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Human Patellar-Tendon Rupture

A KINETIC ANALYSIS*

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ABSTRACT: **The first biomechanical analysis of a human patellar-tendon rupture during actual sports competition is reported. Cinematographic data for analysis were collected at a national weight-lifting championship. Dynamic equations to mathematically model the lifter were developed to compute time course and magnitudes of hip, knee, and ankle-joint moments of force and of tensile loading of the patellar tendon before and during tendon trauma. Results provided evidence that the range of maximum tensile stress of the tendon may be considerably greater during rapid dynamic loading conditions, as in many sports situations, than maximum tensile stress obtained during static test conditions.**

Injuries to ligaments and tendons occur when those structures are subjected to rapidly applied loads of high magnitude [18], but little is known about the magnitudes of loads or loading rates during actual injuries in humans. It generally is not feasible to obtain any useful data during the ordinary course of human activity, and it would be unconscionable to allow human subjects to approach maximum loading conditions experimentally. An unusual opportunity arose for us when we were collecting cinematographic data on weight-lifters for the analysis of joint forces and moments of force. We observed a human patellar-tendon failure during actual sports competition. The purpose of this study is to report the time course and magnitude of knee-joint moments of force and patellar-tendon tensile loading before and during the failure of that patellar tendon.

Methods

Cinematographic data were collected for all weight classes at the 1975 U.S.A. National Weightlifting Championships. The subject of this study was a twenty-nine-year-old man who competed in the light heavyweight division. He then weighed 82.2 kilograms and was of world-class caliber. He had won the championship of his weight class just prior to the attempt in which his right patellar tendon ruptured. During subsequent surgical repair it was found that the patellar tendon was attenuated throughout its course. Portions of its mass had pulled out both from the distal pole of the patella and from the patellar-tendon insertion on the tibia.

The subject had no previous history of injury to the right knee.

A camera had been positioned ten meters to the right of the geometric center of the competition platform and perpendicular to the lifter's plane of motion. The platform was four meters square and the optical axis of the camera passed one meter above the center of the platform. The motor-driven camera was set at fifty frames per second.

Serial film images were projected onto a digitizer with a measurement rounding error of 127 micrometers. Digitized rectangular coordinates were available for the center of gravity of the lifted weight; for positions of the lifter's right hip, knee, ankle, and fifth metatarsophalangeal joint; and for the top of the lifter's head. These coordinates were transferred directly to digital cassette tapes and computer programs were written for subsequent analyses which included the calculation of center-of-gravity locations [5] and segmental inclinations for each twenty-millisecond time interval in the analysis. A least-squares five-point moving arc technique was used when time derivatives of linear or angular position data were required [22,24,25,36].

Film-derived kinematic segmental linear and angular accelerations and mass parameter estimates [5] were incorporated into the kinetic equations of motion for a mathematical model of the lifter. The lifter was modeled in two dimensions as a five-link rigid-body system with the assumption of symmetry about the cardinal sagittal plane. The five constituent links were: (1) a point mass at the center of gravity of the weight being lifted, which was

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connected by a massless rigid link to the hip joint; (2) a rigid body consisting of the head, trunk, and upper extremities; (3) the thigh segment; (4) the leg segment; and (5) the foot segment. A schematic representation of the model and a coordinate reference system are shown in Figure 1. The segments were assumed to be connected by

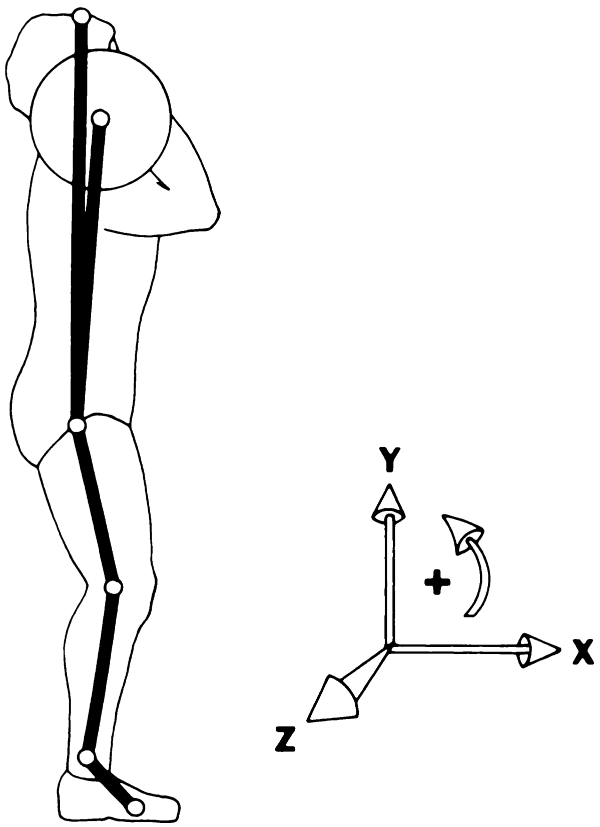

Fig. 1

Schematic representation of the rigid-body model. The lifter was modeled as a planar five-link system. The five links included: (1) a point mass at the center of gravity of the weight; (2) a rigid head, trunk, and upper-extremities segment; (3) the thigh segment; (4) the leg segment; and (5) the foot segment. The rectangular coordinate and moment-of-force reference systems are defined.

hinge joints and friction at the joints was assumed to be negligible[3,20]. The net forces and moments of force at each joint were calculated from the equations of Newtonian rigid-body dynamics. The general equations of motion consisted of two force equations, $\Sigma F_X = m\ddot{x}$ and $\Sigma F_y = m\ddot{y}$, and one moment-of-force equation[16], $\Sigma M_z = I_Z \alpha_Z$. For each rigid body, the sum of the forces, either horizontal (F_x) or vertical (F_y), was equivalent to the product of the segment mass (m) and the corresponding horizontal ($\ddot{x}$) or vertical ($\ddot{y}$) acceleration of that segment's center of mass. Similarly, the sum of the moments of force (M_z) orthogonal to the plane of the rigid-body motion was equal to the product of the segment's angular acceleration (α_z) and the appropriate mass moment of inertia (I_z) at the segment's center of mass. Detailed derivations of the kinetic equations of motion used in this study were presented elsewhere[9,16].

Validation of the model and computational techniques was achieved through an analysis of an experienced weight-lifter executing the same movement under similar technical conditions while standing on a force platform (Kistler 9261A). The resultant vertical reaction forces at the foot, computed using the five-link rigid-body system, were compared with those same forces measured directly with the force platform. The mean agreement between the independent techniques was greater than 93.0 per cent. The absolute and relative reliabilities of calculated joint moments of force were determined for the modeling techniques employed in this study. The same film was redigitized in three months. The correlation between knee-joint moments of force from the original and redigitized data was 0.9. In absolute terms, the average standard error of the mean for an estimate of any moment of force at the knee joint was 24.1 newton-meters.

The moments of force at the knee joint, determined from the mathematical model, were used to calculate tensile loads in the patellar tendon during the lift. Patellar-tendon tension was defined as the quotient of the net knee-joint moment of force and the perpendicular distance from the instant axis of rotation at the knee joint to the line of action of an idealized force vector colinear with the patellar tendon (Fig. 2). Because the injury was unanticipated, direct roentgenographic measurements of requisite moment-arm distances were not available. Smidt[30], however, did report such data, collected using a roentgenographic method similar to that developed by Frankel and Burstein[8]. All of Smidt's measurements were obtained for

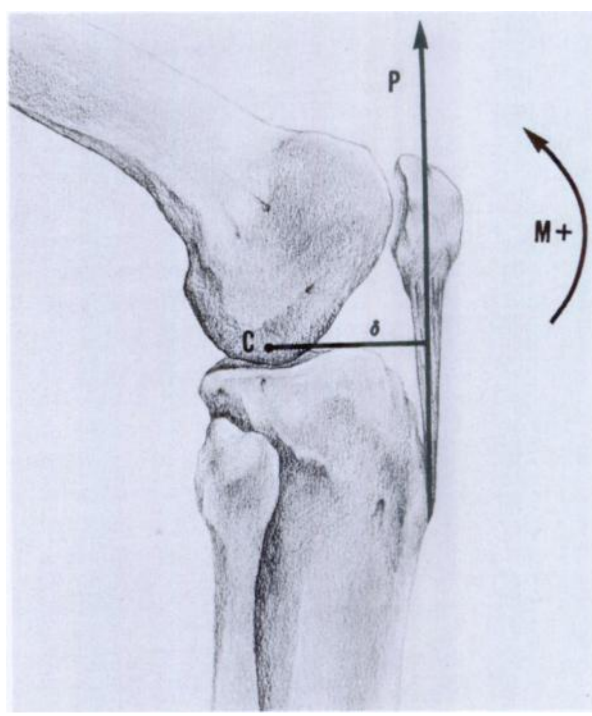

Fig. 2

Method of calculation of patellar-tendon load. The load on the patellar tendon was defined as the quotient of the net knee-joint moment of force (M) and the perpendicular distance (δ) from the instant center (C) of rotation at the knee joint to the line of action of an idealized force vector (P) colinear with the patellar tendon.

the right lower extremity. It is significant to note that his sample consisted of data from twenty-six normal men whose average age was twenty-eight years; average height, 176 centimeters; and average weight, eighty-two kilograms [30]. The lifter in this study was the same height and was only one year older and 0.2 kilogram heavier than Smidt's population means. Therefore, we assumed that the published values of patellar-tendon moment arms were applicable to our analysis.

Results

The lift during which the tendon rupture occurred was the fifth competitive lift the athlete attempted in the meet. The first three attempts were so-called snatch lifts, in which the weight is lifted in one continuous motion from the floor to a position in which both arms are fully extended above the lifter's head. The first snatch was successful at 135.0 kilograms (1.64 times body weight). The second and third lifts were unsuccessful and successful snatch attempts at 142.5 kilograms (1.73 times body weight). In the so-called clean and jerk competition, an individual must first lift (clean) the weight from the floor to a standing posture in which the bar rests partially on the clavicles and shoulders in addition to being supported by the upper extremities. During the jerk, the lifter's body and the weight are lowered gradually by flexion of lower-extremity joints while the trunk, upper extremities, and head remain rigid. A powerful extension of the lower-limb joints begins the upward motion of the weight. As the weight moves upward, the lifter quickly lowers his body under the weight to regrasp the weight overhead, with the upper extremities fully extended. The subject's fourth lift was a successful clean and jerk attempt at 170.0 kilograms (2.1 times body weight). The combined total of 312.5 kilograms from the third snatch and first clean and jerk attempts gave this lifter a total sufficient to win the light heavyweight championship. His second clean and jerk attempt, pictured at the bottom of Figure 3, was at 175 kilograms (2.13 times body weight); the tendon rupture occurred as he attempted to jerk the weight overhead after successfully cleaning it. The analysis of knee-joint moments of force and patellar-tendon tension was conducted from the beginning of the jerk motion until 0.04 second after tendon failure.

The magnitudes and temporal characteristics of resultant moments of force at the lifter's right knee joint during the jerk phase of the traumatic lift are shown in Figure 3. The initial constant portion of the curve (0.0 to 0.08 second) indicated a net positive (extensor) moment of force at the knee joint to maintain knee extension as the weight was held just prior to the initiation of knee flexion. At approximately 0.10 second into the analysis, a slight increase in extensor moment of force was seen as he elevated both shoulders and the weight slightly, to begin the flexion of the lower-extremity joints. A net flexor moment of force was seen at the knee as he continued to flex (0.12 to 0.18 second). During this same period the acceleration of the weight approached -8.5 meters/s^{-2}, which indicated an essential free fall of the weight. During the following 0.24 second, from 0.18 to 0.42 second, the extensor moment at the knee developed from zero to approximately 400 newton-meters. The rapid increase in extensor moment of force was necessary to slow the weight's descent. An initial peak moment of force at 0.42 second was associated directly with a flexure of the bar connecting the weights. The bar flexed concave downward as the lifter attempted to slow the descent of the weights. Oscillation of the bar developed as the lifter and the weights continued their downward motion. As the weights oscillated upward, with the bar straightening, the necessary knee-extension moment of force decreased slightly (0.42 to 0.48 second). A subsequent downward oscillation, bar concave downward, occurred as the lifter simultaneously stopped the downward motion of his body and the weights (0.48 to

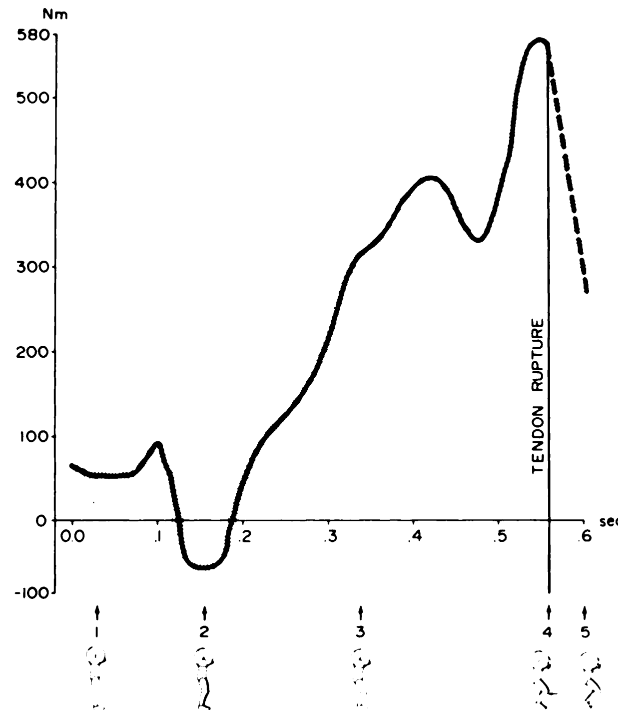

FIG. 3

Mean resultant knee-joint moments are provided from the beginning of the jerk motion until 0.04 second after tendon failure. The solid line represents the mean values of knee moments from the two digitized representations of the data and the shaded area indicates ± 1 standard error of the mean value for each frame. A positive value is a net extensor moment and a negative value is a net flexor moment at the knee joint. Figures of the actual lifter are located temporally along the abscissa to indicate his position during the analysis.

0.56 second). At 0.56 second into the jerk, the moment of force at the knee joint reached a maximum and the patellar tendon ruptured. The knee-joint moments of force dropped precipitously after rupture of the tendon (0.56 to 0.60 second).

The angle at the knee joint was 89.2 degrees when the tendon failed. The maximum knee-extensor moment of force for the 0.04 second prior to rupture was between 550 and 560 newton-meters. The data provided by Smidt [30] indicated that at 90 degrees a value of 3.8 centimeters would provide a reasonable estimate of the moment arm of the patellar-tendon tensile force at the knee joint. Therefore, the patellar-tendon tension was equal to approximately 14.5 kilonewtons, or more than 17.5 times the lifter's body weight at the time of failure.

Discussion

Rupture of the patellar tendon was analyzed previously, but on a qualitative level, and then usually when a pathological condition existed in the patient [21,26-28,35]. The data from the present analysis provide quantitative estimates of loading conditions in actual sports competition.

The absolute magnitudes of joint moments of force and tensile loads must be regarded with proper caution. With the relatively high reliability and validity of the modeling techniques, the calculated resultant moments of force at the knee joint should be valid, but the subsequent calculations of loads on the patellar tendon were sensitive to at least two indeterminate factors: (1) a possible discrepancy between the data of Smidt [30] for moment-arm distances and the true distances for this particular individual, and (2) possible knee-flexor moments of force occurring simultaneously with extensor moments of force.

The nearly identical anthropometric characteristics of Smidt's [30] sample population and those of the lifter in this study support the validity of the calculations of moment-arm distances selected in this analysis. In addition to the anthropometric similarities, our measurements for the lifter and Smidt's [30] measurements were taken from a lateral view of the right knee. Patellar-tendon moment arms were also reported by Haffajee and associates [12] and by Lindahl and Movin [19] for other populations, including men and women. Their moment-arm distances agreed with those of Smidt [30] to within zero to three millimeters. Smith [31] reported moment-arm distances that varied significantly from the other reported values; for example, at a knee-flexion angle of 90 degrees, Smidt's value was 3.8 centimeters and Smith's was approximately 3.1 centimeters. The discrepancy may have been the result of a difference in the technique of measurement; Smith measured the distances using cadaver specimens, while Smidt used a roentgenographic technique with living subjects. In the present analysis, if there was a one-millimeter error in the patellar-tendon moment arm as estimated at the time of tendon failure, it would have caused a 361.5-newton (0.45 times body weight) error in the estimate of tensile load in the tendon.

The second indeterminate factor was inherent in the calculation of net moments of force about a joint. If a flexor moment of force existed at the knee joint at the same time as an extensor moment, the computed result would be a sum of the positive and negative values. In the course of the analyzed movement, a flexor moment may have been caused by the two-joint hamstring muscles. Any flexor moment contributed by the gastrocnemius at the knee joint has been found to be negligible relative to the quadriceps moment of force [2]. During isometric knee extension, Lindahl and Movin [19] determined that the rotatory effect of the anterior cruciate ligament at the knee joint may be disregarded for all practical purposes. They concluded that the rotatory force causing extension may be considered to be a quadriceps mechanism. Any small flexor countermoment at the time of tendon failure would have increased the actual tensile loading on the tendon. If a flexor moment existed, the reported tensile load at rupture is a conservative estimate.

The experiments in which the physical properties of tendons were analyzed may be categorized as either *in vitro* or *in vivo*. The *in vitro* studies can be divided into experiments in which loads were applied at a slow, constant rate and experiments in which the rate of strain was varied.

The physical properties of tendons and ligaments in response to low strain rates have been investigated extensively [1,4,6,7,10,11,14,29,33]. It is difficult, however, to relate these data to loading conditions in normal movements or in injury situations, when stress on tendons can be expected to be applied rapidly [13,17,18]. With advances in experimental equipment, strain rate has become a significant independent variable for *in vitro* analyses of tendon and ligament properties. Normal viscoelastic properties of ligaments and tendons have been demonstrated *in vitro* in both animals [15,23,32] and humans [18], although in one study [34] using normal and regrown canine tendon a decrease in Young's modulus was noted with an increase in strain rate. With human knee ligaments, Kennedy and associates [18] attempted to apply a strain rate that may be expected during an injury. They reported increases of 14 to 42 per cent in maximum load at rupture for different human knee ligaments as strain rates were increased from 2.08 to 8.33 millimeters per second.

Obtaining valid data for the mechanical properties of tendon *in vivo* has been difficult. Measurements [17] of the mechanical strain in tendons in sheep during locomotion indicated that the rate of strain is very rapid, and that the rate increases with increased speed of locomotion. The mean maximum strain rate recorded from the animals was 19.35 per cent strain per second for about 0.10 second during fast walking or trotting.

A direct comparison of the data of the present study with previous experiments determining the mechanical properties of tendons was not feasible. Certain data from this study, however, may be useful in subsequent *in vitro* investigations. In particular, the results indicated that the absolute time from development of a zero net moment of force at the knee joint to patellar-tendon rupture was only 0.38 second.

Opportunities to obtain kinetic data such as are presented in this study will continue to be exceptional situations. The greatest promise for determining the ranges of tendon and ligament mechanical properties during normal and traumatic stress conditions seems to lie in the refinement of *in vivo* measurement techniques.

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