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The clinical assessment of the normal and abnormal foot during locomotion

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Abstract

The clinical assessment of the weight-bearing foot during locomotion is normally based on subjective judgement rather than on quantitative measurement. Although anatomical abnormalities are often apparent at examination, the accurate assessment of an abnormality of function is more difficult to assess particularly if the abnormality is only apparent under dynamic loading conditions.

The many drawbacks in previous methods proposed for the clinical assessment of gait have led to the development of a novel system which allows an immediate quantitative visualization of the magnitude and point of application of the forces applied to the plantar surface of the foot during locomotion. This paper describes the technique and presents visual data on normal locomotion, on abnormal locomotion and the changes induced into a patient's abnormal gait by corrective surgery.

Introduction

Attempts to quantify the performance of the walking foot stretch back as far as the nineteenth century. An early researcher was Beely (1882) who attempted to relate interface pressures with the depth of a foot impression in a thin bag filled with plaster of Paris. Later researchers such as Elftman (1934) and Morton (1935) used the deformability of rubber projections on a walkpath mat to measure localized loads. A similar method has been proposed recently, although here an optical technique was used to display the localized loads as a set of circular interface fringes (Arcan and Brull, 1976).

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All of these methods allow an easily assimilated visualization of the loading pattern under the standing foot to be displayed, but such a pattern is critically dependent upon body posture and is of minimal value in assessing the performance of the walking foot. Dynamic data can be recorded, but the rapidly changing loading patterns across the plantar surface of the walking foot makes data recording and interpretation both difficult and time consuming.

Other quantitative studies have attempted to simplify data reduction by attaching discrete pressure transducers to selected anatomical sites (Schwartz and Heath, 1949; Bauman and Brand, 1963) or inserting them into special shoes (Holden and Muncey, 1953). However, transducer fixation and repetitive transducer positioning both cause problems, the transducers are easily damaged and the method is not suitable for routine use.

Currently the most successful methods for assessing locomotor biomechanics are based on the floor mounted force plate which, when used in conjunction with simultaneously recorded visual data, allows instantaneous forces through the segment of interest to be computed. Unfortunately, the traditional force plate and camera system is unsuitable for assessing the biomechanics of the foot because of the gross data recorded by the plate and also because of the small and subtle movements of the foot during the stance phase. One advance on the normal force plate, as far as the foot is concerned, was described by Stott *et al.* (1973). These authors constructed a force plate from twelve parallel beams and were thus able to record the loading on different segments of the foot during the stance phase. The position of the foot on the plate, the image of the foot/plate contact area and the recorded force distribution all have to be related to each other after a subject has walked across the plate which is again a time consuming procedure. Thus data processing is slow and the system is not suitable for use within the confines of the normal clinical routine.

The instrumentation system used to record the data presented in this paper was designed specifically to enable the performance of the walking foot to be assessed without resorting to elaborate and time consuming data reduction procedures (Manley and Solomon, 1978). The system measures both the magnitude and distribution of the vertical force component acting on the plantar surface of the foot during the stance phase of gait and presents this quantitative clinical data as an easily assimilated visual display.

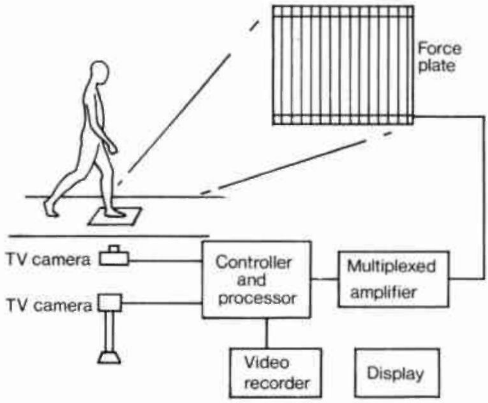

Fig. 1. Schematic representation of the force plate and its signal processing system.

Instrumentation

The instrumentation system is shown schematically in Figure 1. The force plate is made up of 16 transparent beams mounted in a walkway transverse to the direction of walking. Each beam is 260mm long and 20mm wide with a gap of 1mm between adjacent beams. Both ends of each beam are mounted on strain gauged cantilevers which act as load cells to measure the vertical reaction forces at the beam ends. The 32 strain gauge load cells are driven by a multiplexed amplifier. On any one beam the sum of the two vertical reaction forces measured by the load cells at the beam ends gives a measure of the total vertical force applied to that beam, while a ratio calculation from these same two reaction forces gives the position of the "centre of pressure" of the applied load. Both of these functions are performed by the central controller and processor.

Two television cameras are used in conjunction with the force plate system to provide the necessary simultaneous visual data on both the swinging and the planted foot. When the foot contacts the force plate its plantar surface is photographed by the camera beneath the plate while its medial or lateral aspect (and the position of the swinging leg) is photographed by the camera mounted adjacent to the walkway. The central controller and processor generate a composite video picture on a T.V. monitor which shows (i) a lateral view of the planted foot and swinging leg, (ii) a view of the plantar surface of the weight-bearing foot which also clearly shows the boundaries between the sixteen beams running transversely across the force plate with

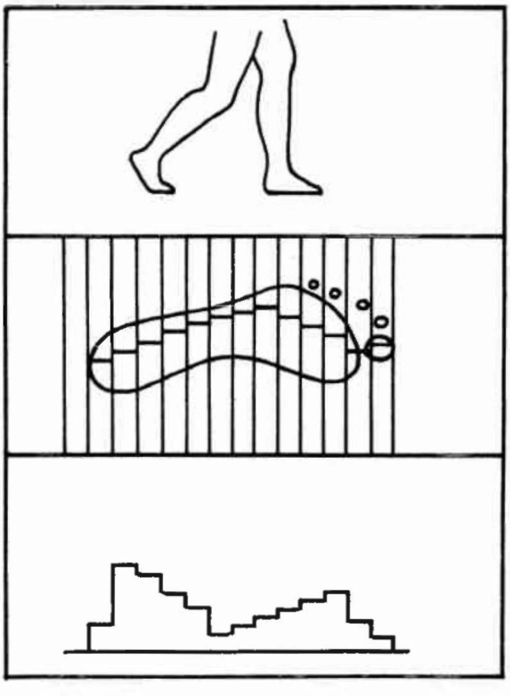

Fig. 2. Schematic representation of one frame from the video output during the stance phase.

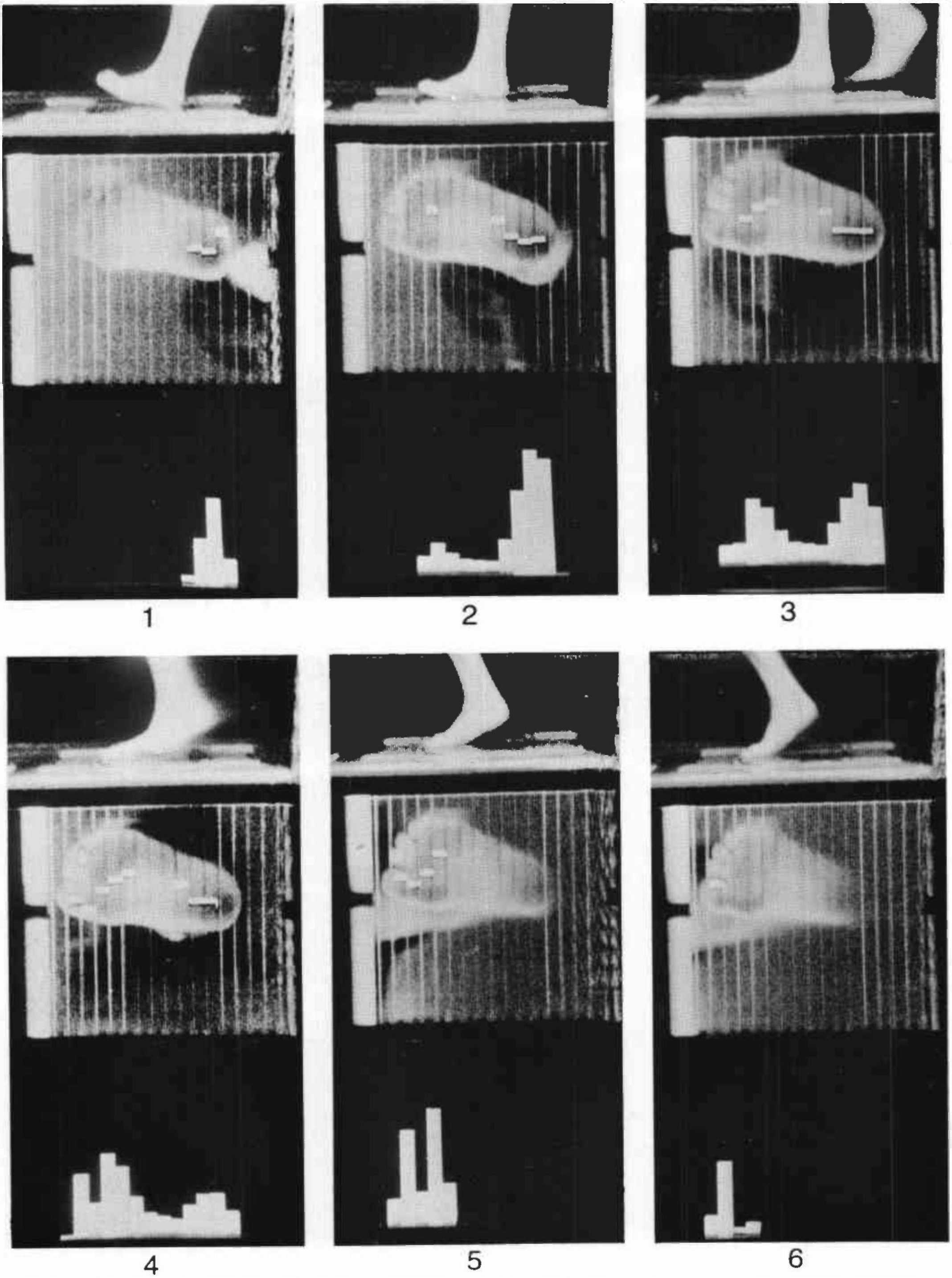

Fig. 3. The force distribution across the plantar surface of the right foot of a normal 78 kg subject during selected instants in the stance phase.

(1) Heel strike. (2) Approaching foot flat. (3) Foot flat.
(4) Mid-stance. (5) Heel rise. (6) Approaching toe off.

Full scale deflection on the histogram (from histogram base line to lower edge of the force plate image) represents 115 Newtons.

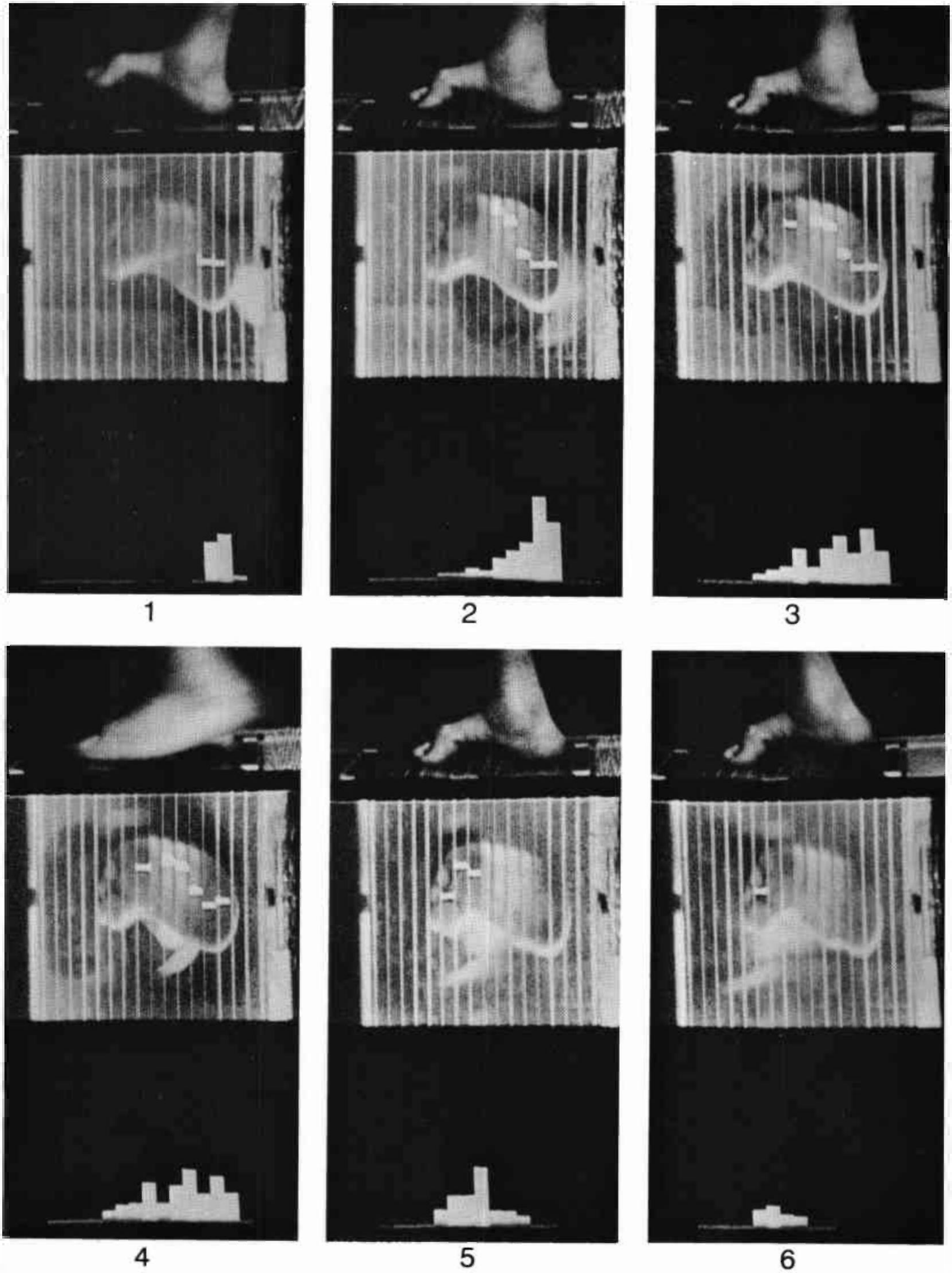

Fig. 4. The force distribution across the plantar surface of a post-poliomyelitis foot with fixed claw toes and restricted plantar flexion at the ankle joint.

(1) Heel strike. (2) Approaching foot flat. (3) Foot flat.
(4) Mid-stance. (5) Heel rise. (6) Toe off.

Patient's body weight 58.3 kg. Full scale deflection on the histogram: 95 N.

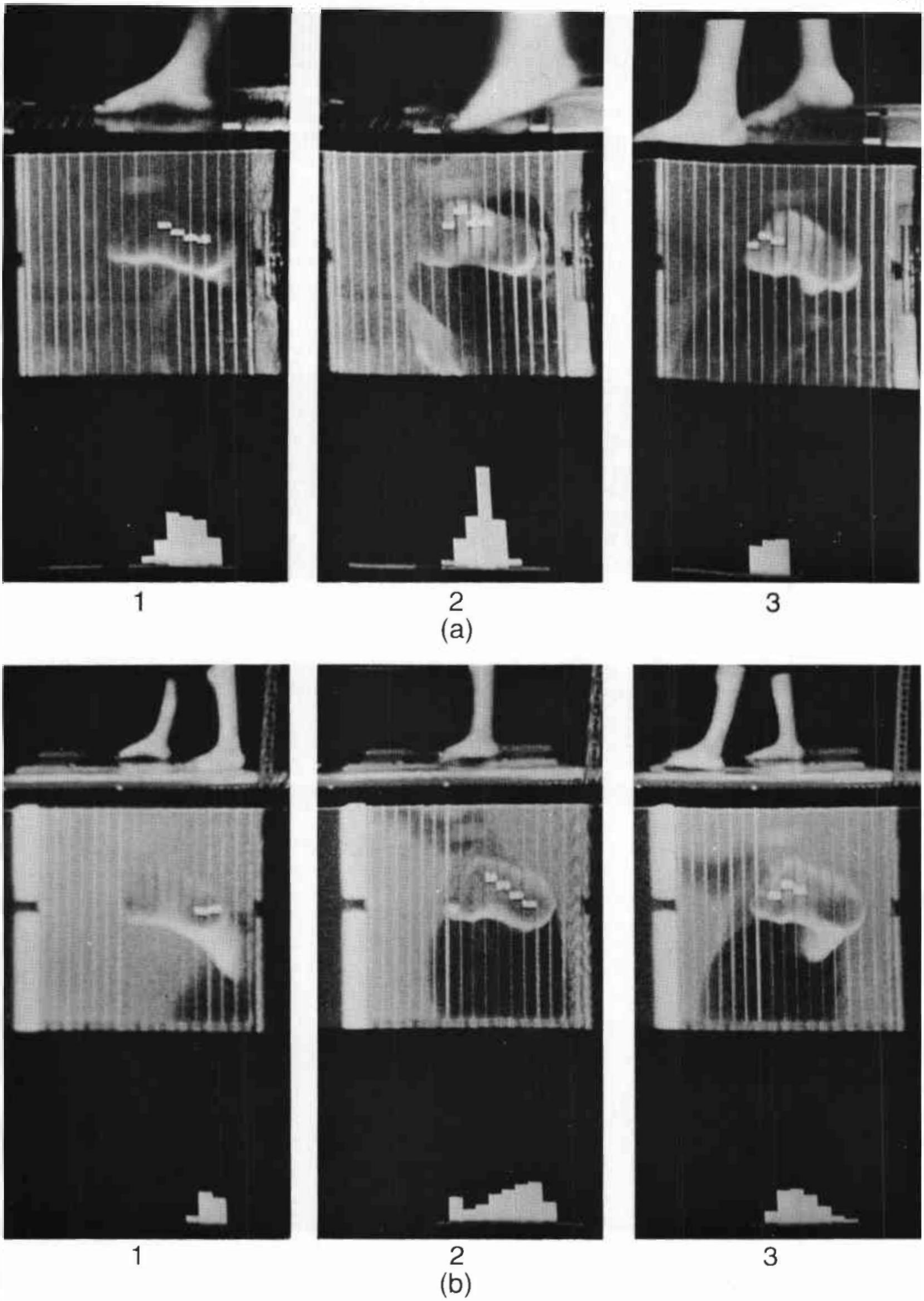

Fig. 5. The force distribution across the plantar surface of an arthrogrypotic foot with arthrogrypotic ankle before and after surgery. (a) Before surgery: (1) Foot strike, (2) Mid-stance, (3) Approaching toe off; (b) Eight weeks after bilateral antero-lateral tibial osteotomies: (1) Heel strike, (2) Mid-stance, (3) Approaching toe off. Patient's body weight 19.8 kg. Full scale deflection on the histogram: 70 N.

the centre of pressure lines superimposed, (iii) a histogram type display of the load carried by each beam.

A video picture, with superimposed histogram and centre of pressure lines is generated every 20 milliseconds (Fig. 2). The multiplexed strain gauge amplifier scans its 32 load cells for the 7 milliseconds that the controller needs to construct the histogram and then switches off for 13 milliseconds until the next histogram commences. Accuracy of the histogram is within 5% at full scale deflection while the natural frequency of the unloaded beams is in excess of 30 Hertz. The histogram can be calibrated with either a known mass or with an electronically generated output signal which corresponds to 50 Newtons. Full scale deflection on the histogram is from the histogram base line to the lower edge of the segmented force plate image in the middle third of the display. Histogram gain is infinitely variable from 50 to 200 N for full scale deflection. Histogram gain does not effect the "centre of pressure" calculation although for calculation stability a centre of pressure line on any one beam is not drawn unless the total load on that beam exceeds 5 Newtons.

When a load is applied to a beam the magnitude of the load is reflected by the height of the relevant bar in the histogram. The centre of pressure of the load on the beam is superimposed as an electronically generated white line upon the image of the beam and thus upon the image of the plantar surface. A lateral shift of load is therefore reflected as a lateral shift in the centre of pressure line relative to the foot, while a longitudinal shift of load (i.e. from hindfoot to forefoot) will be shown as a change in shape of the histogram display. It is thus possible to see at a glance not only the magnitude of the load carried by different sections of the foot, but also the pattern of loading applied to the plantar surface.

Results

In the gait assessment clinic, the records obtained from the segmented force plate system are recorded on to videotape for analysis in real time, in slow motion or frame by frame. The stance phase of gait from heel strike to toe off in most subjects occupies about 35 video frames.

A series of still photographs taken from the video display and illustrating the stance phase of the right foot of an adult, normal subject is shown in Figure 3. The instants in the stance phase depicted here are heel strike, the approach to foot flat, foot flat, mid-stance, heel rise, and the approach to toe off. The upper third of each photograph shows the medial aspect of the planted foot and the lateral aspect of the swinging foot, the middle third shows the plantar surface of the planted foot photographed through the force plate beams with centre of pressure lines superimposed, and the histogram in the lower third displays the total load carried on each segment of the foot.

The series of photographs clearly show that, as predicted by traditional gait studies, the heel is subjected to an impact force of considerable magnitude at heel strike. At the instant depicted the largest force on one beam is 50 N, although the videotape shows that the maximum loading of the segment is 78 N and occurs 40 milliseconds later. At heel strike the ankle joint is in a few degrees of plantar flexion, and the position of the centre of pressure lines shows that the calcaneus is also slightly everted. As the forefoot descends towards the plate and the lower leg moves towards the vertical, the heel still carries the majority of the applied load, but as the foot supinates, increased loading is applied to the mid- and forefoot, somewhat laterally to the longitudinal midline. The ends of the longitudinal arch remain the major weight bearing areas from foot flat to mid-stance, with approximately balanced forefoot loading at mid-stance. After heel rise the ankle joint moves back into plantar flexion and the loading moves medially across the metatarsal heads, along the line of the metatarsal break, towards the big toe. The fifth picture shows the big toe and second metatarsal head carrying the majority of the impulsive load generated by the lower limb, while the approach to toe off shows further medial movement of the weight bearing area with rapid decrease in applied loading, until the final contact occurs between the plate and the big toe.

By comparison, a series of photographs taken at similar instants during the stance phase of the right foot of an adult patient with claw toes and post-poliomyelitis deformity of the foot is shown in Figure 4. The heel strike photograph clearly shows both the medial view and also the plantar aspect of the fixed foot deformity. Initial heel loading is on the mid-line of the calcaneus while the maximum loading of any segment of the foot during the complete stance phase (40 N) is shown in the second photograph. As the cycle moves towards mid-stance, the foot is loaded along the whole of the lateral border; the longitudinal arch, evident in the normal foot sequence, not being present in this grossly abnormal foot. The hindfoot remains in contact with the plate until well after mid-stance, and it can be seen that plantar flexion of the ankle is minimal throughout the stance phase. It is also noticeable that the big toe, despite its fixed claw position, never comes into contact with the plate; forefoot loading being carried by the second and fifth metatarsals until "toe-off" occurs.

The final series of photographs (Figure 5) shows the right foot of a ten year old patient with arthrogrypotic feet and ankles. Figure 5a shows three instants in the stance phase of the patient's gait before treatment of the abnormality and Figure 5b shows the same patient after anterolateral tibial osteotomy.

The first photograph in the sequence taken before corrective surgery depicts the instant of contact between the foot and the force plate. Thus, at "heel strike" contact is made between the lateral border of the foot and the plate, with three segments of the plate carrying similar loading (17 to 20 N). At mid-stance, loading of the plantar surface is entirely on the forefoot with maximum loading in the area of the fifth metatarsal head. As toe off approaches, the loading tracks across the metatarsal break towards the big toe.

By comparison, the three photographs taken from the sequence recorded after tibial osteotomy show a dramatic change in the hindfoot loading applied at heel strike. Initial plantar loading is now in the centre of the heel pad, and as the limb moves towards mid-stance, the loading tracks towards the lateral border of the foot. The mid-stance photograph shows that body weight is well distributed along the foot although there is little sign of a longitudinal arch. As toe off approaches the foot is loaded on the second and third metatarsal heads instead of the fifth metatarsal head as shown in the sequence recorded prior to treatment.

Comparison of the two sequences shows that the patient's plantar loading profiles were certainly improved by surgery. Although the patient now tends to walk slightly flat footed and the foot is still inflexible, the prolonged loading of the forefoot, so evident before surgery, has now been relieved. Additionally the wedge osteotomy has successfully everted the foot, moved plantar loading towards the midline and relieved the loading of the extreme lateral border.

Conclusions

The series of photographs presented, which were taken from videotape records, clearly show the differences between normal and abnormal foot function during locomotion. The sequence depicting the normal foot shows that the movement of the foot and the way in which load is transferred between the foot and ground is not comparable with the tripod-like behaviour that the traditional view of foot function would have us believe. By comparison, the records of the poliomyelitis foot and the arthogrypotic foot clearly show the deficiencies in foot function and allow the quantitative measurement of localized loading as well as visualization of the shifting load patterns throughout the stance phase. Such readily assimulated data can only lead, in the long term, to the prescription of definitive treatment for foot disorders. The short post-operative sequence also shows the ease with which corrective treatment can be assessed.

It is believed that this technique of displaying the foot and its applied loads will facilitate a fuller understanding of this hitherto ignored component of the locomotor system, and will give greater insight into correcting and preventing injuries and damage to the lower limb which result from biomechanical defects in foot function.

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