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Ankle biomechanics

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The ankle is a complex joint that, with the subtalar joint and small joints of the foot, is fundamental for plantigrade, bipedal ambulation. The intact ankle efficiently dissipates the compressive, shear, and rotatory forces that are encountered while adapting to weight-bearing and ground-reaction forces. The ankle joint possesses a large contact area that is inherently stable under static load. This dynamic stability of the ankle is afforded by ligamentous support and balanced muscular forces. Given the ankle's location, orientation, and function, its mechanical efficiency can be inferred from its relative resistance to primary degenerative joint disease. Subtle changes in alignment or cartilage injury caused by trauma or inflammatory disease, however, commonly result in articular degeneration.

End-stage degenerative disease in the ankle is a debilitating malady. Surgical treatment options consist of arthrodesis, the longstanding "gold standard," and arthroplasty. Arthrodesis, however, is associated with several disadvantages, including a high rate of non- or malunion, prolonged immobilization, revision surgery, development of adjacent joint arthrosis, and decreased patient satisfaction [1].

Through its early development, arthroplasty of the ankle was fraught with technical problems and disappointing results; however, a heightened interest in ankle form and function has influenced recent designs. Several prostheses have demonstrated satisfactory intermediate results, and ankle replacement has become a viable alternative to fusion.

Structural considerations

The ankle joint is composed of three articulations: the tibiotalar, fibulotalar, and tibiofibular joints. The talus or trochlea has been described as a frustum of a cone with its apex oriented medially (Fig. 1). When viewed from the top, the body appears shaped like a wedge that narrows posteriorly. The average

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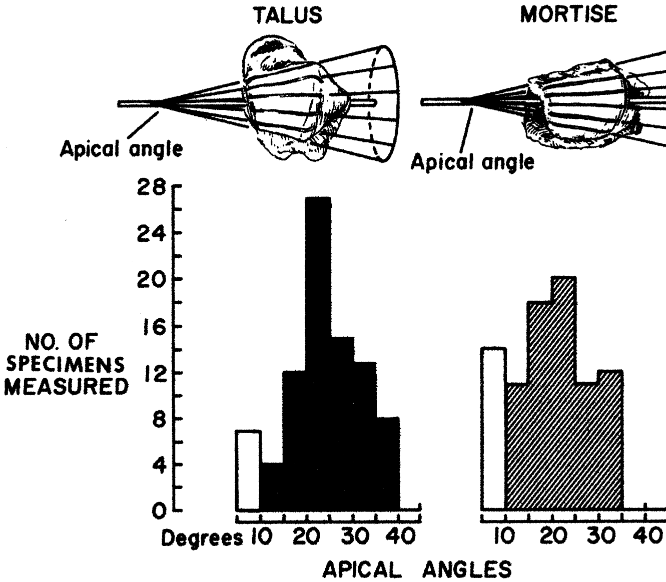

Fig. 1. Variations in apical ankles of conical surfaces of trochleas and casts of mortises. Angles were obtained by extrapolating from the end of the saw cuts toward the medial sides. All specimens whose apical angles were 10° or less were lumped together (*open bars*). (*From* Johnson JE. Functional morphology of the trochlea. In: Stiehl JB. Inman's joint of the ankle. 2nd ed. Baltimore: Williams & Wilkins; 1976, p 7–13; with permission.)

difference in transverse width from anterior to posterior reported by Inman [2] was 2.4 $\pm$1.3 mm (range 0–6 mm). The facets of both the medial and lateral malleoli are parallel to corresponding facets of the talus [2]. There is articular contact at these facets from extreme plantarflexion through complete dorsiflexion. Both tibiotalar and fibulotalar contact area was found to increase significantly with weight bearing and to be maximized at 50% of stance [3].

Inman [2] also evaluated the curvature of the medial and lateral talar facets by comparing the vertical plane of each facet to the "empiric ankle axis" that was measured experimentally. Comparing the lateral plane with the empiric axis revealed an angle of $89.2 \pm 2.8°$ (range 80°–95°). The medial talar facet subtended an angle of $83.9 \pm 5.2°$ (range 70°–93°). From this data, Inman determined that the talus was not a cylinder. Further investigation revealed that the curvature of the lateral facet was consistent with the arc of a circle. Similarly, the curvature of the medial facet approximated a circle in 86 of the 107 tali that were examined. Barnett and Napier [4] also demonstrated two different radii of curvature of the medial talar facet, which these authors attributed to two specific axes of rotation—one serving as the center of rotation from neutral to dorsiflexion, the other from neutral through plantarflexion.

In evaluating the fit of the talus in the mortise, Inman [2] found that, on the lateral side, the radii and arc of curve of the mortise was within 1 mm of the talus.

On the medial side, the average difference was 2.1 mm ± 1.1 mm (range 0–5 mm). In all specimens examined, Inman found the radius of curvature of the mortise to be larger than the talus and concluded that this difference allowed for several degrees of horizontal rotation.

Axial load of the ankle

Weight-bearing forces transmitted through the ankle have been reported to be 4.5 to 5.5 times body weight at heel rise [5]. Earlier studies have reported results based on static models that were loaded at varying sagittal positions [3,6]. The complex geometry of the mortise and trochlea of the talus influences load characteristics [3,6–10]. Reports of contact area have varied from 1.5 to 9.4 cm^2 depending on load and ankle position [11]. Several studies have reported that contact area increases as the ankle dorsiflexes form a plantarflexed position [3,9], which contrasts with reports of a similar contact area from plantarflexion through neutral position decreasing to dorsiflexion [6]. Other authors report that contact area is maximized in approximating a neutral position decreasing in both plantarflexion and dorsiflexion [7]. These discrepancies may be attributed to differences in load, position, and technique [12]. Calhoun et al. [3] found contact surface area to increase from plantarflexion to dorsiflexion and force per unit area to decrease proportionately. No significant difference was seen under loads of 490 and 686 N from dorsiflexion to 15° of plantarflexion; however, contact area decreased and average force per unit area increased (both significantly) when comparing dorsiflexion and 30° plantarflexion with neutral position and 30° plantarflexion. These authors also observed that the medial and lateral facets had the greatest contact with the tibia and fibula in dorsiflexion [3].

Michelson and Helgemo [13] used a dynamic model to determine whether transverse rotation of the talus relative to the tibia associated with sagittal motion was different from that demonstrated by the static model. They reported progressive lateral loading with concomitant unloading medially during dorsiflexion and associated external rotation. The results of these studies substantiated those of the static model.

Axis of rotation

Initially, the ankle joint was thought to function as a simple hinge whose primary axis was transverse and at a right angle to the sagittal plane. Isman and Inman [14] and Inman [2] described the axis being correlated with the transmalleolar plane and being externally rotated an average of 23°. The position and orientation of this axis accounted for coupled ankle motion. As the ankle dorsiflexed, it rotated externally and everted. Conversely, as the ankle plantarflexed, it rotated internally and inverted. Barnett and Napier [4] determined that plantarflexion and dorsiflexion took place about two distinct axes; however, they provided no description of axis transition. Recent studies have discovered that the axis of rotation is not fixed but rather changes direction and position throughout ankle motion [15–17].

Leardini et al. [16] developed a mathematical model to explain the multiaxial motion of the ankle in the sagittal plane. These authors described a four-bar linkage model that shows the talus/calcaneus and tibia/fibula rotating about one another on inextensible line segments that represent the calcaneofibular and tibiocalcaneal ligaments without resistance. The four-bar linkage model describes the talus as polyradial and polycentric with two distinct radii. With the mortise shaped like a circle with a 2.9-cm radius, the shape of the talus can be deduced. The anterior 25% of the articular surface fits a curve with a 2.3-cm radius; the posterior 75% of the talus fits a curve with a 2.8-cm radius (Fig. 2). This model (1) addresses the irregular curve of the talus and its relationship with a transient center of rotation, and (2) accounts for the anterior distraction in plantarflexion and the converse posterior distraction in dorsiflexion.

Using stereophotogrammetry, Lundberg et al. [8] evaluated the axis of rotation in eight healthy ankles. Talocrural motion was evaluated in 30° of dorsiflexion to 30° of plantarflexion, in 10° of lateral rotation to 20° of medial rotation, and in

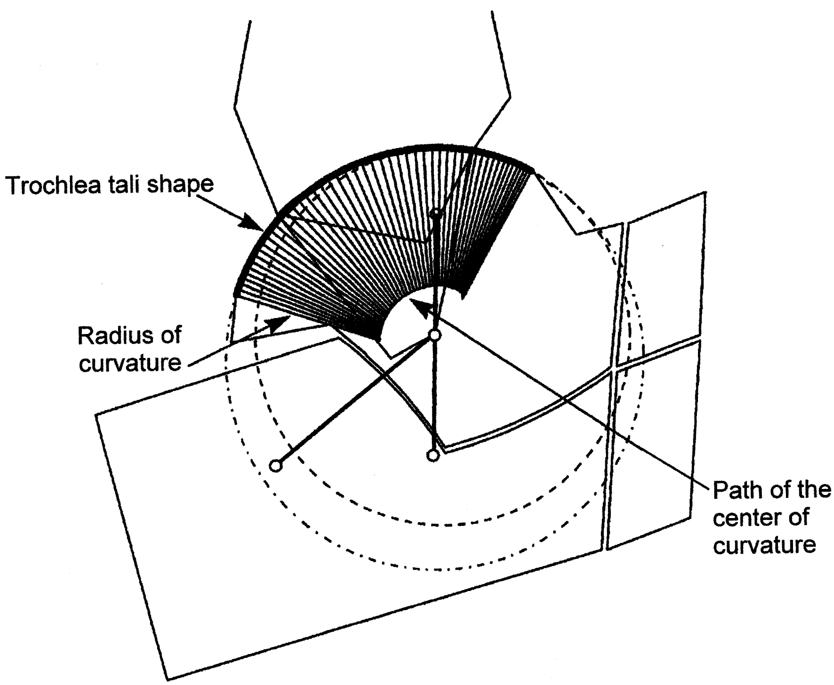

Fig. 2. Shape of the talus articular surface (*solid line*) and the best-fit circumferences for the posterior 75% (*dash-dotted line*) and the anterior 25% of the curve (dash line). (*From* Leardini A, O'Connor JJ, Catani F. A geometric model of the human ankle joint. J Biomech 1999;32:585–91; with permission.)

20° of pronation to 20° of supination. The axes were calculated and projected on the sagittal, coronal, and transverse planes (Fig. 3).

Sammarco [18] studied sagittal plane motion relative to the tibiotalar joint surface and explained that the motion between the tibia and talus takes place about multiple instant centers of rotation (Fig. 4). Between two consecutive positions, a vector depicts the resultant force and can be positioned (1) away from the talus to portray distraction, (2) parallel with the talus to express gliding, or (3) into the talus to illustrate compression. Sammarco also found that distraction was evident in forced plantarflexion, that gliding occurred as the "functional" range of tibiotalar motion was approached, and that compression was associated with forced dorsiflexion. Furthermore, Sammarco noted that an unstable ankle demonstrated normal gliding during weight bearing but that non–weight-bearing motion was grossly abnormal.

Similarly, Leardini et al [16] demonstrated the translation between the tibia and the talus in their four-bar linkage model of non–weight-bearing sagittal plane motion. These authors described that motion between the polycentric, polyradial trochlea consisted of a combination of "rolling" and "sliding" motions. In this model, rotation is dictated by the most anterior fibers of the anterior talofibular and calcaneofibular ligaments. Linardini [19] later observed that these specific fiber bundles were isometric through the range of sagittal motion of the ankle. The function of portions or fascicles of these ligaments has been reported previously [20]. The instantaneous center of rotation translates from a posteroinferior to a superoanterior position, which is consistent with several studies that have suggested that the ankle is incongruent and rotates about a transient center [21–23].

Ankle range of motion

Ankle motion occurs in the sagittal, coronal, and transverse planes [8,19]. Sagittal plane motion, which consists of plantarflexion and dorsiflexion, constitutes the greatest amount of motion in the healthy ankle. The range of sagittal plane motion has been reported to be 13° to 33° of dorsiflexion and 23° to 56° of plantarflexion [8,18,24–26]. Sammarco et al. [27] observed 10° to 15° of plantarflexion and 10° of dorsiflexion during walking. In healthy subjects, peak dorsiflexion during weight-bearing gait was 10° degrees, whereas peak plantarflexion was 25°. Stauffer et al. [5] measured sagittal plane motion to average 24.4°. Climbing stairs requires 37°; descending requires 56° [26,27].

Several factors influence sagittal plane ankle motion. Healthy older individuals demonstrate decreased plantarflexion [25,28,29]. Several authors have reported a significant increase in sagittal motion, primarily dorsiflexion, by assessing the subjects while weight bearing compared with passive measuring [29–31]. The measurement method and landmarks used obviously influence observations.

Rotation of the ankle in the transverse plane usually is reported relative to instability [32,33]; however, transverse-plane motion is coupled with sagittal

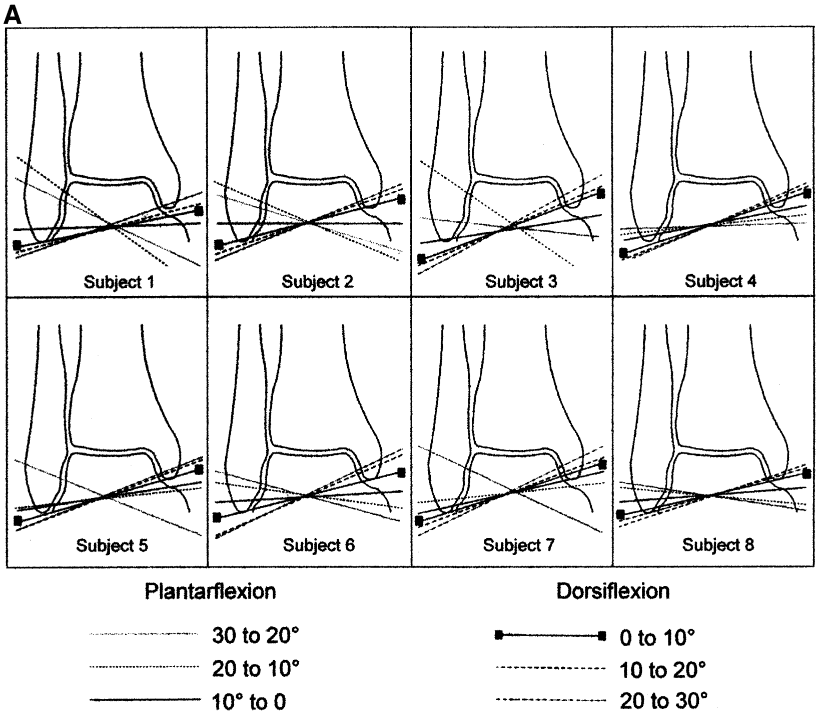

Fig. 3. (A) Individual discrete helical axes of the ankle joint of eight normal subjects for each 10° interval from 30° of plantarflexion to 30° of dorsiflexion projected onto a coronal plane. (B) Individual discreet helical axes projected onto a sagittal plane in eight normal subjects. Variations in inclination are large because axes are viewed nearly on their ends. (C) Individual discrete helical axes projected onto a horizontal plane in eight normal subjects. Axes tend to fall parallel to a transverse plane through the center of the malleoli. (*From* Lundberg A, Svensson OK, Nemeth G. The axis of rotation of the ankle joint. J Bone Joint Surg [Br] 1989;71:94–9; with permission.)

plane motion [8,13,18,34]. Transverse-plane motion was noted during normal gait by several authors [8,17,23,34]. Lundberg et al. [8] observed 8.9° of external rotation of the talus from neutral position to 30° of dorsiflexion. Lundberg et al. [17] reported a small amount of internal rotation with plantarflexion from neutral to 10° followed by external rotation at terminal plantarflexion. Michelson and Helgemo [13] reported that dorsiflexion resulted in an average of 7.2° ± 3.8° of external rotation of the foot relative to the leg with ankle dorsiflexion and 1.9° ± 4.12° degrees of internal rotation with plantarflexion. Siegler et al. [23] observed coupling between the ankle and subtalar joints in unloaded specimens with sagittal plane motion. Although not significant, these authors found that, with ankle dorsiflexion, there was internal rotation of the subtalar joint and external rotation of the tibiotalar joint [23]. This motion is thought to be caused by the tensioning of the deltoid ligament, an idea that is supported by the findings of McCullough and Bruge [32] who described greater external rotation

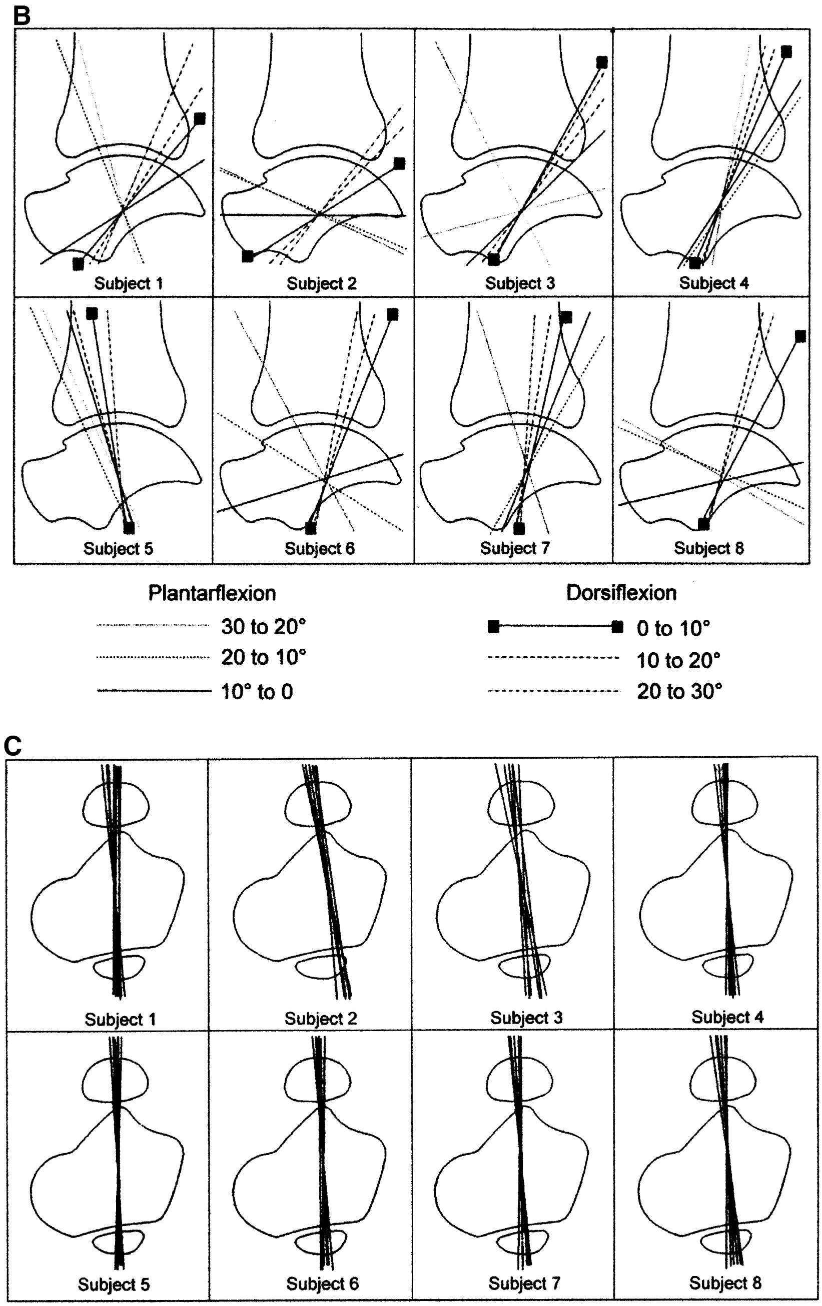

Fig. 3 (*continued*).

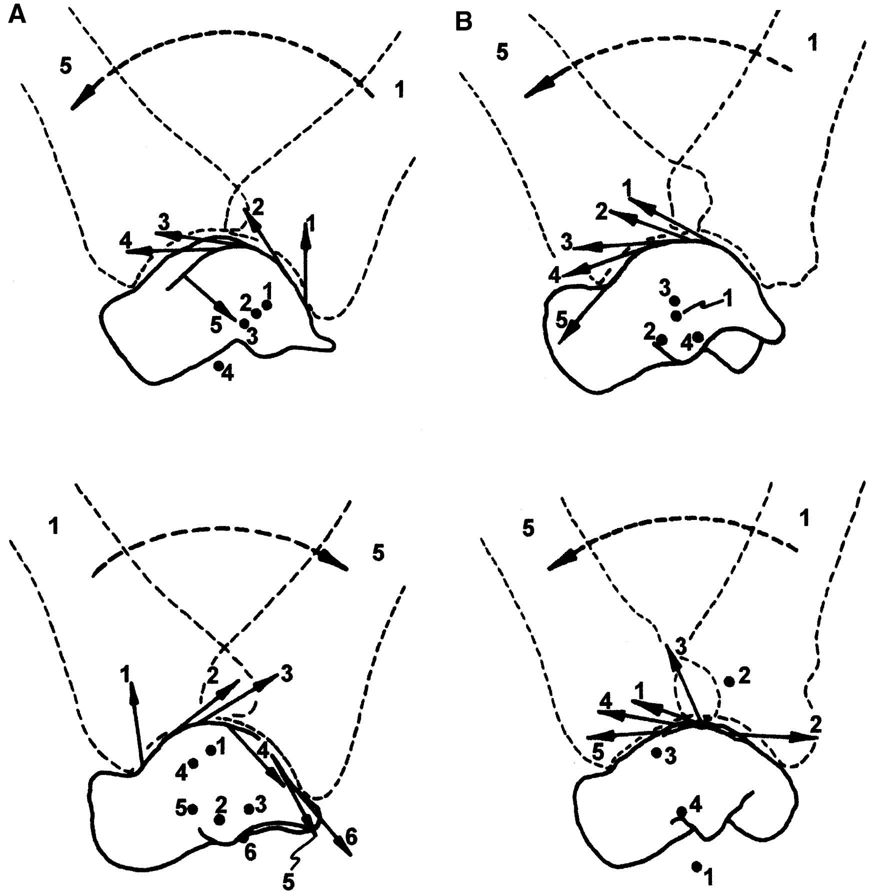

Fig. 4. (A) Surface-velocity diagram that demonstrates changes in surface velocity (*arrows*) and instantaneous centers of rotation (*dots*) during weight bearing. Initially, distraction between the tibia and talus occurs from plantarflexion through dorsiflexion. Through the "functional range," gliding is indicated by the tangential nature of the vectors followed by compression at forced dorsiflexion (*top panel*). Reversing the motion surface results in concomitant change in surface velocities and centers of rotation (*bottom panel*). (B) Diagram that demonstrates changes in surface velocity (*arrows*) and instantaneous centers of rotation (*dots*) as an ankle with lateral instability is moved from plantarflexion through dorsiflexion during weight bearing (*top panel*) and non–weight bearing (*bottom panel*). (*From* Sammarco J. Biomechanics of the ankle: surface velocity and instant center of rotation in the sagittal plane. Am J Sports Med 1973;5:231–4; with permission.)

of the talus after deltoid sectioning [13]. As dorsiflexion occurs, the foot and ankle rotate externally. When the foot is in contact with the ground, the leg rotates about a vertical axis over the fixed foot.

Coronal motion is described as varus or valgus rotation. Michelson et al. [35] observed that both maximal dorsiflexion and plantarflexion of the ankle were associated with internal rotation and inversion of the ankle. They attributed coronal plane motion to the position of the deltoid ligament.

Ligament function

Stability of the ankle joint during weight bearing depends on several factors, including articular congruity, ligament position, and ankle position when the stress is applied [32,33,36,37]. The lateral ligament complex consists of the anterior talofibular, calcaneofibular, and posterior talofibular ligaments. The anterior talofibular ligament originates from the anterior margin of the distal fibula and inserts at the lateral tubercle of the talus. The calcaneofibular ligament originates at the distal tip of the fibula and inserts at the lateral wall of the calcaneus, spanning both the ankle and subtalar joints. Because of its unique position (parallel to the axis of rotation of the subtalar joint), however, the calcaneofibular ligament stabilizes the ankle while not restricting subtalar motion. The anterior talofibular and calcaneofibular ligaments are oriented approximately 90° from one another. Inversion stress is resisted by the anterior talofibular ligament during plantarflexion and by the calcaneofibular ligament during dorsiflexion. Harper [36] determined that these structures were the primary restraint to anterior translation of the talus.

The deltoid ligament, which provides medial support, consists of a superficial and deep portion. The superficial portion takes its origin from the anterior and posterior colliculi of the medial malleolus and inserts broadly into the calcaneus, talar neck, and navicular [2,34]. The deep portion originates from the medial aspect of the medial malleolus and consists of anterior, intermediate, and posterior fascicles that insert at the medial wall of the talus.

Several studies [37,38] have reported the effects of the lateral ligaments on axial rotation of the loaded ankle. Sommer et al [38] observed increased anterolateral rotation of the talus with ankle dorsiflexion but noted no increase in the ratio of tibial rotation to plantarflexion after sectioning lateral ligaments. These authors concluded that, with inversion, articular morphology conferred stability [38]. In an attempt to quantify the effect of sequential ligament transection on tibial rotation and calcaneal eversion–inversion associated with axial loading. Independent of load, sectioning the anterior talofibular ligament resulted in increased internal rotation of the tibia during dorsiflexion. Sectioning the calcaneofibular and posterior talofibular ligaments did not increase rotation further [37]. Hinterman et al [37] observed that rotation of the tibia that occurred after sectioning of the anterior talofibular ligament was more profound from neutral to plantarflexion compared with that observed in 10° and 20° of dorsiflexion. When the deltoid ligament was sectioned, no tibial rotation was observed, which is consistent with the findings of Michelson et al [39] whose report suggested a role for the deltoid in motion coupling in addition to stabilization.

Harper [36] found no lateral displacement or valgus tilt of the talus with an intact deltoid ligament and noted no increase in valgus tilt with sectioning of the deep or superficial portions of the deltoid ligament individually. Valgus tilt, however, was possible with sectioning of both deep and superficial portions. Further sectioning of the posterior ankle-joint capsule increased valgus tilt [36]. Sectioning the superficial deltoid ligament did not result in increased anterior translation or lateral shift of the talus [36], whereas sectioning of the deep deltoid ligament did result in significant lateral shift and anterior translation of the talus [36].

Many authors have reported that the stability of the ankle under axial load is conferred by the bony construct of the ankle mortise [2,17,32,33,36].

Lateral malleolar function

The weight-bearing function of the fibula has been described in numerous reports [2,40–42]. In a simulated weight-bearing model, Lambert [41] reported that the fibula supports approximately 16% of the weight of the leg, a measurement substantiated by Skraba and Greenwald [43] who found that the fibula bore 15% to 20% of a physiologic load. Takebe et al. [42], however, observed that only 6.4% of total weight was borne by the fibula and found that the force borne by the fibula varied with ankle position. These authors also observed that fibular loading increased with dorsiflexion and decreased with plantarflexion.

Where the fibula was thought to provide static support during weight bearing, the dynamics of the lateral malleus during sagittal motion of the ankle also have been described [4,34,40,44]. Fibular motion has been observed to occur in the transverse, sagittal, and frontal planes [44].

Karrholm et al [40] noted lateral and distal translation of the fibula with passive dorsiflexion from neutral, which caused an average lateral displacement of 1.0 to 1.4 mm and from 0.1 to 0.5 mm of distal translation. With active loading during a simulated push-off maneuver, Scranton et al. [45] reported 2.4 mm of distal translation, which was explained by the action of the peronei, flexor hallucis longus, and posterior tibialis muscles [45]. Karrholm et al. [40] also described posterior translation of the fibula in the sagittal plane in eight of nine subjects, who demonstrated posterior translation of the distal fibula from neutral through dorsiflexion.

Close [34] and Barnett and Napier [4] reported Rotation of the fibula about its longitudinal axis. During forced dorsiflexion of the ankle, the latter authors measured 3° lateral rotation in one patient, attributing this to the joint-surface orientation at the proximal tibiofibular syndesmosis. Ogden [46] speculated that the lateral rotation of the fibula was modulated by the position of the knee and the tension of the collateral ankle ligaments.

Gait analysis

The orderly sequence of events that cause weight transfer and energy production during normal walking are divided into weight-bearing (stance) and non–weight-bearing (swing) phases (Fig. 5). The stance phase is further

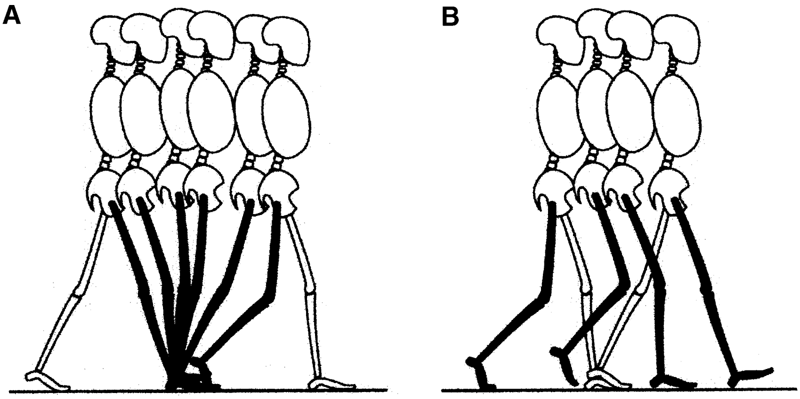

Fig. 5. (A) During stance, the supporting (*solid*) limb provides an advancing base that rolls forward over the foot. (B) During swing, the same limb advances itself to its next position for weight acceptance. (*From* Perry J. Gait analysis: Normal and pathological function. Thorofare, NJ: Slack; 1992,p 19–47; with permission.)

divided into 4 phases. Fig. 6 represents mean ankle sagittal plane motion in normal walking by healthy subjects. The weight-bearing phase begins with heel strike, after which the entire lower extremity rotates internally. The ankle rapidly plantarflexes from a position of slight dorsiflexion under control of the tibialis anterior.

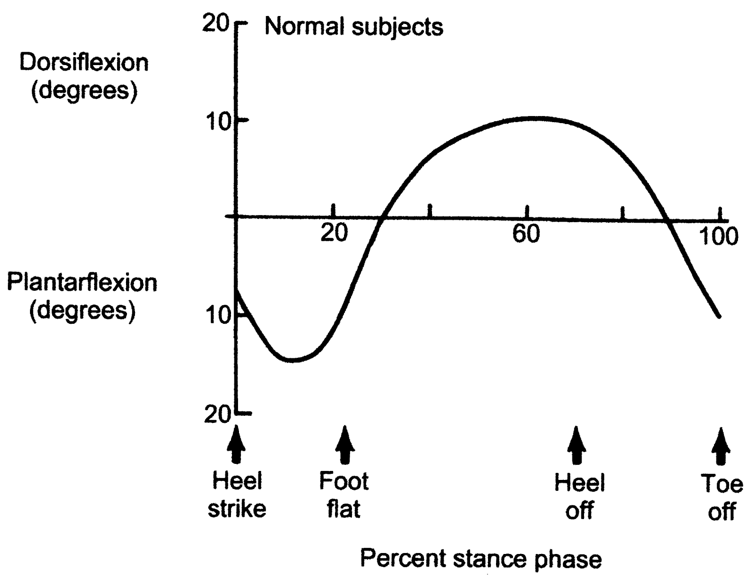

Fig. 6. Mean pattern of ankle joint motion in normal subjects. (*From* Stauffer RN, Chao E, Brewster RC. Force and motion analysis of the normal, diseased, and prosthetic ankle joint. Clin Orthop 1977;127:189–96; with permission.)

During the second phase of stance, or single-limb support, the center of mass passes over the weight-bearing limb. Dorsiflexion of the ankle occurs as the foot remains flat on the ground, and the leg rotates medially about the vertical axis [47]. The third phase of stance is marked by heel rise. The ankle and foot plantarflex, which is accompanied by lateral rotation of the leg until the toe is lifted off the ground. At the time of contralateral heel strike, plantarflexion becomes passive as weight transfer occurs.

Stauffer et al [5] found that shear forces result from the internal tendon forces and the external inertial forces of the body as it moves over the foot. These tangential forces are described as being directed "aft" or negatively as the foot decelerates after contact with the ground and during heel strike and flatfoot stance. The shear forces then become positive, acting in a forward direction at heel rise until the toe is lifted off the ground [5].

Ankle replacement

The problems that plagued early prosthetic designs have been reported extensively [48–52]. Early constrained designs were prone to loosening because all forces other than sagittal rotation were transferred to the bone–cement interface. Conversely, unconstrained designs allowed motion in multiple planes and therefore were more efficient in force distribution. Because these designs relied primarily on soft tissue support, however, they were unstable [50,53,54]. Subsidence of one or both components also has been reported [48,50,55,56].

The systems used today are semiconstrained, which allows greater freedom of motion and provides stability [57–59]. The Alvine Agility (Depuy, Warsaw, IN), which has been approved by the Food and Drug Administration, is a three-component, porous-coated system. The talar component is externally rotated 23° relative to the sagittal plane, and the tibial component has three sides, which essentially resurfaces the ankle mortise. This design allows (1) medial/lateral translation and axial rotation in addition to the rotation and gliding associated with plantar and dorsiflexion, and (2) 60° of sagittal plane motion. The tibial component is supported by the anterior, posterior, and lateral tibial cortex and by the medial cortex of the fibula. The distal tibiofibular syndesmosis is fused at the time of implantation.

The LCS/Buechal-Pappas (New Jersey LCS Total Ankle, Endotec, South Orange, NJ) is a three-component, porous-coated system that has a mobile bearing made of an ultrahigh-molecular-weight polyethylene and a metal-onlay talar component. The tibial component is supported by subchondral bone [57]. The articular facets of the tibia and fibula are maintained for structural support and stability. This design allows 60° of congruent flexion and extension and 30° of congruent axial rotation [57].

The Scandinavian Total Ankle Replacement (STAR, Waldemar Link, Germany) consists of an anatomically shaped stainless steel cap that resurfaces the talus [58]. The tibial component is made of polyethylene and is congruent with the talar component on the articular side. The back side of the component has two bars that are placed through parallel drill holes in the subchondral bone of the tibia. The articular surfaces of the medial and lateral malleoli are maintained.

Summary

The developers of a successful prosthetic implant for ankle replacement must consider the biomechanical properties that are unique to this complex joint. The prosthesis needs to provide structural support while allowing for motion in the sagittal, transverse, and coronal planes. Although the design must conform to and function within the soft tissue constraints of the ankle, it is only a component of the overall success. Paying attention to leg alignment and meticulous soft tissue balancing is essential to a satisfactory outcome.

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