THE UNIVERSITY OF CALGARY

Skeletal Lower Extremity Motions During Running

by

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A DISSERTATION

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES

IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE

DEGREE OF DOCTOR OF PHILOSOPHY

DEPARTMENT OF MEDICAL SCIENCE

CALGARY, ALBERTA

AUGUST 1998

0-612-34702-8

ABSTRACT

The purpose of this study was to quantify the effects of (i) medial foot orthoses and (ii) shoe sole modifications (lateral heel flares) on skeletal movements during the stance phase of running, to compare (iii) barefoot versus shod running, and to describe (iv) the movement coupling between selected segments of the lower extremity. Marker triads were attached to intracortical bone pins inserted under standard local anesthesia into the calcaneus, tibia and femur of five subjects. The subjects ran barefoot, with a normal shoe, with three shoe soles and with two orthotic modifications. The trials were recorded during the stance phase of running using three high speed cine cameras operating at 200Hz. Tibiocalcaneal and tibiofemoral movements were calculated using three-dimensional marker reconstruction and a joint coordinate system approach.

Medially placed foot orthoses did not substantially change tibiocalcaneal movement patterns during running. Differences between subjects were significantly larger (up to 10°) than between orthotic conditions (1° to 4°; $p<.01$). Significant orthotic effects across subjects were found only for total internal tibial rotation ($p<.05$). Lateral heel flare effects on tibiocalaneal movements were small and unsystematic. Total shoe eversion and shoe eversion velocity were found to be approximately twice as large as the respective bone eversion. Tibiocalcaneal movements were similar in barefoot running compared to running with shoes. Only one specific shoe modification (posterior orthosis) showed significant ($p<.01$) differences to barefoot running, all other test variable comparisons were not significant. Movement coupling was observed between various segments at the lower extremity (shoe-calcaneus, calcaneus-tibia, and tibia-femur) and was found to be subject and shoe dependent. Generally, movement coupling varied in distinct phases between heel strike and take-off. It was suggested that the input to calcaneus-tibia coupling was from distal to proximal, and that to tibia-femur coupling (at least partially) from proximal to distal.

A number of possible factors were identified that may have influenced the results, including the test shoes used, the shoe sole construction, and the application of local anesthesia and its possible influence on proprioceptive feedback.

The results of this *in-vivo* study suggest that (a) orthotic effects and effects of shoe sole modifications on skeletal movement patterns are small and subject specific and that these effects may be mechanical as well as proprioceptive, (b) calcaneal and tibial movement patterns may not be changed substantially when using shoes compared to running barefoot, but differences may occur when extreme shoe modifications are used (c) movement coupling varies between heel-strike and take-off and takes place in phases between various segments of the lower extremity during the stance phase of running.

PREFACE

Chapters 4 to 7 are based on the following manuscripts:

- Stacoff, A., Reinschmidt, C., Nigg, B.M., Bogert, A.J. van den, Lundberg, A., Denoth, J., and Stüssi, E. Effects of foot orthoses on skeletal motion during running. Submitted to *Clinical Biomechanics*.

- Stacoff, A., Reinschmidt, C., Nigg, B.M., Bogert, A.J. van den, Lundberg, A., Denoth, J., and Stüssi, E. Effects of shoe sole construction on skeletal motion during running. Submitted to *Medicine and Science in Sports and Exercise.*

- Stacoff, A., Nigg, B.M., Reinschmidt, C., Bogert, A.J. van den, and Lundberg, A. Tibiocalcaneal kinematics of barefoot versus shod running. Submitted to *Journal of Biomechanics.*

- Stacoff, A., Reinschmidt, C., Nigg, B.M., Bogert, A.J. van den, Lundberg, A., Denoth, J., and Stüssi, E. Movement coupling in the lower extremities during the stance phase of running. Submission: Submitted to *Foot and Ankle.*

This thesis has been written as a compilation of (stand-alone) papers arranged in chapters 4 to 7. Thus, these chapters contain redundant information mainly in the "introduction" and "methods" sections, since the rational and the methods of these papers are similar. On the other hand, parts of chapters 1 to 3 can be found in the introduction and methods of chapters 4 to 7.

ACKNOWLEDGEMENTS

I would like to express my sincere thanks to the following individuals and institutions. Without their help this thesis would not have been possible.

- Dr. Benno Nigg, who gave me a second chance to achieve what I wanted to do more than 15 years ago.
- Dr. Edgar Stüssi (Biomechanics Laboratory, ETH Zürich) for his continuous support also during the years of waiting for this second chance.
- Dr. Christoph Reinschmidt, whom I introduced once into biomechanics and who introduced me now into many new techniques of the field.
- Dr. Ton van den Bogert who navigated not only Christoph Reinschmidt through three-dimensional space, but made it also happen that this set of data was not lost in space.
- Drs. Ron Zernicke, Walter Herzog, Peter Goldsmith and Mario Lafortune for serving on my thesis committee.
- Dr. Arne Lundberg, who made sure that all medical difficulties were handled with the necessary experience at the Huddinge Hospital in Stockholm.
- Dr. Norman Murphy and Andrzej Stano for their extraordinary help during the experiments.
- Dr. Jachen Denoth with his critical mind and guidance through the difficult parts of the work.
- Heinz Strebel and Evangelos Avramakis for the big help in adapting computer programs and their patience in providing many new versions of diagrams.
- Russell Lawson in helping me to find the right words between the different styles of the English language.
- Andrea Ming, Jakob Waser, Thomas Umbricht and Marco Demont in helping me to shorten the long hours of digitizing the data.

- Roland Müller, who helped with critical input to straiten out some difficult data and Dr. Hans Gerber and Michael Slameczka for troubleshooting computer problems at the right place and the right time.
- All fellow graduate students at the University of Calgary and the ETH in Zürich for their stimulating discussions.
- All five subjects who participated in the study, taking the risk of an invasive experiment.
- The foilowing institutions helped to support this study: The University of Calgary, The Biomechanics Laboratory of the ETH Zürich, the ADIDAS sport shoe company, the Swedish Defense Materials Administration.

DEDICATION

to Moni, Yvonne and Fabian, and to Hans and Hanni

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1. INTRODUCTION

Running is one of the basic forms of human locomotion and has attracted the interest of scientists for more than 150 years (Weber and Weber, 1836, Muybridge, 1887; Marey, 1895; Braune and Fischer, 1895-1904). Running, as a physical activity, increased in popularity during the 1970's and 1980's (Cavanagh, 1990), but has encountered some recent decline (Stephens and Craig, 1990), although it is still a sport performed by millions of people. With the running boom, the importance of shoe construction, specifically of the shoe sole and of foot orthoses, has been recognized by a number of investigators. Kinematic analyses were performed to quantify the effects of specific shoe sole modifications as well as alterations of the placements of orthoses. The results of studies related to shoe sole modifications (Clarke et al., 1983; Frederick, 1986; Nigg et al., 1987; and Stacoff et al., 1988) showed that an increase in shoe flare on the lateral side of the rearfoot is associated with an enhanced initial shoe eversion but not with maximal shoe eversion. The results of the studies related to the placement of foot orthoses are controversial. Although several studies suggested that medially applied orthoses of various designs and placements as well as varus wedged shoe soles decrease maximum shoe eversion (Cavanagh, 1978; Taunton et al., 1985; Clarke et al., 1984; van Woensel and Cavanagh, 1992; Milani et al. 1995), other studies showed no significant differences (Bates et al., 1979; Rogers and LeVeau, 1982; Nigg et al., 1986; Eng and Pierrynowski, 1994). In addition, it has not been established whether foot orthoses may have effects on other segments of the lower extremities, i.e. on the rotation of the tibia and femur (Kozak, 1991; Eng and Pierrynowski, 1994). Thus, the effects of orthoses on the segmental movements of the lower extremities are not comprehensively understood.

All of these shoe and orthotic related findings are based on skin or shoe mounted marker settings. Recent studies have shown, however, that errors are introduced when skin mounted markers move relative to the underlying bone (Cappozzo et al., 1996; Reinschmidt et al., 1997a). Therefore, it is likely that current knowledge of the effects of shoe alterations and orthoses during running is based on inaccurate data. The relevant movement to be studied to answer shoe and orthotic related questions is the bone movement of the major segments of the lower extremity during running.

Methods to determine the skeletal kinematics at the lower extremities include external fixator devices, percutaneus skeletal trackers, x-ray techniques, video fluoroscopy, and inserted bone pins. External fixator devices are typically attached to a segment of a patient, having improper wound healing (Angeloni et al., 1993; Cappozzo et al, 1996); this approach can therefore not provide kinematic results of several leg segments. Percutaneous skeletal trackers consist of an external fixator ring applied to the malleoli to determine the three-dimensional skeletal movements during walking (Stanhope, 1994; Holden et al., 1994); since the ring is clamping both malleoli and possibly influencing skeletal movements between the tibia and fibula, this technique needs further improvement. The x-ray stereo technique allows to reconstruct three-dimensional positions of implanted markers or bony landmarks and has mostly been used *in-vivo* (Simkin, 1982; van Langelaan, 1983) or requires the surgical implantation of markers (Lundberg, 1989). Video fluoroscopy seems a promising method which allows one to directly measure skeletal movements of identified bony landmarks (Tashman et al., 1995); however, for an exact location of the bony landmarks at each time frame this technique requires also implanted markers. Bone pins, surgically inserted into bony structures of the lower extremity *in-vivo*, have been used by a few investigators to study the transverse rotations of the tibia and the femur during walking (Levens et al., 1948), the kinematics of the patellofemoral joint during running (McClay et al., 1990), the tracking of the patella during sitting and squatting (Koh et al., 1992), the methodology of bone pin application (Karlsson and Lundberg, 1994), the kinematics of the knee during walking (Lafortune et al., 1994), and the skin movement artefact in walking and running (Reinschmidt, 1996). At the present time, the bone pin technique seems best suited for the investigation of skeletal kinematics of the lower extremities during running. However, none of the previous bone pin studies has investigated the effects of shoe sole modifications or foot orthoses on the kinematics of the major bones of the lower extremity during running.

Along with the increasing number of runners and joggers, running injuries and measures to prevent those injuries also increased together with the growing interest of the medical community. It has been estimated that 33-56% of runners are affected by at least one running injury per year (Subotnick, 1977; James et al., 1978; Clement et al., 1981). Running injuries have been associated with excessive foot eversion (pronation), which causes health problems at the knee (Subotnick, 1977; James et al., 1978; Noakes, 1991), at the tibia (Segesser and Nigg, 1980; Viitasalo and Kvist, 1983), at the Achilles tendon (Smart et al., 1980; Clement et al., 1981), and at the iliotibial band (Messier and Pittala, 1988). As a consequence, numerous shoe sole constructions and orthotic designs have been developed. It is not known, however, whether these shoe sole and orthotic designs do have the desired effects on the kinematics of the lower extremities during running.

Furthermore, it has been suggested that foot eversion affects tibial rotation and that excessive tibial rotation is associated with the development of knee injuries (Clement et al., 1981; James et al., 1990; van Mechelen, 1992). One mechanism which has been proposed to explain the overloading of the knee joint is associated with patellar malalignment caused by excessive internal rotation of the tibia with respect to the femur (Bahlsen, 1988; Stergiou, 1996). Tibial rotation has been shown *in-vivo* and *in-vitro* to be influenced by foot eversion as well as anatomical characteristics of the ankle joints (Hicks, 1953; Inman, 1969; Inman and Mann, 1978; Nigg et al., 1993; Hintermann, 1994). This coupling of movements at the foot, tibia and femur typically occurs in heel-toe running but has not been studied comprehensively during the stance phase of running up until now.

The concept of the coupling of eversion and internal tibial rotation was proposed by Hicks (1953), was later expanded by Inman (1969 and 1976) and has been discussed with respect to knee injuries by Subotnick (1977) and Segesser et al. (1980). Movement coupling during walking was first investigated by Levens et al. (1948) and later by Wright et al. (1964) who concluded that "during stance phase (in walking) the tibia rotates about its long axis ... therefore, this tibial rotation must be resolved at the subtalar or ankle joint axis". Nigg et al. (1993) studied the movement coupling with regard to foot arch height. In each of the investigated subjects arch height explained a substantial (27%)

amount of the variation in the transfer of movement. The authors proposed a transfer coefficient for a quantitative description of the movement transfer.

Movement coupling between calcaneus and tibia was studied *in-vitro* for systematically varied axial loads on the tibia, as well as systematically varied foot positions of plantarflexion and dorsiflexion, changes in ligament integrity, and selected ankle joints fusions with the movement input from either the tibia or from the calcaneus (Hintermann, 1994). The variation in these variables influenced the movement coupling substantially depending on the movement input (i.e. from the tibia to the calcaneus compared to from the calcaneus to the tibia) which led to the conclusion that different coupling mechanisms were used. While studies have concentrated on the ankle joints, most of the running injuries occur at the knee joint and are substantially influenced by the relative motion between the tibia and the femur (Siegler et al., 1988; Lundberg, 1989; Hintermann, 1994). However, movement coupling between tibia and femur has not been studied in depth yet.

Tibial rotation has been quantified *in-vivo* in walking and running. Levens and co-workers (1948) quantified tibial rotation using bone pins in a laboratory coordinate system and measured an average internal tibial rotation, during the first part of ground contact, of about 20 degrees using bone pins. Similar measurements were conducted by Lafortune and co-workers (1994) who showed an average tibial rotation of about 13 degrees with respect to the femur. Internal tibial rotation with respect to the foot during running of 22 degrees (mean maximum values) using skin markers were reported by Nigg and co-workers (1993).

Barefoot running is often looked upon as the baseline for normal running (Clarke et al., 1984). Running with shoes may change foot and leg kinematics compared to running barefoot. Comparing barefoot with shod running showed that in barefoot running the foot exhibited less inversion at touchdown and less maximum eversion velocity and total eversion (Bates et al., 1978; Nigg et al., 1980; Stacoff et al., 1991; Vagenas et al., 1992). To date, tibial rotation of barefoot running has not been documented in the literature. Tibial rotations are likely to be decreased in barefoot running compared to shod running when assuming that the coupling mechanism at the ankle remains unchanged.

Consequently, it can be postulated that barefoot running could lead to fewer running injuries than shod running, provided there are no additional injuries from the lack of protection. However, the comparison of barefoot versus shod running lacks information with respect to skeletal movements during locomotion, possible changes in muscle activity and epidemiological data.

Biomechanical factors, which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. It has been suggested that excessive eversion forces the Achilles tendon to bend laterally, hereby producing an asymmetric stress distribution across the tendon, which could lead to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981; Denoth, 1986). Excessive eversion velocity produces eccentric loading of the muscles of the posterior tibial compartment, which control/reduce eversion after touchdown. Forces acting on muscle-tendon units during eccentric loading are increased compared to concentric loading. Excessive eversion velocity has been associated with overloading and injury of the muscles of the posterior tibial comparment, e.g. medial tibial stress syndrome (Segesser et al., 1980; Viitasalo et al., 1983; Messier et al., 1988; Stacoff et al., 1988; DeWit et al., 1995). Excessive tibial rotation has been associated with the changes in the tracking of the patella, hereby changing the contact pressure and possibly the friction of the articulating surface of the patella, which may be related to the occurrence of the patellafemoral pain syndrome (Stergiou, 1996). These variables indirectly describe the movement at those structures of interest, but do not directly describe the load within these structures. However, they are relatively easy to quantify. Furthermore, it has been argued that running shoes may be a possible cause of these injuries (e.g. Cavanagh, 1980; Clarke et al., 1983), and that shoe sole constructions and orthoses can align the underlying skeleton in such a way that running injuries can be avoided successfully (James et al., 1978; Eggold, 1981; Segesser et al., 1987). However, it has been shown that skin and/or shoe mounted markers do not represent actual bone movements (Cappozzo et al., 1996; Reinschmidt et al., 1997a). It remains presently unknown, whether the above mentioned excessive movements do in fact take place at the bone level, and whether shoe sole constructions and orthoses do have a measurable effect at the skeleton. Thus, biomechanical factors, related to injuries but measured with skin and shoe markers, should be tested on the skeletal level. Knowledge about skeletal movement during running is crucial for the understanding of excessive movements and shoe interventions with respect to the development of pain and injury.

In summary, the effects of modifications of orthoses and shoe soles on three-dimensional movements of the calcaneus, and tibia during running are not well understood. Additionally, kinematic skeletal movement differences of barefoot versus shod running are currently unknown. Furthermore, investigations so far performed have not clearly established *in-vivo* movement coupling between the calcaneus and the tibia and between the tibia and the femur during the ground contact of running. However, this information seems important for a better understanding of the movement coupling at the lower extremity, the etiology of running injuries and/or the construction of the appropriate orthoses and footwear.

Therefore, the purposes of this study were to quantify:

(i) the influence of modifications of foot orthoses on three-dimensional skeletal movements of the calcaneus and tibia, and

(ii) the influence of shoe sole modifications on three-dimensional skeletal movements of the calcaneus and tibia, and

(iii) the influence of shoes compared to barefoot running on three-dimensional skeletal movements of the calcaneus and tibia, and

(iv) the movement coupling between shoe and calcaneus, between calcaneus and tibia, and between tibia and femur during heel-toe running.

With respect to purpose (i) the specific interest is focused on the question of whether medially applied orthoses can decrease eversion of the calcaneus relative to the tibia as well as internal tibial rotation relative to the calcaneus from touchdown to midstance during running. Furthermore, with two different placements of orthoses being tested, the posterior orthosis is thought to have a decreasing effect on eversion compared to the anteriorly placed orthosis. The specific interest in purpose (ii) lies in the question of whether an increased lateral heel flare enhances initial and/or maximum eversion of the calcaneus relative to the tibia during running. Purpose (iii) is aimed towards the question of whether in barefoot running total eversion and total tibial rotation are decreased when compared to shod running. Purpose (iv) focuses on the question of whether a movement coupling between the calcaneus and the tibia as well as between the tibia and the femur takes place from touchdown to midstance and from midstance to take-off during running. The main hypotheses of the study were:

(i) Medially placed orthoses decrease maximum calcaneal eversion and internal tibial rotation compared with no orthoses.

(ii) Large lateral heel flares increase maximum calcaneal eversion velocity and maximum internal tibial rotation velocity compared to with systematically reduced heel flares.

(iii) Skeletal movements of barefoot running, when compared with shod running, show decreased total calcaneal eversion.

(iv) Skeletal movements of barefoot running, when compared with shod running, show unchanged tibiocalcaneal coupling.

The goal of the present study was to provide kinematic results on the effects of shoe sole and orthotic modifications on skeletal movements during the stance phase of running. The present study can be regarded as the first step towards the understanding of load related injuries at the lower extremities. Once available, these results can be used for kinetic analyses which could attempt to simulate and explain the loading of internal structures (muscles, tendons, ligaments) during running. Thus, the present study may provide information which could be useful in further studies to come.

2. REVIEW OF LITERATURE

This literature review is divided into five sections, covering the kinematics of the lower extremities during running, the effects of running shoes and foot orthoses on the kinematics of running, the kinematics of the lower extremities in view of possibly related injuries, and is concluding with a summary and a reiteration of the questions being addressed in this thesis.

2.1 Lower Extremity Kinematics of Running

2.1.1 Introductory Comments

The functional anatomy of the normal ankle and knee joint will be briefly presented in this review and the definitions of various motions will be provided.

The ankle-joint complex. The ankle-joint complex, linking the foot and the shank consists of two joints, the talocrural or ankle joint between the shank (tibia and fibula) and talus, and the talo-calcaneo-navicular or subtalar joint between the talus and the foot (Inman, 1976).

Movements at the ankle-joint complex. Foot movements at the ankle joint complex are typically described by using a "clinical" coordinate system with the three axes coinciding with a longitudinal, a medio-lateral and an inferior-superior axis. Movements about these axes can be quantified using a joint coordinate system (JCS) (Grood and Suntay, 1983; Cole et al., 1993; Nigg et al., 1993) with three defined successive rotations. However, the rotations around the actual joint axes of the ankle and subtalar joint, which would correspond to functional movements at these joints are difficult to determine *in-vivo* and have been attempted only by a few investigators (Areblad et al., 1990; Bogert et al., 1994). The ankle joint complex has been described in a simplified way as a ball-and-socket joint (Soutas-Little et al., 1987; Areblad et al., 1990; Moseley et al., 1996) and the three rotations about this joint can be described with the three variables plantar/dorsiflexion, ab/adduction, and ev/inverison.

The knee joint. The knee joint allows movements in three planes. The dominant movement component is flexion/extension in the sagittal plane about a medio-lateral oriented axis. Rotations about the longitudinal axis are internal and external rotations of the tibia relative to the femur (or vice versa). Rotations about an antero-posterior axis are abduction/adduction of the tibia relative to the femur (or vice versa).

2.1.2 Historical Aspects of the Kinematics of Running

One of the early detailed work on human locomotion was published by W. Weber and E. Weber (1836). The Weber brothers made their observations on different variations of walking and running using a scale, a watch and a telescope. Their goal was to provide the basis for a theory of locomotion. Among their 150 hypotheses towards this theory, they suggested that "the leg has not only to flex and extend at the knee joint, but also to rotate at the shank about itself" (p.161). This statement is likely to be the first of its kind which is concerned with the longitudinal rotation of the tibia and/or the femur during locomotion. In addition, they reported that the axial rotation of the shank increased when changing from an extended to a flexed knee position (p.204), an important observation for understanding human walking and running.

Additionally, the Weber brothers provided a literature review on human locomotion including work of Aristotle, Galen, Gessendi, Borelli, and Haller and suggested "What is missing are investigations which provide a precise description and measurements upon which a theory of these movements can be based" (p.383). In their final remarks, they suggested that segmental movements of the body in walking and running should be quantified using measurement techniques assessing the magnitude, form and connection of these segments (p.421). This interesting statement shows their foresight for future research in the area of locomotion.

About twenty years after the Weber bothers, Meyer (1853) developed a theoretical idea of human walking observing subjects from three different views introducing a terminology which is still in use to describe human locomotion. Meyer (1853) studied many different forms of gait (including running) and concluded that it is impossible to describe one single form of walking as the typical one (p. 548).

In the second half of the 19th century a number of research techniques were developed to describe the kinematics of the lower extremities devices which are still used today. Marey (1895/1972) invented a pressure measurement devices for the shoe sole, a dynamometer which later improved into a force plate and a "chronophotographe" using a photo camera which allowed to study human gait with the relatively high frequency of 100 Hz to document human gait. Marey's work on locomotion concentrated mostly on sagittal plane movements, providing the first "stick figures" of humans during gait. Muybridge (1887/1979), was the first to make pictures of up to three orthogonal directions of the same subject during locomotion using serial photographs which could be animated by means of re-photographing them with modern cine-cameras. Muybridge, however, did not use his serial photographs to undertake any measurements.

The first to conduct a three-dimensional study of locomotion were Braune and Fischer (1904/1987). Fischer's work (Braune died shortly after the experiments) concentrated mainly on movements in the sagittal plane but made reference to rotations about the longitudinal axis of the leg segments when discussing the "tridimensional co-ordinates of the joint centers". Fischer stated that "...the axis of the lower leg and axis of the foot can be used as indicators of the rotation of the thigh and the lower leg, respectively" (Braune and Fischer, 1904/1987, p.59). Fischer, proposed a typical hiking step which soldiers "can perform for hours without tiring"; the corresponding stick figures were regarded by Fischer as very similar to those provided by Marey (p. 319).

In the first half of the 20th century Bernstein (1967) studied human locomotion using various biomechanical tools to derive theories on motor control. He developed the cyclocraphic method to produce stick figures of a running man at 187 Hz. In Bernstein's view locomotion movements ..."are amongst the most highly automated of movements.

The most rigid succession of details are extremely receptive for each particular subject" (p. 60).

Levens et al., (1948) performed the first study using inserted bone pins at the lower extremities allowing for the first time the quantification of actual skeletal movements during locomotion. Further work with bone pins was performed by Close and co-workers (1967) showing tibial rotation in space of 10° to 11° during the stance phase of walking. Wright et al., (1964) used goniometers at the ankle joints to quantify human locomotion. The number of publications on walking and running has steadily increased since the 1970s. Progress resulting from the development of opto-electronic devices has led to the kinematic analysis of walking and running being widely used now for the characterization of normal and pathological gait (Berme and Cappozzo, 1990; Cavanagh, 1990; Perry, 1992).

In summary the methods used to describe the kinematics of the lower extremity during human locomotion have conceptually changed little since the days of Braune and Fischer. Modern technology allows 3-D quantification of segmental movements and the manipulation of large amounts of data using computers. However, previous work on running concentrated primarily on segmental movements with the use of external markers while actual skeletal movements are still not well understood. Today the statement provided by the Weber brothers more than 150 years ago still holds true, that "... investigations (are) missing which provide the measurements and a precise description upon which a theory (of locomotion) can be based" (p. 383).

2.1.3 Determination of Three-Dimensional Movements

For the representation of three-dimensional (3-D) movements at the lower extremity a number of parameterizations are available (Cole et al. 1993): Cardan/Euler angles (Grood and Suntay, 1983), helical axis (Woltring, 1994), finite helical axis descriptors (Lundberg, 1989), and instantaneous helical axis (Murphy, 1993). For the specific purpose of a gait analysis one commonly used parameterization of the rotation matrix is Cardan angles with a predefined sequence of rotations about a "Joint Coordinate System" (JCS) (Grood and Suntay, 1983). On each segment of interest, three (or more) markers are attached and their position is calculated using primarily optical methods. These markers are used to measure the movement of the anatomical coordinate system of each segment. Different methods have been used to define these anatomical coordinate systems including the use of standing trials in a standardized neutral position (Areblad et al., 1990; Nigg et al., 1993; Moseley et al., 1996), roentgen-stereophotogrammetry (Lafortune, 1992b; McClay, 1990), and the relationship between external markers and internal markers (bone embedded) on anatomical landmarks (Cappozzo et al. 1995).

Grood and Suntay (1983) proposed a non-orthogonal JCS for the knee joint in which one axis is fixed in the femur segment, a second axis is fixed in the tibial segment, an the intermediate or "floating" axis is normal to the two body fixed axes. Grood and Suntay (1983) proposed that flexion/extension occurs about the femur fixed axis in medio-lateral direction, interal/external knee rotation about the tibia fixed longitudinal direction, and abduction/adduction about the floating axis. Cole et al. (1993) proposed plantar/ dorsiflexion for the ankle joint complex (AJC) to occur about the tibia fixed axis in medio-lateral direction, in/everison about the calcaneus fixed longitudinal axis and ab/adduction about the floating axis. This sequence was different than previously proposed where abduction/adduction and inversion/eversion was reversed (Areblad et al., 1990; Soutas-Little et al., 1987; Engsberg, 1987). The different choice of sequence by Cole et al., (1993) was based on the argument that rotations about the distal segments should be around the longitudinal axis. Nigg et al. (1993) proposed yet another sequence with tibial rotation about a tibia fixed longitudinal axis, then in/eversion about the floating axis, and plantar/dorsiflexion about a calcaneus fixed medio-lateral axis having the advantage that the calculation of the tibial rotation was first in the sequence. Furthermore, Cole et al. (1993) showed that for running the sequence of rotation produced differences which were less than one degree.

In this thesis the rotations and translations are expressed by a 4x4 matrix, containing a 3x3 direction cosine matrix and a 3x1 translation vector (see chapter 3 and

Reinschmidt, 1996). Cardan angles (and translations) can be extracted from this matrix by decomposition in the defined sequences of rotations.

2.1.4 Skin and Shoe Movement Artefacts

In kinematic studies of human locomotion, generally external (i.e. skin or shoe mounted) markers are used to describe the segmental movements of the underlying bone. In this procedure, two main sources of inaccuracy can be introduced which result from (Cappozzo et al., 1996):

- Methodology (or instrument) errors, representing the errors with which marker coordinates are reconstructed in a film (or global) reference frame.
- Skin and shoe movement artefacts, representing the errors which are introduced by the relative movement between the external markers and the underlying bone.

The methodology errors related to this thesis are presented in chapter 3.3, skin and shoe movement artefacts are discussed below.

2.1.4.1 Skin Movement Artefacts

A number of measuring techniques have been developed and used to quantify the skin movement artefact during locomotion. These include external fixator devices, percutaneus skeletal trackers, x-ray techniques, video fluoroscopy, and inserted bone pins. Currently the most useful results for the determination of skin movement artefacts originate from studies with bone pins and external fixators. Bone pins, surgically inserted into bony structures of the lower extremity *in-vivo*, have been used by a few investigators to study the transverse rotations of the tibia and femur during walking (Levens et al., 1948; Close et al., 1967), the kinematics of the patellofemoral joint during running (McClay et al., 1990), the tracking of the patella during sitting and squatting (Koh et al., 1992), the methodology of bone pin application (Karlsson and Lundberg, 1994), the kinematics of the knee during walking (Lafortune et al., 1994), and the skin movement artefact in walking and running (Reinschmidt, 1996). Generally, these results can be summarized as follows:

- Skin movement artefacts are generally larger than the errors introduced by methodology errors (Cappozzo et al., 1996).
- Tibiocalcaneal rotations are generally well represented by skin markers, but absolute values have to be interpreted with caution since tibiocalcaneal rotations are expected to be overestimated (Reinschmidt, 1997a).
- Knee rotations other than flexion/extension may be affected with substantial errors when using skin markers (Reinschmidt, 1997a).

2.1.4.2 Shoe Movement Artefacts

To date, two methods, shoe windows and bone pins, have been applied to study the relative movements between shoe and underlying heel and/or calcaneus. Windows cut into the heel counter of a shoe have been used to estimate the relative movement between the skin covering the posterior part of the calcaneus and the heel of the shoe in running (Nigg et al., 1986; Stacoff et al., 1992). In these two-dimensional studies it was found that eversion (projected into the frontal plane) was similar, but not identical, for the shoe and skin markers; the shoes overestimating the heel movement by 2°-3°. However, this procedure does not provide information how much the skin markers move with respect to the underlying bone.

Reinschmidt (1997a) provided the first data in which eversion was compared between shoe markers and calcaneal bone pin markers in running. The results from five subjects showed that maximum shoe eversion ***overestimated*** calcaneal eversion by 4.3° to 13.1° depending on the subjects and the shoes (Figure 2-1). There appeared to be a relationship between maximum shoe eversion and bone eversion which was likely to be influenced by the shoe and the location of the markers attached to the shoe. The difference between external and internal markers was small at touchdown and take-off and largest around the time of maximum eversion which was associated with the possible decrease of accuracy towards the edges of the defined three-dimensional space. It was concluded, that generally shoe markers overestimated eversion of the calcaneus and that the fitting of the heel to the heel counter may have been different between the subjects. What could not be detected, however, was the relative movement between bone pin markers and skin markers inside the shoe.

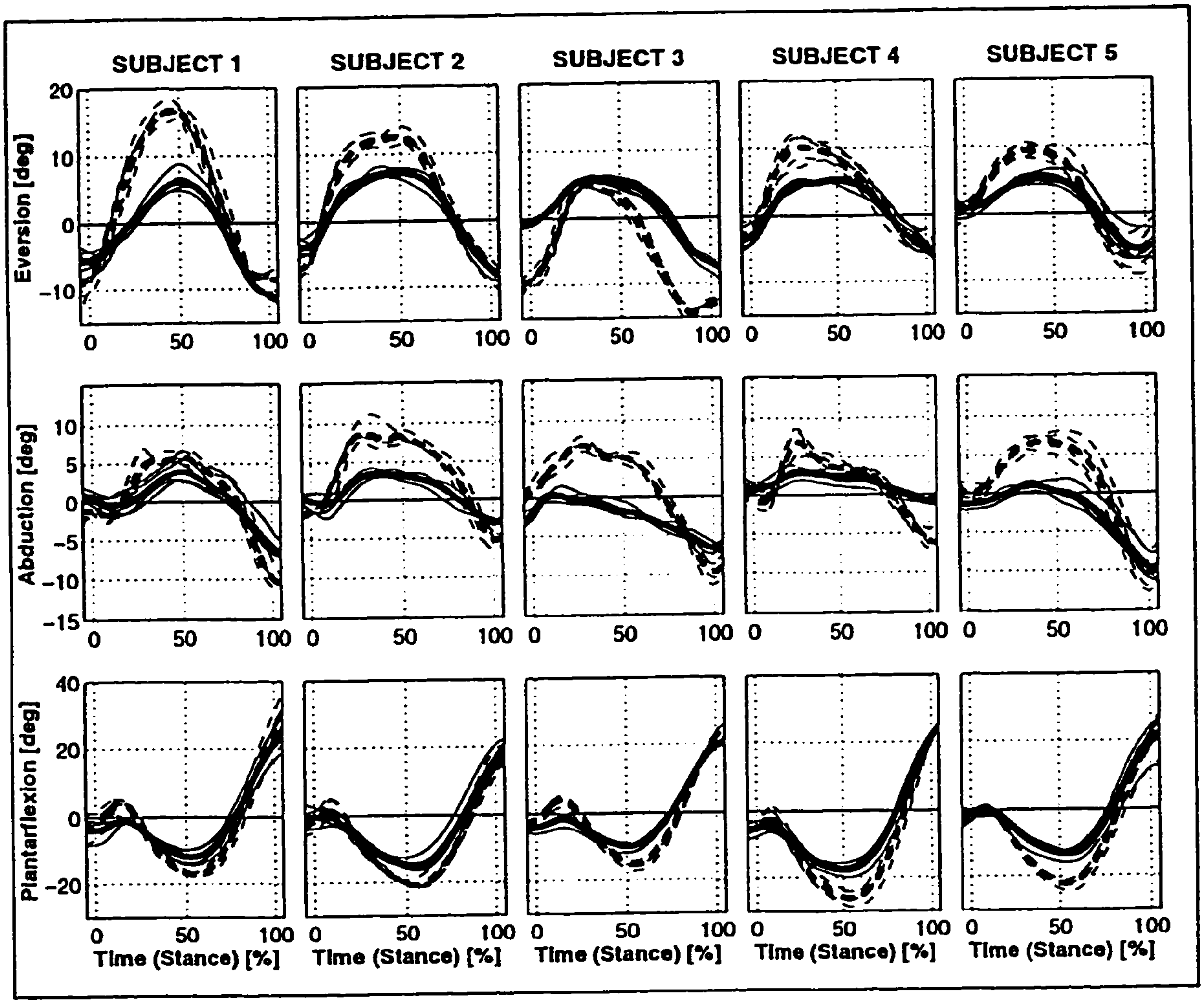


Figure 2-1: Skin and shoe movement artefacts (after Reinschmidt, 1996). Solid lines (—) represent bone pin based kinematics, dashed lines (---) represent skin/shoe marker based kinematics. Thick lines represent averages of five trials.

2.1.5 Movements at the Ankle Joint Complex During Running

This review focuses on ankle and rearfoot movements of 2-D and 3-D analyses of running barefoot and with shoes. The majority of kinematic investigations of the ankle during running are two-dimensional and/or biplanar (Table 2-1). Effects of shoe sole and orthotic modifications are discussed in chapters 2.2 and 2.3.

2.1.5.1 Inversion and Eversion

In heel-toe running the period of foot-ground contact starts typically with the touchdown of the lateral aspect of calcaneus. In barefoot running the foot is only slightly inverted, in running with shoes this initial inversion is on average about 10° (Table 2-1). Immediately after touchdown the foot and shoe evert with the barefoot eversion velocity being smaller than the shoe eversion velocity (Stacoff et al., 1991). Several possible reasons have been proposed to explain this difference. The most probable one is that shoe eversion is overestimated (Reinschmidt, 1997a) and, consequently the results are primarily influenced by methodological problems.

Maximum barefoot eversion is generally reported smaller (8°-12°) than maximum shoe eversion (7°-19°; Bates et al., 1978; Nigg et al., 1986). Maximum shoe eversion is depending on the subjects, on the leg (dominant versus non-dominant, Vagenas et al., 1992), the running speed (Nigg et al., 1986) and the running shoes (Table 2-1).
Differences between 2-D and 3-D analyses of eversion were first described by Soutas-Little et al., (1987). The authors concluded that the 2-D approach overestimated the 3-D results. Areblad et al., (1990) showed that most two-dimensional angular values measured from a posterior view are sensitive to the alignment angle (angle between the longitudinal axis of the foot and the camera axis). The Achilles tendon angle, (relative angle between the rearfoot and the tibia, $\Delta\beta_{max}$), was the variable that was least sensitive to the alignment angle. The result of the 2-D rearfoot angle corresponded well to the 3-D results during stance phase only when the camera axis was aligned with the longitudinal axis of

Authors, Year	Type of study	n	γ_o [°]	γ_{max} [°]	$\Delta t_{\gamma max}$ [ms]	$\Delta\gamma_{max}$ [°]	$\dot{\gamma}_{max}$ [°/s]	$\Delta t_{\dot{\gamma} max}$ [ms]	β_o [°]	β_{max} [°]	$\Delta t_{\beta max}$ [ms]	$\Delta\beta_{max}$ [°]	$\dot{\beta}_{max}$ [°/s]	$\Delta\beta_{10}$ [°]
Barefoot														
Bates et al., 1978	2D	10							-1.9	8.6	95	10.5		
Nigg et al., 1980	2D	54	-0.8			17.5			4.3			21.3		3.9
Stacoff et al., 1991	2D, left right	9 9										8.9 11.5	527 641	
Vagenas et al., 1992	2D, dom. Non-dom.	29 29	-5.1 -4.2	5.5 6.0	82 77				3.6 3.0	11.3 10.3		7.7 7.4		
Shoes														
Bates et al., 1978	2D	10							-10.4	7.2	82	17.6		
Nigg et al., 1980	2D	45	-7.5			27.0			-2.7			34.6		6.3
Clarke et al., 1980	2D	15	-3.7	10.8	45	14.5								
Clarke et al., 1983	2D	10	-4.9	-11.7	93	16.7	532	26.6					532	
Nigg et al., 1986	2D, A45, 3m/s A45, 6m/s A25, 3m/s A25, 6m/s	16 16 16 16	-9.8 -11.6 -9.8 -11.9	2.1 2.5 1.5 1.2	137 65 125 69	12.0 14.1 11.3 13.2			-1.1 -0.1 -1.1 -1.0	14.0 19.0 13.6 17.4	134 71 136 83	15.0 19.1 13.8 18.4		7.4 8.6 2.9 5.8
Soutas-Little, 1987	3D 2D 3D 2D	A A B B	-4 -8 -6 -4	9.0 7.5 9.5 8.5	85 105 130 130	15.0 15.5 17.5 14.5	330 300 310 250	50 30 50 75						
Stacoff et al., 1991	2D, left right	9 9										25.4 22.9	1087 963	

(continued)

Authors, Year	Type of study	n	γ_o [°]	γ_{max} [°]	$\Delta t_{\gamma max}$ [ms]	$\Delta\gamma_{max}$ [°]	$\dot{\gamma}_{max}$ [°/s]	$\Delta t_{\dot{\gamma} max}$ [ms]	β_o [°]	β_{max} [°]	$\Delta t_{\beta max}$ [ms]	$\Delta\beta_{max}$ [°]	$\dot{\beta}_{max}$ [°/s]	$\Delta\beta_{10}$ [°]
Stacoff et al., 1992	2D,sm	15										14.1		
	win	15										12.1		
	bf	15										13.7		
Vagenas et al., 1992	2D, dom.	29	-9.4	8.7	92				-1.2	10.8		12.1		
	Non- dom.	29	-7.5	8.6	88				-1.0	10.0		11.0		
Nigg et al., 1993	3D	30										28.0		
DeWit et al., 1995	2D, C65, left	7	-10.7	3.1			526		-2.4			16.9		8.0
	C65, right	7	-11.8	1.4			503		-2.7			16.4		8.2
	C40, left	7	-8.3	7.5			403		0.1			18.9		5.5
	C40, right	7	-10.3	5.0			347		0.9			19.0		4.7
Reinschmidt, 1996	3D, sm	5										16.0		
	bp											8.6		

Table 2-1: Findings of investigations of ankle eversion during running. Key:

bp/sm	results based on bone pin markers / shoe markers	win	results based on heel measurements with shoe
2D	two-dimensional study in the frontal plane	3D	windows
bf	barefoot	dom.	three-dimensional study
A45,C65	Shore A 45, Shore C 65	β_o	dominant leg
γ_o	calcaneal inversion (-) at touchdown	β_{max}	inversion (-) calcaneus relative to tibia at touchdown
γ_{max}	max. eversion of calcaneus during ground contact	$\Delta t_{\beta max}$	max. Achilles tendon angle
$\Delta t_{\gamma max}$	time diff.: touchdown to max. calcaneal eversion	$\Delta\beta_{max}$	time diff.: touchdown to max. Achilles tendon angle
$\Delta\gamma_{max}$	$= \gamma_{max} - \gamma_o$	$\dot{\beta}_{max}$	$= \beta_{max} - \beta_o$
$\dot{\gamma}_{max}$	max. velocity during eversion of the calcaneus	$\Delta\beta_{10}$	max. velocity of the Achilles tendon angle
$\Delta t_{\dot{\gamma} max}$	time diff.: touchdown to max. eversion calcaneus vel.		initial eversion within the first 10th of ground contact

the foot. Since 2-D studies generally do not specify the alignment angle, all 2-D results reported in Table 2-1 should, therefore, be used carefully.

Three-dimensional studies, on the other hand, have the problem that different sequences in determining the Cardan angles will typically produce different results, which when carefully chosen, can be reduced to less than one degree (Cole et al., 1993). There are only a limited number of three-dimensional studies on inversion/eversion being published. Nigg et al., (1993) reported an average maximum eversion of 28° which was considered rather high compared to Reinschmidt (1996) and older 2-D studies (see Table 2-1). It was suggested (Nigg et al., 1993) that this was because the marker positions were on the entire foot measuring "total foot eversion" rather then "rearfoot eversion".

Engsberg (1987) provided a unique contribution towards the understanding of 3-D movements of the rearfoot during (barefoot) running. The movements of the AJC were expressed in direction cosine vector projection angles and the relative contribution in each of the three major planes around the axis of rotation (equivalent screw displacement axis, ESD) were expressed in ESD parameter values with a maximum of 1. The comparison of the relative contributions in the three planes showed that among the five study subjects there was no common way of moving the rearfoot from touchdown to take-off. Plantarflexion was dominant from touchdown to footflat in three subjects, eversion in one subject and abduction also in one subject. Four subjects showed most movement in eversion in the period of footflat to maximum eversion, but in one subject dorsiflexion and abduction dominated. Dorsiflexion during take-off was found to be largest in four subjects, but inversion was found to be greater in one subject. It was concluded that during running different subjects may exhibit different sets of dominant rotations at different times. This finding is supported by *in-vitro* studies (van Langelaan, 1983; Lundberg, 1989) who found different movement solutions depending on the location of the point of force application (see 2.1.7). In other words, the movements at the AJC can differ greatly between individuals and there seems to be no distinct movement order of rotations in the human ankle.

Edington et al. (1990) attempted to identify three types of runners based on a 2-D analysis, according to initial inversion/eversion patterns: "normals" with a initial eversion, "supinators" showing initial inversion and "excessive pronators" with a large initial eversion.

2.1.5.2 Plantar/Dorsiflexion

Movements in plantar/dorsiflexion at the AJC are substantially influenced by the landing strategy, i.e. rearfoot, midfoot or forefoot touchdown (Cavanagh, 1990). This review focuses on rearfoot touchdown. At the beginning of the rearfoot touchdown, an initial plantarflexion of a few degrees takes place (Milliron and Cavanagh, 1990). After around 20% of the stance time dorsiflexion occurs in the order of magnitude of 20°-25° reaching its maximum at around 50% of stance time. During take-off the foot goes again into plantarflexion (Mann et al., 1980; Soutas-Little, 1987; Milliron and Cavanagh, 1990). Comparing plantar/dorsiflexion measured with bone pin and skin markers, Reinschmidt (1997a) concluded that the skin markers overestimated the movement by an average of 4,7° (14,1%) over the entire stance phase, but that the general shape of the curve was similar between the two marker settings. It can be argued that this result is based on tibial and calcaneal bone and skin markers which were not placed at the exact same location at the tibia and foot and that some small differences occurred at touchdown and at midstance (Figure 2-1).

2.1.5.3 Ab/Aduction (internal/external tibial rotation)

During ground contact, when the foot is flat on the ground it can be regarded as approximately "fixed". Hence, ab/adduction movements, calculated as ab/adduction of the foot relative to the shank, and internal/external tibial rotation, calculated as rotation about the longitudinal axis of the shank, may be regarded as describing the same movement. Internal rotation of the shank relative to the heel between touchdown and midstance have been reported to be 15° in barefoot running using skin markers (Soutas-Little, 1987) and 22° using skin and shoe markers (Nigg et al., 1993). During the second phase of stance, an external tibial rotation was reported to be 10° (Soutas -Little, 1987) and 19° (Nigg et al., 1993). Reinschmidt et al., (1997a) found an average range of motion of 7.5° internal tibial rotation measured on the bone level and 11.1° measured with external markers, although skin and shoe markers were not fixed at the exact same location at the tibia and foot.

2.1.6 Tibiofemoral Joint Motion During Running

2.1.6.1 Flexion/Extension

During the stance phase of running the knee flexes between touchdown and about 40% ground contact. Then, flexion is followed by extension until shortly before take-off (McClay, 1990). Reinschmidt et al. (1997c) found flexion/ extension curves derived from skeletal and external marker settings similar for all three subjects. Flexion at the knee from touchdown to midstance has been reported to be about 30°-45° (Milliron and Cavanagh, 1990) at an average flexion velocity of 450°/s. Knee extension from midstance to take-off is reported to be about 20°-30° at an extension velocity of around 300°/s (Milliron and Cavanagh, 1990).

2.1.6.2 Leg Ab/adduction

Abduction of the leg during running has been reported by a number of 2D and 3D studies. In 2D studies leg abduction between touchdown and midstance was reported to be between 5° and 10° (tibia relative to the vertical) with various running speeds and different shoe sole hardnesses (Clarke et al., 1984; Nigg et al., 1986; Stacoff et al., 1989). Leg abduction was found to show left-right differences, the dominant leg being significantly more abducted than the non-dominant leg (Vagenas et al., 1992). Results based on 3D measurements showed considerable differences between subjects. The total range of motion (tibia relative to the femur) averaged 8° (McClay, 1990) and between 5°

and 10° and the agreement between bone and skin marker based results was found to be poor (Reinschmidt et al., 1997c).

2.1.6.3 Internal/external tibiofemoral rotation

McClay et al., (1990) reported 10° of internal tibiofemoral rotation in barefoot running using bone pins from touchdown to midstance followed by an external rotation of about the same magnitude to take-off. Reinschmidt (1997a) found the agreement between skin and bone markers highly subject dependent. His three subjects showed an average internal tibial rotation relative to the calcaneus of around 5°-10° (bone) and 8°-15° (skin) when running in shoes.

These findings on tibial rotations during running are supported by studies on walking. Internal tibial rotation relative to the floor has been reported to be an average of 7° (Levens et al., 1948) using shoes and bone pins and 6° (Lafortune et al., 1992) under the same shoe and bone pin conditions. Moseley, (1996) reported 7° when walking barefoot using skin markers, and Reinschmidt (1997b) found 8° of internal tibial rotation relative to the ground under the same conditions.

In summary, knowledge about tibiocalcaneal and tibiofemoral rotations during running is rather limited. Most results are based on skin and/or shoe mounted marker settings, which are known to overestimate movements. However, if orthotic and shoe sole modification effects are to be discussed, the knowledge about the effects of external versus bone mounted marker settings is important.

2.1.7 Movement Coupling

Movement coupling during locomotion can occur between all segments of the lower extremity, which has been described by Hicks (1953) as follows: "The lateral rotation of the tibia was seen to occur when the posterior part of the foot supinated as a result of pronation twist of the forefoot, this in turn being brought about by raising the medial arch. By this chain of actions it comes about that the femur laterally rotates at the hip joint when the big toe is lifted off the ground".

The concept of coupling between foot eversion and internal tibial rotation dates back to x-ray observations of Lovett and Cotton (1898) who concluded that pronation consists of a number of changes of positions, one of them being "... the obvious change of the malleoli, which (...) move with it in this rotation, the inner malleolus backward and the outer forward". In 1917, Strasser commented that the rotation of the tibia about its vertical axis is only possible at the subtalar joint and as a consequence of the rotation of the foot about its longitudinal axis. Manter's results (1941) show that the subtalar joint causes the talus to move as a right-handed screw in the right foot.

It has been gradually understood that internal and external tibial rotation takes place as a result of the oblique orientation of the subtalar joint. The approximate position of the subtalar joint axis was already described by Meyer (1853), and was confirmed by Henke (1859), R. Fick (1911) and Elftman and Manter (1935) who all agreed, generally, that this axis passes through the lateral side of the calcaneus and extends upward, forward and medially to the head of the talus.

Jones (1945) was the first to use a wooden model (Figure 2-2, (a)) to explain that "... on account of the obliquity of the axis of the subtalar joint a lateral torque in the tibia produces a component of force which acts to invert the foot". Hicks (1953) provided yet another wooden model (Figure 2-2, (b)) demonstrating that "when the foot tends to supinate, the leg rotates laterally, and when the foot tends to pronate, the leg rotates medially (...). This apparently double movement is nothing other than simple hinge movement at the talo-calcaneo-navicular joint". Inman's model (1969) demonstrates how "rotation of the leg is transmitted to the foot" (Figure 2-2, (c)). In his later work Inman (1976) referred to the action of talocalcaneal joint to that of a "mitered joint", a simple hinge joint connecting the leg to the foot. Inman (1976) stated that the mitered joint model shows that "within a limited range, longitudinal rotation of the leg imposes a longitudinal rotation of the foot". Inman (1976) showed the effects of different orientations of the subtalar and ankle joint axes on leg and foot movements using an entire series of different wooden models (Figure 2-2 (d)). However, the concept of the "mitered joint" has recently been questioned by Engsberg (1987) and by *in-vitro* studies (see next paragraph) concluding that external tibial rotation and supination and internal tibial rotation and pronation are not identically coupled as it would be in the case of a

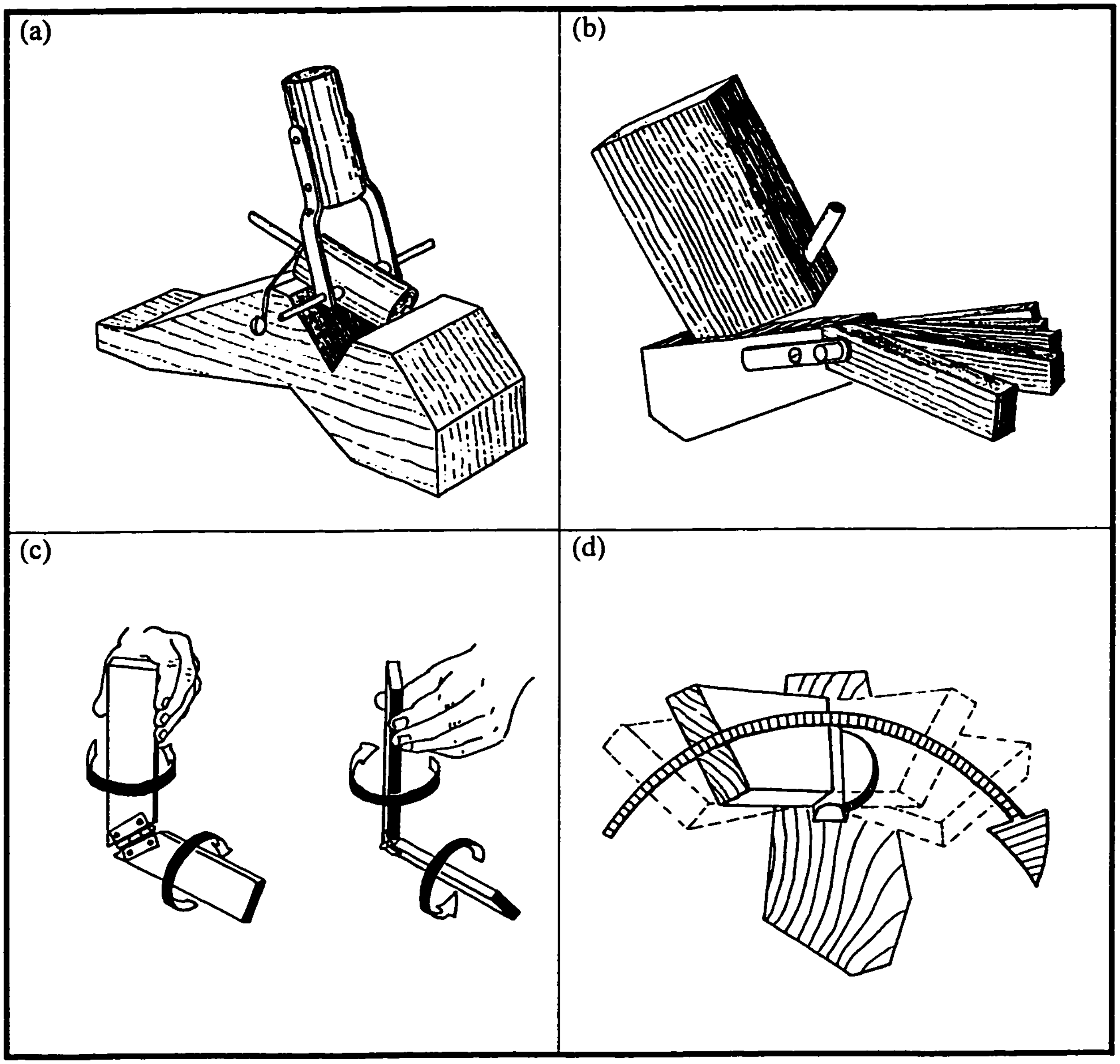


Figure 2-2: Various models of the human foot explaining movement coupling of the rearfoot relative to the tibia: redrawn after (a) Jones (1945), (b) Hicks (1953), (c) and (d) Inman (1969 and 1976).

mitered joint. Furthermore, due to the constrains of the hinge, a mitered joint does not allow only rotation about the longitudinal axes of the leg and foot, as proposed by Inman (1969), since the hinge causes the foot and the leg to move also laterally hereby changing the leg ab/adduction. Mechanically, for rotations about the longitudinal axes only, a bevel gear would be a more appropriate joint model to show that each degree of rotation of the foot is transferred to the tibia. Since during human locomotion inversion/eversion, abduction/adduction and plantar/dorsiflexion take place simultaneously, none of the models in Figure 2-2 correctly demonstrate the actual movement coupling between the rearfoot and the tibia. For a more realistic representation of the movements at the ankle during gait further models need to be developed.

Investigations using a test apparatus for the movement coupling between the calcaneus and the tibia. A number of *in-vivo* and *in-vitro* studies are related to the relative movements of the tarsal bones and the orientation of the ankle axes (Hicks, 1953; Inman, 1976; Langelaan, 1983; Lundberg, 1989; Siegler, 1988; Murphy, 1993), but only a few investigators (Olerud and Rosendahl, 1985; Lundberg, 1989; Hintermann, 1994) have looked closely at the coupling mechanisms between the rearfoot and the tibia. Recent studies using test apparatus have shown that the ankle and subtalar joint are not "simple hinge joints", because the movements were found to take place about axes which move continuously (Langelaan, 1983; Siegler, 1988; Lundberg, 1989, Murphy, 1993).

The torsion-transmitting capacity was studied *in-vitro* by Olerud and Rosendahl (1985). A transfer coefficient (or coupling coefficient: change of tibia rotation over change of inversion) of 0.42 was found for the unloaded foot in plantigrade position when moving the calcaneus. Hintermann (1994) found 0.74 for the same condition. He argued, that the differences were because of different specimen fixation techniques. Lundberg (1989) found *in-vivo* a coupling coefficient of 0.2 using again another fixation device. There seems to be a general agreement that movement coupling between the calcaneus and the tibia does exist and that it is more pronounced from inversion to external leg rotation than from eversion to internal leg rotation (Lundberg, 1989; Hintermann, 1994). Furthermore, the coupling mechanism from the calcaneus to the tibia with a coupling coefficient of 0.3-0.4 for eversion/internal tibial rotation and 0.7-0.8 for inversion/external tibial rotation (depending on the vertical load at 0° plantarflexion) is larger than from the tibia to the calcaneus, where coupling coefficients of 0.05-0.1 for eversion/internal tibial rotation and 0.4-0.5 for inversion/external tibial rotation (Hintermann, 1994) have been reported. This excludes the comparison of the ankle joint complex to a hinge or universal joint as previously stated by Inman (1976) and Olerud et al. (1985). In addition, movement coupling is substantially dependent on the vertical loads on the tibia, positions of plantarflexion and dorsiflexion, changes in ligament integrity, selected ankle joint complex fusions and muscle-tendon forces (Hintermann, 1994).

Investigations of movement coupling during locomotion. Presently, there are only very few studies on movement coupling during locomotion. Perry and Lafortune (1992) concluded that the results of their walking study supported the idea of coupling between the foot and leg. Nigg et al. (1993) proposed a transfer coefficient for a quantitative description of the coupling between eversion and internal tibial rotation and found a high correlation ($p<0.0001$) between internal shank rotation and eversion. Also, movement coupling was studied with regards to the foot arch. Arch height explained a substantial (27%), yet incomplete, proportion of the variation in the transfer of movement between subjects. The mean coupling coefficient in the study by Nigg et al. (1993) was 0.76 ± 0.16 which was later supported by the work of McClay et al. (1997) who found a mean coefficient of a pronator group of 0.81 ± 0.29 and one for a normal group of 0.65 ± 0.26. Both of these investigations used skin and shoe mounted markers which may have overestimated the underlying bone movements.

Investigations on movement coupling during locomotion between tibia and femur. Movement coupling at the knee has not specifically attracted locomotion research. However, movements of the tibia relative to the femur about the respective longitudinal axes have been reported by a few investigators using bone pins. Levens et al. (1948), found an average rotation of the tibia relative to the femur of 8.7° during walking. Lafortune et al. (1994) reported an average of 4.5° of relative rotation during walking.

Also in walking, Reinschmidt (1996) found relative internal rotation of less than 2 degrees in one subject, 5 degrees in another, and relative external rotation of approximately 4 degrees in a third subject. During running, relative internal rotation was found to be between 5° and 10° in the same study. These results suggest that, similar to the ankle, movement coupling at the tibia and femur may be very subject dependent.

Lafortune et al (1992) and Perry and Lafortune (1995) noted that rotational changes during walking, induced by 10° varus and valgus wedged shoes were larger at the tibia than at the knee, suggesting that muscular and ligamentous structures of the knee play an important role in transferring tibial rotation to the femur. Further results on movement coupling between the tibia and femur during locomotion are not available at the present time.

Summary. *In-vitro* and *in-vivo* studies using test apparatus suggest that movement coupling between the calcaneus and the tibia depends on various factors, including vertical loading, foot positions, ligament integrity and muscular activity. Movement coupling at the ankle during running, is significant and seems to be dependent (among other factors) on arch height. The coupling mechanism at the knee during running is not well established. The effects of shoe sole modifications and orthoses on movement coupling at the ankle and knee are currently not well understood (see 2.2 and 2.3).

2.2 Influence of Shoe Soles on the Kinematics of Running

2.2.1 Definitions

Lateral heel flare: The prominence of the shoe sole deviating laterally from the vertical by a given degree (see Figure 3-4).

Midsole hardness: The hardness of the sole material located between the upper and that part of the sole which is in contact to the ground (which is defined the outer sole). The hardness is measured statically with a device called Shore durometer.

Touchdown angle: The angle between the shoe sole and the ground at the instant of touchdown viewed in the frontal plane.

2.2.2 Kinematic Effects of Shoe Soles

The effect of various shoe sole modifications on rearfoot movement during running have been shown in a number of studies. Two important modifications are changes in shoe sole geometry, specifically in flare on the lateral side of the sole, and changes in midsole hardness.

Early running shoes typically showed a prominent heel flare at the lateral side which was later discussed in sports medical and biomechanical circles to be related to running injuries (Cavanagh, 1980). A number of investigations have been carried out in the search of an optimal shoe sole geometry where the lateral heel flare and/or modifications of sole hardness were tested systematically.

The lateral aspect of the shoe sole touches the ground first at a touchdown angle of 5°-10° when running with rearfoot touchdown (Table 2-1). The point of application of the ground reaction force is influenced by this touchdown angle as well as by the prominence and the deformation of the shoe sole. A prominent and hard heel flare increases the lever about the subtalar joint axis, hereby increasing the moment arm of the acting forces. It has been shown in a computer simulation (Stacoff et al., 1988) that the theoretically maximum eversion and eversion velocity of the rearfoot increases compared to a shoe sole with a non-prominent and soft shoe sole construction.

Clarke and co-workers (1983) studied the influence of lateral heel flare (30°, 15°, 0°) on maximum eversion combined with three different shoe sole hardnesses on 10 subjects running on a treadmill. It was found that for hard midsoles (Shore A45) maximum eversion did not change between the different heel flares. For medium (Shore A35) and soft (Shore A25) midsole hardnesses, maximum eversion increased with decreasing flare (30° flare=11.1° eversion; 15° flare=11.4°; 0° flare=12.6°), which is the opposite of the theoretically achieved result provided by Stacoff et al. (1988). The results of the hard midsole were supported later by Nigg and Morlock (1987a), but the results of the medium and soft midsole were not. Nigg and Morlock (1987a) suggested that the discrepancy of the results of the two studies may be because of the different shoe types, another definition of maximum eversion and/or overground versus treadmill running. Furthermore, it was suggested, that maximum eversion is reduced with an enhanced medial midsole hardness hereby contributing to an improved medial stability of the shoe. In their study (Nigg and Morlock, 1987a) heel flares of 0°, 16° and a negative flare (rounded off) were tested on 14 subjects. It was found that with decreasing heel flare initial eversion ($\Delta\beta_{10}$: 16° flare=8.1°; 0° flare=6.7°; rounded=4.9°) and initial eversion velocity ($\dot{\beta}_{10}$: 16° flare=301°/s ; 0° flare=255°/s; rounded=191°/s) were decreased. It was argued that this was related to a decrease in lever arm about the subtalar joint.

Similar results were observed in a further study (Nigg and Bahlsen, 1987b) between the flared (16°) and the neutral (0°) heel construction with hard midsole materials, but not with soft materials. The lever arm can be reduced easily since soft materials are compressible at first ground contact. It was concluded, that in order to decrease initial eversion the sole should have either a neutral (0°) or rounded heel, or soft material at the lateral side of the heel.

Hamill et al. (1992) demonstrated that shoes with softer sole material produce significantly larger maximum eversion but no changes in tibial ab/adduction and knee flexion/extension. Further, with soft soles the rearfoot began to supinate for take-off while the knee was still flexing. It was suggested that such "asynchronous timing could be the root cause of some knee injuries to runners".

In a recent study (DeWit et al., 1995), an enhanced and faster initial eversion was found for hard midsoles compared to soft midsoles, but maximum eversion was larger for the soft midsole compared to the hard one (Table 2-1). It was concluded, that at touchdown the point of application of the hard midsole is located more laterally thereby increasing the lever arm of the everting moment about the subtalar joint axis, but during midstance a hard sole would offer more stability to foot movements in the frontal plane, thus, producing arguments which are quite similar to the ones stated before.

All of these studies are based on shoe or skin mounted marker settings. It has been shown recently, that errors are introduced when the skin or the shoe moves over the underlying bone (see 2.1.4). Furthermore, the comparison of 2-D versus 3-D studies shows that the 2-D data tends to overestimate eversion compared to the results based on 3-D studies (see 2.1.5). Current knowledge of the effects of shoe sole modifications during running is based on shoe or skin data while knowledge of skeletal motion changes due to shoe sole modifications is not available. Furthermore, it is not understood whether and how shoe sole modifications may affect tibial and femoral rotations during running.

2.3 Influence of Orthoses on the Kinematics of Running

Foot orthoses (insoles, arch supports, etc.) date back to the 18th century and have been industrially produced since the turn to the 20th century, both in America and Europe (Lovett and Cotton, 1898; Ata-Abadi et al., 1974; Cavanagh, 1980). Traditionally, foot orthoses have been applied in clinical environments treating patients with severe gait problems (Philps, 1990; Wu, 1990; Bowker et al., 1993). Running injuries have been treated with orthoses since the mid 1970s and have attracted biomechanic investigators. Hence, the literature on the effects of orthoses in runners is located both in the orthopaedic and biomechanical areas. The summaries provided here include the results of outcome studies from the orthopaedic area as well as the results of quantified effects of orthoses on movements at the lower extremities in running from the biomechanical area.

2.3.1 Outcome Studies for Orthoses

James et al., (1978) were the first to report the outcome of orthoses applied to runners (n=83) in a retrospective study. The authors found beneficial results in 78% of the runners using rigid (n=39) and flexible (n=44) orthoses. Eggold (1981) reported on 146 runners using individually applied orthoses with a complete relief of the problem in 40%

and partial relief in 35% of the runners. Segesser et al. (1987) reported on 578 orthoses applied to a sports population with long-lasting health problems and a nearly complete reduction of symptoms in 75%. Lohrer (1989) treated 62 runners with medially applied orthoses finding very good results with 34% and good results with 42% using orthoses with a thermoplastic upper and individual cork bottom. Gross et al., (1991) reported 31% with complete resolution of the symptoms and 45% with great improvement based on the feedback from 347 runners; the different orthoses being classified as flexible (63%), semi-rigid (23%) and rigid (14%). Positive results have been achieved in all of these outcome studies applying orthoses on runners with injuries such as medial tibial stress syndrome, chondromalacia patellae, Achilles tendon problems, patellafemoral pain syndrome and others.

In summary, outcome studies report 70-80% positive results from the use of orthoses in runners. This successful outcome is achieved using orthoses of different shapes, different material properties (ranging from flexible to rigid), and different placements inside the shoe. Consequently no consensus or guideline is available which may improve the application of orthoses. Additionally, outcome results may be influenced by subjective assessment and/or subjects not reporting back when orthoses did not alleviate the problem. Therefore, further research quantifying the effects of systematically varied orthoses on movement changes and reduction of symptoms at the lower extremities seems necessary.

Furthermore, the fact that a successful treatment of running injuries can be achieved with flexible, semi-rigid or rigid orthoses leads to the conclusion that orthotic effects may be caused by mechanical and/or proprioceptive reasons. In the first case it is thought that an orthoses exerts a force on the foot which causes it to move, or to reduce an imposed movement, e.g. eversion. In the second case, the application of the orthoses may increase the afferent feedback from cutaneous receptors in the foot (Feuerbach et al., 1994), which may lead to a decresssed eversion because of the muscular contraction of foot inverting muscles. Hence, it is currently unclear whether the positive results of outcome studies are based on mechanical and/or proprioceptive effects during running.

2.3.2 Orthoses in Biomechanical Studies

The study of the effects of orthoses during running has attracted researchers in biomechanics for almost 20 years. The related literature is focused mainly on the influence of orthoses on eversion (pronation), and only a few investigations report on the influence of movements at the tibia and knee.

Effects of Orthoses on Eversion Movements. Nigg et al. (1977) were the first to study effects of orthoses during running. The method developed by this group was adopted by most other subsequent investigators in this area. Nigg et al. (1977) concluded that orthoses should change the gait characteristics (maximum eversion, initial eversion) of subjects with pain towards those of healthy subjects. Cavanagh (1980) found a reduction of maximum eversion with increased orthotic thickness of 2° with each layer added. Bates et al. (1979) and Rodgers et al. (1982), however, did not find significant differences in maximum eversion when running with and without orthoses; but with orthoses, eversion was found to start later and end sooner.

Taunton et al. (1985) reported a significant decrease of maximum eversion with orthoses between 1.6° and 2.5° (± 2.7°) but not for the time needed to achieve that maximum. In addition, a left-right difference was detected for maximum eversion, a finding which was not thoroughly discussed. Smart et al. (1985) found also a significant decrease (of 6.3°±6.2°) of ankle eversion using medially placed orthoses using electrogoniometers. Nigg et al., (1986) found that with orthoses initial eversion of the rearfoot at touchdown was reduced (4.4°) significantly, but maximum eversion was not reduced significantly (2.9°). Furthermore, the comparison between different placements of orthoses showed, that the most posterior position inside the shoe (support at the sustentaculum tali) was the most effective in reducing initial heel eversion.

Kozak et al. (1991) found in a 3-D study a 20% decrease in total eversion with the use of orthoses designed to reduce pronation at the midtarsal joints with no rearfoot posting. In another 3-D study Eng and Pierrynowski (1994) found only modest (and not significant) effects of soft orthoses on subjects with patellofemoral pain syndrome.

Eversion was reduced by 2.5° at foot-ground contact (a parameter related to initial eversion) and 0.8° at midstance in slow running. However, it remains open whether these small effects are biologically relevant with a reported standard deviation of ± 2-3 degrees.

Van Woensel and Cavanagh (1992) showed that during running a varus shoe (a 10° ramp towards the medial side built in the midsole of the shoe) reduced maximum eversion significantly by 9.5° when compared to a normal shoe (no ramp). The valgus shoe (a 10 degrees ramp towards the lateral side) increased maximum eversion by 9.9°. Perry and Lafortune (1995) showed that with the same three shoe modifications the varus shoe reduced maximum eversion in running significantly by 6.7°, the valgus shoe increased maximum eversion by 8.8° when compared to a normal shoe. This result was supported by a study by Milani et al. (1995) who showed that a 8° varus shoe restricted eversion significantly by 5° compared to a normal shoe and that a 8° valgus shoe increased eversion by 8.9°. In all of these three studies the size of shoe sole modification, ranging from the rearfoot to (at least) the base of the first metatarsale was larger when compared to shoe orthoses built inside the shoe; hence a larger effect of the wedged soles compared to the orthotic correction on the lower extremity kinematics may be expected. In the most recent study, Nigg et al. (1998) found an average group effect of less than 1° (not significant) of inserts on total shoe and foot eversion. Furthermore, individual results showed substantial differences between subjects to different tested inserts (soft versus hard).

Effects of Orthoses on the Tibia and Knee. The effect of orthoses on internal tibial rotation or movements at the knee during running are not yet well understood, and the reported results are controversial. Taunton et al. (1985) used triplanar electrogoniometers and found no significant differences for tibiofemoral rotation, knee valgus displacement (which can be interpreted as knee adduction), ankle dorsiflexion and leg abduction with and without orthoses. On the other hand, significant differences were reported on plantarflexion, and on the time intervals between foot strike and maximum ankle dorsiflexion and on various variables describing left-right differences. Eng and Pierrynowski (1994) reported no change in maximum internal tibiofemoral rotation in midstance as a result of soft orthoses in slow running. A similar result was reported by Smart et al. (1985) who found no significant effects on internal tibiofemoral rotations due to orthoses using a 3D goniometer. Kozak et al. (1991) reported a 8% increase of total internal tibial rotation with the use of orthoses. Lafortune et al. (1994) reported that the effect of 10° varus and valgus shoes on internal tibial rotation during walking was not significant, the effect being only between 1°-3° on average. Nigg et al. (1998) found an average group reduction on total internal tibial rotation due to orthoses of less than 1° (not significant).

In summary, the results of studies related to the placement of foot orthoses are controversial. Although some studies suggest that medially applied orthoses of various designs and placements as well as varus wedged shoe soles decrease maximum shoe eversion (Cavanagh, 1978; Clarke et al., 1983; Taunton et al., 1985; van Woensel and Cavanagh, 1992; Perry and Lafortune, 1995; and Milani et al. 1995), other studies showed no significant differences (Bates et al., 1979; Rogers et al., 1982; Nigg et al., 1986; Eng and Pierrynowski, 1994; Nigg et al., 1998). In addition, it has not been established whether foot orthoses have effects on other segments of the lower extremities, i.e. on the rotation of the tibia and femur (Taunton et al., 1985; Smith et al., 1985; Eng and Pierrynowski, 1994, Nigg et al., 1998). Thus, the effects of orthoses on the segmental movements of the lower extremities are currently not well understood.

2.4 Lower Extremity Kinematics and Running Injuries

Running activities are often associated with running injuries. It has been estimated that between 33% and 56% of runners are affected by at least one injury per year (Subotnick, 1977; James et al., 1978; Brody, 1980; Clement et al., 1981; Nigg et al., 1986; van Mechelen, 1992). Among the most common running injuries are Achilles tendon pain, medial tibial stress syndrome (shin splints), and patellofemoral pain syndrome (PFPS). A large number of possible factors are proposed to contribute to the development of these injuries are listed in the related medical literature. In biomechanical studies, excessive eversion and tibial rotations have been associated with the etiology of running injuries.

Since running involves about as much eccentric as concentric contractions (McClay et al., 1990), it has been speculated that the force-velocity relationship during eccentric contractions of muscles may provide some insight into possible mechanisms of injury (Faulkner et al., 1984). However, there is relatively little knowledge about the details of the force-velocity curve as the muscle lengthens (Winter, 1990). Generally, it has been shown, that the slope of the force-velocity curve is about six times greater for slow lengthening than for slow shortening (McMahon, 1984). Therefore, it has been speculated that a velocity variable (i.e. maximum velocity of the Achilles tendon angle $\dot{\beta}_{max}$) may be linked to muscular injuries or injuries at the origin or insertion of a muscle, whereas a variable which describes the total range of motion at a given joint independently of the movement velocity (i.e. total range of Achilles tendon movement, $\Delta\beta_{max}$) may be linked to ligament loading at that joint (Nigg et al., 1986).

2.4.1 Achilles Tendon Pain

Pain at the Achilles tendon occurs in 6-20% of all running injuries (Clement et al., 1984; Johansson, 1986; Lysholm et al., 1987; Marti et al., 1988) and includes problems at the Achilles tendon associated with a variety of clinical symptoms. The etiology of the pain has been related to a large number of factors ranging from training errors to rheumatic origin (Brukner and Khan, 1993; Clement et al., 1981; Kvist, 1994; Archambault et al., 1995). Biomechanical factors which have been proposed to be associated with the onset of Achilles tendon problems include excessive movements from touchdown to take-off (Segesser et al., 1980; Clement et al, 1984; Nigg et al., 1986). At touchdown Smart et al. (1980) and Clement et al.(1984) observed a "whipping action of the tendon". During stance phase and take-off the Achilles tendon has been observed to bend whilst tensile forces are acting (Segesser et al., 1980; Nigg et al., 1986). It has been argued that this bending may produce an asymmetric loading over the medio-lateral cross section of the tendon (Denoth, 1986) which would provide a mechanical explanation for pain at the Achilles tendon. Perhaps linked to this bending of the tendon is the decreased blood supply at the height of the malleoli which has been hypothesized by Clement et al. (1984) and Wilson and Goodship (1994). However, at the present time, no well controlled studies on the relationship between excessive eversion and Achilles tendon pain are available (Kvist, 1994).

2.4.2 Medial Tibial Stress Syndrome (Shin Splints)

In running, muscles of the posterior tibial compartment (tibialis posterior, flexor hallucis longus and flexor digitorum longus) and the soleus muscle are thought to contribute to control eversion during running . These muscles have been discussed with respect to the medial tibial stress syndrome (shin splints) by Segesser et al. (1980), Viitasalo and Kvist (1983), Messier et al. (1988), and James et al. (1990). It was suggested that excessive, eccentric eversion during running may lead to an inflammation of the fascial attachments of these muscles at the tibia and consequently to pain at the origin of these muscles. The terminology of this pain is still under discussion (Batt, 1995), it will be referred to here as medial tibial stress syndrome.

Viitasalo et al. (1983) compared the 2-D rearfoot kinematics of a control group with a moderate and severe pain group. For both pain groups, the following significant differences were found compared to the control group: decreased touchdown angle (the differences being 3.9° to the severe group and 2.4° to the moderate group), increased maximum eversion (differences being 2.4° and 1.4° accordingly), and decreased total eversion (differences being 1.5° and 1.5° accordingly). These results were supported by a further 2-D study (Messier et al., 1988) in which the pain group showed significantly larger maximum eversion (difference 2.7°), larger total eversion, higher maximum eversion velocity (difference 163°/s) and a shorter time to maximum eversion (difference 7.9 ms) than in the control group. It was concluded that maximum eversion and maximum eversion velocity are significant discriminators between groups of runners with and without medial tibial stress syndrome. However, the association of the variables of total eversion and the time to maximum eversion to the medial stress syndrome was not discussed in both of these investigations (Viitasalo et al., 1983; Messier et al., 1988) and needs further clarification.

2.4.3 Patellofemoral Pain Syndrome

The injury of the patella tendon, the patellofemoral pain syndrome (PFPS) has been shown to be one of the most common injuries in runners as well as within the injuries to the knee (25.8%) as seen in a sports medicine clinic (Clement et al., 1981; James, 1990; van Mechelen 1992). However, the etiology of the PFPS is not well understood. A number of factors have been associated with PFPS, one of them being excessive eversion and as a consequence excessive internal tibial rotation (James, 1990). Excessive tibial rotation has been shown *in vivo* and *in vitro* to be influenced by the eversion movement of the foot as well as some anatomical characteristics of the ankle joint complex (see 2.1.7). This combination of movements at the foot, tibia and femur occurs typically in heel-toe running and may be related to the development of PFPS. Specifically, it has been suggested that excessive foot eversion (pronation) "causes a twisting force to develop at the knee" (James et al., 1978; Noakes, 1991; Subotnick, 1977). This force is speculated to pull the knee cap out of its correct alignment which could lead to PFPS, (Clement et al., 1981). Bahlsen, (1988) suggested in a longitudinal prospective study that excessive pronation is associated with excessive internal tibial rotation which may contribute to PFPS.

In a recent study, Stergiou (1996) compared three groups of runners, an ansymptomatic group, a group experiencing PFPS, and a group who had experienced PFPS in the past with no pain at the time of study. Group comparisons revealed that no variable was different across the groups (the variables being maximum tibial rotation with respect to the foot and femur, maximum patellofemoral joint contact force, Q-angle and weekly training difference). However, a stepwise discriminant function analysis revealed that maximum tibial rotation with respect to the femur was able to predict 55% of all cases, which was considered a low prediction percentage. Individual subject analyses showed that a combination of both Q-angle and maximum tibial rotation with respect to the femur and gender may be a good predictor of PFPS. It was concluded, that PFPS is a multi-facorial problem in which several factors play a role in the development of the pain.

2.4.4 Summary

The etiology of most running injuries is not well understood, although excessive eversion and internal tibial rotation have been associated with some injuries among runners. Kinematic variables such as maximum eversion and maximum eversion velocity have been reported to be larger in pain groups (medial tibial pain syndrome) compared to non-pain groups, differences being 2°-3° for maximum eversion and 150-200°/s for maximum eversion velocity. It can be argued that if foot orthoses and/or shoe sole modifications were to have an effect on the etiology of running injuries these effects should be at least of the same order of magnitude . In the search of a possible relationship between shoe and orthotic constructions and running injuries the definition of relevant effects are important. Nigg and Bobbert (1990) proposed an empirical approach, analogous to the present work, which includes the analysis of the possible relationships between (i) shoe construction (and/or orthoses) and the frequency of sport injuries, (ii) kinematics and injuries, (iii) shoe construction (and/or orthoses) and kinematics, and (iv) the demands for shoe construction (orthotic construction) and (v) the verification that the injury frequency in the shoes (orthoses) developed following biomechanical findings is in fact reduced. This thesis focuses on step (iii) of the empirical approach proposed by Nigg and Bobbert (1990) and may provide some conclusions related to step (iv). Hence, in view of the final goal towards the understanding of the etiology of running injuries the present study may contribute some important information.

2.5 Summary of the Review of Literature

The biomechanics of human locomotion has attracted researchers for more than 150 years. Although the measuring techniques have been improved with time, the principle methodology is still the same. Improvements have been made in the analysis of joint and segment motion and in the recognition of the limitations of skin and shoe mounted versus bone pin markers. Also, the limitations of 2-D studies versus 3-D have been quantified and the introduction of joint coordinate systems (JCS) now allows the description of the relative movements between different segments of the lower extremity in three dimensions.

The artefacts introduced by skin and/or shoe movement relative to the underlying bone are considerable. Good agreement between skin and bone is limited to knee flexion/extension. Skin movement artefact may exceed the skeletal range of motion for abduction/adduction and internal/external rotation. Maximum shoe eversion can overestimate calcaneal eversion by 4.3° to 13.1° and is highly shoe dependent. It is likely that current knowledge of the effects of shoe sole modifications and foot orthoses during running is based on studies with inherent projection errors as well as skin and shoe movement artefacts.

Results from movement studies of the ankle joint complex conclude that movements at the AJC can differ greatly between individuals and that there seems to be no distinct movement order of rotations in the human foot. Movement coupling at the AJC has attracted some investigators, but is not yet well understood. During running movement coupling is significant but depends on a variety of different factors.

Movements of the tibia relative to the femur during running take place in all directions, i.e. flexion/extension, leg ab/adduction and internal/external rotations. The coupling mechanism at the knee during running is not well established. Further, effects of shoe sole modifications and orthoses upon movement coupling at the knee have not been investigated yet.

The influence of shoe soles on the kinematics of running has lead to a mechanical explanation upon which most investigators agree: At touchdown the point of application of the sole is located lateral to the subtalar joint axis producing a lever arm which can be influenced by the hardness of the sole material as well as by the heel flare of the shoe sole. As a result, initial eversion and eversion velocity is reduced with a decreased heel flare. This is in contrast to maximum eversion which is reduced during midstance with hard and flared shoe soles. Further, it is currently not well documented whether shoe sole modifications may also affect movements at the tibia and the knee.

Orthoses show 70-80% positive results in clinical outcome studies which is achieved using orthoses of different shapes, material properties and placements. In biomechanical studies the results of the kinematic effects of orthoses are controversial. Although some studies suggest that medially applied orthoses decrease maximum shoe eversion, others show no significant differences. Furthermore, the effects of orthoses on the tibia and femur during running have not been measured yet.

The exact etiology of running injuries (i.e. Achilles tendon pain, medial tibial stress syndrome, patellofemoral pain syndrome) is currently unknown, although these injuries have been associated with excessive eversion, enhanced eversion velocity and internal tibial rotation. Some investigations provide the information that kinematic variables such as maximum eversion and maximum eversion velocity are larger in pain groups compared to non-pain groups, the differences being 2°-3° (for maximum eversion) and 150-200°/s (for maximum eversion velocity). Further, it has been proposed that asynchronous timing of maximum eversion and knee flexion could be the cause of some knee injuries and that the development of pain may be a multi-factorial problem in which several factors play a role.

Therefore, in view of the related literature summarized above, there are four main questions that will be addressed in this thesis:

1. How are the three-dimensional skeletal kinematics of the calcaneus and tibia influenced as a result of modifications in the placement of foot orthoses ?

2. How are the three-dimensional skeletal kinematics of the calcaneus and tibia influenced as a result of shoe sole modifications at the lateral heel ?
3. How are the three-dimensional skeletal kinematics of the calcaneus and tibia influenced as a result of shoe sole and orthotic modifications compared to running barefoot ?
4. How does a movement coupling between shoe and calcaneus, calcaneus and tibia, and tibia and the femur take place from touchdown to midstance and from midstance to take-off during heel-toe running ?

Question 1 is addressed in chapter 4, question 2 in chapter 5, question 3 in chapter 6 and question 4 in chapter 7.

3. METHODS

This chapter explains the details of the methodology used in this study including data collection, data analysis, data reduction, and methodological considerations with respect to rigid body kinematics.

3.1 Data Collection

3.1.1 General Project Description

The experiments were performed at the Department of Orthopaedics, Karolinska Institute at the Huddinge University Hospital, Stockholm, where surgical experience from previous bone pin studies was available (Lundberg, 1989; Karlsson and Lundberg, 1994). This project was part of a larger study which addressed issues of skin movement artefacts (Reinschmidt, 1996). The combination of these research projects allowed optimum use of the time the subjects were under local anesthetic. Ethical approval for the experiments was obtained from the Ethics committee of the Karolinska Hospital and by the Medical Ethics Committee of The University of Calgary.

3.1.2 Subject Selection

Five healthy male volunteers participated as test subjects (age 28.6 ± 4.3 yrs., mass 83.4 ± 10.2 kg, and height 185.1 ± 4.5 cm), who were all injury free with no previous injury history that might have influenced their locomotion patterns. The subjects gave their informed consent to participation in the study, and the entire procedure was explained to them before the testing began. Running trials were performed before the bone pins were surgically implanted and immediately after surgery, to accustom the subjects to walking and running with inserted bone pins.

3.1.3 Preparation of the Subjects (Surgery)

Figure 3-1: Surgical procedure. Left: application of antiseptic (before the application of anesthesia). Right: application of bone pins.

Antiseptic was applied to the posterior lateral aspect of the calcaneus, the anterior aspect of the tibial condyle and the lateral aspect of the femoral epicondyle (Figure 3-1) in the Huddinge Hospital surgery. The skin, subcutaneous tissue and periosteum at these locations on the right leg were then anesthetized locally (Citanest 10 mg/ml). The anesthesia was active for 2-3 hours, leaving enough time for the experiments. Three slots, each about a centimeter long, were cut into the prepared areas of the skin and underlying soft tissues. A 2.5 mm external fixator pin (mini-Hofmann) was inserted with a hand drill approximately one centimeter into the bone of each location. The bone pin tightness was tested by manual manipulation and the three small wounds were covered with sterile gauze. Unlike previous studies (Lafortune et al., 1992; Koh et al., 1992), the femur pin was inserted in the lateral aspect of the femoral epicondyle to decrease the risk of the opposite leg touching the markers during running. The lateral iliotibial band was cut longitudinally (10 to 15 mm in length) for the femur pin insertion, in an attempt to decrease bone pin impingement problems. The surgical application of the three bone pins took about half an hour. The subject walked over to the gait laboratory, where the experiments took place. The pins were removed with a hand drill and the wounds covered in the surgery after testing. Follow-ups of the subjects several months after testing showed no complications except for one subject who took part in a running competition a week after testing and subsequently suffered knee pain for a few weeks during running exercises.

3.1.4 Marker Positioning

The direction of the drill hole is important when drilling holes into the bone. The hole in the calcaneus has to be directed downward such that the pin does not interfere

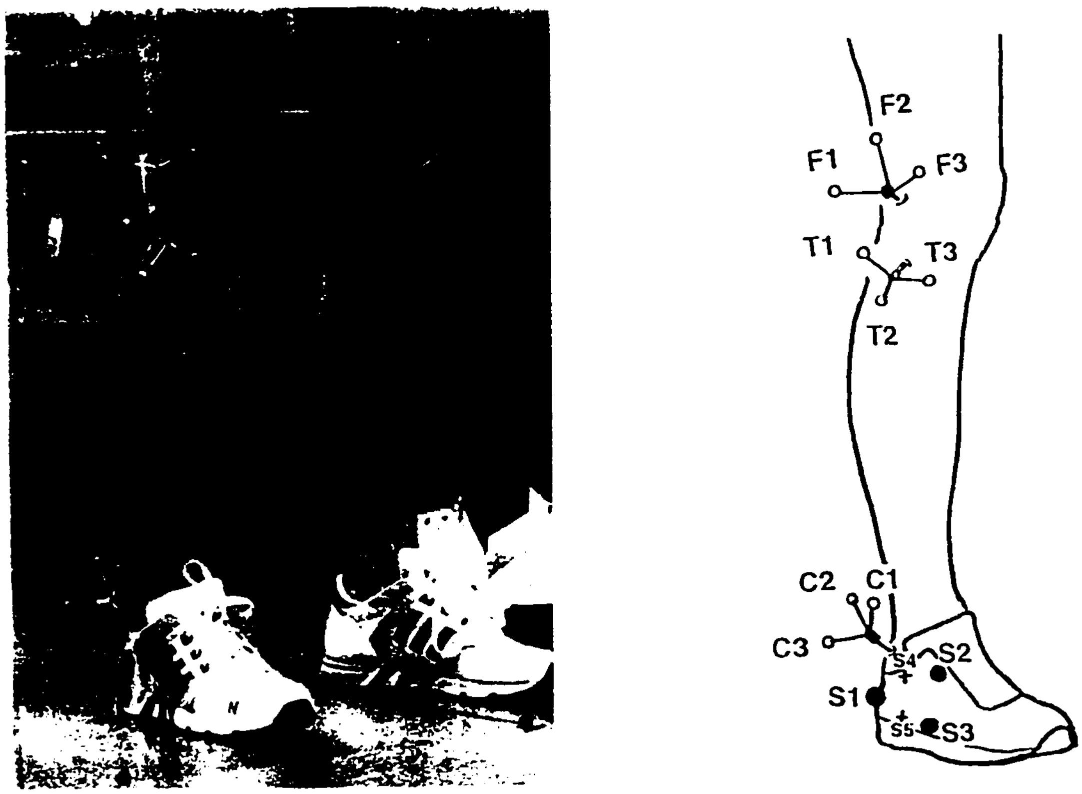


Figure 3-2: Positioning of markers attached to the bone pins and shoe: femur markers (F1 to F3), tibia markers (T1 to T3), calcaneus markers (C1 to C3), and shoe markers (S1 to S5).

with the shoe heel counter (see 3.1.6). The tibial and femoral pins were positioned such that they would not interfere with each other during motion. Immediately after surgery a reflective marker triad was screwed to each bone pin (Figure 3-2). These triads were applied in a similar fashion as in previous investigations (Karlsson et al., 1994). Three markers were glued onto the test shoes, one at the posterior lateral aspect of the calcaneus, and two in the midfoot to avoid marker merging (marker 2 at the dorsum of the foot (location of the lateral cuneiform) and marker 3 at the lateral tuberosity of the fifth metatarsal, Figure 3-2). The effect of shoe marker configuration on eversion was tested in one subject using the auxiliary markers S4 and S5 identified by felt pen marking on the shoe over the calcaneal region. It was found that total shoe eversion was decreased in the order of 2-4° using the auxiliary markers, indicating that marker 2 and 3 may have shown eversion movement including midfoot movement.

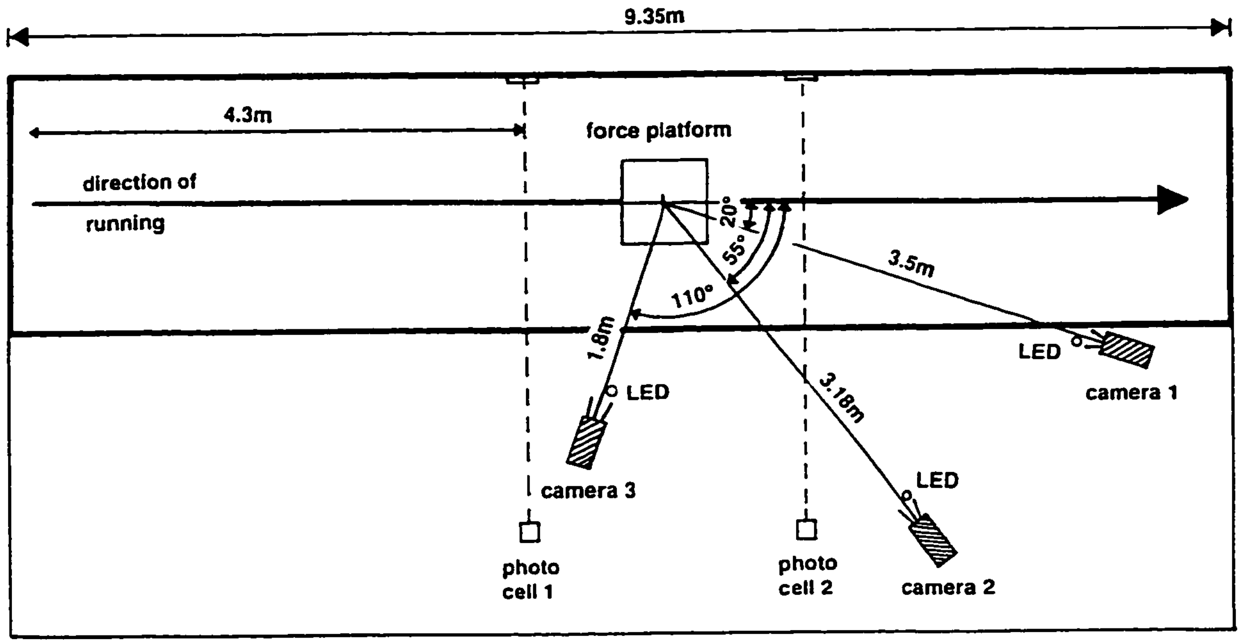


Figure 3-3: Experimental set-up and dimensions

3.1.5 Testing Site and Experimental Set-up

Three high speed cine cameras (LOCAM) were focused in umbrella form on a force platform (KISTLER, Winterthur, Switzerland) which was mounted flush to the runway (Figure 3-3). The camera speed was set at 200 Hz and three LED's (flashing at 200 Hz) triggered by a threshold detector connected to the force plate, were used to synchronize the cameras during the subject's time of contact with the force plate. Fluctuations in camera speed were corrected using the signals of internal camera LED's. A calibration frame with 6 control points (0.5 x 0.5 x 0.5 m^3) was used for the three-dimensional reconstruction. The frame was elevated from the floor by about 15 cm spanning the defined volume from ground to mid-thigh, thus, the foot markers were at the edge or about 5cm below that volume. The total length of the runway was 9.35 m, allowing a 4.3 m run-up to the filming area and enough room to continue and stop.

3.1.6 Testing Procedure and Test Conditions

The subjects performed heel-toe running trials with an average speed of 2.5 to 3.0 m/s with different shoe and orthotic conditions (Table 3-1). The running speed was monitored by photo cells placed 0.7m in front of and behind the force platform. Each test condition was repeated three times except the one with the normal shoe and normal insert, which was repeated five times. Trials were repeated if the subjects did not land with their right foot on the force plate and/or if an obvious modification of the gait pattern occurred. Each of the five subjects performed 23 running trials for this project and a further three running and six walking trials for the other part of this project (see 3.1.1).

Test Condition	Type of Shoe Sole	Type of Orthoses	No. of Trials
1	normal	normal	5
2	normal	anterior	3
3	normal	posterior	3
4	barefoot	---	3
5	straight	normal	3
6	flared	normal	3
7	round	normal	3

Table 3-1: Table of test shoes and orthoses used in the study.

Seven test conditions were used (Table 3-1). An Adidas Equipment Cushion shoe with a 2.8 cm dual density midsole with a midsole hardness of Shore A 35 laterally and Shore A 45 medially was used in the first, "normal", condition (Figure 3-4). The standard insole (of the normal shoe) was assumed to have minimal mechanical foot support. Special orthoses were mounted on the manufacturer's insole in the conditions 2 and 3 (Figure 3-5). These orthoses were made from cork with a 1cm maximum thickness and were thought to support the foot medially at two different locations. The "anterior" orthosis supported the foot arch, whilst the "posterior" orthosis supported the calcaneus at the sustentaculum tali, vertically beneath the medial malleolus. The subject performed the trials barefoot in condition 4. Conditions 5 to 7 (Table 3-1) used the same shoe upper

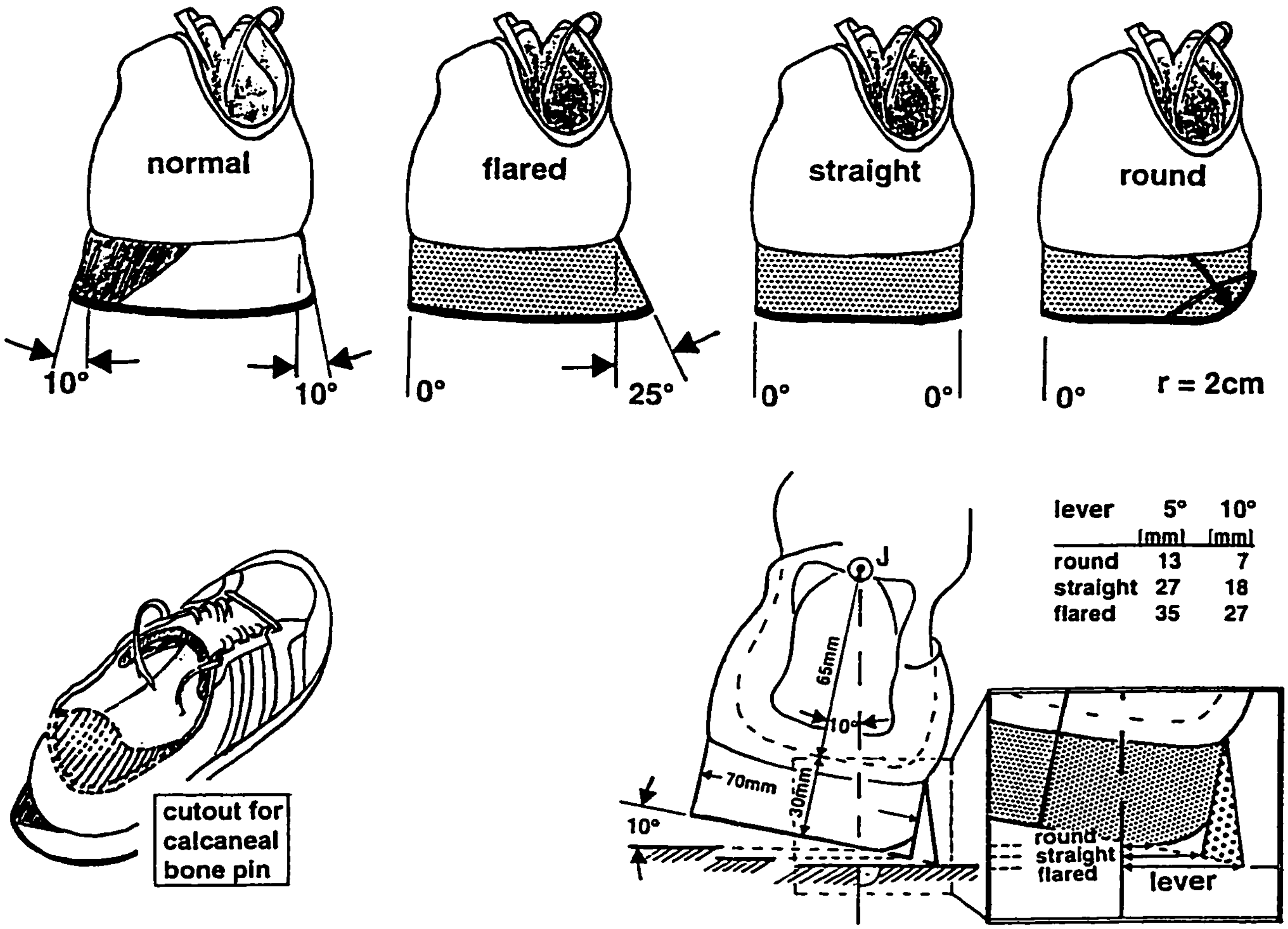


Figure 3-4: Test shoes used in the study (right shoe)

but a single density midsole of Shore A 45 with a hard outer sole of Shore A 65 and with three modifications at the rearfoot. The lateral side of the shoe soles was constructed with a systematic change with a wide flare, a neutral flare or straight sole and a rounded sole. These shoe sole modifications were assumed to produce a different leverage during the initial landing phase and possibly affect foot eversion, since runners typically land on the lateral side of the heel. The heel counter of all shoes used had a lateral cut-out to prevent impingement with the calcaneal bone pin during running.

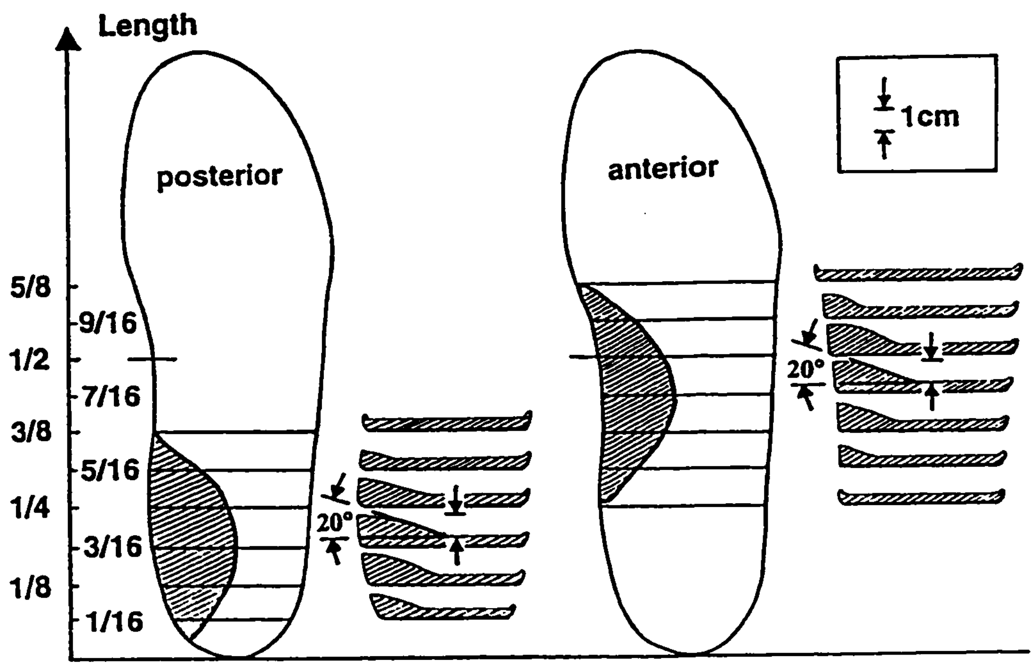


Figure 3-5: Test orthoses used in the study

3.2 Data Analysis and Reduction

3.2.1 Film Analysis

The procedure used to analyze the film followed the specifications developed and described by Reinschmidt (1996). All three camera film sequences, consisting of the running trial, the calibration frame, and the standing barefoot trial, were manually digitized for each trial using the film analysis system of the Biomechanics Laboratory of the ETH Zürich, Switzerland. This system consisted of a projection head (VANGUARD), a digitizing table (HEWLETT PACKARD) connected to a VAX workstation, and a software package developed at the laboratory. Five frames before touchdown and five frames after take-off were digitized in addition to the stance phase to reduce possible filtering artefacts. Three stationary points on the force plate were digitized in addition to the nine pin markers on the subjects, to control single film frame shifts relative to the digitizing board. Markers that were not visible, because of merging or insufficient contrast to the background, were replaced with linear interpolation, although this was never necessary for more than five consecutive frames. Fluctuations in camera speed were corrected using internal camera timing LED signals running at 200Hz which produced markers on the edge of the film and were analyzed during the digitizing session. Generally, each trial was analyzed once, and the entire trial repeated if problems arose with the identification of markers in more than five consecutive frames. One operator digitized the same trial five times each, in order to assess the reliability of the digitizing procedure. The results of this repeatability test are presented in subsection 3.4.3. The raw data of each camera and each trial was visually inspected for outliers and corrected by linear interpolation if necessary.

The marker coordinates were normalized to 101 data points with 0% defining the touchdown and 100% the take-off. The camera coordinates were filtered with a bi-directional 4^{th} order low-pass Butterworth filter with 10 Hz cut-off frequency determined from a residual analysis (Reinschmidt, 1996, after Winter, 1990).

3.2.2 Coordinate transformations

A set of programs was written (Reinschmidt, 1997d) in MATLAB™ (MathWorks, Natick, MA) and adapted for the specific needs of this investigation to reconstruct the three-dimensional marker positions and to calculate the relative segmental movements. The 3-D reconstruction, based on a standard direct linear transformation (DLT) method (Abdel-Aziz and Karara, 1971), was performed for the running and standing trials. All three cameras were used to reconstruct the 3-D position of all markers at any point. The coordinate systems followed the proposal of the ISB Standardization and Terminology Committee (Wu and Cavanagh, 1995). A right handed orthogonal coordinate system was used for the film coordinate system with the x-axis in the forward running direction, the y-axis in the vertical (positive upward) and the z-axis in the lateral direction (for a right limb) defined by the cross product of x and y. The transformation of marker coordinates in one coordinate system, A, to the identical marker coordinates in a coordinate system, B, was expressed as described by Reinschmidt (1996):

$$\mathbf{r_B} = \left[T_{A \to B}\right] \mathbf{r_A} \tag{1}$$

where $\mathbf{r_A}$, $\mathbf{r_B}$ are the location vectors of the coordinate system A and B, respectively. The location vector is:

$$\mathbf{r_A} = \begin{bmatrix} x_A \\ yA \\ z_A \\ 1 \end{bmatrix}$$

and the 4x4 transformation matrix containing rotational (R) and translational (t) information is:

$$\left[T_{A \to B}\right] = \left[\begin{array}{c|c} \left[\mathbf{R}_{3x3}\right] & \left[\mathbf{t}_{3x1}\right] \\ \hline 0 \quad 0 \quad 0 & 1 \end{array}\right]$$

This transformation matrix can be calculated with three or more known markers in the two coordinate systems A and B. The transformation matrix represents the solution of a least square problem defined by Söderquist and Wedin (1993):

$$\min\left(\sum_{i=1}^{n}\left\|\left[T_{A\to B}\right]\cdot r_{Ai} - r_{Bi}\right\|^2\right);\ n \geq 3 \tag{2}$$

The singular value decomposition method described by Söderquist and Wedin (1993) was used for the computation of the transformation matrices.

The barefoot standing trials were used to define the segment-fixed coordinate systems of the calcaneus, tibia and femur. The subjects were instructed to stand with straight knees and the ankle in the "neutral" position at 90° dorsiflexion (after Debrunner, 1982); the subject's feet were aligned parallel to the force plate, which represented the film coordinate system. The investigators checked this procedure visually. It was assumed that during the barefoot standing trials the segment-fixed coordinate systems were aligned with the film coordinate system. This implied that the local coordinates of the markers at the calcaneus ($\mathbf{r}_{cal}$), the tibia ($\mathbf{r}_{tib}$), and the femur ($\mathbf{r}_{fem}$) during the barefoot standing trial were identical to the markers in the film coordinate system, referred to as $\mathbf{r}_G$. This trial was used as the reference for all seven test conditions (Figure 3-6).

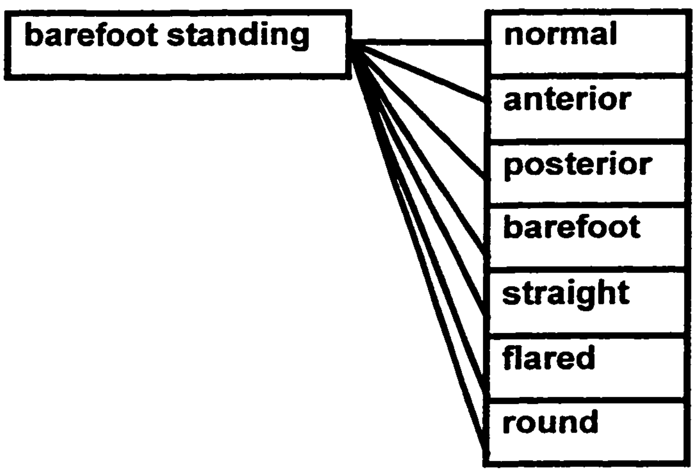


Figure 3-6: The barefoot standing trial used as reference for all test conditions of the running trials.

The following relations were known at any time during running stance between the segment-fixed coordinate systems and the film coordinate system:

$$\mathbf{r}_G = \left[T_{cal\to G}\right]\mathbf{r}_{cal} \tag{3}$$

$$\mathbf{r_G} = \left[T_{tib \rightarrow G}\right] \mathbf{r_{tib}} \tag{4}$$

$$\mathbf{r_G} = \left[T_{fem \rightarrow G}\right] \mathbf{r_{fem}} \tag{5}$$

In the current investigation these relations are referred to as the absolute orientations of the calcaneus, tibia and femur segments in the film coordinate system, leading to absolute movements of these segments relative to the barefoot standing trial. The transformation matrices of the tibia relative to the calcaneus and the femur relative to the tibia were calculated from equations (3) to (5). This led to the description of the relative movements at the ankle complex (AJC) and at the knee.

$$\mathbf{r_{cal}} = \left[T_{cal \rightarrow G}\right]^{-1} \left[T_{tib \rightarrow G}\right] \mathbf{r_{tib}} \equiv \left[T_{cal \rightarrow tib}\right] \mathbf{r_{tib}} \tag{6}$$

$$\mathbf{r_{tib}} = \left[T_{tib \rightarrow G}\right]^{-1} \left[T_{fem \rightarrow G}\right] \mathbf{r_{fem}} \equiv \left[T_{fem \rightarrow tib}\right] \mathbf{r_{fem}} \tag{7}$$

The rotations were calculated as Cardanic angles for the stance phase of all test conditions using a joint coordinate system (JCS) at the ankle joint complex. In/eversion was calculated with the following sequence of rotations of (1) plantar/dorsiflexion about a tibia fixed medio-lateral axis, (2) calcaneus ab/adduction about the floating axis, and (3) in/eversion about the antero-posterior axis of the calcaneus (Cole et al., 1993). Tibial rotation (corresponding to ab/adduction in the above sequence) was calculated with a different sequence to avoid calculations about the floating axis that have limited anatomical relevance: (1) tibial rotation about a tibia fixed proximal-distal (longitudinal) axis (2) in/eversion about the floating axis, and (3) plantar/dorsiflexion about a calcaneus fixed medio-lateral axis (Nigg et al., 1993).

3.3 Definition of Variables

In/eversion and tibial rotation variable definitions are explained in Table 3-2 and depicted Figure 3-7. The variables were defined between touchdown and midstance of running. The inversion positions at touchdown (β_o and ρ_o) were considered to detect possible adaptations to shoe interventions before touchdown. Biomechanical factors, which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. It has been suggested that excessive eversion (i.e. maximum eversion (β_{max}) and total eversion ($\Delta\beta_{max}$)) forces the Achilles tendon to bend laterally, hereby producing an asymmetric stress distribution across the tendon, which could lead to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981; Denoth, 1986). Excessive eversion velocity ($\dot{\beta}_{max}$) produces eccentric loading of the muscles of the posterior tibial compartment, which control/reduce eversion after touchdown. Forces acting on muscle-tendon units during eccentric loading are increased compared to concentric loading. Excessive eversion velocity has been associated with overloading and injury of the muscles of the posterior tibial compartment, e.g. medial tibial stress syndrome (Segesser et al., 1980; Viitasalo et al., 1983; Messier et al., 1988; Stacoff et al., 1988; DeWit et al., 1995). Eversion velocity was defined in the window of 10% to 40% of ground contact time, excluding values before 10% because of possible inaccuracies, e.g. filtering effects and/or markers close to the edge of the defined three dimensional space. After 40%, eversion velocities had reached their maximum in all trials of the test.

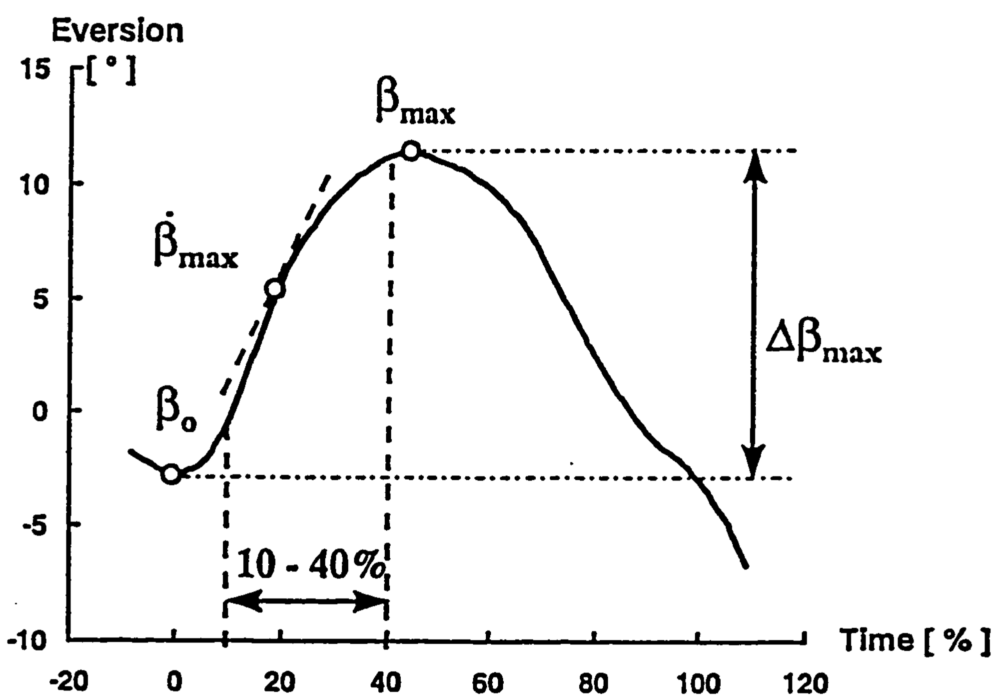


Figure 3-7: Definitions of the study variables. The tibia and shoe variables were defined accordingly.

Excessive tibial rotation ($\Delta\rho_{max}$) has been associated with the changes in the tracking of the patella, hereby changing the contact pressure and possibly the friction of the articulating surface of the patella, which may be related to the occurrence of the patellafemoral pain syndrome (Stergiou, 1996). Tibial rotation is thought to take place as a result of the movement coupling from the calcaneus to the tibia. These variables indirectly describe the movement at those structures of interest, but do not directly describe the load within these structures. However, they are relatively easy to quantify.

VARIABLE	SYMBOL	DEFINITION	JUSTIFICATION
• Touchdown in/eversion • Touchdown tibial rotation	β_o ρ_o	In/eversion position of calcaneus (relative to tibia) and tibial rotation position (relative to calcaneus) at touchdown	• Shoe modifications may affect calcaneal and tibial position before touchdown changing the initial conditions
• Maximum eversion • Total eversion	β_{max} $\Delta\beta_{max}$	Maximum eversion of calcaneus (relative to tibia) during ground contact $= \beta_{max} - \beta_o$	• Excessive eversion has been associated with Achilles tendon problems
• Maximum eversion velocity	$\dot{\beta}_{max}$	Maximum eversion velocity of calcaneus between 10% and 40% of ground contact	• Excessive eversion velocity has been associated with medial tibial stress syndrome
• Maximum internal tibial rotation	ρ_{max}	Internal tibial rotation (relative to calcaneus) during ground contact	• Excessive eversion transferred to excessive internal tibial rotation
• Total internal tibial rotation	$\Delta\rho_{max}$	$= \rho_{max} - \rho_o$	• Excessive tibial rotation has been associated with patella-femoral pain syndrome
• Max. internal tibial rotation velocity	$\dot{\rho}_{max}$	Internal tibial rotation velocity between 10% and 40% of ground contact	• Excessive eversion velocity transferred to excessive internal tibial rotation velocity

Table 3-2: Definition and functional explanation of the study variables. The shoe variables were defined accordingly.

In addition to these variables, eversion of the shoe relative to the tibia was also determined, with the standing trial of each shoe condition being used for the definition of the neutral position for this purpose. The shoe variables were determined to compare the results of this investigation with previous studies using external markers.

The testing procedure (3.1.6) was organized such that test conditions were independent from each other. The residuals of the test variables were inspected for normal distribution. All variables were tested with two-tailed ANOVA techniques with repeated measures, the one-way ANOVA to test subject independent orthotic and shoe sole effects, the two-way ANOVA to test subject dependent effects, as well as possible interactions between the subjects and the orthotic and sole conditions. In cases of contradicting results between the one-way and two-way ANOVA, the more conservative result of the one-way ANOVA was accepted. The power analysis conducted on the kinematic variables suggested that there was a 80% chance of detecting any differences in these variables between the test conditions which were greater than 3.5°.

3.4 Methodological Considerations

This section discusses the limitations and assumptions inherent in the invasive procedure associated with a kinematic study of three-dimensional bone pin movements.

3.4.1 Effect of Insertion of Bone Pins

None of the subjects suffered severe pain or discomfort during the experiment because of the local anesthesia, although one subject experienced minor calcaneus pain during the running barefoot trials. It may be argued, however, that running with inserted bone pins and local anesthesia alters the running pattern of different lower extremity segments. For that purpose, Reinschmidt et al. (1997a) compared three pre-operative running trials of two subjects (2 and 4) with the corresponding post-operative trials (with bone pins). It was concluded that the pre/post-operative knee and ankle joint rotations showed graphs which were similar in shape and magnitude, maximum differences between the two conditions being 2°. This was supported by the observation of the investigators which did not find any obvious differences between pre-operative and post-operative running trials. However, local anesthesia may have disturbed cutaneous feedback which may have altered or diminished the response to changes in shoe and orthotic constructions.

3.4.2 Stability of Femur Pin Attachment

The femur pin fixation encountered some problems in three subjects. The pin in subject 2 became loose at the beginning of testing, thus no femur data are available for this subject. A sudden rotation, of about 10° flexion/extension, of the femur pin in subject 4 occurred during the swing phase when the knee underwent relatively large flexion angles. It was thought that this "popping" rotation was possibly related to an interference between the pin and the iliotibial band, and was obviously faster than the femur rotation. It was decided to discard the femur pin data of subject 4. The femur pin in subject 5 became loose during the last set of trials with the round shoe sole; thus the data of these trials were not included in the study. Therefore, the femur pin data of all subjects should be interpreted with caution. Femur pin problems are described in more details elsewhere (Reinschmidt, 1996).

3.4.3 Accuracy of Spatial Reconstruction

There are a number of possible errors that might occur in a three-dimensional kinematic study (Nigg and Cole, 1994). Those relevant to this study are:

1. Digitizing errors
2. Errors in the three-dimensional reconstruction process
3. Calibration errors
4. Kinematic cross-talk errors

This section provides some possible explanations as to why these errors occurred, how large they might be and what their impact might be on the interpretation of the results.

1. Digitizing errors

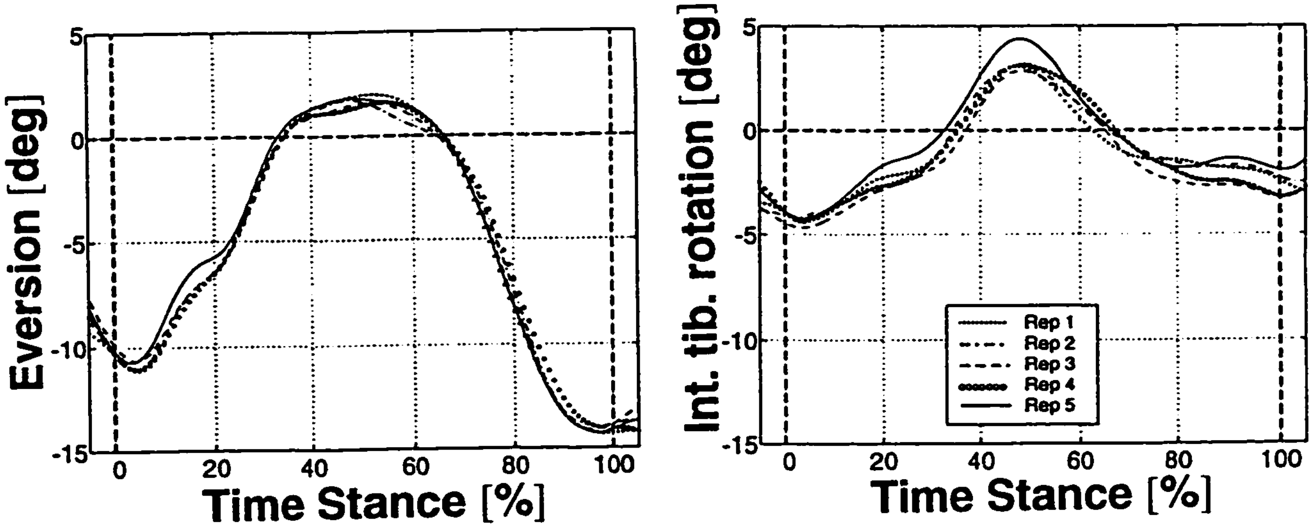


Figure 3-8: Digitizing errors: Eversion and internal tibial rotation of the same trial (subject 1, trial 3, normal shoe) digitized five times by the same operator.

One error source is the digitizing process. The reflective markers may not be digitized constantly at their center, which leads to inaccuracies in the data, since they can change their optical definition during digitizing as a result of a light or dark background. The same trial was digitized five times by the same operator in order to estimate the digitizing process repeatability. The variations of the repetitive digitizing on in/eversion and tibial rotation led to a random error shown in Figure 3-8. The error between the repetitions was found to be between ±0.5° and ±1.0°. The result of this inaccuracy with respect to the three-dimensional reconstruction is discussed below.

2. Errors in the three-dimensional reconstruction process: The DLT method (Abdel-Aziz and Karara, 1971) provides the relationship between 2D marker coordinates on the film from each camera and its 3D location in space. The determination of the spatial coordinates of the markers is a least square problem with residuals. The mean residual errors are calculated from the sum of the squared residuals:

$$\text{mean residual error} = \sqrt{\frac{\sum \text{residuals}^2}{(2n-3)}} \tag{8}$$

where "n" is the number of cameras (in this study n = 3). The denominator is set as such because 2n DLT equations were used to solve the three unknowns X, Y, & Z. The errors can be examined when plotted over time, as shown in Figure 3-9. The plots show that the errors were generally larger at the beginning and the end of each stance phase. This resulted from the camera synchronization problem and from the spatial reconstruction accuracy. The synchronization used allowed the cameras to be synchronized within one frame (200 Hz corresponding to 5ms), thus the time difference between two cameras was never larger than 5ms. The resulting error is expected to be larger during touchdown and take-off where faster marker movements can occur than during the mid-stance phase (Figure 3-9) which explains the larger errors at the beginning and end of each diagram.

The mean residual error during the stance phase was calculated from Figure 3-9 (and those of all other shoe conditions) to be 3mm. Thus, between two markers of a marker triad (with an average distance of 80mm between them), the mean rotational error was ±3.0° (= arc tan $\frac{3\text{mm}}{80\text{mm}}\sqrt{2}$). This error can be regarded as the digitizing error variance in the reconstruction of a constant 80mm distance between two markers of the same triad. Thus, the total error between two marker triads would be:

$$\text{error}_{\text{total}} = \sqrt{\text{error}^2_{\text{tibia}} + \text{error}^2_{\text{calcaneus}}} \tag{9}$$

Allying equation (9) for the example above, the total mean error between the tibia and calcaneal triads was found to be ±4.0° (±4.3° rounded).

However, this estimated error is probably too large because it includes a systematic lens distortion error which should be eliminated if the two triads are within the

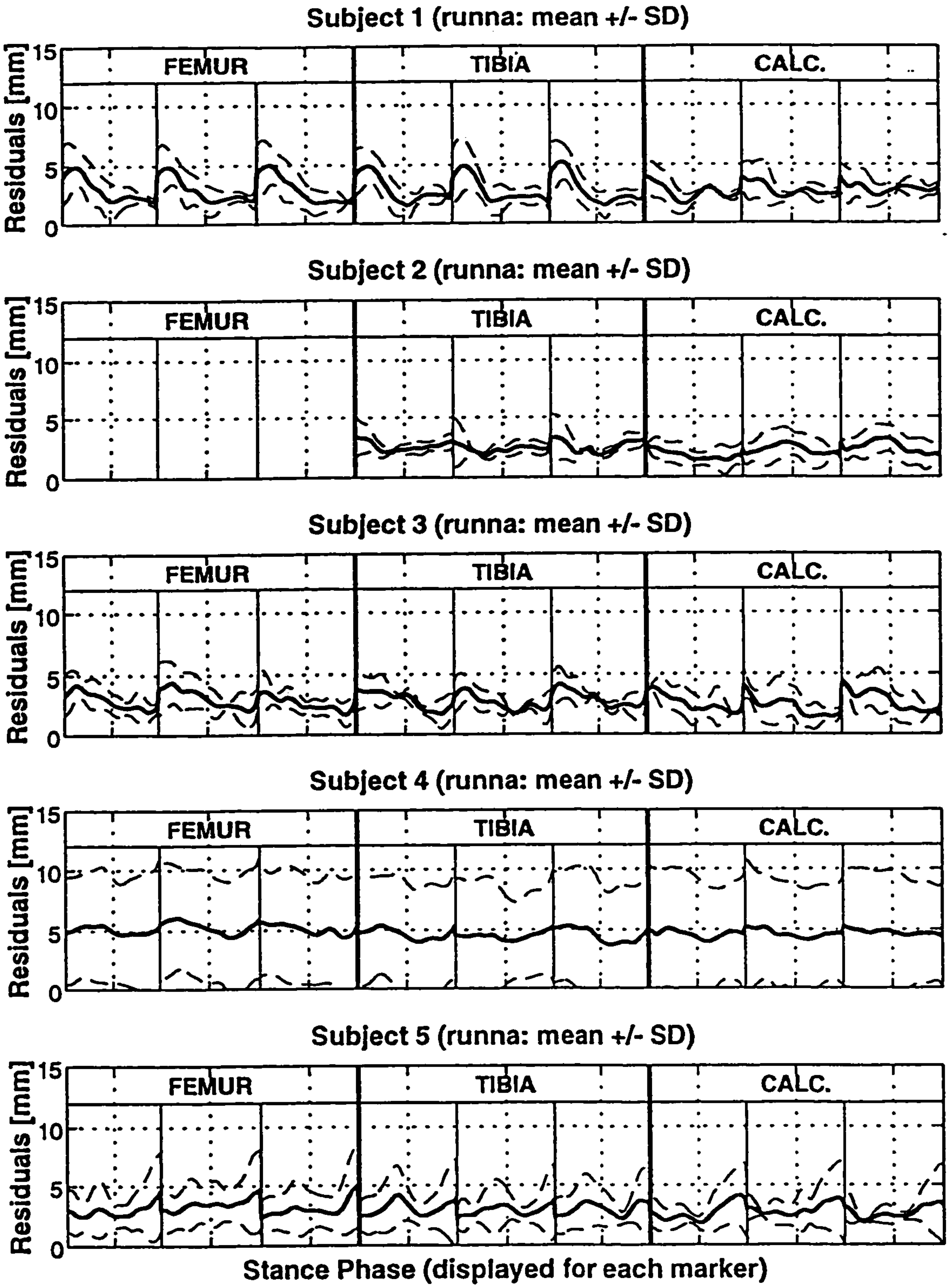


Figure 3-9: The norm of residual errors of the spatial reconstruction (DLT) displayed for all subjects of the anterior orthoses condition. The mean residuals (± SD) of three trials of each marker during the stance phase are plotted. The residual errors are in laboratory units.

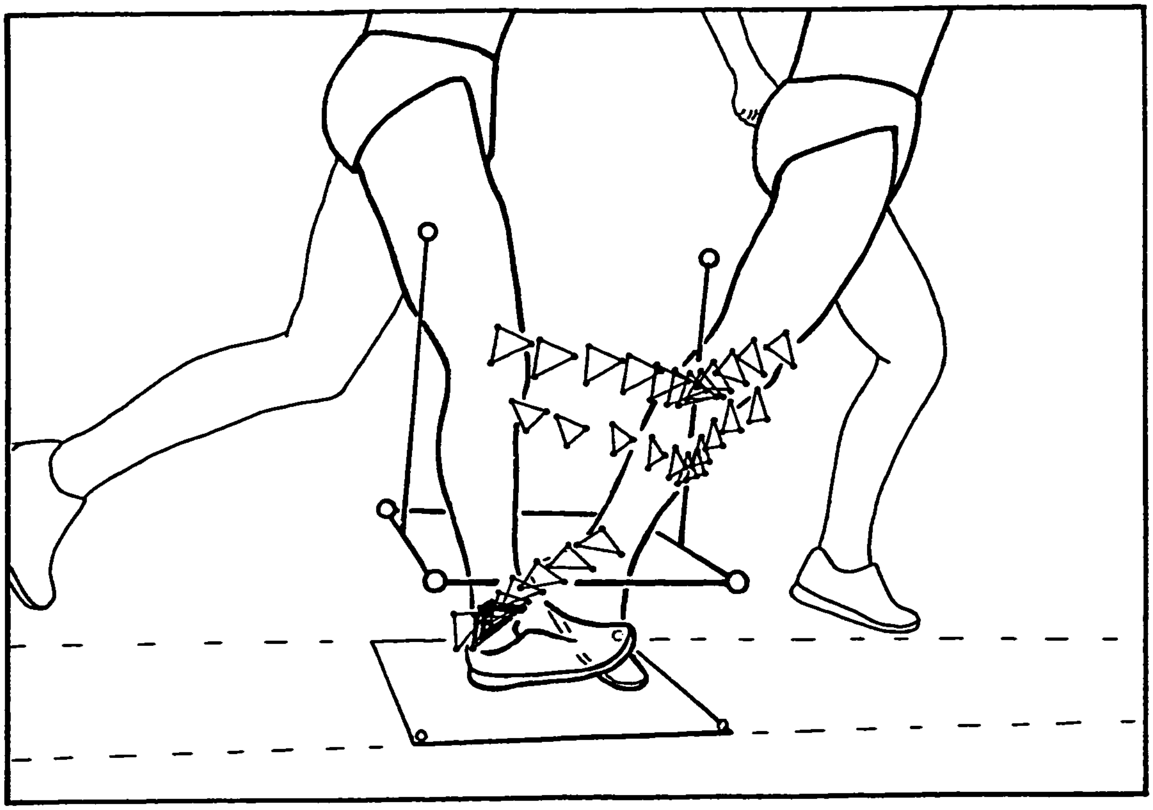


Figure 3-10: Example of one running trial (subject 4) running through the three-dimensionally calibrated space shown as the superimposed calibration frame. The triangles (also superimposed) indicate the femur, tibia and calcaneus marker triad traces from touchdown to takeoff.

same area which primarily applies to the knee joint. At the ankle joint the two triads are farther away such that this error would partially be eliminated. Each triad had a similar lens distortion error since all three markers of the same triad moved in the same area of the film image (Figure 3-10) during testing and consequently the true error within a triad was probably smaller than the residual error shown above. Consequently another error estimation was performed. The marker triads were considered to keep a constant spatial distance between the three markers during running, since they were made from small aluminum rods. Thus, this marker distance in space was calculated using the same procedure as for all other data of the study, that is the correction of outliers (3.2.1), and

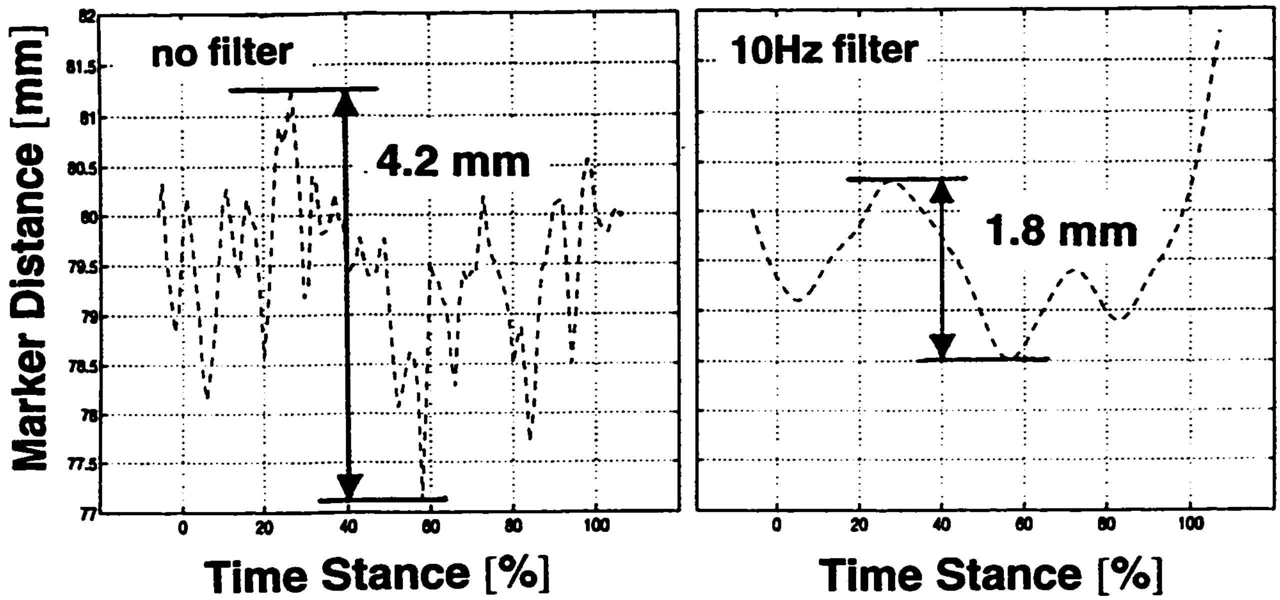


Figure 3-11: Distance between two markers of one triad in space from touchdown to takeoff (distance for marker 4 to 6, trial 3, subject 3, normal shoes). Left: outlier corrected, unfiltered data. Right: filtered with a 4th order low pass Butterworth filter with a 10 Hz cut-off frequency. The example is taken from a low noise trial.

the 3D reconstruction using a least square method (3.2.2). However, as a result of the errors due to digitizing and the 3D reconstruction process, this (constant) distance showed a variation (Figure 3-11). When using the 10Hz filter (the same filter that was used for all other data of this study) the maximum amplitude was reduced (Figure 3-11). Furthermore, it can be observed that the error increased dramatically at the end of the filtered curve, which influenced one variable of the study (movement coupling of the unloading phase, chapter 7). This marker distance in space was calculated for three selected running trials (one trial each with low, medium and high noise; Figure 3-12). The maximum amplitude can be regarded as the worst-case error, as shown in Figure 3-11, not an average inaccuracy over the entire stance phase. Hence, the RMS (the mean deviation) was calculated between 0% and 100% of the stance phase for the outlier corrected (unfiltered) data as well as for the filtered data. All of these error estimations

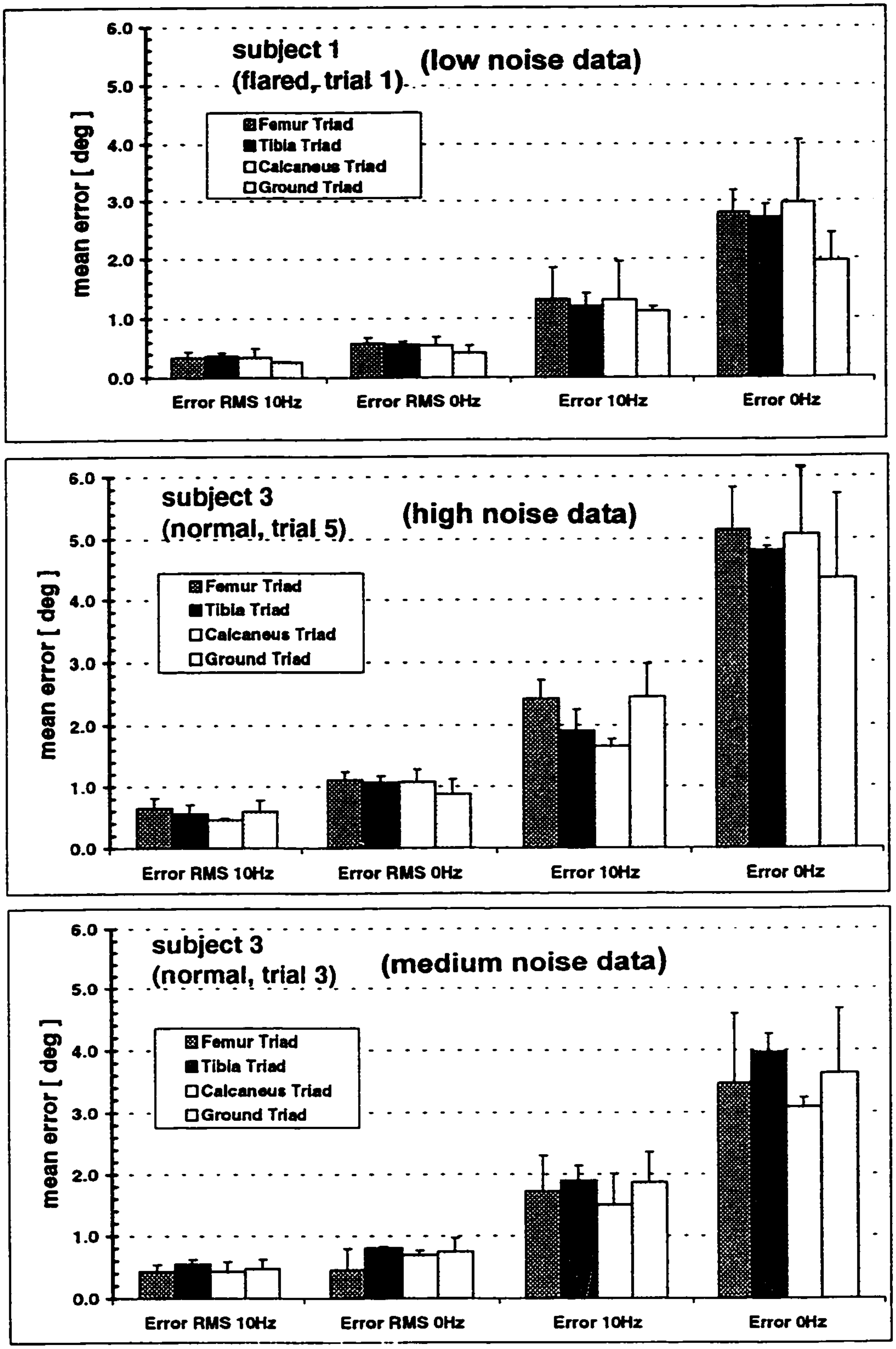


Figure 3-12: Mean errors of the marker distances 1-2, 1-3 and 2-3 of the femur, tibia, calcaneus and ground triads (force plate) under different conditions (from left to right): RMS error of the 10Hz filtered data, RMS error of unfiltered data, mean error of maximum amplitudes filtered with 10Hz, and mean error of maximum amplitudes (unfiltered).

were then translated into rotational errors using a marker distance of 80mm, as for the example above (residual error). The mean error of each triad is represented in Figure 3-12, i.e. the distances between marker 1 to 2, marker 1 to 3, and marker 2 to 3 of each triad (femur, tibia, calcaneus) and the ground (i.e. the three markers on the force plate). It is evident that the unfiltered error was largest and the filtered RMS error was smallest. The Figure shows that the average error (RMS filtered with 10Hz) produced for a constant marker distance was on average about ±0.5° for one marker, thus, ±0.7° for two markers ($0.5\sqrt{2}$), and ±1.0° for two marker triads (using equation (9)). Therefore, compared to the mean residual error (±4.0°, see above) which included lens distortion, the error estimation without lens distortion was reduced.

3. Calibration errors: The coordinates of the available calibration frame was measured exactly to 0.001mm, but was limited because of its size (0.5m x 0.5m x 0.5m) and the number of calibration points (6 only). In general, the accuracy of a spatial reconstruction is reduced when small numbers of calibration points are used (Hatze, 1988) and/or when markers are reconstructed that are outside the calibrated volume (Wood et al., 1986). This was the case for the calcaneal markers (Figure 3-11), and partially for the femur and tibia markers during touchdown and take-off. However, it was not possible to estimate by how much these errors could be reduced, if another calibration frame spanning a larger volume with more calibration points was used.

4. Kinematic cross- talk: Reinschmidt (1996) defined cross-talk, in the context of Cardan angle measurements, as the amount of rotation being registered for a specific angle caused exclusively by another rotation as a result of an ill-defined anatomical coordinate system. When a pure knee flexion occurred during midstance phase of running, for example, ab/adduction movement and internal/external knee rotation of 5° to 7° would be registered (Reinschmidt, 1996).

The barefoot standing trial was used as a reference for all shoe conditions (Figure 3-6), in order to "reduce" the effect of the cross-talk. This resulted in the cross-talk error within each subject becoming systematic and enabled comparisons between the different shoe conditions based on bone marker data. However, the results based on shoe markers remained influenced by cross-talk because of the different standing trials of the various shoe conditions. The following procedural steps were taken to find the magnitude of this cross-talk relative to shoe eversion.

First, the standing trial was perturbed by ±5° ab/adduction and the effect of this perturbation was studied for arbitrarily selected five running trials. Figure 3-13 shows the result and the effects of the rotated data on the original data (shown as 0° rotation). This perturbation was thought to simulate the range of different standing positions and the possible cross-talk effects on eversion. The results show that the cross-talk effect on total eversion between two 5° increments was in the order of ±1°. The largest effect was found between midstance and takeoff, which was irrelevant to this study.

Then the range of foot positions of all the study standing trials had to be found. This was done by examining the film and measuring the alignment of the foot relative to the force plate (see 3.2.2). The long axis of the foot (defined by the center of the heel to the second toe) was compared with direction of the force plate for this purpose. It was found that all subjects placed the long axis of their feet within an accuracy of ±1.5 cm relative to the force plate which corresponded to alignment errors of between 1.9° ± 0.8° (smallest, subject 4) and 3.2° ± 2.1° (largest, subject 3). Thus, the errors due to ab/adduction of foot positions in the standing trials of all subjects fell within one 5° increment which led to a maximum cross-talk error on shoe eversion in the order of ±1°.

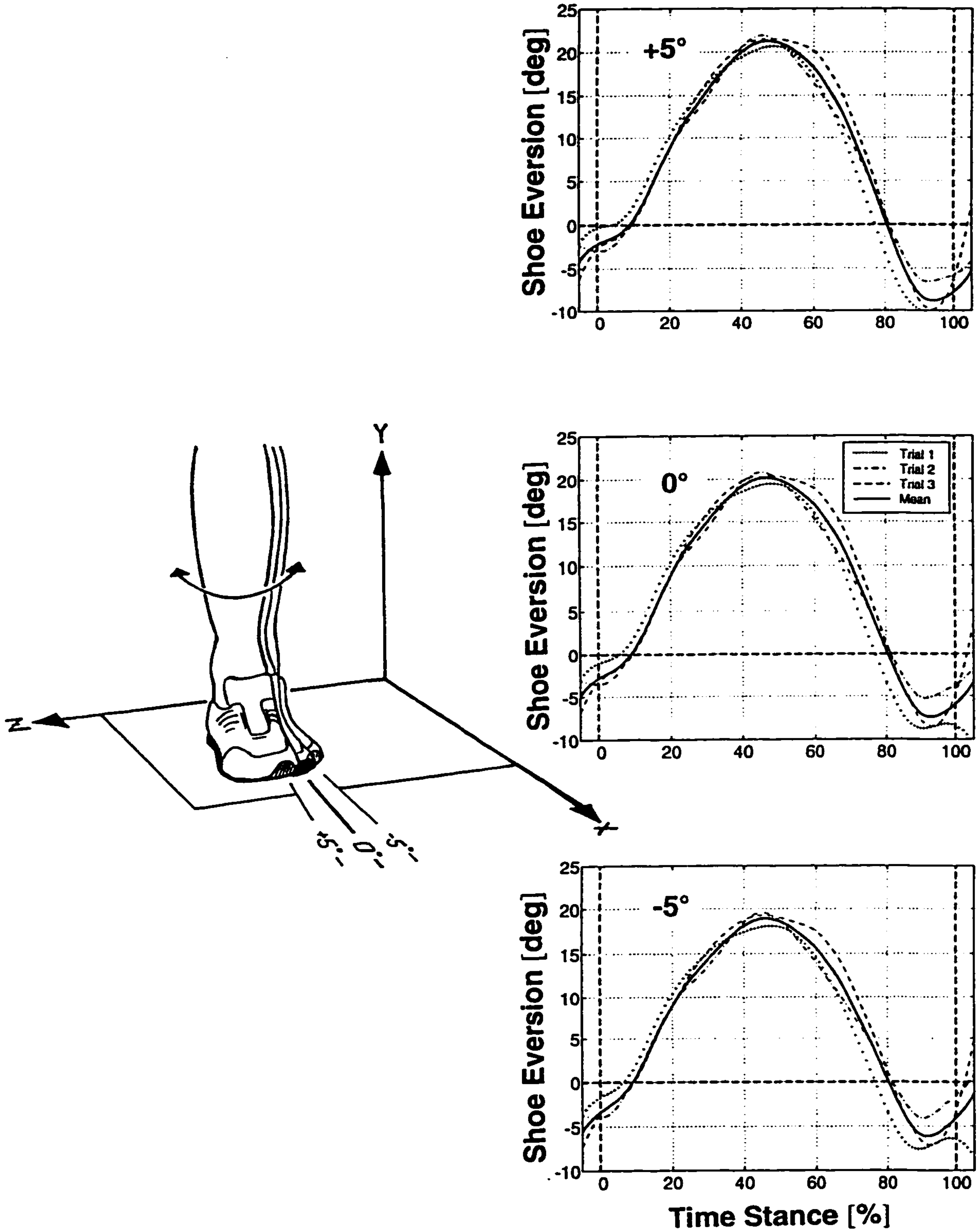


Figure 3-13: Perturbation (ab/adduction) of the standing trail to study the cross-talk effect on shoe eversion.

5. Summary: Inaccuracies in the entire filming, digitizing and 3D reconstruction process must be considered as limitations in the interpretation of the results. The estimated magnitudes of the possible errors using the residual approach and the marker distance approach between two sets of marker triads are listed in Table 3-3. In conclusion, the accuracy of the spatial reconstruction between two marker triads was determined (i) based on the residuals of the DLT equations averaged over the entire stance phase for all markers and (ii) based on the deviations of the inter-marker distances of the same trials. The mean error based on DLT residuals was found to be in the order of ±4° which included noise error and lens distortion error. The mean error based on marker distances (RMS) was found to be ±1.0° including noise error only. Thus, a realistic error estimation of the present study was likely to be between the two errors given above. The error of the shoe data was about ±1.0° higher than that at the bone, because it included inaccuracies of different standing trials with different shoes.

Type of error analysis	**estimated error [°]**
• repeatability	±0.5° to ±1.0°
• mean residuals (with lens distortion)	±4.0°
• mean triad distances (without lens distortion)	±1.0°
• cross-talk (shoe markers)	±1.0°

Table 3-3: Estimation of errors between two sets of marker triads.

4. EFFECTS OF FOOT ORTHOSES ON SKELETAL MOTION DURING RUNNING

4.1 Introduction

Excessive eversion and excessive tibial rotation have been associated with various running injuries (James, 1978; van Mechelen, 1992). Excessive eversion has been linked to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981) and to shin splints (Segesser et al., 1980; Viitasalo et al., 1983), whereas excessive tibial rotation has been associated with the development of knee injuries (Bahlsen, 1988; James et al., 1990; van Mechelen, 1992).

To reduce and control excessive movements, foot orthoses or shoe inserts are often applied medially inside the shoes. Studies analyzing the effect of such orthoses administered to injured runners generally report 70-80% positive outcomes (James et al., 1978; Eggold, 1981; Segesser et al., 1987; Lohrer, 1989; Gross et al., 1991). This is a rather surprising result since the orthoses tested in these studies differed considerably in shape, material properties (ranging from flexible to rigid) and placements. The lack of a consensus on the appropriate application of shape, material properties and placement of foot orthoses indicates that the knowledge on which these decisions were based is small and incomplete.

Effects produced by orthoses may be the result of mechanical and/or proprioceptive mechanisms. Orthoses are thought to reduce foot eversion and/or increase the afferent feedback from cutaneous receptors in the foot (Feuerbach et al., 1994), which is assumed to change the innervation pattern and, consequently, the movement. However, the quantification of these cause and effect processes is not trivial and the determination of the actual skeletal foot movement is difficult, since skeletal kinematics are masked by soft tissue movements (Cappozzo et al., 1996; Reinschmidt et al., 1997a).

Several groups have studied the effect of foot orthoses on rearfoot movement using various orthotic designs, materials, and placements as well as varus wedged shoes, but the results were inconsistent. Some authors found significant differences in rearfoot movements as a result of these interventions (Cavanagh, 1978; Taunton et al., 1985; Clarke et al., 1984; van Woensel et al., 1992; Milani et al., 1995) whilst others did not (Bates et al., 1979; Rodgers et al., 1982; Eng et al., 1994). Nigg et al. (1986) reported a reduction of initial pronation (eversion) as a result of medial orthoses but not of total pronation, and found that a posterior support inside the shoe (support beneath the sustentaculum tali) was more effective in reducing initial eversion than more anterior placements. However, the reasons for these results are not well understood.

Foot movement is transferred to the tibia by a coupling mechanism (Hicks, 1953; Inman, 1969; Olerud et al., 1985; Lundberg, 1989; Hintermann et al., 1994). Consequently, it has been proposed that excessive eversion may be transferred into excessive tibial rotation (Subotnick, 1977; Segesser et al., 1980; Clement et al., 1981; James et al., 1990). Thus, it may be concluded that orthoses may have an effect on this movement coupling and may consequently affect tibial rotation. However, effects of orthoses on the transfer of the foot movement to the tibia during running have not yet been studied, and hence, orthotic effects on the kinematics of the lower extremities are currently not well understood.

Studies related to the kinematics of running and orthotic effects are based on skin or shoe mounted marker settings. Recent studies comparing skin/shoe markers with bone pin markers indicate that externally mounted markers overestimate the movements of the underlying bone (Cappozzo et al., 1996; Reinschmidt et al., 1997a). Therefore, external markers cannot be used to obtain skeletal kinematics information. Hence, the purpose of this study was to quantify the effect of medially placed orthoses on calcaneal eversion and tibial rotation using markers mounted on bone pins. The hypotheses to be tested in this study were:

I. Medially placed orthoses (anterior and posterior) decrease maximum calcaneal eversion and internal tibial rotation compared with no orthoses.
II. Posterior orthoses are more effective in decreasing maximum calcaneal eversion and internal tibial rotations compared with anterior orthoses.

4.2 Methods

4.2.1 General Project Description

The experiments were performed at the Department of Orthopaedic, Karolinska Institute at Huddinge University Hospital, Stockholm, where previous bone pin studies have been carried out (Lundberg, 1989; Karlsson and Lundberg, 1994). The project was part of a larger study performed at the University of Calgary, Canada (Reinschmidt, 1996). Ethical approval for the experiments was obtained from the Ethics committee of the Karolinska Hospital and by the Medical Ethics Committee of The University of Calgary.

Five healthy male volunteers participated in this study (28.6 ± 4.3 yrs., mass 83.4 ± 10.2 kg, height 185.1 ± 4.5 cm); they were all injury free and had no previous injury history which may have influenced their locomotion patterns. The subjects gave their informed consent to participate in the study, and the entire procedure was explained to them before testing. The subjects familiarized themselves with the running procedure before surgery and again before being filmed with inserted bone pins. Intracortical Hofmann pins with reflective marker triads were inserted under standard local anesthetic (Citanest 10 mg/ml) which was active for 2-3 hours, leaving enough time for the experiments. Two bone pins were drilled into the posterior lateral aspect of calcaneus and the anterior lateral aspect of the tibial condyle. Immediately after surgery, reflective marker triads were screwed onto each bone pin (Figure 3-2). Subject follow-ups, several months after testing, showed no complications except for one subject who took part in a running competition a week after testing and subsequently suffered knee pain for a few weeks during running exercises. Three markers were glued onto the test shoes, one at the posterior lateral aspect of the calcaneus, and two in the midfoot, to avoid marker merging (marker 2 at the location of the lateral cuneiform and marker 3 at the lateral tuberosity of the fifth metatarsal, Figure 3-2). The effect of shoe marker configuration on eversion was tested on one subject using auxiliary markers 4 and 5 identified by felt pen marking of the shoe over the calcaneal region.

4.2.2 Experimental Set-up and Testing Procedure

Three high-speed cine cameras (LOCAM) were focused in umbrella form on a force platform (KISTLER) which was mounted flush to the runway (Reinschmidt, 1996). The camera speed was set at 200 Hz and a LED, triggered by a threshold detector during the time of contact of the subjects on the force plate, was installed in the field of view of each camera. The signal was visible during the entire stance phase, determining its duration and allowing camera synchronization. Fluctuations in camera speed were corrected using the signals of internal camera timing LED signals (chapter 3, Reinschmidt, 1996). A calibration frame with six control points (0.5 x 0.5 x 0.5 m^3) was used for the three-dimensional reconstruction.

The accuracy of the spatial reconstruction between two marker triads was determined (i) based on the residuals of the DLT equations averaged over the entire stance phase for all markers and (ii) based on the deviations of the inter-marker distances of the same trials. The mean error based on DLT residuals was found to be in the order of ±4° which included noise error and lens distortion error. The mean error based on marker distances (RMS) was found to be ±1.0° including noise error only. Thus, for the present study, a realistic estimation of the error was likely between the two errors given above. The error of the shoe data was about ±1.0° higher than that at the bone, because it included inaccuracies of different standing trials with different shoes.

The subjects performed heel-toe running trials with a running speed of between 2.5 and 3.0 m/s measured with two photo cells placed 0.7m in front and behind the force platform. Each of the test conditions was repeated three times with the exception of that with no orthoses, which was repeated five times. Trials were repeated if the subjects did not land with their right foot on the force plate or if they obviously modified their step length in order to hit the force plate.

4.2.3 Orthoses used in the Study

The tests were performed with three orthotic conditions. The test shoes (Adidas Equipment Cushioning) had a 2.8 cm dual density midsole with a midsole hardness of Shore A 35 laterally and Shore A 45 medially. The heel counter of the right shoe had a specially constructed cutout to prevent impingement with the calcaneal pin (Figure 3-4). In the first condition, the standard manufactured insole was used, which was assumed to have minimal mechanical support. In the second and third conditions, special orthoses were mounted onto the manufacturers insole (Figure 3-5). The orthoses were made from cork with a 1cm maximum thickness and were thought to support the foot at two different locations: The anterior orthosis supported the foot arch, the posterior orthosis supported the calcaneus at the sustentaculum tali, vertically beneath the medial malleolus.

4.2.4 Data Analysis and Reduction

The procedure used to analyze the film followed the specifications developed and described by Reinschmidt (1996). The films were manually digitized including five frames before and after the stance phase for filtering purposes. Camera coordinates were filtered with a bi-directional 4th order low-pass Butterworth filter with a 10 Hz cut-off frequency. The marker coordinates were normalized to 101 data points with 0% defining the touchdown and 100% the take-off. KineMat, a set of programs written in MATLAB™, was adapted from Reinschmidt and van den Bogert (1997b) for the specific needs of this investigation to reconstruct the three-dimensional position of the markers and to calculate the relative segmental movements. The 3-D reconstruction, based on a standard direct linear transformation method (Abdel-Aziz and Karara, 1971), was performed for the running trials and one standing barefoot trial of each subject. The barefoot trial was used as the neutral position, to define the segment-fixed coordinate systems of the calcaneus and tibia. For that purpose the subjects were instructed to stand with straight knees, the ankle in the neutral position of 90° dorsiflexion and the feet aligned parallel to the force platform representing the laboratory coordinate system. This implied that during barefoot standing all joint rotations equaled zero. The standing trials with the respective shoe condition was used for the shoe marker analysis.

The rotations were calculated as Cardanic angles for the stance phase of all test conditions using a joint coordinate system (JCS) at the ankle joint complex. In/eversion was calculated with the following sequence of rotations: (1) plantar/dorsiflexion about a tibia fixed medio-lateral axis, (2) foot ab/adduction about the floating axis, and (3) in/eversion about the antero-posterior axis of the foot (Cole et al., 1993). Tibial rotation ("corresponding" to ab/adduction in the above sequence) was calculated with a different sequence to avoid calculations about the floating axis which has limited anatomical relevance: (1) tibial rotation about a tibia fixed proximal-distal (longitudinal) axis (2) in/eversion about the floating axis, and (3) plantar/dorsiflexion about a calcaneus fixed medio-lateral axis (Nigg et al., 1993).

4.2.5 Definition of Variables

In/eversion and tibial rotation variable definitions are explained in Table 3-2 and depicted Figure 3-7. The variables were defined between touchdown and midstance of running. The inversion positions at touchdown (β_o and ρ_o) were considered to detect possible adaptations to shoe interventions before touchdown. Biomechanical factors, which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. It has been suggested that excessive eversion (i.e. maximum eversion (β_{max}) and total eversion ($\Delta\beta_{max}$)) forces the

Achilles tendon to bend laterally, hereby producing an asymmetric stress distribution across the tendon, which could lead to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981; Denoth, 1986). Excessive eversion velocity ($\dot{\beta}_{max}$) produces eccentric loading of the muscles of the posterior tibial compartment, which control/reduce eversion after touchdown. Forces acting on muscle-tendon units during eccentric loading are increased compared to concentric loading. Excessive eversion velocity has been associated with overloading and injury of the muscles of the posterior tibial compartment, e.g. medial tibial stress syndrome (Segesser et al., 1980; Viitasalo et al., 1983; Messier et al., 1988; Stacoff et al., 1988; DeWit et al., 1995). Eversion velocity was defined in the window of 10 to 40% of ground contact time, excluding values before 10% because of possible inaccuracies, e.g. filtering effects and/or markers close to the edge of the defined three dimensional space. After 40%, eversion velocities had reached their maximum in all trials of the test. Excessive tibial rotation ($\Delta\rho_{max}$) has been associated with the changes in the tracking of the patella, hereby changing the contact pressure and possibly the friction of the articulating surface of the patella, which may be related to the occurrence of the patellafemoral pain syndrome (Stergiou, 1996). Tibial rotation is thought to take place as a result of the movement coupling from the calcaneus to the tibia. These variables indirectly describe the movement at those structures of interest, but do not directly describe the load within these structures. However, they are relatively easy to quantify. In addition to these variables, eversion of the shoe relative to the tibia was also determined, with the standing trial of each shoe condition being used for the definition of the neutral position for this purpose. The shoe variables were determined to compare the results of this investigation with previous studies using external markers and to quantify the relative movement of the shoe and the calcaneus caused by slipping of the heel inside the shoe. However, it has to be kept in mind, that two of the shoe markers were placed at the midfoot. Thus, strictly spoken, shoe eversion of the present study was a combination of shoe eversion at the calcaneus and at the midfoot.

The testing procedure (4.2.2) was organized such that test conditions were independent from each other. The residuals of the test variables were inspected for normal

distribution. All variables were tested with two-tailed ANOVA techniques with repeated measures, the one-way ANOVA to test subject independent orthotic and shoe sole effects, the two-way ANOVA to test subject dependent effects, as well as possible interactions between the subjects and the orthotic and sole conditions. In cases of contradicting results between the one-way and two-way ANOVA, the more conservative result of the one-way ANOVA was accepted. The power analysis conducted on the kinematic variables suggested that there was a 80% chance of detecting any differences in these variables between the test conditions which were greater than 3.5°.

4.3 Results

Eversion and tibial rotation movement patterns are presented in Figure 4-1 (single curves of a typical subject) and Figure 4-2 (mean curves of each condition for each subject). Eversion and internal tibial rotation took place from touchdown until midstance, thereafter, the movements reversed to inversion and external tibial rotation until take-off. These general movement patterns were found to be consistent for all subjects and test conditions.

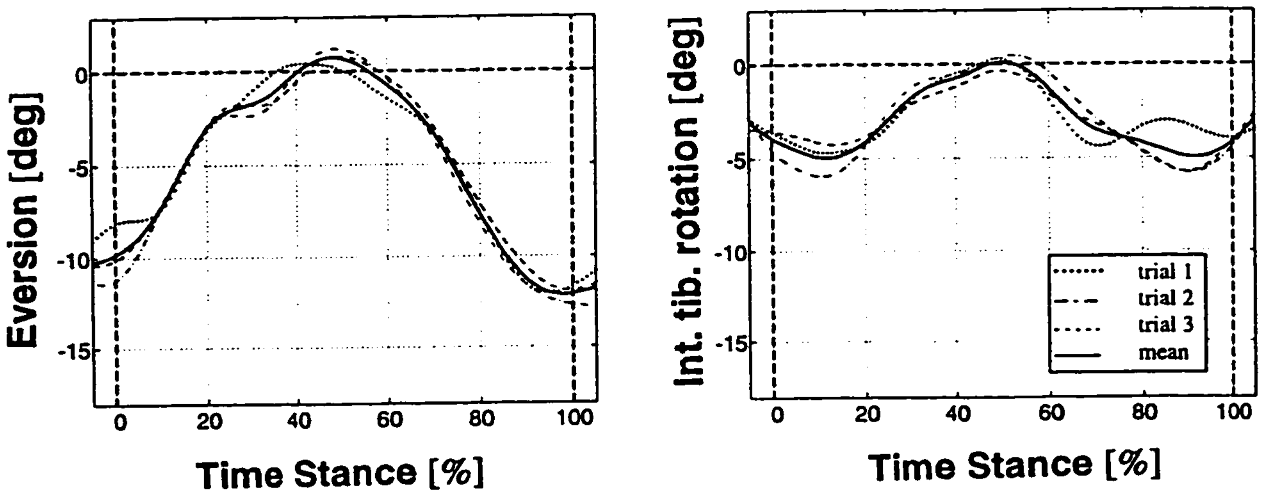


Figure 4-1: Example of in/eversion and tibial rotation (subject 1 with anterior orthoses). Thin lines: three repetitions, thick lines: mean curve. Labels on the vertical axes indicate movements in the positive direction.

The ankle at touchdown was on average inverted and the tibia externally rotated by a few degrees. Maximum eversion averaged between 3.1° and 4.1° (mean β_{max}) and total eversion averaged between 7.9° and 8.2° (mean $\Delta\beta_{max}$) for all conditions and subjects (Table 4-1).

Total internal tibial rotation averaged between 3.2° and 4.5° (mean $\Delta\rho_{max}$). The maximum eversion velocity averaged between 130°/s and 137°/s and the difference between total skeletal and total shoe eversion averaged between 5.8° and 7.3°, hence the calcaneus moved relative to the shoe markers in all subjects under all shoe conditions.

These results show that during the stance phase of running at 2.5 to 3 m/s foot orthoses had no substantial effects on skeletal calcaneal and tibial kinematics. Mean differences between the test conditions were less than 1.6° and 10°/s, but differences between subjects were up to 10° (Table 4-1). The only variable that showed a significant difference independent of the subjects was the total internal tibial rotation ($\Delta\rho_{max}$) which was reduced as a result of medial orthoses ($p<.05$) (Table 4-2). Although subject 5 showed a very low value (Table 4-1: 1.27°) influencing the test results, all other subjects

Variable	condition	subject 1	subject 2	subject 3	subject 4	subject 5	mean SD
β_0 [°]	normal	-8.37 ± 1.89	-6.80 ± 1.20	0.10 ± 0.61	-3.53 ± 2.24	-2.16 ± 1.47	-4.15 ± 3.44
	posterior	-9.64 ± 0.97	-7.66 ± 0.64	0.58 ± 0.32	-2.98 ± 1.13	-4.93 ± 0.51	-4.92 ± 3.99
	anterior	-9.83 ± 1.60	-7.02 ± 0.61	-0.53 ± 1.49	-1.91 ± 0.52	-3.90 ± 1.14	-4.64 ± 3.79
βmax [°]	normal	1.92 ± 0.93	4.05 ± 1.00	8.80 ± 1.48	3.53 ± 1.56	2.09 ± 1.80	4.08 ± 2.79
	posterior	1.82 ± 0.67	1.54 ± 0.74	7.21 ± 0.30	5.52 ± 1.02	-0.85 ± 1.25	3.05 ± 3.26
	anterior	0.85 ± 0.40	3.06 ± 0.71	7.50 ± 1.05	4.41 ± 2.41	0.42 ± 2.28	3.25 ± 2.88
$\Delta\beta max$ [°]	normal	10.29 ± 1.89	10.85 ± 1.94	8.70 ± 1.44	7.05 ± 3.49	4.26 ± 2.46	8.23 ± 2.67
	posterior	11.45 ± 0.30	9.19 ± 1.20	6.63 ± 0.29	8.50 ± 2.13	4.08 ± 1.73	7.97 ± 2.78
	anterior	10.68 ± 2.00	10.08 ± 1.29	8.03 ± 1.32	6.32 ± 1.98	4.32 ± 3.19	7.89 ± 2.64
$\dot{\beta} max$ [°/s]	normal	151.68 ± 49.69	157.41 ± 51.66	138.18 ± 39.29	141.25 ± 82.46	73.17 ± 20.59	132.34 ± 33.98
	posterior	171.44 ± 24.45	122.13 ± 34.76	110.78 ± 21.42	148.40 ± 18.93	96.87 ± 10.36	129.92 ± 29.94
	anterior	168.15 ± 16.26	152.31 ± 23.82	133.97 ± 53.00	146.83 ± 29.74	85.44 ± 29.94	137.34 ± 31.50
$\Delta\rho max$ [°]	normal	4.91 ± 1.85	6.09 ± 1.07	4.24 ± 1.82	3.86 ± 0.50	4.97 ± 1.29	4.81 ± 0.85
	posterior	3.64 ± 0.77	4.93 ± 0.74	3.06 ± 2.51	3.21 ± 2.35	1.27 ± 0.22	3.22 ± 1.32
	anterior	4.08 ± 0.92	4.93 ± 0.46	3.91 ± 3.43	4.43 ± 2.22	3.99 ± 2.19	4.27 ± 0.42

Table 4-1: Values and standard deviation of the study variables. Positive values represent eversion, and internal tibial rotation; negative values denote inversion.

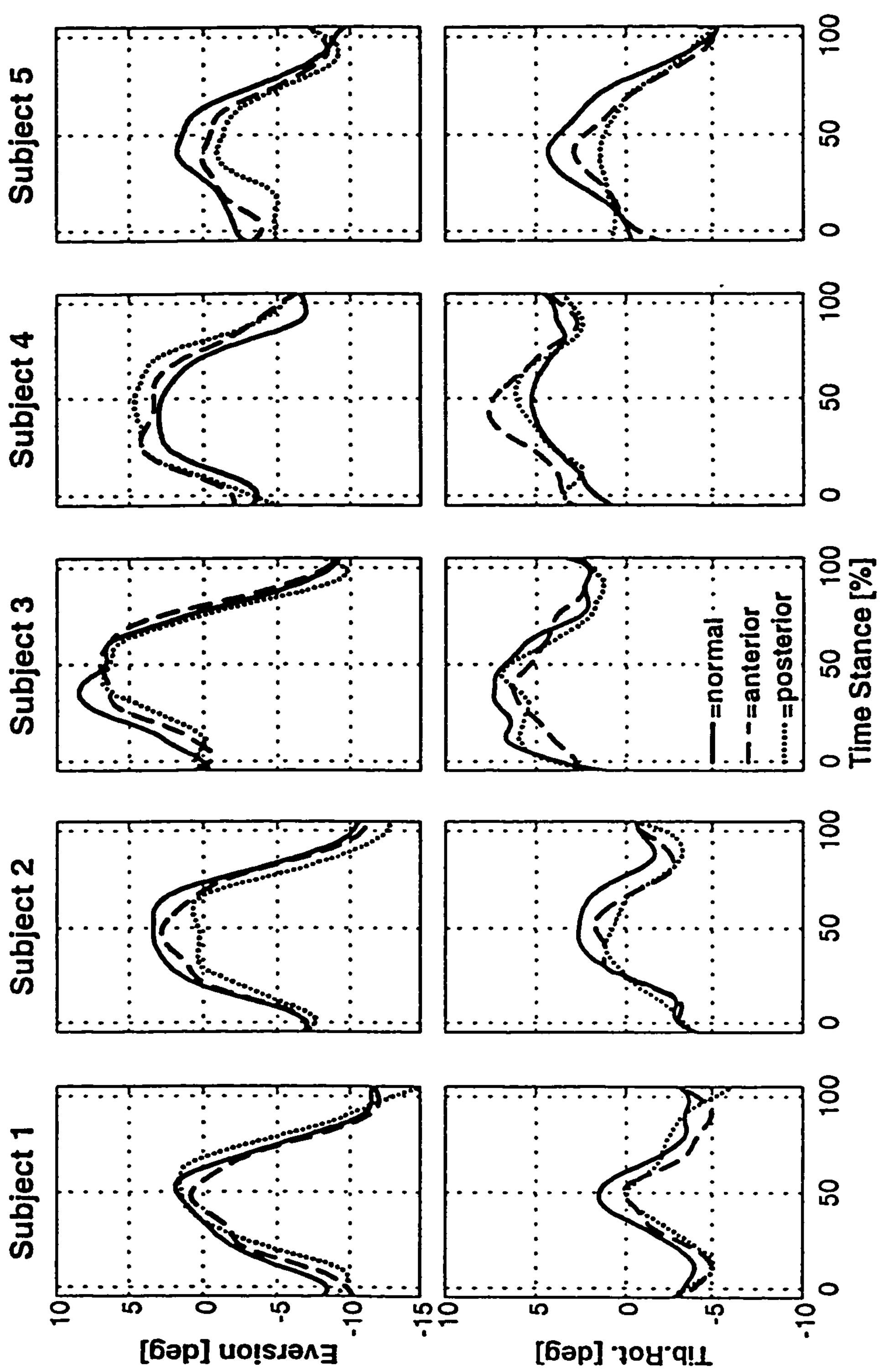


Figure 4-2: Mean curves of in/eversion and tibial rotation of all conditions and all subjects: (—) normal, (- -) anterior orthoses, (····) posterior orthoses. The standard deviation during the stance phase was on average ± 1.2° for eversion and ± 1.5° for tibial rotation.

BONE MARKER VARIABLES				
	normal-anterior-posterior		anterior-posterior	
Variable	1-way ANOVA	2-way ANOVA	1-way ANOVA	2-way ANOVA
β_o	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
β_{max}	n.s.	1 <.05 2 <.01 3 <.05	n.s.	1 n.s. 2 <.01 3 n.s.
$\Delta\beta_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\beta}_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\Delta\rho_{max}$	<.05	1 <.05 2 n.s. 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.

SHOE MARKER VARIABLES				
	normal-anterior-posterior		anterior-posterior	
Variable	1-way ANOVA	2-way ANOVA	1-way ANOVA	2-way ANOVA
$\beta_{o/shoe}$	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.05
$\beta_{max/shoe}$	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 <.05
$\Delta\beta_{max/shoe}$	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\beta}_{max/shoe}$	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 <.05 2 <.01 3 n.s.

Table 4-2: Study variable statistics (1 = difference between orthotic conditions; 2 = difference between subjects; 3 = interaction between orthoses and subjects).

showed consistently the lowest tibial rotation result with the posterior orthoses. Thus, hypothesis I was supported for total internal tibial rotation, but not for the eversion variables. Furthermore, no significant differences were found between the two orthotic conditions, thus hypothesis II could not be supported.

4.4 Discussion

Due to the invasive character of the study the number of subjects was limited to five, which did not allow an extensive statistical analysis. However, the general rotation patterns during running were consistent and generally found to be similar to previous investigations using external markers in running (Soutas-Little et al., 1987; Areblad et al., 1990; Nigg et al., 1993; Stergiou, 1996; McClay et al., 1997), bone markers in running (McClay, 1990) as well as bone markers in walking (Levens et al., 1948; Lafortune et al., 1994). Differences between those investigations and this study are discussed below, including the results of the test variables (Tables 4-1 and 4-2, Figure 4-2):

4.4.1 Inversion at touchdown (β_o)

All subjects consistently lowered their feet toward the ground in an inverted position, except for subject 3. The differences in orthotic conditions were small ranging from subject 2 (between 0.22° and 0.86°) to subject 5 (between 1.73° and 2.76°) whereas the differences between the subjects was up to 10° ($p<.01$). Hence, the degree of inversion appeared to be independent of the test condition, indicating that the orthotics did not significantly affect the tibiocalcaneal position prior to touchdown. Furthermore, there was no significant interaction between the orthotic conditions and the subjects. Generally, the range of touchdown inversion values was found to agree with previous investigations using shoe markers (Nigg et al., 1986; Soutas-Little, 1987; DeWit et al., 1995).

4.4.2 Variables of total movement ($\Delta\beta_{max}$, $\Delta\rho_{max}$)

The differences in total eversion ($\Delta\beta_{max}$) between the orthotic conditions for each subject were in the order of 1-2°, but the differences between the subjects were of the order of 6-7° ($p<.01$; Table 4-1). Whereas subjects 2 and 3 showed an expected decrease of eversion with anterior and/or posterior orthoses, subject 1 showed an increase and subjects 4 and 5 no consistent change, resulting in no significant differences between test conditions. These results suggest that there were no systematic differences resulting from the use of medial orthoses. Furthermore, no significant interactions between subjects and orthotic conditions were found. The results of this study are in agreement with previous investigations (Bates et al., 1979; Rogers et al., 1982; Nigg et al., 1986; Eng et al., 1994; Nigg et al., 1998) where small and insignificant decreases of eversion with orthoses of 1° to 4° were reported. However, other investigations (Clarke et al., 1984; Taunton et al., 1985) showed significant changes as a result of medially placed orthoses of 2° to 4° and of varus wedged shoe soles of 5° to 9° (van Woensel et al., 1992; Milani et al., 1995;

Perry et al., 1995). These conflicting results may be explained by differences in shape and material properties of the tested orthoses and varus wedged soles as well as methodological differences such as marker placements. All these previous studies were based on skin and shoe mounted marker settings which have been shown to overestimate the bone movements (Reinschmidt et al., 1997a). Hence, previous studies reporting on orthotic effects on eversion have to be interpreted with caution.

The differences in total internal tibial rotation ($\Delta\rho_{max}$) between the posterior orthotic condition and the normal shoe and between the subjects were small (between 1° to 4°; exact values: 0.75° and 3.7°) but significant ($p<.05$). All subjects showed a decrease in total internal tibial rotation with orthoses compared with the normal condition, with one exception (subject 4). Thus, it is concluded that posteriorly placed orthoses may significantly decrease total internal tibial rotation which is in contrast to previous studies using electrogoniometers (Taunton et al., 1985, no consistent orthotic effect; and Smart et al., 1985, reduction of 2°, insignificant) using external markers (Kozak et al., 1991, increase of 8%; Eng et al., 1994, no effect) and using bone pins in walking (Lafortune, 1994, reduction of 1° to 3°, not significant).

Summarizing the results on the two variables, total eversion and internal tibial rotation, it can be concluded that small decreasing effects were apparent in the order of 1° to 2° (eversion) and 1° to 4° (tibial rotation). The reduction in eversion was not systematic over all subjects and insignificant, in contrast to internal tibial rotation where the reduction was systematic (one exception: subject 4 anterior orthoses) and significant. However, the limitations of this study (only five subjects, limited repetitions, accuracy and repeatability) narrow the certainty to which a true result may be detected to about 1°. Thus, orthoses may produce small differences (i.e. tibial rotation) where it is currently unknown if they are associated with the occurrence or treatment of pain and injury, hence, whether or not they are relevant. Grau (1997) recently reported eversion differences between subjects with and without Achilles tendon pain of 1° to 2° only. This seems to indicate that relevant kinematic differences may in fact be very small or that the quantified variables are functionally or biologically not relevant.

4.4.3 Maximum eversion (β_{max})

It was expected that, with the support of shoes and orthoses, maximum eversion would be reduced. Maximum ankle eversion (β_{max}) was moderately reduced with both orthoses in subjects 1,2,3,5, but was increased in subject 4 compared with the normal shoe condition. The differences between the normal shoe condition and both orthoses were not significant in the one-way ANOVA ($p<.05$) showing significant interactions between orthoses and subjects ($p<.05$). Hence, maximum eversion showed a reduction with orthoses of about 1° to 3° (exception subject 4 with an increase of 1° to 2°) and hypothesis I could not be supported. Furthermore, the posterior orthoses was no more effective in decreasing maximum eversion than the anterior orthoses, thus hypothesis II was rejected.

4.4.4 Maximum eversion velocity ($\dot{\beta}_{max}$)

The differences of maximum eversion velocity between the subjects (smallest in subject 5 with 73-96°/s and largest in subject 1 with 152-171°/s) was again larger ($p<.01$) than the differences between the orthotic conditions (maximum of 25°/s for subject 2). It was expected that orthoses would decrease maximum eversion velocity, which was found in subjects 2 and 3. The other three subjects, however, showed an increased velocity, thus suggesting that there was no systematic orthotic effects on the maximum eversion velocity. As expected $\dot{\beta}_{max}$ measured at the bone level (between 73°/s and 171°/s, Table 4-1) was smaller compared with studies using shoe markers where eversion velocities have been reported between 408°/s and 532°/s (Clarke et al., 1984; Williams et al., 1991; van Woensel et al., 1992). Thus, considerable differences between skeletal velocities and velocities measured with shoe markers were observed which indicated a relative movement between the bone markers and the shoe markers.

4.4.5 Shoe eversion versus bone eversion

The comparison of total eversion measured at the shoe ($\Delta\beta_{max/shoe}$) with that at the bone (calcaneus, $\Delta\beta_{max/bone}$) is shown in Figure 4-3. It is evident that the differences between the subjects are larger than those between the shoe conditions. Therefore, not only was bone movement found to be typical for each subject but also the shoe movement, even though all five subjects used the same running shoe model and the same orthotics.

The shoe variables showed no significant differences based on the one-way ANOVA (Table 4-2). Significant interactions were found for the touchdown variable and the maximum shoe eversion. Although not significant, eversion velocity of the posterior orthotic was reduced in all five subjects compared to the anterior orthosis (subject 1: 361°/s posterior versus 393°/s anterior; subject 2: 292°/s versus 345°/s; subject 3: 196°/s versus 292°/s; subject 4: 178°/s versus 217°/s; subject 5: 130°/s versus 184°/s; mean SD ± 52°/s). This trend is in agreement with Nigg et al. (1986) who concluded that initial pronation (a variable related to eversion velocity) is reduced in posterior orthoses compared with anterior orthoses. It has to be considered, however, that these shoe results depend on the variability between the standing trials of each subject with each shoe condition. Thus, these shoe results have to be interpreted with caution.

Intraindividually, all five subjects (except subject 4) showed a significantly larger eversion of the shoe markers compared with the bone markers ($p<0.01$), indicating that the shoes moved relative to the underlying calcaneus (Figure 4-3). This relative movement (difference of $\Delta\beta_{max/shoe}$ - $\Delta\beta_{max/bone}$) was not found to be constant for the entire period of ground contact which is supported by Van Gheluwe et al. (1995), using skin markers viewed through windows cut into the shoes. The relative movement was smallest at touchdown (between 1°-3°) and largest at maximum eversion, where it was as large as 11-12° for subject 1, 6-10° for subjects 2 and 3, 3-6° for subject 5, but also as small as 0-3° for subject 4, (see Figure 4-3). Previously reported values of relative movements were 2-4° between skin markers and shoe markers (Clarke et al., 1984; Nigg et al., 1986; Stacoff et al., 1992; Van Gheluwe et al. 1995).

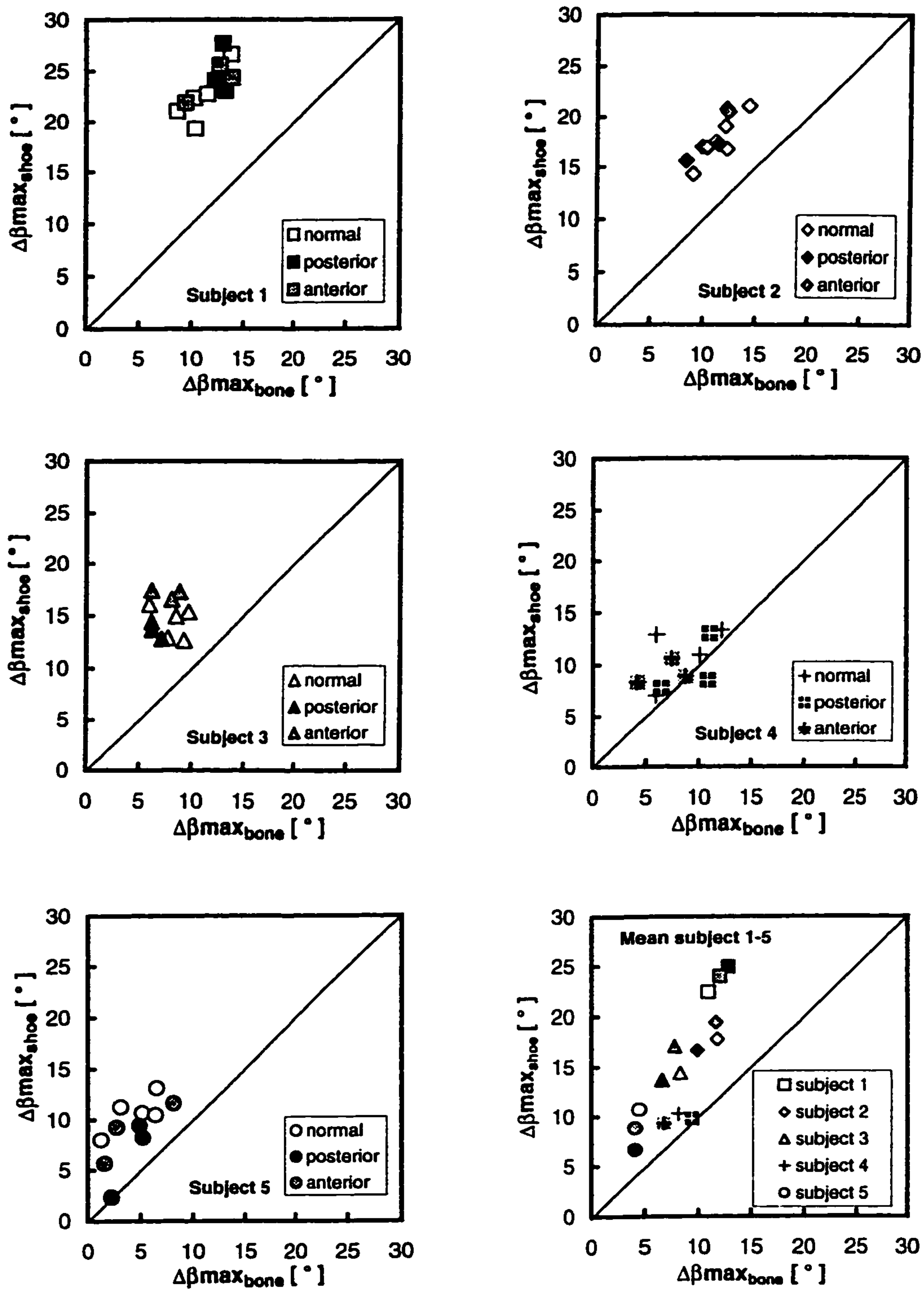


Figure 4-3: Total eversion of the shoe relative to the bone. Diagrams "Subject 1-5": For each trial of all subjects. Diagram "Mean subject 1-5": Mean values of each shoe condition of all subjects. (Note: Bone values may differ slightly from Table 4-1 because of different standing trial results from different shoes).

Maximum eversion velocity measured at the shoe varied between 250-500°/s for subjects 1 and 2 and 75-300°/s for subjects 3, 4 and 5, showing moderately smaller values compared with previous studies with shoe markers where maximum eversion velocity was between 408°/s and 532°/s for running speeds between 3 and 4 m/s (Clarke et al., 1984; Williams et al., 1991; van Woensel et al., 1992). The lower angular velocities of this study may be explained by the slower running speeds (2.5 to 3m/s).

Furthermore, in a small additional study it was found that the total shoe eversion might be 2-4° larger using the marker setting S1-S2-S3 compared with the setting S1-S4-S5 (Figure 3-2). It is likely that the anteriorly placed markers S2 and S3 include more midfoot motion compared with the more posteriorly placed markers S4 and S5. Thus, the relative movement between shoe and bone may depend on the shoe marker configuration. This may also explain discrepancies between values reported in different investigations.

In summary, the relative movement between the shoe and underlying bone may be larger than previously reported. It may consist of slipping inside the shoe, may depend on skin and fat pad movements but also may depend on shoe marker configuration. Presently it is unclear, how this relative movement might be related to running injuries and to fit and comfort inside a shoe.

4.5 Interpretation

The results of this study show that there were no substantial or systematic orthotic effects over all subjects and test variables. Only hypothesis I could be supported for the total internal tibial rotation variable. There are a number of reasons that may explain these results; (1) use of orthoses, (2) use of test shoes, (3) proprioceptive effects, and (4) individual responses. These aspects are discussed below.

4.5.1 Orthoses

The orthoses applied in this study were not individually constructed. Hence, possible anatomical differences between the test subjects were not accounted for, a factor which has been pointed out by Kozak et al., (1991). Posteriorly placed orthoses are believed to support the calcaneus beneath the sustentaculum tali hereby producing an inverting moment about the subtalar joint. However, whether this mechanical support works in all subjects is currently not known. In view of the present results it may be suggested that the same orthoses applied on different feet may produce different results.

4.5.2 Shoes

The cutout on the lateral heel counter of the tests shoes (Figure 3-4) was necessary to prevent impingement with the calcaneal bone pin. However, this cutout may have reduced heel counter rigidity and the fit of the heel inside the shoe, consequently influencing the results. Thus, a good heel counter fit may be needed for the orthoses to be effective. On the other hand, Van Gheluwe et al. (1995) concluded that a stiff heel counter increases shoe eversion, but not heel eversion (inside the shoe) which was regarded as independent of the rigidity of the heel counter. However, heel inversion may include skin movement artifacts. Therefore, the effects of varied heel counter rigidity on the bone movement of the rearfoot remain unknown for the time being.

Shoe markers mounted at the heel counter have been used in most previous investigations related to the kinematics of running. However, markers placed in other areas of the foot quantifying midfoot and/or forefoot movements have not been used systematically to date. Thus, it is possible that the calcaneus may not be the relevant bone to be assessed; and midfoot and forefoot movements may be more important to the understanding of orthotic effects.

4.5.3 Proprioception

The invasive character of this study required the application of local anesthesia at the bone pin insertion site. Reinschmidt et al. (1997a), using the same subjects and shoes in their investigation, studied the effect of bone pin insertion (and local anesthesia) on skin-marker-based tibiocalcaneal rotations in subjects 2 and 4 comparing three trials with and without bone pins. It was concluded that the pre/post-operative knee and ankle joint rotations showed graphs which were similar in shape and magnitude, maximum differences between the two conditions being 2°. However, it remains unknown if subjects would adapt their individual running patterns when using orthoses if the local anesthesia was not present.

Furthermore, the fact that successful injury treatment can be achieved with flexible, semirigid or rigid orthoses suggests that orthotic effects may be caused by proprioceptive reasons. It is possible that orthoses placed in the midfoot and forefoot may increase the afferent feedback from cutaneous receptors (Feuerbach et al., 1994) of the m. tibialis posterior, m. digitorum communis, and m. hallucis longus which may lead to decreased eversion because of muscular contraction of these inverting muscles. Indirect evidence from cadaver studies shows that when pulling forces are applied on m. tibialis posterior, eversion is reduced and the movements at the midfoot joints are changed (Müller et al., 1997; Stähelin et al., 1997). Further support is provided by Fromme and co-workers (1997) who found that with increasing fatigue pronation increased; suggesting that muscular activity may play an important role in the control of eversion during the stance phase of running.

Hence, orthotic effects may have to be looked at in a combination of a mechanical and a proprioceptive view. Furthermore, research is necessary to improve the understanding between proprioception and muscular effects during running. For that purpose kinematics may only be one of several necessary tools to study these effects.

4.5.4 Individual responses

The results of the variables used in this study show that the differences between the subjects were larger than the differences between the different test conditions. This suggests that each subject may have moved within his individual movement pattern despite the different orthotic conditions. Comparing individual subject data showed that only subject 2 had a significant decrease of maximum eversion ($p<.05$) and subject 5 a significant decrease of total internal tibial rotation ($p<.05$) as a result of medially placed orthoses. It is suggested that for the given running task there may be various solutions with respect to the magnitude of rotations between different segments of the lower extremity, an observation which is supported by the work of Engsberg et al. (1987) and Lafortune et al., (1994). Thus, a specific movement, such as running, may be associated with individual movement patterns such that a mechanical support, i.e. with medially placed orthoses, cannot change them substantially. Alternatively, even a small decrease of any rotation (as observed in this study) may reduce the risk of injury considerably. However, it is presently not established how large such a relevant difference might be.

4.6 Conclusion

In conclusion, this *in-vivo* study showed that medially placed foot orthoses did not substantially change tibiocalcaneal movement patterns during running. Orthotic effects on eversion and tibial rotations were found to be small and unsystematic over all subjects. Differences between the subjects were significantly larger (up to 10°) than between the orthotic conditions (1° to 4°). Significant orthotic effects across the subjects were found only for total internal tibial rotation ($p<.05$).

The results of this study suggest that (a) the effects of the tested foot orthoses on tibiocalcaneal movement patterns during running may be small, (b) the orthotic effects may be mechanical as well as proprioceptive, and that (c) the calcaneus may not the relevant bone to be assessed, i.e. mid-foot, fore-foot movements may be more important to the understanding of orthotic effects.

4.7 Summary

Previous studies using shoe and skin markers concluded that medially placed orthoses control/reduce foot eversion and tibial rotation. However, it is currently unknown if such orthoses also affect skeletal motion at the lower extremities. The purpose of this study was to quantify the effects of medially placed foot orthoses on skeletal movements of the calcaneus and tibia during the stance phase in running.

Intracortical Hofman pins with reflective marker triads were inserted under standard local anesthetic into the calcaneus and tibia of five healthy male subjects. The three-dimensional tibiocalcaneal rotations were determined using a joint coordinate system. Eversion (skeletal and shoe) and tibial rotation were calculated to study the orthotic effects of anterior and posterior orthoses compared with no orthotic. The results showed that orthotic effects on eversion and tibial rotations were small and unsystematic over all subjects. Differences between the subjects were significantly larger than between the orthotic conditions. Significant orthotic effects across subjects were found only for total internal tibial rotation ($p<.05$). This *in-vivo* study showed that medially placed foot orthoses do not change tibiocalcaneal movement patterns substantially which contrasts with previous studies using external markers to estimate bone movements.

The results of this study suggest that (a) the effects of the tested foot orthoses on skeletal movement patterns of the calcaneus and tibia during running may only be small and subject specific, (b) the orthotic effects may be mechanical as well as proprioceptive, thus, that the muscular control of eversion during the stance phase of running may play an important role and that (c) the calcaneus may not the relevant bone to be assessed, i.e. that mid-foot and fore-foot movements may be more important to the understanding of orthotic effects.

5. EFFECTS OF SHOE SOLE CONSTRUCTION ON SKELETAL MOTION DURING RUNNING

5.1 Introduction

The increasing number of runners and consequently of running injuries of the last decades has produced an interest in studying the effects of shoe sole constructions on the kinematics of running and their effects on the development of running injuries (Bates et al., 1978; Cavanagh, 1980 and 1990; Nigg et al., 1986; Frederick, 1986). Biomechanical factors which have been associated with the development of running injuries include excessive foot eversion and excessive tibial rotation (James et al., 1978; Segesser et al., 1980; Clement et al., 1981; Viitasalo and Kvist, 1983; van Mechelen, 1992). The effects of shoe sole modifications, specifically the change of sole geometry on the lateral side of the rearfoot, are thought to be important with respect to eversion and consequently with respect to running injuries (Clarke et al., 1983; Nigg et al., 1987a, 1987b; Stacoff et al., 1988; Hamill et al., 1992; DeWit et al., 1995). Cavanagh (1980) pointed out that early running shoes typically showed a prominent lateral heel flare producing excessive pronation/eversion that may be associated with running injuries.

When running with heel landing, the lateral aspect of the shoe sole touches the ground typically first with a touchdown angle of 5°-10°. It has been postulated that a prominent and hard heel flare would increase the lever about the subtalar joint, causing an increased initial eversion and/or maximum eversion velocity which has been confirmed experimentally (Nigg et al., 1987a, 1987b; Stacoff et al., 1988; DeWit et al. 1995). However, during midstance, kinematic effects of lateral heel flares have been reported to be small and dependent on the midsole hardness which lead to controversial results (Clarke et al. 1983 ; Nigg et al. 1987a; Hamill et al., 1992). It was suggested, that the discrepancies in the results might be because of differences in the methodologies used

(Nigg et al., 1987a). Thus, heel flare effects are substantial at touchdown but small for total eversion.

Foot movements are transferred to the tibia by a coupling mechanism (Hicks, 1953; Inman 1976; Lundberg, 1989; Nigg et al., 1993; Hintermann et al., 1994). Consequently, it has been suggested that excessive eversion may be transferred into excessive tibial rotation (Subotnick, 1977; Segesser et al., 1980; Clement et al., 1981; James 1990). Generally, tibial rotation depends on foot eversion, the vertical force, plantar/dorsiflexion, ligament integrity, and on muscle-tendon forces (Hintermann, 1994). Furthermore, shoe sole constructions may influence the movement of the foot and/or the orientation of the subtalar joint axis which may change the movement coupling in the ankle joint complex. This may affect tibial rotation resulting in an increase of loading at the knee. Therefore, to understand the effects of shoe sole modifications on internal loading one has to study movements of the calcaneus and tibia during running.

Previous studies of shoe sole effects during running are based on shoe or skin mounted marker settings. It has been shown recently, that externally mounted markers overestimate the movements of the underlying bone (Cappozzo et al., 1996, Reinschmidt et al., 1997a), but it can be assumed that there exists a relationship between shoe eversion and bone eversion. This relationship is currently unknown. The purpose of this study was to quantify the effects of shoe sole modifications on tibiocalcaneal eversion and tibial rotation using bone pins, and to compare eversion measured at the bone level with eversion determined from shoe mounted markers. The hypotheses to be tested in this study were:

I. Large lateral heel flares increase maximum eversion velocity and maximum internal tibial rotation velocity compared with systematically reduced heel flares.
II. Large lateral heel flares increase maximum eversion and maximum internal tibial rotation compared with systematically reduced heel flares.
III. An increase in shoe eversion variables (maximum eversion velocity and total eversion) is related to an increase in bone eversion variables.

5.2 Methods

5.2.1 General Project Description

The experiments were performed at the Department of Orthopaedics, Karolinska Institute at Huddinge University Hospital, Stockholm, where previous bone pin studies have been carried out (Lundberg, 1989; Karlsson and Lundberg, 1994). The project was part of a larger study (Reinschmidt, 1996, 1997a). Ethical approval for the experiments was obtained from the Ethics committee of the Karolinska Hospital and by the Medical Ethics Committee of The University of Calgary. The experimental set-up, testing procedure, data analyses and data reduction has been described previously in more detail (chapter 3; Reinschmidt, 1996).

Five healthy male volunteers participated in this study (28.6 ± 4.3 yrs., mass 83.4 ± 10.2 kg, height 185.1 ± 4.5 cm); they were all injury free at the time of the experiments and had no previous injury history which may have influenced their locomotion patterns. The subjects gave their informed consent to participate in the study. Intracortical Hofmann pins with reflective marker triads were inserted under standard local anesthetic (Citanest 10 mg/ml) which was active for 2-3 hours, leaving enough time for the experiments. Two bone pins were drilled into the posterior lateral aspect of calcaneus and the anterior lateral aspect of the tibial condyle and reflective marker triads were screwed onto each of these pins (Figure 3-2).

5.2.2 Experimental Set-up and Testing Procedure

Three high-speed cine cameras (LOCAM) were focused in umbrella form on a force platform (KISTLER) which was mounted flush with the runway. The camera speed was set at 200 Hz. Three LED's, triggered by a threshold detector connected to the force plate, were used to synchronize the cameras and to determine the time of contact on the force plate. The synchronization was possible within one frame, corresponding to 5ms in the worst case. A calibration frame with six control points (0.5 x 0.5 x 0.5 m^3) was used for the three-dimensional reconstruction. The subjects performed heel-toe running trials with a speed of between 2.5 and 3.0 m/s. Each of the test conditions was repeated three times and trials were repeated if the subjects did not land with their right foot on the force plate and/or if an obvious modification of the gait pattern occurred.

5.2.3 Test Shoes

The tests were performed with standard shoes (Adidas Equipment Cushioning) where the rearfoot geometry was systematically changed. The original sole was changed to a single density (Shore A45) midsole and was modified at the lateral side to a wide flare, a neutral flare or straight sole and a rounded sole (Figure 3-4). The outer sole was constructed with a hard material of Shore A 65. These shoe sole modifications were thought to produce different lever arms during the initial landing phase under the following conditions: Touchdown angle being between 5° and 10° (Clarke et al., 1984; Nigg et al., 1986; Stacoff et al., 1988; Edington et al., 1990), shoe and sole geometry based on the test shoes (Figure 3-4). The theoretical levers at 5° and 10° inversion from the point of application to an assumed subtalar joint center were determined to be 7 and 13mm for the round shoe sole, 18 and 27mm for the straight, and 27 and 35mm for the flared. The heel counter of all shoes had a lateral cutout to prevent impingement with the calcaneal bone pin during running.

5.2.4 Data Analysis and Reduction

KineMat, a set of programs written in MATLAB™, was adapted from Reinschmidt and van den Bogert (1997d) for the specific needs of this investigation which allowed the reconstruction of the three-dimensional position of the markers and the calculation of the relative segmental movements. The barefoot standing trial was used as the neutral position to define the segment-fixed coordinate systems of the calcaneus and tibia. For that purpose the subjects were instructed to stand with straight knees, the ankle in the neutral position of 90 degrees dorsiflexion and the feet aligned parallel to the force platform representing the laboratory coordinate system. The standing trials with the respective shoe conditions were used for the shoe marker analysis. The intersegmental rotations were calculated for the stance phase of all test conditions as Cardanic angles using a joint coordinate system (JCS) at the ankle joint complex. In/eversion was calculated with the following sequence of rotations: (1) plantar/dorsiflexion about a tibia fixed medio-lateral axis, (2) foot ab/adduction about the floating axis, and (3) in/eversion about the antero-posterior axis of the foot (after Cole et al., 1993). Tibial rotation (corresponding to ab/adduction in the above sequence) was calculated with a different sequence to avoid calculations about the floating axis having limited anatomical meaning: (1) tibial rotation about a tibia fixed proximal-distal (longitudinal) axis, (2) in/eversion about the floating axis, and (3) plantar/dorsiflexion about a calcaneus fixed medio-lateral axis (after Nigg et al., 1993).

The accuracy of the spatial reconstruction between two marker triads was determined (i) based on the residuals of the DLT equations averaged over the entire stance phase for all markers and (ii) based on the deviations of the inter-marker distances of the same trials. The mean error based on DLT residuals was found to be in the order of ±4° which included noise error and lens distortion error. The mean error based on marker distances (RMS) was found to be ±1.0° including noise error only. Thus, for the present study, a realistic estimation of the error was likely between the two errors given above. The error of the shoe data was about ±1.0° higher than that at the bone, because it included inaccuracies of different standing trials with different shoes.

5.2.5 Definitions of Variables

In/eversion and tibial rotation variable definitions are explained in Table 3-2 and depicted Figure 3-7. The variables were defined between touchdown and midstance of running. The inversion positions at touchdown (β_o and ρ_o) were considered to detect possible adaptations to shoe interventions before touchdown. Biomechanical factors, which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. It has been suggested that excessive eversion (i.e. maximum eversion (β_{max}) and total eversion ($\Delta\beta_{max}$)) forces the Achilles tendon to bend laterally, hereby producing an asymmetric stress distribution across the tendon, which could lead to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981; Denoth, 1986). Excessive eversion velocity ($\dot{\beta}_{max}$) produces eccentric loading of the muscles of the posterior tibial compartment, which control/reduce eversion after touchdown. Forces acting on muscle-tendon units during eccentric loading are increased compared to concentric loading. Excessive eversion velocity has been associated with overloading and injury of the muscles of the posterior tibial compartment, e.g. medial tibial stress syndrome (Segesser et al., 1980; Viitasalo et al., 1983; Messier et al., 1988; Stacoff et al., 1988; DeWit et al., 1995). Eversion velocity was defined in the window of 10 to 40% of ground contact time, excluding values before 10% because of possible inaccuracies, e.g. filtering effects and/or markers close to the edge of the defined three dimensional space. After 40%, eversion velocities had reached their maximum in all trials of the test. Excessive tibial rotation ($\Delta\rho_{max}$) has been associated with the changes in the tracking of the patella, hereby changing the contact pressure and possibly the friction of the articulating surface of the patella, which may be related to the occurrence of the patellafemoral pain syndrome (Stergiou, 1996). Tibial rotation is thought to take place as a result of the movement coupling from the calcaneus to the tibia. These variables indirectly describe the movement at those structures of interest, but do not directly describe the load within these structures. However, they are relatively easy to quantify. In addition to these variables, eversion of the shoe relative to the tibia was also determined, with the standing trial of each shoe condition being used for the definition of the neutral position for this purpose. The shoe variables were determined to compare the results of this investigation with previous studies using external markers and to quantify the relative movement of the shoe and the calcaneus caused by slipping of the heel inside the shoe.

The testing procedure (5.2.2) was organized such that test conditions were independent from each other. The residuals of the test variables were inspected for normal

distribution. All variables were tested with two-tailed ANOVA techniques with repeated measures, the one-way ANOVA to test subject independent orthotic and shoe sole effects, the two-way ANOVA to test subject dependent effects, as well as possible interactions between the subjects and the orthotic and sole conditions. In cases of contradicting results between the one-way and two-way ANOVA, the more conservative result of the one-way ANOVA was accepted. The power analysis conducted on the kinematic variables suggested that there was a 80% chance of detecting any differences in these variables between the test conditions which were greater than 3.5°.

5.3 Results

The general patterns of eversion and tibial rotation are presented in Figure 5-1 (single curves of a typical subject) and Figure 5-2 (mean curves of each condition for each subject). In all subjects eversion and internal tibial rotation took place from touchdown until midstance, and inversion and external tibial rotation from midstance to take-off.

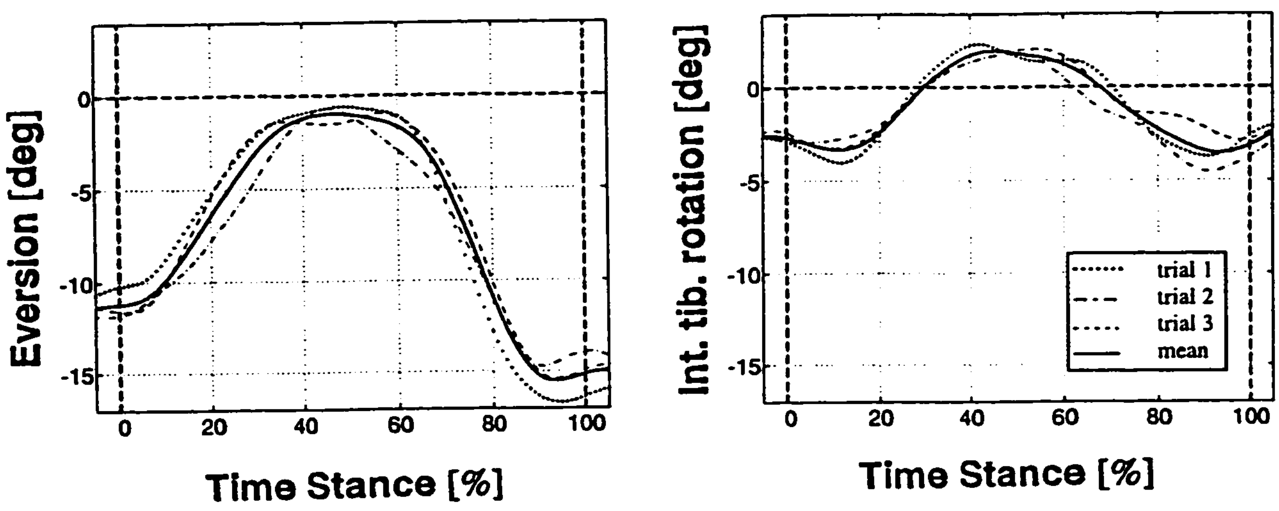


Figure 5-1: Example of in/eversion and tibial rotation (subject 1 with the straight shoe sole). Thin lines: three repetitions, thick lines: mean curve. Labels on the vertical axes indicate movements in the positive direction.

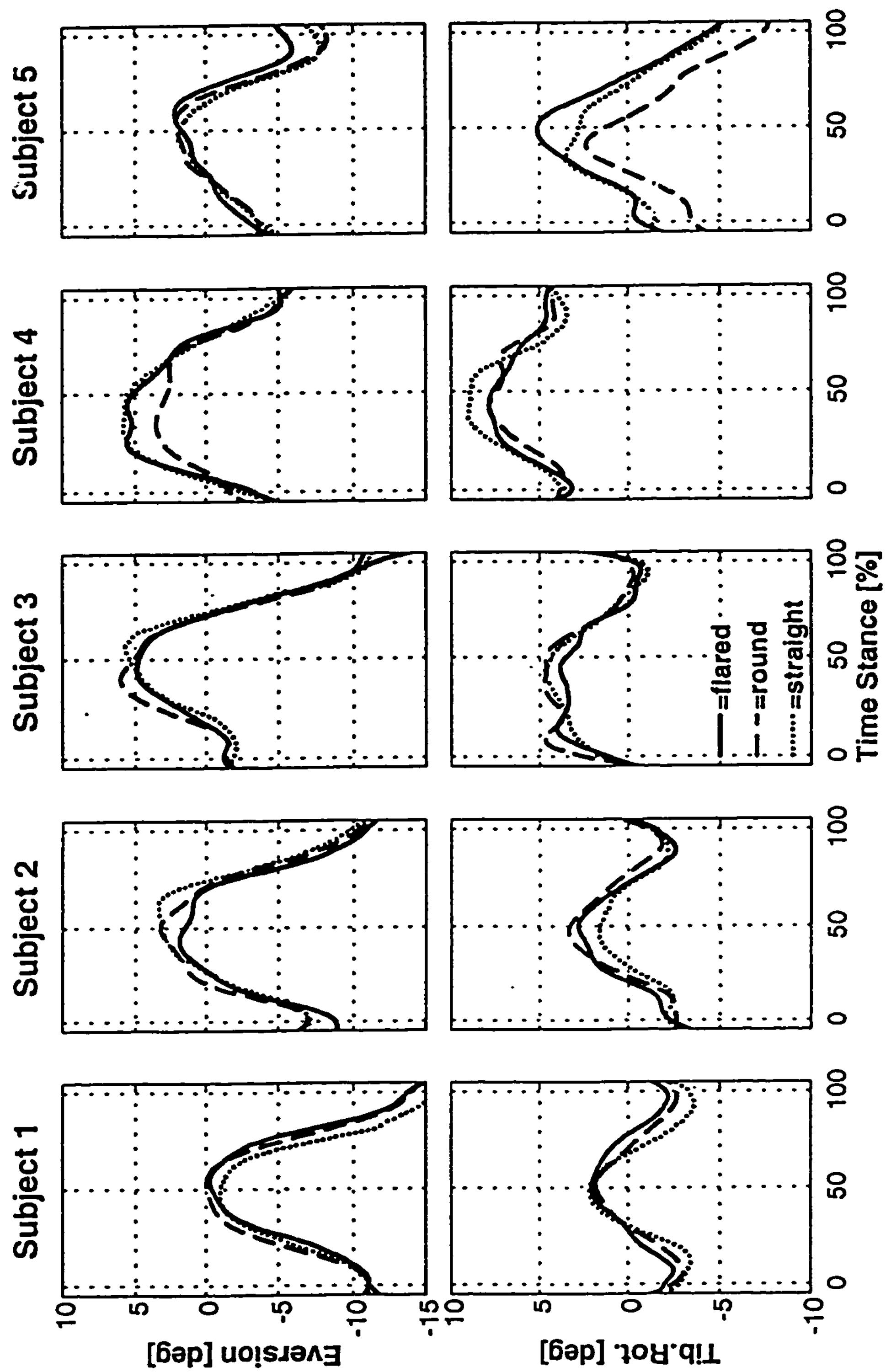


Figure 5-2: Mean curves of in/eversion and tibial rotation of all conditions and all subjects: (···) straight, (—) flared, (- -) round shoe sole. The standard deviation during the stance phase was on average ±1.1° for eversion and tibial rotation.

Variable	condition	subject 1	subject 2	subject 3	subject 4	subject 5	mean SD
β_0 [°]	straight	-11.22 ± 0.87	-7.07 ± 1.08	-1.97 ± 1.66	-2.17 ± 1.72	-4.01 ± 1.21	-6.29 ± 3.90
	flared	-11.05 ± 2.51	-8.72 ± 1.66	-1.28 ± 1.48	-3.29 ± 0.40	-3.29 ± 1.40	-5.53 ± 4.15
	round	-11.15 ± 1.45	-6.93 ± 0.59	-1.50 ± 0.43	-1.71 ± 0.71	-3.82 ± 0.48	-5.02 ± 4.08
β_{max} [°]	straight	-0.77 ± 0.20	4.02 ± 0.91	5.87 ± 0.82	6.45 ± 0.52	1.97 ± 1.29	3.51 ± 2.96
	flared	-0.09 ± 0.42	2.46 ± 1.94	5.25 ± 1.97	6.56 ± 0.44	2.32 ± 0.51	3.30 ± 2.63
	round	0.23 ± 1.09	3.38 ± 1.79	6.39 ± 0.06	3.60 ± 1.26	2.59 ± 2.22	3.24 ± 2.21
$\Delta\beta_{max}$ [°]	straight	10.45 ± 0.76	11.09 ± 1.09	7.84 ± 1.34	8.62 ± 2.16	5.98 ± 2.51	8.80 ± 2.05
	flared	10.96 ± 2.73	11.18 ± 3.05	6.52 ± 1.17	9.86 ± 0.28	5.61 ± 1.89	8.83 ± 2.59
	round	11.37 ± 2.41	10.31 ± 1.43	7.89 ± 0.50	5.31 ± 0.59	6.41 ± 2.70	8.26 ± 2.56
$\dot{\beta}_{max}$ [°/s]	straight	144.47 ± 22.67	138.25 ± 21.23	125.10 ± 33.77	170.95 ± 94.39	80.84 ± 36.97	131.92 ± 33.08
	flared	172.49 ± 44.10	160.66 ± 54.11	108.64 ± 17.36	211.97 ± 50.70	67.72 ± 24.54	144.30 ± 58.62
	round	191.47 ± 41.80	162.55 ± 15.16	149.43 ± 22.42	113.44 ± 30.11	90.06 ± 3.42	141.39 ± 40.11
$\Delta\rho_{max}$ [°]	straight	4.86 ± 0.38	3.88 ± 0.10	3.29 ± 0.92	5.50 ± 0.75	4.95 ± 1.50	4.50 ± 0.89
	flared	4.02 ± 1.08	5.26 ± 1.55	3.46 ± 1.46	4.87 ± 1.00	5.93 ± 0.70	4.71 ± 0.98
	round	4.29 ± 0.74	6.19 ± 1.75	2.76 ± 0.53	4.28 ± 2.13	5.73 ± 1.03	4.65 ± 1.36
$\dot{\rho}_{max}$ [°/s]	straight	94.69 ± 24.88	78.31 ± 20.97	53.98 ± 22.19	86.26 ± 22.48	107.20 ± 30.21	84.09 ± 19.95
	flared	87.38 ± 22.64	82.14 ± 24.73	45.78 ± 24.19	98.21 ± 11.73	85.32 ± 12.38	79.77 ± 19.93
	round	68.72 ± 22.41	84.25 ± 25.09	44.27 ± 3.35	80.52 ± 21.88	84.51 ± 12.46	72.46 ± 17.01

Table 5-1: The results and SD of the study variables based on bone pin data. Positive values represent eversion, and internal tibial rotation; negative values denote inversion.

Variable	condition	subject 1	subject 2	subject 3	subject 4	subject 5	mean SD
β_0 [°]	straight	-2.55 ± 1.37	-6.03 ± 1.56	-3.99 ± 1.07	0.84 ± 2.77	-6.56 ± 1.74	-3.66 ± 2.98
	flared	-3.56 ± 4.84	-7.47 ± 0.72	-2.75 ± 1.55	-1.10 ± 1.44	-3.20 ± 2.68	-3.62 ± 2.35
	round	-3.75 ± 1.10	-6.22 ± 1.18	-0.76 ± 1.04	2.81 ± 1.49	-2.04 ± 0.95	-1.99 ± 3.37
β_{max} [°]	straight	20.26 ± 0.70	12.59 ± 0.84	9.33 ± 1.39	15.80 ± 1.69	5.71 ± 1.14	12.74 ± 5.63
	flared	17.56 ± 1.00	12.67 ± 1.61	10.42 ± 2.64	14.87 ± 1.39	8.41 ± 1.06	12.79 ± 3.61
	round	22.37 ± 1.09	13.98 ± 0.71	17.02 ± 0.48	16.73 ± 1.30	9.90 ± 1.66	16.00 ± 4.57
$\Delta\beta_{max}$ [°]	straight	22.81 ± 2.06	18.62 ± 1.02	13.33 ± 2.43	14.96 ± 4.42	12.27 ± 1.55	16.40 ± 4.32
	flared	21.12 ± 5.76	20.14 ± 1.82	13.17 ± 1.96	15.97 ± 1.38	11.61 ± 2.77	16.40 ± 4.18
	round	26.11 ± 2.00	20.20 ± 1.47	17.77 ± 0.78	13.91 ± 0.31	11.94 ± 1.18	17.99 ± 5.67
$\dot{\beta}_{max}$ [°/s]	straight	321.12 ± 16.92	291.81 ± 61.15	226.84 ± 90.20	274.85 ± 165.32	218.10 ± 21.43	266.54 ± 45.62
	flared	328.26 ± 49.11	393.07 ± 81.33	195.43 ± 76.10	318.46 ± 92.01	157.99 ± 48.67	278.64 ± 98.26
	round	454.92 ± 39.82	371.75 ± 77.39	286.75 ± 56.62	302.49 ± 59.73	191.91 ± 47.38	321.56 ± 98.35

Table 5-2: The results and SD of the study variables based on shoe mounted markers. Positive values represent eversion, and internal tibial rotation; negative values denote inversion.

5.3.1 Inversion at touchdown (β_o)

Results based on bone pin markers. All subjects consistently lowered their feet towards the ground in an inverted position. The differences between the shoe conditions (straight, flared, round) within each subject were small (ranging between 0.17° in subject 1 and 1.79° in subject 2, Table 5-1) and statistically not significant (Table 5-3). The differences between the subjects were as large as 10°.
Results based on shoe markers. Shoe sole modifications showed no significant and no systematic differences in touchdown inversion (Table 5-3). Subjects 2 and 4 showed the largest inversion with the flared sole, subject 3 and 5 with the straight sole, and subject 1 with the round shoe (Table 5-2). The smallest inversion was found with the round sole in subjects 3, 4 and 5 and with the straight sole in subjects 1 and 2. Thus, there was no consistent pattern of shoe inversion at touchdown across the five subjects.

5.3.2 Variables of maximum velocity ($\dot{\beta}_{max}, \dot{\rho}_{max}$)

Results based on bone pin markers. The differences in maximum eversion velocity ($\dot{\beta}_{max}$) between the heel flare modifications (between 23°/s in subject 5 and 98°/s in subject 4) were smaller than the differences between the subjects (smallest in subject 5 with 68°/s to 90°/s and largest in subject 1 with 144°/s to 191°/s, Table 5-1). The flared shoe showed enhanced eversion velocity in subjects 1, 2 and 4, but not in subjects 3 and 5, compared to the straight shoe condition. The round shoe showed a reduced eversion velocity only in subject 4, but an increased velocity in all other subjects, compared to the straight shoe condition. Hence, measured on the bone level, there were no systematic effects on maximum eversion velocity because of the heel flare modifications.

The differences in maximum internal tibial rotation velocity ($\dot{\rho}_{max}$; between 6 °/s in subject 2 and 26 °/s in subject 1) were smaller than the differences between the subjects (smallest in subject 3 with 44°/s to 54 °/s and largest in subject 5 with 85 °/s to 107°/s). The flared shoe showed enhanced internal tibial velocity in subjects 2 and 4, but not in subjects 1, 3 and 5 compared to the straight shoe condition, and the round sole was found with reduced internal tibial velocity in all subjects except subject 2. None of these comparisons were statistically significant (Table 5-3).

Results based on shoe markers. The flared shoe was found with an increased maximum eversion velocity in subject 2 and 4 only (Table 5-2). Subjects 1 and 3 showed the largest velocity with the round shoe and subject 5 with the straight shoe. Thus, shoe modification effects on maximum eversion velocity were unsystematic across the five subjects. The comparison of the maximum eversion velocity measured at the shoe ($\dot{\beta}_{max/shoe}$) with that at the bone ($\dot{\beta}_{max/bone}$) is shown in Figure 5-3. The maximum eversion velocity of the shoe varied between 160-320°/s (subjects 3, 4, 5) and 300-450°/s (subjects 1, 2).

BONE MARKER VARIABLES		
	flared-straight-round	
Variable	1-way ANOVA	2-way ANOVA
β_o	n.s.	1 n.s. 2 <.01 3 n.s.
β_{max}	n.s.	1 n.s. 2 <.01 3 n.s.
$\Delta\beta_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\beta}_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.
$\Delta\rho_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\rho}_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.

SHOE MARKER VARIABLES		
	flared-straight-round	
Variable	1-way ANOVA	2-way ANOVA
β_o	n.s.	1 <.05 2 <.01 3 n.s.
β_{max}	n.s.	1 <.01 2 <.01 3 <.01
$\Delta\beta_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\beta}_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.

SHOE MARKER VS. BONE MARKER VARIABLES		
	flared-straight-round	
Variable	1-way ANOVA	2-way ANOVA
β_o	<.01	1 <.01 2 <.01 3 n.s.
β_{max}	<.01	1 <.01 2 <.01 3 <.01
$\Delta\beta_{max}$	<.01	1 <.01 2 <.01 3 <.01
$\dot{\beta}_{max}$	<.01	1 <.01 2 <.01 3 n.s.

Table 5-3: Study variable statistics. The following abbreviations are used: 1=difference between shoe sole conditions; 2=difference between subjects; 3=interaction between shoe soles and subjects; n.s.= not significant.

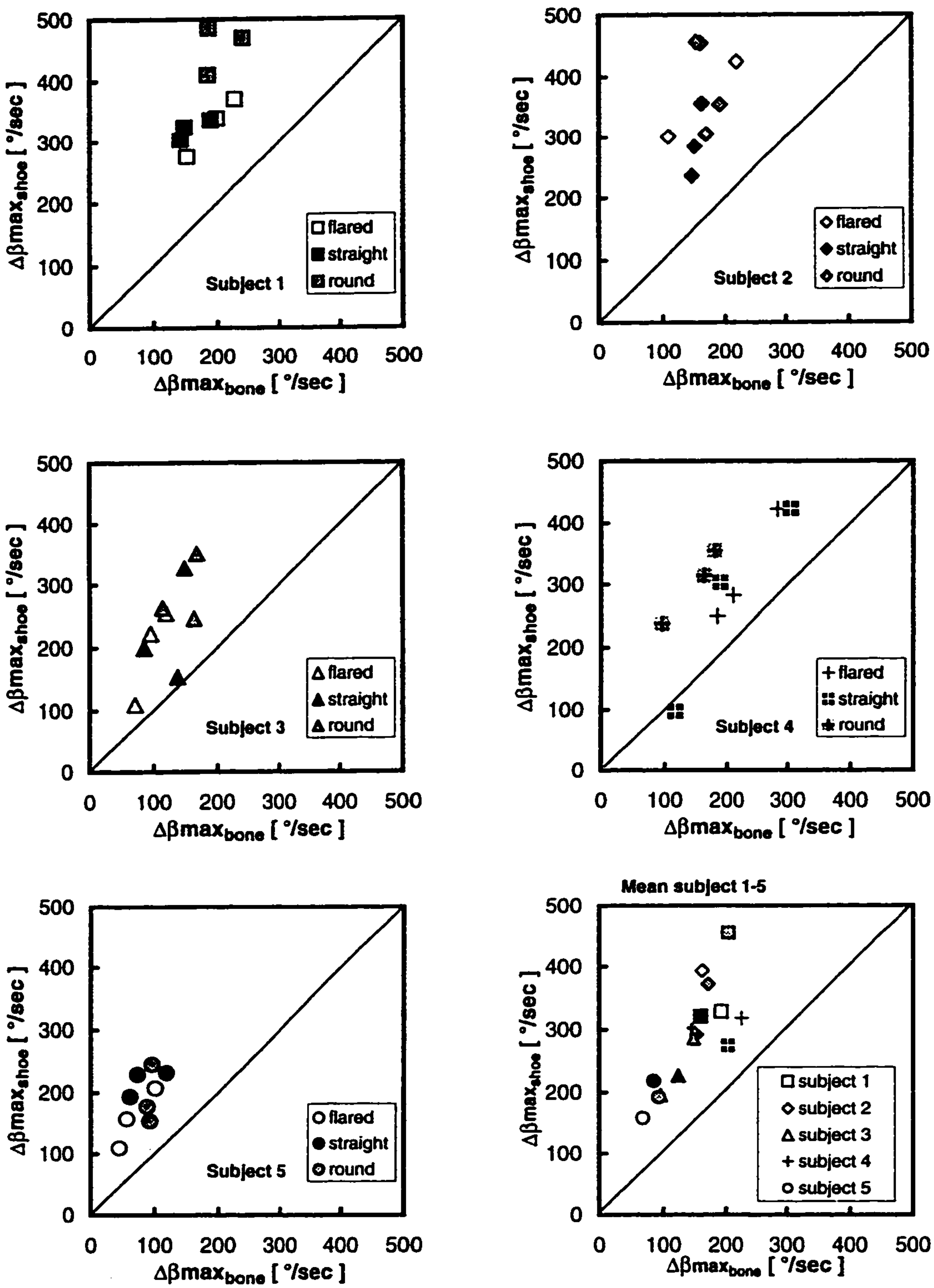


Figure 5-3: Maximum eversion velocity of the shoe relative to the bone. Diagrams "Subject 1-5" show each trial of all subjects. Diagram "Mean subject 1-5" shows the mean values of each shoe condition of all subjects. (Note: Bone values may differ slightly from Table 5-1 because of different standing trial results from different shoes.)

The shoe eversion velocity was about twice as large as that at the bone (varying between 70-150°/s in subjects 3, 5 and 150-225°/s in subjects 1, 2, 4). Increased shoe eversion velocities correlated with increased bone eversion velocities ($r=0.79$; Figure 5-3: mean subject 1-5) showing a good relationship between internal (bone) and external (shoe) movements.

5.3.3 Variables of maximum (β_{max}) and of total movement ($\Delta\beta_{max}$, $\Delta\rho_{max}$)

Results based on bone pin markers. The differences in maximum (β_{max}) and total eversion ($\Delta\beta_{max}$) between shoe sole modifications were in the order of 1-3°, but the differences between the subjects were up to 7° (Table 5-1). The flared shoes showed an increased total and maximal eversion in subject 1 and 4, a decreased eversion in subject 3, and an inconsistent result in subjects 2 and 5. Total and maximal eversion was decreased with round soles in subjects 2 and 4, but increased in all other subjects. Hence, there were no systematic shoe sole effects with respect to maximum and total eversion on the bone level.

Total internal tibial rotations ($\Delta\rho_{max}$) between shoe sole modifications were in the order of 1° to 2° and the differences between the subjects were in the order of 0° to 3.5° ($p<.01$). Thus, there were no systematic shoe sole effects with respect to total internal tibial rotation.

Results based on shoe markers. Although significant shoe modification effects were found on shoe eversion ($p<0.01$, Table 5-3), the results were not as expected (Table 5-2). All subjects had the largest maximum eversion with the round shoe and the smallest maximum eversion with either the flared sole (subjects 1 and 4) or the straight sole (subjects 2,3 and 5). Furthermore, interactions between the shoes and the subjects were found to be significant. Thus, the main effect was unsystematic across the five subjects.

The comparison of total eversion measured at the shoe ($\Delta\beta_{max/shoe}$) with that at the bone ($\Delta\beta_{max/bone}$) is shown in Figure 5-4. Eversion at the shoe level varied between 11° and 15° (subjects 3, 4, 5) and between 18° and 26° (subjects 1 and 2). Shoe eversion was

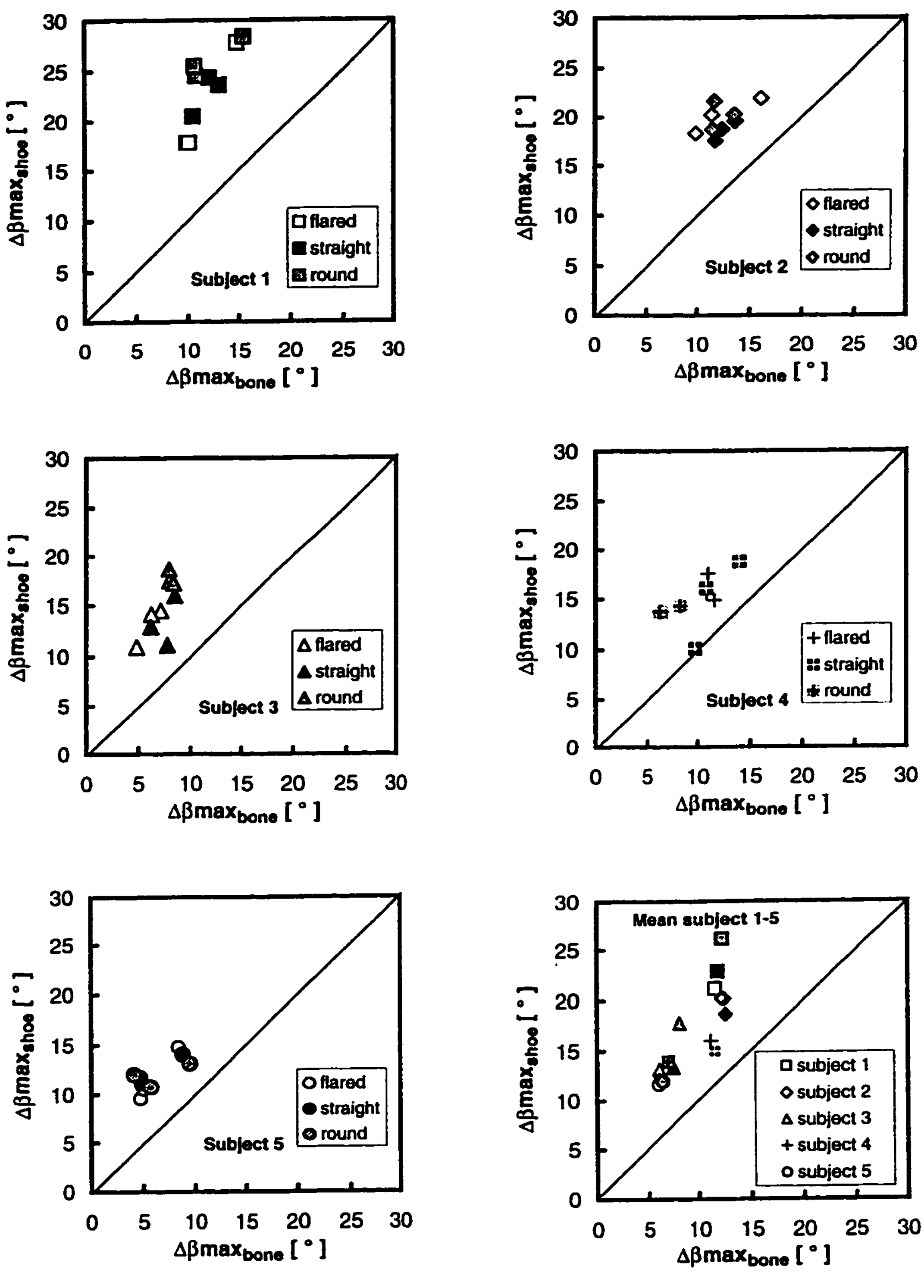


Figure 5-4: Total eversion of the shoe relative to the bone. Diagrams "Subject 1-5" show each trial of all subjects. Diagram "Mean subject 1-5" shows the mean values of each shoe condition of all subjects. (Note: Bone values may differ slightly from Table 5-1 because of different standing trial results from different shoes.)

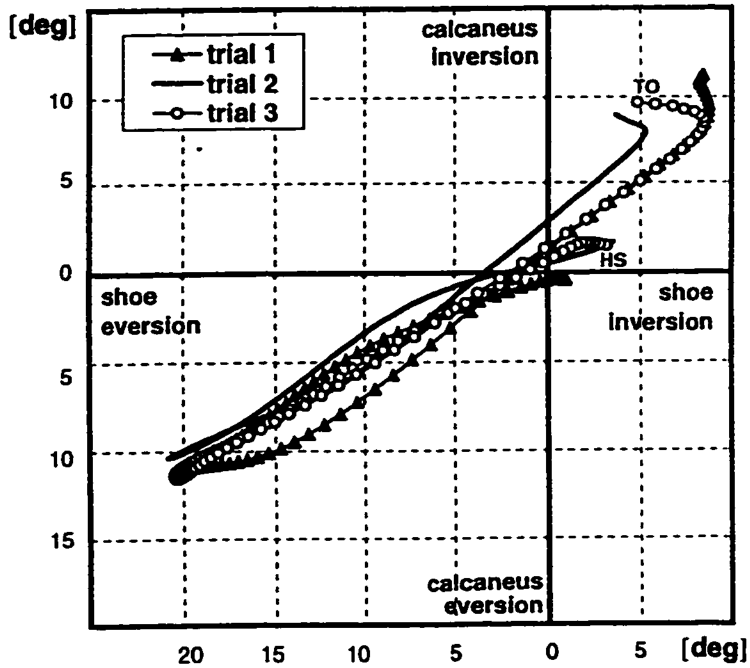


Figure 5-5: Shoe eversion as a function of calcaneus eversion. Presented are three trials of subject 1 with the straight shoe. HS = heel strike; TO = takeoff.

found to be about twice as large as bone eversion (r=0.88; Figure 5-4: mean subject 1-5), total eversion at the shoe was related to total eversion at the bone.

Generally, the shoe and bone eversion movement patterns differed between touchdown to take-off. Immediately after heel strike the shoe moved considerably more than the bone, then towards midstance the shoe and bone moved together until shortly before take-off (Figure 5-5).

5.4 Discussion

The results of this study showed that the shoe sole modifications did not produce the expected systematic effects on the test variables. Thus, hypothesis I and II of this study were not supported. Evidence was found that a relationship or coupling effect between shoe and bone eversion occurred during running. Thus, hypothesis III was accepted.

Large lateral heel flares were found neither to increase eversion velocity nor internal tibial rotation velocity compared to reduced heel flares. Similarly, neither maximum nor total eversion and tibial rotation were increased with large heel flares. It is possible that these results (rejection of hypothesis I and II) were influenced by several factors which could have played a role during testing. These factors are discussed below.

5.4.1 Factors possibly influencing the results

The test shoes had a cutout in the lateral heel counter (Figure 3-4) which was necessary to prevent impingement with the calcaneal bone pin. This cutout may have reduced heel counter rigidity and the fit of the heel inside the shoe, consequently reducing the coupling effect between the shoe and the foot. On the other hand, van Gheluwe et al. (1995) provided evidence that heel eversion is independent of the rigidity of the heel counter. Thus, whether or not heel counter rigidity or lateral cutouts had a systematic effect during testing cannot be answered conclusively.

It is possible that shoe eversion results may depend on the shoe marker configuration. In a small additional study, on one subject, the anterior marker configuration S1-S2-S3 was compared with the posterior configuration S1-S4-S5 (Figure 3-2). The results showed that total shoe eversion with the anterior configuration was 2-4° larger than with the posterior configuration suggesting that anteriorly placed markers are likely to include midfoot rotations compared with posteriorly placed markers. Thus, the present shoe eversion results may have be influenced by midfoot rotations.

The application of local anesthesia at the insertion site of the bone pins may be one further factor. Anesthesia was necessary because of the invasive character of the study, but it may have changed the proprioceptive feedback and consequently, possible adaptations and modulations of the movement pattern to different shoe conditions may not have taken place. In order to test this, Reinschmidt et al. (1997a), using the same subjects at the same test date compared three trials with and without bone pins in subjects

2 and 4. It was concluded that pre/post-operative knee and ankle joint rotations showed graphs which were similar in shape and magnitude, maximum differences being 2°. Thus, it is unlikely that the local anesthesia had a substantial effect on the results and it remains speculative whether the subjects would have adapted their individual running patterns towards the test shoes if the local anesthesia was not present.

5.4.2 Eversion and tibial rotation

The above discussion shows that there is no strong evidence that the test shoes and the local anesthesia dramatically influenced the kinematics during testing. Therefore, the findings of this study are discussed under the assumption, that the present results are real (and would be repeated with another set of shoes and subjects).

Generally, the movement patterns of this study (Figure 5-1 and 5-2) were found to be similar to previous investigations using external markers in running (Soutas-Little et al., 1987; Areblad et al., 1990; Nigg et al., 1993; Stergiou, 1996; McClay et al., 1997), using bone markers in running (McClay, 1990), as well as bone markers in walking (Levens et al., 1948; Lafortune et al., 1994).

It was expected that the large lever of the flared shoe sole would produce the largest eversion velocity and that the decreased lever of the round sole the smallest. This was not the case and is in contrast to previous investigations using shoe markers where it was concluded that prominent lateral heel flares cause an increased initial eversion or eversion velocity (Clarke et al., 1983; Nigg et al., 1987a, 1987b; DeWit et al., 1995). Generally, shoe eversion velocities of this study were found comparable to previous studies using shoe markers (Clarke et al., 1984: 532°/s; Williams et al., 1991: 475°/s; van Woensel et al., 1992: 408°/s) considering the slightly faster running speed of these studies (3-4m/s) compared to the present one (2.5-3m/s).

The results on total internal tibial rotations were found consistent with those reported by Lafortune et al. (1994) who found no significant differences in internal tibiofemoral rotation between normal and varus wedged shoes. Total internal tibial rotation of this study was found to be smaller than previously reported by McClay et al., 1997 (8.9° to 11.1°) using external markers that may have overestimated tibial rotation. The results are further supported by a bone pin study on walking, where total internal rotation was reported to be between 6° and 8° (Levens et al., 1948).

It was expected that the flare would increase maximum shoe eversion and the round shoe sole would decrease maximum shoe eversion. However, all subjects had the largest maximum eversion with the round shoe and the smallest maximum eversion with either the flared sole or the straight sole. One possible explanation is that the hard outer sole deformed very little at touchdown which may have favoured a rolling action of the round shoe resulting in a large maximum eversion.

Previously reported differences in maximum eversion between shoe and skin were between 2° and 4° (Clarke et al., 1984; Nigg et al., 1986; Stacoff et al., 1992; van Gheluwe et al., 1995) which is considerably smaller than the differences found in this study (between 5° and 20°, Table 5-1 and 5-2). One possible explanation for this discrepancy may be the differences in the methodologies used. The previous studies used shoe and skin markers and a two-dimensional approach, whereas the present study used a three-dimensional approach and shoe eversion was calculated relative to the tibia using bone markers. The observed relative movement between the bone and shoe is suggested to consists of slipping inside the shoe but also of deformations of the fat pad and shoe materials.

5.4.3 Interpretation

The results of the present study were not as expected but open the possibility for interpretations. One possibility being, that running is a movement pattern which is difficult to alter. More specifically, locomotion is thought to be controlled by a central pattern generator (Zernicke and Smith, 1996). During running a basic locomotor-like pattern is modulated by input from supraspinal centers and motion related feedback (Zernicke and Smith, 1996). One may therefore argue that a running pattern is predetermined and that muscular activity during running is used to adapt to shoe modifications. Although muscle activation (i.e. EMG) was not measured in this study, muscular activity as a response to shoe modifications may have been present during testing. This possible explanation is supported by the following arguments:

- A number of authors suggest that for a given task, there may be various solutions with respect to the rotations between different segments of the lower extremity (Engsberg et al., 1987; Lundberg et al., 1989; Lafortune et al., 1994). Thus, a specific movement, such as running, may be associated with individual movement patterns such that an external input (i.e. shoe sole modifications) may have only small and varying effects on the kinematics of the calcaneus and tibia.
- The large shoe eversion of the round shoe were not reflected by the bone eversion results. Thus, the increased shoe eversion (induced by the round sole construction) may have been compensated by muscular activation such that the kinematics at the bone level remained unchanged.
- In an additional study (with the same test subjects) the results of a normal shoe were found to be very similar to those of the straight shoe (chapter 6). The normal shoe was constructed with softer material on the lateral side (Shore A35). This indicates that the subjects may have altered their muscular activity not only to shoes with different sole geometry (as in present study), but also to shoes with different midsole hardnesses.

5.5 Summary

The purpose of this study was to quantify effects of shoe sole modification on skeletal kinematics of the calcaneus and tibia during the stance phase of running. Intracortical bone pins with reflective marker triads were inserted under standard local anesthesia into the calcaneus and tibia of five healthy male subjects. The three-dimensional tibiocalcaneal rotations were determined using a joint coordinate system approach. Three shoe sole modifications were tested with different sole geometry: a lateral heel flare of 25° (flared), no flare 0° (straight), and a rounded sole. The results

showed that shoe sole modifications did not change tibiocalcaneal rotations substantially. The following conclusions can be drawn from the results of this study:

- Mean shoe sole effects were found to be less than 1° which was smaller than the differences between the subjects (up to 7°; $p<.01$). Thus, on the bone level, shoe sole effects on tibiocalaneal movements may be small and unsystematic.
- Total shoe eversion and shoe eversion velocity were found to be approximately twice as large as bone eversion, the correlation being $r=0.88$ (between total shoe and bone eversion) and $r=.0.79$ (between maximum shoe and bone eversion velocity). This suggests that there may be a relationship or coupling effect between the shoes and bone. This relationship is possibly influenced by the shoes and the configuration of the markers attached to the shoe.
- Simultaneously measured shoe markers showed no systematic shoe sole effects on shoe eversion, which is in contrast to previous studies. It can be argued that if systematic shoe sole effects were present at the shoe level, then bone level effects could be expected. However, since the present results do not support this argument, it is possible that local anesthesia, individual muscular responses and/or the test shoe construction influenced the calcaneus and tibia kinematics during running.

6. TIBIOCALCANEAL KINEMATICS OF BAREFOOT VERSUS SHOD RUNNING

6.1 Introduction

Lower extremity movements of excessive eversion and tibial rotation have been associated with various running injuries (Clement et al., 1981; James et al., 1990; Segesser et al., 1980; Viitasalo and Kvist, 1983; van Mechelen, 1992). A coupling mechanism linking in/eversion and tibial rotation has been described extensively in the literature (Hicks, 1953; Inman 1976; Olerud et al., 1985; Lundberg, 1989; Hintermann et al., 1993). This movement coupling between foot and shank results in the tibia rotating internally between touchdown and midstance and has recently been associated with running injuries (Nigg et al., 1993; Stergiou, 1996; McClay et al., 1997).

Foot orthoses and shoe sole modifications have been proposed to reduce excessive movements of foot and shank (James et al., 1978; Bates et al., 1978; Clarke et al., 1984; Segesser et al., 1980; van Woensel et al., 1992; Nigg et al., 1987; Milani et al., 1995). Most of these studies used a two-dimensional analysis that has been shown to be affected by the alignment of the foot with respect to the camera (Areblad et al., 1990). Additionally, the majority of these studies used shoe- and skin-mounted markers that are known to overestimate the skeletal movements (Cappozzo et al., 1996; Reinschmidt et al., 1997a). Thus, the results of these studies may not have reflected the kinematics of the underlying bone.

Running with shoes may change foot and leg kinematics compared to running barefoot. Hence, barefoot running is often looked upon as the baseline for normal running (Clarke et al., 1984). In barefoot running the foot has been shown to invert less at touchdown (between 5° to 8°), decrease maximum eversion velocity (about 300°/s) and total eversion (between 4° to 13°) compared to shod running (Bates et al., 1978; Nigg et al., 1980; Stacoff et al., 1991; Vagenas et al., 1992). Thus, tibial rotations may be assumed to be decreased in barefoot running compared to shod running (assuming that movement coupling in barefoot and shod running does not differ). Consequently, it may be suggested that barefoot running could lead to fewer running injuries than shod running, provided there are no additional injuries from the lack of foot protection. To date, tibial rotation of barefoot running has not been documented in the literature and the suggestion about possible advantages and/or disadvantages of barefoot running lacks information on the skeletal movement during barefoot or shod movements, possible changes in muscle activity and epidemiological data.

The purpose of this study was to quantify three-dimensional skeletal movement differences between barefoot and shod running using markers fixed to bone pins during the stance phase of running. The following hypotheses were to be tested: Skeletal movements of barefoot running, when compared to shod running, show

I. decreased total calcaneal eversion,
II. decreased total tibial rotation,
III. unchanged tibiocalcaneal coupling.

6.2 Methods

6.2.1 General Project Description

The experiments were performed in the Department of Orthopaedics, Karolinska Institute at Huddinge University Hospital, Stockholm. The experiments were approved by the Ethics committees of the Karolinska Hospital and The University of Calgary. The experimental set-up, test procedure, data analysis and data reduction have already been described earlier (chapter 3; Reinschmidt et al., 1997a; 1997c).

In brief, five healthy male volunteers, all injury free with no previous injury history that might influence their locomotion patterns, participated as test subjects (28.6 ± 4.3 yrs., mass 83.4 ± 10.2 kg and height 185.1 ± 4.5 cm). None of the subjects had a clinically conspicuous foot. Intracortical Hofmann pins with reflective marker triads were inserted under standard local anesthetic which was active for 2-3 hours, leaving enough time for the experiments. The subjects gave their informed consent to participate in the study and performed heel-toe running trials at a speed between 2.5 to 3.0 m/s. Trials were repeated if the subjects did not land with their right foot on the force plate and/or if they made an obvious gait pattern modification in order to hit the force plate.

Three high-speed cine cameras (LOCAM) were placed around a force platform (KISTLER) mounted flush to the runway. The camera speed was set at 200 Hz and three LED's, triggered by a threshold detector connected to the force plate, were used to synchronize the cameras. A calibration frame with 6 control points ($0.5 \times 0.5 \times 0.5\ m^3$) was used for the three-dimensional reconstruction.

6.2.2 Test Conditions, Shoes and Orthoses

The tests were performed barefoot, with a normal test shoe and with five modifications of the normal test shoe. The normal test shoe (Adidas Equipment Cushioning) had a dual density midsole of Shore A 35 on the lateral and Shore A 45 on the medial side and a standard insert which was assumed to have no mechanical support for the foot. The standard insert was exchanged with two orthoses, one anterior to support the foot arch (modification 1) and one posterior to support the sustentaculum tali of the calcaneus (modification 2). The remaining three modifications concerned the shoe sole which was changed to a single density midsole (Shore A 45). The lateral heel flare was modified with a wide flare (modification 3), a neutral flare (modification 4) and a rounded sole (modification 5). The outer sole consisted of a hard rubber sole of Shore A 65. The different shoes and orthoses used have been explained in chapters 4 and 5 respectively. Each test condition was repeated three times (except for the normal shoe condition with five repetitions). The subjects performed a total of 23 running trails. All shoe heel counters had a lateral cutout to prevent impingement with the calcaneal bone pin during running as described previously (chapter 4).

6.2.3 Data Analysis and Reduction

KineMat, a set of programs written in MATLAB™, was adapted from Reinschmidt and van den Bogert (1997d) for the specific needs of this investigation. The programs served to reconstruct the three-dimensional marker positions and to calculate the relative segmental movements. The barefoot trial was used as the neutral position to define the segment-fixed calcaneus and tibial coordinate systems (Reinschmidt et al., 1997a). The rotations between the segments were calculated as Cardanic angles for the stance phase of all test conditions using a joint coordinate system (JCS) at the ankle joint complex, with the defined sequence of rotations of plantar/dorsiflexion about a tibia fixed medio-lateral axis, calcaneal ab/adduction about the floating axis, and in/eversion about the antero-posterior axis of the calcaneus (Cole et al., 1993). Tibial rotation was calculated using the sequence: tibial rotation about a tibia fixed proximal-distal axis, in/eversion about the floating axis, and plantar/dorsiflexion about a calcaneus fixed medio-lateral axis (Nigg et al., 1993).

The accuracy of the spatial reconstruction between two marker triads was determined (i) based on the residuals of the DLT equations averaged over the entire stance phase for all markers and (ii) based on the deviations of the inter-marker distances of the same trials. The mean error based on DLT residuals was found to be in the order of ±4° which included noise error and lens distortion error. The mean error based on marker distances (RMS) was found to be ±1.0° including noise error only. Thus, for the present study, a realistic estimation of the error was likely between the two errors given above.

6.2.4 Definitions of Variables

In/eversion and tibial rotation variable definitions are explained in Table 3-2 and depicted Figure 3-7. The variables were defined between touchdown and midstance of running. The inversion positions at touchdown (β_o and ρ_o) were considered to detect possible adaptations to shoe interventions before touchdown. Biomechanical factors, which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. It has been suggested that excessive eversion (i.e. maximum eversion (β_{max}) and total eversion ($\Delta\beta_{max}$)) forces the Achilles tendon to bend laterally, hereby producing an asymmetric stress distribution across the tendon, which could lead to Achilles tendon problems (Smart et al., 1980; Clement et al., 1981; Denoth, 1986). Excessive eversion velocity ($\dot{\beta}_{max}$) produces eccentric loading of the muscles of the posterior tibial compartment, which control/reduce eversion after touchdown. Forces acting on muscle-tendon units during eccentric loading are increased compared to concentric loading. Excessive eversion velocity has been associated with overloading and injury of the muscles of the posterior tibial compartment, e.g. medial tibial stress syndrome (Segesser et al., 1980; Viitasalo et al., 1983; Messier et al., 1988; Stacoff et al., 1988; DeWit et al., 1995). Eversion velocity was defined in the window of 10 to 40% of ground contact time, excluding values before 10% because of possible inaccuracies, e.g. filtering effects and/or markers close to the edge of the defined three dimensional space. After 40%, eversion velocities had reached their maximum in all trials of the test. Excessive tibial rotation ($\Delta\rho_{max}$) has been associated with the changes in the tracking of the patella, hereby changing the contact pressure and possibly the friction of the articulating surface of the patella, which may be related to the occurrence of the patellafemoral pain syndrome (Stergiou, 1996). Tibial rotation is thought to take place as a result of the movement coupling from the calcaneus to the tibia. The coupling variable T_{CT} ($= \frac{\Delta\rho \max}{\Delta\beta \max}$) described how much of total eversion is transferred to total internal tibial rotation and indicated how much movement coupling at the ankle was influenced by the shoe modifications.

The above mentioned variables indirectly describe the movement at those structures of interest, but do not directly describe the load within these structures. However, they are relatively easy to quantify.

The testing procedure (5.2.2) was organized such that test conditions were independent from each other. The residuals of the test variables were inspected for normal

distribution. All variables were tested with two-tailed ANOVA techniques with repeated measures, the one-way ANOVA to test subject independent orthotic and shoe sole effects, the two-way ANOVA to test subject dependent effects, as well as possible interactions between the subjects and the orthotic and sole conditions. In cases of contradicting results between the one-way and two-way ANOVA, the more conservative result of the one-way ANOVA was accepted. The power analysis conducted on the kinematic variables suggested that there was a 80% chance of detecting any differences in these variables between the test conditions which were greater than 3.5°.

6.3 Results

The results are presented in two parts: (6.3.1) A comparison between barefoot and shod running using the "normal" shoe, and (6.3.2) a similar comparison using all shoe modifications.

6.3.1 Barefoot versus Shod Running

Eversion and internal tibial rotation took place from touchdown until midstance in all subjects, and inversion and external tibial rotation took place from midstance to take-off (Figure 6-1).

The in/eversion and tibial rotation diagrams illustrate that movement coupling for the barefoot and shod conditions were similar (Figure 6-2). Both curves start in inversion and external tibial rotation at heel strike, move to eversion and internal tibial rotation in midstance and return to inversion and external tibial rotation at takeoff.

The individual differences in total calcaneal eversion and total internal tibial rotation during ground contact between shod and barefoot running were found to be small (less than 1.6°) and not systematic (Figure 6-3).

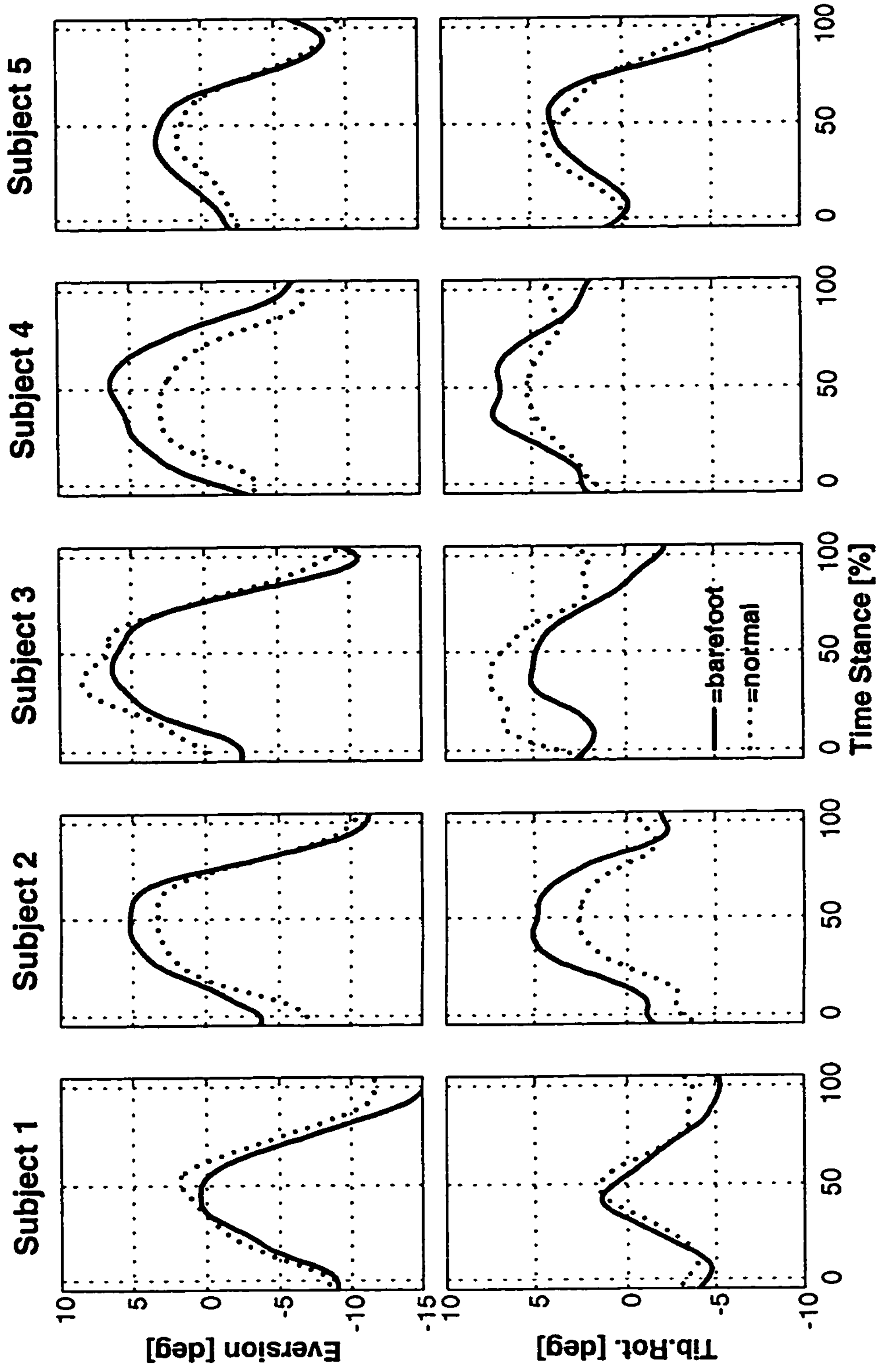


Figure 6-1: Mean curves of in/eversion and tibial rotation of all conditions and all subjects: (—) barefoot, (• • •) normal shoe. The standard deviation during the stance phase was on average ±1.1° for eversion and tibial rotation.

The statistical comparison between barefoot and shod running showed no significant difference in any of the test variables. However, differences between subjects were significant for all but the coupling variable (Table 6-2).

6.3.2 Barefoot versus Shoe Modifications

The comparison between barefoot and shod running including all orthotic and shoe sole modifications showed some selected significant differences:

Touchdown: Calcaneal inversion for the modified shoe conditions showed significant interactions between subjects and shoes (p<.01; except for the straight shoe). Thus the differences at touchdown were not significant (one-way ANOVA). Comparison of a single subject's data showed small barefoot versus shod differences (2° to 3°). The differences across subjects were up to 8° and unsystematic (Table 3, appendix). Only

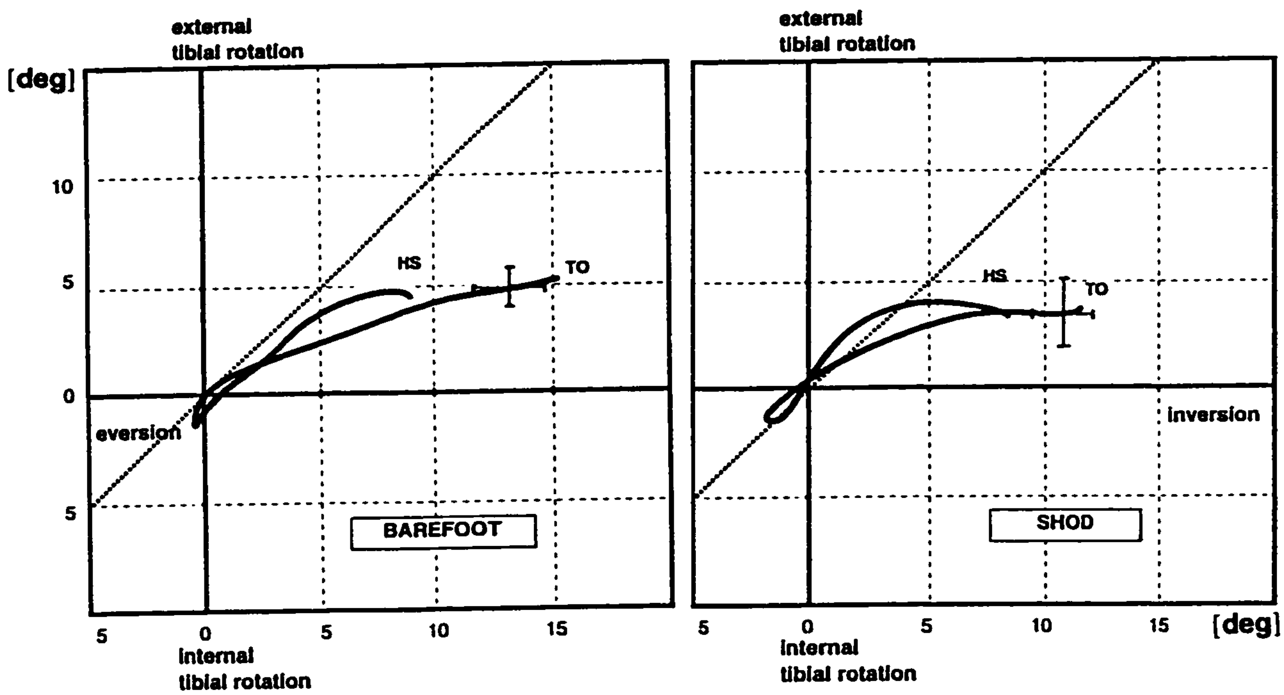


Figure 6-2: Example of the relationship between eversion and tibial rotation (subject 1) for barefoot running and running with the normal shoe. The thick line represents the mean curve, mean SD are indicated.

subject 3 inverted more in the barefoot condition than in the shod condition. The position of the tibia at touchdown was inconsistent across subjects. The tibia was externally rotated in subjects 1 and 2, internally rotated in subjects 3 and 4 and varied in subject 5. The differences between the shoe conditions were between 1° to 3°, but the differences between the subjects were up to 7°. There was no significant difference between the barefoot and shod conditions.

Maximum velocity: Differences of maximum eversion velocity between subjects often exceeded 100°/s for the same shoe condition. Barefoot running showed the lowest eversion velocity in subjects 1, 2 and 4, but not in subject 3 and 5. Maximum internal tibial rotation velocity was consistently slower than maximum eversion velocity (except in subject 3). The slowest internal tibial velocity was that of the posterior orthotic shoe condition which was the only barefoot versus shod comparison to be significant ($p<.01$).

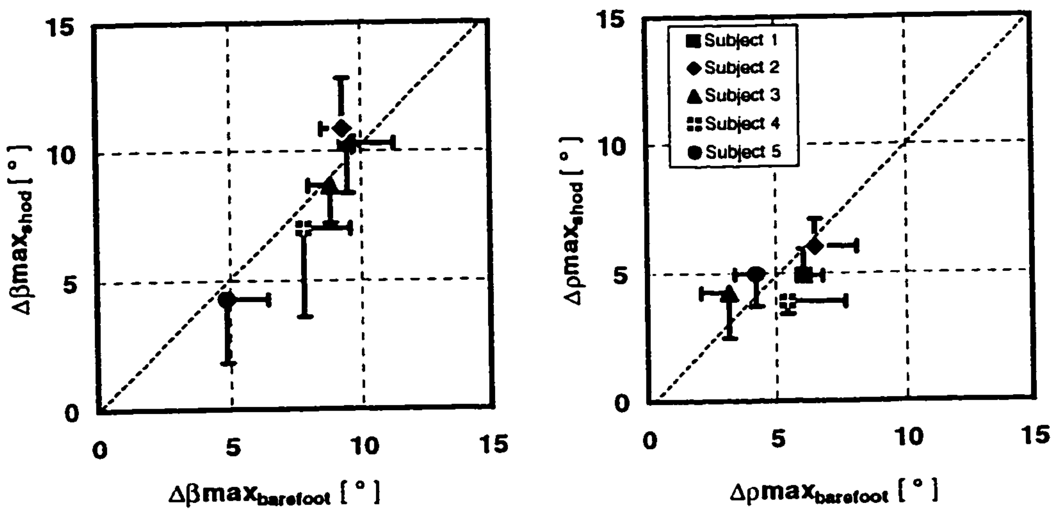


Figure 6-3: Total eversion of shod running (normal shoe) versus total eversion of barefoot running (left) and total internal tibial rotation of shod running versus total internal tibial rotation of barefoot running (right) of all five test subjects.

Maximum and total movement: The differences in maximum and total calcaneal eversion between barefoot and the shoe modifications were found to be small (in the order of 1° to 3°) and inconsistent over the five subjects. Maximum barefoot eversion was greater than when shod (in subject 2, 4 and 5, but not in subjects 1 and 3), and not significant. Total barefoot eversion was found to be very similar to total shod eversion because in barefoot running the increased maximum eversion was compensated by a smaller touchdown inversion. Internal tibial rotation differences were small, in the order of 1° to 3° and inconsistent over the five subjects. The differences between the subjects were up to 7° ($p<.01$). The barefoot versus posterior orthosis was the only significant comparison ($p<.01$).

Vari-able	PART I		PART II									
	BAREFOOT VS. NORMAL		MODIFICATIONS OF ORTHOSES				MODIFICATIONS OF SHOE SOLES					
			barefoot - posterior		barefoot - anterior		barefoot - straight		barefoot - flared		barefoot - round	
	1-way	2-way	1-way	2-way	1-way	2-way	1-way	2-way	1-way	2-way	1-way	2-way
β_o	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 n.s.	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.01
β_{max}	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 <.05 2 <.01 3 n.s.	n.s.	1 <.05 2 <.01 3 n.s.	n.s.	1 <.05 2 <.01 3 n.s.	n.s.	1 <.05 2 <.01 3 n.s.
$\Delta\beta_{max}$	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\beta}_{max}$	n.s.	1 n.s. 2 <.05 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.	n.s.	1 <.05 2 <.01 3 <.05	n.s.	1 <.05 2 <.01 3 n.s.
ρ_o	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 <.05	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 n.s.
ρ_{max}	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 <.01 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 <.05	n.s.	1 n.s. 2 <.01 3 <.01	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\Delta\rho_{max}$	n.s.	1 n.s. 2 <.05 3 n.s.	<.01	1 <.01 2 <.05 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.	n.s.	1 n.s. 2 <.05 3 n.s.	n.s.	1 n.s. 2 <.05 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.
$\dot{\rho}_{max}$	n.s.	1 n.s. 2 <.05 3 n.s.	<.01	1 <.01 2 <.05 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.	n.s.	1 n.s. 2 <.05 3 n.s.	n.s.	1 n.s. 2 n.s. 3 n.s.
T_{CT}	n.s.	1 n.s. 2 n.s. 3 n.s.	<.01	1 <.01 2 n.s. 3 n.s.	n.s.	1 n.s. 2 <.05 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.	n.s.	1 n.s. 2 <.01 3 n.s.

Table 6-1: Statistics of variables of this study. The following abbreviations are used: 1 = difference between barefoot and shoe condition; 2 = difference between subjects; 3 = interaction between barefoot/shoe and subjects; n.s. = not significant.

Movement coupling between the calcaneus and tibia: Movement coupling between the barefoot and shod conditions was inconsistent over the five subjects. With shoes compared to barefoot, two subjects showed decreased coupling (subject 1 and 2), one an increased coupling (subject 3) and two inconsistent results (subjects 4 and 5). The transfer coefficient over all test conditions were subject dependent and varied between 0.44 ± 0.07 (subject 1 and 3), 0.54 ± 0.11 (subject 2), 0.62 ± 0.15 (subject 4), and 0.89 ± 0.27 (subject 5). All subjects, except subject 3, showed decreased coupling with the posterior orthosis compared to barefoot running.

Interactions: Interactions between test conditions (barefoot/shod) and the test subjects were found to be significant mainly for the touchdown variables (9 out of 12), some for the maximum variables (6 out of 12), but none for velocity variables or coupling variables.

6.4 Discussion

The purpose of this study was to quantify the differences in calcaneal and tibial movements during ground contact for running barefoot, with a normal shoe, and with shoe modifications. It was suggested that total eversion and total internal tibial rotation for barefoot running would be smaller than for running with shoes. These suggestions, which have been formulated in hypothesis I and II, could not be confirmed. The results showed that the differences between skeletal movements in barefoot versus shod running were small and not systematic, and that the differences between subjects were larger than the differences between shoe and barefoot conditions. Furthermore, it was found that the coupling mechanism between calcaneus and tibia was only minimally affected by normal shoes, shoe sole modifications and orthoses (with the exception of the posterior orthoses). Thus, hypothesis III of this study was accepted for all barefoot versus shod comparisons excluding the posterior orthoses which significantly decreased movement coupling between the calcaneus and the tibia compared to the barefoot condition.

The investigation was limited by the fact that the test shoes had a cutout in the lateral heel counter that was necessary to prevent impingement with the calcaneal bone pin and that local anesthesia was applied at the bone pin insertion sites. However, there is evidence (chapter 4 and 5) that these factors (test shoes and/or local anesthesia) did not substantially influence the kinematics during testing. Therefore, the findings of this study will be interpreted under the assumption that the present results are real and would be repeated with another set of shoes and subjects.

6.4.1 Barefoot versus Shod Running

Previous studies using skin and shoe mounted markers have shown substantial and significant differences between barefoot and shod running with respect to eversion (Bates et al., 1978; Nigg et al., 1980; Stacoff et al., 1991; Vagenas et al., 1992). However, these results can not be supported by the present study. The differences in calcaneal eversion between barefoot running and running with normal shoes were small, not systematic and not significant. The same result was found for internal tibial rotation and for movement coupling. Consequently, one can conclude (a) that previous studies described the movements of the shoe and/or skin and did not reflect the movement of the underlying bone or (b) that the results of the present study represent only an exception, or (c) that both results were correct, but maybe caused by a third factor, e.g. the running surface. Thus, if one is interested in the actual skeletal movement, skin marker studies may not provide the appropriate information.

6.4.2 Barefoot versus Shoe Modifications

Tibiocalcaneal bone movement differences between barefoot and shod running did occur only with extreme shoe modifications. One such modification was the posterior orthoses which showed a significant decrease in several variables compared with the barefoot condition. This result seemed to contradict the work of Lafortune et al. (1994)

who found no significant differences on internal tibiofemoral rotations between normal and 10° varus wedged shoes during walking and using bone pins. However, when considering Lafortune's result of tibial rotation in space (where a varus wedge effect was shown) both results come to a similar conclusion.

Extreme shoe modifications were the flared and the round shoe soles which showed an increased maximum eversion velocity in subjects 1,2,4,5 (not significant). Both of these shoe sole conditions were special. The flare of the sole was 20° lateral (about twice that of an average running shoe) which may have acted as a lever that forced the foot into eversion. The round shoe sole may have acted as a ramp as a result of the combination of the round sole geometry and the hard outer sole. This suggests, that only extreme shoe modifications may affect tibiocalcaneal movement patterns during running. Normal shoes, or less extreme changes, like those in the anterior orthosis and the straight shoe condition, seem not to affect the kinematics measured on the bone level.

The results of this study were achieved with clinically inconspicuous feet. However, McClay et al (1997) showed that the coupling at the ankle may depend on the foot type (normal versus pronator). Thus, it can be argued that kinematic responses of shoe sole modifications may be changed when different types of clinically conspicuous feet are tested. Furthermore, it is suggested that the results of the present study may be different if higher forces were acting (e.g. at higher running speeds or during cutting movements) and consequently the effect of the shoe modifications would be more prominent. However, at the present time it is not possible to draw conclusions with respect to these suggestions.

6.4.3 Interpretation of Movement Coupling

The transfer coefficient T_{CT} is the variable used to describe the coupling movement between calcaneal eversion and tibial rotation (Table 3, appendix). The only shoe condition found to be significantly different to barefoot running was the posterior orthosis. All subjects, except subject 3 (possibly with an effect of inaccurate touchdown data), showed decreased coupling with the posterior orthosis compared to barefoot running. This suggests that posterior orthoses may reduce tibia loading in some subjects. The individual differences of the average transfer coefficient varied between 0.44 and 0.89 indicating that each of the five subjects had a distinct coupling mechanism (with a certain variation) between the movements of the calcaneus and tibia.

The study results compare well with the *in-vitro* studies by Olerud et al., (1985; coefficient 0.42) and Hintermann (1993; coefficient 0.74), but are larger than the *in-vivo* study using internal markers by Lundberg (1989; coefficient 0.2). Studies using shoe and skin-mounted markers report average coefficients of 0.76 (Nigg et al., 1993) and 0.72 (McClay et al., 1997). There is evidence that the coupling coefficient is dependent on the vertical load, the amount of plantar/dorsiflexion, changes in ligament integrity and muscle-tendon forces (Hintermann, 1994). Thus, comparisons of the coefficients between different studies are limited because of the different methodologies used.

6.5 Conclusion

This *in-vivo* study showed that bone movements during barefoot running are generally very similar to those inside shoes. The normal shoe condition showed no significant difference relative to barefoot running in either of the test variables. The results of this *in-vivo* study contrast with previous investigations using skin and shoe-mounted markers, and suggests that these discrepancies may be the result of the overestimation of externally mounted markers.

The posterior orthosis showed significant differences ($p<.01$) to barefoot running (variables: internal tibial rotation, internal tibial rotation velocity and tibiocalcaneal coupling). All other barefoot versus shod test variable differences were found to be not significant. It is concluded that, measured at the bone level, calcaneal and tibial movement patterns do not change substantially between shod and barefoot running.

6.6 Summary

Barefoot running kinematics has been described to vary considerably from shod running. However, previous investigations were typically based on externally mounted shoe and/or skin markers, which have been shown to overestimate skeletal movements. Thus, the purpose of this study was to compare calcaneal and tibial movements of barefoot versus shod running using skeletal markers. Intracortical bone pins with reflective marker triads were inserted under standard local anesthetic into the calcaneus and tibia of five healthy male subjects. The subjects ran barefoot, with a normal shoe, with three shoe soles and two orthotic modifications. The three-dimensional tibiocalcaneal rotations were determined using a joint coordinate system. Test variables were defined for eversion and tibial rotation. The results showed that the differences of bone movements of barefoot versus shod were small and not systematic. However, the differences between the subjects were significant for most study variables ($p<.01$). Only one specific shoe modification (posterior orthosis) showed significant ($p<.01$) differences to barefoot running with three variables (internal tibial rotation, internal tibial rotation velocity and calcaneal-tibial coupling). All other comparisons were not significant. It is concluded that calcaneal and tibial movement patterns do not change substantially when using shoes compared to running barefoot, but differences may occur when extreme shoe modifications are used and that the effects of these interventions are subject specific. The result of this *in-vivo* study contrasts with previous investigations using skin and shoe mounted markers and suggests that these discrepancies may be the result of the overestimation of externally mounted markers.

6.7 Appendix

Variable	condition	subject 1	subject 2	subject 3	subject 4	subject 5	mean	SD
β_0 [ ]	[illegible]	-8.92 ± 0.42	-3.61 ± 0.65	-2.47 ± 1.18	-0.92 ± 1.14	-1.41 ± 0.23	[illegible]	[illegible]
	[illegible]	-8.37 ± 1.89	-6.80 ± 1.20	0.10 ± 0.61	-3.53 ± 2.24	-2.16 ± 1.47	[illegible]	[illegible]
	[illegible]	-9.64 ± 0.97	-7.66 ± 0.64	0.58 ± 0.32	-2.98 ± 1.13	-4.93 ± 0.51	[illegible]	[illegible]
	[illegible]	-9.83 ± 1.60	-7.02 ± 0.61	-0.53 ± 1.49	-1.91 ± 0.52	-3.90 ± 1.14	[illegible]	[illegible]
	[illegible]	-11.22 ± 0.87	-7.07 ± 1.08	-1.97 ± 1.66	-2.17 ± 1.72	-4.01 ± 1.21	[illegible]	[illegible]
	[illegible]	-11.05 ± 2.51	-8.72 ± 1.66	-1.28 ± 1.48	-3.29 ± 0.40	-3.29 ± 1.40	[illegible]	[illegible]
	[illegible]	-11.15 ± 1.45	-6.93 ± 0.59	-1.50 ± 0.43	-1.71 ± 0.71	-3.82 ± 0.48	[illegible]	[illegible]
βmax [ ]	[illegible]	0.56 ± 1.47	5.71 ± 1.41	6.36 ± 1.82	6.88 ± 0.70	3.45 ± 1.34	[illegible]	[illegible]
	[illegible]	1.92 ± 0.93	4.05 ± 1.00	8.80 ± 1.48	3.53 ± 1.56	2.09 ± 1.80	[illegible]	[illegible]
	[illegible]	1.82 ± 0.67	1.54 ± 0.74	7.21 ± 0.30	5.52 ± 1.02	-0.85 ± 1.25	[illegible]	[illegible]
	[illegible]	0.85 ± 0.40	3.06 ± 0.71	7.50 ± 1.05	4.41 ± 2.41	0.42 ± 2.28	[illegible]	[illegible]
	straight	-0.77 ± 0.20	4.02 ± 0.91	5.87 ± 0.82	6.45 ± 0.52	1.97 ± 1.29	[illegible]	[illegible]
	flared	-0.09 ± 0.42	2.46 ± 1.94	5.25 ± 1.97	6.56 ± 0.44	2.32 ± 0.51	[illegible]	[illegible]
	round	0.23 ± 1.09	3.38 ± 1.79	6.39 ± 0.06	3.60 ± 1.26	2.59 ± 2.22	[illegible]	[illegible]
$\Delta\beta$max [ ]	barefoot	9.48 ± 1.81	9.31 ± 0.81	8.82 ± 0.86	7.80 ± 1.80	4.86 ± 1.56	[illegible]	[illegible]
	normal	10.29 ± 1.89	10.85 ± 1.94	8.70 ± 1.44	7.05 ± 3.49	4.26 ± 2.46	[illegible]	[illegible]
	posterior	11.45 ± 0.30	9.19 ± 1.20	6.63 ± 0.29	8.50 ± 2.13	4.08 ± 1.73	[illegible]	[illegible]
	anterior	10.68 ± 2.00	10.08 ± 1.29	8.03 ± 1.32	6.32 ± 1.98	4.32 ± 3.19	[illegible]	[illegible]
	straight	10.45 ± 0.76	11.09 ± 1.09	7.84 ± 1.34	8.62 ± 2.16	5.98 ± 2.51	[illegible]	[illegible]
	flared	10.96 ± 2.73	11.18 ± 3.05	6.52 ± 1.17	9.86 ± 0.28	5.61 ± 1.89	[illegible]	[illegible]
	round	11.37 ± 2.41	10.31 ± 1.43	7.89 ± 0.50	5.31 ± 0.59	6.41 ± 2.70	[illegible]	[illegible]
$\dot{\beta}$max [/s]	[illegible]	145.94 ± 27.28	96.71 ± 29.26	144.46 ± 38.34	113.93 ± 25.68	82.16 ± 14.19	[illegible]	[illegible]
	normal	151.68 ± 49.69	157.41 ± 51.66	138.18 ± 39.29	141.25 ± 82.46	73.17 ± 20.59	[illegible]	[illegible]
	posterior	171.44 ± 24.45	122.13 ± 34.76	110.78 ± 21.42	148.40 ± 18.93	96.87 ± 10.36	[illegible]	[illegible]
	anterior	168.15 ± 16.26	152.31 ± 23.82	133.97 ± 53.00	146.83 ± 29.74	85.44 ± 29.94	[illegible]	[illegible]
	straight	144.47 ± 22.67	138.25 ± 21.23	125.10 ± 33.77	170.95 ± 94.39	80.84 ± 36.97	[illegible]	[illegible]
	flared	172.49 ± 44.10	160.66 ± 54.11	108.64 ± 17.36	211.97 ± 50.70	67.72 ± 24.54	[illegible]	[illegible]
	round	191.47 ± 41.80	162.55 ± 15.16	149.43 ± 22.42	113.44 ± 30.11	90.06 ± 3.42	[illegible]	[illegible]

Variable	condition	subject 1	subject 2	subject 3	subject 4	subject 5	mean	SD
[illegible]	barefoot	-4.55 ± 0.67	-1.16 ± 0.43	2.19 ± 0.82	2.33 ± 0.78	0.01 ± 0.67	[illegible]	[illegible]
	normal	-3.35 ± 0.24	-2.95 ± 0.54	3.97 ± 1.03	1.63 ± 0.32	-0.12 ± 0.44	[illegible]	[illegible]
	posterior	-3.52 ± 0.24	-3.44 ± 0.30	4.11 ± 0.99	3.41 ± 0.99	0.67 ± 0.24	[illegible]	[illegible]
	anterior	-3.94 ± 0.56	-3.09 ± 0.40	2.79 ± 2.31	3.47 ± 1.20	-0.58 ± 0.88	[illegible]	[illegible]
	straight	-2.75 ± 0.27	-2.02 ± 0.23	1.45 ± 1.11	3.85 ± 2.05	-1.41 ± 0.83	[illegible]	[illegible]
	flared	-1.98 ± 0.32	-2.25 ± 0.64	1.22 ± 1.45	3.10 ± 0.64	-0.64 ± 1.18	[illegible]	[illegible]
	round	-2.47 ± 0.29	-2.52 ± 1.03	2.62 ± 0.27	3.70 ± 1.13	-3.46 ± 5.16	[illegible]	[illegible]
ρmax [illegible]	barefoot	1.55 ± 0.29	5.37 ± 1.21	5.38 ± 1.32	7.79 ± 1.69	4.26 ± 0.28	[illegible]	[illegible]
	normal	1.55 ± 1.87	3.14 ± 0.94	8.21 ± 1.40	5.50 ± 0.47	4.85 ± 1.42	[illegible]	[illegible]
	posterior	0.12 ± 0.55	1.49 ± 0.55	7.17 ± 1.55	6.62 ± 1.52	1.94 ± 0.03	[illegible]	[illegible]
	anterior	0.14 ± 0.44	1.84 ± 0.64	6.70 ± 1.12	7.90 ± 1.37	3.40 ± 2.29	[illegible]	[illegible]
	straight	2.11 ± 0.27	1.86 ± 0.18	4.74 ± 0.41	9.35 ± 1.29	3.54 ± 1.91	[illegible]	[illegible]
	flared	2.04 ± 1.15	3.02 ± 1.00	4.68 ± 0.52	7.97 ± 0.75	5.29 ± 0.63	[illegible]	[illegible]
	round	1.82 ± 0.67	3.67 ± 0.80	5.38 ± 0.55	7.98 ± 1.00	2.27 ± 4.27	[illegible]	[illegible]
$\Delta\rho max$ [illegible]	barefoot	6.09 ± 0.76	6.53 ± 1.64	3.19 ± 1.11	5.46 ± 2.24	4.25 ± 0.82	[illegible]	[illegible]
	normal	4.91 ± 1.85	6.09 ± 1.07	4.24 ± 1.82	3.86 ± 0.50	4.97 ± 1.29	[illegible]	[illegible]
	posterior	3.64 ± 0.77	4.93 ± 0.74	3.06 ± 2.51	3.21 ± 2.35	1.27 ± 0.22	[illegible]	[illegible]
	anterior	4.08 ± 0.92	4.93 ± 0.46	3.91 ± 3.43	4.43 ± 2.22	3.99 ± 2.19	[illegible]	[illegible]
	straight	4.86 ± 0.38	3.88 ± 0.10	3.29 ± 0.92	5.50 ± 0.75	4.95 ± 1.50	[illegible]	[illegible]
	flared	4.02 ± 1.08	5.26 ± 1.55	3.46 ± 1.46	4.87 ± 1.00	5.93 ± 0.70	[illegible]	[illegible]
	round	4.29 ± 0.74	6.19 ± 1.75	2.76 ± 0.53	4.28 ± 2.13	5.73 ± 1.03	[illegible]	[illegible]
$\dot{\rho} max$ [S]	barefoot	101.75 ± 11.81	91.09 ± 20.07	77.13 ± 20.87	99.35 ± 26.06	61.60 ± 28.87	[illegible]	[illegible]
	normal	100.06 ± 20.09	91.64 ± 14.42	50.99 ± 8.87	69.93 ± 30.32	74.42 ± 29.24	[illegible]	[illegible]
	posterior	72.08 ± 13.20	74.00 ± 24.71	47.18 ± 16.41	75.35 ± 11.96	47.16 ± 17.49	[illegible]	[illegible]
	anterior	91.84 ± 32.23	101.04 ± 26.97	54.91 ± 32.65	88.78 ± 32.33	61.13 ± 28.03	[illegible]	[illegible]
	straight	94.69 ± 24.88	78.31 ± 20.97	53.98 ± 22.19	86.26 ± 22.48	107.20 ± 30.21	[illegible]	[illegible]
	flared	87.38 ± 22.64	82.14 ± 24.73	45.78 ± 24.19	98.21 ± 11.73	85.32 ± 12.38	[illegible]	[illegible]
	round	68.72 ± 22.41	84.25 ± 25.09	44.27 ± 3.35	80.52 ± 21.88	84.51 ± 12.46	[illegible]	[illegible]
T_{CT}	barefoot	0.65 ± 0.06	0.70 ± 0.16	0.37 ± 0.15	0.68 ± 0.13	0.91 ± 0.22	[illegible]	[illegible]
	normal	0.49 ± 0.22	0.58 ± 0.15	0.50 ± 0.23	0.66 ± 0.30	0.99 ± 0.41	[illegible]	[illegible]
	posterior	0.32 ± 0.07	0.54 ± 0.03	0.46 ± 0.37	0.34 ± 0.22	0.36 ± 0.17	[illegible]	[illegible]
	anterior	0.39 ± 0.08	0.50 ± 0.12	0.45 ± 0.38	0.69 ± 0.26		[illegible]	[illegible]
	straight	0.47 ± 0.07	0.35 ± 0.03	0.43 ± 0.15	0.68 ± 0.26	0.96 ± 0.51	[illegible]	[illegible]
	flared	0.39 ± 0.16	0.47 ± 0.04	0.53 ± 0.20	0.49 ± 0.09	1.13 ± 0.34	[illegible]	[illegible]
	round	0.38 ± 0.03	0.62 ± 0.24	0.35 ± 0.09	0.78 ± 0.32	0.97 ± 0.35	[illegible]	[illegible]

Table 6-2: The results and SD of the eversion (β) and tibial rotation variables (ρ), and the results of movement coupling (T_{CT}). Positive values represent eversion and internal tibial rotation; negative values denote inversion and external tibial rotation.

7. MOVEMENT COUPLING IN THE LOWER EXTREMITIES DURING THE STANCE PHASE OF RUNNING

7.1 Introduction

The knee is the most common site of running injuries (Clement et al., 1981; James et al., 1990; Mechelen, 1992). However, the etiology of knee injuries is presently not well understood (McClay et al., 1997). It has been suggested, that excessive tibial rotation is associated with the development of knee problems, for instance by altering the tracking of the patella (Subotnick, 1977; James et al., 1978; Clement et al., 1981). It has been further suggested that excessive tibial rotation is a result of excessive foot eversion determined by a coupling mechanism at the ankle (Inman, 1976; James et al., 1978; Clement et al., 1981; Lundberg, 1989). There is evidence, that this coupling mechanism is increased in high-arched and stiff feet (Nigg et al., 1993).

Movement coupling has been discussed for many decades. Coupling between foot eversion and internal tibial rotation has been associated with the oblique orientation of the subtalar joint axis (Meyer, 1853; Henke, 1859; Fick, 1911; Elftman and Manter, 1935; Jones, 1945; Hicks, 1953; Inman, 1969, 1976). Movement coupling during walking was first investigated by Levens et al. (1948) and by Wright et al. (1964) who concluded that the tibial rotation must be resolved at the subtalar joint. Later, Lundberg (1989), using implanted markers *in-vivo*, found that moving the foot from pronation to supination induced some vertical axis rotation of the lower leg, indicating the existence of a rotation transferring mechanism. The work of Levens and co-workers (1948) quantifying tibial rotation using bone pins started a series of *in-vivo* investigations aimed at the understanding of skin and shoe marker artefacts that may mask the kinematics of the underlying bone (McClay, 1990; Lafortune et al., 1994; Cappozzo, 1996; Reinschmidt et al., 1997a, 1997b). Up to the present, most studies on kinematics of running have used externally mounted skin and shoe markers which have been demonstrated to overestimate skeletal movements (Reinschmidt et al., 1997a, 1997b) and thus do not represent the movements of the underlying bones.

Movement coupling between foot in/eversion and tibial rotation has also been described by *in-vitro* studies (Inman, 1976; Olerud et al., 1985; Siegler et al., 1988; Hintermann, 1994). The coupling was found to be higher between inversion and external tibial rotation than between eversion and internal tibial rotation and was found to depend on vertical load, plantar/dorsiflexion, ligament integrity and muscle-tendon forces.

Movement coupling occurs between the shoe and the foot, the foot and the shank, and the shank and the thigh. *In-vivo,* movement coupling between foot/shoe and shank has been described by Nigg et al. (1993) and McClay et al. (1997) using external markers during running. The coupling between shoe eversion and calcaneus eversion was first described by Reinschmidt et al. (1997a) using bone pins in running. Relative movements between tibia and femur have been described by Lafortune et al. (1994) and Reinschmidt et al. (1997b) using bone pins in walking. However, the cited bone pins measurements did not (or only minimally) address the coupling aspect between neighboring segments starting from the shoe to the calcaneus, then to the tibia and to the femur.

Therefore, the purpose of this study was to quantify the movement coupling between (i) shoe and calcaneus, (ii) calcaneus and tibia, and (iii) tibia and femur during the stance phase of running.

7.2 Methods

7.2.1 General Project Description

The experiments were performed in the Department of Orthopaedics, Karolinska Institute at Huddinge University Hospital, Stockholm. Ethical approval for the experiments was obtained from the Ethics committees of the Karolinska Hospital and The University of Calgary. The experimental set-up, test procedure, data analysis and data reduction have been described previously in detail (chapter 3; Reinschmidt et al., 1997a; 1997b; 1997c).

In brief, five healthy male volunteers, all injury free with no previous injury history that might influence their locomotion patterns, participated as test subjects (28.6 ± 4.3 yrs., mass 83.4 ± 10.2 kg and height 185.1 ± 4.5 cm). Bone pins with reflective marker triads were inserted under standard local anesthetic (Citanest 10 mg/ml) which was active for 2-3 hours, leaving enough time for the experiments. The pins were inserted at the femur (lateral aspect of femoral epicondyle), tibia (anterior lateral aspect of tibial condyle) and calcaneus (posterior lateral aspect). Femoral data were only available for subjects 1, 3 and 5, since the femur pin became loose for subjects 2 and 4 during testing. Thus femur pin data of all trials should be interpreted with caution. The subjects gave their informed consent to participate in the study and performed three heel-toe running trials with a speed of between 2.5 to 3.0 m/s. Trials were repeated if the subjects did not land with their right foot on the force plate and/or if they made an obvious gait pattern modification in order to hit the force plate.

The tests were performed with a running shoe (Adidas Equipment Cushioning) with a rearfoot shoe sole modification (straight shoe sole, Shore A45). This sole modification was regarded as a neutral shoe sole design with no extremes of lateral heel flare or midsole hardness (chapter 5). The shoe markers were placed at the posterior lateral aspect of the heel counter, at the dorsum of the foot (lateral cuneiform), and at the lateral tuberosity of the fifth metatarsal. Three high-speed cine cameras (LOCAM) were placed around a force platform (KISTLER) mounted flush to the runway. The camera speed was set at 200 Hz and three LED's, triggered by a threshold detector connected to the force plate, were used to synchronize the cameras. A calibration frame with 6 control points (0.5 x 0.5 x 0.5 m^3) was used for the three-dimensional data reconstruction.

7.2.2 Data Analysis of Tibiocalcanal and Tibiofemoral Rotations

KineMat, a set of programs written in MATLAB™, was adapted from Reinschmidt and van den Bogert (1997d) for the specific needs of this investigation. The programs served to reconstruct the three-dimensional marker positions and to calculate the relative segmental movements. The barefoot standing trial was used as the neutral position to define the segment-fixed calcaneus and tibia coordinate systems. The rotations between the segments were calculated as Cardanic angles for the stance phase of all test conditions using a joint coordinate system (JCS) at the knee and ankle joint complexes. The defined sequence of rotations for the knee joint was flexion/extension about a medio-lateral femur fixed axis, knee ab/adduction about a floating axis and internal/external rotation about a proximal-distal tibia fixed axis (Grood and Suntay, 1983). The sequence at the ankle joint complex was plantar/dorsiflexion about a tibia fixed medio-lateral axis, foot ab/adduction about the floating axis, and in/eversion about the antero-posterior axis of the foot (Cole et al., 1993). Tibial rotation was calculated using the sequence: tibial rotation about a tibia fixed proximal-distal axis, in/eversion about the floating axis, and plantar/dorsiflexion about a calcaneus fixed medio-lateral axis (Nigg et al., 1993).

The accuracy of the spatial reconstruction between two marker triads was determined (i) based on the residuals of the DLT equations averaged over the entire stance phase for all markers and (ii) based on the deviations of the inter-marker distances of the same trials. The mean error based on DLT residuals was found to be in the order of ±4° which included noise error and lens distortion error. The mean error based on marker distances (RMS) was found to be ±1.0° including noise error only. Thus, for the present study, a realistic estimation of the error was likely to be between the two errors given above. The error of the shoe data was about ±1.0° higher than that at the bone, because it included inaccuracies of different standing trials with different shoes.

7.2.3 Definitions of Variables

The variables used in this study are summarized and defined in Tables 7-1 and 7-2. Total eversion is equivalent to the range of motion in eversion of the calcaneus relative to the tibia between touchdown and maximum, based on bone marker measurements. Analogous definitions to those in table 7-1 were used for the shoe markers and the skeletal markers at the tibia and femur. Thus, total shoe eversion was defined as

$\Delta\beta_{max/shoe}$, total internal tibial rotation (tibia relative to calcaneus) as $\Delta\rho_{max}$, total internal knee rotation (tibia relative to femur) as $\Delta\psi_{max}$. These definitions follow the nomenclature of previous investigations on the kinematics of running (Nigg et al., 1986; Lafortune et al., 1994; Reinschmidt et al., 1997b).

VARIABLE	SYMBOL	DEFINITION
in/eversion at touchdown	β_o	In/eversion position of calcaneus relative to tibia at touchdown
maximum eversion	β_{max}	Maximum eversion position of calcaneus relative to tibia during ground contact
total eversion	$\Delta\beta_{max}$	Total eversion of the calcaneus relative to the tibia between touchdown and maximum (= β_{max} - β_o)

Table 7-1: Definitions of bone marker variables. Analogous definitions were used for the shoe markers and the skeletal markers at the tibia and femur.

The coupling coefficient describes the movement transfer from one segment to another. This coefficient has been used in previous *in-vitro* investigations (Olerud et al., 1985; Lundberg, 1989; Hintermann, 1994) as well as *in-vivo* studies (Nigg et al., 1993; McClay et al., 1997). It was defined after Nigg et al. (1993) as the transfer from A to B:

$$T_{AB} = \frac{\Delta\varphi(\text{output})}{\Delta\varphi(\text{input})}$$

where the rotation ($\Delta\varphi$) at A is considered the input (reason), and that at B the output (effect). As an example, it was expected that the input at the calcaneus would cause some rotational effect at the tibia (output), thus that the coefficient T_{CT} would become less than 1 (Table 7-2). When the coefficient would remain the same between two running phases it would indicate a "stable" coupling, when the coefficient would decrease between phases it would indicate that some other input (i.e. a rotation about another axis, muscular forces) may have taken place. At the knee, the rotation of interest was tibial rotation relative to femoral rotation (JCS after Grood and Suntay, 1983), thus the rotation about the tibia fixed longitudinal axis. Thus, if there was no rotation of the tibia relative to the femur (i.e. high coupling) the coupling coefficient was expected to be 0.

VARIABLE	SYMBOL	DEFINITION
coupling from shoe to calcaneus	T_{SC}	$\dfrac{\Delta\beta\, max}{\Delta\beta\, max/shoe}$
coupling from calcaneus to tibia	T_{CT}	$\dfrac{\Delta\rho\, max}{\Delta\beta\, max}$
coupling from tibia to femur	T_{TF}	$\dfrac{\Delta\psi\, max}{\Delta\rho\, max}$

Table 7-2: Definition of coupling coefficients used in this study.

The stance phase in running can be divided in a loading phase (heel strike to midstance) and an unloading phase (midstance to take-off). *In-vitro* studies (Inman, 1976; Olerud et al., 1985; Siegler et al., 1988; Hintermann, 1994) have shown that movement coupling may not be identical in these two phases. Thus, the coefficients of Table 7-2 (defined for the loading phase) were computed also for the unloading phase. For this purpose the take-off variables β_{TO} , ρ_{TO} and ψ_{TO} were used. Total inversion was then defined as $\Delta\beta_{max}$ - β_{TO} and in analogy to that the variables of the tibia and femur. Thus, the coupling coefficients were computed for the loading phase ($T_{SC\text{-}1}$, $T_{CT\text{-}1}$, $T_{TF\text{-}1}$) and the unloading phase ($T_{SC\text{-}2}$, $T_{CT\text{-}2}$, $T_{TF\text{-}2}$) respectively. The coupling results of this study were illustrated with angle-angle diagrams.

7.3 Results

7.3.1 Coupling Coefficients and Repeatability of Test Movements

The results of three running trials of one subject are presented in Figure 7-1. Generally, the shape of the curves showing movement coupling between shoe and calcaneus and between calcaneus and tibia were found to be very similar.

Most coupling coefficients (Table 7-3) were consistent within each condition, illustrated by the small standard deviation, but varied considerably between segments and between subjects.

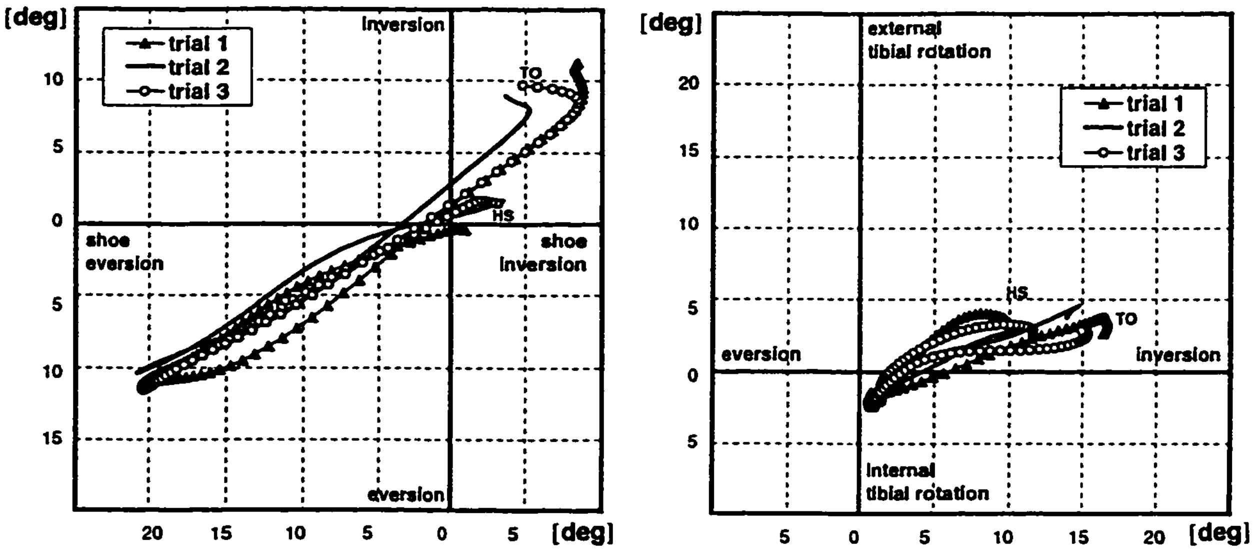


Figure 7-1: Three running trials of subject 1 illustrating the repeatability of the testing procedure. Movement coupling is presented between shoe and calcaneus (left) and between calcaneus and tibia (right). Subject 1, straight shoe; HS = heel strike; TO = takeoff.

7.3.2 Coupling between Shoe and Calcaneus

Movement coupling between shoe and calcaneus showed two distinct phases during the ground contact of running (Figure 7-2). From heel strike to midstance the shoe and the calcaneus moved into (shoe/calcaneus) eversion, from midstance to take-off into (shoe/ calcaneus) inversion. Coupling differed considerably between the subjects. During the loading phase shoe eversion was about twice that of calcaneal eversion for subjects 1, 3 and 5, with the coupling coefficient being around 0.5 (Table 7-3). The coupling coefficient for subjects 2 and 4 was between 0.7 and 0.8. The average coupling coefficient for the loading phase was 0.61 and was increased by 34% in the unloading phase (0.83). Shoe eversion relative to the tibia was between 12° and 23° and calcaneus eversion between 7° and 12° (Figure 7-2).

Phase	Coefficient	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Mean SD
Loading	$T_{SC\,1}$	0.52 ± 0.03	0.68 ± 0.02	0.57 ± 0.11	0.80 ± 0.15	0.49 ± 0.12	0.61 ± 0.13
	$T_{CT\,1}$	0.47 ± 0.07	0.35 ± 0.03	0.43 ± 0.15	0.68 ± 0.26	0.96 ± 0.51	0.58 ± 0.24
	$T_{TF\text{-}1}$	1.71 ± 0.45		3.44 ± 1.09		1.09 ± 0.38	2.08 ± 1.22
Unloading	$T_{SC\,2}$	0.80 ± 0.02	0.86 ± 0.03	0.86 ± 0.06	0.98 ± 0.09	0.66 ± 0.07	0.83 ± 0.11
	$T_{CT\,2}$	0.36 ± 0.02	0.24 ± 0.03	0.29 ± 0.05	0.45 ± 0.11	0.95 ± 0.35	0.46 ± 0.29
	$T_{TF\text{-}2}$	1.71 ± 0.34		2.13 ± 0.15		0.97 ± 0.26	1.60 ± 0.59

Table 7-3: Coupling coefficients from shoe to calcaneus (T_{SC}), from the calcaneus to the tibia (T_{CT}) and from the tibia to the femur (T_{TF}) for the loading and unloading phase.

7.3.3 Coupling between Calcaneus and Tibia

Movement coupling from the calcaneus to the tibia showed loading and unloading phases similar to the shoe-calcaneus movement coupling. From heel strike to midstance the calcaneus everted and the tibia rotated internally, from midstance to take-off the calcaneus inverted and the tibia rotated externally. The coupling coefficients between calcaneus and tibia ranged between 0.35 (subject 2) to 0.96 (subject 5; Table 7-3). The coupling coefficients decreased in the unloading phase compared to the loading phase for all subjects, except subject 5 where it remained unchanged. Additionally, the coupling coefficient between calcaneus and tibia was smaller than the coefficient between shoe and

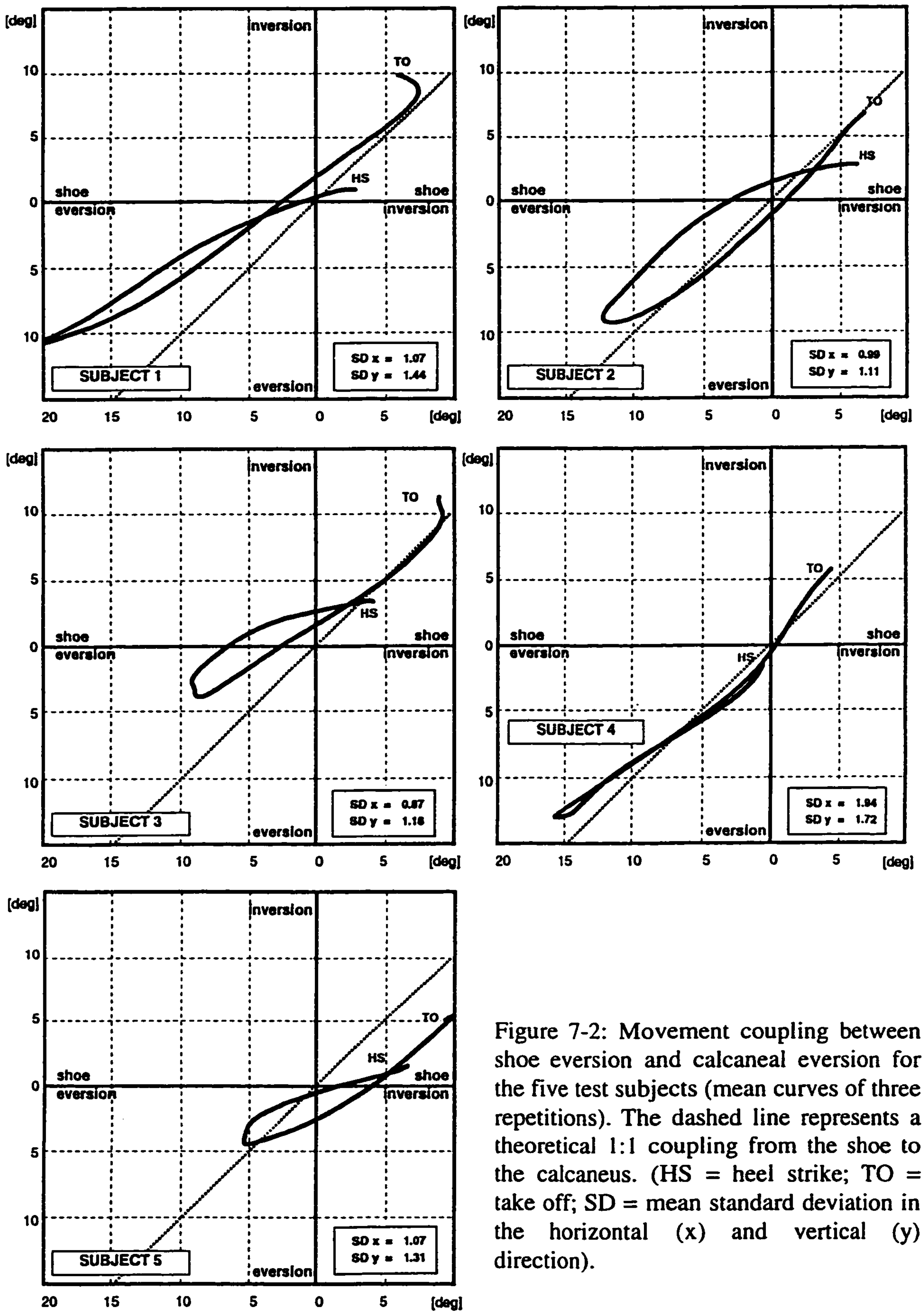
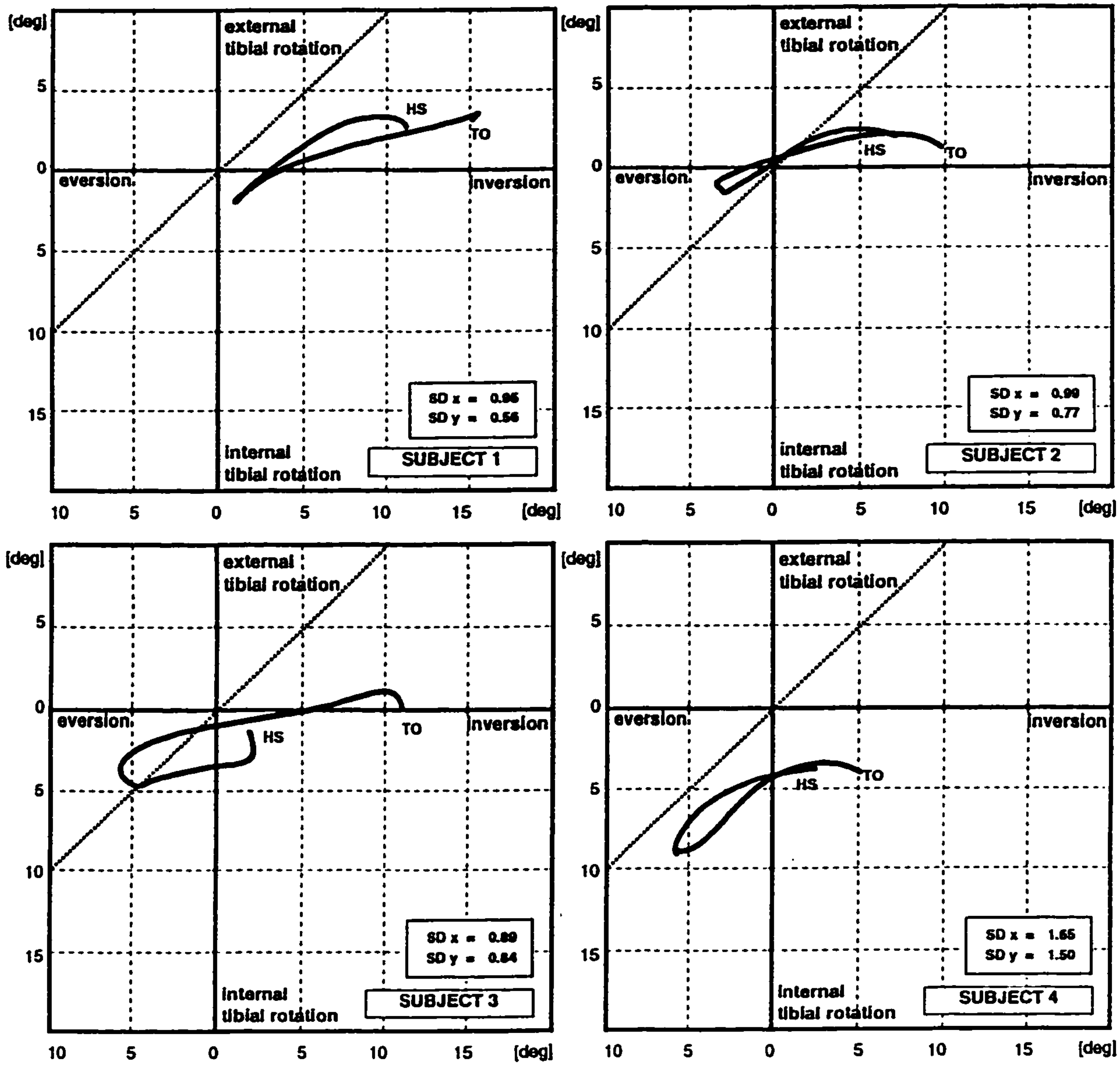


Figure 7-2: Movement coupling between shoe eversion and calcaneal eversion for the five test subjects (mean curves of three repetitions). The dashed line represents a theoretical 1:1 coupling from the shoe to the calcaneus. (HS = heel strike; TO = take off; SD = mean standard deviation in the horizontal (x) and vertical (y) direction).



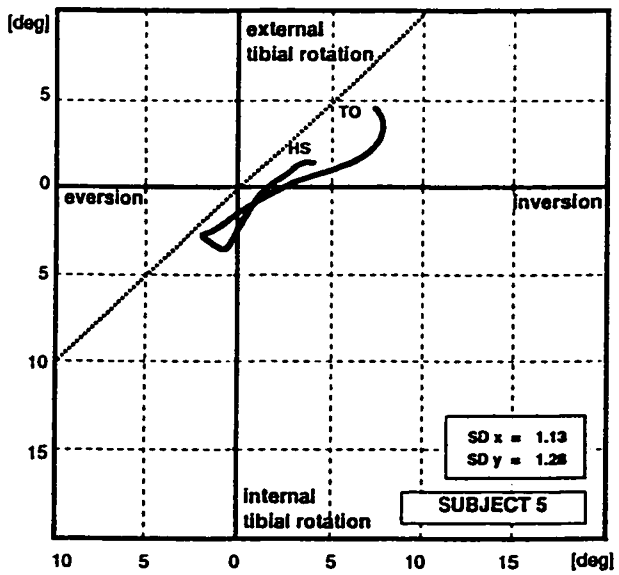


Figure 7-3: Movement coupling between calcaneal eversion and tibial rotation for the five test subjects (mean curves of three repetitions). The dashed line represents a theoretical 1:1 coupling from the calcaneus to the tibia. (HS = heel strike, TO = take off; SD= mean standard deviation in the horizontal (x) and vertical (y) direction).

calcaneus for all subjects but one (subject 5). Calcaneal eversion relative to the tibia was found between 6° and 11° and internal tibial rotation relative to the foot was between 3° and 6° (Figure 7-3).

7.3.4 Coupling between Tibia and Femur

All three subjects rotated their tibia internally relative to the femur during the loading phase and externally during the unloading phase. Internal and external movements were of the same order of magnitude (between 5° and 12°; Figure 7-4). Since knee rotation was defined about a proximal-distal tibia fixed axis (Grood and Suntay, 1983), the movement of the tibia relative to the femur was presented rather than an angle-angle diagram (Figure 7-4). The coupling coefficient between tibia and femur was about the same for the loading and the unloading phase (Table 7-3). However, the intraindividual differences were of about the same order of magnitude as the tibia-femur coupling coefficients.

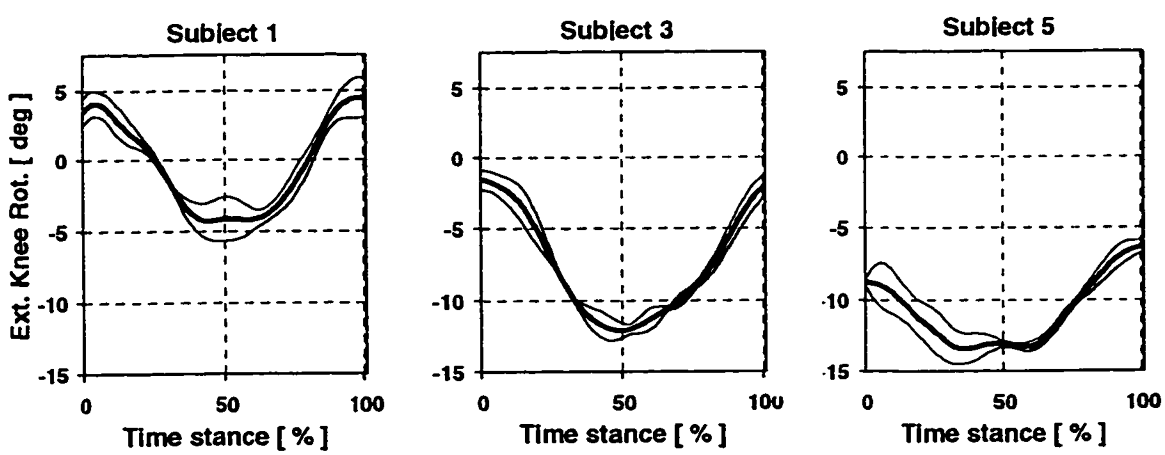


Figure 7-4: Knee rotation during the stance phase of running for subjects 1, 3 and 5 (mean curves and SD of three trials). Knee rotation corresponds to tibial rotation relative to femoral rotation.

7.4 Discussion

7.4.1 Coupling between Shoe and Calcaneus

The coupling between shoe and calcaneus showed two distinct phases, one from heel strike to midstance, and one from midstance to take-off with an increase in coupling in the second half (Figure 7-2). Additionally, the first phase might be divided in two sub-phases, because immediately after touchdown shoe eversion increased more than that of the calcaneus in all but subject 4. Thus, during landing the shoe and the calcaneus were found with lower coupling coefficients in four of the five subjects. At touchdown lower coupling coefficients are welcomed, because higher coefficients would increase eversion of the calcaneus which could increase the risk of injury.

During the second half of the first phase coupling increased in all subjects (except subject 4, who already had a high coupling) compared to the first half. When attempting to reduce excessive eversion of the calcaneus, a high shoe-calcaneus coupling coefficient seems advantageous (in the second half), provided the shoe has the capacity to reduce eversion. One such shoe construction which has often be attributed to reduce eversion is the heel counter. However, van Gheluwe et al. (1995) reported that with rigid heel counters more relative movement between the shoe and the foot should be expected, and that a rigid heel counter would decrease shoe-calcaneus coupling. Hence, the effect of heel counters on calcaneus movement may have to be revised using the shoe-calcaneus coupling coefficient as a possible indicator. Furthermore, the large subject differences in shoe-calcaneus coupling of the present study indicated that coupling between the shoe and calcaneus is subject dependent. In Figure 7-2 subjects 1 and 4 showed a "stable" coupling between heel strike and takeoff, whereas all other subjects varied their coupling between the loading and unloading phase.

From midstance to take-off all subjects showed a high shoe-calcaneus c oupling coefficientand the differences between the subjects became smaller. During this phase high coupling may be advantageous in order to reduce relative movements of the foot inside the shoe. It is suggested, that relative movements (i.e. slipping inside the shoe) would be a disadvantage during the propulsion phase of running.

It may be argued that shoe eversion/inversion was overestimated in this study, since two shoe markers were set at the dorsum of the foot recording midfoot eversion rather than pure heel eversion. It has been estimated that this marker setting increased maximum eversion by about 2-4 degrees compared to shoe markers set at the heel only (chapter 4). However, given the large shoe movements of this study, it is speculated that the general shape of the present coupling curve would change only minimally, when using shoe/heel markers.

7.4.2 Coupling between Calcaneus and Tibia

The results of this *in-vivo* study showed that the coupling between calcaneus and tibia (i) was generally less than 1 and (ii) changed throughout the stance phase of running. These findings are in contrast to the "mitered joint" or "universal joint" model, which has been used to visualize the coupling between the foot and shank (Inman, 1976) and which implies a constant coupling. Thus, the present study confirms previous *in-vitro* and *in-vivo* studies that the ankle joint complex does generally not work like a universal joint (van Langelaan, 1983; Olerud et al., 1985; Engsberg et al., 1987; Siegler, 1988; Lundberg, 1989; Murphy, 1993; Hintermann, 1994).

Furthermore, the present study showed that the coupling coefficients between eversion and internal tibial rotation were higher than between inversion and external tibial rotation. This is in contrast to the results of the *in-vitro* study of Hintermann (1994), with low eversion-internal tibial rotation coupling and high inversion-external tibial rotation coupling. Hintermann (1994) identified vertical loading as one factor which increased eversion of the foot. Thus, it is possible that the low eversion-internal tibial rotation coupling of Hintermann's study (1994) may have been due to the low vertical loading and that the coupling results of the present study are due to the different vertical loads applied *in-vivo*.

At heel strike and take-off subject 3 showed a few degrees of tibial rotation without any rotation at the calcaneus. This results suggested that other movements than eversion/inversion provided an input which may have caused the tibia to rotate. This argument is supported by Engsberg et al. (1987) who concluded that although eversion is the major rotational component, ab/adduction or plantar/dorsiflexion may contribute to the coupling mechanisms at the ankle. Thus, future studies should analyze movement coupling at the ankle joint complex with a 3D input and a 3D output.

Tibiocalcaneal motion during running has been measured using skin and shoe mounted markers. Nigg et al. (1993) reported a mean coupling coefficient of 0.76 ± 0.16 using external markers. McClay et al., (1997) found a mean coupling coefficient of 0.81 ± 0.29 for a pronator group and 0.65 ± 0.26 for a normal group. These values corresponded well with the present study where a mean coupling coefficient of 0.58 ± 0.24 was found (Table 7-3). The small discrepancies between the previous study and the present one may be the result of overestimations due to external marker settings at the dorsum of the foot.

In conclusion, coupling between the calcaneus and tibia is far more complex than a simple mitered or universal joint model. The coupling mechanism between eversion and internal tibial rotation is only one coupling component. In future studies the coupling at the ankle joint complex should be studied with 3D input and output variables for each subject individually. The recent study by McClay et al., (1997) showing tibial rotation differences between excessive pronators and normal subjects, showed the possibility in this respect to describe differences between categories of runners using the coupling coefficient as an indicator.

7.4.3 Coupling between Tibia and Femur

The coupling coefficients between tibia and femur were all larger than 0, and the total tibial rotation relative to the femur ($\Delta\psi max$) was larger (between 5° and 12°) than the total tibial rotation relative to the calcaneus ($\Delta\rho max$; between 3° and 6°). This results suggests that during the stance phase tibial rotation may be more constrained at the ankle than at the knee. Furthermore, previous bone pin studies on walking have shown, that the femur rotates externally relative to the tibia from touchdown until midstance (McClay, 1990; Lafortune et al., 1994). This external femoral rotation may be initiated by a pelvic rotation and/or muscular activity of the thigh and hip muscles. Thus, it is suggested that the input to the coupling at the knee may not be from distal to proximal, but at least partially from proximal to distal.

The coupling coefficients remained almost constant between the loading and the unloading phase (Table 7-3). Thus, the coupling at the knee may be more stable than at the ankle. It is likely to be supported by muscular and ligamentous structures which maintain the integrity of knee joint function (Lafortune et al.,1994). Lafortune et al. (1994) found that during walking with varus-wedged shoes compared to normal shoes tibiofemoral rotation revealed only minimal differences in midstance. However, from midstance to takeoff Lafortune et al. (1994) reported internal tibiofemoral rotation, whereas in the present study all three subjects showed external tibiofemoral rotation. The reason for this discrepancy is currently unknown.

7.5 Summary and Conclusion

The present *in-vivo* study related to movement coupling in the lower extremity during the stance phase of slow running (2.5m/s to 3.0m/s) showed the following findings:

- Movement coupling was observed from the shoe to the calcaneus, from the calcaneus to the tibia, and from the tibia to the femur. Between the segments coupling was found to take place in distinct phases. Coupling between the shoe and the calcaneus showed low coupling coefficients during the loading phase between heel strike and midstance and increased coupling coefficients during the unloading phase from midstance to take-off. It was suggested that future studies related to shoe development may use shoe-calcaneus coupling coefficients as an indicator to test the effects of shoe constructions during the loading and unloading phase.

• The coupling coefficient between the calcaneus and the tibia decreased from the loading phase to the unloading phase of running and indicated that the coupling input is at the calcaneus, and the output at the tibia (distal to proximal). Thus, the results of the present study (using bone markers) supported previous *in-vitro* investigations as well as *in-vivo* measurements using external markers. Consequently, the present study confirmed that the ankle joint complex does not work like a universal joint.

• Coupling between the tibia and femur remained almost constant between the loading and unloading phase which suggested that bony, muscular and ligamentous structures at the knee were able to maintain the integrity of knee joint function. It was suggested that the input to coupling at the knee may take place (at least partially) from proximal to distal which would be the opposite to that at the ankle joint complex.

• Movement coupling at all three examined levels (shoe-calcaneus, calcaneus-tibia, and tibia femur) showed a large intersubject variability which indicated that movement coupling at the lower extremity is subject dependent.

8. IMPLICATIONS FOR FUTURE STUDIES

The purpose of this chapter is to discuss implications of the results of this investigation and to provide suggestions for future studies. The issues to be discussed are related to: (1) foot orthoses, (2) shoe sole modifications, (3) movement coupling, (4) adaptation, (5) running injuries, and a summary (6).

8.1 Implications and Suggestions for Foot Orthoses

- *The effects of medially placed foot orthoses on skeletal movements were found to be small and unsystematic (chapter 4). Thus, an induced mechanical change (e.g. an applied orthosis) did not produce the same skeletal reaction in all five subjects.*

These study results led to the suggestion that orthotic effects may be mechanical as well as proprioceptive. There is some indirect evidence that supports this suggestion. Feuerbach et al. (1994) argued that orthoses may increase the afferent feedback from cutaneous receptors which may lead to decreased eversion because of muscular contraction of inverting muscles. Indirect evidence from cadaver studies showed that when pulling forces are applied on m. tibialis posterior, eversion is reduced and the movements at the midfoot joints are changed (Müller et al., 1997; Stähelin et al., 1997). Further support is provided by Fromme and co-workers (1997) who found that pronation increased with increasing fatigue; indicating that muscular activity may play an important role in the control of eversion during the stance phase of running. In a recent study Nigg et al. (1998) found substantial differences in the individual reactions to soft and hard inserts. Thus, there is some evidence that orthotic effects may have to be looked at as a combination of mechanical and proprioceptive effects that may lead to muscular responses. Further research may be necessary to improve the understanding of the interaction between proprioception and muscular activity as well as between proprioception and kinematics during running.

- *Orthotic effects on tibialcalcaneal movements were not apparent in some subjects, but in the order of 1° to 2° (eversion) and 1° to 4° (tibial rotation) in others. Thus, orthoses may produce unsystematic subtle differences in some individuals.*

This suggests, that in contrast to the present study the administration of foot orthoses for running may have to be tuned for each individual foot. However, whereas guidelines for the administration of orthoses for everyday activities, e.g. walking, have been established for severe foot deformities (McPoil et al., 1990; Wu, 1990; Condie et al., 1993; Philps, 1995; Donatelli et al., 1996), guidelines for the administration of orthoses for running shoes are not well established (Nigg et al., 1998). On the other hand, McClay et al. (1997) described kinematic differences between subjects denoted as pronators and normals. Thus, further research should focus on individual orthotic effects as well as group effects with respect to the administration of foot orthoses for running.

In a recent investigation, Grau (1997) reported eversion differences between subjects with and without Achilles tendon pain of 1° to 2° only. The comparison was based on a large study group of over 300 runners (150 with, 150 without pain). This leads to the question whether differences as small as 1° to 2° may functionally be relevant or not with respect to the occurrence of pain and injury. One can speculate that such small differences are not relevant. However, presently there is no evidence to support this speculation.

- *The position of orthoses inside running shoes may has to be revised.*

The small effects of foot orthoses on calcaneal movements lead to the suggestion that mid- and forefoot movements may be more important to the understanding of orthotic effects than that of the calcaneus. Traditionally, most orthoses for runners are placed medially and posteriorly to support the stustentaculum tali and/or medially and anteriorly to support the longitudinal foot arch. In contrast to rearfoot orthoses, the effects of forefoot orthoses during running are currently mostly unknown, except for rare investigations (i.e. Kogler et al., 1995). Furthermore, Hicks (1953) pointed out that the "twist in the forefoot may be primary and the tilt of the whole foot into supination or pronation is secondary". Thus, it may be speculated that there exists a relationship or coupling between forefoot and tibia during the stance phase of running that may be influenced by forefoot orthoses. Thus, future studies should focus on the effects of forefoot orthoses on the kinematics of the shank during running.

- *The role of the calcaneus during running may has to be reconsidered.*

The most common research method to study rearfoot kinematics in running is to use markers fixed to the heel counter of the shoe and to the shank. The present study showed that shoe markers overestimated the movements of the underlying calcaneus, but also that the two movements were related to each other. Orthotic effects on the calcaneus (as well as on the shoe) were found to be small and unsystematic. Thus, it may be speculated that the calcaneus may not be the only relevant bone to be assessed. Although the calcaneus may be important during the initial landing phase of heel-toe running, it may be less important during the stance phase and take-off phase. However, presently there is no evidence which can support this speculation.

8.2 Implications and Suggestions for Shoe Sole Modifications

- *Kinematic effects of shoe modifications were found to be small and unsystematic across the five test subjects (chapter 5). It was suggested that the results depended on several factors such as adaptations and the use of local anesthesia (see subchapter 8.4), and/or the use of the test shoes as discussed below.*

It was expected that shoe sole constructions would influence the kinematics of the calcaneus and tibia during running. The expectation was based on previous studies which have shown that prominent lateral heel flares cause an increase in initial eversion velocity and round shoe soles cause a decrease (Clarke et al., 1983; Nigg et al., 1987a; 1987b; DeWit et al., 1995). However, the present study results did not support this expectation. On the other hand, the shoe constructions of the present study had some features which may provide some possible explanations. All test shoes were constructed with the same hard outer sole material of Shore A65. But the structural differences between the flared, straight and round sole may have produced different effects. For example, the edge of the flared sole may have deformed during touchdown, whereas the round sole may have kept its shape. Thus, it is speculated that the round sole construction may have acted as a ramp which forced the shoe and the calcaneus into eversion. One other investigation related to the effects of round shoe soles did not find the same effect, possibly because a softer sole material was used (Nigg et al., 1987a). Thus, future studies related to effects of shoe sole modifications may have to take great care in designing the shoe sole geometry and shoe sole hardness.

One further explanation is related to the lateral cutout on the heel counter that was necessary to prevent impingement with the calcaneal bone pin. It was suggested that this cutout may have reduced the fit inside the shoe, thereby influencing the test results (chapters 4-6). Consequently, future studies with calcaneal bone pins may have to change the design of the shoe and/or the mounting of the marker triad on the bone pin to ensure that the shoe upper remains intact.

8.3 Implications and Suggestions with Respect to Movement Coupling

- *Movement coupling between shoe and calcaneus showed distinct phases of higher and lower coupling. It was suggested to use shoe-calcaneus coupling as an indicator for the development of future shoe designs (chapter 7).*

Immediately after heel-strike shoe eversion increased more than that of the calcaneus in four of the five subjects. Thus, during landing the shoe and the calcaneus were found to show low coupling. This was considered as advantageous, because high coupling could lead to excessive eversion. After this initial phase, coupling was found to increase until midstance in all subjects (except in one subject who already had high coupling). In this phase high coupling is welcomed, provided the shoe has a built-in capacity to reduce eversion. During the last phase, from midstance to take-off, all subjects of this study showed a high coupling which was considered as an advantage in order to keep relative movements of the foot inside the shoe to a minimum. Therefore, movement coupling may be considered to be used as an indicator to compare the effects of different shoe designs in different phases of the ground contact of running.

- *Movement coupling between calcaneus and tibia changed throughout the stance phase of running. This finding is in contrast to the mitered joint or universal joint model.*

Movement coupling between calcaneus eversion and internal tibial rotation was found to be higher than between inversion and external tibial rotation. This finding did not support the results of the *in-vitro* study of Hintermann (1994), with low eversion-internal tibial rotation coupling and high inversion-external tibial rotation coupling. However, Hintermann (1994) identified vertical loading as one factor which increased eversion of the foot. Thus, the discrepancy between the two studies may be explained as a result of different loads applied. However, the results of Hintermann's study as well as of the present study are in contrast to the concept of a mitered joint or universal joint which implies that the coupling remains constant. Further support is provided by previous *in-vitro* and *in-vivo* studies (Olerud et al., 1985; Engsberg et al., 1987; Lundberg, 1989) which suggests that the ankle joint complex does generally not work like a universal joint.

8.4 Implications and Suggestions with Respect to Adaptations

- *During testing, further adaptations of the test subjects may have taken place which could not be accounted for in this study.*

In order to test the effects of bone pin insertion (and local anesthesia) Reinschmidt et al. (1997a), using the same test subjects at the same test date, compared three trials with and without bone pins in subjects 2 and 4 (using skin mounted markers). It was concluded that the pre/post-operative knee and ankle joint rotations showed graphs which were similar in shape and magnitude, maximum differences being 2°. Thus, whether or not proprioceptive and/or adaptive effects to the shoe sole modifications took place in this study cannot be answered conclusively. However, there are a number of arguments that may have to be considered in future studies:

Locomotion is generally thought to be controlled by a central pattern generator (CPG) that issues a basic locomotor-like pattern modulated by input from supraspinal centers and motion related feedback (Patla, 1991a; Zernicke and Smith, 1996). Feedback in this study may have been absent or changed because of the local anesthesia used at the calcaneus, tibia, and femur. Consequently, possible adaptations and modulations of the locomotor-like pattern to different shoe conditions may not have taken place. However, the literature related to adaptation towards shoe modifications is controversial. Frederick (1986) concluded that "shoes may elicit a kinematic adaptation that in turn has secondary consequences on kinetics and on injury and performance". Reinisch et al. (1991) and Hartwig et al. (1997) concluded that there may be different categories of runners, those who do adapt, those who do not and possibly even a third group who adapt wrongly to a given shoe condition. This is in contrast to Milani et al. (1997) who concluded that there may be surprisingly good relationships between biomechanical variables and perception of subjects running with shoes of different sole hardness. Robbins et al. (1987; 1990) on the other hand concluded that, in barefoot, weight bearing adaptations to varied ground conditions can take place that are otherwise insulated by the soft soles of modern running shoes. Finally, Konradsen et al (1993) found that a missing afferent input (due to local anesthesia) from ligaments and joint capsules might be replaced by afferent information from active calf muscles unimpaired by the local anesthesia. Thus, the controversial arguments with respect to adaptation are difficult to summarize. Patla (1991b) concluded that "it is evident that the sensory modalities are primed during locomotion to provide appropriate reactive and proactive modulations of gait patterns for safe navigation through varied environments". However, how these modulations and/or adaptations take place during running appears not to be very well understood at the present time.

8.5 Suggestions with Respect to Running Injuries

- *Although the present study is not directly related to running injuries, the results in chapters 4-7 allow the following suggestions:*

Excessive tibial rotation has been associated with various running injuries (Clement et al., 1981; van Mechelen, 1992). The present study showed that during midstance internal tibial rotation changed to external tibial rotation. It may be argued that this movement direction change may put an eccentric load onto the shank muscles. It has been suggested that eccentric contractions (Winter et al., 1983; Faulkner et al., 1984) as well as the timing between the segmental movements (Hamill et al., 1992; Stergiou N. et al., 1997; McClay et al, 1997) may be important with respect to injury mechanisms related to muscles at the lower extremity during running. Furthermore, eccentric muscle forces are larger than concentric muscle forces as a result of the force-velocity relationship of skeletal muscles. Thus, it is speculated that running injuries may be associated with eccentric loading of shank muscles.

As discussed above (8.4), a running movement pattern controlled by the central pattern generator (CPG) may receive proprioceptive inputs from shoes and orthoses that may cause modifications of muscular activities. It is speculated that such muscular activity changes may change muscular loading patterns. Some indirect evidence for this possibility is provided by Fromme et al. (1997) who showed that pronation during running increased over time with increasing fatigue. Furthermore, Feltner et al. (1994) demonstrated that strength training can reduce total eversion significantly after eight weeks of training. Thus, although the present study did not show systematic kinematic effects, it may be argued that muscular activity changes may have taken place which were not measured in this study. It can be further speculated that such muscular activity changes may eventually lead to running injuries (Figure 8-1).

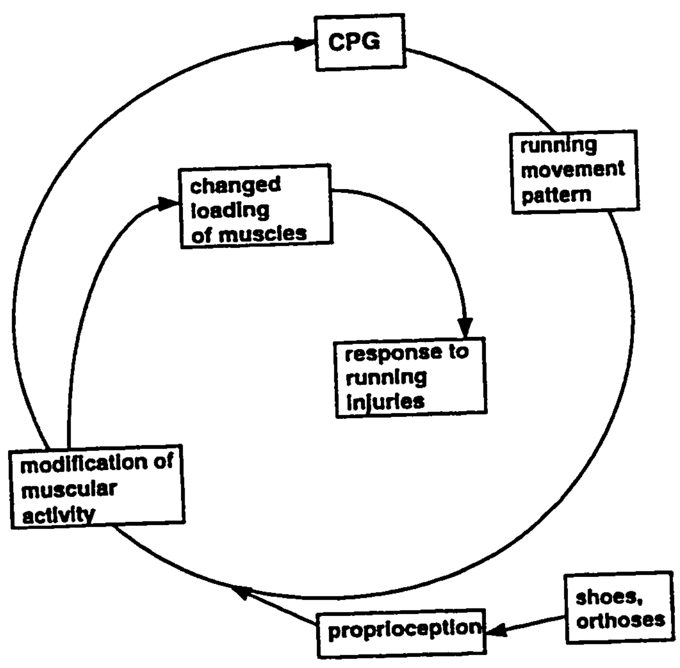


Figure 8-1: Schematic diagram of how the input from the central pattern generator (CPG) may be modified and may lead to injuries. Adapted from Patla (1991b).

8.6 Summary

The discussion of the implications of this study's results with respect to future studies show various possibilities for further improvements. In short, foot orthoses could be administered individually (and/or in groups) using newly designed guidelines, studies on shoe modifications may focus on the effects of round shoe sole designs, and movement coupling studies may try to focus on input factors that influence movement coupling at the knee. Investigations related to adaptations may find a way to describe how movement patterns adapt to external inputs, such as those from shoe modifications. Thus, the possibility provided in this investigation to measure kinematic effects on the bone level led to a number of further challenging research questions. However, whatever the goals for future studies may be, it should be kept in mind that one deals with subjects and/or patients and that *"Thoughtful intervention implies correcting what one can correct, accepting what cannot be changed and having the wisdom to know the difference."* (Patla, 1991b).

9. SUMMARY AND CONCLUSION

Biomechanical factors which have been associated with specific running injuries include excessive eversion, excessive eversion velocity, and excessive tibial rotation. A coupling mechanism between the foot and the leg (calcaneus and tibia) has been shown *in-vivo* using shoe and skin mounted markers. However, skin and shoe movement artefacts mask skeletal kinematics and conclusions from studies using shoe and/or skin markers, suggesting that specific shoe sole modifications and foot orthoses can be used to control/reduce foot eversion and tibial rotation may be incorrect. Therefore, the purpose of this study was to quantify (i) the effects of medial foot orthoses and (ii) the effects of lateral heel flares on skeletal movements during the stance phase of running, to compare (iii) barefoot running versus shod running, and to describe (iv) the movement coupling between selected segments of the lower extremity.

Intracortical bone pins with reflective markers triads were inserted under standard local anesthesia into the calcaneus, tibia and femur of five healthy male subjects (age 28.6 ± 4.3 years). The subjects were injury free at the time of testing and had no injury history that may have resulted in abnormal gait. The subjects ran barefoot, with a normal shoe, and with five shoe modifications. Shoe sole modifications were tested with lateral heel flares of 25° (flared), 0° (straight), and a rounded sole. The effect of medial foot orthoses was tested with two 1cm thick cork supports, one anteriorly to support the foot arch and one posteriorly to support the sustentaculum tali; the standard insert had no medial support. The trials were recorded during the stance phase using three high speed cine cameras operating at 200Hz. The barefoot standing trial was used to define the neutral position and the corresponding segment-fixed calcaneal, tibial and femoral coordinate systems. Tibiocalcaneal and tibiofemoral movements were calculated for the stance phase using three-dimensional marker reconstruction and a joint coordinate system approach. Test variables were defined at touchdown, at movement maximum, for the total range of movement and the maximum movement velocity. Test results were discussed with respect to orthotic and shoe sole effects, barefoot versus shod running, as well as movement coupling.

Effects of medial foot orthoses: Medially placed foot orthoses did not substantially change tibiocalcaneal movement patterns during running. Orthotic effects on calcaneal eversion and tibial rotations were found to be small and unsystematic over all subjects. Differences between the subjects were significantly larger (up to 10°) than between the orthotic conditions (1° to 4°; $p<.01$). Significant orthotic effects across subjects were found only for total internal tibial rotation $p<.05$). The results suggest that orthotic effects on skeletal movement patterns are small and subject specific, and that they may be due to mechanical effects as well as proprioceptive adaptations. Based on these results, it is speculated that the calcaneus may not be the only relevant bone to be assessed, and that skeletal movements of the midfoot and forefoot should be included in an analysis for the understanding of orthotic effects.

Effects of lateral heel flares: Shoe sole effects on tibiocalcaneal movements were found to be small and unsystematic. The effects were found to be less than 1°, which was significantly smaller than the differences between the subjects (up to 7°; $p<.01$). Total shoe eversion and shoe eversion velocity were found to be approximately twice as large as bone eversion, the correlation being $r=0.88$ (between total shoe and bone eversion) and $r=0.79$ (between maximum shoe and bone eversion velocity). It was concluded that a number of factors such as individual muscular responses and/or the test shoe construction may have influenced the tibiocalcaneal kinematics during the stance phase of running.

Comparison of barefoot versus shod running: Generally, bone movements during barefoot running were found to be similar to bone movements inside shoes. Only one specific shoe modification (posterior orthosis) showed significant ($p<0.01$) differences to barefoot running (internal tibial rotation, internal tibial rotation velocity and tibiocalcaneal coupling). All other test variable comparisons were not significant. The results suggest that calcaneal and tibial movement patterns do not change substantially when using shoes compared to running barefoot, but differences may occur when extreme shoe modifications are used and that the effects of these interventions are subject specific.

Movement coupling: Movement coupling was observed between various segments of the lower extremity: between the shoe and the calcaneus, between the calcaneus and the tibia, and between the tibia and the femur. Between the segments coupling was found to take place in distinct phases. The shoe-calcaneus coupling coefficient was low in the loading phase (between heel strike and midstance), and high in the unloading phase (between midstance and take-off). The coupling coefficient between the calcaneus and tibia decreased from the loading phase to the unloading phase. The results suggest that the input to calcaneus-tibia coupling was at the calcaneus and the output at the tibia (distal to proximal). Tibia-femur coupling remained almost constant between the loading and unloading phase and the results suggested that the input to coupling was (at least partially) from proximal to distal (which is the opposite to that at the ankle joint complex). Generally, movement coupling varied between heel strike and take-off and was found to be subject and shoe dependent.

There are a number of factors that may have influenced the results, including the test shoe used, proprioception and individual responses. The test shoes had a cutout in the lateral heel counter that was necessary to prevent impingement with the calcaneal bone pin. This cutout may have reduced heel counter rigidity and the fit of the heel inside the shoe. Furthermore, the shoe sole construction of the round shoe may have favored a rolling action resulting in a large maximum eversion. In addition, locally applied anesthesia may have changed the proprioceptive feedback and consequently, possible adaptations to different shoe and orthotic conditions may not have taken place. Finally, for a given task, such as running, there may be various solutions with respect to the rotations occurring at the ankle. Thus, running may be associated with individual movement patterns such that an external input (i.e. shoe modifications) may only have small and varying effects on the kinematics of the calcaneus and tibia.

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IMAGE EVALUATION TEST TARGET (QA-3)

150mm

6"

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