PROBLEMS IN THE DESIGN OF A COSMETIC CALIPER[a]

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INTRODUCTION

The clinical caliper [orthosis] shown in Figure 1 makes no pretense at cosmesis. However well made, it is ugly and disfiguring and has been so since its introduction a century or so ago. The functional design also leaves much to be desired, but the psychological distress caused to the wearer by the ugliness of the appliance is perhaps its greatest fault. In an attempt to improve this situation collaboration between the Salford Royal Hospital and the University of Salford on the design of a cosmetic caliper started in 1964 and this paper records the progress to date.

There have, of course, in past years been several attempts to improve the appearance of these appliances, but the total research effort has been small and largely unsuccessful. More recently, a number of research projects using sheet plastic have been undertaken mainly to provide cosmesis for non-weight-bearing and below-knee calipers and, in one or two cases, for weight-bearing calipers with fixed ankle. In the case of the Salford caliper, however, a full length, fully weight-bearing structure and a moving foot were essential requirements.

Initial study showed that the design would have to be a radical departure from either the existing caliper or the artificial limb. As seen on Figure 1 the struts of existing calipers are clear of the limb and cannot be successfully disguised, while in the case of a lower limb prosthesis there is no human limb to inhibit the design of the structure and very effective cosmesis is possible. The cosmetic caliper is, therefore, in a different category to both as the structure has to be worn under the stocking and should fit comfortably into a normal shoe. The new design would therefore have to be a compromise between cosmesis, comfort, and strength with the need for light weight, resistance to

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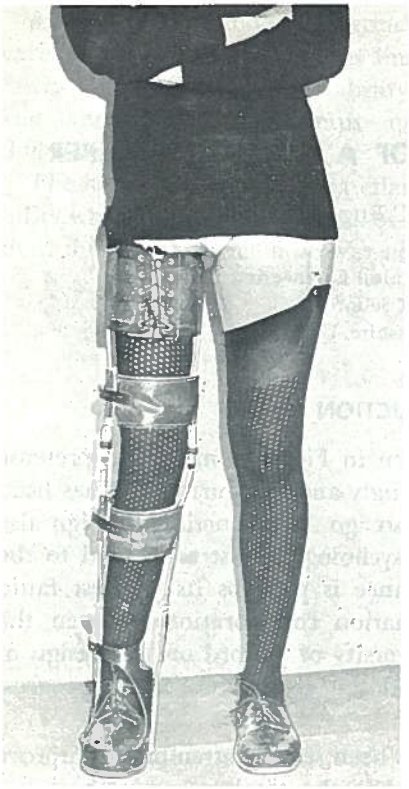


FIGURE 1.—Existing caliper.

metal fatigue, and complete reliability as vital engineering considerations.

MEDICAL REQUIREMENTS

Designed originally for patients paralyzed by poliomyelitis, the caliper structure was required to support the full body weight and provide stability for the knee and ankle. The thigh corset should be arranged to take the load from the ischial tuberosity and any knee or ankle straps necessary should be fitted. Dorsiflexion of 5 deg. and plantar flexion of 10 deg. must be permitted and controlled with suitable "drop-foot" stops and some inversion and eversion with medial and lateral rotation were desirable. Flexion at the knee of at least 90 deg. was also necessary with suitable locking.

The poor circulation usually present in cold weather in limbs paralyzed by poliomyelitis was another medical consideration and it was believed that this condition would be improved if the affected limb was protected by the caliper and its cosmetic cover.

The major anatomical difficulties were those presented by the frontal width at the ankle and knee joints. While a paralyzed limb is usually

thin and wasted, the bones continue to grow and although the lengths of the bones in the affected limb may be much shorter than those of the sound limb the frontal widths at the joints are usually little different. The structure must, therefore, pass the ankle and knee with the minimum acceptable clearance if the width of the cosmetic caliper is to compare reasonably with that of the sound limb.

ENGINEERING DESIGN

The engineering problems were more complex. An early study of the probable loads and the design problems involved showed that plastic construction was unsuitable.

The strongest plastic found in the survey of possible materials had an ultimate tensile strength of 14,000 lb./sq. in. which should be compared with the high tensile steel eventually used for this project with its maximum tensile strength of over 150,000 lb./sq. in. Furthermore, plastics, being prone to cracking in areas where there is a concentration of stress, are unsuitable for structures which are required to pick up load at discrete points; e.g., the joints at the foot and the knee on the cosmetic caliper. Although this can be overcome by a carefully designed laminating in the affected areas, specialized production techniques are involved, and furthermore any increase in the thickness of the ankle was not acceptable for cosmetic reasons.

While, therefore, appreciating the advantages of light weight and pleasant appearance which the use of plastics might have given, the structure of the cosmetic caliper was made of metal for reasons of structural efficiency, using modern lightweight design techniques to provide a light, stiff structure.

A lightweight structure in any material is liable to be unstable and subject to metal fatigue. The problem of instability would be aggravated if light (Duralumin) alloys were used because of their relatively low modulus of elasticity (10×10^6 lb./sq. in.) and there seemed to be no possibility of improving this by using titanium ($E = 15 \times 10^6$ lb./sq. in.) because of the very high cost of this material. It was, therefore, a straight choice between the readily available materials of Duralumin alloy or steel ($E = 30 \times 10^6$ lb./sq. in.). In the further assessment the vulnerability of the light alloys to metal fatigue when compared with the known good qualities of a high strength steel had to be carefully considered. Furthermore, the possibility that steel struts would be little, if any, heavier than struts of Duralumin alloy made the use of steel struts attractive, particularly if thin metal sections were used.

A design consideration of vital importance in gaining the patient's confidence was the need for the structure to be "fail safe." The experimental nature of the new caliper and the fact that the structural members were close to the human limb, together with the possibility of

fatigue failure in the thin metals likely to be used, were sufficient reasons for trying to ensure that if failure should occur the wearer would not be injured. As an additional safety precaution it was desirable that any fatigue cracks which might occur should propagate slowly and that, if possible, vulnerable areas should be left uncovered by leather or padding to facilitate inspection.

Simplicity of production was also essential if the caliper was to be produced commercially and there was, of course, the need to ensure that the appliance was continuously reliable, easily maintained, and that it would have a reasonable life.

Apart from the above considerations, the engineering design itself is complex in that the basic structure is redundant and consists of struts of varied shape and irregular cross section. These features combine to make strut calculations even less reliable than is usual, a situation which, combined with a varying loading pattern, dictates that academic analysis and laboratory testing be used to indicate the direction of design rather than to prove it.

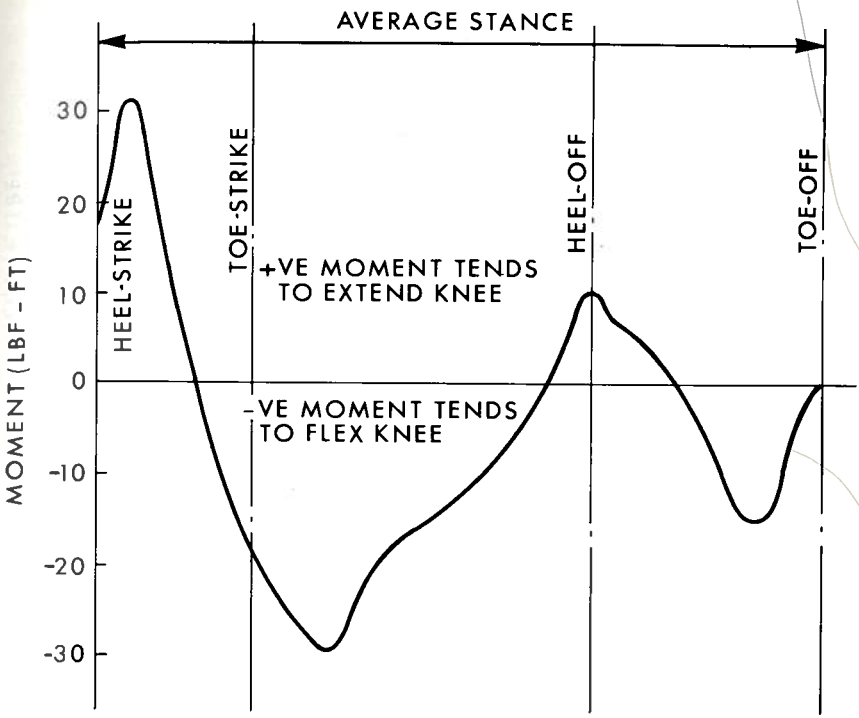
LOADING

The external forces acting on the body during walking are of only two types: a. gravitational, due to the weights of the various parts of the body, and b. reactions from the ground on one or both feet. The pattern of this loading will vary with gait and other peculiarities of movement and will, therefore, tend to vary from person to person. This is so for normal people and it can be expected that this situation will be aggravated for people who have disabled lower limbs.

In normal gait the knee flexes with the effects indicated on Figure 2. The gait of the monoplegic will not follow this pattern as the knee will be stiff, but a study of available data suggests that the maximum (vertical) load will be similar for both normal and monoplegic gait. Horizontal and torsional loads at the foot may be expected to compare similarly and, as loading data are available in greater detail for normal gait, it is reasonable to use this for design analysis. Furthermore, because of the varied patterns of gait, particularly of the monoplegic, the loading can only be regarded as a guide to design with the desirable probability that the loading will be somewhat pessimistic if that for normal gait is used. The typical loading curves shown were therefore drawn up, each being modified to suit a body weight of 140 lb.

Vertical Component

For the vertical component of the force created when walking in a straight line, most investigators show curves with two peaks and a trough as shown on Figure 3, the first peak occurring shortly after toe-strike when the body rolls over the supporting leg, and the other



TYPICAL MOMENT AT KNEE JOINT
(BODY WEIGHT - 140 LBF)

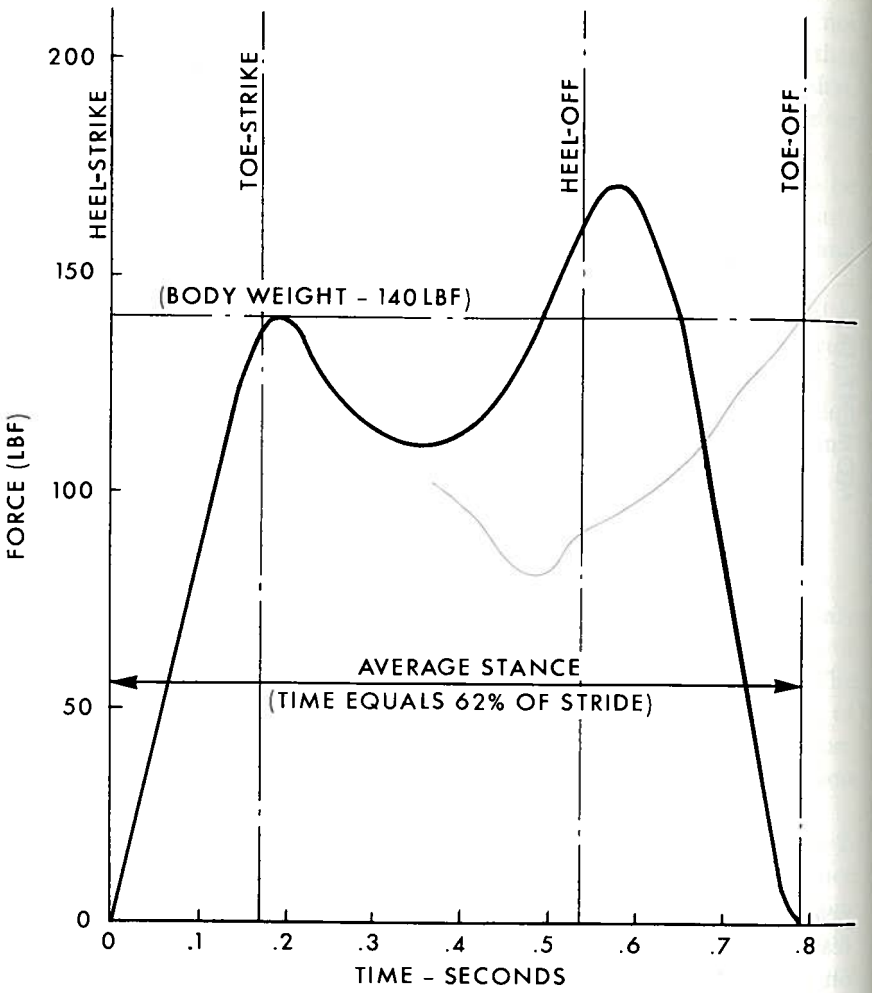
FIGURE 2

peak occurring when the leg "pushes" the body up just after the heel leaves the ground. The 20 percent increase over bodyweight indicated on the curve has been used for straight walking and hence for fatigue considerations, but as it is found that on turning a right angled corner a maximum vertical force of 150 percent of body weight may be recorded this latter figure has been used for the unfactored static design load.

HORIZONTAL COMPONENT

The horizontal component of the load is found to follow the pattern shown on Figure 4a. For straight walking the horizontal component seldom exceeds 20 percent of the body weight although where the subject is turning a corner this tends to rise to a maximum of 30 percent.

The lateral component of the horizontal force is found in all cases to be very small and, when it does occur, to be usually directed outwards from the foot. On account of its small magnitude it was considered, however, that this component could be neglected.



TYPICAL VERTICAL FORCE ON LOWER LIMB
(NORMAL GAIT)

FIGURE 3

Torque

The amount of torque in straight walking is also small, but the pattern varies considerably with individual gait. A typical pattern is shown on Figure 4b. As might be expected the torque maxima when turning a right-angled corner are greater than those for straight walking and are found, on average, to be nearly twice as high.

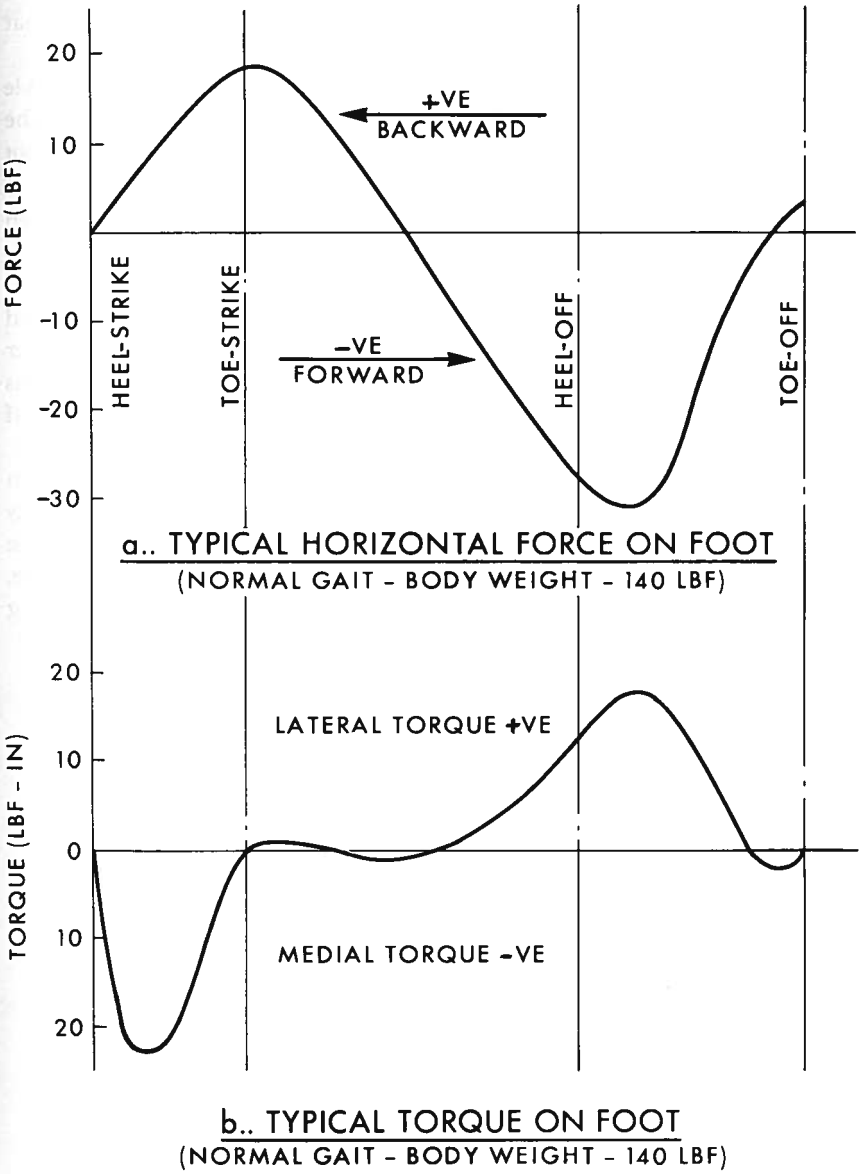


FIGURE 4

THE FIRST PROTOTYPE

With these requirements in mind the first prototype was designed and made primarily to investigate the loading and fitting problems, and also to acquire the "feel" of the project. A specially selected 18-

year-old model who could be relied upon to cooperate intelligently was chosen, and plaster casts were made of both her lower limbs so that comparison between the two could be made throughout the design.

This prototype is shown on Figure 5. It will be noted that the ankle joint is positioned in the region of the natural ankle while below the joint an 18 S.W.G. (.048 in.) mild steel stirrup, shaped to fit the foot and ankle, is embedded in a fiber glass heel piece.

Side struts of 18 S.W.G. (.048 in.) commercial Duralumin, stabilized by flanging the edges at 45 deg., were used between the knee and ankle joints, and two stabilizing arches were fitted behind the calf. A single knee lock was fitted. Adequate leather-covered padding was provided as shown to protect and support the ankle. The steel stirrup and fiber glass foot fitted comfortably into the existing shoe and cosmesis was provided by a plastic calf cover, split like a gaiter down the inside of the leg, and secured to the inner strut with Velcro.

It was later found that a design of foot similar to this had been produced by earlier designers, but had not been successful and, early trials of the Salford caliper showed the reasons for this. Although the appliance fitted the model well and greatly improved her appearance, regular use proved disappointing and gave rise to the following criticisms:

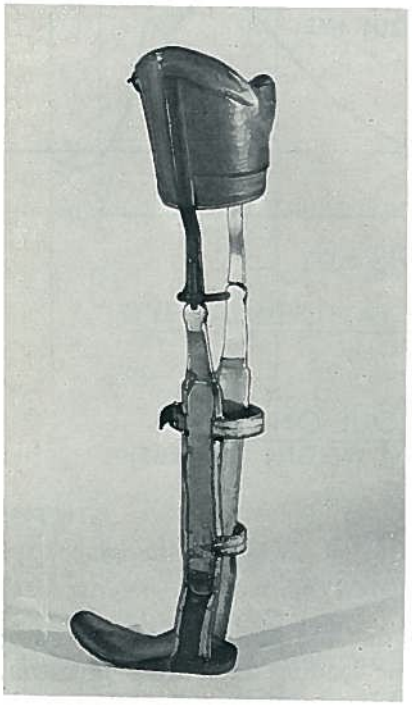


FIGURE 5.—First prototype.

1. The presence of the joint at the ankle prevented the struts being manipulated to follow the contour of the malleolus and so hindered the fitting of the appliance to the patient and also created an unnecessarily bulky ankle.
2. The ankle joint, positioned on a strut, was inefficient structurally, particularly as the necessary slimness of the joint weakened it laterally and restricted the bearing surfaces.

The relatively brief trials of the first prototype, however, provided valuable information enabling the real problem of the project to be evaluated.

THE SECOND PROTOTYPE

In the reappraisal of the design it became clear that the solution to the problem lay mainly in the design of a lower strut which would allow ample clearance at the ankle (without being bulky) and which could be manipulated to suit a variety of deformities. It was, therefore, decided that the ankle joint as designed on the first prototype must be eliminated both because of its probable unreliability and because of its intrusion into an area which would require to be manipulated to accommodate different shapes and sizes of ankle, and also because of its bulk.

Study of the problem showed that the ideal joint would have the following advantages:

1. It would be clear of the area around the ankle where the side struts would require to be manipulated.
2. It should permit the side struts to enter the shoe at any angle (vertical or horizontal) necessitated by the shape of the ankle or foot.
3. It should not create a fixing moment which would cause high stresses in bending about the minor axis of the side struts. This last is important as earlier study of the problem had shown the probable need to flatten the strut section where it entered the shoe thereby impairing the stability of the section and possibly weakening it to a point where it would fail under the stress of a fixing moment.

Consideration of the above requirements led to the conclusion that the ideal joint would:

1. be positioned close to the end of the side strut,
2. have a universal action.

This reasoning led to the development of a design using two ball joints. A cheap standard type of hardened steel ball joint as used extensively in car production was checked for strength and mounted in a 1-in.-thick foot block, the joint now being positioned below the heel.

Rigid attachment of the ball joint stubs to the side struts required welding and it was therefore necessary to change the material of the side struts from light alloy to stainless steel. Furthermore, as manipulation of the struts adjacent to the ankle was now considered to be essential for comfortable fitting it was decided that steel, with its higher modulus of elasticity, would be preferable to light alloy to maintain maximum stability of the thin metal.

18 S.W.G. stainless steel (Specification S521) was therefore used in the softened condition to facilitate manipulation. The cross section was also changed to a circular arc 1¼ in. wide and 1½ in. internal radius to give a higher degree of stability under compressive loads and a better fit under the gaiter.

The lower stabilizing arc behind the calf was eliminated as this had been chafing, and the upper arc was doubled in depth (to 2 in.) to compensate for this. Above the lower structs the same components as used on the prototype caliper were fitted and a similar calf mold was again attached with Velcro. On fitting to the model easy clearance was allowed at the ankle and adequate leather-covered padding was used.

This second prototype caliper, with the above modifications incorporated (shown on Fig. 6) was put into daily use and after 9 months of continuous use fatigue cracks were found in the 18 S.W.G. steel struts. After investigation, it was decided to modify the side struts to prevent further failure. This was done by increasing the metal thickness to 16 S.W.G. (.064 in.) and by using the same steel in the hardened condition (Specification S520).

Following promising trials of the second prototype caliper described above, calipers were made for three other models, all teenaged girls and all using fully weight-bearing calipers. These appliances were basically the same as the second prototype and the extended trials which we were able to carry out over some 2 years proved that the concept of a cosmetic caliper was practicable.

DEVELOPMENT

Following the trials recorded, field trials of another 20 calipers were sponsored by the Department of Health and further development took place.

The Foot Joints

The original ball joints described earlier stood up well to the job. They were, however, subject to hard usage and it was clear that several replacements would be necessary during a normal caliper life. As the ball component of the bearing had to be welded to the steel side struts replacement was difficult and it was, therefore, considered desirable to provide a more serviceable joint.

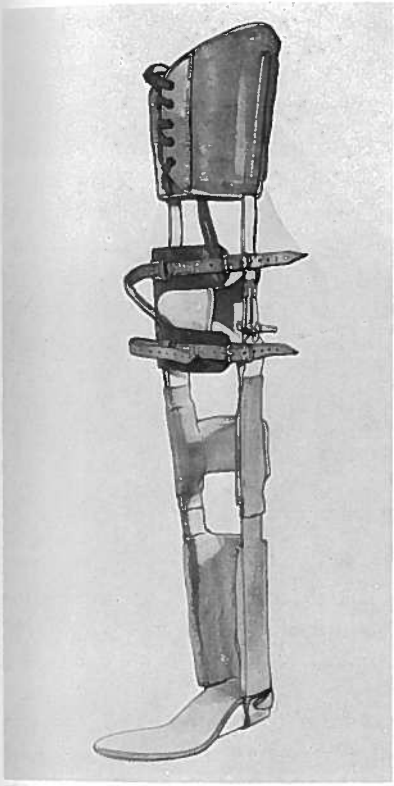


FIGURE 6.—Second prototype.

As a result of this study the Rose "Unilink" joint shown on Figure 7 was adopted. These were cheaper and the phosphor bronze cup and cone which constitute the bearing were readily replaceable. In use this joint has been most satisfactory showing little wear after long periods of rough use.

Foot Blocks

Using the above joint, foot blocks in several materials were tried out; viz., fiber glass, aluminum alloy, and plastic with and without metal inserts. Of these, the light alloy block was the most satisfactory, but its weight at 7 oz. was higher than desired and, as the trials advanced, it became clear that a foot block thickness of 1 in. was too great where the patient had a raised instep or similar foor deformity.

A study was, therefore, made into the possibility of reducing the thickness of the foot block. This was eventually achieved by embodying the inner "cup" of the bearing in a foot block of impregnated material (Nylatron G.S.) and modifying the "drop-foot" stop as shown on Figure 8. In trials to date, this foot block has been found to be very

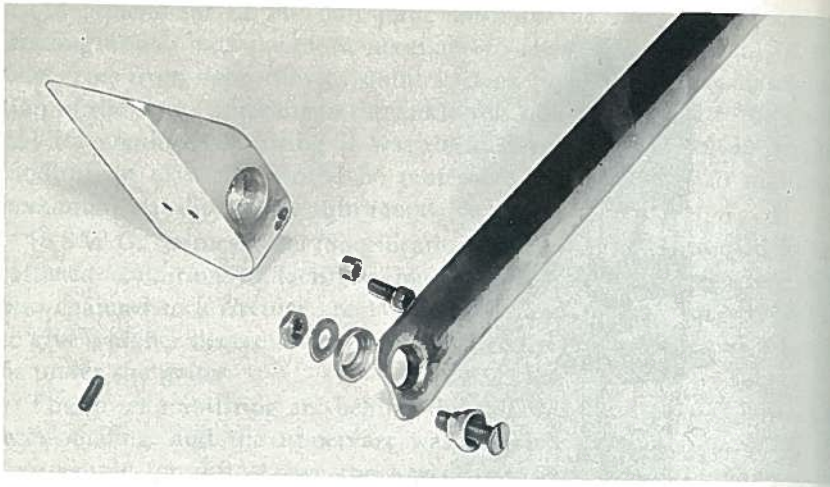


FIGURE 7.—Foot joint.

satisfactory and is expected to replace the thicker block of the earlier models giving a marked improvement to the fit in the shoe.

Knee Joints and Locks

The knee joints and locks were also modified and improved. On the earlier models only one knee lock was fitted. From the experience gained in the trials and on the advice of the Technical Branch of the Department of Health two locks were later fitted. This aggravated the problem insofar as stocking wear was concerned as the operating lever shown on Figure 6 had been a nuisance in this respect. Cable operation of encased locks as shown on Figure 9 solved this problem and permitted individual seating of each lock, a useful safety feature as one lock failing to close could not hold out the other.

These improvements are embodied in the latest caliper, designed for a weight of 140 lb. (64 kg.) and shown being worn by one of the models in Figure 10.

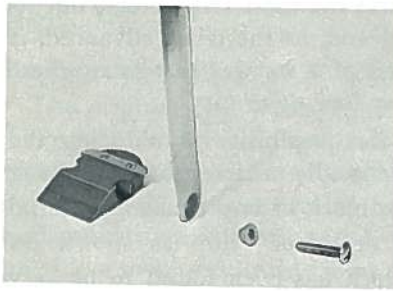


FIGURE 8.—Foot block.

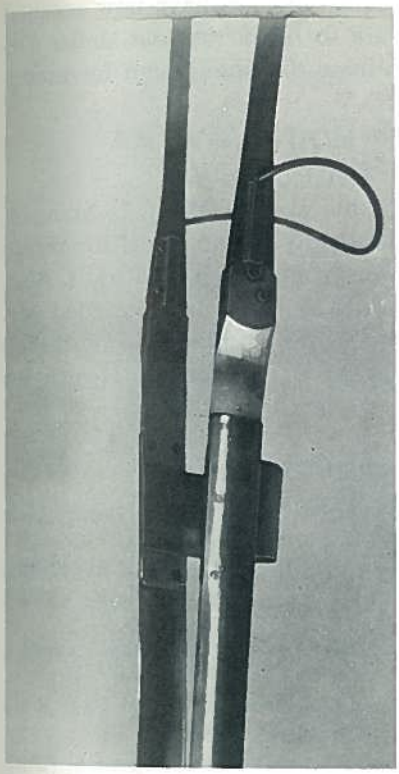


FIGURE 9.—Knee joint assembly and lock.

FIGURE 10.—Cosmetic caliper in use.

CONCLUSION

The Salford caliper has now been worn successfully by some 15 patients with various disabilities for up to 4 years and has been shown to have several advantages including improved function, a warmth which eliminates chilblains, and a design which permits normal footwear to be worn. Case histories of the models, particularly of women, have shown that to relieve disfigurement by improved cosmesis is to make the person less self conscious and so more able to fit happily into society both socially and as a wage earner.

The design has been carried out using academic analysis and laboratory testing as a guide to the optimum structure and proof of design efficiency has been gained from extensive and carefully monitored prototype trials carried out in conjunction with the Salford Royal Hospital and the Department of Health.

The research has, therefore, shown that a cosmetic caliper is a practical and worthwhile proposition and while it is still early to claim

success, development has been encouraging, and we have now reached a stage where general hospital trials are to be carried out under Department of Health sponsorship to evaluate the final design for manufacture and supply purposes.

ACKNOWLEDGMENTS

I would like to acknowledge the valuable work of Mr. J. L. Mangan, F.R.C.S.(I)., the originator of the Salford caliper who, until his recent retirement, supervised the medical aspects of the design and trials. Also, I would like to thank Mr. P. Frank, F.R.C.S., who has succeeded Mr. Mangan; the Nursing Staff of the Salford Royal Hospital; the Workshop Staff of the University of Salford; and the Department of Health for their valuable help.

GENERAL REFERENCES

1. Bresler, B. and J. P. Frankel: The Forces and Moments in the Leg During Level Walking. A.S.M.E. Trans., 72:27–36, 1950.
2. Harper, F. C. et al.: The Forces Applied to the Floor by the Foot in Walking on a Level Surface. D.S.I.R. Building Research Station, National Building Studies, Research Paper 32, H.M.S.O. 1961.
3. Williams, Marian and H. P. Lissner: Biomechanics of Human Motion. Saunders, 1962.
4. Allen, C. E. L.: Biomechanics in Normal and Prosthetic Stance and Gait. South African Mechanical Engineer, 14 (2) :27–42, Sept. 1964.
5. Thomas D. P.: The Effect of Loading on Normal Standing in Man. J. of Anatomy, 93:75–96, 1959.
6. Whitney, R. J. and D. P. Thomas: Postural Movements During Normal Standing in Man. J. of Anatomy, 93:524–539, 1959.
7. Moffat, C. A., E. T. Haslam, and E. H. Harris: An Experimental Determination of Prosthetic Knee Moment for Normal Gait. A.S.M.E. Pub. No. 66 WA/BHF–8.
8. Marks, M. and G. C. Hirschberg: Analysis of the Hemiplegic Gait. Annals of the New York Academy of Sciences, 74 Art. 1, Sept. 1958.