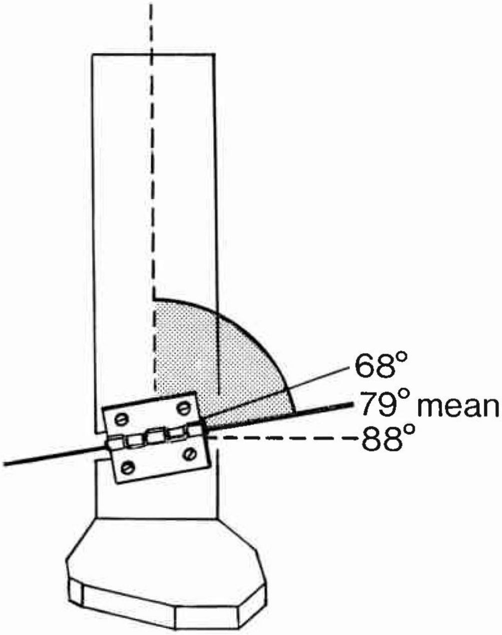

Fig. 2. Variations in inclination of axis of ankle joint, derived from anthropometric studies of 107 cadaver ankles at the University of California, San Francisco. (After Inman.)

fixed, the free foot deviates medially in plantarflexion and laterally in dorsiflexion (Figure 3). When the foot is fixed, as during the stance phase of walking, the tibia has to rotate medially or laterally (Figure 4).

The subtalar joint, to my mind, is the key joint in the mechanism of the foot. It has several roles to play.

(1) It allows adaptation of the foot to uneven ground or slopes. This not only results in an even distribution of load on the sole of the foot but also protects the more proximal segments of the limb. This is brought out very clearly when we see the ground reactions transmitted up to the stump-socket interface and bruising of the stump when our villagers use prostheses with conventional uniaxial ankles or SACH feet. In a way, therefore, the subtalar joint acts as a shock absorber.

(2) It acts like a mitred hinge with its axis inclined about 45 degrees to the horizontal plane. It, therefore, functions as a torque converter, with a one-to-one ratio (Figure 5). When the leg is rotated inwards, the foot pronates; when the leg is rotated outwards, the foot supinates (Figure 6).

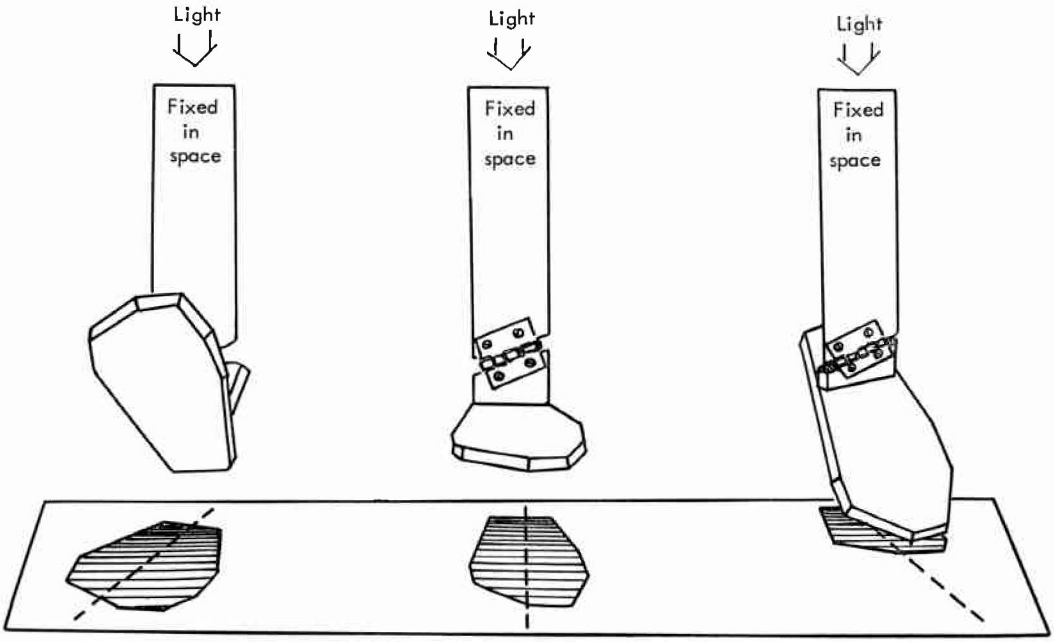

Fig. 3. Effect of obliquely placed ankle axis upon rotation of foot in horizontal plane during plantarflexion and dorsiflexion with foot free. The displacement is reflected in the shadows of the foot. (After Inman.)

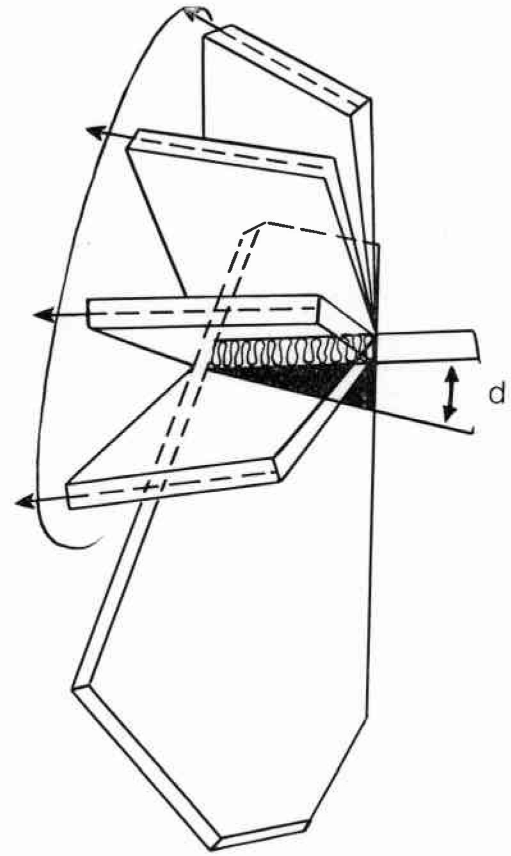

Fig. 4. Foot fixed to floor. Plantarflexion and dorsiflexion of the ankle produce horizontal rotation of the leg because of the obliquity of the ankle axis. (After Inman.)

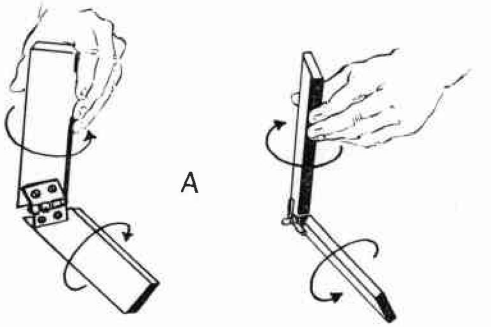

Fig. 5A. Action of mitred hinge. If the axis of the hinge is at 45 degrees, a simple torque converter is created. Rotation of the vertical member causes equal rotation of the horizontal member, with a one-to-one relationship.

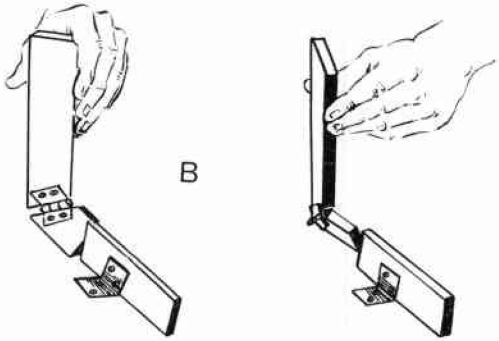

Fig. 5B. To prevent the entire horizontal segment from rotating, the horizontal member has been divided into a short proximal and a long distal segment with a pivot in between. (After Inman.)

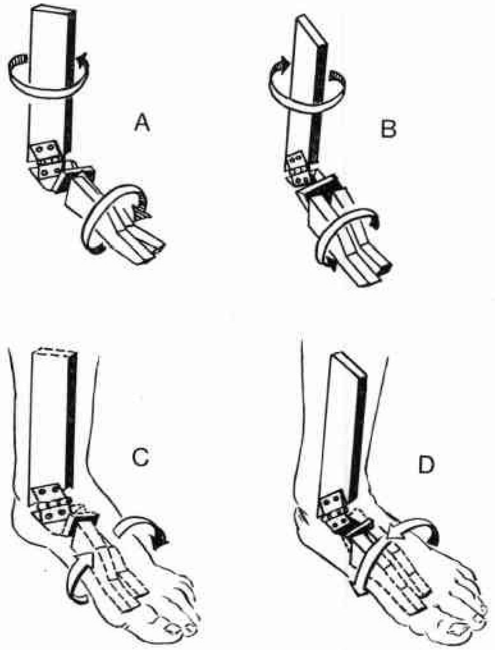

Fig. 6. Distal portion of horizontal member replaced by two structures A and B. Mechanical analogue of principal components of foot C and D. Mechanical components inserted into foot and leg. External rotation of the leg causes inversion of the heel, elevation of the medial side of the foot and depression of the lateral side. Internal rotation of the leg produces the opposite effect on the foot. (After Inman.)

(3) If there were no subtalar joint, the obliquity of the metatarsophalangeal break would cause the leg to tilt out during heel rise and toe-off. Here, the subtalar joint acts as a correcting device, allowing the leg to remain vertical (Figure 7).

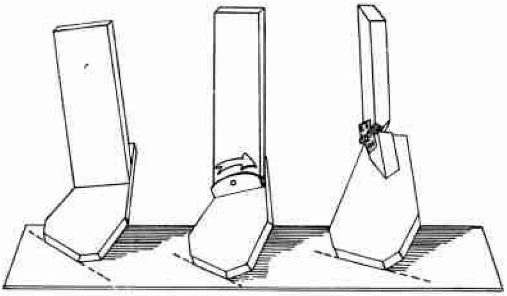

Fig. 7. In spite of the obliquity of the metatarsophalangeal break, raising of the heel allows the leg to remain vertical. This is another function of the subtalar joint. (After Inman.)

The close relationship between subtalar movement and horizontal rotation of the tibia during gait cycle has been demonstrated in several studies (Figure 8). From heel strike to foot flat, the foot pronates and the tibia rotates medially. During mid-stance, the foot supinates and the tibia reverses its rotation. At heel rise and toe-off, the heel is in inversion, the tibia is rotated out. This tibial rotation has not received the attention it deserves in the design of foot orthoses and prostheses. I would like to emphasize that most quantitative data on subtalar movement and tibial rotation has been collected on level walkway studies. It is inevitable that the dimensions of these movements would be considerably greater when walking on irregular ground.

Any footwear, shoe modification, or a prosthetic foot should respect this system of linkage between the various articulations of the human foot.

In unshod communities the foot muscles get freedom for exercise and the joints remain supple. This is why functional disorders of the foot are so rarely seen in such people.

However, footwear becomes necessary to protect the skin of the sole against rough ground or hard pavements and to keep the foot warm in cold climates. Over the ages, various communities have used locally available raw materials to design footwear which suit their own needs, and it is unnecessary and unwise to dictate to them that only a particular design of footwear is correct. In fact I sometimes like to look upon closed shoes as braces, a necessary evil, which take up the work of muscles, causing them to atrophy from disuse and make the joints stiff. I feel it is wise, at least for some time during the day, to take the shoes off and allow the foot to function unhampered. Use of an open sandal without a heel strap, where the toes have to grip to prevent their slipping off, is an excellent exercise for the intrinsic muscles. For those who may argue that this cannot be applicable for the cold climates, I suggest the example of a Japanese Geta and Tobi, a very sensible combination to provide warmth as well as freedom of foot action (Figure 9).

In short, for normal feet, the use of a shoe is optional. But when a foot is abnormal a shoe becomes compulsory, to realign the foot, to redistribute pressure especially in an anaesthetic sole, to accommodate deformities, to equalize leg lengths, or to act as a foundation over which an orthosis has to be erected.

I feel a little puzzled about the long and repetitive list of shoe modifications which are illustrated in most texts on foot ailments. Whenever many alternatives are available for a particular situation it usually means that none is really effective. Also, I have never been sure that when I prescribe a wedged heel or an arch support, I am really being rational. I am particularly sceptical about arch supports which are based on a static concept of the foot. The advertising media exploit such devices, and I do feel that some long term controlled clinical trials should be held to evaluate these devices. All useless and occasionally harmful modifications should be pruned. I must confess, however, that there is a great temptation to use them as placebos.

There is a need for controlling an abnormal inclination of the heel such as in valgus feet. Plastic heel seats were described by Helfet (1956) and later modified by Rose. For some reason, I have not seen them being used. I wonder why? The UCBL foot support (Henderson and Campbell, 1969) developed at the Biomechanics Laboratory, University of California appeals to me. Its fabrication, however, is not such as to be used except by skilled orthotists and it requires raw materials not available to us in the underdeveloped countries. I do feel that there has been no clear understanding of this problem and its solution so far.

From my basic theme of emphasizing the mobility of the foot, I now move in the opposite direction, that is, the need in some situations, for rigid footwear. Further work on the design of

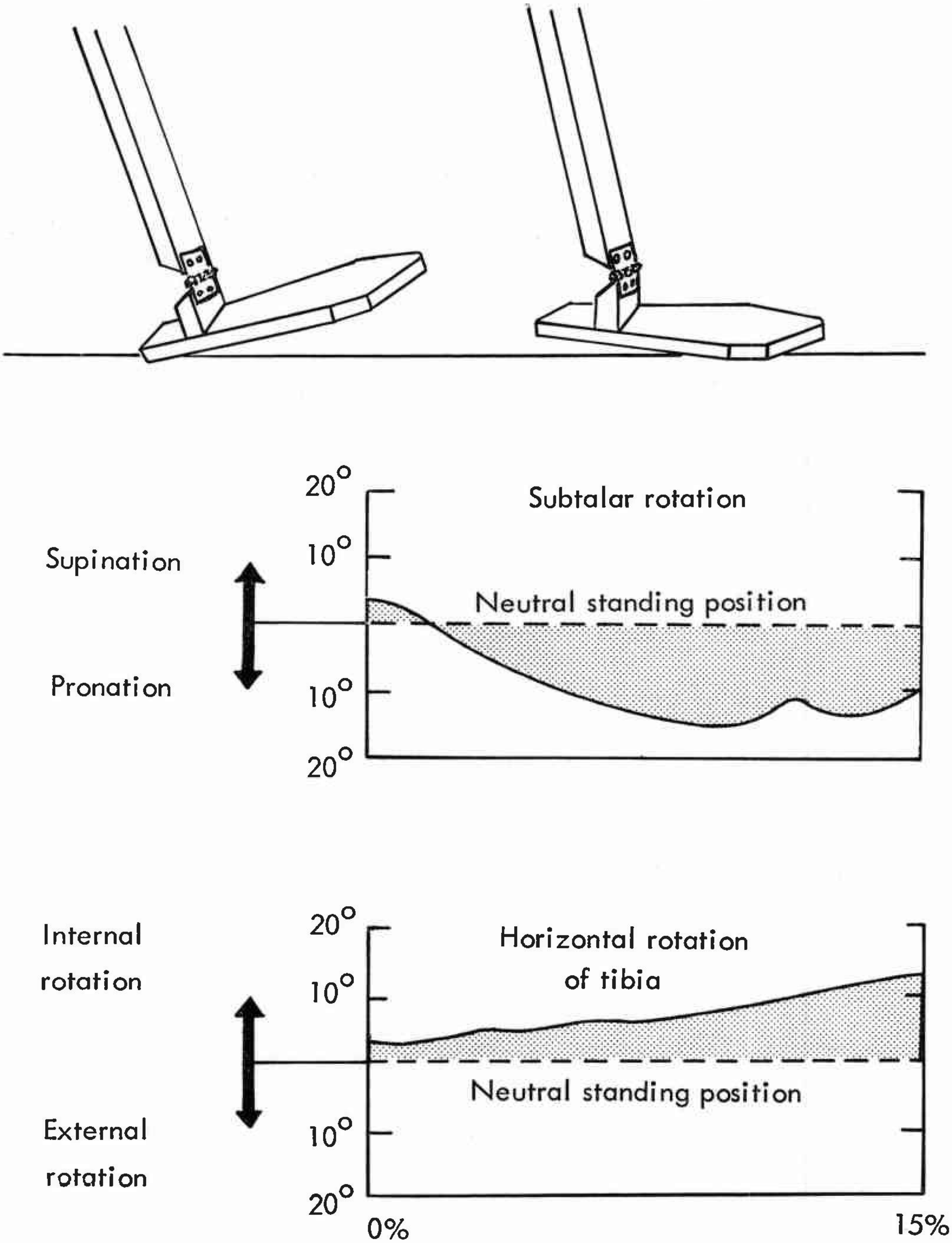

Fig. 8a. Subtalar rotation and horizontal rotation of tibia in gait cycle. Heel-strike to foot-flat—the heel is elevated and the tibia rotates medially.

rigid footwear has become necessary. This is so for several reasons. An absolute indication for such rigid footwear is the anaesthetic foot with healed ulcers or with toe deformities. Once the toes have lost their bite during push off, the metatarsal heads are exposed to tremendous

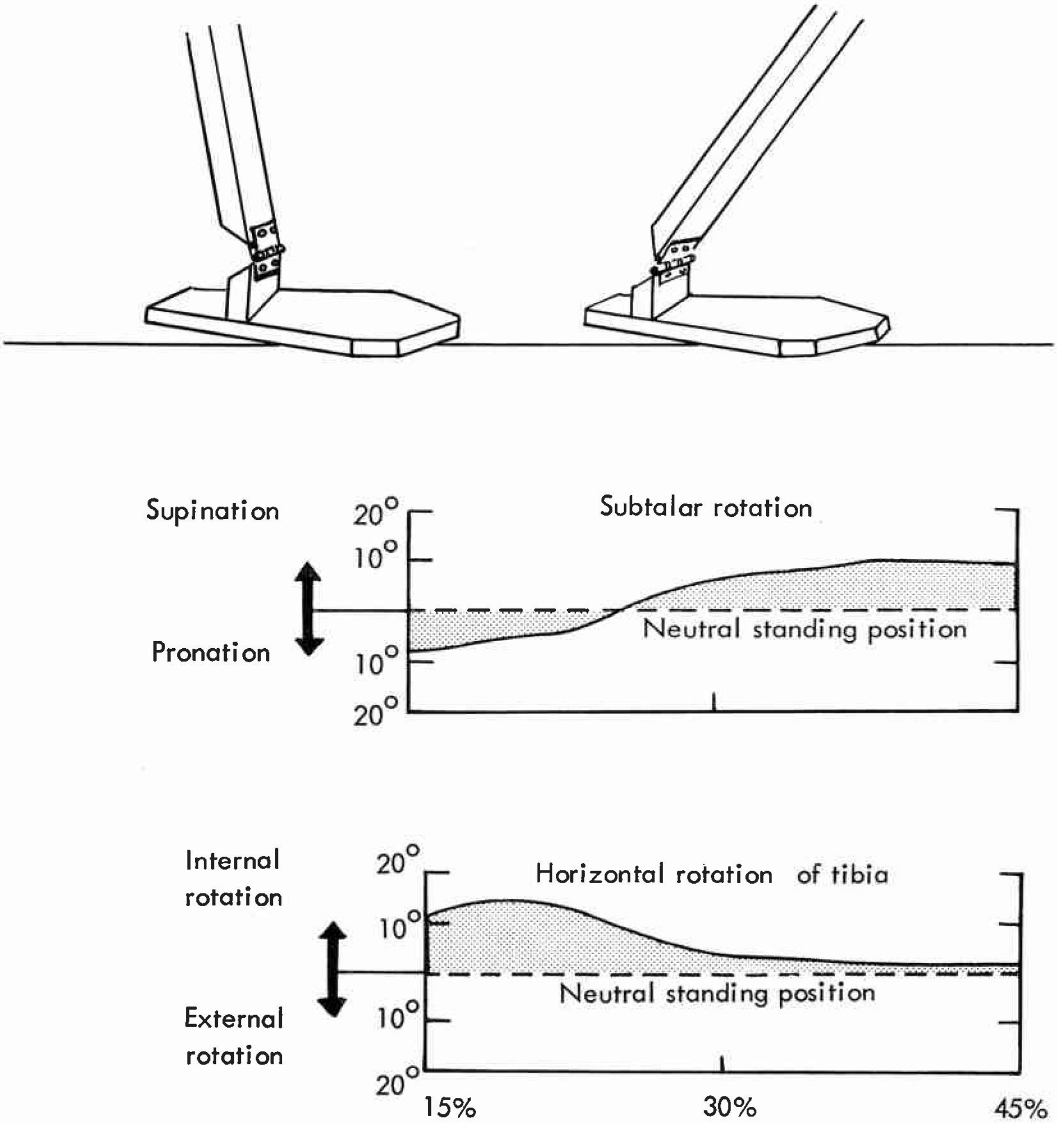

Fig. 8b. During mid-stance—the foot supinates and the tibia reverses its rotation.

pressure. There is also the shearing effect due to the roll of the metatarsal heads on the soft tissues underneath them. A microcellular rubber insole of 15 shore hardness, as described by Paul Brand (Ward 1964) can only prevent ulcers in an unscarred balanced foot. This resilient insole, over a wooden clog with a rockered sole, abolishes peak pressures on the sole during the walking cycle. I think there is still considerable scope for research in the problem of redistribution of load on the anaesthetic skin. A resilient insole is only one approach. There can be other approaches.

Such rigid wooden clogs have also been used in a completely different context by Huckstep (1974) in Uganda for polio patients. Huckstep was concerned about the cost of leather shoes, the long waiting lists piling up in his workshop, and the lack of trained orthotists. Therefore, he tried to simplify the fabrication of calipers by using wooden soles and a rocker bar as a substitute for orthodox shoes. This is a realistic

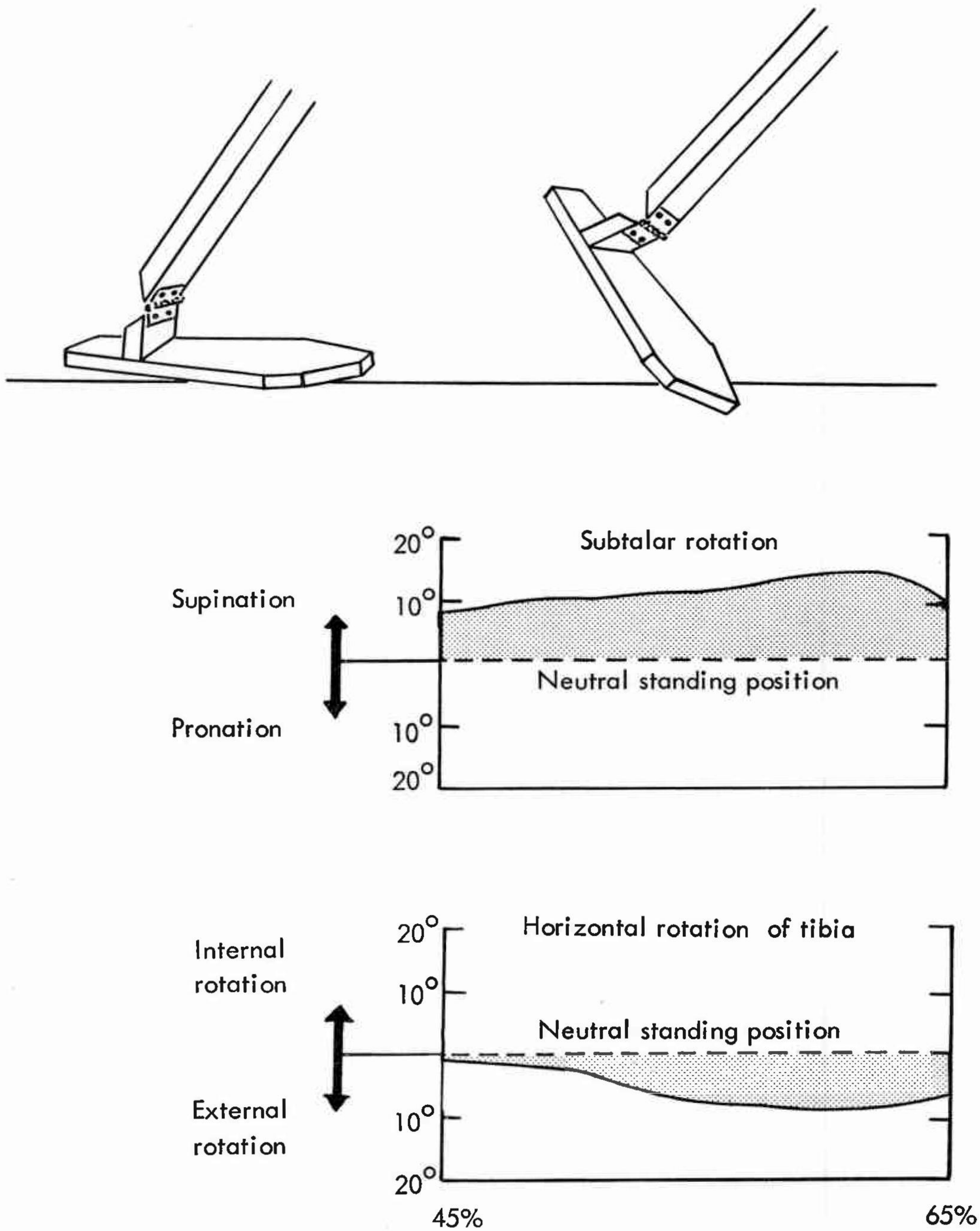

Fig. 8c. At heel rise and toe-off, the heel is in inversion and the tibia rotates out.

approach in underdeveloped countries.

Rigid footwear with a rocker sole has been advocated in a more sophisticated setting by John Glancy (1971) where he has used it to programme the heel rise by the body tripping over a rocker bar. A preset toe-off metal shoe plate can convert the entire length of the shoe into an efficient "resistance arm" which not only prolongs the life of a shoe, but allows a "booster" spring to act efficiently.

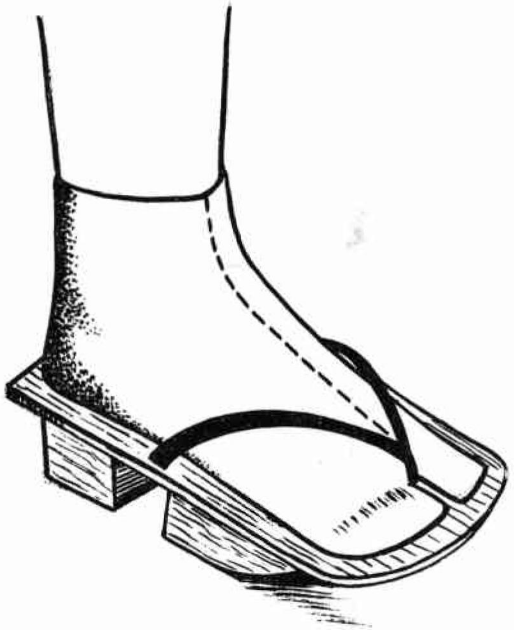

Fig. 9. A Japanese Geta and Tobi provides a very sensible combination of warmth and freedom of foot action.

The crux of the matter lies in the design of the rocker bar. If a narrow rocker bar is used, the tip-off is very abrupt (Figure 10). Its height is critical; the further back it is placed, the higher it has to be to prevent the front edge of the clog from striking the ground. A Lancashire clog gives a more smooth roll. However, when side bars are fixed to rigid footwear, as in Huckstep's design for paralytic feet, the line of progression of the limb is difficult to match with the pathway of the rolling clog. The arc of curvature of the rocker would have to vary with the length of the step, the length of the leg and whether or not the knee was affected. It appears well nigh impossible to devise a standard formula. John Girling has devised an ankle-foot orthosis for leprosy patients which has many thoughtful features (Wollstein and Girling, 1972). It takes into consideration the angle of inclination at heel strike, a broad area of support in mid-stance, and a suitable take-off angle. Is it possible to utilize the SACH wedge principle to simplify the solution? If an easy bio-mechanical solution can be worked out, it would be a great boon to the poorer countries (Figures 11, 12). We have some very skilled craftsmen who can handle wood with amazing dexterity and we can then make full use of them to produce a cheap and sturdy appliance.

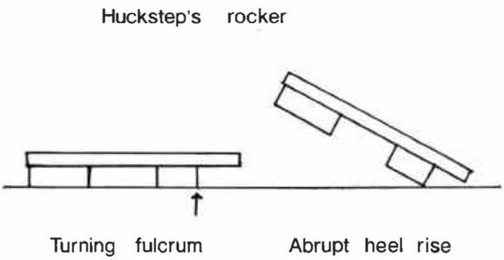

Fig. 10. Huckstep's wooden rocker provides an abrupt heel rise.

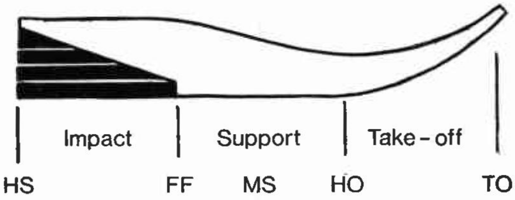

Fig. 11. Design of a rigid wooden footwear currently being used at Jaipur, utilizing a SACH wedge for the impact surface, a broad surface for mid-stance and a gentle rockered surface for take-off.

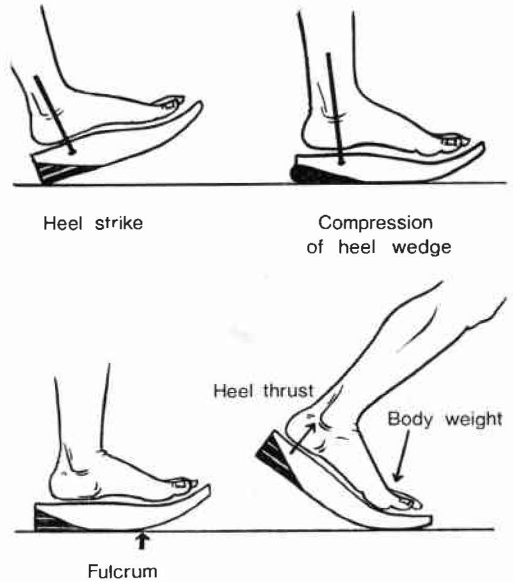

Fig. 12. With the foregoing design, an ankle joint is not needed. The orthosis can be quickly and very economically fabricated.

Top, compression of SACH wedge leads to grounding of foot. Bottom, body weight can be harnessed for simulating action of plantarflexors. Excellent performance characteristics are seen in paralytic calcaneus feet.

Finally a word about foot and ankle orthoses. There has been an unfortunate "orthotic lag". In spite of rapid progress in the field of prosthetics, the design of orthoses has witnessed, till lately, no basic change over the last half century. Only recently, a break-through has appeared. There is a greater realization that any mechanical analogue should match the biomechanical system. I have already referred to the value of subtalar movement in the walking cycle and I would like to have a device which preserves it, especially when most of my patients have to walk on a rugged terrain. The UCBL dual-axis ankle was, in my opinion, a step in the right direction. But the recent work by Lehneis (1972) at New York University, such as the use of a plastic spiral ankle-foot orthosis, is much more appealing because these orthoses rely on the natural anatomic axes rather than some arbitrarily imposed external joints of the traditional brace systems. While plastics seem to have a great future, they are still very expensive and being rigid, require a great deal of expertise in the fabrication of such orthoses to avoid the creation of pressure points. Also, I am not quite sure whether these can withstand the merciless beating in our rugged terrain. It is in this context that I want you to consider the use of a cord-cum-rubber bracing, which is more akin to the properties of skeletal muscle

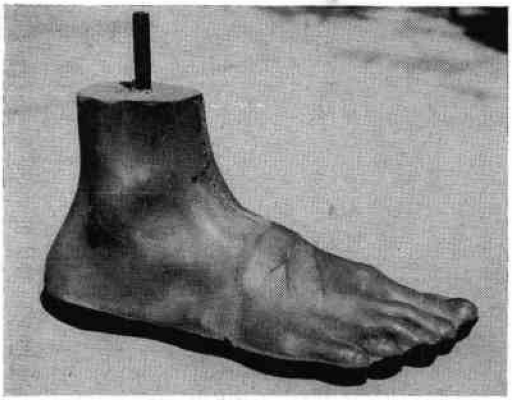

Fig. 13. Jaipur Foot—The external covering is vulcanized rubber with rayon cord reinforcement.

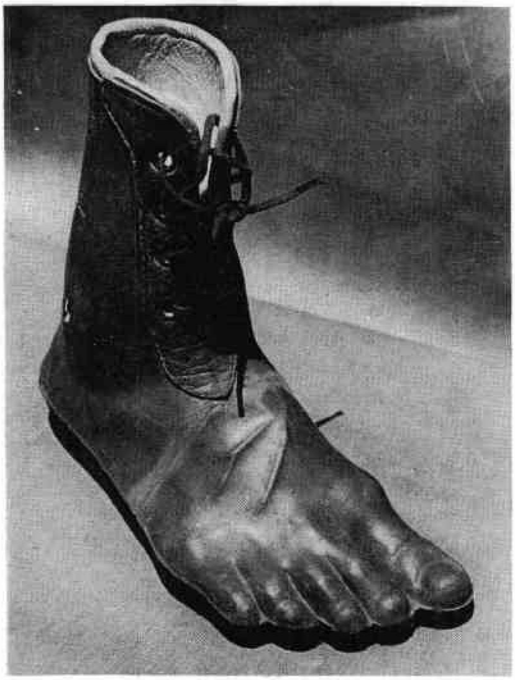

Fig. 14. Jaipur version of Syme's prosthesis. The cord-cum-rubber provides a combination of resilience and stability.

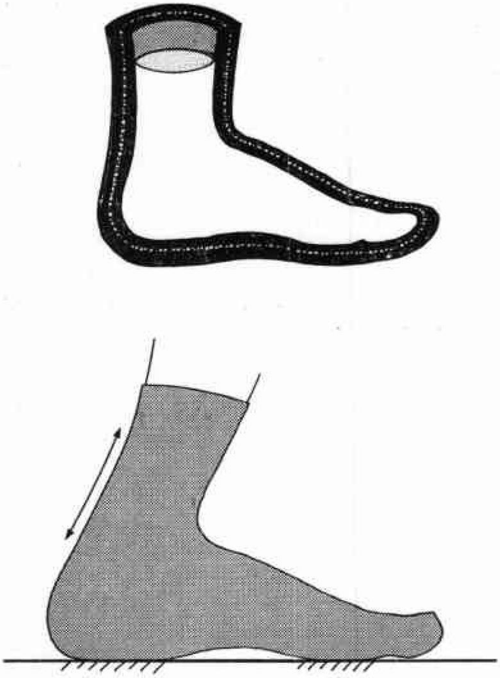

Fig. 15. Schematic design of a foot orthosis made of vulcanized rubber and cord reinforcement. Movements are permitted at natural joint axes. Reasonable stability is conferred on a paralytic foot. The orthosis is worn like a gym shoe.

and is not subject to the failure encountered in all mechanical spring systems. As a "spin-off" effect of our work on the Jaipur Foot (Figure 13) and its modification for Syme's amputation (Figure 14), it has recently occurred to us that we could extend the same principle for orthotic purposes. If a Syme's amputee can feel secure in this resilient container, I see no reason why a flail ankle and foot cannot be stabilized by this device. I am sure it can be used to programme motion in any direction. We intend to pursue this idea and have a feeling that, at least for problems of the foot and ankle, it may answer many of our needs (Figure 15).

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