EXTERNAL CONTROL OF LOWER EXTREMITIES

DYNAMIC GAIT STUDIES APPLIED TO THE DESIGN OF LOWER LIMB PROSTHESES

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The Bioengineering Unit at the University of Strathclyde was established in 1963 to carry out certain basic and applied research projects, one of which involved the estimation of the force transmitted between the femoral head and the acetabulum during normal activity. This information was required in connection with the design of internal hip prostheses. Accordingly an experimental procedure was devised to enable all the force actions involved to be measured or calculated.

The first stage in the investigation requires the calculation of the force actions transmitted to the trunk by the leg. These actions are fully defined by three components of force and three components of moment as shown in Figure 1.

The forces producing this action are external forces (ground to foot), gravity forces, and inertia forces. A walk-path was set up so that these three force actions might be simultaneously evaluated during normal level walking. The variations with time of the ground-to-foot force action were measured by a force plate, of the form shown in Figure 2, which was set in the path of progression. This is a platform supported on four strain-gauged pillars by which it is possible to measure all the components which fully define the force action between the foot and the ground. The gravity forces were estimated using Fischer's coefficients which give approximate weights of limb segments in terms of the total weight of the subject. To enable the inertia forces to be calculated it is necessary to obtain the displacement/time relationship of the leg segments during walking. This was done by filming the subject by means of two synchronized cameras filming at 50 frames/sec., one in the plane of progression and one at right angles to the plane of progression (i.e., from in front and from one side). The films were then re-exposed with grids composed of 5 in. squares situated at the reference axes, at the centre of the force plate. It was then possible to obtain the necessary displacement information by measuring the positions of skin markers situated as required on the bony prominences.

The necessary measurements were taken from each exposure of the film, i.e., at intervals of .02 seconds, measuring from the reference axes and by use of the superimposed grid which minimized optical distortion. A further correction was made to allow for the difference between the true position of any point and its position as viewed by the camera (Fig. 3). The displacement/time relationship so obtained was then double differentiated by a numerical method to give the segment accelerations and hence, again using Fischer's coefficients, the inertia forces.

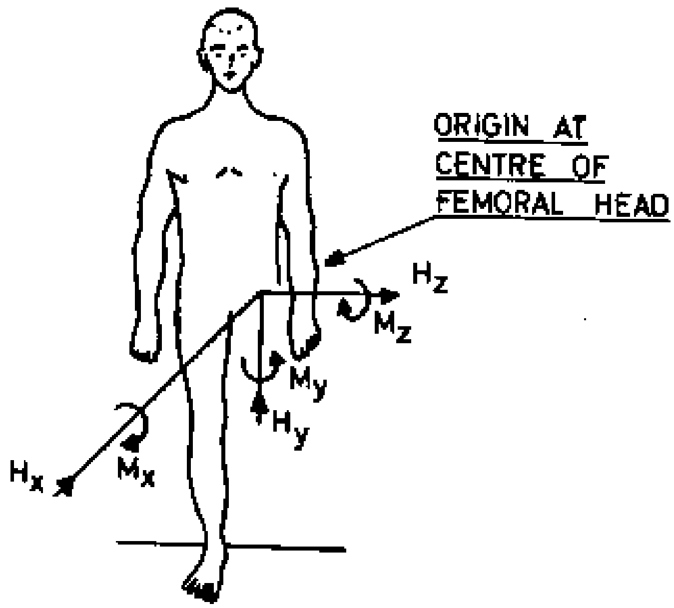

Figure 1. Force actions transmitted to trunk at a section through the left hip

The force transmitted between the femoral head and the acetabulum is fully represented by three components in mutually perpendicular planes as shown in Figure 4. This force is due to the above force and moment actions of the leg on the trunk. The moments are transmitted across the joint by tension developed in the appropriate muscle group and this muscle force is balanced by an equal and opposite compressive force at the joint. Having obtained the leg/trunk action it is then necessary to estimate the muscle action. There are 22 muscles acting across the hip joint, and to enable the calculation of the joint force several assumptions have to be made:

The muscle force is assumed to act along a line joining the centre of areas of the origin and insertion of the muscle.

The hip rotators are ignored in the calculation as their position relative to the hip joint and their size minimize the force effect they could produce at the joint.

The muscles are grouped according to their action at the hip. This is permissible as the muscles in a group have been shown to be generally synchronous in their action by Marks and Hirschberg[1]. The muscle action may then be represented by the action of

Long Extensors
Gluteus Maximus
Abductors
Adductors
Long Flexors
Illiopsoas

The equations of equilibrium at the hip, however, only allow the calculation to involve three unknown muscle forces and so to reduce the variables electromyographic recordings are also taken during walking thus eliminating from the calculation those muscles which display no significant electrical activity.

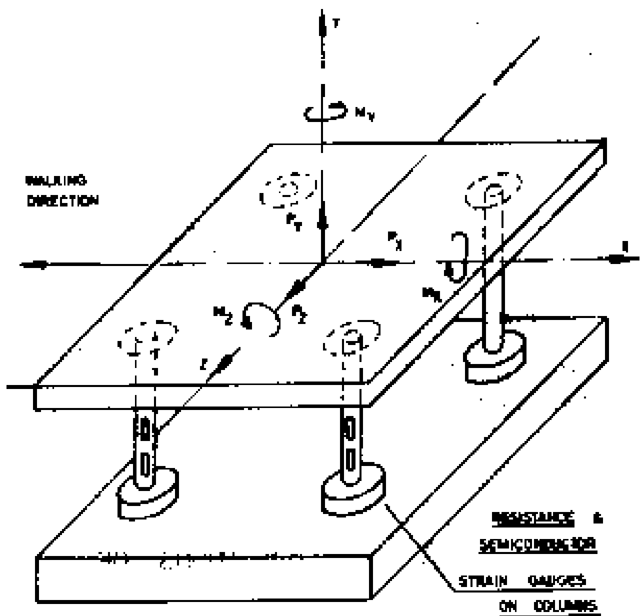

Figure 2. Force plate. x, y, z — coordinate system axes, P_x, P_y, P_z, M_x, M_y, M_z — measured force actions

All the above information is recorded, translated to punch tape, and the subsequent calculations handled by computer.

This procedure is decribed in detail by Paul.[2,3]

The information which is obtained then constitutes:

a) A complete displacement/time record of the subjects' mode of progression.

b) The force and moment actions transmitted to the trunk by the leg.

c) The force transmitted by the femoral head to the acetabulum.

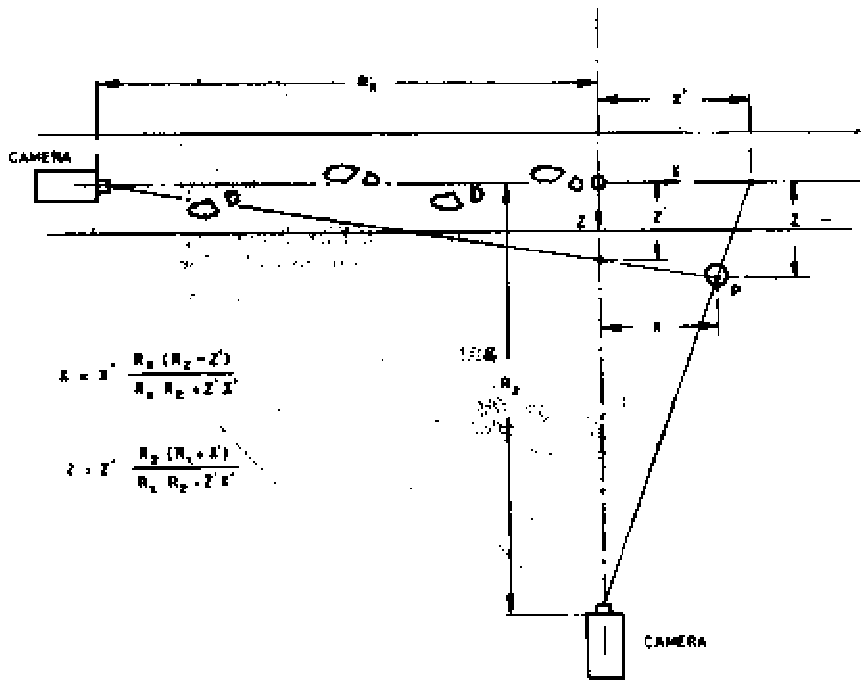

Figure 3. Walk-path viewed from above

There is then the facility to study and to specify the forces and motions involved in normal or pathological gait. The studies so far undertaken have been concerned with level walking by apparently

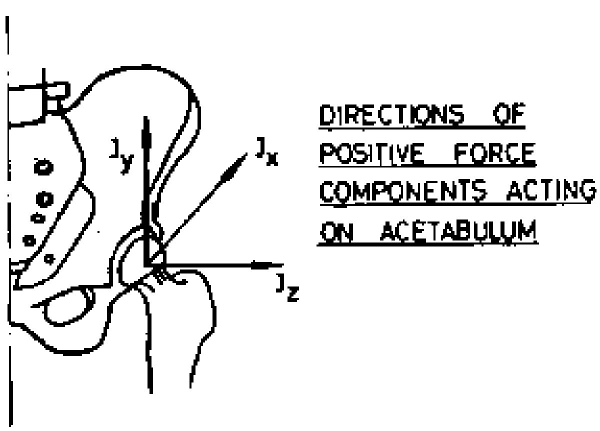

Figure 4. Components of force transmitted between femoral head and acetabulum

normal young adults. This investigation is now being extended to the study of amputee gait. A typical analysis of the gait of an above-knee amputee wearing a conventional suction-socket limb fitted with a friction-brake knee mechanism follows. In the results presented the typical »normal« data for a subject of comparable build and body weight are also shown.

Figure 5 shows the vertical component of the ground-to-foot force. The two major differences from the normal individual's record are, firstly, that the normal leg of the amputee remains on the ground longer than the prosthesis, and secondly, the pattern of force corresponds to a less vigorous mode of walking — the accelerations and decelerations of the body in a vertical direction being less marked.

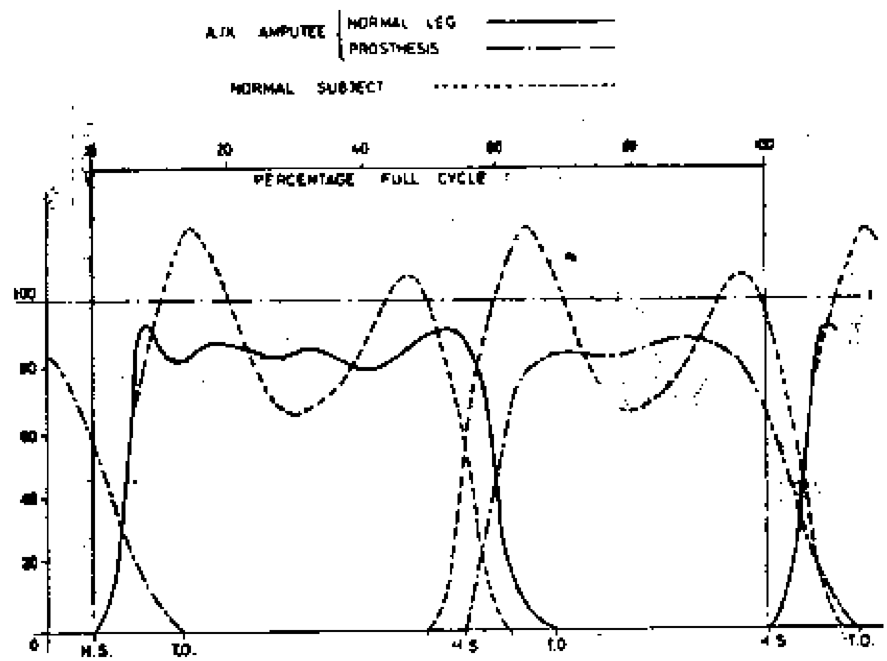

Figure 5. Ground to foot force (vertical component)

Figure 6 shows the leg/trunk moment corresponding to action of the abductor/adductor muscle groups. The general picture of moment is similar for the normal individual and the amputee's normal leg and somewhat less so for the prosthetic side. The difference in the moment on the prosthetic side is largely explained by the greater amount of body sideways which is present.

The leg/trunk moment illustrated in Figure 7 corresponding to flexion-extension muscle action shows a large difference in general pattern existing at both hips of the amputee. The difference at the normal leg is due to the amputee rising quickly onto his toe after heel strike, presumably to propel his prosthesis forward and provide ground clearance. This moves his point of support forward relative to his hip and retains the moment requiring extensor muscle action for a longer extent of the stance phase than is normal. The moment on the prosthetic side is again abnormal because the stride is shorter and the hip is quickly passed beyond the point of support. Also the peak values are very much lower, the large peak after heel strike being almost absent on the prosthetic side.

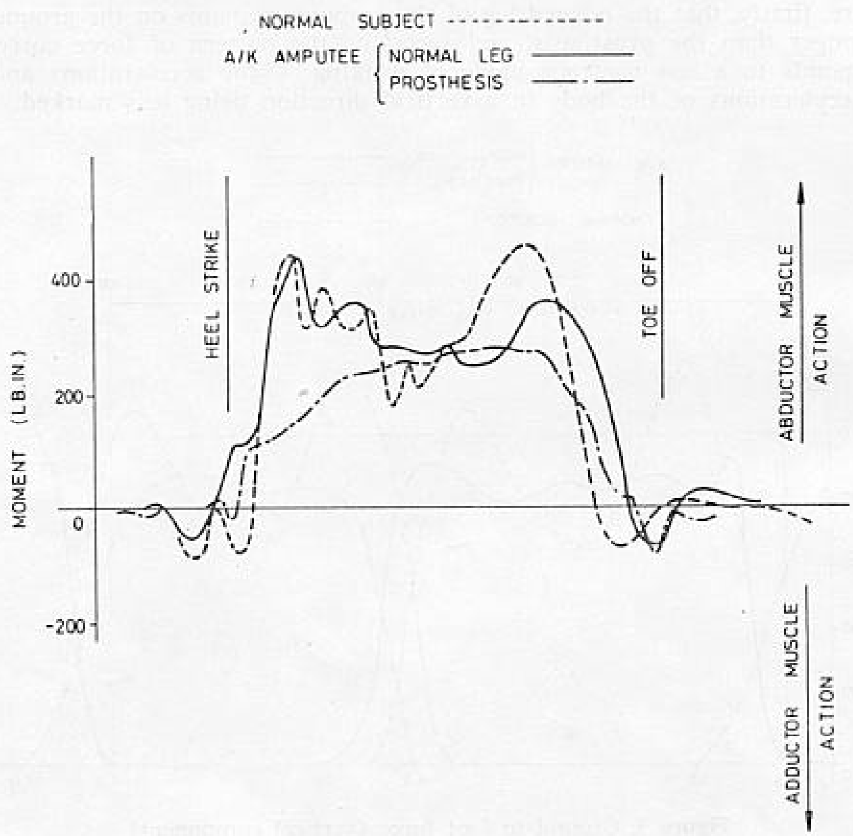

Figure 6. Leg/trunk moments (abduction — adduction)

Figure 8 shows the estimated resultant hip-joint force. During the stance phase this force is largely due to body weight and flexor-extensor and abductor-adductor muscle action. The force on the hip joint on the normal side is of similar form to the normal individuals but having a lower peak value after heel strike and no peak before toe-off. On the prosthetic side the force is much lower than normal and has no characteristic double peak.

The general picture then is one of force actions which are abnormal and are not symmetrical in that the joints have dissimilar force actions imposed upon them.

The amputee whose analysis has been considered had what would be described as a »good« walk. That is, using his conventional above-knee prosthesis he was able to walk in such a manner that his gait appeared to the observer to be fairly close to that of a normal individual. This, of course, is the model used during walking training. It has been shown that the resulting force actions are neither normal nor symmetrical as regards external or internal actions. The reason, of course, is that the missing limb has been replaced by a prosthesis which has an articulated skeletal layout but no musculature and completely different mass properties. *Thus symmetrical movement would necessarily imply unsymmetrical force actions* — kinematic »symmetry« — kinematic »dissymmetry«.

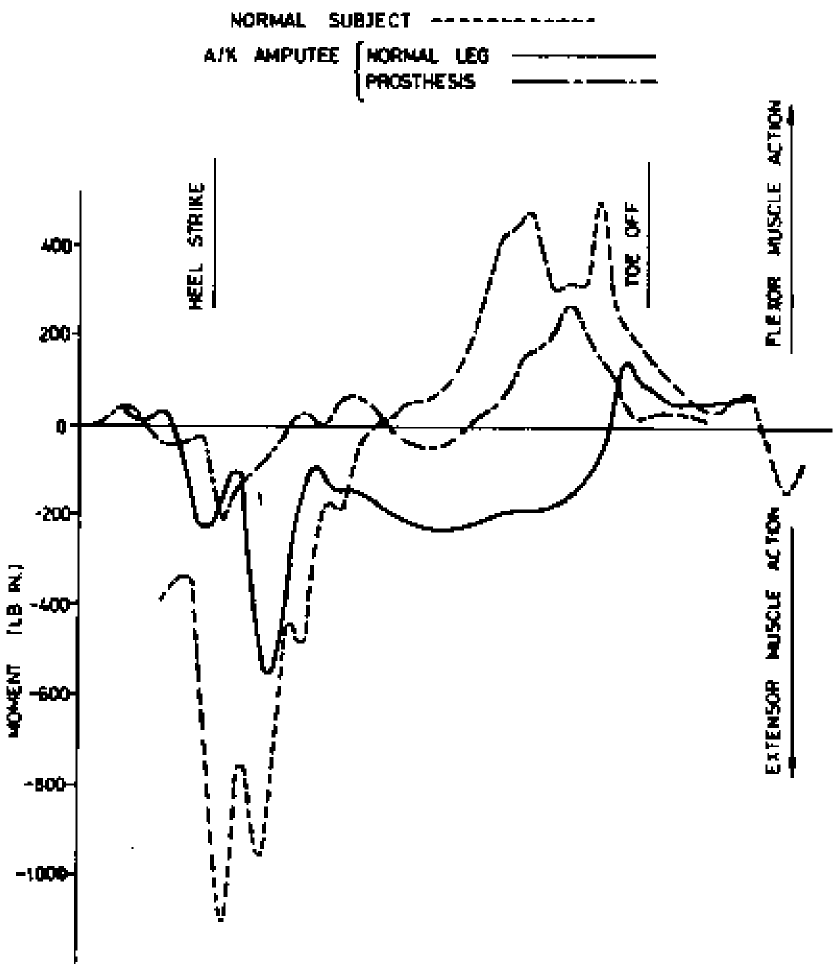

Figure 7. Leg/trunk moments (flexion — extension)

This abnormality of joint force would seem to be a most undesirable feature when considered in conjunction with clinical evidence presented by Solonen, Rinne, Vikeri and Karvinen[4] who examined 65 above-knee and 157 below-knee amputees from the 1939—45 war. They used a control group of 92 individuals of the same age group who served in the armed forces during the war but were otherwise randomly selected. They showed that in the AK amputees arthrosis of the hip joint was more common in the intact lower limb than in the amputated limb and in both significantly more common than in the control group. In the BK-amputee group arthrosis of the ankle joint in the intact leg was again statistically more common to a significant degree than in the control group as also was arthrosis of the joints of the foot.

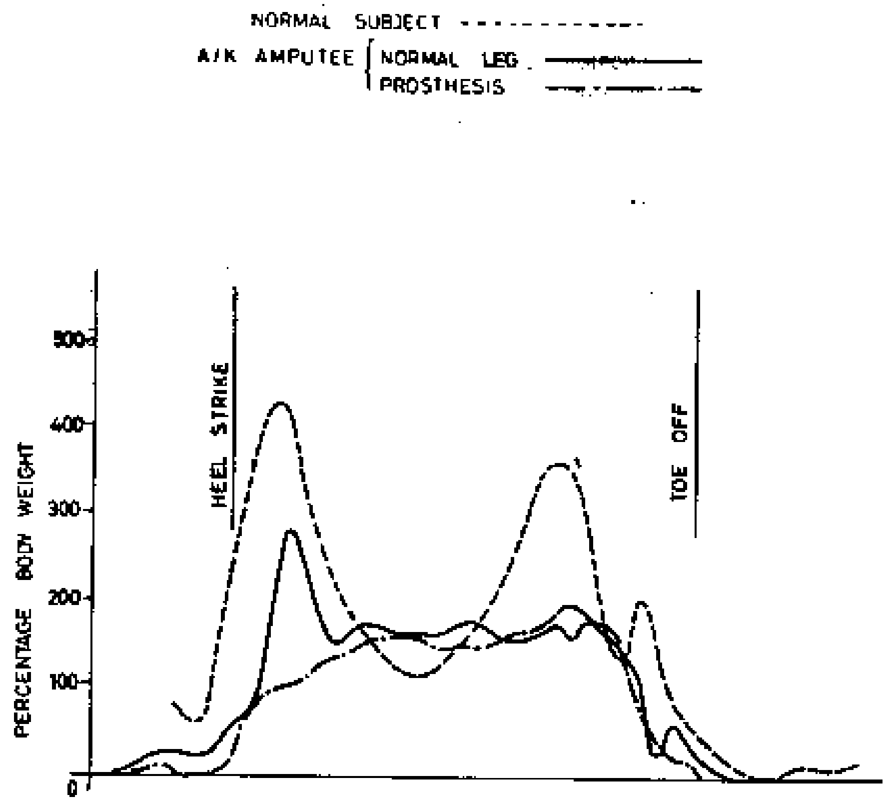

Figure 8. Resultant hip joint force

It is questionable on these findings whether, in fact, the amputee with a conventional prosthesis should be trained to walk in a symmetrical fashion as the long-term effects would appear to be detrimental. It is an observed fact that many amputees who are able to reproduce a gait pattern similar to the normal individuals reserve this gait for the physiotherapist or limb-fitting centre and lapse into a completely different gait during normal unobserved activity—possibly for reasons of comfort, coupled with an adaptation to produce a minimum energy-expenditure condition. This would seem to support the view that the best gait pattern for the amputee with the conventional prosthesis may very well not be that of the normal individual.

A more far-reaching question, however, concerns the whole basis of lower limb prosthesis design. The first requirement of the conventional prosthesis is that in the static condition it bears a close resemblance to the normal limb. Locomotion is achieved by a process of adaptation of remaining musculature. The result is often a gait pattern which is by no means »cosmetic« and, as has been demonstrated, produces joint and muscle forces which are abnormal. A further important point relates to the elderly amputee who accounts for a very high percentage of the amputee population and for whom amputation is only one sympton of a general health picture. Here the inherent instability of the prosthesis and the difficulty of adapting remaining musculature renders the conventional prosthesis with its cosmetic basis of design totally inadequate; a view borne out by Hansson's findings in a clinical follow-up study of leg amputees at Gothenburg[4] in 1964. Hansson described the functional attainment of elderly amputees as discouraging and ascribed this to the fact that »the types of prosthesis available today are unsuitable for the aged«.

It is suggested that a more rational approach to the design of lower limb prostheses would be to base this on functional requirements and subsequently supply the maximum cosmesis attainable. The primary functional requirements are that prosthesis should supply to the trunk the *same motion and the same force and moment conditions* as were imparted by the normal limb before amputation. There are other conditions which it might also be thought desirable to incorporate into the design, again on a functional basis. The stump is not the most suitable area to carry the transmitted loads and it might be possible to support at the hip region or, in other words, provide a »seat« with a greater area of support and tissue well able to sustain the lower pressures which would obtain. Further to this it might be possible to allow the stump to move in a similar manner to the pre-amputation movement even if this meant movement relative to the prosthesis. Another consideration might well be the need to provide greater inherent stability, particularly required by the elderly amputee.

It is quite conceivable then that a design evolved on this sort of basis would bear little or no resemblance to the normal limb. Although this would mean poor cosmesis in a static sense, it should mean that the gait pattern would be much closer to the normal individual's, implying maximum cosmesis in a dynamic sense and providing normal joint conditions.

If such a design of prosthesis proved practicable, there is no reason why the amputee should not have a wardrobe of different limbs the conventional cosmetic limb for dress occasions, theatre, dinner and the like, and a functional cosmesis for work and more arduous occasions. No doubt the utilisation factor would be dependent on the amputee's occupation, age group, social position, etc. The youthful amputee would probably use the conventional limb to a greater extent accepting the higher energy cost while the elderly amputee might only use his cosmetic limb on the odd occasion.

It may be that this whole concept appears fanciful at first sight but it is no more so than supplying an articulated, skeletal limb with no musculature and expecting functional effectiveness.

Thus it is submitted that before proceeding on the design of control systems for lower limb prostheses, two questions require to be answered urgently:

1) Should control systems for conventional prostheses be oriented to reproducing the gait pattern for a normal individual?

2) Should the whole basis of the design of lower limb prostheses not be given a critical re-appraisal?

References

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