The Dynamics of the Subtalar Joint in Sudden Inversion of the Foot

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The human subtalar joint was modelled as a quasi-linear second-order underdamped system to simulate sudden inversion motion of the foot relative to the shank. The model was fed with experimental data obtained from six subjects on a specially constructed apparatus. A total of 35 deg inversion was produced on the tested leg rapidly enough (lasting less than 40 ms) in order to ensure that the protective muscles are not activated. The parameters of the joint were evaluated and the following ranges were obtained at 35 deg inversion: elastic stiffness 14-52 Nm rad^{-1}, damping coefficient 1.4 – 2.9 Nms rad^{-1}, and natural frequency 78-125 Hz. The effects on the test parameters of weight bearing amount, foot dominance, and protective footwear were studied on one subject.

Introduction

The dynamic properties of the human foot-shank articulation complex are of great interest if the motion of the lower limb is to be studied as a multisegmented system. This articulation complex consists of two major joints, the talocrural (ankle) and the subtalar. Neither of these joints act as ideal hinges. In fact, Gray's Anatomy defines the subtalar joint as a shallow ball and socket joint (Warwick and Williams, 1973) and recently Siegler et al. (1989) reported an average of about 5 degrees of inversion/eversion and about 13 degrees of abduction/adduction occurring at the talocrural joint. In addition, these joints are considerably coupled together. Siegler et al. (1989) also studied the load-displacement and flexibility characteristics of the joint and identified the effect of injury on these characteristics by retesting the load-displacement behaviour after sectioning the ankle ligaments in the specimens studied (Chen et al., 1989).

Whilst motion of the ankle joint has been studied in the past, the subtalar inversion-eversion motion has seldom been measured, despite its frequent involvement in everyday activity and in trauma such as ankle sprain, when the foot is forced into rapid inversion motion. It would therefore be of interest to study the dynamic properties of the subtalar joint. It should be noted though that there is a limitation of examining the subtalar joint in vivo without invasive or radiographic techniques. This is so since the talocrural joint can account for 20–25 percent of the total inversion (Siegler et al., 1989). Most of the previous works on the mechanics of the ankle joint complex in vivo were thus done with the motion in the sagittal plane, i.e., in plantar and dorsi flexion of the joint. The relationship between applied external moment and resulting angular displacement of the joint was studied in oscillatory motion, in which the foot was driven relative to the shank (Agarwal and Gottlieb, 1977; Gottlieb and Agarwal, 1978). The moment was applied by means of a band-limited Gaussian signal of 0–50 Hz. This group also studied the effects of the activation level of the muscles around the joint, and of the dynamic reflex as a result of applying the torque suddenly, on the results they obtained.

A very similar method was also used by another group of investigators (Kearney, 1978; Kearney and Hunter, 1982; Hunter and Kearney, 1982, 1983; and, more recently by Weiss et al., 1986) with variations in the resting posture of the tested subject, the mode of attachment of the foot, and the manner of applying the displacement perturbations to the foot, which had a power spectrum with a flat region from 0.5 to 20 Hz. Additionally, the amplitude of motion, which was considered an important parameter, differed between the two groups. Fatigue of the muscles during exercise was studied as well. These experimental considerations may be relevant in the design of methodology for studying the subtalar joint motion.

Both the above groups made use of a second-order quasi-linear underdamped system for analyzing the ankle joint in plantar and dorsi flexion, by which the following parameters could be evaluated: elastic stiffness, damping coefficient, natural frequency, and moment of inertia of the foot about the ankle joint. The results obtained were found to be affected by the level of activation of the muscles. These methods of modelling and of indentification of parameters to characterize the dynamics of the ankle may also be applicable to the subtalar joint.

Among the few studies made in vivo on the foot-shank joint in planes of motion other than the sagittal, the following should be mentioned: Johnson et al. (1976) statically evaluated the stiffness of various football boots in inversion-eversion motion. They showed that when using rigidly attached high boots, ligamentous load on the subtalar joint was reduced considerably. This finding demonstrates the condi-

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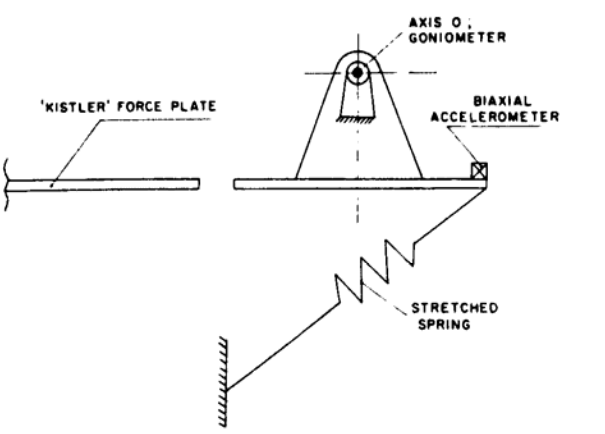

Fig. 1 Scheme of specially constructed apparatus, consisting of a swivelling platform driven by a stretched spring and made to rotate with the tested leg about axis 0. The platform is equipped with a goniometer and biaxial accelerometer, and is colaterally installed with a "Kistler" piezoelectric forceplate, on which the other leg was positioned.

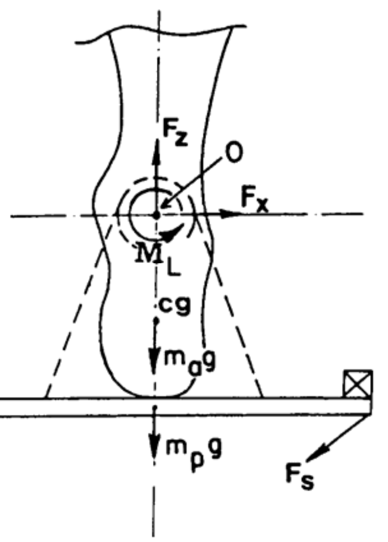

Fig. 2 Free body diagram of the swivelling platform including the tested foot. All external loads are shown for the combined system including the platform and foot up to the level of the subtalar joint.

tions under which footwear may protect the joint. Mote and Lee (1982) studied the dynamics of the ankle in rotation motion in the medial-lateral plane in both laboratory and skiing apparatuses and identified stiffness, damping and inertia parameters for this motion. Weight bearing on the foot and muscle-induced torsion were used as test variables, and were found important in studying the dynamics of the ankle.

In a previous study, Isakov et al. (1986) found that in unexpected and sudden inversion motion of the foot, the stretch reflex on the peroneal muscles remains unelicited for a period of 70 ms approximately from the onset of motion. In the present study, the dynamic properties of the human subtalar joint in sudden inversion were studied in vivo on a specially designed apparatus. This type of motion is likely to occur in conditions of inversion sprain of the ankle joint. It is thus important to quantify the kind of anatomical restraints involved in the joint. The purpose of this study was to measure these restraints. The subtalar joint was modelled as a second order system, from which the elastic stiffness and damping coefficient of the joint, as well as the natural frequency of the foot and its moment of inertia about the joint were evaluated in inversion-eversion motion.

2 Mechanical Model and Apparatus

The human subtalar joint was modelled as a quasi-linear second-order underdamped system to simulate sudden inversion motion of the subtalar joint. Inversion motion is defined as supination of the foot relative to the leg, about an anteroposterior horizontal axis (Inman, 1976). In the motion studied in this work, the foot was positioned at 10 deg external rotation from this axis. In sudden enough motion, the muscles are not actively involved in protecting the joint (Isakov et al., 1986). The internal moment $M_L(\theta)$ at the subtalar level may be written as

$$M_L(\theta) = k(\theta)\,\theta + c(\theta)\,\dot{\theta} \tag{1}$$

where θ is the inversion angle of the subtalar joint, and $k(\theta)$ and $c(\theta)$ are the elastic and damping coefficients of the joint, respectively, during constant-level weight bearing.

Evaluation of $M(\theta)$ during the process of inversion $\theta(t)$ is essential to solve equation (1) and to determine the elastic and damping coefficients of the joint. A special apparatus, shown schematically in Fig. 1, was thus constructed. It consisted of a swivelling platform, on which the tested leg and foot were positioned and made to purely rotate at the level of the subtalar joint (axis 0, Fig. 1), from the neutral position up to 40 deg inversion angle. The effective range of the experiment was however only 35 deg, due to the presence of a bumper in the last 5 deg of rotation. The average normal range of motion for inversion was reported to be 40 deg ± 7 deg (Inman, 1976). All the individuals tested had a larger range of motion and there was therefore no potential for injury. It should be noted that this fixture is measuring the amount of inversion for the combined talocrural and subtalar joints. The level of the subtalar joint was determined as 1 cm below the lateral malleolus. Vertical adjustment for each subject was made by using inserts between the feet and the ground. Horizontally, the swivel platform axis was aligned with the center of the line connecting the medial and lateral malleoli. It was estimated that the maximal resulting error from this alignment is less than 5 mm. The platform was driven by stretched linear springs with known stiffnesses to allow a rapid enough complete trip (lasting less than 40 ms) and to ensure that during rotation the protective muscles are not activated by the stretch reflex. The swivelling platform was collaterally installed with a "Kistler" piezoelectric force platform, on which the other leg was made to stand in the same height as the tested leg, and by which weight bearing on the tested leg prior to and until the onset of the test could be controlled.

The kinematics of the swivelling platform was measured by means of a goniometer located at the axis 0. From the readings of a biaxial accelerometer attached to the platform (Fig. 1), the angular acceleration as well as the linear acceleration components could be measured directly. Direct measurement of acceleration is advantageous if the rather noisy double differentiation procedure of the goniometer data is to be avoided.

The free body diagram of the swivelling platform, including the tested foot up to the level of the subtalar joint, is shown in Fig. 2. The external loads on this system include the spring driving force F_s, gravitational forces of the platform $m_p g$ and of the foot $m_a g$ (up to the level of the subtalar joint), the reactive force components at the joint and at the axis of rotation combined together and denoted F_x and F_z, and the opposing moment M_L in the subtalar joint. At any angle θ, the sum of moments ΣM_0 about the axis 0 may be written as follows:

$$\Sigma M_0(\theta) = M_s(\theta) + M_p(\theta) + M_a(\theta) + M_L(\theta) \tag{2}$$

In equation (2) M_s is the driving moment of the stretched springs and M_p, M_a, M_L are the opposing moments due to platform gravity, foot gravity and internal moment around the subtalar joint, respectively. In the right-hand side of equation (2), the only unknown is $M_L(\theta)$, the others being directly

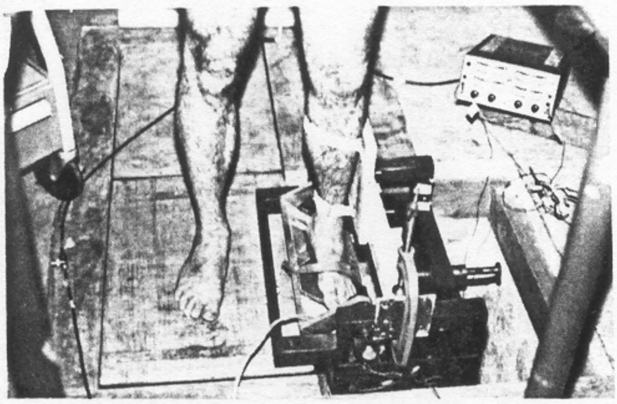

Fig. 3 Subject standing with the tested leg on the swivelling platform and other leg on the "Kistler" forceplate, during a test

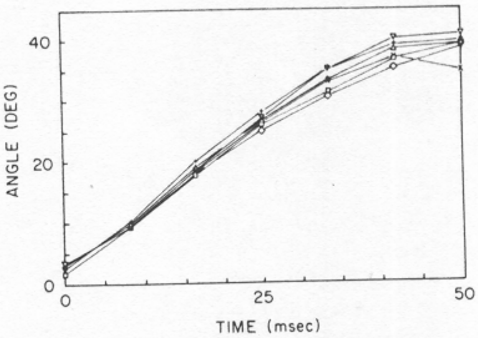

Fig. 4 Variation of inversion angle versus time in standard test for all subjects

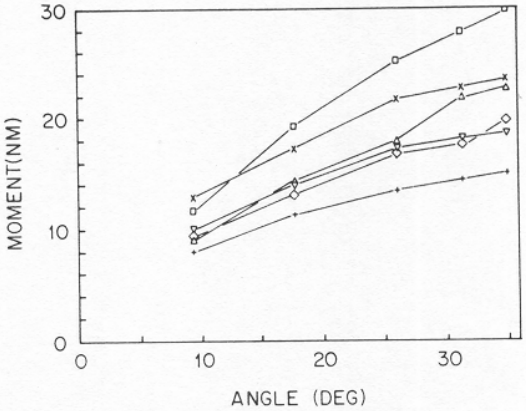

Fig. 5 Variation of the mean subtalar moment during inversion of the foot in standard test for all subjects

determined from the data of the system and from anthropometric data (Hatze, 1979; Winter, 1979).

Additionally,

$$\Sigma M_0 = (I_p + I_a)\ddot{\theta} \quad (3)$$

where I_p and I_a are the moment of inertia about the axis 0 of the platform and of the foot, respectively.

The procedure for solving equations (1), (2), and (3) is as follows: The sum of moments ΣM_0 is first obtained from equation (3), from the geometry and the measured kinematics of the swivelling platform and from anthropometric data. Equation (2) is next used to solve the subtalar moment $M_L(\theta)$, which in turn is used in equation (1) to solve $k(\theta)$ and $c(\theta)$, for each interval angle, using the same method described previously by Mizrahi and Susak (1982). Apart from the elastic stiffness and damping coefficient, the following were also calculated from the moment $M_L(\theta)$:

– Overall stiffness $k_1(\theta)$ defined as:

$$k_1(\theta) = M_L(\theta)/\theta(t) \quad (4)$$

– Natural frequency w_n of the tested system

$$w_n = [k(\theta)/(I_p + I_a)]^{1/2} \quad (5)$$

– Evaluation of the moment of inertia of the foot I_a up to the level of the subtalar joint.

The data collected were on-line digitized at 1000 Hz and fed into an IBM-XT computer, where they were processed.

3 Subjects and Procedure

Six healthy subjects, one female and five males, with no orthopaedic deficiencies and without known history of previous ankle sprains, took part in this study. Average data were as follows: age 33 yr (range 27-42), height 172 cm (range 165-185), and mass 66 kg (range 55-70). During each experiment the feet were positioned at 10 deg external rotation from the center line. The tested foot was tightened to the platform by means of a strap passing over the dorsal aspect of the foot. The shank of the tested leg was also tightened to a vertical support, to avoid compensatory movement of the shank of that leg during the experiment. The knee joints were in full extension. As stated earlier, placement of the opposite leg on a forceplate, at the same height as the tested leg, enabled the subject to control weight bearing on each foot, prior to and until the onset of each experiment. This was done by means of an oscilloscope displaying the force reading, and placed in front of the subject.

In the standard tests, each subject was asked to stand with bare feet in a relaxed position, with an even distribution of his weight between the legs (Fig. 3). The swivelling platform was then unexpectedly released to rotate, under the influence of the stretched springs. EMG of the peroneal muscle showed that there was no anticipation of the motion. Each experiment was digitized on the computer and repeated at least three times for every subject. After each experiment, the foot was removed from the test apparatus. Before retested, the foot was repositioned and the straps reattached.

On one subject a supplementary series of tests were performed to verify the trend of variation of the results due to the following parameters: amount of weight bearing, dominant against opposite foot and different protective footwear. As previously stated, each of these experiments was repeated at least three times. A student t-test was performed for significance of the differences between the standard and varied testing conditions.

4 Results

Variation of the inversion angle versus time in standard testing conditions for all the six subjects is shown in Fig. 4. The average inversion rate in the full range of motion studied was 15.69 ± 0.66 rad.s^{-1}.

Curves of the mean subtalar moment against the inversion angle for all the tested subjects are shown in Fig. 5. All the curves ascended monotonically, reaching maximal values at the end of the range of motion, i.e., at 35 deg, with a slight tendency to level-off there. By comparison with the curves presented in Fig. 4, it can be seen that the moment curves tend to level-off in correspondence with the decrease in the inversion rate.

The overall stiffness curves defined in equation (4) are shown in Fig. 6 and they are found to generally have a descending pattern, with the lowest values at the end of the range of motion. Variation of the elastic stiffness is presented in Fig.

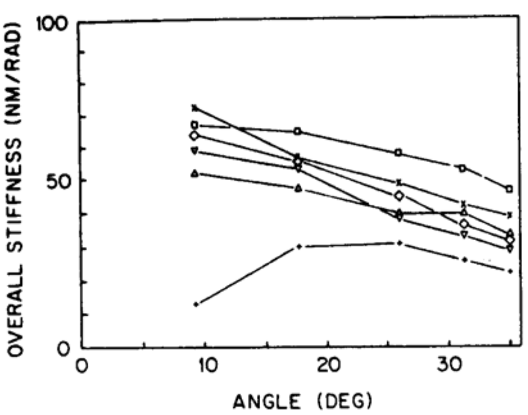

Fig. 6 Variation of the mean overall stiffness during inversion of the foot in standard test for all subjects

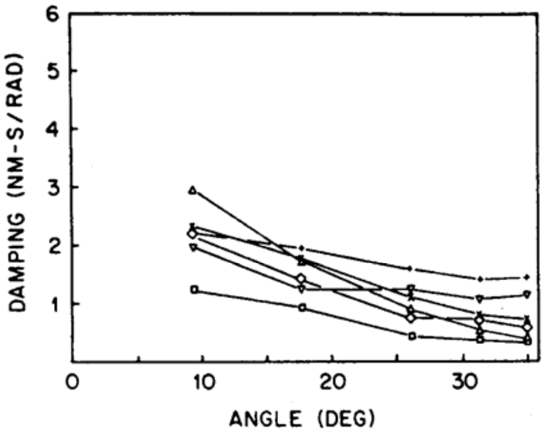

Fig. 8 Variation of the mean damping coefficient during inversion of the foot in standard test for all subjects

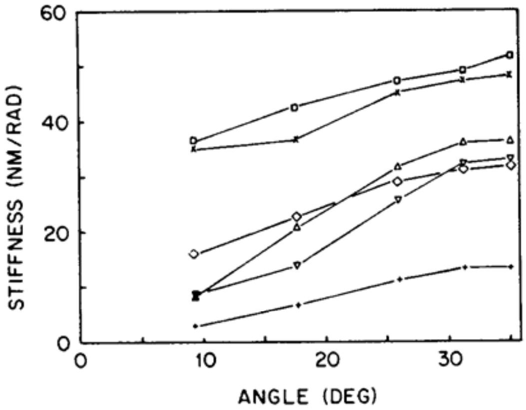

Fig. 7 Variation of the mean elastic stiffness during inversion of the foot in standard test for all subjects

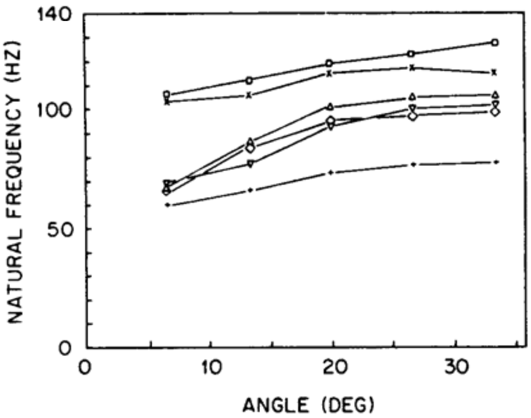

Fig. 9 Variation of the mean natural frequency during inversion of the foot in standard test for all subjects

7. The curves of elastic stiffness are found to somewhat mimic the moment curves. The damping coefficient, shown in Fig. 8 demonstrates a rather opposite pattern to that of the elastic stiffness. The natural frequency of the system including the foot up to the level of the subtalar joint, as defined in equation (5) is shown in Fig. 9.

Calculation of the moment of inertia of the foot about the subtalar joint gave steady values ranging between 0.0030 and 0.0037 kg m^2. These values were invariable with respect to the inversion angle.

Intra-subject variability is demonstrated in Table 1 for one of the subjects. The averages of four different tests are presented together with their standard deviations. It should be remembered that between the tests, the foot was removed from the apparatus and reattached.

The effects of distribution of weight bearing between the legs, foot dominance and protective footwear on the average values of the tested parameters are shown in Table 2. The reference condition to which these effects are compared corresponds to testing of the dominant leg with bare feet and with even distribution of the weight between the legs.

It is seen that the moment, overall stiffness and elastic coefficient, all increased with increasing load, the differences being statistically significant ($p < 0.05$). On the other hand, the damping coefficient gave, in comparison to the reference condition, higher values both in the high and low weight-bearing tests. This difference, however, was statistically significant only in the high weight bearing ($p < 0.05$). Comparing the dominant and the opposite feet together shows that the subtalar moment, overall stiffness and elastic coefficient were higher in the former case, this difference, however, not being statistically significant. The effect of footwear protection was pronounced only in the elastic stiffness parameter. Wearing of high boots alone did not show any measurable difference. However, if an elastic dressing was inserted between the foot and boot to provide a tighter fit, this difference was noticed ($p < 0.05$). The effect of footwear on the other parameters was not significant.

5 Discussion

The in vivo parameters of the subtalar joint in sudden inversion motion of the foot relative to the shank were the focus of this investigation. This study thus differs from the previous studies concerned with the in vivo dynamic properties of the ankle joint complex in two respects: 1. Direction of the major motion studied, inversion in our case and plantar-dorsi flexion in the other studies; 2. Motion is applied unexpectedly to provoke the full inversion motion desired in a rather short period of time. In this way, activation of the protective peroneal muscles is not involved (Isakov et al., 1976). Under these conditions, the reported experiments may be considered to mimic ankle sprain motion in controlled conditions and more restricted total rotation.

The purity of the motion applied may be questioned in view of the coupling existing between the joints of the ankle (Siegler et al., 1989). Furthermore, the deliberate positioning of the foot in 10 deg external rotation to the midline might have caused a further increase in crosstalk between the ankle and subtalar joints, resulting in plantar flexion. This effect was nevertheless considered secondary to the major inversion motion applied at the subtalar joint.

In spite of the above-mentioned differences between this

Table 1 Intra-subject variability as demonstrated on one subject (denoted Δ in Figs. 4-9). The averages of four different tests and their standard deviations (in parentheses) are presented. As described in the text, the foot was removed from the apparatus and reattached between the tests.

Inversion Angle °	M Nm	k_1 Nm rad^{-1}	k Nm rad^{-1}	C Nms rad^{-1}	W_n Hz
9	9.11 (3.0)	51.56 (8.5)	8.33 (2.5)	2.96 (0.75)	55.70 (3.6)
18	14.22 (3.0)	46.98 (5.7)	20.55 (5.0)	1.68 (0.50)	83.85 (8.2)
26	18.01 (3.2)	39.53 (6.3)	31.66 (5.5)	0.88 (0.20)	104.12 (10.1)
31	22.02 (2.5)	39.53 (5.7)	36.11 (5.0)	0.53 (0.11)	111.23 (8.3)
35	22.40 (3.5)	33.80 (11.4)	37.55 (4.0)	0.41 (0.20)	112.40 (7.5)

Table 2 Effects of weight bearing on the tested leg, foot dominance, and protective footwear for one subject on the average maximum of the tested parameters. Results are referred to the reference condition, corresponding to an even distribution of body weight between the legs and testing of the dominant foot with barefeet. Values presented are averages (standard deviations, in parentheses).

	M Nm	k_1 Nm rad^{-1}	k Nm rad^{-1}	C Nms rad^{-1}
Reference condition	30.01 (8.5)	67.4 (22.0)	51.76 (12.5)	1.32 (0.50)
75% W-Bearing	37.7* (2.7)	96.5* (5.5)	69.01* (2.5)	2.81* (0.06)
25% W-Bearing	23.2* (4.6)	62.3* (9.2)	10.6* (5.8)	2.83 (0.20)
Opposite foot	23.0 (2.2)	48.0 (6.3)	34.6 (5.3)	1.30 (0.16)
High boots	31.6 (2.8)	67.6 (4.9)	51.8 (4.4)	1.08 (0.29)
High boots + Elastic dressing	43.5 (1.7)	105.5* (5.5)	84.1* (7.1)	1.92 (0.50)

*Statistically significant difference from the reference condition ($p < 0.05$).

Table 3 Comparison of the parameters obtained in this study (for 10 - 35° range of motion) and those found in the literature for the ankle joint in vivo

Parameter compared	This study	Kearney and associates*	Agarwal and Gottlieb**
Range of motion (deg)	10 - 35	-51 - +17	±12
Moment (Nm)	8 - 30	5 - 35	0 - 12
Elastic stiffness (Nm/rad)	3 - 52	0 - 100	13 - 54
Damping coefficient (Nms/rad)	0.3-2.9	0.6-1.0	0.2-0.7
Natural frequency (Hz)	58-125	10-20	5

*Kearney, 1978; Kearney and Hunter, 1982; Hunter and Kearney, 1982, 1983; Weiss et al., 1986.
**Agarwal and Gottlieb, 1977; Gottlieb and Agarwal, 1978.

study and previous studies on the ankle joint, it may be of interest to compare the actual results obtained. Such a comparison may indicate whether the magnitudes obtained are of comparable values. Firstly, it should be noted that the intensities of the moment applied to the joint were higher in our study as compared to the those reported earlier. Chen et al. (1989) measured moment values within the range of 5 Nm for static eversion of 20° of cadaver specimens. For in vivo plantar-dorsi flexion of the foot, Weiss et al. (1986) reported moment values of up to 10 Nm, the range of motion being 50 deg. Similar values were also reported by Agarwal and Gottlieb (1977). In all these studies it was found that the moment increased with increasing the angle of rotation of the joint from the neutral position. In our study, the moments by which the joint was loaded reached values of as high as 30 Nm. This difference can be attributed to the fact that in our experiments there was a muscle tone due to weight bearing during standing, producing higher moments as compared to in vitro conditions, or in vivo tests conducted on seated individuals in the previous studies. Another difference is the high deformation rate applied; the latter is easily seen in the slopes of the angle-time curves (Fig. 4). It is observed that as long as the deformation rate is high, the moment rate of increase is also high. This occurs in the range of 5-20 deg inversion angle. At angles higher than 20 deg inversion, the inversion curves start to level off, indicating a decrease in deformation rate, accompanied by a corresponding levelling-off in the moment curves. Thus, this dependence on deformation rate somewhat moderates the increasing trend of the static moment with rotation angle, at inversion angles higher than 20 deg. To examine this dependence quantitatively, we calculated separately the two components of the total internal moment, i.e., the elastic $M_k(\theta) = k(\theta)\theta$, and the damping $M_c(\theta) = c(\theta)\dot{\theta}$. This showed that M_c has a significant contribution at low inversion angles, but this contribution gradually decreases as the rotation angle increases. M_k, however, is initially small, but it monotomically increases with the inversion angle.

The differences found between the inversion angle curves of the different tested subjects may be explained as follows: In the onset of the curves, the slight difference as well as the 2 - 3 deg offset from the origin of the axes can be attributed to the accuracy in setting the initial starting position of the platform. As inversion motion proceeds, the curves somewhat fan out from one another, reflecting the differences in M_L (moment produced by the joint), of the different subjects on the driven platform.

It may be argued whether the obtained moments, stiffnesses and damping coefficients represent the intrinsic values for the subtalar joint, or are only "apparent" values for the particular test fixture kinetics studied. To answer this question it should be remembered that the role of the apparatus used was twofold: (a) to provide a dynamic analysis of the whole system, including that of the subtalar joint, and (*b*) to provide a stable support for the shank and foot, so as to allow a rapid and controlled motion of the subtalar joint. As in previous studies on the ankle joint where specific types of motion were prescribed to the joint (Gottlieb and Agarwal, 1978; Kearney and Hunter, 1982), also in this case the solution of the dynamic system dissociates the properties of the joint from those of the appartus. It is thus believed that the resolved internal moment $M_L(\theta)$ pertains to the subtalar joint itself, for the motion analysed. Relating to the error involved in evaluating this moment, it is estimated to be of the order of 2 N-m for an estimated maximal excursion of 5 mm of the subtalar joint axis of rotation relative to the swivel platform axis of rotation (for a 70 kg subject). Comparison of parameter values obtained in this study with those obtained previously for the ankle joint in plantar and dorsi flexion is presented in Table 3. The elastic stiffness values were comparable, whereas the damping coefficient was higher in our study. As for natural frequency, it should be remembered that the values presented in Table 3 reflect both the testing platform and foot. Values of this parameter for the foot alone should thus be calculated and compared. This is easily done by taking the average stiffness and moment of inertia of the foot, using $w = (k/I_a)^{1/2}$. Doing this calculation, we obtain 120 Hz for the results in our study and 127 Hz for the results reported by Weiss et al. (1986).

There are several indicators for the reliability of the system used and the results obtained. Firstly, reproducibility of the experiments as reflected by the intra-subject variability. It should be remembered that over the repeated testing episodes for each subject, the foot was removed from the test apparatus, repositioned and restrapped, before being tested. Secondly, the calculated moment of inertia, ranged between 0.0030 and 0.0037 kg m^2 for the range of inversion angles used. This compares very well with the value of 0.0035 kg m^2 obtained from anthropometric data. In addition were the effects on the results obtained for weight bearing and protective footwear. The increase of elastic stiffness with weight bearing is in agreement with previous work by Greene and McMahon (1979). Increase in stiffness with load may be attributed to the increase in muscle tone, which is necessary for the extra weight bearing, as well as for maintaining balance in the joint in this condition.

The increase in stiffness as a result of wearing well-fitted high boots correlates with the findings of Johnson et al. (1976), who showed that when using rigidly attached high boots, ligamentous load on the joint was reduced considerably.

To confirm the above effects of the parameters tested and to provide a quantitative relationship, it is clear that both the number of experiments and the number of tested subjects should be increased.

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