J. Biomechanics Vol. 19, No. 5, pp. 369–373, 1986.
Printed in Great Britain

0021-9290/86 $3.00 + .00
Pergamon Journals Ltd

THE PASSIVE ELASTIC MOMENT AT THE KNEE AND ITS INFLUENCE ON HUMAN GAIT

J. M. MANSOUR and M. L. AUDU

Department of Mechanical and Aerospace Engineering, Case Western Reserve University, Cleveland, OH 44106, U.S.A.

Abstract—The elastic component of the passive moment at the knee was measured *in situ*. The force needed to manually range the knee from approximately 90° of flexion to full extension was measured. Hip and ankle angle were held fixed. The passive knee moment, computed from the force and knee angle data, was compared to the total knee moment required for normal gait. This comparison suggested that the passive moment can contribute a significant portion of the total joint moment during some phases of the gait cycle.

INTRODUCTION

Passive joint moment is being defined as the moment acting at a joint when all muscles crossing the joint are relaxed. The passive joint moment arises from the deformation of all tissues which surround the joint including skin, ligaments, tendons, relaxed muscles etc. These tissues exhibit a complex mechanical behavior (Wright and Johns, 1961). Only the elastic part of this response was investigated.

In a previous study we showed that the passive hip moment was a potentially significant part of the total hip moment at toe off and heel strike in normal gait (Yoon and Mansour, 1982). The passive hip moment was an exponential function of hip position. The magnitude of the passive hip moment was modulated by the knee position through two joint muscles: rectus femoris and the hamstring group.

The purpose of this study was to determine the passive knee moment, evaluate the effects of two joint muscles on this moment and evaluate its importance with respect to the total knee moment required for walking.

METHODS

The measurement technique employed was a modification of that used for the passive hip moment measurement (Yoon and Mansour, 1982). All tests were performed with the subject in the side-lying position (Fig. 1). The subjects were positioned in the apparatus with their back against a support (B). The thigh was held in position by vertical supports (V). The shank was strapped to a splint (S) and was supported distally by a load cell (L). The shank support was on a wheeled cart (C) which was manually rotated to produce knee flexion/extension. The rotational speed was approximately 0.20 rad s^{-1}. This speed was chosen to minimize the influence of the viscous response of joint tissues on our measurements. It is much slower than the relative shank-thigh speed in normal gait. The knee

Received April 1984; *in revised form October* 1985.

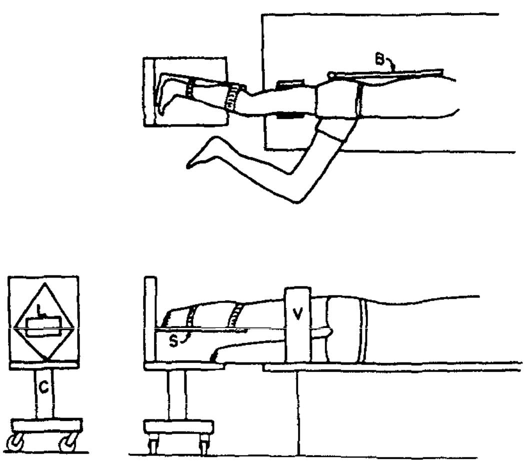

Fig. 1. Top, front and end view of the apparatus used to measure the passive elastic moment at the knee. An electrogoniometer, mounted laterally on the right leg was used to measure the knee angle during a test. This electrogoniometer is not shown in the diagram.

angle was measured by an electrogoniometer not shown in Fig. 1. The range of knee angles employed was from extension to approximately 90° of flexion. The extended position of the knee was determined by the most extended position at which the subject felt comfortable.

Since the knee is crossed by a series of two joint muscles (rectus femoris, hamstring group, gastrocnemius among others) it was necessary to control both the hip and ankle angle during a test. Hip position was measured at the superior aspect of the greater trochanter as the acute angle between the line from the anterior superior iliac spine to the greater trochanter and the extension of the line from the center of the knee to the greater trochanter. A reference value for the hip angle was measured while the subject was standing straight with their heels and back against a wall. This reference value was used as the zero for all hip angle measurements in the passive moment test. During a test the hip angle was held fixed by adjusting the back support. Two discrete values of the hip angle were employed, approximately zero degrees (the standing reference angle) and ten degrees of flexion (beyond the standing reference angle). In practice it was not always possible to position the subject's hip at 0°. In such instances a comfortable position near zero degrees was chosen. The second hip position was then obtained by flexing approximately 10° from the first position. The ankle angle was measured between the plantar surface of the foot and the perpendicular to a line through the center of the knee and the lateral maleolus (in the sagittal plane). During a test the ankle angle was held fixed by a brace on the shank support. Three discrete values of the ankle angle were employed, 10° of dorsiflexion, neutral and 10° of plantarflexion.

The test protocol for each subject consisted of four cycles of knee rotation (extension–flexion–extension) for each set of fixed hip and ankle angles. Thus, for each subject, a total of 24 cycles of data were recorded (two hip angles × three ankle angles × four cycles per hip and ankle angle).

Data from the load cell and goniometer were recorded on an x–y plotter. From this force vs angle data the knee moment, as a function of angle, was computed using the measured distance from the load cell to the tibial surface of the knee. For each set of four data cycles the average and standard deviation of the passive moment was computed. The sign convention for the moment was chosen such that a moment which caused knee extension was positive.

The measured passive moment was compared to the total (computed) knee moment required for normal gait. This comparison was made at selected times during the gait cycle using published hip, knee and ankle angle data (Murray, 1967) and total joint moment data (Cappozzo *et al.*, 1975). The gait cycle time corresponding to a prescribed hip angle (in the passive moment measurement) was determined from the data given by Murray (1967). The knee and ankle angles, at this time in the gait cycle, were then determined. The corresponding passive knee moment was then obtained from our experimental data. It was not always possible to determine the passive moment corresponding to a time in the gait cycle due to the ankle angle being out of the range in which our measurements were made. In general moments were compared in the range from 20 to 45% and 55 to 60% of the gait cycle where the gait cycle was measured from heel strike to the next heel strike of the same leg.

Four adult male subjects were tested (Table 1). The subjects had no previous history of orthopaedic problems with their leg.

RESULTS

The relationship of the passive knee moment to the knee angle was qualitatively similar to the moment–angle relationship found at the hip (Yoon and Mansour, 1982). The average and standard deviation of the moment data for four cycles, on subject CST, are shown in Fig. 2. Each cycle represents a displacement of the knee from extension to a flexed position and back to extension. The upper branch of the curve was recorded while rotating the knee from extension to flexion and the lower branch from flexion to extension. The fixed hip angle was 18° of flexion (with respect to the standing reference value) and the ankle was held fixed at 10° of dorsiflexion. The force

Table 1. Age, height and weight for the four male subjects tested

Subject	Age (yr)	Height (m)	Weight (N)
CST	24	1.81	543
MSJ	27	1.65	574
MA	30	1.70	592
MT	23	1.72	681

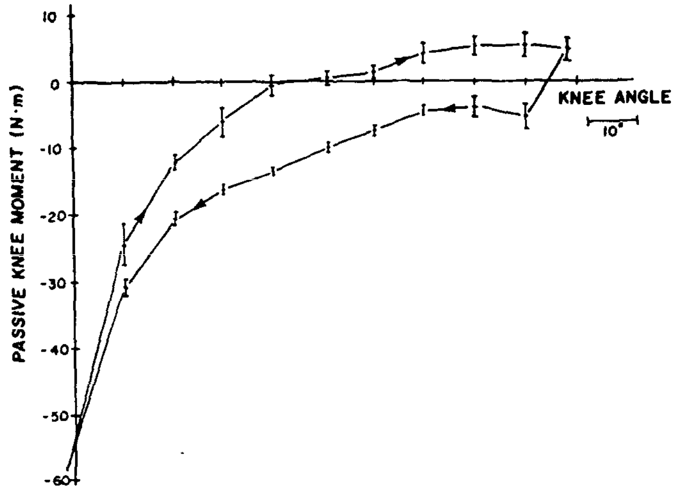

Fig. 2. Average and standard deviation of the measured passive knee moment for subject CST. The hip angle was 18° and the foot was dorsiflexed 10°.

and angle data were recorded continuously, but the average and standard deviation we computed at discrete knee positions.

The value of the fixed angle at both the hip and ankle affected the measured knee moment. Changing the ankle angle from ten degrees of dorsiflexion to ten degrees of plantarflexion changed the knee moment (at extension) from 55 Nm to 40 Nm respectively (Fig. 3). Changing the hip angle from twelve degrees to eighteen degrees of flexion decreased the knee moment through most of the range tested (Fig. 4). At knee extension, however, the measured moment was independent of the hip angle.

While the data for all subjects was qualitatively similar, there were quantitative differences. For example, subject MT was 'stiffer' than the others (Fig. 5). Comparing this data to that of CST (Figs 2 and 3) shows a greater change in moment over a smaller range of knee rotation.

When compared to the total moment required for normal gait, the passive moment was found to provide a significant contribution (Fig. 6 and Table 2). In

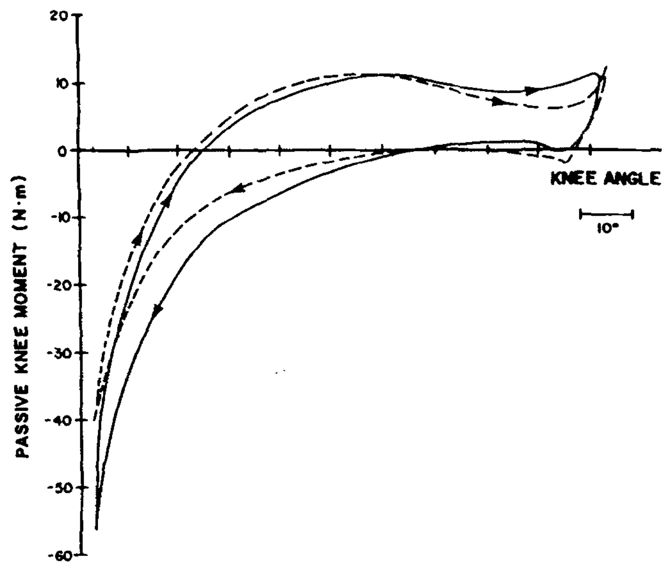

Fig. 3. Average passive knee moment for the foot in 10° of dorsiflexion (solid line) and 10° of plantarflexion (dashed line). The hip angle was 12°. Subject CST.

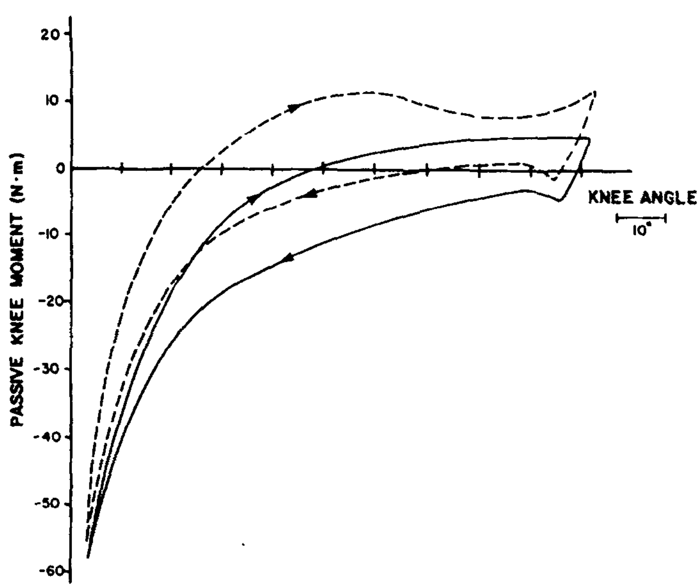

Fig. 4. Passive knee moment for the hip at 12° (dashed line) and 18° (solid line). The foot was in 10° of dorsiflexion. Subject CST.

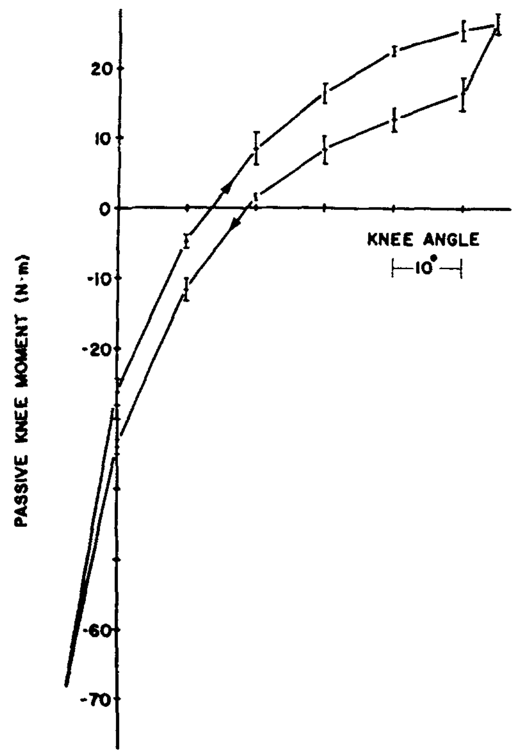

Fig. 5. Average passive knee moment data for the foot at 0° and the hip at −1° for subject MT.

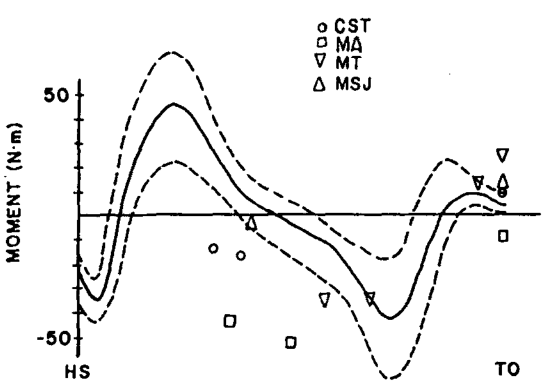

Fig. 6. Comparison of the total moment required for normal gait (continuous line), plus/minus two 'indeterminants' (dashed line), (from Cappozzo *et al.*, 1975) to the passive moment. The average passive moment as measured on four subjects is shown.

general the passive and total moments had the same sign. The passive moment had, however, a greater absolute value than the total moment.

Table 2. Average and 1 S.D. of the passive moment at specific intervals of the gait cycle. These average values are plotted in Fig. 6. The standard deviation was not computed for those cases where the passive moment was changing rapidly. In these cases the standard deviation was small, but could not be determined accurately from the chart recordings

Subject	Percent gait cycle	Passive moment (Nm)
CST	23	-17.3 ± 1.6
	19	-13.5 ± 1.1
	Toe off	9.9 ± 1.4
MA	22	-42.0
	30	-52.9
	Toe off	-9.0 ± 2.1
MSJ	25	-3.5 ± 0.9
	Toe off	15.2 ± 3.8
MT	35	-36.7
	42	-36.7
	57	13.6 ± 3.6
	Toe off	24.5 ± 1.6

DISCUSSION

The results have shown that the passive knee moment varies with knee angle in a qualitatively similar manner as the moment–angle relationships found at other joints (Hatze, 1975; Yoon and Mansour, 1982). For a given knee angle the passive knee moment was influenced by both hip and ankle angle. As the ankle changes position from plantarflexion to dorsiflexion the length of the gastrocnemius and its tendons is increased. This lengthening causes an increase in the passive force in the muscle–tendon unit with a concomitant increase in the moment that this muscle produces at the knee. The major effect of lengthening the gastrocnemius, on the passive knee moment, is in the region of knee extension. Increasing hip flexion shifted the passive knee moment to generally lower values throughout the range of knee angles studied. Flexing the hip causes the hamstring and its tendons to lengthen and the rectus femoris and its tendons to shorten. At knee extension the knee moment is independent of hip position. This suggests that the hamstrings are having little effect on the passive moment, i.e. the hamstrings are behaving as if they were slack for the range of joint angles investigated.

The passive moment generally exceeded the total moment required for locomotion. If this were true, antagonist muscles would have to be activated to obtain dynamic equilibrium with external forces. For example, in the range from 20 to 45 % of the gait cycle the rectus femoris or muscles of the vasti group would have to be active to diminish the effect of the large passive moment. The rectus is inactive in this range and the vasti group ceases its activity before 30 % of the cycle which is inconsistent with the observed relationship between the total and passive moments. Several factors may be responsible for this inconsistency. The data used for passive moment, total moment and joint angles were obtained from different sources rather than all being measured on a single subject. These data were related through a dimensionless time parameter; percent of gait cycle. Small differences between these sources could greatly change the nature of the results.

For example if the passive moment data for subject MT is shifted approximately 6% (to the right, Fig. 6) this data would lie within the range of the total joint moment. It is also possible that muscles were not relaxed during a test, yielding some active component to the measured 'passive' moment. However, our previous investigation at the hip suggests that active muscle force is not influencing our results. The hysteresis present in the experimentally obtained moment-angle relationship (Figs 2–5) introduces some error when relating the passive moment to the total moment. The passive moment at any knee angle depends on the history of the displacement, i.e. the passive knee moment is different when going from extension to flexion or from flexion to extension for a given knee angle. To properly assess the effect of hysteresis the knee would have to be rotated through the same sequence of angles as used during gait. Clearly, additional work is needed to verify the seemingly high contribution of the passive moment.

The fact that the passive moments were of the same magnitude as the moments required for gait suggests that the passive moments may be important when estimating muscle and joint forces. Most muscle force analyses, based on dynamic equilibrium or optimization of some function, have neglected the passive contributions to total joint moment (Morrison, 1970; Seireg and Arvikar, 1975; Crowninshield, 1978; Hardt, 1978 among others). Neglecting the passive moment could lead to an overestimation of muscle force and incorrectly predict the time when a muscle is active.

In summary, this study has shown that the passive moment at the knee is of the same magnitude as the total moment required for normal human locomotion. It appears that the passive moment may make a significant contribution to the total moment. Further study, where all data is obtained from a given subject, is needed to verify these results.

Acknowledgement—This work was supported by the NIH, grant number AM 32366.

REFERENCES

Cappozzo, A., Leo, T. and Pedotti, A. (1975) A general computing method for the analysis of human locomotion. *J. Biomechanics* **8**, 307–320.

Crowninshield, R. D. (1978) Use of optimization techniques to predict muscle forces. *J. biomech. Engng* **100**, 88–92.

Hardt, D. E. (1978) Determining muscle forces in the leg during normal human walking—an application and evaluation of optimization methods. *J. biomech. Engng* **100**, 72–78.

Hatze (1975) A control model of skeletal muscle and its application to a time-optimal bio-motion. Ph D thesis, university of South Africa.

Morrison, J. B. (1970) The mechanics of the knee joint in relation to normal walking. *J. Biomechanics* **3**, 51–61.

Murray, M. P. (1967) Gait as a total pattern of movement. *Am. J. Phys. Med.* **46**, 290–332.

Seireg, A. and Arvikar, R. J. (1975) The prediction of muscle load sharing and joint forces in the lower extremity in walking. *J. Biomechanics* **8**, 888–894.

Wright, V. and Johns, R. J. (1961) Quantitative and qualitative analysis of joint stiffness in normal subjects and in patients with connective tissue diseases. *Ann. Rheum. Dis.* **20**, 36–46.

Yoon, Y. S. and Mansour, J. M. (1982) The passive elastic moment at the hip. *J. Biomechanics* **15**, 905–910.