J. Biomechanics Vol. 15, No. 12, pp. 905-910, 1982.
Printed in Great Britain

0021-9290/82/120905-06 $03.00/0
Pergamon Press Ltd.

THE PASSIVE ELASTIC MOMENT AT THE HIP*

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Abstract—The passive elastic moment at the hip was measured in normal male subjects *in situ*. The influence of two joint muscles crossing the hip was evaluated by performing hip moment measurements over a continuous range of hip angles and at prescribed knee angles. The experimentally acquired data was fitted by exponential functions which separately modeled the moment contribution of tissues deformed by hip flexion and extension. The results of the investigation are discussed with regard to the possible role of the passive joint moments as an energy storage and release mechanism during human walking.

INTRODUCTION

To form a complete model of musculoskeletal dynamics it is necessary to know all the forces and moments acting on the limb segments. Resultant internal moments at a joint arise from active muscular contraction as well as from passive deformation of tissues surrounding and crossing the joint. The resultant moment across a joint can be determined for any activity for which kinematic and external force data can be obtained; a typical example being joint moments during walking (Bresler and Frankel, 1951; Aleshinski and Zatsiorski, 1978). However, in all such cases the relative contribution to the resultant moment from active (muscular contraction) and passive (inactive muscle and fibrous tissue deformation) joint moments is not known.

The term passive joint moment is being used to describe the total resistance to joint movement due to the deformation of all tissues that cross the joint when the muscles crossing the joint are inactive. This passive resistance is viscoelastic. It is not meant to imply that the passive resistance arises solely or in any significant way from the sliding of the opposing joint surfaces as the resistance arising in this way is known to be quite small in normal joints (Linn, 1967).

Several investigations of passive joint moment have been performed. The second metacarpophalangeal joint was extensively analyzed with regard to passive moment resistance in normal joints and the changes which occur in arthritic joints (Wright and Johns, 1960a, b, 1961). Passive knee resistance has received considerable attention in both normal and arthritic joints, as well as in cases of knee laxity due to ligament injury (Fenn and Garvey, 1934; Such *et al.*, 1975; Crowninshield *et al.*, 1976; Markolf *et al.*, Amstutz, 1976; Piziali and Rastegar, 1977). The specific problems addressed in previous studies generally did not require evaluating the influence of passive two joint muscles on the resistance to joint motion. Yet if the effects of passive joint moment are to be included in general models of musculoskeletal dynamics the effects of two joint muscles need to be evaluated (Hatze, 1975).

The purposes of this investigation were twofold. First the passive elastic component of the moment at the hip was determined as a function of hip and knee angles, thereby accounting for the influence of two joint muscles. Secondly, a mathematical function fitting the experimentally determined passive hip moment data was developed.

METHODS

The experimental apparatus was designed to measure the passive elastic hip moment as a function of hip and knee angles. All tests were performed with the subject lying on his side. Referring to Fig. 1, padded supports at the abdominal and mid-thoracic regions (S) provided bracing to keep the subject lying on his side and prevented lateral displacement of the trunk during testing. The subject's leg was strapped to a brace (B) with a hinge at the knee. The braced leg was then suspended at two points in the force measuring apparatus. One point of suspension was a vertical strap (V) located near the mid-thigh. A simple structural analysis of the effects of this strap on the calculated hip moment was performed assuming that its deviation from the vertical was limited to 10°. The other suspension was through a custom load cell (L) mounted distally in the testing apparatus (Advanced Mechanical Technology Inc., Newton, MA, U.S.A.). The load cell measured the force tangential to the circular path of movement of the leg. The position of the hip was measured relative to the load cell in a coordinate system fixed to the apparatus. Rotation of the leg about the hip was performed by a human operator. The speed of rotation was controlled by the

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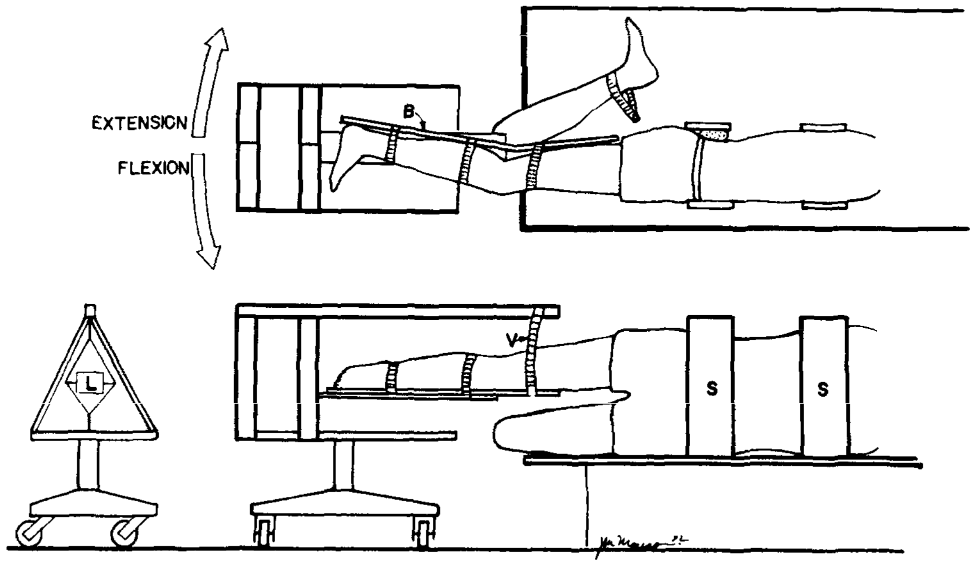

Fig. 1. Front, top and end views of the apparatus used to measure the passive elastic moment at the hip. An electrogoniometer mounted laterally on the right leg was used to measure the hip angles during a test. The electrogoniometer has been omitted from the figure. See text for details of the measurement procedure.

operator who tracked a pointer moving at a known speed. Angular displacement at the hip and knee was measured using an electrogoniometer mounted laterally on the leg but not shown in Fig. 1 (University of British Columbia - Canadian Arthritis and Rheumatism Society). For all tests the leg was rotated in the horizontal plane.

First a series of tests were performed to determine the effects, if any, of the rate of leg rotation on the passive hip moment. It was important to rotate the hip slowly so that the viscous component of the passive joint moment would be negligible as compared to the elastic component. Passive moment measurements were made at angular speeds of 0.017, 0.20, 0.41, 0.81, and 1.62 rad/s. In these tests the knee was held fixed at 0° as measured on the knee brace. In a separate series of tests the presence of possible active muscular contraction during testing was evaluated. Surface electrodes were placed over the quadriceps femoris, medial hamstrings, lateral hamstrings and gluteus maximus of the leg being tested.

For the routine subject testing protocol the hip was exercised through a range from approximately 0° (extension) to 60° flexion. The actual limits were set for each subject by noting the point at which observable pelvic rotation was induced by the leg rotation. A complete cycle of data consisted of rotation of the leg from hip extension to hip flexion and back to the extended position. The cycle was halted at full flexion for a few seconds before starting the return to extension. For each cycle of data the knee was fixed. Eight fixed knee angles were used ranging from 0° (extension) to 70° flexion, in 10° increments as measured on the leg splint. The actual knee angle on the subject was measured by the electrogoniometer. Five cycles of data were taken at each knee angle, making a total of 40 individual cycles.

Data from the load cell and electrogoniometer were acquired in real time using a PDP 11/10 mini computer in conjunction with an LPS-11 analog to digital convertor (Digital Equipment Corporation). Initially all data was sampled at 500 Hz. From this detailed record a 100 point data file was constructed for each cycle. Data files were written onto a disk for later calculation of the passive hip joint moment and graphic presentation of the results. The data were plotted in raw form and after 6 Hz low pass digital filtering (Lesh *et al.*, 1979).

For each knee angle two functions were developed to fit each cycle of the experimental data, one for passive rotation of the hip from hip extension to flexion (M_{HEF}) and one from flexion to extension (M_{HFE}, Fig. 2):

$$M_{\mathrm{HEF}} = A_1\{e^{k_1(\theta-\theta_1)}-1\} + A_2\{e^{k_2(\theta-\theta_2)}-1\} \quad (1)$$

$$M_{\mathrm{HFE}} = A_3\{e^{k_3(\theta-\theta_3)}-1\} + A_4\{e^{k_4(\theta-\theta_4)}-1\}. \quad (2)$$

A_i and k_i ($i = 1, 2, 3, 4$) are constants for each curve. θ_i are the origins of the coordinate systems for each exponential term. The values of θ_1 and θ_3 were set from the maximum hip extension and flexion respectively. The values of θ_2 and θ_4 were offset from θ_1 and θ_3 by the distance AB as shown in Fig. 2. The A_i and k_i ($i = 1, 2, 3, 4$) were determined using an iterative least squares search procedure. First k_1 and k_2 (or k_3 and k_4) were assigned values and A_1 and A_2 (or A_3 and A_4) were determined using a least squares analysis. For each set A_1, A_2, k_1, and k_2 (or A_1, A_2, k_1 and k_2) the sum of the squared error or deviation between the experimental and fitted data was determined. The

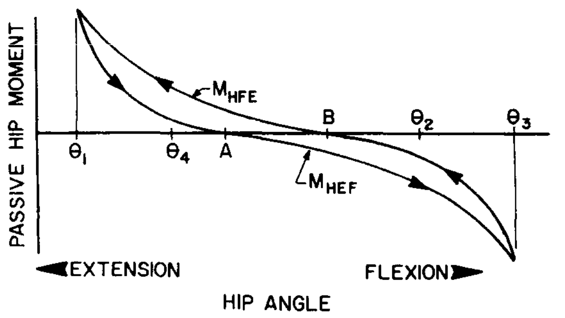

Fig. 2. Idealized representation of the measured passive elastic hip moment obtained from one complete cycle of testing. The moments M_{HEF} and M_{HFE} are those being fitted by equation (1) and (2). The origins of the four exponential functions in equations (1) and (2) are denoted by θ_1, θ_2, θ_3 and θ_4.

sequence of assigning values to the k_i, determining the A_i from a least squares analysis and computing the deviation between the experimental and fitted data was continued until the change in the k_i was less than 10^{-4}. When this tolerance was met, the A_i and k_i values were accepted and were used in equations (1) and (2).

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

Subject	Age (yr)	Height (m)	Weight (N)
221	36	1.85	756
225	23	1.60	734
226	34	1.68	778
228	34	1.85	694

A_i and k_i ($i = 1, 2, 3, 4$) were also determined as a function of knee angle. A second order polynomial was fitted to each of the constants determined at each knee angle. A least squares analysis was used to fit the data.

Four adult male subjects were employed (Table 1). These subjects had no previous history of trauma or disease which would effect the measurements.

RESULTS

No differences were found among the passive hip moments measured at different rates of hip rotation. Thus, all subsequent measurements were made at 0.20 rad/s. This speed was chosen since it was relatively easy for the operator to follow. The myoelectric signal recorded at this speed showed no change in electrical output, above the resting level, in muscles tested, i.e. quadriceps, medial and lateral hamstrings, and gluteus maximus. Analysis of the vertical strap suspending the leg showed that it would carry a moment of approximately 2–4 N m which was not considered to be a significant error. Comparison of the low pass filtered vs unfiltered data for one complete test cycle shows little distortion of the raw data, but a clear removal of noise (Fig. 3). The noise in the raw data arises mainly from vibration of the apparatus as it is moved over the floor. All hip moment data is plotted according to the sign convention that a positive moment is one that would cause hip flexion, while a negative moment is one that would cause hip extension.

Two joint muscles had an anatomically consistent effect on the passive hip moment (Fig. 4). With the knee set in the maximum flexed position the hip moment was greatest in hip extension and least in hip flexion.

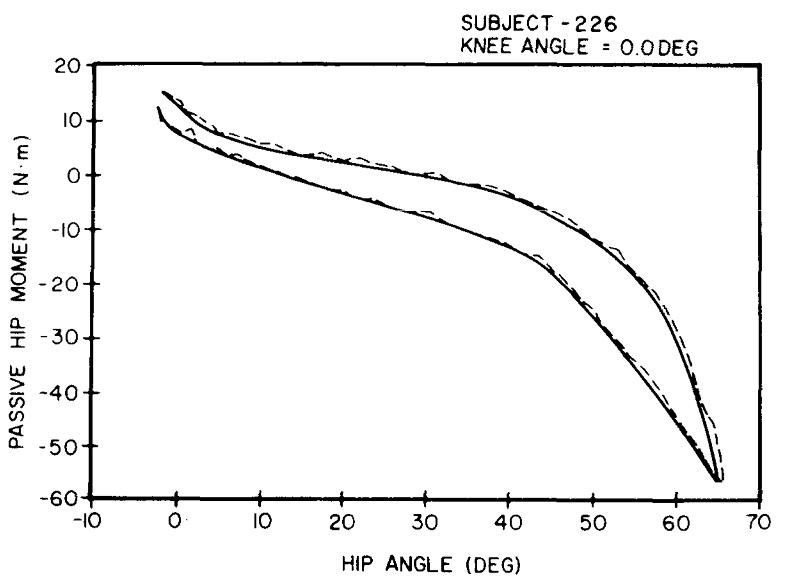

Fig. 3. Comparison of the measured passive hip moment (dashed line) to the same data after low pass filtering at 6 Hz (solid line). Data are from a single cycle of measurement.

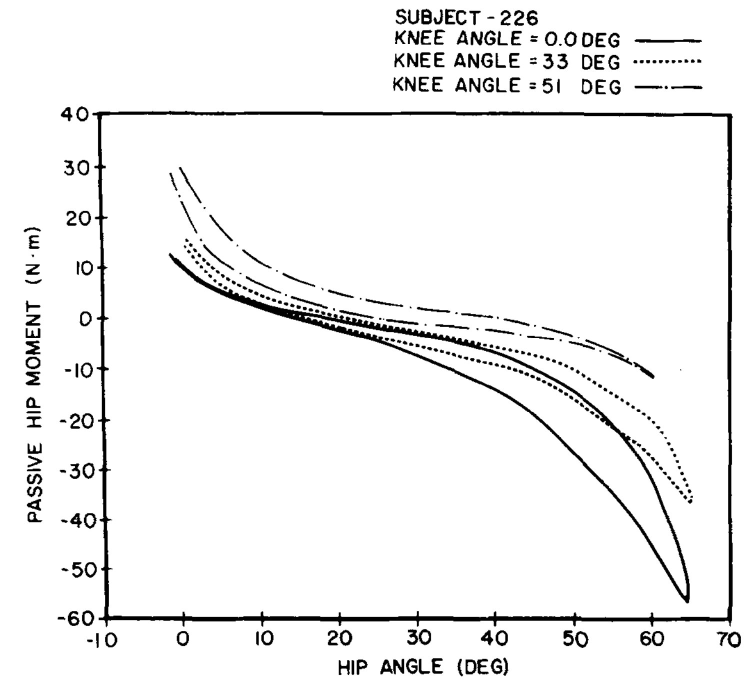

Fig. 4. Average measured passive hip moment at three different knee angles after 6 Hz low pass filtering. Five cycles of raw data at each constant knee angle were averaged after filtering.

This was attributed to the biasing moment induced at the hip due to the stretching of the rectus femoris by flexing the knee. Conversely, with the knee set in an extended position the hip moment was greatest in hip flexion and least in hip extension. This was attributed to the biasing moment induced at the hip due to the stretching of the semi-tendinosus, semi-membranosus and long head of the biceps femoris by extending the knee. Fixed knee angles between the maximum knee flexion and full extension yielded a hip moment bounded by those measured at the extremes of the knee angle.

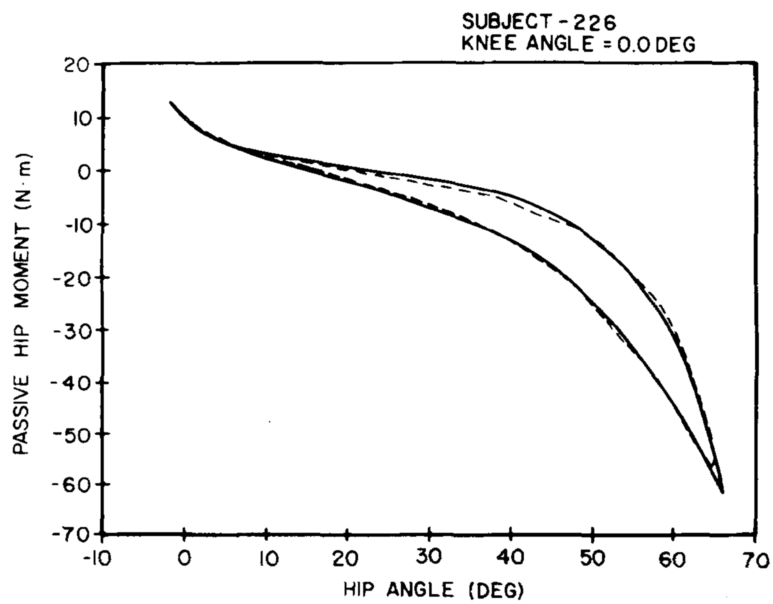

Fig. 5. Comparison of the average of five cycles of data at a constant knee angle (dashed line) to the least squares fit of the data using equations (1) and (2) (solid line).

Table 2. Constants in equations (1) and (2) for each of the four subjects tested. Data are from tests at selected specific knee angles

Subject	Knee angle	A_1	K_1	θ_1	A_2	K_2	θ_2	A_3	K_3	θ_3	A_4	K_4	θ_4
	15	−1.68	0.0425	5.80	0	−0.376	79.5	0	−0.210	81.1	−1.59	0.0425	7.40
221	38	−3.08	0.0300	7.30	0	−0.195	77.9	0.00197	−0.120	81.1	−1.22	0.0425	10.5
	50	−2.34	0.0300	5.7	0.00570	−0.120	73.4	0.00364	−0.120	78.5	−1.00	0.0425	10.8
	19	−0.276	0.0800	3.0	0.000627	−0.170	62.4	0.0112	−0.120	63.9	−0.221	0.0831	4.51
225	38	−0.557	0.0613	3.6	0.00680	−0.139	60.6	0.0165	−0.120	62.6	−0.0952	0.0925	5.56
	53	−0.629	0.0550	3.4	0.0143	−0.145	54.3	0.0414	−0.109	62.5	−0.187	0.0863	11.6
	17	−2.85	0.0425	−0.900	0.000255	−0.176	60.2	0.000269	−0.165	65.8	−0.334	0.0800	4.66
226	40	−0.661	0.0550	−0.400	0.0111	−0.145	50.8	0.0489	−0.0975	62.8	−0.178	0.0925	11.6
	50	−0.387	0.0550	−1.60	0.0938	−0.120	45.3	0.0952	−0.0947	60.8	−0.0774	0.105	13.9
	15	−0.286	0.0800	9.9	0.0276	−0.0950	69.8	0.0101	−0.109	72.6	−0.0156	0.130	12.7
228	40	−0.172	0.0800	11.1	0.484	−0.0575	70.8	0.124	−0.0750	74.0	−0.0313	0.105	14.3
	52	−1.037	0.0425	9.8	0.299	−0.0700	67.8	0.181	−0.0750	71.0	−0.109	0.0800	13.0

Values less than 1×10^{-5} were recorded as 0.

Applying the least squares fitting procedure to determine the constants in equations (1) and (2) yielded a good fit of the experimental data (Fig. 5). While the qualitative characteristics of the passive hip moment were the same among the subjects tested there were quantitative differences. A summary of the constants A_i and k_i and the origins θ_i for selected knee angles for each of the subjects tested is given in Table 2.

While the constants in equation (1) could always be determined for each knee angle they could not be reliably determined as a continuous function of knee angle. The least squares analysis always yielded a second order polynomial, but the fit to the experimental data using the constants determined in this way was not as good as that obtained by fitting each curve separately. Therefore no further attempt to determine the constants as a function of knee angle was made.

DISCUSSION

The results obtained in this investigation show the passive resistance at the hip to be similar to that recorded at other joints. The insensitivity of this moment measurement to the rate of hip rotation is similar to that found at the knee (Fenn and Garvey, 1934). The absence of electrical activity in the muscles tested supports the idea that a truly passive measurement was being made. However, the myoelectric signal was recorded only from muscles near the surface so the effects of active contraction in deeper muscles, such as the iliopsoas, was not evaluated.

The passive hip moments take on considerable potential significance when they are evaluated relative to the total joint moment in a common activity such as walking. Using typical joint angles for toe-off (hip less than 10° flexion, knee 40° flexion) and incipient heel strike (hip 60° flexion, knee 0°, Murray, 1967) the corresponding measured passive elastic hip moments for all subjects tested are 10–15 Nm at toe-off and 20–40 Nm at incipient heel strike. The passive hip moment at toe-off is approximately 30–50% of the total computed hip moment at this time (Bresler and Frankel, 1951). The passive moment just before heel strike is more difficult to evaluate, relative to the total moment, as the total moment is changing rapidly at this time. Using maximum and minimum values of the total moment in the neighborhood of heel strike the passive hip moment can be calculated to be approximately 60–100% of the total moment at this time. An implication of this analysis is the possibility of passive elastic energy storage and release during human gait. The storage of elastic energy in tissues has been shown to be an important element in many human activities such as running (Cavagna *et al.*, 1964) and some exercises (Thys *et al.*, 1972). Similar utilization of elastic energy storage has been shown for certain classes of animal locomotion such as hopping (Dawson and Taylor, 1973; Alexander, 1977) and insect flying (Goldspink, 1977). Clearly, considerable approximation has been introduced in making these evaluations of the relative contributions of the passive and active joint moments since the data employed have been gathered from three independent sources. However, the apparently high contribution of the passive joint moment to the total moment does not appear to be insignificant.

In abnormal gait fixed joint contractures further limit the range of motion at a joint to within that found in normal gait. In these pathological cases the passive storage and release of elastic energy in tissues may be greater than that found in normal gait. Further, more detailed study of passive energy storage and release in normal and pathological gait is needed, and is currently under way.

In summary, the passive elastic moment at the hip was measured *in situ*. The experimental data was fitted by exponential functions representing the separate contributions of tissue deformed by hip flexion and hip extension. The storage and release of elastic energy during gait in passive structures was shown to be a possible mechanism employed in normal gait.

Acknowledgement—This work was supported by NIH musculoskeletal diseases program grant No. 5 RO1 AM 22165.

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