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TECHNICAL NOTE

A METHOD TO DETERMINE BONE MOVEMENT IN THE ANKLE JOINT COMPLEX *IN VITRO*

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Abstract An experimental set-up has been developed to quantify motion of bone structures in the ankle joint complex of human cadaver specimens under conditions approximating physiological joint loading. The device allows to load the foot/leg specimen along the axis of the tibia, and muscle forces can be simulated by clamping the extrinsic tendons of the foot. Additionally, an axial moment can be applied to the tibia. A variety of foot movements can be induced by rotating a foot plate around an arbitrary axis in the horizontal plane. The input force which produces the movement at the foot is applied to the entire sole of the foot. A forefoot fixation allows for the natural adaptation of the midfoot and hindfoot which occurs during loading of the specimen. Bone pins were placed in the tibia, talus, calcaneus and navicular, and three reflective markers were attached to each pin in order to record the bone movements with a video system. Intersegmental rotations in the talo-crural, talo-calcaneal, and talo-navicular joints were calculated in three dimensions, compared for different loading and ligament integrity conditions, and related to a functionally/anatomically described foot position. Repeated measurements of relative bone orientations indicated a reproducibility better than 2°; the slope of the curves, representing the kinematic coupling, was virtually identical between repetitions. It is proposed that this method simulates multidirectional AJC compression similar to loading situations during locomotion. © 1997 Elsevier Science Ltd. All rights reserved.

Keywords: Ankle; *In vitro*; Kinematics; Multidirectional load; Muscle.

INTRODUCTION

Cadaveric studies investigating the ankle joint complex (AJC) have typically attempted to assess the effect of selected anatomical structures for simulated natural foot movements and physiological joint loadings. However, all previous studies had at least four of the following limitations: (1) Muscle forces were neglected. (2) Weight bearing was not taken into account. (3) Movement patterns were derived from a series of static measurements. (4) Foot movements were studied around the three coordinate axes in isolation. (5) The axes of rotation of the fixation device were aligned with the anatomical axes of the AJC, thereby unrealistically decreasing the moments produced by vertical loading. (6) The movement of the AJC was restricted. (7) The calcaneus was rigidly fixed, which may have changed the transmission of load throughout the foot. (8) Either the movement was not three-dimensionally analysed or the three-dimensional presentation of movement was difficult to correlate to a certain foot position.

The purpose of this project was to develop an experimental set-up for studying movement and functional stability of the AJC which overcomes the limitations associated with previous studies.

METHODS

The leg/foot specimen was mounted in a fixation device (Fig. 1) consisting of a distal part to induce movement to the foot and a proximal part to apply vertical loading to the leg. The fixation device distal to the specimen (Fig. 2) consisted of a foot plate (FP), a ground plate (GP), and a block (B) placed between the two plates. Shortened thumbtacks were glued to the top of the foot plate to provide friction between the sole of the foot and the foot plate. The foot plate rotated around a horizontal axis (A) located 11 mm below the plate. The orientation of the horizontal axis could be changed by rotating the block (B) around a vertical axis which was secured by two screws (US, LS). A medial-lateral orientation (0°) of the horizontal axis was used to mimic foot flexion; an anterior-posterior orientation (90°) to mimic eversion/inversion; (3) an axis between those two axes (45°) to mimic pronation/supination shown in Fig. 2; and an axis between the first two axes (135°) to mimic a combination of dorsal-extension and inversion or plantar-flexion and eversion. Each axis could be reproduced by means of cursor markings on the ground plate and an arm-like indicator (I). The fixation device fixed the tibia proximally with a stem (S) which was inserted into the medullary channel and connected to a rod (R). A coupling mechanism (CM) described earlier (Hintermann *et al.*, 1993) allowed alignment of the rod with the longitudinal axis of the tibia. The rod could be adjusted to the length of the tibia and terminated in a ball-and-socket joint (J). Linear bearings (LB) allowed for vertical translation of the proximal loading device.

In order to simulate muscle forces, nine extrinsic tendons (triceps surae, flexor hallucis longus, flexor digitorum longus, tibialis posterior, tibialis anterior, extensor digitorum longus, extensor hallucis longus, peroneus longus, and peroneus brevis) were connected to wires with mini-clamps. Tension was applied along the lines of action of the muscles using an estimated origin of the corresponding muscles, pulleys mounted on a rigid frame (PD) and weights suspended from the end of the wires. The muscles were divided into four functional groups: dorsal-extensors (extensor digitorum and hallucis longus, and tibialis anterior), plantar-flexors (triceps surae, flexor digitorum and hallucis longus), pronators (peroneus longus and brevis) and a supinator (tibialis posterior). Each group was loaded with a weight which could be selected freely. A bar system (BS) allowed for distribution of the applied forces to the individual muscles in one muscle group, according to the relative physiological cross-sectional areas of the corresponding muscles taken from the literature. Fibular loading through muscle forces was simulated as follows: three pulleys were attached to the fibula (PF) at the estimated centres of origin of the peroneus longus, the flexor

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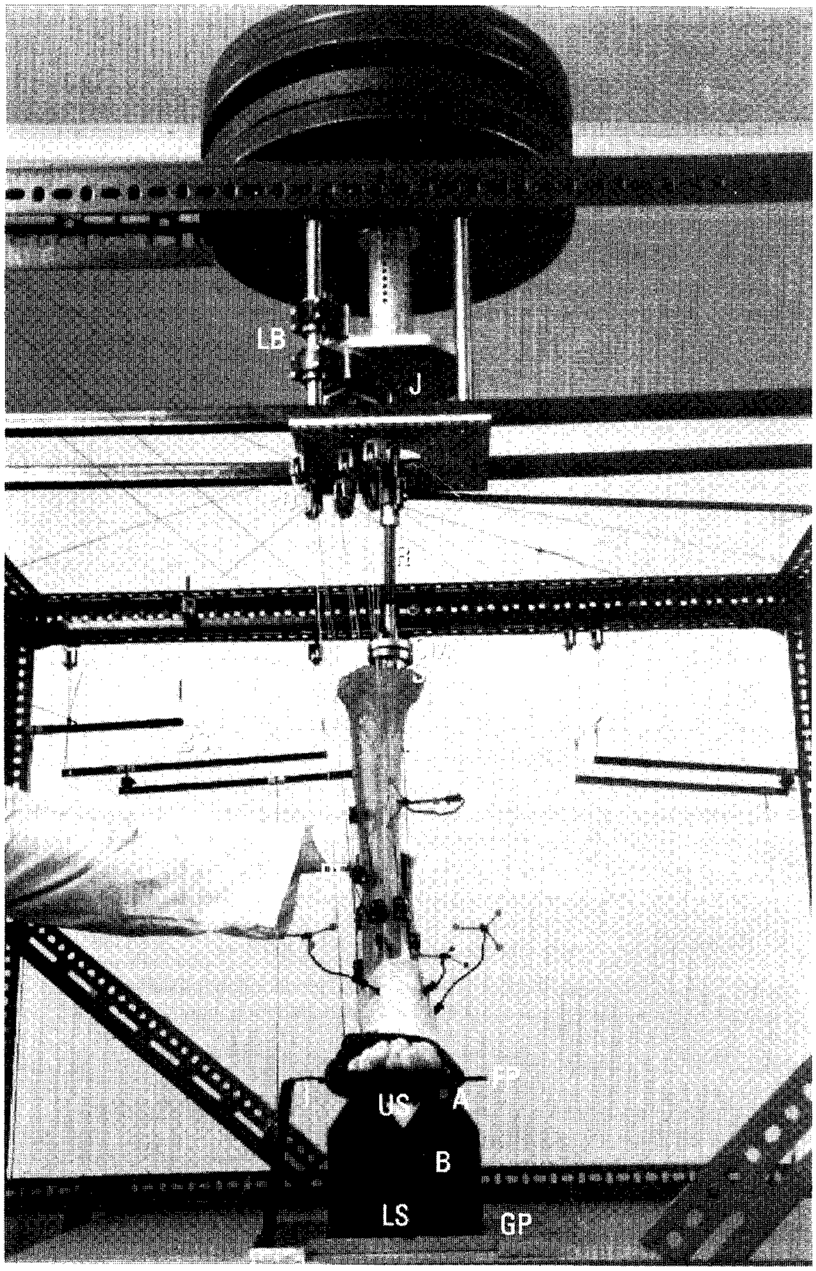


Fig. 1. Experimental set-up. The symbols used represent: FP, foot plate; GP, ground plate; B, block; A, axis; US, upper screw; LS, lower screw; I, indicator; IL, input lever; R, rod; CM; coupling mechanism; J, ball-and-socket joint; LB, linear bearings; LR, lever fixed to the rod; PD, pulley at the device; PF, pulley at the fibula; BS, bar system.

hallucis longus, and the extensor hallucis longus (Fig. 3). The wire of the three respective muscle–tendon units was redirected at the fibula pulley by approximately 180°. By applying tension to a wire in such a direction, the force at the fibular origin was doubled with respect to the force at the insertion. This doubling approximated the effect of forces of muscles originating at the fibula for which fibular loading was neglected. In order to simulate weight bearing, the leg/foot specimen could be vertically loaded. Additionally, a moment of up to 5 N m could be applied to the tibia about its longitudinal axis with a lever fixed to the rod (LR) to simulate rotational torques originating from the knee joint.

Ten through-the-knee amputees of human cadavers were studied [8 different bodies; 6 female legs (median age 62, range 55–68), 4 male legs (median age 69, range 63–74)]. Proximally, the tibial plateau was cut off. The fat pad of the heel was removed, leaving the plantar aponeurosis intact. Upon loading the leg/foot, the shortened thumbtacks were pressed into the cortical bone of the calcaneus preventing translation but allowing rotation of the calcaneus relative to the foot plate. Steinmann pins were drilled into the tibia, talus (lateral head–neck border), calcaneus, and navicular with the skin and fascia widely incised. The pins were not constrained by soft tissue or adjacent bones in any of the foot positions tested.

The anterior-posterior (y) axis of the foot was defined as a straight line through the centre of the calcaneal tuberosity and the centre of the distal metatarsal head of the second digit. In the frontal plane the proximal-distal (z) axis of the tibia was defined as a straight line passing half-way between the tips of the malleoli distally, and between the medial and lateral border of the bone proximally. In the sagittal plane, the references used were the distal and proximal midpoints of the tibia, determined as the anterior and posterior border. The anatomical neutral position was defined with the foot plate horizontal and the proximal-distal axis of the tibia vertical. In this position, the medial-lateral (x) axis was defined as the vector cross-product of the anterior-posterior (y) axis of the foot and the proximal-distal (z) axis of the tibia.

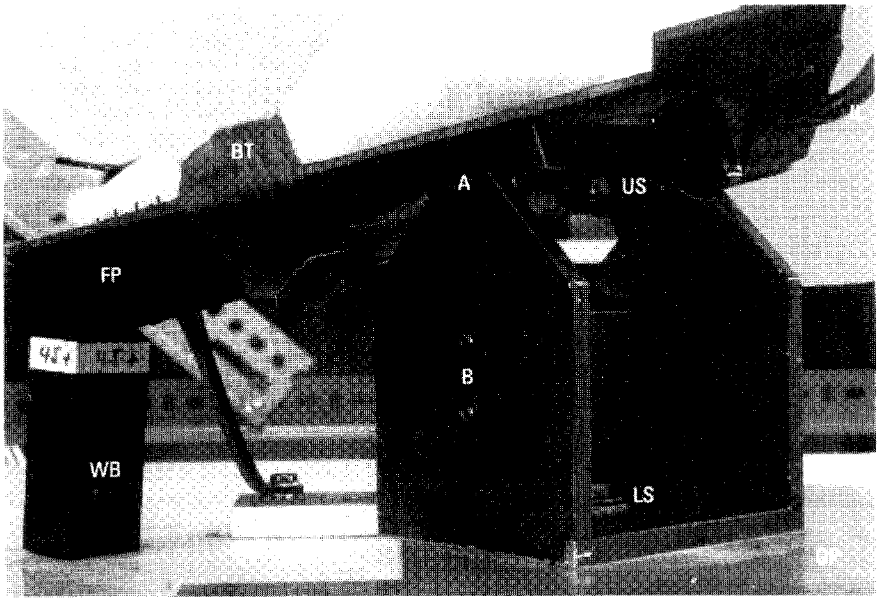


Fig. 2. Sideview of the device distal to the foot. The symbols used represent: FP, foot plate; GP, ground plate; B, block; A, axis; US, upper screw; LS, lower screw; I, indicator; IL, input-lever; WB, wooden block; BT, bicycle tube.

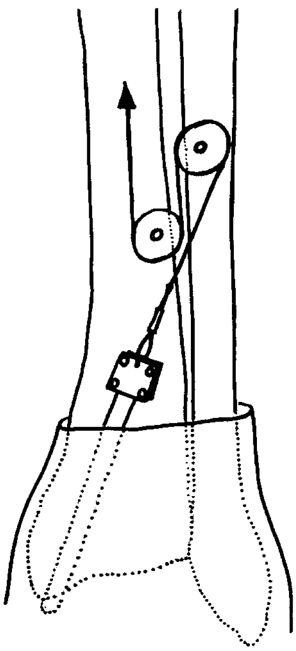


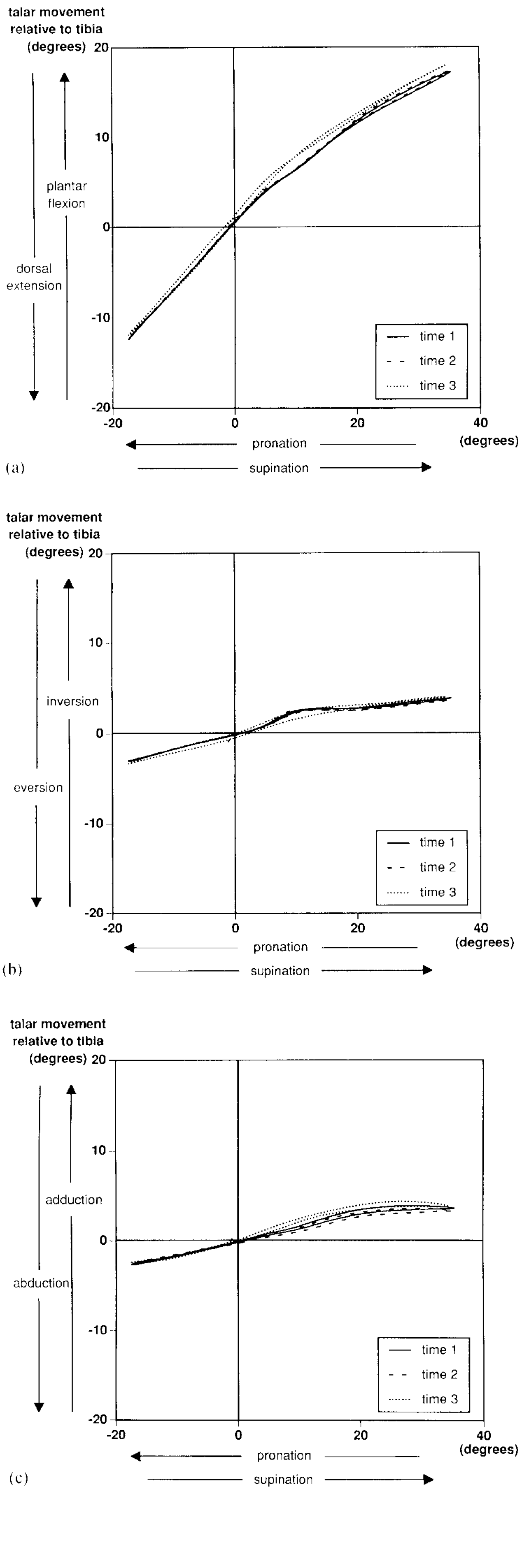
Fig. 3. Illustration of force application to the tibia. In a posterior view of the right mortise inclusion of fibular loading is shown. Tension is applied to the flexor hallucis longus tendon. The wire is redirected at the fibula-pulley for approximately 180°. Applying tension to a wire in such a direction the force at the fibular origin is doubled with respect to the force at the insertion. The wire between the two pulleys is vertical.

The anterior-posterior axis of the foot was aligned parallel to the inversion/eversion axis (A) below the foot plate. The in/eversion axis was set for displacing the foot plate medially-laterally until it neither everted or inverted under vertical loading. The flexion axis was set for displacing the foot plate anteriorly-posteriorly until it neither plantar-flexed or dorsal-extended under vertical loading. The procedure was repeated until all equilibriums were achieved. Once the equilibrium position was found, the ground plate was fixed to the table. In order to achieve controlled input kinematics at one point of the forefoot, the distal metatarsal head of the fifth digit was fixed to the foot plate with a screw. A bicycle tube (BT) was then tightened around the digits at the level of the distal metatarsal heads and the foot plate.

A three-dimensional motion acquisition system (Expert Vision EV HiRes) was used to quantify the movement of selected bones. Four video-cameras (HSC-250, Motion Analysis Corp., Santa Rosa, CA) with 12.5 mm zoom lenses calibrated for lens distortions (EVa 1.08 software) recorded the movement of 12 reflective spherical markers (three for each bone pin) at a sampling frequency of 30 Hz. Three cameras viewed the markers from the anterior, medial, and lateral side, respectively; the fourth camera provided a top view. A high precision calibration frame (volume 600 cm^3) with 16 control points was used to calibrate the field of views. The raw data were stored on a SUN SPARC classic workstation. The neutral position of each vertically loaded specimen was recorded where the segmental coordinate systems of the tibia, talus, calcaneus, and navicular coincided with the global coordinate system. Joint orientations were determined with joint coordinate systems (Cole *et al.*, 1993; Grood and Suntay, 1983). In the talo-crural joint, the proximal segment was the tibia, and the distal segment the talus. In the talo-calcaneal joint, the proximal segment was the talus, and the distal segment the calcaneus. In the talo-navicular joint the proximal segment was the talus, and the distal segment the navicular. The first body-fixed (flexion) axis was the x-axis of the proximal segment of the joint, the second body-fixed (long) axis was the y-axis of the distal segment of the joint.

Rotation of the foot plate was induced manually through an input lever (IL). The movement started in the anatomical neutral position, rotated first in one direction, then in the opposite direction through an entire arc of motion, and ended in the anatomical neutral position. Typically, the duration of such movements was about 5 s. The arcs of motion of the foot plate were set arbitrarily: −15–25° for dorsal-extension/plantar-flexion, −15–25° for pronation/supination, −15–25° for eversion/inversion, and −15–20° for dorsal-extension and inversion/plantar-flexion and eversion. The arc of motion was limited by wooden blocks (WB) located under the foot plate.

The loading conditions for the repeat measurements were as follows: vertical load 430 N; tendon load 4 × 180 N; no moment at the tibia. To evaluate the ability of this device to reproduce the input movement, to maintain the foot orientation relative to the foot plate, and to reproduce the kinematics in ten specimens, measurements were taken 0 (time 1), 20–40 (time 2) and 60–120 min (time 3) after applying the vertical load. Several measurements with different loading conditions and axes orientations were performed in between the described trials.



Intersegmental rotation for the selected segments in the joint coordinate systems were calculated using the Kintrak motion analysis software (Motion Analysis Corp., Santa Rosa CA, version 5.0 alpha). The data were filtered with a second order, low-pass Butterworth filter (cut-off frequency 0.7 Hz). The relative orientation between tibia and calcaneus described the input movement of the foot. In order to describe pronation/supination and dorsal-extension and inversion/plantar-flexion and eversion as a major movement around one axis, a coordinate system transformation for the calcaneus was calculated such that one cardanic axis was parallel to the axis (A) of the device. Rotations in the talo-crural, talo-calcaneal, and talo-navicular joints were considered responses to the input.

RESULTS AND DISCUSSION

All tested specimens showed retest differences which were < 2° for the three rotations quantified for each joint [illustrated in Fig. 4(a)–(c) for the typical specimen #1]. The extreme position of repeated movements showed an average difference of about 1.3° for all specimens. Loading of the specimens produced a systematic shift throughout the entire arc of motion leaving the slope (kinematic coupling) constant.

The results showed high repeatability for the measured joint orientations as long as the anatomical structures in the specimen did not change their mechanical properties. However, the foot and leg changes its mechanical properties as a function of time of applied load depending on the creep behaviour of its soft tissue structures. Consequently, repeated measurements of the neutral positions are required to account for creep dependent changes in the mechanical behaviour of the specimen.

The present method used only constant muscle-tendon loading. Future developments should include variable muscle-tendon load to allow an even more realistic simulation of a specific movement pattern during locomotion.

CONCLUSION

A method has been developed to quantify movement in the AJC *in vitro*. It has been shown that the method is reliable. It is proposed that this method simulates multidirectional AJC compression similar to loading situations during locomotion.

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Fig. 4(a)–(c). Angle–angle diagrams illustrating the repeatability of talo-crural joint angles of specimen #1 in three dimensions [medial-lateral axis in (a); anterior-posterior axis in (b); proximal-distal axis in (c)] for movement around the pronation/supination axis.