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Artificial Limbs

A Review of Current Developments

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COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT

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NATIONAL ACADEMY OF SCIENCES—NATIONAL RESEARCH COUNCIL
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Artificial Limbs is a publication of the Committee on Prosthetics Research and Development, National Academy of Sciences—National Research Council, issued randomly two numbers to the volume, in partial fulfillment of Veterans Administration Contract V1005M-1914, Office of Vocational Rehabilitation Contract SAV-1000, and National Institutes of Health Grant RG-5057. Library of Congress Catalog Card No. 55-7710.

The Patellar-Tendon-Bearing Prosthesis

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OBIOUSLY there is no "ideal" leg substitute short of regenerating or transplanting another normal leg. The surgeon, the prosthetist, and the amputee alike have long accepted major deficiencies in leg prostheses as inescapable concomitants of aid in a situation demanding drastic compromise. Substitution of an artificial leg for a natural one involves not only manual skills and the principles of inanimate mechanisms but is also dependent on anatomy, physiology, and biomechanics. Mutual application of these disciplines toward the advancement of leg prosthetics was slow in coming. As the science of astronomy emerged from the superstitions of astrology, so too there is sound reason to hope that the profession of prosthetics will continue to grow increasingly rapidly beyond the great dependence on "experience in the finger tips" of the ancient skill of limbmaking by adding to its art more general application of the discoveries of science.

Time after time, like recurrent approaches of a comet from its far-reaching orbit, dazzling prospects of improvements in prostheses for below-knee amputees have illuminated the prosthetics scene. The slip socket, the many attempts at end-weight-bearing, the "muley" leg without side joints or corset, the single sidebar, the various polycentric joints, and the several attempts at below-knee suction sockets have been spectacular objects visible for varying periods in Europe, the United States, or alternately in both regions. Unhappily, these phenomena, like comets, have often receded into outer darkness as abruptly as they appeared, leaving the typical amputee with crutches, peg leg, or the centuries-old "conventional" prosthesis.

Pads, straps, locks, and similar devices often reflect either lack of knowledge or incomplete application of such knowledge as there is to control pressure or to overcome instability. Freedom of the human knee joint, distribution of forces in proportion to tolerance of tissues, improved rather than constricted circulation, and better kinesthetic appreciation—all major goals in recent years—demand simplicity of mechanism and reduction of the false joint between the prosthesis and the body by use of an intimate fit.

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The patellar-tendon-bearing (PTB) prosthesis developed by the Biomechanics Laboratory of the University of California, to which much of this issue is devoted, combines many long-controversial features—each long used by some, yet rejected by others. PTB is almost a code name integrating a long list of elements which the prosthetist through logical principles and teachable techniques employs to distribute forces comfortably. Because of individual variations, not all so-called "PTB prostheses" contain all the major features. The name implies weight-bearing on the patellar tendon, more properly called the patellar ligament. Because in fact the nearby retinacula also share weight, perhaps the name might well be the "patellar-tendon*s*-bearing" prosthesis! Actually, as later pages of this issue describe, many other areas of the socket (notably the closed distal end) are at least in contact with the stump, and some (*e.g.*, the flares supporting the tibial condyles) share substantial portions of body weight.

Because of its typical use of cuff suspension, with consequent freedom from thigh corset, the PTB prosthesis is often erroneously identified with the "muley" leg, which has stomped the field for as much as a century and yet has so often developed complications during prolonged use. One may speculate that the common complaints of instability of the knee attributed to the "muley" principle were at least partially related to poor alignment between socket and foot, excessive extension or even hyperextension of the socket axis and hence of the human knee, and needlessly low brim levels offering less than maximum stability to the stump. Careful prescription and medical supervision, not available for the earlier "muley," should also characterize use of the PTB and greatly enhance its chances of success.

This writer's personal observations, from visits to the birthplace of PTB and to numerous clinics throughout the United States, have indicated misconceptions of the role of knee flexion in initial alignment of the socket axis. Certainly hyperextension is to be avoided and mild flexion sought. Because the *cast* is taken with the knee in substantial (possibly excessive?) flexion, some newly trained prosthetists initially aligned the socket bore similarly but with a very large angle of flexion. The horizontal components of forces on the condyles were reduced; but the resulting extreme bent-knee gait was tiring, the quadriceps were unduly stressed in their atrophied state immediately after their release from bondage within the thigh corset, and the unique mechanical stability of the extended human knee was transformed into the capability of substantial horizontal rotation of the flexed knee. In the below-knee amputee lacking an actively steerable ankle and foot, an unimpaired but controlled horizontal rotation in the knee joint must be considered of added importance. Thus neither the rigid "screw-home" of final extension nor the gross instability of major flexion will be as suitable as mild flexion with control of unencumbered hamstrings as internal and external rotators.

In many past efforts too little attention has been paid to the popliteal space.

The PTB includes logical principles allowing a higher brim in the popliteal space (and indeed on all aspects) than has been customary in a majority of cases yet freedom for action of the hamstrings and avoidance of bulging of tissue during sitting. The high brims medially and laterally, reflecting better appreciation of anatomy and of the force patterns dictated by biomechanics, should give greater mediolateral stability than was typically available with a "muley" limb. Eventual use of brims of tapering flexibility, by avoiding sharp pressure points at the very edge, may ultimately allow still better fitting.

No one, especially among its developers, would acclaim the PTB as the ultimate solution. Some of its features represent successive reincarnations over a century, each with a higher survival percentage. Yet the PTB is only an evolutionary step toward greater mechanical freedom under better neuromuscular discipline. Many apparent failures can be salvaged by careful adherence to the principles and techniques enunciated in the UCB manual and its recent revision and in the following papers of this issue of ARTIFICIAL LIMBS.

The conveniences which the PTB leg accords its wearer are so numerous that continued efforts seem assured. Though a single breaker may recede, the tide is surely coming in.

Anatomical and Physiological Considerations in Below-Knee Prosthetics

EUGENE F. MURPHY, Ph.D.[1]
AND
A. BENNETT WILSON, JR.[2]

ONE of the most difficult problems in the design of prostheses is the development of the best means of attaching the prosthesis to the wearer. In lower-extremity cases, transmission of forces between stump and prosthesis is of primary importance. To effect efficient transmission of forces, a stable connection between stump and prosthesis is necessary. At the same time comfort and freedom of motion must be maintained to as high a degree as possible. All of these goals are affected by anatomical and physiological characteristics of the stump and the next proximal joint, and often of the joint above that.

Stability is provided most often by encasing the stump in a socket to a point near the first proximal joint. The soft tissues of the stump are not especially ideal for providing resistance to the torques and moments imposed on them by a socket during use of a prosthesis. If the tissues are compressed in an attempt to provide maximum stability, circulation will be impaired; if the socket is too loose, a false-joint effect is produced resulting in abnormally high unit pressures at proximal and distal points, chafing, and a reduction in ability to control the prosthesis. Thus, extreme care must be exercised in socket design and fabrication if the optimum condition is to be obtained.

When weight-bearing can be achieved through the long bones, as in the case of many disarticulations and certain special types of amputation, the socket is designed to permit loads to be carried through the end of the bone in the stump. If most of the weight-bearing needed cannot be achieved through the end, some other areas must be found to provide the transmission of forces necessary during standing. For all of these reasons, then, it is extremely important that prosthetists and others responsible for the design of sockets take into consideration certain anatomical and physiological factors in the management of the amputee. In no other case is it more important than in that of the below-knee amputee.

FUNCTION OF THE BELOW-KNEE STUMP

Because most of the insertions of the muscles and ligaments that control the knee are located on the tibia and fibula at points close to the knee joint (Figs. 1, 2, 3), amputation below the knee rarely affects the function of the knee joint. An exception is the gastrocnemius which originates from the posterior portion of each of the femoral condyles and has for its insertion the Achilles tendon, thus acting as a flexor. Upon amputation, however, the distal end of the gastrocnemius often becomes reattached to the tibia, and the remaining musculature is thus available to assist the flexors and perhaps to aid in preventing dislocation of the fibula with respect to the tibia. Thus the moment that can be generated about the knee in the parasagittal plane by a typical below-knee amputee is approximately the same as that before amputation. Because, in general, the ligaments are left untouched, mediolateral stability of the below-knee amputee usually is not affected.

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Those muscles which have origins on the tibia and fibula, and which control ankle and foot motion, have been severed and consequently atrophy, resulting generally in a bony, conical-shaped stump (Fig. 4). The amount and type of atrophy that takes place depend of course upon surgical technique and postoperative care.

In very short below-knee stumps, removal of the fibula (Fig. 5) is sometimes performed to prevent lateral and posterior deviation with uncomfortable protrusion at the distal end. Such deviation is generally thought to be caused by frictional engagement on the socket wall (with inadequate relief) or by action of the biceps femoris. In any below-knee amputee, the distal ligamentous attachment near the ankle is missing, and in short stumps the interosseus membrane (Fig. 6) between the remnants of the tibia and the fibula is presumably inadequate, partly because the proximal opening for the vessels leaves only a small amount of the membrane, and particularly because atrophy of intervening muscles leaves some slack in the membrane. Removal of the fibular head, though, implies that the tendon of the biceps femoris, as well as the fibular collateral ligament, should be reattached with appropriate lengths and at suitable centers on the tibia. A bone bridge from fibula to tibia (p. 90) that would restore stability between tibia and fibula as well as increase the possibilities for bearing weight on the end of the stump would seem to be preferable to removal of the fibula.

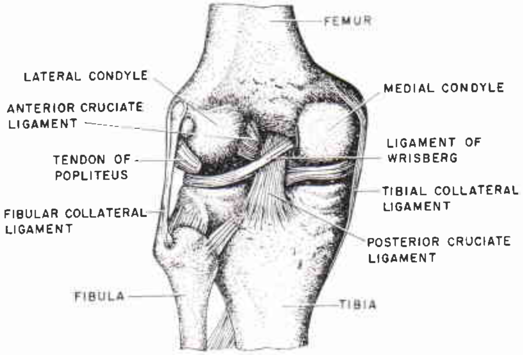

Fig. 1. Posterior view of left knee joint, showing anterior ligaments. Redrawn from Gray's *Anatomy*.

THE KNEE JOINT

The knee joint formed by the condyles of the femur and tibia (Figs. 3, 7) allows about 160 deg. of flexion. It is classified as a synovial joint, or one that is provided with synovial fluid, and the friction developed between the moving surfaces of an unimpaired joint is of an unusually low magnitude as compared with

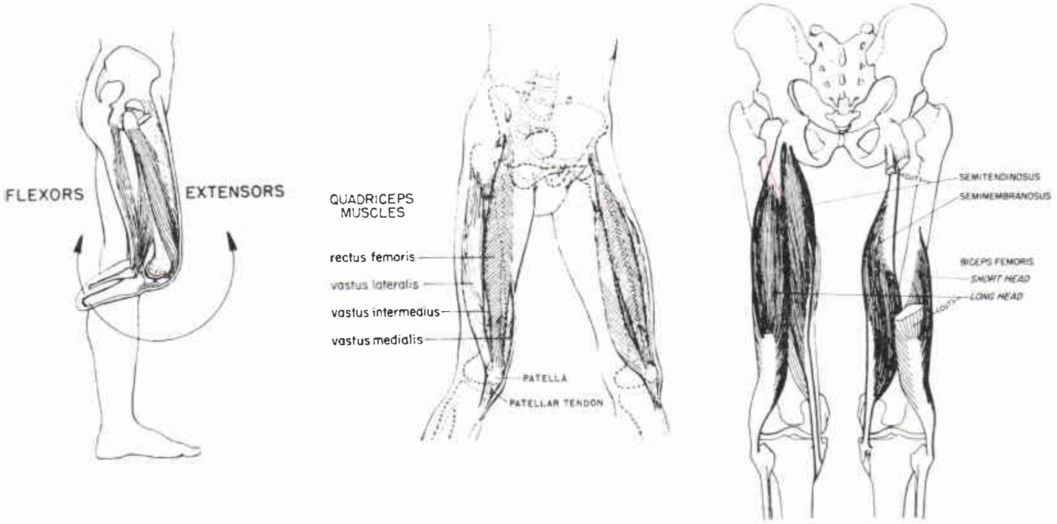

Fig. 2. The major muscles that flex and extend the knee joint. From *The Patellar-Tendon-Bearing Below-Knee Prosthesis* (*4*).

Fig. 3. X-rays of a typical below-knee stump. *A*, Anterior view; *B*, medial view. Courtesy *Veterans Administration Prosthetics Center*.

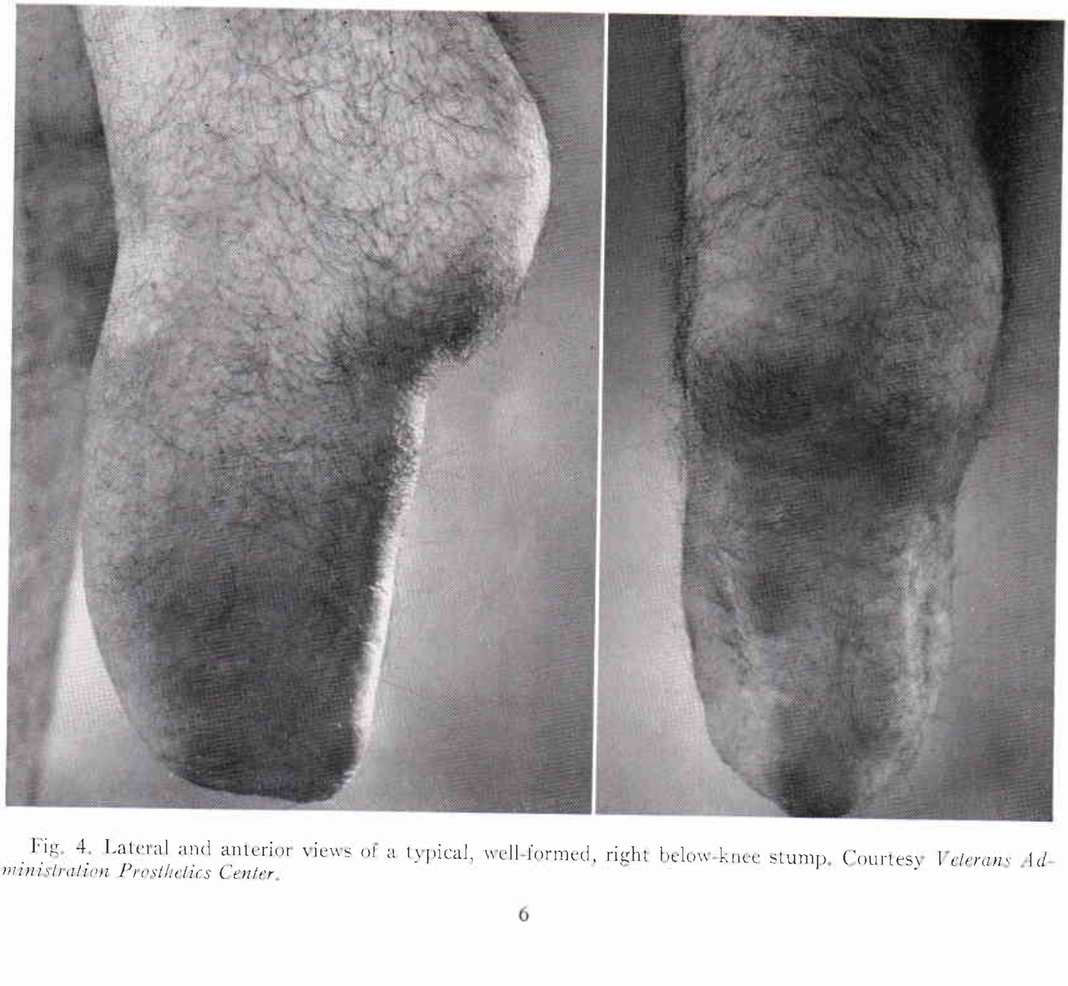

Fig. 4. Lateral and anterior views of a typical, well-formed, right below-knee stump. Courtesy *Veterans Administration Prosthetics Center*.

Fig. 5. Roentgenogram of a short below-knee stump in which lateral deviation and rotation of the fibula have taken place. *Courtesy University of California Medical School.*

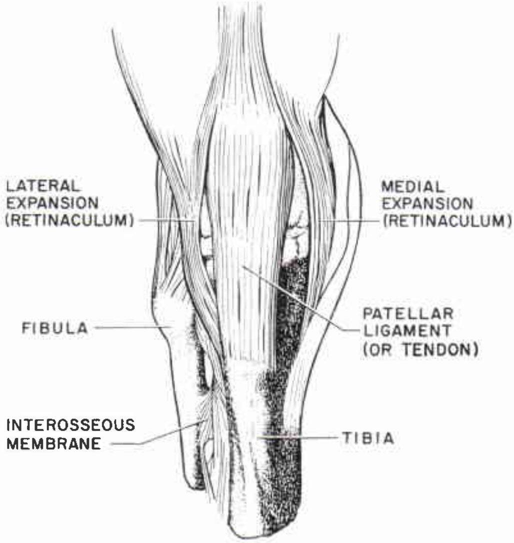

Fig. 6. Anterior ligamentous structure of the right knee.

moving joints in machinery (*1*). It is not a simple hinge joint with a single axis of rotation. Because movement of the tibia with respect to the femur is a combination of gliding and rolling actions, and because of the shape of the contacting surfaces, the instantaneous center of rotation of the knee varies with each degree of flexion. Though the exact course of the instantaneous centers for different individuals cannot be described with present knowledge, a general idea of the typical area through which they move can be had (Fig. 8).

For many years it has been common practice to divide the responsibility of weight-bearing between the below-knee stump and the thigh by use of simple hinge joints (located along the medial and lateral aspects of the knee) connecting a thigh corset to the socket and shank (Fig. 9). But, because the center of rotation of the knee moves constantly while flexion or extension takes place, any artificial joint attached on the outside of the leg and thigh that does not follow the complex pattern of the human joint will cause relative motion between the body parts and the prosthesis. Since there is not available an artificial joint that simulates normal movement, it appears highly desirable to provide the below-knee amputee with a prosthesis that does not require side joints, even though the tissues in the stump and thigh are capable of absorbing the effects of some relative motion.

Weight-Bearing

If sidebars are to be avoided, obviously all of the weight-bearing loads must be transmitted through the stump to the skeletal system. Some areas on the stump are better suited to assume these loads than others. In the light of present knowledge and technology it is necessary to design and construct the socket so that the pressures imposed on specific areas, whether by normal repeated loads encountered during walking or whether by single emergency loads, are not of values that exceed the varying tolerances of the different tissues of the stump. And just as obviously some means other than sidebars and thigh corset must be found to maintain the limb on the stump. If, however, the necessary mediolateral stability

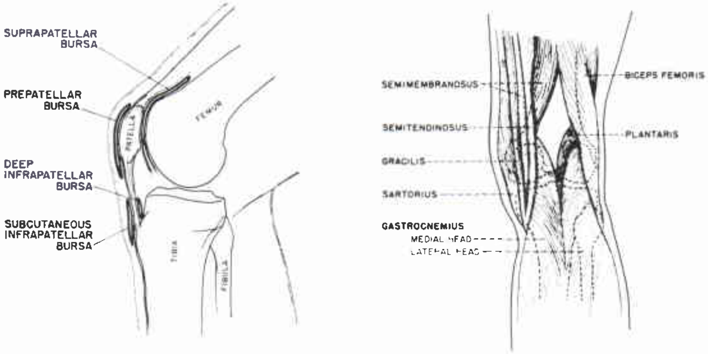

Fig. 7. Major structures that form the knee joint. From *The Patellar-Tendon-Bearing Below-Knee Prosthesis* (*4*).

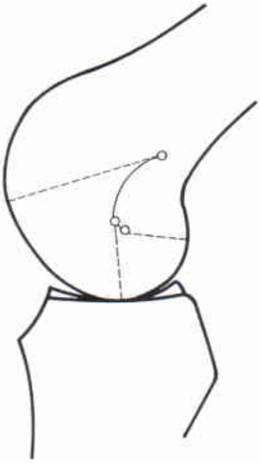

Fig. 8. Section through the medial condyle of the femur and through the tibia. The center of curvature is shown for three parts of the articular surface. As gliding occurs in the joint, the instantaneous center moves along the curve connecting these centers of curvature. From Elftman (*2*).

is not present, there is no known recourse except to use at least one sidebar and generally two.

THE PATELLAR LIGAMENT

Extension of the knee is effected by the contraction of the quadriceps muscle, so named because it has four distinct components. However, they merge into a single tendon which inserts on the anterior portion of the tibia just below its head (Fig. 6). Embedded in this tendon is the patella (Fig. 7), which is therefore a sesamoid bone, the largest in the body. Its function is twofold. While acting as a guide for the quadriceps tendon by following the vertical groove between the femoral condyles, it also tends to increase the lever arm of the quadriceps acting about the knee axis. Its cartilaginous underbody tends to produce very little friction as it slides over the anterior surface of the femur. That part of the quadriceps tendon between the patella and the insertion, frequently referred to as the patellar ligament (Fig. 10), is composed of extremely tough fibers which stretch insignificantly under normal tensile loads along the long axis and is particularly suited to take compressive loads anteroposteriorly. Because of the inextensible quality of the quadriceps tendon, there can be little or no relative motion between the patella and the tibia when the quadriceps develops tension, a condition which permits compressive loads over the quadriceps tendon, perpendicular to the fibers, up to the proximal edge of the patella. The sharp lower edge of the patella, though, is relatively unsuited for weight-bearing.

Branching out from the quadriceps tendon on each side above the patella are the lateral and medial retinacula (Fig. 6), which insert on the flares of the tibia. Like the patellar ligament, these tendons are capable of weight-bearing.

If the socket wall contains an indentation (Fig. 11) between the lower edge of the patella and the tendinous insertion, some initial tension is placed on the tendon. The upper surface of the indentation also permits the tendon to assume a load with a larger vertical component than would be the case if the indentation were not present (Fig. 12). Moreover, when the socket is aligned so that a slight amount of initial flexion is present when the wearer is in the standing position, both initial tension in the quadriceps tendon and the vertical components of load-bearing are enhanced.

FLARES OF THE TIBIAL CONDYLES

By virtue of its wedgelike shape and the nature of its thin, tough, overlying tissues, the upper portion of the tibia can assume part of the weight-bearing load by distribution of pressure over the medial and lateral flares of the condyles. Because part of the lateral flare of the tibial condyle is obscured by the head of the fibula, the medial flare offers most of the weight-bearing area.

Figure 13 shows horizontal cross sections of the tibia below the condyles superimposed on each other. Thus it can be seen that there is available potentially a considerable difference in horizontal area over which to distribute vertical forces to balance body weight. If the socket is aligned so that the stump is forced into a slightly flexed position when the wearer is standing erect, the horizontal components are reduced, the requirements for counterpressure over the posterior wall are less, and therefore the risk of pressure over the major vessels and nerves in the rear is reduced. Proximity to relatively sensitive zones like the head of the fibula (typically present under the lateral flare), the sharp tibial crest, and the rough tibial tubercle greatly reduces the useful area on the anterolateral portion. The medial flare, though seemingly smaller than the lateral, is quite effective in providing support.

Fig. 9. Some examples of the so-called "conventional" below-knee prosthesis offered by prosthetists for more than a century. Note the sidebars, corset, relatively low brim, and free space at distal end of socket.

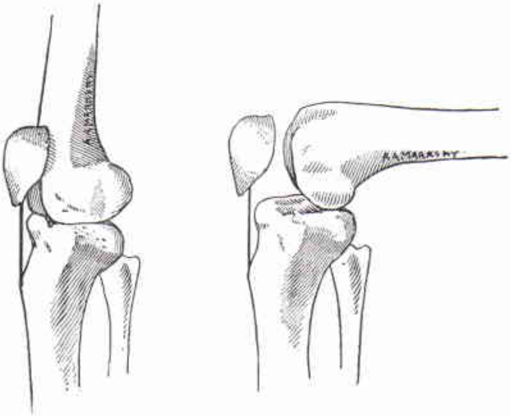

Fig. 10. Schematic drawing showing the nearly complete lack of relative motion between patella and tibia during flexion of the knee. The inextensibility of the patellar ligament prevents the patella from moving proximally with respect to the tibia. From Marks (*3*).

THE TIBIAL CREST

The shaft of the tibia is roughly triangular in horizontal section, one apex, the tibial crest, lying in the anterior portion of the leg (Fig. 13). The anteromedial wall of the tibia is covered with a thin layer of tissues and is admirably suited to assume some of the weight-bearing stresses. In the normal limb, the anterolateral wall of the tibia is covered by the tibialis anterior, which inserts in the region