Journal of Orthopaedic Research
2:49–54, Raven Press, New York
© 1984 Orthopaedic Research Society

Force Ratios in the Quadriceps Tendon and Ligamentum Patellae

H. H. Huberti, W. C. Hayes, J. L. Stone, and G. T. Shybut

Orthopaedic Biomechanics Laboratory, Charles A. Dana Research Institute, Beth Israel Hospital and Harvard Medical School, Boston, Massachusetts

Summary: We measured forces in the quadriceps tendon and ligamentum patellae in six human cadaver knee joints loaded through a range of flexion angles from 30 to 120°. Using standardized loads based on one-sixth of maximum isometric quadriceps moments reported in human volunteers, we measured maximum average forces at 60° flexion of 532 N in the quadriceps tendon (F_Q) and 470 N in the ligamentum patellae (F_L). Linear extrapolation to full maximum extensor moments results in estimates of 3,200 N for F_Q and 2,800 N for F_L. The force ratio (F_L/F_Q) reached a maximum value of 1.27 at 30° and minimums of 0.7 at 90 and 120° knee flexion. Contrary to prevailing opinion, our results indicate that the patella is not a simple pulley that serves only to change the direction of equal forces in the quadriceps tendon and ligamentum patellae. Instead, the ratio of forces in these structures varies significantly with flexion angle. The ratio appears to be determined by the changing location of the patellofemoral contact area relative to the insertions of the tendon and ligament. These findings emphasize the biomechanical importance of patellar length and of the vertical dimensions and locations of the patellofemoral contact area. Attempts at surgical intervention for the treatment of disorders of the extensor apparatus should recognize these variations. Procedures that tend to move the contact area more proximally (at a particular flexion angle) will also tend to decrease the F_L/F_Q ratio. **Key Words:** Patellar forces—Buckle transducers—Knee—Joint forces.

The high frequency of chondromalacia and osteoarthritis of the patellofemoral joint has led to many attempts to calculate the extensor forces acting on the loaded knee. Most authors have assumed that the forces in the ligamentum patellae and the quadriceps tendon are equal (2,6,15,18,19,21,22,25). Maquet (14) and other workers (3,5) have suggested that the force in the patellar ligament is less than in the quadriceps tendon or that the force ratio varies with flexion angle. The patella is actually a complex geometric structure subjected to a three-dimensional force system. In addition, the patellofemoral contact area shifts its location from the distal end of the patellar articular cartilage to the proximal pole as the flexion angle increases (6). The possibility thus exists that the ratio of ligamentum patellae and quadriceps tendon forces is a function of flexion angle.

Force magnitudes in both structures have never been measured directly in intact knees. Most data available are obtained from calculations based on kinematic measurements (4,19), measurements of extensor moments (22), or measurements of floor reaction forces (19). These predictions of force mag-

Presented in part at the 29th Annual Meeting of the Orthopaedic Research Society, Anaheim, California, March 8–10, 1983.
Address correspondence and reprint requests to Dr. Hayes at Orthopaedic Biomechanics Laboratory, Beth Israel Hospital, 330 Brookline Avenue, Boston, Massachusetts 02215.

nitudes are usually based on the assumption that forces in the quadriceps and ligamentum patellae are equal. In addition, the variety of loading conditions used in these calculations makes it difficult to compare previous studies.

Our objective with these experiments was to measure directly forces in the quadriceps tendon and the patellar ligament in intact cadaver knee joints subjected to controlled external forces through a range of flexion angles.

MATERIALS AND METHODS

We tested six fresh frozen human cadaver knee joints (age 62 to 74, three men and three women). All specimens were normal radiographically, had macroscopically intact cartilage, and were free from previous surgery. Both tibia and femur were sectioned transversely approximately 25 cm from the joint space. With the additional length provided by the loading fixtures, the resulting functional lengths of both tibia and femur were approximately 35 cm. After bilateral release of the capsule, the quadriceps was clamped 5 cm above the proximal pole of the patella, including the layered sections of the rectus and intermedius tendons. The clamp was attached through a force transducer to the proximal end of the femur. This allowed knee flexion angles of up to 120° without interference between clamp and femoral groove. The joints were mounted into a materials test system (Model 1331, Instron Corp., Canton, MA 02021) using a special loading fixture (Fig. 1). Knee joint resultant moments were generated by the application of collinear compressive loads to low-friction spherical bearings at locations approximating the hip and the ankle joint. Through the natural knee stabilizing mechanisms, the application of these external forces resulted in forces in the quadriceps tendon and the ligamentum patellae.

To standardize loading conditions, the joints were loaded with bending moments equivalent to one-sixth of the maximal isometric quadriceps moments measured in human volunteers (7,9,12,13,22). These literature data indicate a consistent pattern of quadriceps moment versus flexion angle with the greatest moments generated at 60° flexion. Based on these data from normal populations, we applied moments of 15.4 Nm at 30° flexion, 23.6 Nm at 60°, 17.5 Nm at 90°, and 11.3 Nm at 120°. This was achieved by applying collinear compressive forces of 150 N at 30°, 115 N at 60°, 62.5 N at 90°, and 34 N at 120° flexion. Forces were applied as truncated,

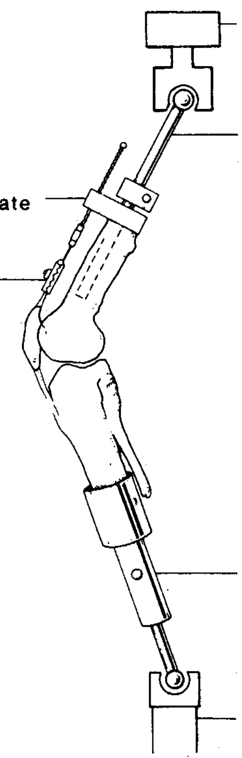

FIG. 1. Lateral view of loading fixture for *in vitro* force measurements in quadriceps tendon and ligamentum patellae. A tensile load cell is mounted between *Q*-angle adjustment plate and quadriceps tendon clamp, and a buckle transducer is attached directly to the ligamentum patellae.

symmetric, triangular impulses with a plateau of 3 s and a total duration of 13 s.

We measured quadriceps forces with a strain gaged tension load cell mounted between the quadriceps clamp and the proximal bracket used for *Q*-angle adjustment. Forces in the ligamentum patellae were measured with a buckle transducer as described by Lewis and Fraser (11) (Fig. 2). Calibration of both transducers was performed after each experiment by applying collinear tensile forces directly to the quadriceps transducer and to a Steinmann pin placed mediolaterally through the tibial tuberosity. Both transducers were linear (R^2 = 0.999) throughout the entire loading range from 50 to 1,000 N.

RESULTS

Forces in the quadriceps and ligamentum patellae varied with knee flexion angle. Both were greatest at 60° flexion (Table 1). For the applied forces corresponding to one-sixth of maximum isometric extensor moments, average (number = 6) maximum forces at 60° were 532 N (±155, 95% CI) in the

TABLE 1. *Measured forces in the ligamentum patellae* (F_L) *and the quadriceps tendon* (F_Q) *and their ratio*

Flexion angle (°)	F_L (N)	F_Q (N)	Force ratio F_L/F_Q
30	309 ± 81	245 ± 63	1.27 ± 0.26
60	470 ± 118	532 ± 155	0.91 ± 0.16
90	356 ± 105	511 ± 92	0.70 ± 0.17
120	207 ± 93	300 ± 102	0.70 ± 0.30

Based on applied moments of one-sixth of maximum isometric extensor moments in an average population. Means for n = 6 (±95% CI).

quadriceps and 470 N (±118, 95% CI) in the ligamentum patellae (Table 1). Linear extrapolation to full maximum extensor moments results in average maximum force estimates of approximately 3,200 N in the quadriceps and of approximately 2,800 N in the ligamentum patellae. These values are between four and five times standard body weight of 700 N. At 30 and 120°, forces in both structures were roughly one-half of these peak values.

The force ratios (F_L/F_Q) of ligament force (F_L) to quadriceps force (F_Q) varied significantly ($p < 0.01$ for an analysis of variance for force ratios against flexion angle; and $p < 0.05$ for grouped *t* tests between the ratio at 30° and the values at each of 60, 90, and 120°). The ratios exhibited a typical pattern in all six joints tested (Fig. 3). At 30°, the smallest flexion angle we tested, forces in the ligamentum patellae were always greater than forces in the quadriceps tendon. The average ratio (F_L/F_Q) at 30°

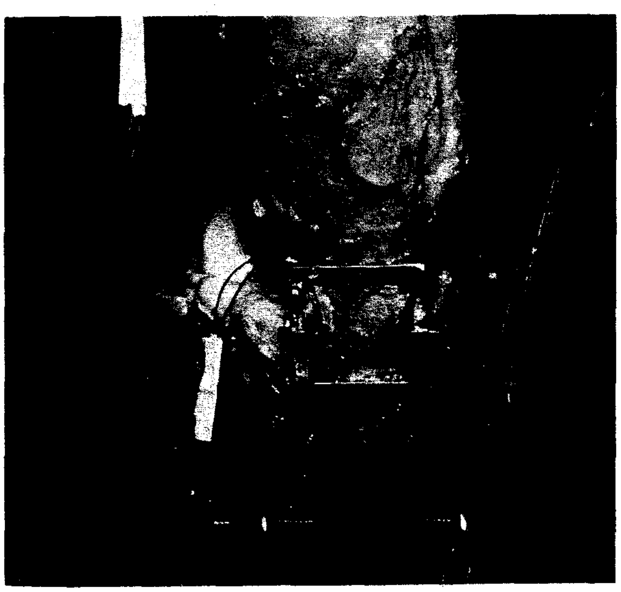

FIG. 2. Frontal view of the buckle transducer *in situ* on the ligamentum patellae.

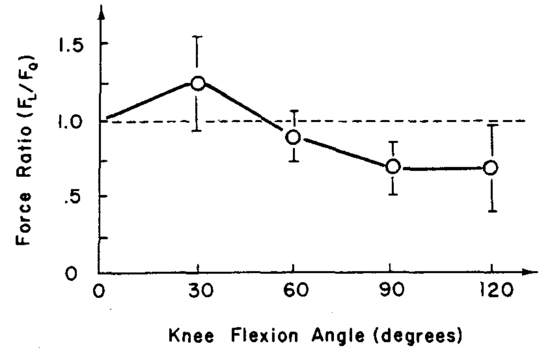

FIG. 3. Force ratio (F_L/F_Q) of ligamentum patellae force (F_L) to quadriceps tendon force (F_Q) as function of flexion angle. At 30° F_L is approximately 30% greater than F_Q. At 90 and 120° F_L is approximately 30% less than F_Q.

was 1.27 (±0.26, 95% CI), which indicates that ligamentum patellae forces were approximately 30% greater than quadriceps forces at this flexion angle. At flexion angles greater than approximately 45° the ratio was less than one. At 90 and 120°, the average force ratios were 0.70, with the ligamentum patellae force approximately 30% less than quadriceps force.

DISCUSSION

Force Ratios

These data demonstrate that the force ratio in the ligamentum patellae and quadriceps tendon varies with the flexion angle. This finding confirms what had been suspected previously on theoretical and clinical grounds (14). However, the result is in contrast with results from previous experimental studies (3,5) at similar flexion angles. Ellis et al. (5) found a ratio of close to one (0.97) at 30° flexion. Bishop and Denham (3) found a ratio of 0.7 at 40°, the smallest flexion angle tested in their study. Both findings were based on experimental preparations that included only femur and patella. The large differences between these earlier studies and ours may be related to the variations in patellofemoral alignment associated with the different experimental configurations.

In our experiments the F_L/F_Q ratio was greater than 1.0 at flexion angles less than approximately 45°. This finding can be explained by a characteristic shift in the location of the patellofemoral contact area at different flexion angles. The proximal-distal dimension of the contact area varies from 6 to 10 mm depending on the flexion angle (8) and is only a fraction of the total patellar length (approx-

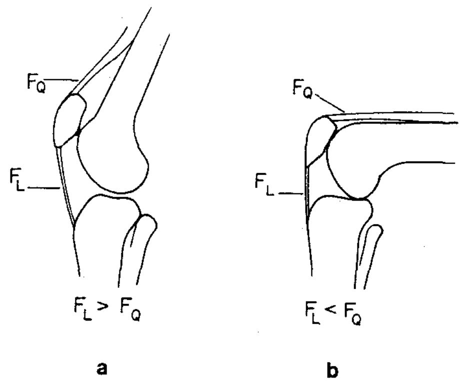

FIG. 4. Shifting location of patellofemoral contact area at different flexion angles determines the force ratio. **a:** At 30° flexion the contact area is located distally. The quadriceps (F_Q) acts at a mechanical advantage and generates a greater ligamentum force (F_L). **b:** At 90° the opposite situation occurs.

imately 5 cm) (6). At small flexion angles, the patellofemoral contact area is located at the distal end of the patella cartilage (6,7). This distal location causes the quadriceps force to act at a mechanical advantage to the ligamentum patellae force (Fig. 4a). The result is that the ligamentum patellae force is larger than the quadriceps force at low flexion angles. At higher flexion angles, the opposite situation is true (Fig. 4b). The patellofemoral contact area thus occurs near the proximal pole of the patella and the ligamentum patellae force acts at a mechanical advantage to the quadriceps force. Thus, the varying location of the patellofemoral contact area on the patellar cartilage is the primary determinant of the force ratio, with the patella acting as a three-dimensional lever around the patellofemoral contact area.

This finding diverges from a view of the patella as a simple pulley (14,18,20,22). Such a pulley model accounts only for the anteroposterior thickness of the patella in increasing the effective moment arms of forces in both the quadriceps and ligamentum patellae. As an example, it has been emphasized repeatedly in the literature that patellectomy leads to a reduction of the extensor moment arm and a decrease in maximum extensor moment. It seems, however, that the length of the patella and the relative location of the patellofemoral contact area are also important biomechanically. These parameters define proximal and distal lever arms of different magnitudes, which result in unequal forces in the ligamentum patellae and the quadriceps tendon. At flexion angles less than approximately 45°, the distal location of the contact area results in F_L/F_Q ratios greater than one. At flexion angles greater than 45°, the contact area is located more proximally and the force ratio is less than one.

Thus, it is no longer tenable to consider the patella as a simple pulley serving to change the direction but not the magnitude of the forces in the quadriceps tendon and ligamentum patellae. Instead, the patella must be considered as a three-dimensional body acted on by a system of three forces. Two of these forces (F_L and F_Q) change magnitude and direction (but not their points of application) and one (the patellofemoral contact force) changes magnitude, direction, *and* point of application. Equilibrium of this force system therefore requires that the F_L/F_Q ratio changes as the patellofemoral contact force changes position during flexion of the knee. It is important to note that this dependence of the force ratio on flexion angle (and thus on the pattern of applied moments used to determine the ratio) is not influenced by the loading magnitudes. For a given flexion angle, as long as the position and size of the patellofemoral contact area is unchanged at different load magnitudes [a fact we have demonstrated from direct patellofemoral contact pressure measurements (8)], the F_L/F_Q ratio will be invariant with the load magnitudes. Thus, had we chosen different patterns of applied resultant knee moments to determine the ratio, equilibrium conditions require that the measured force ratios remain the same.

The clinical implications of our findings relate to the changing F_L/F_Q ratio as a function of flexion angle and to the role of the location of the patellofemoral contact area in determining the force ratio. Thus, should it be necessary to minimize forces on the ligamentum patellae but not on the quadriceps tendon, our findings would suggest that the application of large isometric or isokinetic knee moments at nearly full knee extension should be particularly avoided. Under such conditions, it might be useful to restrict knee extension exercises to flexion angles greater than approximately 45°, both to avoid increased knee moments, which occur near full extension, and to minimize the amplification of the ligamentum patellae force that our F_L/F_Q ratio measurements indicate. Of course, during normal activities, such as gait, where the resultant bending moments (and thus the quadriceps forces) are not in themselves large, this restriction on knee activities near full extension would not apply.

The changing location of the patellofemoral contact area may also be important clinically. For in-stance, conditions such as patella alta, which in turn may influence the location of the contact area, can also be expected to influence the F_L/F_Q ratio. In addition, certain surgical reconstructive procedures, such as capsular release, advancement of the tibial tubercle, or even resurfacing of the patella during total knee replacement, may change not only the anteroposterior location of the patella but also the proximal-distal location of the contact area. The importance of the anteroposterior location of the patella has long been appreciated, whereas the influence of the proximal-distal location of the contact area has not.

Force Levels and Patterns

The forces measured here were based on one-sixth of maximal isometric quadriceps moments in average populations. Extrapolation to full isometric moments suggests that maximum forces in both the quadriceps muscle and the ligamentum patellae can be generated at 60° flexion. Both are predicted to reach approximately 3,000 N in average populations. In women, who exert on average 60% of the average maximum moments in men (12), maximal forces of 1,800 N can be expected. In young trained men, maximum isometric moments are approximately twice as high as in the average population (10,12,17,24,26), reaching maximum values of approximately 250 Nm. In these subjects, maximum forces of 6,000 N (approximately 10 times the body weight) can be expected in the quadriceps and the ligamentum patellae. From isokinetic dynamometer measurements, these quadriceps force levels required to generate maximum isometric knee moments are also the highest forces that can be generated by the quadriceps muscles (24).

Our results agree with a number of previous theoretical predictions of force levels in the knee. Smith's (23) calculations of ligamentum patellae forces were based on moment variations similar to ours and thus can be compared directly. Smith measured maximum isometric moments of approximately 110 Nm at 60° flexion, which is approximately 20% less than our maximum moment of 141 Nm. For this quadriceps moment he calculated a ligamentum patellae force of 2,710 N, very close to the 2,800 N we estimated for maximum isometric moments. Similar close agreements were found by comparing our results at 60° flexion with force predictions for climbing stairs (1,16).

For deep knee bends, Reilly and Martens (19) calculated quadriceps forces that reached nearly maximum isometric levels at flexion angles of 120° but not at smaller flexion angles. In three young men they predicted an average maximum quadriceps force of approximately 3,500 N at 120° flexion and approximately 1,000 N at 60°. The predicted force at 120° is very close to the maximum isometric force at this flexion angle (10,12), and this agrees well with our extrapolated force value of 3,750 N at 120°.

For rising unaided by the hands from a chair, Ellis et al. (4) calculated a quadriceps force of approximately 1,800 N at approximately 120° flexion. This result, for mixed male and female subjects, is equivalent to our extrapolated predictions of 1,800 N quadriceps force at the same flexion angle under maximum isometric conditions. At smaller flexion angles, however, Ellis et al. calculated much smaller forces. At 60°, they calculated quadriceps forces of only 600 N. Seedhom and Terayama (21) found a similar force pattern, but even smaller force values for rising unaided from a chair. They found a maximum quadriceps force at 110° of only 1,260 N. The force at 60° was 450 N. Both values are approximately two-thirds of those obtained by other workers. Their data, however, were obtained in only two individuals of unspecified sex.

These comparisons show that there is a generally close agreement between our measured results and previously calculated force values based on similar loading conditions. Our results suggest, however, that at most flexion angles it is necessary to differentiate between quadriceps muscle force and ligamentum patellae force since they cannot be assumed to be equal. Additionally, the results show that basing *in vitro* loading conditions on maximum isometric extensor moments from average populations appears to be an adequate way of loading standardization. Resulting forces represent the highest forces occurring *in vivo* and can be easily extrapolated to forces in women and men.

CONCLUSIONS

1. The ratio F_L/F_Q varies with flexion angle, reaching a maximum value of 1.27 at 30° flexion and minimum values of 0.7 at 90 and 120°. The proximal-distal location of the patellofemoral contact area appears to be the main factor in determining this ratio. With increasing flexion angle, the patellofemoral contact area moves from distal to proximal on the patellar surface causing the quadriceps force to act first at an increased and then at a de-

creased mechanical advantage to the patellar ligament force. This result emphasizes the biomechanical importance of the length of the patella and of the longitudinal dimensions and location of the patellofemoral contact area.

2. Comparative investigations of knee joint extensor forces require the use of standardized knee joint loading patterns. We suggest here the use of standard loadings based on maximum isometric moments measured *in vivo* in average populations at different flexion angles. This standard allows direct extrapolation to maximal *in vivo* forces in both men and women.

3. Our direct measurements of forces in the quadriceps tendon and the ligamentum patellae suggest maximum forces in both structures at 60° flexion. Extrapolating our measurements to full maximum isometric extensor moments previously measured in human volunteers indicates maximum quadriceps force values of 3,200 N in average mixed populations, of 1,800 N in women, and of approximately 6,000 N in young trained men.

Acknowledgment: This study was supported by NIH RCDA AM 00749, by AM 26740, and by an Orthopaedic Research Fellowship from the Deutsche Forschungsgemeinschaft (Federal Republic of Germany). We thank Paul Phillips for his patience in the preparation of this manuscript.

REFERENCES

1. Andriacchi TP, Andersson BJ, Fermier RW, Stern D, Galante JO: A study of lower-limb mechanics during stair climbing. *J Bone Joint Surg [Am]* 62:749–757, 1980
2. Bandi W, Brennwald J: Degenerative joint disease. The significance of femoropatellar pressure in the pathogenesis and treatment of chondromalacia patellae and femoropatellar arthrosis. In: *The Knee Joint,* ed by OS Ingwersen et al., Amsterdam, Excerpta Medica, 1974, pp 63–68
3. Bishop RED, Denham RA: A note on the ratio between tensions in the quadriceps tendon and infra-patellar ligament. *Eng Med* 6:53–54, 1977
4. Ellis MI, Seedhom BB, Amis AA, Dowson D, Wright V: Forces in the knee joint whilst rising from normal and motorized chairs. *Eng Med* 8:33–40, 1979
5. Ellis MI, Seedhom BB, Wright V, Dowson D: An evaluation of the ratio between the tensions along the quadriceps tendon and the patellar ligament. *Eng Med* 9:189–194, 1980
6. Ficat RP, Hungerford DS: *Disorders of the Patello-Femoral Joint.* Baltimore, Williams and Wilkins, 1977, pp 29–32
7. Haffajee D, Moritz U, Svantesson G: Isometric knee extension strength as a function of joint angle, muscle length and motor unit activity. *Acta Orthop Scand* 43:138–147, 1972
8. Huberti HH, Hayes WC: Patello-femoral contact pressures. The influence of Q-angle and tendo-femoral contact. *J Bone Joint Surg [Am]* 66:715–724, 1984
9. Inman VT, Ralston HJ, Todd F: *Human Walking,* Baltimore, Williams and Wilkins, 1981, pp 88–90
10. Knapik JJ, Ramos MU: Isokinetic and isometric torque relationships in the human body. *Arch Phys Med Rehabil* 61:64–67, 1980
11. Lewis JL, Fraser GA: On the use of buckle transducers to measure knee ligament forces. *Am Soc Mech Eng* AMD-32:71–73, 1979
12. Lieb FJ, Perry J: Quadriceps function. An electromyographic study under isometric conditions. *J Bone Joint Surg [Am]* 53:749–758, 1971
13. Lindahl O, Movin A, Ringqvist I: Knee extension. Measurement of the isometric force in different positions of the knee joint. *Acta Orthop Scand* 40:79–85, 1969
14. Maquet PGJ: *Biomechanics of the Knee.* Berlin, Springer, 1976
15. Morrison JB: Bioengineering analysis of force actions transmitted by the knee joint. *Biomed Eng* 3:164–170, 1968
16. Morrison JB: Function of the knee joint in various activities. *Biomed Eng* 4:573–580, 1969
17. Murray MP, Baldwin JM, Gardner GM, Sepic SB, Downs WJ: Maximum isometric knee flexor and extensor muscle contractions. Normal patterns of torque versus time. *Phys Ther* 57:637–643, 1977
18. Perry J, Antonelli D, Ford W: Analysis of knee-joint forces during flexed-knee stance. *J Bone Joint Surg [Am]* 57:961–967, 1975
19. Reilly DT, Martens M: Experimental analysis of the quadriceps muscle force and patello-femoral joint reaction force for various activities. *Acta Orthop Scand* 43:126–137, 1972
20. Seedhom BB: Biomechanics of the lower limb. In: *An Introduction to the Biomechanics of Joints and Joint Replacement,* ed by D Dowson, V Wright, London, Mechanical Engineering, 1981, pp 73–82
21. Seedhom BB, Terayama K: Knee forces during the activity of getting out of a chair and without the aid of arms. *Biomed Eng* 11:278–282, 1976
22. Smidt GL: Biomechanical analysis of knee flexion and extension. *J Biomech* 6:79–92, 1973
23. Smith AJ: A study of forces on the body in athletic activities with particular reference to jumping [PhD Thesis]. Leeds, 1972
24. Thorstensson A, Grimby G, Karlsson J: Force-velocity relations and fiber composition in human knee extensor muscles. *J Appl Physiol* 40:12–16, 1976
25. Walker PS: Contact areas and load transmission in the knee. In: *AAOS Symposium on Reconstructive Surgery of the Knee,* St Louis, CV Mosby, 1978, pp 26–36
26. Zernicke RF, Garhammer J, Jobe FW: Human patellar-tendon rupture. A kinetic analysis. *J Bone Joint Surg [Am]* 59:179–183, 1977