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Forces predicted at the ankle during running

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ABSTRACT

BURDETT, RAY G. Forces predicted at the ankle during running. *Med. Sci. Sports Exercise*, Vol. 14, No. 4, pp. 308-316, 1982. A biomechanical model of the ankle joint was developed and was used to predict the forces at the ankle during the stance phase of running. Measurements from five cadavers were averaged to obtain insertion points and directions of pull of equivalent tendons with respect to the assumed center of the ankle joint. A minimum joint force solution was obtained by assuming that only two equivalent muscle groups could exert force at one time. Three subjects ran at 4.47 m/s across a force platform that recorded the external forces and moments acting on the foot. Cinematography was used to measure the foot and leg positions during stance. Peak resultant joint forces ranging from 9.0 to 13.3 times body weight and peak Achilles tendon forces ranging from 5.3 to 10.0 times body weight were predicted. Small variations in some cases resulted in large differences in predicted forces. The highest tendon forces predicted exceeded those reported to cause damage to cadaver tendons in other studies.

ANKLE JOINT, BIOMECHANICAL MODEL JOINT FORCES, MUSCLE FORCES, RUNNING

During sports activities the ankle joint is a frequent site of injury. An accurate knowledge of the forces occurring in the tendons, ligaments, and bones of the ankle joint is important for understanding, treating, and preventing injuries to this area. However, direct measurements of internal forces are difficult to obtain. Force transducers must be placed in series with the tendon or ligament to obtain such measurements. Measurement of forces acting between the bones requires the replacement of one or both of the bony components with force-measuring devices. Direct measurement of the forces acting between the acetabulum and the head of an instrumented hip prosthesis has been done in this manner (10). Such measurements require surgical placement of the transducers, a procedure that is impractical and unethical if performed on healthy humans. It may be possible, however, in the future to include such measurement devices in ankle joint prostheses. Until this is done, indirect methods are necessary to estimate the forces at the ankle.

Indirect methods involve the calculation of the forces from a biomechanical model of the joint based on its anatomical structure. Simplifying assumptions are made concerning the motion, geometry, and the activity of the muscles crossing the joint so that a solution of the forces in the model is possible. Because of the many assumptions and approximations involved, these models produce estimates of actual forces. The joint and muscle forces occurring at the ankle during walking (11,13) and jumping (12) have been estimated by using biomechanical models, but the forces that occur during running have not been investigated.

The purposes of this study were 1) to develop a biomechanical model of the ankle joint useful in predicting joint and muscle forces during the stance phase of gait, 2) to use this model to predict the forces that occur during the stance phase of running, and 3) to determine the effects of changes in some of the assumptions of the model on the forces predicted by the model.

METHODS

Model. At any instant in time the foot, consisting of the tarsals, metatarsals, and phalanges, and the leg, consisting of the tibia and fibula, were assumed to be rigid bodies. The ankle joint was assumed to be a uniaxial joint joining these two rigid bodies. Motion at this joint actually occurs about an axis that changes position and direction with ankle motion, but it can be approximated as occurring about a fixed axis passing less than 1 cm distal to the tips of the medial and lateral malleoli and less than 1 cm anterior to the tip of the lateral malleolus (6). However, this axis of rotation passes distal to the actual surface of contact between the talus and the tibia (4). Therefore, contact forces between the two rigid bodies of this model were assumed to occur superior to this axis of rotation at a point mid-way along a line between the most prominent parts of the medial and lateral malleoli. These reference points were used because they closely approximate the position of the superior surface of the talus (4) and because they are easily located surface landmarks. Other forces that can occur on the foot include forces from muscles exerted through tendons, ligamentous and soft tissue forces, and external forces caused by the environment. Ligaments and soft tissues can cause significant forces when stretched. Nevertheless, to decrease the number of unknown forces in the model, forces in these structures were assumed to

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be small enough to neglect during movements within normal limits of ankle motion. The assumption that forces carried by ligaments can be neglected has also been made in other joint models to decrease the number of unknown forces (9,12,13).

There are 11 muscles that cross the ankle joint and insert into the foot, excluding the plantaris which can be neglected because of its small size. To simplify the model, these muscles were assigned to five functional groups on the basis of their proximity to each other and the types of motion each can cause at the foot. These five equivalent muscles and their individual members are 1) the plantar flexion group: gastrocnemius and soleus; 2) the plantar flexion and inversion group: tibialis posterior, flexor hallucis longus, and flexor digitorum longus; 3) the plantar flexion and eversion group: peroneus longus and peroneus brevis; 4) the dorsiflexion and inversion group: tibialis anterior and extensor hallucis longus; and 5) the dorsiflexion and eversion group: peroneus tertius and extensor digitorum longus. This technique of describing equivalent muscles has been used in many joint models (5,8,9,15) to reduce the number of unknown forces in the model and to insure that muscles known to function together from electromyographic evidence are predicted to act together by the model.

Leg and foot coordinate systems were defined so that the directions of pull and the insertion points of tendons could be measured with respect to these coordinate systems (Figure 1). The foot coordinate system was defined with its origin at the mid-point of a line joining the most prominent parts of the malleoli. Its Y-axis was directed anteriorly along the mid-line of the foot and parallel to the plantar surface of the foot. The Z-axis was directed superiorly, was within a plane formed by the Y-axis and a line bisecting the calcaneus, and was perpendicular to the Y-axis. The X-axis was directed laterally and was perpendicular to these two axes. The leg coordinate system was defined with the same origin as the foot coordinate system. Its Z-axis was directed superiorly along the longitudinal axis of the leg. Its Y-axis was directed anteriorly, was perpendicular to the Z-axis, and was within the sagittal plane. The X-axis was perpendicular to the Y- and Z-axes and was directed laterally.

To obtain the anatomical data necessary for formation of the model, a leg, ankle, and foot of five cadavers were dissected to expose the 11 ankle muscles and their tendons. The foot and leg of each cadaver were then positioned so that the coordinate systems defined for the foot and leg were parallel. Photographs of the medial, lateral, anterior, and posterior aspects of the ankle were taken. Measurements of the angle between the shank and the Achilles tendon of each cadaver were made; this angle was less than 10° for each cadaver throughout the range of plantar flexion and dorsiflexion. Therefore, the gastrocnemius and soleus were assumed to pull from the Achilles tendon attachment parallel to the longitudinal axis of the shank.

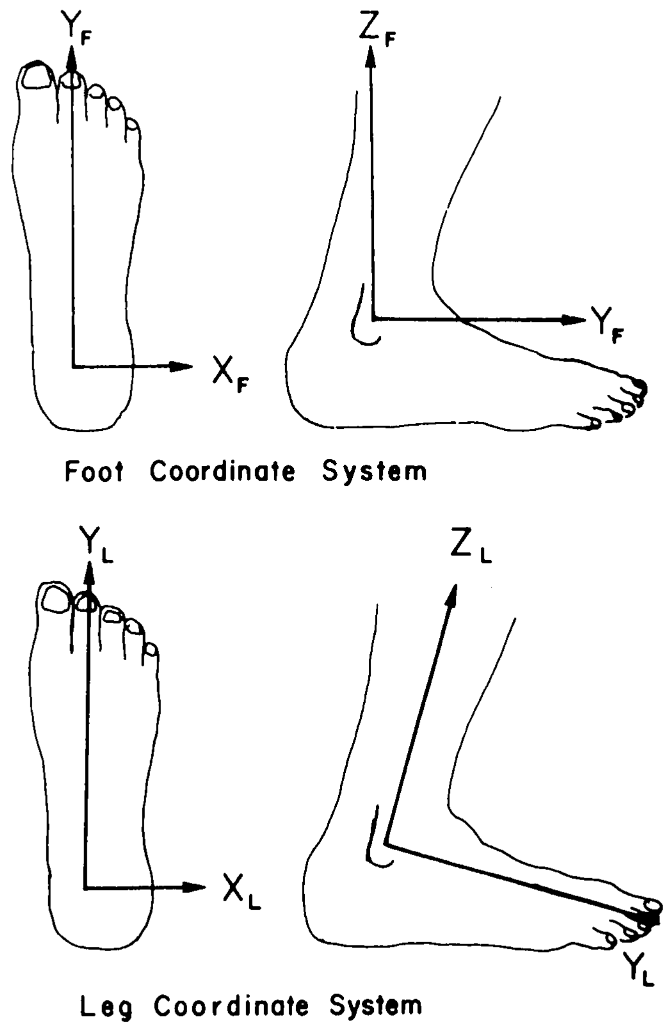

Figure 1—Leg and foot coordinate systems.

The directions of pull of the other muscles were assumed to exist between the exit of each tendon from either the flexor or extensor retinaculum just above the ankle and the entrance of each into its retinaculum at the foot or its insertion point into the foot. The three-dimensional coordinates of these exit and entrance points of the tendons and the positions of the malleoli were then measured on enlarged photographs of the four aspects of the ankle. These coordinates were divided by foot dimensions for normalization and were transformed to leg and foot coordinate systems, respectively. The X-coordinates were divided by the medial-lateral distance between malleoli, the Y-coordinates were divided by foot length, and the Z-coordinates were divided by the vertical distance between the plantar surface of the foot and the origin of the coordinate systems. Table 1 shows both the non-dimensionalized coordinates of each tendon at the point it leaves the leg and at the point it inserts into the foot and the non-dimensionalized coordinates of the medial and lateral malleoli with respect to the leg coordinate

TABLE 1. Non-dimensionalized coordinates of cadaver tendons and malleoli.

Structure	Coordinates X	Y	Z
		Insertion into Foot	
Tibialis Anterior	−0.240(0.058)	0.174(0.023)	−0.054(0.082)
Extensor Hallucis Longus	−0.033(0.104)	0.158(0.021)	0.145(0.062)
Extensor Digitorum Longus	0.152(0.110)	0.144(0.027)	0.131(0.078)
Peroneus Tertius	0.247(0.079)	0.131(0.030)	0.073(0.030)
Tibialis Posterior	−0.391(0.050)	0.122(0.033)	−0.429(0.055)
Flexor Hallucis Longus	−0.197(0.031)	−0.069(0.022)	−0.222(0.027)
Flexor Digitorum Longus	−0.355(0.026)	0.098(0.031)	−0.476(0.094)
Peroneus Brevis	0.378(0.047)	0.138(0.099)	−0.522(0.126)
Peroneus Longus	0.370(0.038)	0.115(0.103)	−0.572(0.123)
Achilles Tendon	−0.027(0.028)	−0.161(0.045)	0(0)
		Exit from Leg	
Tibialis Anterior	−0.148(0.078)	0.131(0.014)	0.286(0.082)
Extensor Hallucis Longus	0.022(0.107)	0.123(0.027)	0.359(0.075)
Extensor Digitorum Longus	0.170(0.124)	0.123(0.027)	0.357(0.081)
Peroneus Tertius	0.259(0.086)	0.111(0.032)	0.266(0.094)
Tibialis Posterior	−0.437(0.023)	0.034(0.031)	−0.209(0.083)
Flexor Hallucis Longus	−0.095(0.063)	−0.093(0.028)	0.116(0.087)
Flexor Digitorum Longus	−0.394(0.055)	0.011(0.018)	−0.285(0.127)
Peroneus Brevis	0.406(0.037)	0.017(0.050)	−0.283(0.124)
Peroneus Longus	0.385(0.037)	−0.001(0.053)	−0.315(0.159)
Medial Malleolus	−0.500(0.000)	0.026(0.016)	0.029(0.019)
Lateral Malleolus	0.500(0.000)	−0.026(0.016)	−0.029(0.019)

Values are means (SD), N=5.

system when the two coordinate systems were parallel. The coordinates of these points were assumed to remain fixed within the leg or foot during ankle movements. The action line of each tendon was assumed to be the line formed by the entrance and exit points of that tendon.

To calculate the lever arm length and the direction of pull of each equivalent tendon for the model, a method of evaluating the force contribution that each individual muscle made to its equivalent muscle was devised. Joint models have generally used either muscle mass (5,15) or anatomical cross-sectional area (9,11) as the reference criterion. In this study the reference criterion used was the physiological cross-sectional area, which was defined as the volume of a muscle divided by the average length of its muscle fibers. Since most fibers do not extend the entire length of the muscle, this method gave a better indication of the number of fibers contributing to the force of a muscle than did the anatomical cross-sectional area.

After dissection, the individual muscles of each cadaver were weighed and their volumes were determined by water immersion. Each muscle was divided into five regions of equal length, and the length of two fibers from each region were measured. These lengths were then used to calculate the average muscle fiber length for that muscle. The physiological cross-sectional areas were normalized by dividing the physiological cross-section of the gastrocnemius. There were variations in the angle of insertion of individual fibers into tendons, both within and between muscles. Because of the variation in angle of pull of the fibers within a muscle, it was difficult to determine the average angle of pull of fibers for some muscles. Therefore, the effect of differences in angle of pull of fibers between muscles were not included in the physiological cross-sectional area. The normalized coordinates and physiological cross-sectional areas obtained from the five cadavers were used to calculate average normalized insertion and exit coordinates of each of the 11 tendons and the average normalized cross-sectional area of each of the 11 muscles for use in the model.

Application. This model was used to estimate the joint forces during the stance phase in the right ankles of three young-adult male subjects running at approximately 4.47 m/s (a 6-min mile pace). Informed consent was obtained from these subjects.

As the subjects ran across a force platform, they were filmed at 100 frames/s with two 16-mm Locam cameras, one with the film plane located parallel to and lateral to the sagittal plane of the runner and one with the film plane located parallel to and behind the frontal plane. A photoelectric timer was used to obtain a measure of the average velocity of each runner. Several practice runs were made by each subject until each achieved the proper speed with a right foot strike on the force platform.

A Bendix film analyzer was used to obtain the coordinates of specific points on the leg and foot of each runner with respect to the center of the force platform. The coordinates of the lateral malleolus, the head of the fibula, and two points along the lateral border of the shoe posterior to the metatarsal-phalangeal joints were measured in the plane parallel to the direction of travel from the lateral film. The coordinates of the medial and lateral malleoli, the base of the heel, and two points along the posterior mid-line of the shank were measured in a plane perpendicular to the plane of travel from the posterior film. A Kistler force platform sampled the three components of force and moment acting on the foot from the floor at a rate of 500 Hz. Two reference points, coated with chalk dust, were placed on the bottom of each runner's right shoe along the mid-line so that the angle of the foot, with respect to the direction of the run, could be determined from the position of these markers on the force platform. All digitized coordinates were smoothed by a three-point moving average. Since no differentiation of the data was performed, more complicated smoothing techniques were not used. Stance time was defined for the force platform data as being the interval during which a vertical force of greater than 50 N was exerted. For the film data, stance time was defined as the interval during which the foot was in contact with the platform. Since the filming was done at 100 frames/s, differences of up to 10 ms could have existed between the initiation of stance as determined from film and from the force platform. To decrease the errors caused by this possible asychronism between force and position data, each stance time was divided into 100 equal intervals, and the coordinates of each digitized point and the magnitude of the ground reaction forces and moments were found by linear interpolation for each interval.

Since the anatomical data used in developing this model were three-dimensional, a three-dimensional analysis of the motion of the foot and leg should be done during application of the model to utilize its capabilities fully. However, since only two cameras were used in the present study, all points needed to define the coordinate systems of the foot and leg were not in view in both cameras. Therefore, a true three-dimensional analysis of the motion could not be made.

Instead, so that the positions of the two coordinate systems in space could be calculated, it was assumed that 1) the Y-axis of the leg remained within a parasagittal plane and 2) the projection of the Y-axis of the foot on the transverse plane remained parallel to the line formed by the two chalk marks left on the force platform at foot contact. This second assumption was valid since little or no rotation of the foot about an axis perpendicular to the force platform surface occurred during stance. For the first assumption, concerning the Y-axis of the leg, to be valid, little rotation of the leg about its longitudinal axis should occur during stance. During stance phase, rotation of the leg about its longitudinal axis is coupled with inversion and eversion movements of the foot at the subtalar joint (6). Therefore, the amount of transverse rotation of the leg should be indicated by the amount of inversion and eversion that occur during stance phase. In this study, during the period from 15-75% of stance time, the angle of inversion or eversion of the foot was measured from the films to be less than 5° for each of the three runners. Therefore, the assumption that the Y-axis of the leg remained in a parasagittal plane was held to be valid for this portion of the stance. During early and late stance, this assumption was less valid since inversion and eversion angles of up to 20° were measured from the films for these periods of stance.

The directions of the axes of the leg and foot coordinate systems were then calculated from the following information: 1) the projection of the Y-axis of the foot on the sagittal plane was parallel to the line formed by the two points on the lateral border of the shoe measured from the lateral film, and its projection in the transverse plane was parallel to the line between the two chalk marks on the force platform; 2) the projection of the Z-axis of the foot on the sagittal plane was perpendicular to the Y-axis projection on the sagittal plane, and its projection on the frontal plane was parallel to a line bisecting the heel measured from the rear film; 3) the X-axis of the foot was perpendicular to both of the other axes of the foot; 4) the projection of the Z-axis of the leg on the sagittal plane was parallel to the line between the lateral malleolus and the head of the fibula measured from the lateral film, and its projection in the frontal plane was parallel to the line along the posterior mid-line of the leg measured from the rear film; 5) the Y-axis of the leg was perpendicular to the Z-axis of the leg, and it was within the sagittal plane; and 6) the X-axis of the leg was perpendicular to both of the other axes of the leg.

The origin of these two coordinate systems was also calculated from the two films. The coordinates of this point within the frontal plane were determined from the rear film. These coordinates were calculated to be located at the mid-point of a line between malleoli. The anterior-posterior coordinate of the lateral malleolus was measured from the lateral film, and then one-half the anterior-posterior distance between malleoli was added to this coordinate along the Y-axis of the leg to obtain the anterior-posterior coordinate of the origin.

When the relative directions of the axes of the foot and leg coordinate systems were known as functions of time, the direction of pull and lever arm lengths of each equivalent muscle group were calculated as functions of time with respect to the leg coordinate system. The dimensions of the foot of each subject were used to convert from normalized coordinates to actual coordinates. Then the equivalent tendon directions of pull were found by 1) calculating unit vectors in the directions of the individual tendons making up the group, 2) multiplying each unit vector by the weighting factor for that muscle, 3) adding these vectors together, and 4) calculating direction cosines of the resultant vector. The lever-arm lengths of each equivalent tendon were found by calculating the perpendicular distances from the joint center to the lines representing the equivalent tendons. The forces and moments measured by the force platform with respect to its center were transformed to the coordinate system of the leg.

During the stance phase of most activities, the foot is usually experiencing only small accelerations. Because the foot also has a small mass, inertial and gravitational forces can be neglected. Therefore, the foot in this model was assumed to be in equilibrium at any instant in time so that the equations of equilibrium could be applied to the model. Because of the inaccuracies in the modeling process, it was held that moments about the longitudinal axis could not be accurately balanced. Therefore, the equation for the moments about the longitudinal axis was not used. Five equations remained to be solved for eight possible unknown forces, the five possible equivalent tendon forces and the three components of the joint force.

It was mathematically possible to balance the external moments and forces acting on the foot with only two tendon forces in this model. However, it was physiologically possible for more than two tendons to exert force. This would have resulted in a greater joint force than if only two tendons had been exerting force. Therefore, the minimum joint force possible for a given movement was determined by assuming that only two tendons were exerting force at any instant. This then reduced the number of unknowns to five, and the equations could then be solved. Ten possible solutions were obtained by letting all combinations of five tendons, taken two at a time, be the only tendons exerting force. Some of these possible solutions

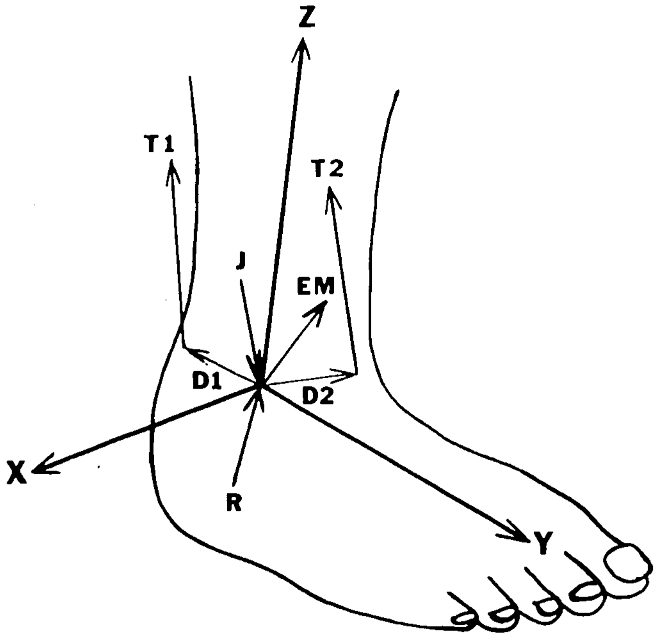

Figure 2—Free body diagram of foot.

were eliminated because negative values were obtained for the tendon forces. The remaining solutions were compared to determine which solution yielded the lowest bone-on-bone joint force.

The vector equations used to solve for the muscle and joint forces were the following:

$$\Sigma F = 0 = T1 + T2 + J + R \qquad [1]$$

$$\Sigma M = 0 = D1xT1 + D2xT2 + EM \qquad [2]$$

where T1 and T2 are two tendon forces, J is the resultant joint force on the foot, R and EM are the resultant external force and moment on the foot transformed to the leg coordinate system, and D1 and D2 are the position vectors of the points of application of T1 and T2 with respect to the leg coordinate system. Figure 2 shows a free-body diagram of the foot with these forces and moments acting.

To determine the effects of variations of some of the major assumptions of the model, the data collected from one of the subjects were used in various modified models, and the results were compared to the forces predicted by the original model. As each modification was tested, all other assumptions of the original model were held constant. These modifications were 1) the use of a two-dimensional rather than three-dimensional model, 2) the variation of the assumed joint center 1 cm in various directions, 3) the use of anatomical data from one cadaver rather than five, and 4) the weighting of the contribution of each individual muscle to its equivalent muscle by muscle mass or by an equal weighting rather than by physiological cross-sectional area.

RESULTS

All the forces predicted by this model were calculated with respect to the coordinate system moving with the leg. The three components of the joint force predicted by the model for the three subjects during the stance phase of running are shown in Figure 3. The compressive forces on the foot along the longitudinal axis of the leg reached maximum values during mid-stance of between 8 and 13 times body weight for the three subjects. The anterior shear force on the foot in a direction perpendicular to the longitudinal axis of the leg reached peak values of between 3.3 and 5.5 times body weight, and medial-lateral shear forces ranged from a medial force of 0.8 times body weight to a lateral force of 0.5 times body weight.

Figure 4 shows the forces predicted in the equivalent tendons for the three subjects. The largest forces were predicted to occur in the plantar flexion group, consisting of the gastrocnemius and soleus muscles. The peak forces in this group ranged from 5.3 times body weight in Subject 2 to 10 times body weight in Subject 1. The plantar flexion and inversion group, consisting of the tibialis posterior, flexor digitorum longus, and flexor halucis longus, also was predicted to exert large forces, ranging from 4 times body weight in Subject 1 to 5.3 times body weight in Subject 2. One muscle group, the dorsiflexion and eversion group, consisting of the peroneus tertius and extensor digitorum longus, was predicted to exert no force during stance for any of the subjects.

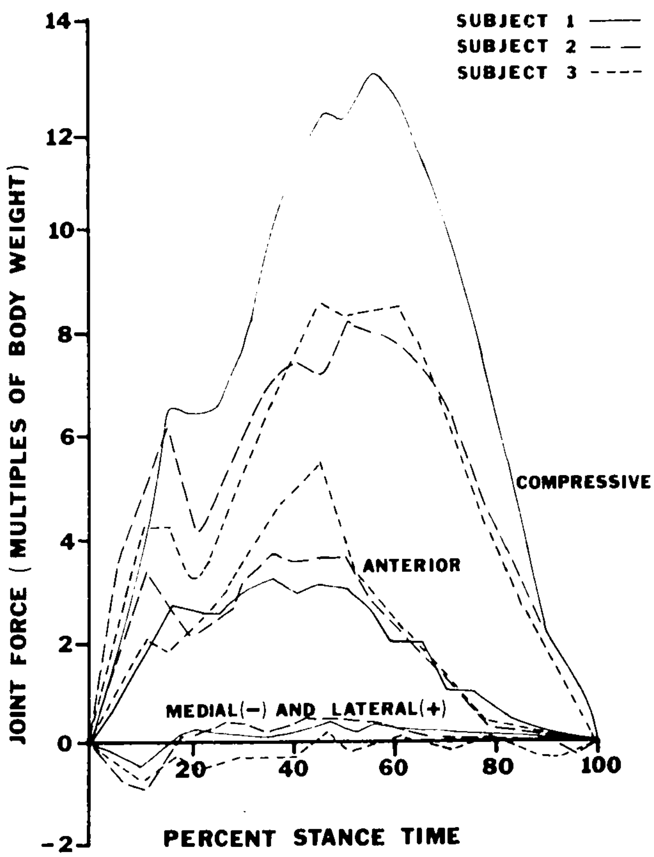

Figure 3—Three components of joint force predicted by the model.

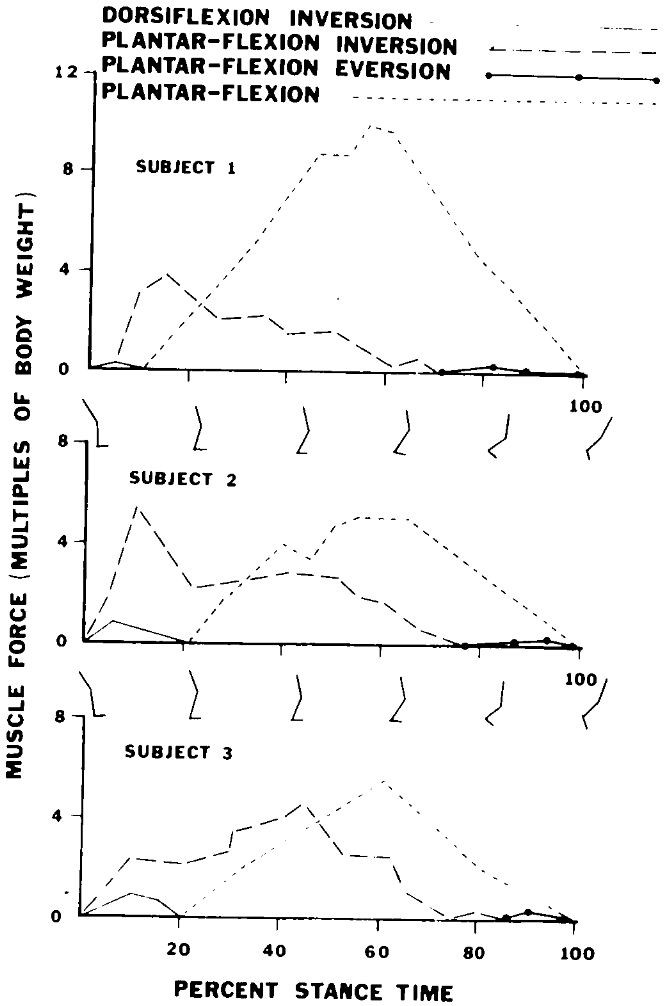

Figure 4—Equivalent muscle forces predicted by the model.

Using the data obtained for Subject 1, a two-dimensional model based on the same anatomical data and assumptions as the original three-dimensional model was used to predict forces. Figure 5 shows a comparison of the joint forces predicted by the two models. The compressive forces predicted by the two models were similar. However, the shear force predicted by the three-dimensional model was somewhat higher than that predicted by the two-dimensional model. By the nature of the two-dimensional model, no medial or lateral forces could be predicted. Also, only one equivalent muscle group at a time could be predicted to exert force with the two-dimensional model. This resulted in the predicted muscle forces shown in Figure 6, which are similar to those predicted by the three-dimensional model but with only one group active at a time.

The effects on the predicted forces of 1-cm differences in the assumed position of the joint forces are shown in Figure 7. Placement of the joint center 1 cm medially resulted in a very large peak in the predicted absolute joint force of almost 22 times body weight, while placement

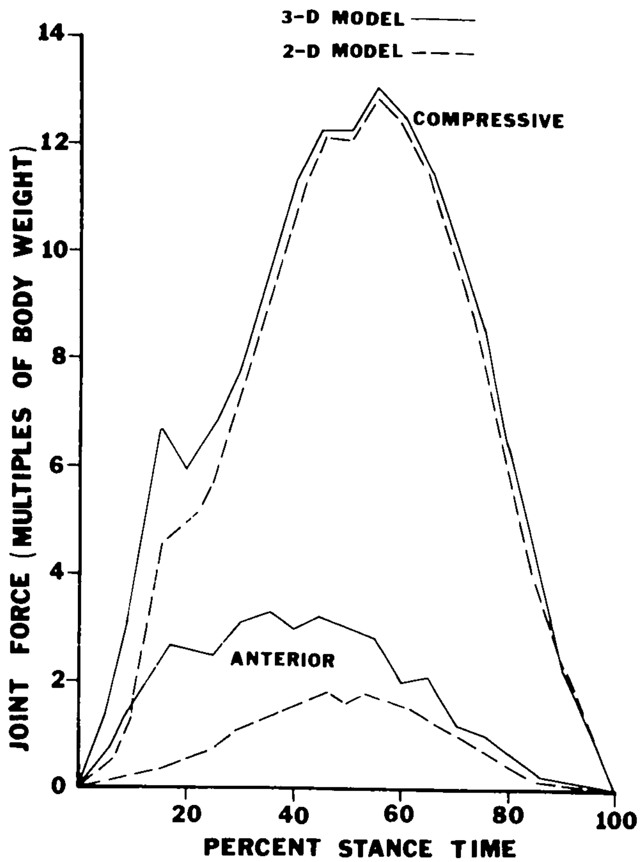

Figure 5—Comparison of joint forces predicted by the three-dimensional and two-dimensional models for Subject 1.

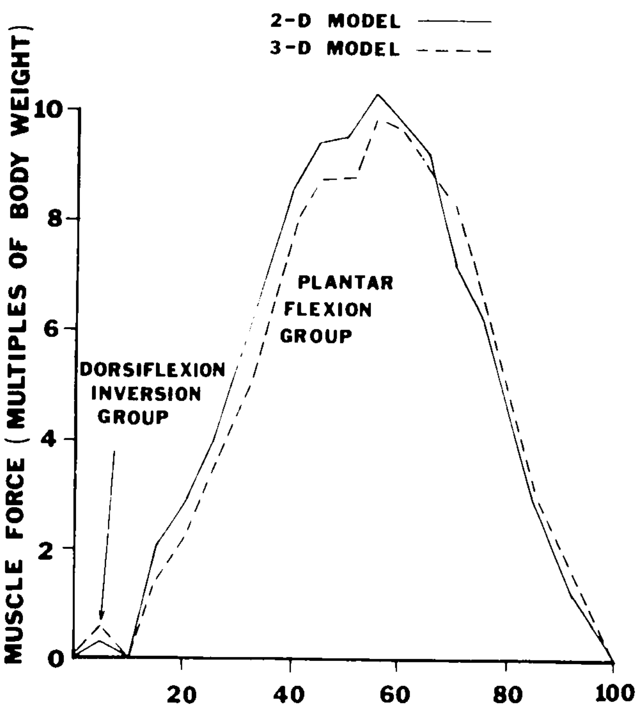

Figure 6—Comparison of muscle forces predicted by the three-dimensional and two-dimensional models for Subject 1.

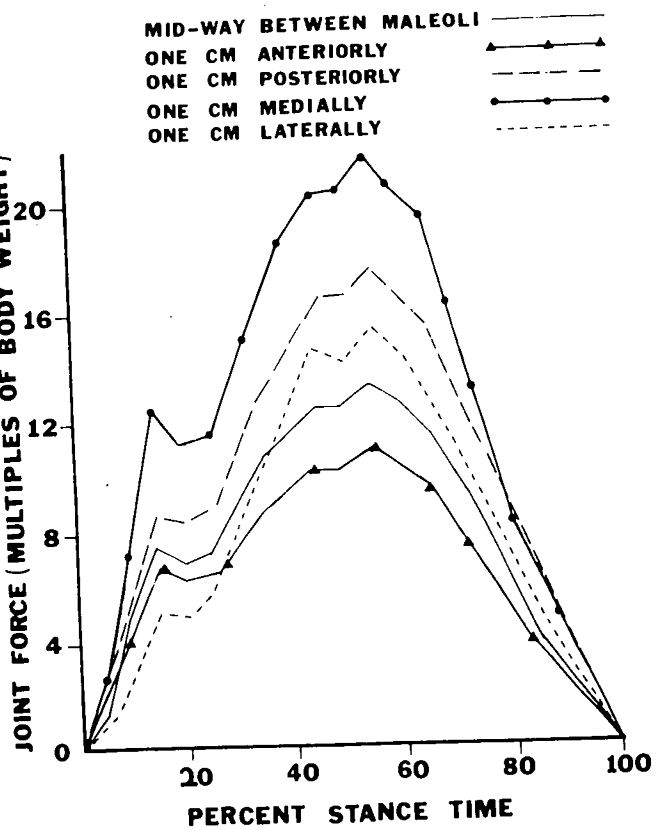

Figure 7—Effects of changes in the assumed ankle joint center on the absolute joint force of Subject 1.

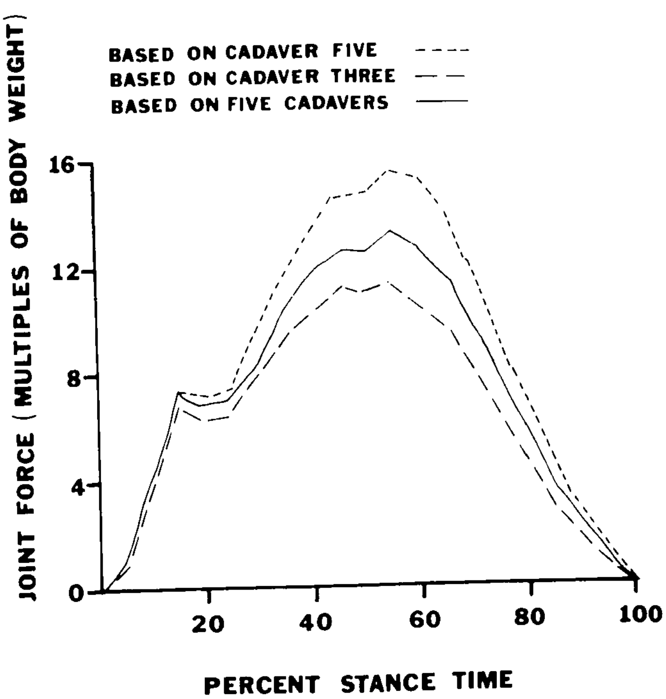

Figure 8—Effects of using different cadavers for modeling purposes on the absolute force of Subject 1.

of the joint center 1 cm anteriorly resulted in a decrease in this peak force to about 11 times body weight. Other changes of 1 cm resulted in forces within these extremes.

Using the anatomical data from one cadaver at a time, rather than the average of all five cadavers, resulted in the range of predicted joint forces shown in Figure 8. Peak absolute joint forces ranged from 11 times body weight when using one particular cadaver to 15.5 times body weight when another was used.

The use of three different methods of weighting the contribution of each individual muscle to the equivalent muscles resulted in only small differences in the joint and muscle forces predicted by the model. These three methods were 1) by physiological cross-sectional area, 2) by muscle mass, and 3) by equal weighting. This occurred because muscles within an equivalent muscle group share similar insertion points and lines of action. The insertion point and action line of an equivalent tendon formed from tendons with similar insertions and action lines are not very sensitive to changes in weighting methods.

DISCUSSION

For this model to be useful, it must be shown to produce valid results. Validation of joint models, however, is difficult because of the problem of directly measuring the joint and muscle forces. Indeed, joint models are usually developed because of this problem of obtaining measurements of forces internal to the body.

Previous ankle joint models, which had been applied to walking (11,13), predicted maximum compressive joint forces of approximately five times body weight, as compared to the maximum compressive joint forces of 10-13 times body weight predicted by this model for running at 4.47 m/s. Much of the difference between the force values predicted during walking and running are probably due to the differences in magnitude of the ground reaction forces in the two gait modes. Walking usually produced vertical ground reaction forces of approximately 1.0-1.2 times body weight (7), while moderately paced running produced vertical reaction forces of 2.5-3.2 times body weight in this study. This ratio of approximately 2.5 of vertical reaction forces during running to those during walking is approximately equal to the ratio of the various forces predicted during the two gait modes.

The center of pressure of the ground reaction force on the stance foot during running has been shown to progress from the posterior-lateral or medial-lateral border of the foot at foot strike to the anterior mid-line of the foot during mid-stance and push-off (2). This center of pressure progression tends to produce external moments about the ankle that are predominantly in the direction of eversion and dorsiflexion. Therefore, the internal moments exerted by the muscles to balance this external moment should be predominantly in the direction of inversion and plantar flexion.

The equivalent muscle forces predicted by this model, shown in Figure 4, were consistent with this pattern of the position of the ground reaction forces. Initially, a short low-level force was predicted in the dorsiflexion and inversion group of muscles for all three runners. Simultaneously, the plantar flexion and inversion group was predicted to exert a relatively high force. This was consistent with a foot strike with the center of pressure of the reaction forces on the lateral border of the foot, which would produce an external eversion moment. The relative amount of force exerted by the dorsiflexion and inversion group compared to the plantar flexion and inversion group depends on the anterior-posterior position of the center of pressure at foot strike. Rear-foot strikers would require a larger relative dorsiflexion muscle force than mid-foot strikers. During mid-stance all three runners were predicted to have high plantar flexion and plantar flexion and inversion muscle forces. This was consistent with a center of pressure that moved anteriorly but still remained lateral to the mid-line of the foot. During late stance, all three runners were predicted to have decreased plantar flexion muscle forces and low levels of plantar flexion and eversion muscle forces. This was consistent with a center of pressure that moved slightly medial to the mid-line of the foot but which had decreased in magnitude. The dorsiflexion and eversion muscle group was predicted to exert no force during stance phase for any of the runners because the position of the center of pressure of the reaction force was such that no external moments in the direction of plantar flexion and inversion were exerted on the foot during stance.

The relative magnitudes of the forces predicted for the three runners were also consistent with the relative magnitudes of the ground reaction forces exerted on the feet of the runners. The peak vertical reaction force for Subject 1, who was predicted to have the highest joint and muscle forces, was 3.2 times body weight, while it was only 2.5 and 2.6 times body weight for the other two subjects. While these comparisons do not provide proof of the validity of the model, they at least show the results to be consistent within itself, with other ankle joint models, and with the normal progression of the center of pressure of the ground reaction forces.

Another method for testing the validity of the forces predicted by this model is to compare them to the forces found experimentally to cause injuries. A compressive joint force of 13 times body weight, or 8678 N for Subject 1, seems very high, but it is only 25% of the compressive breaking load of the shaft of the tibia found experimentally (16). The maximum muscle force predicted by this model was approximately 10 times body weight or 6680 N for the plantar flexion group of Subject 1. Experimental values for the ultimate tensile strength of human tendons range from 34.0-147 MPa (1,2,14,16). If a circular cross-sectional area with a diameter of 1 cm is assumed for the Achilles tendon of Subject 1, the force of 6680 N would cause a stress within this range of stresses found experimentally to cause damage to tendons.

This comparison to find ultimate strengths experimentally seems to indicate that the model predicts tendon forces that could result in injury during a moderate running pace. However, these ultimate strengths have been found in cadaveric specimens loaded under static conditions. The stress produced in biological materials has been shown to be dependent on the rate at which the load is applied (1). Running is a dynamic situation in which the load is applied rapidly and for a short duration. Also, cadaveric specimens may exhibit different properties than living tissues. Therefore, it is possible that the actual stresses associated with bone and tendon injury could be much higher than the experimental values reported. It is also possible that the many simplifications and assumptions introduced into the model have resulted in the prediction of invalid forces. On the basis of this comparison between predicted forces and forces shown to cause injury, no final conclusion can be reached concerning the validity of the model.

This joint model and joint models in general suffer from several fundamental limitations that may affect their validity. These include 1) the limitation of the possible forces to make the solution determinate, 2) the accuracy of the location of the position at which the joint forces act, and 3) the accuracy of the location of insertion points and action lines of tendon and ligament forces. Also, separate from the modeling process itself, the precision of the data collected during motion and force analysis can critically affect the validity of the forces predicted by the model.

Even though the assumption was made that only tendon and bone-against-bone forces were significant, it is possible that the medial and lateral collateral ligaments of the ankle could exert significant forces during the stance phase of running. Because of their location near the plantar flexion and dorsiflexion axis of motion, forces in these ligaments would have little effect on the forces predicted for the plantar flexion muscle group by this model. However, they could share some of the load of the invertor or evertor muscle groups. This would make the predicted forces in these muscle groups higher than the actual forces. The effect that significant ligament forces would have on the joint force is more difficult to predict since the lever arms and action lines of the ligaments have not been defined.

The large variations in predicted forces that occurred when the assumed point of application of the joint forces was varied one centimeter in various directions showed that the accurate location of this point is important in developing a valid model. However, differences of one centimeter between the actual location and assumed location of this point are likely to occur, both because of the limitations in accuracy of film analysis and other methods of obtaining the position of external anatomical landmarks and because of lack of knowledge concerning the position of the joint force with respect to any external or internal landmarks.

Large differences in predicted forces also occurred when individual cadavers were used for the anatomical data of the model rather than the average of the five cadavers. It seems likely that subjects could differ as much in their relative tendon insertion and exit points as did the cadavers. Therefore, differences between actual and predicted forces for a subject could be as large as those obtained when using different cadavers. This shows the importance of basing the model on the body dimensions of the subject being studied rather than text-book or average values. In order to use tendon insertions and action lines of individual subjects, some method must be developed to determine these internal measurements accurately for each subject tested.

The differences between the joint forces and Achilles tendon forces predicted by the two- and three-dimensional models were small. This seems to indicate that a two-dimensional model may be just as accurate as a three-dimensional model in predicting joint forces and the forces within the plantar flexion muscle groups, at least for normal subjects running in a constant direction. However, for activities in which significant motion occurs out of a single plane, a three-dimensional model may be necessary.

This model utilized many assumptions common to joint models. Small changes in some of these assumptions were shown to produce large changes in the forces predicted by the model. Because of the many assumptions used to develop joint models and because of the limitations in the accuracy of measurement systems, the forces predicted by this model and other joint models can only be viewed as order of magnitude estimates of actual forces. In any future joint models, attempts should be made to obtain direct validation of the model. This could be done for human or animal joint models by using instrumented artificial joints to measure joint forces. For animal joint models, force transducers could be placed in series with a tendon to measure the tendon forces directly. Then the effects that various assumptions have on the validity of the model could be directly examined by comparing predicted forces to measured forces. Until this is done, the validity of this and any other joint model remains unknown.

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