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Abstract

Accurate assessment of motion occurring around the ankle joint turns out to be rather difficult because of the various joints, muscles and other anatomical structures involved. Establishing plantar flexion torque is specifically challenging due to the high forces arising. Herzog (1988) mentions influencing factors like gravitational effects, inertial effects or non-rigidity of the dynamometer arm/shank-foot system, contributing to the discrepancy between measured and resultant joint moments. Not taken into account this leads to unreliable conclusions about the properties of the involved muscles and tendons. The present study tackles the specific shortcoming of non-rigidity of the dynamometer system, while maximal voluntary isometric plantar flexion contractions (MVCs).

Eighteen male volunteers participated in the study. They underwent a predefined test sequence, consisting of multiple MVCs, performed at different measuring positions on the isokinetic dynamometer HUMAC® NORM™ (sampling frequency set to 2000 Hz). Kinematic data were collected simultaneously by eight cameras using the Vicon MX™ system operating at 200 Hz. In addition a pedar®-x insole was placed on the dynamometer footplate, recording pressure distribution data (at 100 Hz) in order to estimate the point of force application.

The main finding of this study is a significant effect of subject positioning on the exerted torque. Ankle joint rotation during MVC seems inevitable to at least some extent. Yet rotation during trials could be reduced by about 50 % due to the tightened positioning of the subjects. In addition, torque levels reached at positions with the participants brought closer to the dynamometer were approximately 14 % to 28 % greater.

Changes in muscle-tendon architecture or inner joint motion throughout the contractions were not monitored in the present study and can therefore only be hypothesized. This drawback should be considered in future research also because substantial joint axes misalignment, despite careful consideration, appeared in the conducted study. The novel measuring approach might be improved via affordable methods dealing with those emerged difficulties.

However, there is evidence for an improved setting of how participants should be adjusted on the dynamometer for “true” plantar flexion torque detection. With considerable preload pressure exerted on the ankle adapter footplate the non rigidity of the dynamometer arm/shank-foot system seems to be partly diminishable.

Zusammenfassung

Bewegungen im Sprunggelenk sind ein komplexes Zusammenspiel von vielen Gelenken, Muskeln und weiteren anatomischen Strukturen. Eine seriöse Untersuchung dieser Bewegungen muss darauf Rücksicht nehmen. Besonders anspruchsvoll ist die Erfassung des Drehmoments für die Plantarflexion, wegen der hohen auftretenden Kräfte. In der Studie von Herzog (1988) werden drei Hauptfaktoren genannt, welche zu Abweichungen zwischen gemessenem und resultierendem Moment bezüglich der jeweiligen Gelenkachse führen. Hierbei sind vor allem Gravitation, Trägheit sowie die Nachgiebigkeit des Systems aus Dynamometerarm und Schaftfuß ausschlaggebend. Werden die auftretenden Unterschiede zwischen gemessenem und resultierendem Drehmoment nicht berücksichtigt, führt dies zu Fehlschlüssen bezüglich der involvierten Muskeln und deren Eigenschaften. In der vorliegenden Studie wurde die Bewegung der Plantarflexion bei maximalen willkürlichen isometrischen Kontraktionen (MVC) untersucht, unter besonderer Berücksichtigung der Elastizität des Dynamometerarm-Schaftfuß Systems.

An der Studie teilgenommen haben achtzehn Probanden. In einem vorgegebenen Messablauf wurden mehrere MVC bei unterschiedlichen Positionen am isokinetischen Dynamometer HUMAC® NORM™ (mit einer Abtastrate von 2000 Hz) ausgeführt. Synchron dazu wurden kinematische Daten mit dem Vicon MX™ System (200 Hz) erfasst. Die Bestimmung des Kraftangriffspunktes erfolgte mit einer an der Fußplatte des Dynamometeradapter angebrachten Druckverteilungsmesssohle (pedar®-x System), welche mit einer Abtastrate von 100 Hz Druckdaten aufzeichnete.

Das wichtigste Ergebnis der durchgeführten Untersuchung ist ein signifikanter Einfluss der Probandenpositionierung auf das erreichte Drehmoment. Bis zu einem gewissen Maß unvermeidbar scheint die Rotation im Sprunggelenk während der MVC. Nichts desto trotz konnte eine Veränderung des Sprunggelenkwinkels während der Kontraktionen durch die engere Positionierung um ca. 50 % verringert werden. Zusätzlich war das erreichte Drehmoment in den Positionen näher am Dynamometer um etwa 14 % bis 28 % höher. Über Veränderungen in der Architektur der Muskel-Sehnen-Einheit bzw. über Bewegungen innerhalb der Gelenke während der MVC können nur Vermutungen angestellt werden; diese wurden in der durchgeführten Studie nicht erfasst. In zukünftigen Untersuchungen könnte das durch geeignete (bildgebende) Verfahren mitberücksichtigt werden. Auch wegen der in vorliegender Studie

aufgetretenen Fehlausrichtung der Sprunggelenkachse mit der Dynamometerachse gilt es eine verbesserte und dennoch kostengünstige Messmethodik zu entwickeln, welche diesem Defizit zuvorkommt und über mögliche Abweichungen in der Achsenausrichtung schnellstmöglich informiert.

Dennoch liegt die Vermutung nahe, dass bei adaptierter Positionierung der Probanden am Dynamometer verlässlichere Werte des „tatsächlichen“ Spitzendrehmoments gemessen werden. Die Fußplatte des Dynamometeradapters wurde dabei zu Beginn der MVC mit beträchtlichem Druck vorbelastet. Damit eine weitere Biegung des Systems aus Dynamometerarm und Schaftfuß im Verlauf der Messungen verringert und solidere Messergebnisse erzielt.

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List of Abbreviations

MVC maximal voluntary isometric plantar flexion contraction

AJC ankle joint complex

TCJ talocrural joint

STJ subtalar joint

TTJ transverse tarsal joint

ROM range of motion

TS triceps surae

GAS gastrocnemius

SOL soleus

MA moment arm

EMG electromyography

GM gastrocnemius medialis

ADC analog-to-digital converter

RTD_{max} peak rate of torque development

1 Introduction

1.1 Motivation

Proper functioning of the ankle joint complex is crucial for daily life activities. Because of the many different joints, muscles and other anatomical structures involved, accurate measurement of motions around the ankle turns out to be anything but trivial. The following work addresses the specific exercise of isometric plantar flexion at maximal voluntary contraction (MVC). Applied on a dynamometer arm the resultant torque was obtained. However, maximal voluntary contractions “evaluation requires several precautions to be taken, because mechanical and neural factors could greatly influence torque output” (Turpin, Costes, Villeger, & Watier, 2014).

In a study of Herzog (1988), the author pledged for cautious usage of moments data acquired by torque dynamometers. Conclusions about muscle properties shouldn’t be rashly derived, since the moments measured usually differ from the resultant joint moment. There can be found various influencing factors, contributing to those differences. Herzog (1988) named: “(a) gravitational effects, (b) inertial effects, and (c) non-rigidity of the Cybex [dynamometer] arm/shank-foot system.” Moreover the importance of proper alignment of joint axis and dynamometer axis of rotation was mentioned.

Examining plantar flexion, Arampatzis, Morey-Klapsing, et al. (2005) considered these limitations. Even if carefully adjusted between measurements, the axes of rotation did not remain aligned throughout the contractions. In average, differences between measured and resultant (corrected) moment ranged between 6 to $10\,\%$. There seems to occur an inevitable ankle joint motion (Karamanidis et al., 2005), which also affects the involved muscle’s architecture and, if not taken into consideration, leads to unreliable conclusions.

For those mentioned reasons, the present study tackles the specific shortcoming of non-rigidity of the dynamometer arm/shank-foot system, while maximal torque exerted. In a first attempt the joint motion induced discrepancy between measured and resultant moments during plantar flexion was tried to be minimized. Therefore a modified protocol came into use, with adapted posture throughout the measurement process. Maximal voluntary contractions were quantified at the recommended, as well as three more positions with participants positioned closer to the dynamometer axis of rotation.

The benefits of a potentially greater accordance of the observed and resultant moment are various. However, conclusions about muscular properties would be most notable, since also serving as input parameters for modeling and simulation approaches (Menegaldo & Oliveira, 2009; Cheung, 2008; Hoy, Zajac, & Gordon, 1990). Lastly but of at least equal importance Haskell and Mann (2014) mention that a better understanding about the biomechanics of the foot and ankle may also improve success of postoperative treatment and contribute for surgical decision making.

1.2 Thesis outline

Chapter 2 illustrates the basic functioning of the human ankle joint complex, mainly focusing on its plantar flexion. The anatomical essentials about the different joints and muscles involved will be provided.

Chapter 3 further describes the previously mentioned purpose of this study. Additional literature and recent findings will be presented, to also make the motivation for the work in hand more transparent. Conclusively the to-be-examined hypotheses are outlined.

Chapter 4 introduces the sample group as well as the main instruments used throughout the study. Scheduled test sequence and experimental setup alongside with key information about the statistical analysis and data processing are presented subsequently.

Chapter 5 contains the results of the conducted study. Measurements at the different position groups are compared and statistically significant findings indicated.

In Chapter 6 the discussion is given. It peruses the established results and contrasts them with other findings. In a final outlook suggestions are given of how the research could be continued.

The thesis is concluded in Chapter 7 with a final summary in form of the take home massage.

2 Functional anatomy and mechanics of the ankle

The following Chapter provides the anatomical basics about the human ankle joint complex (AJC). Primarily with respect to motion in the sagittal plane since plantar flexion of the foot was examined in the carried out study.

2.1 The ankle joint complex

“Although frequently referred to as the ‘ankle joint’, there are a number of articulations which facilitate motion of the foot” (Brockett & Chapman, 2016). In general terms Nigg and Herzog (2007) define the whole AJC consisting of calcaneus, talus, tibia, fibula as well as all ligamenteous and muscle-tendon structures crossing the joints between the four bones. Brockett and Chapman (2016) describe the AJC as made up of the tibiotalar (talocrural), talocalcaneal (subtalar) and transverse-tarsal (talocalcaneonavicular) joint, shown in Figure 2.1.

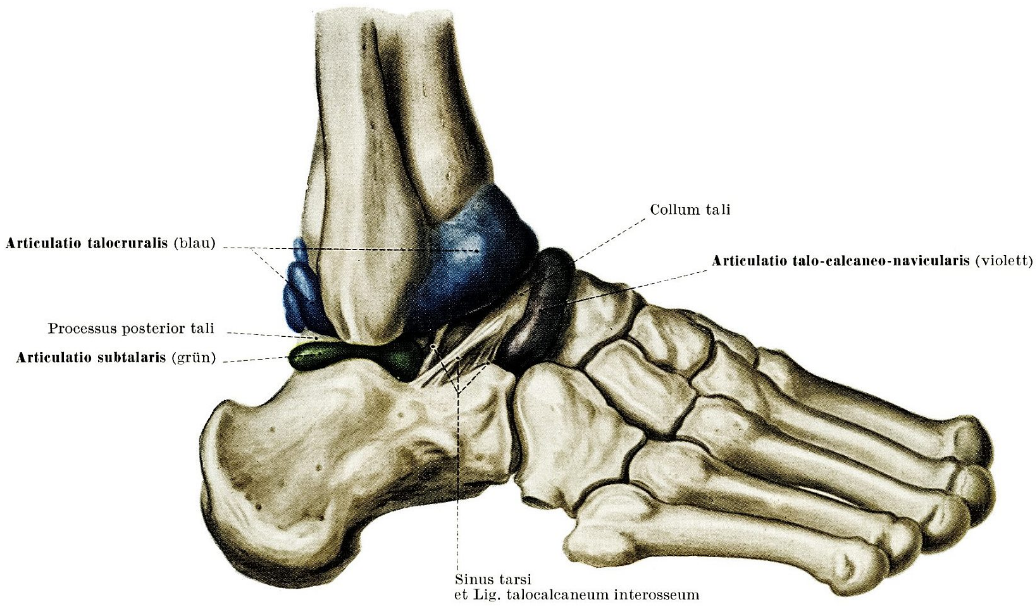


Figure 2.1: The ankle joints of the foot (with colored synovial membrane), lateral view.
Note. Adapted from Lang and Wachsmuth, 1972, p. 364.

2.1.1 The talocrural joint

As illustrated in Figure 2.1 the talocrural joint (TCJ) is made up of the connections between tibia, fibula and talus. These connections are also referred to as tibiofibular, tibiotalar and fibulotalar joints (Donatelli, 1990). “The malleoli of the tibia and fibula act to constrain the talus, such that the joint functions as a hinge joint, and primarily contributes to the plantar- and dorsiflexion motion of the foot” (Brockett and Chapman, 2016). That is why the talocrural or ankle joint is most relevant for the performed study.

In terms of the joint’s function Brockett and Chapman (2016) mention indices, suggesting the TCJ may not strictly function as a hinge. This will be discussed in Section 2.2 in more detail, aiming for a better understanding of the AJC’s motion as a whole.

2.1.2 The subtalar joint

The talocalcaneal or subtalar joint (STJ) consists of the junctions between talus and calcaneus as shown in Figure 2.2. The prominent calcaneus bone is, besides other purposes, providing the attachment area for the Achilles tendon. Movement between the talus and calcaneus occurs around an oblique axis (Donatelli, 1990). Because of the subtalar joint’s geometry, most of eversion and inversion of the foot is provided here (Brockett & Chapman, 2016). However, motion about the ankle and subtalar joint is complex.

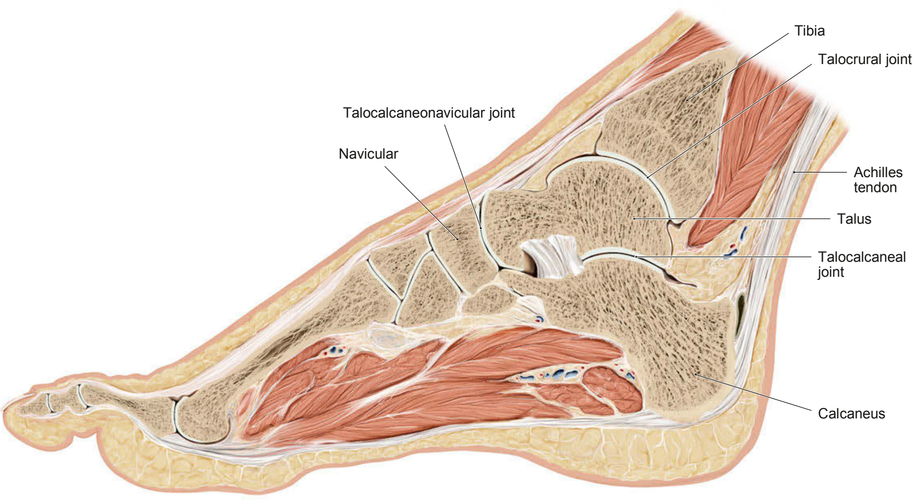


Figure 2.2: Talocrural and subtalar joints in a sagittal section. Right foot, medial view. *Note.* Adapted from Schünke et al., 2014, p. 470.

"The difficulty in understanding the function of the subtalar joint and its relation to the ankle and transverse tarsal joint is that it cannot be easily understood biomechanically as a simple machine. Additionally, the talus is an intercalated segment, without muscular attachment and a paucity of external landmarks which also makes clinical examination difficult." (Jastifer and Gustafson, 2014)

2.1.3 The transverse tarsal joint

The transverse tarsal joint (TTJ) combines the junction between talus and navicular (the talocalcaneonavicular joint, see Figure 2.2). As well as the calcaneocuboid joint between the calcaneus and the cuboid, best seen in Figure 2.3, showing the whole bone anatomy of the lower leg and foot. In respect of its motion the TTJ is considered as part of the same functional unit as the subtalar joint; also contributing to eversion-inversion of the foot since they share a common axis of motion (Brockett & Chapman, 2016).

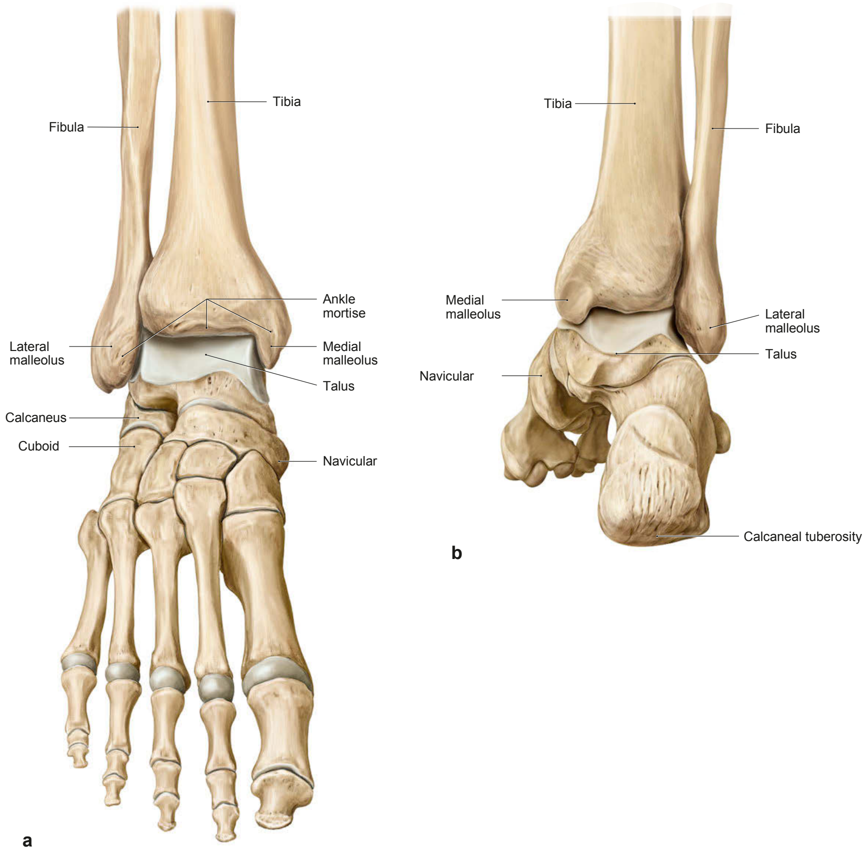


Figure 2.3: Bone anatomy of the ankle joint complex (**a**) Anterior view with talocrural joint in plantar flexion (**b**) Posterior view with foot in neutral (0-degree) position.
Note. Adapted from Schünke et al., 2014, p. 458.

2.2 Ankle joint mechanics

The key movements of the foot about the AJC are shown in Figure 2.4. Described according to the three body planes with: Dorsiflexion and plantar flexion occurring in the sagittal plane, abduction and adduction in the transverse plane, inversion and eversion in the frontal plane (Donatelli, 1990). Yet movement in the foot not necessarily relates to those planes as to be seen in the following.

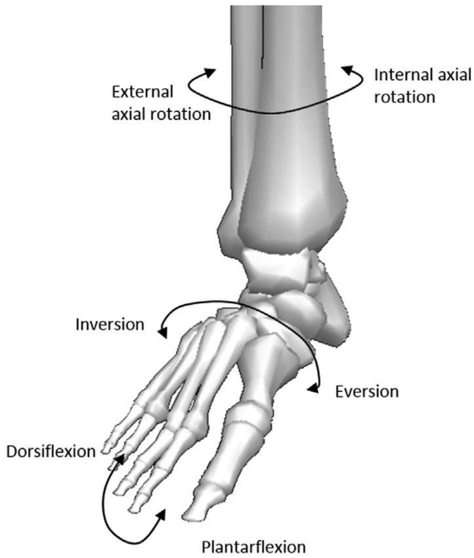


Figure 2.4: Relative motions of the ankle joint complex.
Note. From Brockett and Chapman, 2016.

“Joints of the body are often described based on related simple machines in order to help understand their function such as a ball and socket (hip) or a sloppy hinge (elbow and knee)” (Jastifer and Gustafson, 2014). For illustration purposes the function of a revolute or hinge joint may serve as a rough comparison. However, at a closer look the simplified mechanical model does not hold up, already indicated by the joint’s geometry, such as the cone-shaped trochlea surface and the oblique rotation axis (Brockett and Chapman, 2016). Studies of Siegler, Chen, and Schneck (1988) or Lundberg, Svensson, Nemeth, and Selvik (1989) confirm: “Neither the ankle joint nor the subtalar joint are acting as ideal hinge joints with fixed axis of rotation” (Siegler et al., 1988). The range of motion (ROM) of both joints is rather a triplanar movement of rotations in all three directions (dorsiflexion/plantar flexion; inversion/eversion and internal rotation/external rotation). For the subtalar joint in specific these combined motions of dorsiflexion–abduction–eversion and plantarflexion–adduction–inversion are also referred to as pronation and supination.

2.2.1 Axes of rotation

"The subtalar joint axis is less easy to conceptualize because there are few external landmarks and the axis of rotation is oblique to the traditional anatomic orthogonal planes" (Jastifer and Gustafson, 2014). In studies of Isman and Inman (1969) the axis of rotation of the STJ was found to be inclined about $41°$ to the horizontal and $23°$ to the midline of the foot (see Figure 2.5). Nonetheless there seem to occur extensive individual variations for these values, outlined in a more recent study by Jastifer and Gustafson (2014). As mentioned earlier, the oblique axis of rotation results in a triplanar movement. It is most significant for the subtalar joint with equal amounts of sagittal, transverse, and frontal planar motion during movement (Donatelli, 1990).

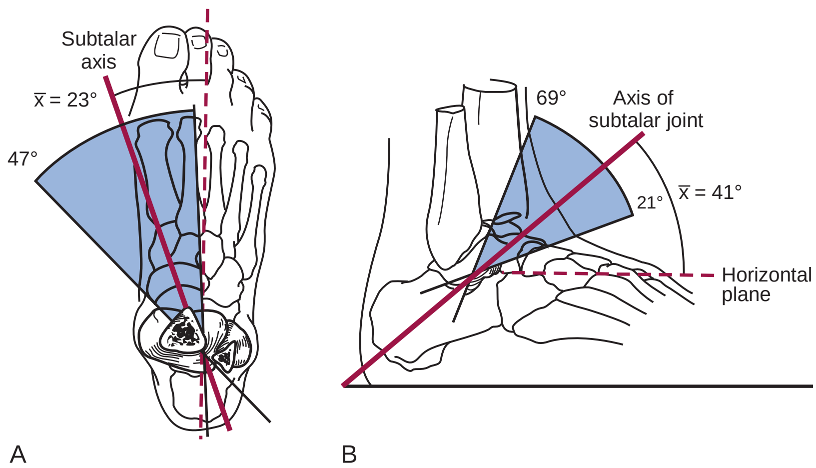


Figure 2.5: Variations in the STJ axes. **A** In transverse plane, **B** In horizontal plane, $\bar{x}$ arithmetic mean.
Note. From Haskell and Mann, 2014, p. 20 modified from Isman and Inman, 1969.

As for the ankle joint, estimation of the axis position is neither trivial. During movement additional changes of its orientation may occur, as shown by Barnett and Napier (1952) or Hicks (1953). Main findings of the mentioned studies are outlined in Klenerman and Wood (2006, p. 86) as follows: "The axis of the ankle joint is not fixed and horizontal, but is inclined downwards and laterally during dorsiflexion, and downwards and medially during plantarflexion. The change occurs within a few degrees of the neutral position of the talus."

That issue taken into consideration Figure 2.6 illustrates the resulting main axis of the ankle joint. There is evidence for a centre of rotation in the ankle joint where all axes, irrespective of their inclination, cross. These findings by Lundberg et al. have practical implications for the conducted study, since "the central point seemed to be located at, or slightly lateral to, the midpoint of a line drawn between the tips of the malleoli."

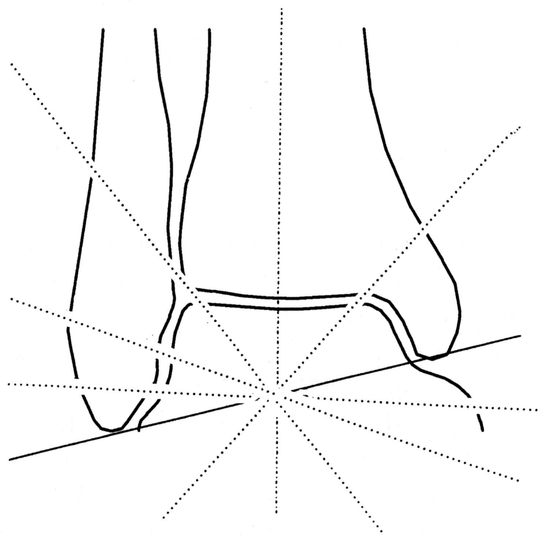


Figure 2.6: The main axis of the ankle runs close to the tips of the malleoli. There is also rotation about an axis close to parallel to the tibia (dashed line), and combined movement can relate to any axis between these (dotted lines).
Note. From Lundberg, 1997, p. 38.

2.2.2 Flexion and extension of the foot

Flexion and extension is usually defined for each joint according to its physiology. "In joint complexes the terms may be contradictory; thus in the ankle and foot, clarity is usually reached through use of the terms dorsiflexion and plantarflexion" (Lundberg, 1997, p. 42).

Having mentioned the difficulties occurring for estimating different momentary axes of rotation in the AJC, it is "usually possible to arrive at a 'compromise axis' that is the mean of all the momentary axes" (Klenerman & Wood, 2006, p. 85). The AJC's main functional axes are shown in Figure 2.7 below. Since this introduction is aiming for a better understanding of the dynamics examined in the present study, mainly plantar flexion motion will be covered in the following.

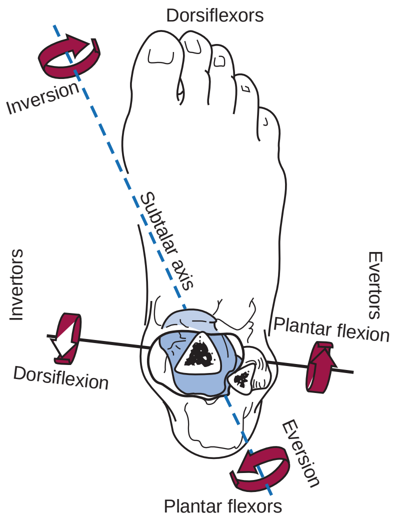


Figure 2.7: Illustration of the rotations that occur about the subtalar and ankle axis.
Note. Adapted from Haskell and Mann, 2014, p. 32 from Haskell and Mann, 2008.

At neutral position the foot sole and longitudinal leg axis are perpendicular to each other, shown in Figure 2.8 A. From that position the instep is either being moved towards the leg, resulting in a dorsiflexion motion (Figure 2.8 B). Or moved away from the leg, resulting in a plantar flexion motion respectively (Figure 2.8 C). The ROM for dorsiflexion lies in between $20°$ to $30°$ with $10°$ of individual differences (striped area). As for plantar flexion the typical ROM is consider between $30°$ to $50°$ with $20°$ of individual variations (Kapandji, 2006).

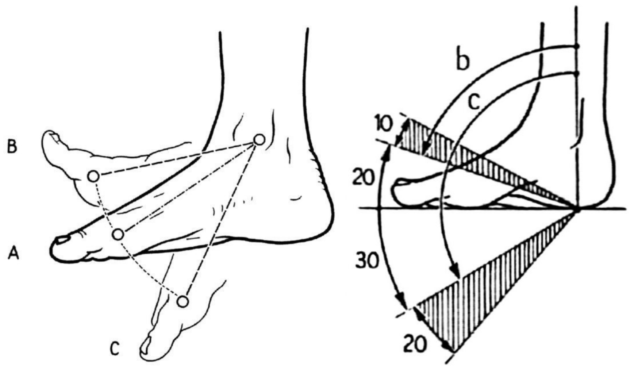


Figure 2.8: Ankle joint's movement in the sagittal plane.
Note. Adapted from Kapandji, 2006, p. 153.

The first part of this introductory Chapter concludes with some general considerations by Lundberg, about the functional anatomy of the ankle and foot.

> "The anatomical obstacles of detailed kinematic assessment of the ankle and foot are substantial. The main problems relate to the relatively small segments and intersegmental distances, and to the difficulty of defining the talus from

> the exterior. There is also large variation in the anatomy of the foot between individuals, which limits the value of universal models of this region. Because of this, in most studies of in vivo function, the foot is seen as connected to the lower leg either through a single hinge joint representing the combined action of all joints between the marker segments, or through two hinge joints (representing the talocrural and talocalcaneal joints)." (Lundberg, 1997, p. 41)

2.3 Muscle involvement

"To be able to influence movement, a muscle obviously has to span at least one joint. A large number of the muscles involved in locomotion span two (biarticular muscles) or several (multiarticular muscles) [joints]" (Lundberg, 1997, p. 29). The muscles for motion within both ankle joints (TCJ and STJ) are shown in Figure 2.9. All muscles are usually involved, either to induce or to decelerate movement. The contraction of each individual muscle is causing the foot to perform a triplanar motion. "A muscle, of course, can (and will) create both an ankle joint and a subtalar joint moment simultaneously. . . . An understanding of the position of the muscle with respect to the axis is critical for understanding its function" (Martin, 2011, p. 471). This is why the oblique axes of rotation (Section 2.2) were discussed in more detail.

Muscles that pass anterior to the talocrural joint axis will cause dorsiflexion, whereas those passing posterior to the axis will make the foot undergo plantar flexion (Figure 2.9). As for the subtalar joint the true axis of rotation is orientated quiet differently than any of the three anatomical axes. Triplanar motion is most significant here and usually referred to as pronation and supination (Donatelli, 1990). With respect to Figure 2.9, muscles that insert medial to the subtalar joint axis will cause supination torques, whereas those inserting laterally will cause pronation torques (Martin, 2011).

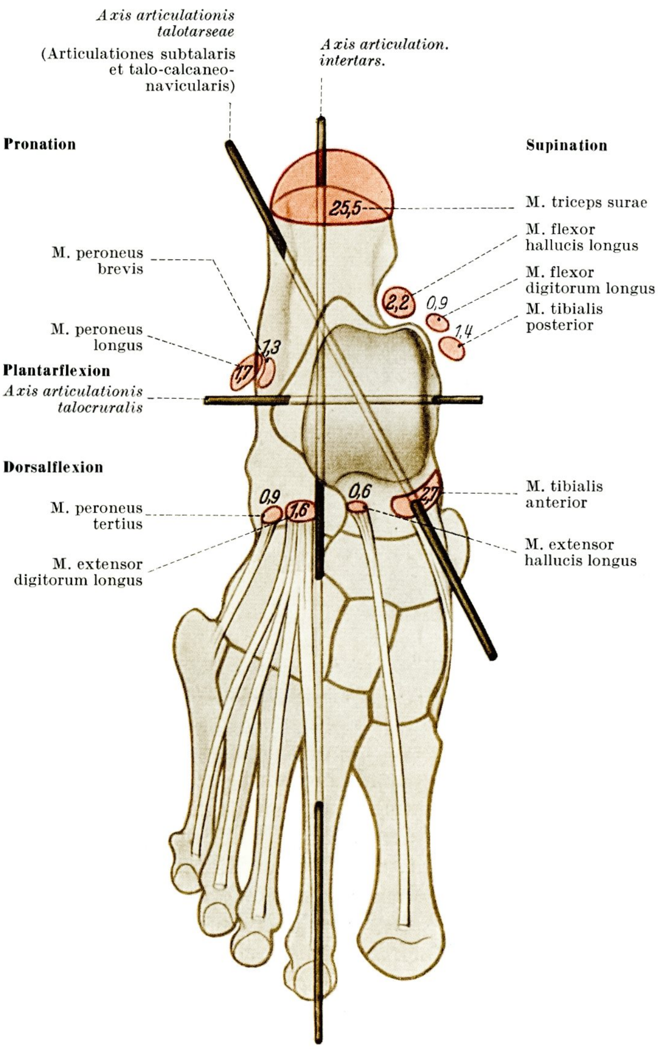


Figure 2.9: Location of muscle insertions in relation to the ankle joint axes.
Note. Adapted from Lang and Wachsmuth, 1972, p. 372.

However, not all of a muscle's work potential contributes for motion around one specific axis only. Depending on the area the muscle is attached to as well as how the muscle is aligned, the influence on the foot's movement will be different. As stated by Lundberg (1997, p. 29) "The relationship between contraction and joint movement is influenced primarily by:

- the length and cross-sectional area of the muscle
- the distance between muscle insertion and joint axis/joint center of rotation
- the fiber direction and composition of the muscle"

2.3.1 Muscle's contribution to the plantar flexion-motion

Muscles illustrated in Figure 2.9 are also compared regarding the size of their cross-sectional areas, correlating with their expected strength (Lang and Wachsmuth, 1972). In accordance with Knuttgen and Komi (2003, p. 6) the term strength will be employed to "identify the maximal force or torque that can be developed by the muscles performing a particular joint movement (e.g. elbow flexion, knee extension)." Numbers in Figure 2.9 relate to kilopondmetre ($\mathrm{kp\,m}$), an obsolete unit of torque. Conversion to the SI unit newton metre results in: $1\,\mathrm{kp\,m} \approx 9.81\,\mathrm{N\,m}$.

The triceps surae (TS) is the strongest plantarflexor of the ankle and most relevant for the conducted study. As shown in Figure 2.10, it consists of the gastrocnemius (GAS) and the soleus (SOL) muscle. "The gastrocnemius muscle arises from two heads of origin on the condyles of the femur and inserts via the Achilles tendon into the most posterior aspect of the calcaneus" (Martin, 2011, p. 471).

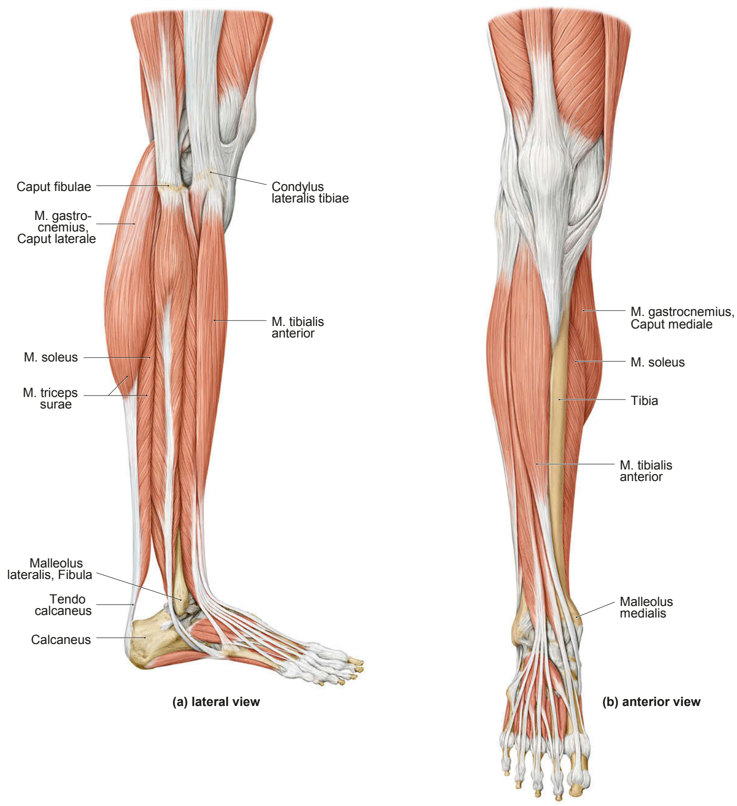


Figure 2.10: Muscles of the right leg.
Note. Adapted from Schünke et al., 2014, p. 516.

Because of that attachment, the gastrocnemius is functioning as a biarticulate muscle, as shown in Figure 2.11. The two indicated moment-arms have a length of $5\,\mathrm{cm}$ at the ankle and

3.5 cm at the knee. This rather small 1.5 cm difference has vital implications, resulting in a higher contribution to the ankle extensor moment than to the knee flexor moment. “The net effect of these two contributions is to cause the leg to rotate posteriorly and prevent the knee from collapsing” (Winter, 2009, p. 103).

“The soleus muscle is deep to the gastrocnemius, originating on the tibia and fibula and inserting with the gastrocnemius into the posterior calcaneus via the Achilles tendon” (Martin, 2011, p. 471). As indicated in Figure 2.10, the large plantar flexion torque of the TS also originates from the insertion point of the Achilles tendon, which is perpendicularly on the calcaneus, relatively far from the ankle joint axis thus resulting in a large moment arm (MA).

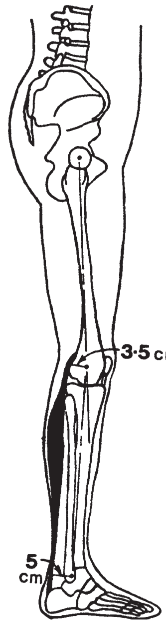


Figure 2.11: The biarticulate gastrocnemius with its moment-arm lengths at the proximal and distal end.
Note. Adapted from Winter, 2009, p. 104.

Since in the present study isometric plantar flexion was examined, this basic illustration of the triceps sure complex should be sufficient. From the muscles inserting anterior to the talocrural joint axis, the tibialis anterior is the strongest (dorsiflexor). Yet having only a fraction of the plantarflexor’s strength its antagonistic coactivation during plantar flexion could underestimate the resultant ankle joint moment and should therefore be considered (Arampatzis, Stafilidis, et al., 2005).

2.3.2 Muscle architecture

Influencing factors for the relationship between contraction and joint movement were listed at the beginning of this Chapter. Not yet discussed was the fibre direction and composition of the muscle. Knowledge about those properties is quiet useful to validate drawn conclusions about muscle forces since they cannot be measured directly. For illustration purposes Figure 2.12 shows ultrasound scans of the gastrocnemius muscle at different contraction levels. From measured electromyography (EMG) activity corresponding torque levels might be deduced, and with known moment arms also the acting forces (Magnusson, Aagaard, Rosager, Dyhre-Poulsen, & Kjaer, 2001). However, one must cautiously look out for unexpected observations. As an example Arampatzis et al. (2006) revealed a decrease in EMG activity of the biarticular gastrocnemius medialis (GM) during MVC at flexed knee-joint position despite no measured differences in GM's fascicle length.

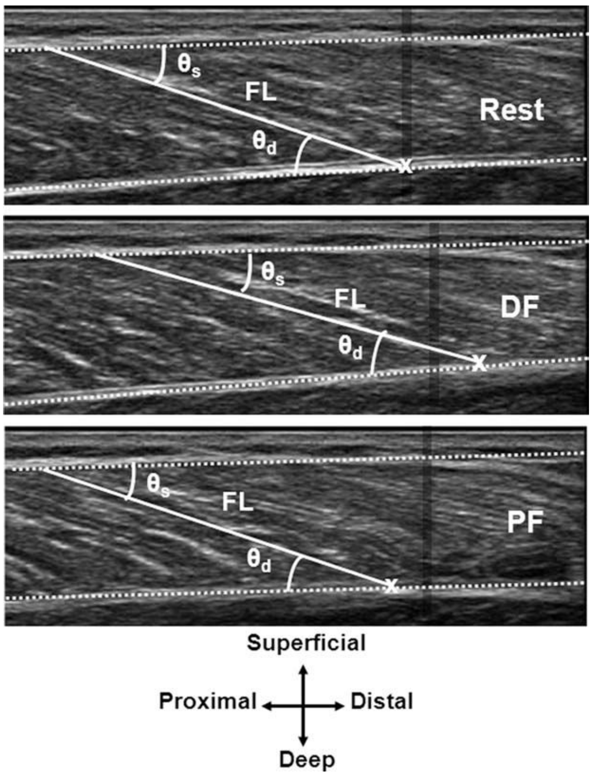


Figure 2.12: Ultrasound scans of the GM obtained at rest, during maximal dorsiflexion (DF) and submaximal plantarflexion (PF) isometric contractions, at a given level of EMG activity of the GM. FL, fiber length; θ_s, superficial pennation angle; θ_d, deep pennation angle.
Note. From Simoneau et al., 2012.

3 Purpose of the study

With the anatomical and physiological basics introduced, all necessary background information should now be provided to further describe the purpose of the conducted study. Besides the mentioned implications of ankle joint rotation during maximal voluntary contraction efforts (Section 1.1) further entanglement was revealed in biomechanics research. In several cases (Maganaris, 2002; Kubo, Kanehisa, & Fukunaga, 2002, 2005) the functional characteristics of the triceps surae were examined in their theoretical optimal length (Kubo et al., 2005). As previously stated, it is known that the joint rotation alters the morphological properties of the muscle fibers and therefore could also alter the theoretical optimum of the muscle's force length relation. This possible outcome might reduce the exerted torque and therefore underestimate the force production capability of the respective muscles. In that case, corrections that had been previously (Karamanidis et al., 2005) proposed could not adjust for the absent torque. Taken into account that the mentioned averaged differences of the exerted torque due to axis misalignment ranges from 6 to 10 % (Arampatzis, Morey-Klapsing, et al., 2005) and adding also a possible additional torque loss due to alterations of the force length relationship of the respective muscles, the corrected torque could be higher than previously reported.

Implications on estimation of the tendon's load bearing capability is another important aspect. According to the literature (Maganaris, 2002) is it is often attempted to assess the force length characteristics of the Achilles tendon during maximal voluntary contractions. Also in conjunction with the Achilles tendon morphological properties (tendon area) the Young's modulus gives an estimation of the tendon mechanical capabilities (Maganaris, 2002). This information can help physicians to modulate the tendon properties in respect to the movement task through various interventions and training programs. Although an underestimation of the Achilles tendon mechanical properties can be seen as a safety margin, in case of repeated measurements the additional error in the estimation of the exerted torque, as stated above, could lead to erroneous results and therefore to training interventions that could increase the actual load acting on the tendon and furthermore the risk of injury. Not only the stress but also the strain of the tendon could be underestimated in case of lower force production through the improper positioning. Such an outcome could have major implications in the understanding of functional characteristics of the Achilles tendon and its contribution on the energy storage and recoil at various moving (running, sprinting, hoping) tasks. It is reported (Maganaris, 2002;

Butler, Grood, Noyes, & Zernicke, 1978) that the force elongation property of the tendon has s curve linear form and that the stiffness of the tendon is estimated in the upper portion (about 50 to $100\,\%$) of that relation, where a linear region is expected. If we assume that with the modified positioning in the present study the exerted torque will be higher, it is still unclear how the tendon structure will behave and how different its stiffness compared to the normal positioning will be. In a first approach this present research aims to identify if the exerted torque of the triceps surae muscles is affected by the positioning of the subject on the isokinetic device and its possible effects on other muscle tendon mechanical parameters.

Another important aspect of the muscle's force generation mechanism is its electromyographic activity during maximal voluntary contractions. In an earlier study (Stafilidis & Arampatzis, 2007) it has been shown that at maximal voluntary contractions the EMG activity of the plantar flexor muscles was increasing with increased muscle fiber length towards optimum muscle fiber length. In this perspective, the rotation of the ankle joint during voluntary contractions could alter the operating length of the muscle fibers and possibly the activation level of the respective muscles. The positioning of the subjects on the isokinetic dynamometer seemed to have an substantial effect on the exerted torque not only for the plantar flexors but also for the knee extensors (Stafilidis & Arampatzis, 2007). Rotation of the knee joint as well as misalignment of the joint axis to the axis of the isokinetic dynamometer was also evident while maximal voluntary extension contraction efforts (Stafilidis & Arampatzis, 2007). Similar to ankle joint correction, adjustments for joint axis misalignment are necessary in order to acquire reliable information about the function of the quadriceps muscles. In many cases physicians as well as researchers have to examine the functional characteristics of the triceps surae or the quadriceps femoris muscle group in a pre-post assessment design. If the positioning of the subject can influence the torque outcome it would be necessary to standardize that proper subject position prior to examination.

With the known shortcoming of plantar flexion peak torque assessment, a novel setup arrangement and experimental protocol were conceptualized for the conducted study. Targeting the specific limitation of a non rigid dynamometer arm/shank-foot system. It should be revealed weather positioning of the participants has a significant effect on the established parameters.

3.1 Hypotheses

The purpose of this study was to examine the following hypotheses:

Hypothesis 1: Plantar flexion peak torques measured at the dynamometer during maximal voluntary isometric contractions at *different reclining chair positions* differ significantly.

Hypothesis 2: Rates of torque development measured at the dynamometer during maximal voluntary isometric contractions at different chair positions differ significantly.

The term different reclining chair positions is identified with varied horizontal distances between dynamometer axis and chair.

4 Material and methods

In this chapter the sample group as well as testing equipment used throughout the study are being introduced. Furthermore the setup of the measuring instruments will be described alongside with the experimental protocol of the actual testing. Lastly, key information about statistical analysis and data processing will be summarized.

4.1 Participants

The study was conducted on 18 healthy adult males (age: 29.26 ± 6.07 years, body mass: 76.99 ± 9.81 kg, height: 181.43 ± 4.65 cm). Randomly acquired at the Centre for Sport Science and University Sports where they regularly participate in physical activity. All volunteers gave their consent to take part in the study. The Ethics Committee of the University of Vienna[1] approved all procedures. None of the participants had any major or recent musculoskeletal injury in the examined leg at the time of testing.

4.1.1 Motion Capture System

"*Joint kinematics data* are generally obtained by using motion capture systems such as VICON and motion analysis. These systems use high-resolution infrared cameras to track and obtain the trajectories of external skin-mounted markers using the optical principle" (Dao & Tho, 2014, p. 23). The emitted (stroboscopic) radiation is "reflected in the incident direction by markers whose surface is made of a retro-reflective material. A filter positioned on the lens makes the cameras sensitive to a specific wavelength, and only the markers are, therefore, detected" (Chèze, 2014, p. 18). An eight-camera motion capture system (Vicon MX™ , OMG plc, Oxford, UK) was used to collect kinematic data. The system consists of six MX13 cameras and two MX40 cameras, set at a sampling frequency of 200 Hz. Illustrated in Figure 4.1 is the basic Vicon MX™ architecture, yet for actual testing a slightly different assembled setup was used. Instead of the MX Ultranet one MX Link, one MX Control as well as two units of MX Net were implemented in the system.

[1]http://ethikkommission.univie.ac.at/en/mission-statement/

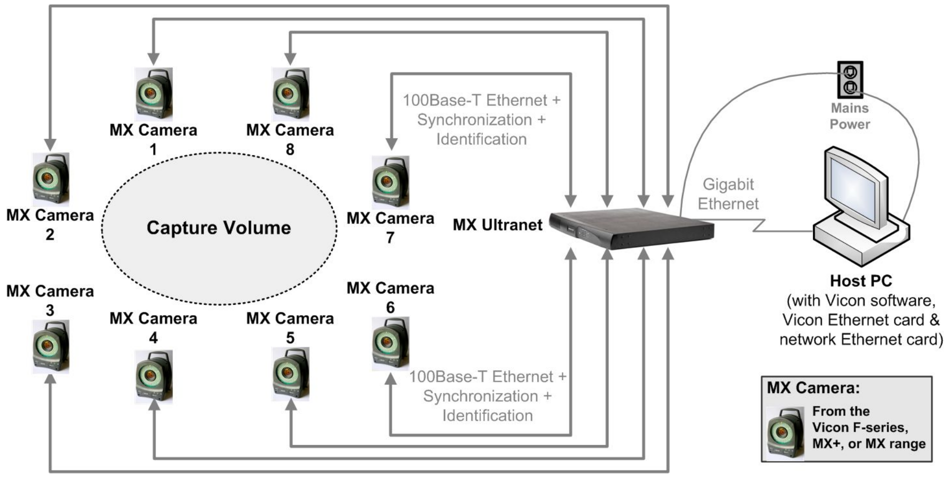


Figure 4.1: Generic motion capture system architecture.
Note. Obtained from the Vicon MX™ Hardware System Reference, Revision 1.6
http://www.evl.uic.edu/sjames/mocap/resources/Doc/MXhardware_Reference.pdf

4.1.2 Dynamometer

A dynamometer in the context of human biomechanics might be defined as "a device that measures force or torque for muscular performance testing" (Knudson, 2007, p. 286). Muscle strength is measured via maximal voluntary contraction utilizing a dynamometer under isometric conditions. As mentioned in Section 2.3 the SI unit of torque is newton metre. The terms moment and torque are often used interchangeably, not without controversy though (Rodgers & Cavanagh, 1984). This issue will not be discussed in more detail, yet should be taken into account since the terms used by cited authors may vary.

The device used for the present study was a HUMAC® NORM™ Model 770 (CSMi, Stoughton, MA, USA). The main components are shown in Figure 4.2.

Figure 4.2: Illustration of the isokinetic dynamometer system used.
Note. Retrieved from http://www.datateknikticaret.com.tr/datateknikdir/data/storage/attachments/72fa7727cf25893fabaade32affb3a93.jpg

System accuracy retrieved from the User's Guide (©2006 Computer Sports Medicine, Inc.) is listed in Table 4.1. Conversion of torque range to the SI unit yields in: $500\,\mathrm{lbf\,ft} \approx 678\,\mathrm{N\,m}$. The raw torque signal was captured with the Vicon measuring system (ADC card, 16bit) at a sampling frequency of $2000\,\mathrm{Hz}$ and stored for further analysis.

Table 4.1: Measurement accuracy of the HUMAC® NORM™ System.
Note. Retrieved from http://www.csmisolutions.com/sites/default/files/300004d-409_humac_norm_user_guide_english_0.pdf

Channel	Range	Accuracy
Torque	$\pm 500\,\mathrm{lbf\,ft}$	$0.5\,\%\,\mathrm{FS}$
Position [2]	$\pm 3600°$	$\pm 0.25°$

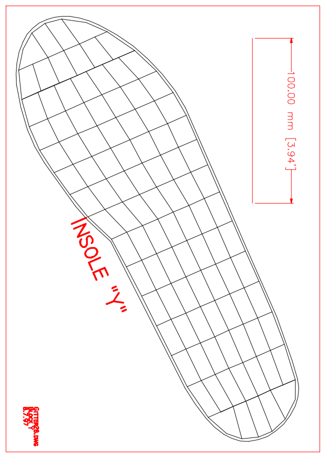
4.1.3 Pressure distribution measuring insoles

Pressure is calculated as force divided by the contact area on which this force acts. The SI unit of pressure is pascal ($1\,\mathrm{Pa} = 1\,\frac{\mathrm{N}}{\mathrm{m}^2} = 1\,\frac{\mathrm{kg}}{\mathrm{m\,s}^2}$). In biomechanical studies, forces (ranging around

[2]Position range of ten revolutions in both directions

orders of magnitude of the body weight) are distributed over a rather small area. "Thus, the units of kilopascals (kPa) and megapascals (MPa) are frequently used" (Rodgers & Cavanagh, 1984). Yet different technologies exist, capacitive transducers were being utilize for the present study. "An electrical capacitor typically has two metal plates in parallel with each other with a dielectric material sandwiched in-between. . . . Applied forces can be determined through the compression of the elastic material (dielectric) between the plates" (Hennig & Lafortune, 1997, p. 116). For the conducted study the pedar®-x system (novel GmbH, Munich, Germany) was called into action, with its components illustrated in Figure 4.3.

(a) The PX290 main device connected to an insole.
Note. Retrieved from http://www.novel.de/novelcontent/images/stories/download/novelpictures/web/pedar-insole-shoe(c)novel.jpg



(b) Sensor matrix of the used insole model Y.
Note. Retrieved from http://www.novelusa.com/assets/pdf/pedar/y.pdf

Figure 4.3: Components of the pressure distribution measuring system used.

The properties and metrics of the specific insole used are shown in Table 4.2, sampling frequency was set to 100 Hz. Throughout processing both the point of force application (for establishing moment arm lengths) as well as the net loading at the beginning of each contraction were of main interest and obtained from the gathered data.

Table 4.2: Pedar®-x system insole properties.
Note. Retrieved from http://www.novel.de/novelcontent/sensors

Product ID	Name	EU Shoe Size	US Shoe Size	Number of Sensors	Pressure Range [kPa]
Y	Adult Insole Standard	44–45	10.5–11	99	20–600

4.2 Experimental setup

Arrangement of the measuring instruments is illustrated in Figure 4.4. Infrared cameras were placed around the observed volume of the dynamometer system with height and perspective adjusted for best possible detection of the retroreflective markers. Aligned in a way that each of the markers was detected by at least two cameras throughout measurement. One marker was mounted on the dynamometer axis center. In order to define the line of force application, two markers were placed on the foot plate of the ankle adapter attached to the dynamometer. Visual feedback was provided with a laptop placed on the swivel monitor arm (Figure 4.1). As stated above, via the DB-9 connector at the rear of the HUMAC interface, analog output data was collected from the Vicon MX Control unit's analog-to-digital converter (ADC) card in sync with joint kinematics data.

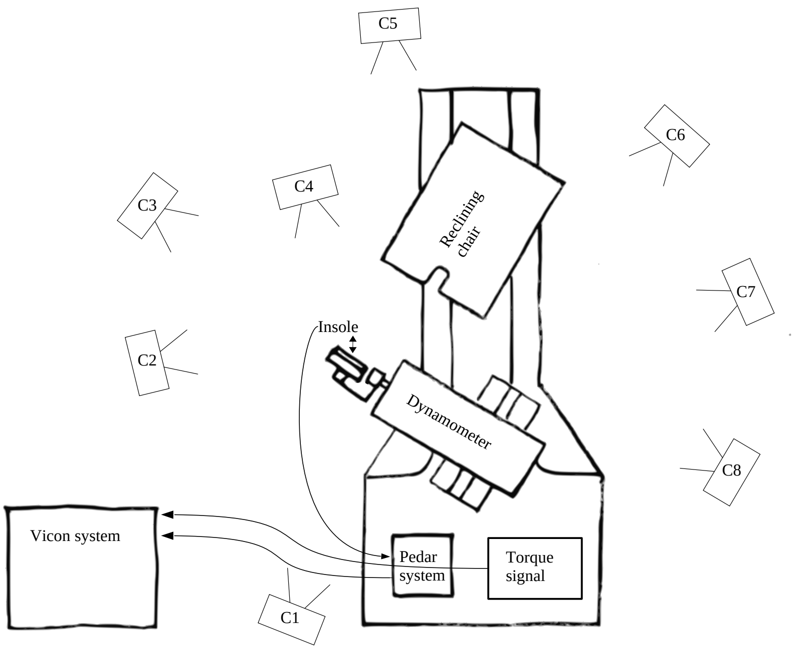


Figure 4.4: Schematic view of the experimental setup.

For an even surface, belt slide supports as well as the threaded stud weldment were detached from the ankle adapter. Thereby exposed notches were covered with a thin wooden panel of the same shape as the foot plate. The pressure distribution measuring insole could now be secured on the adapter, with one marker attached to it at the most anterior superior point. Measurement procedure was coordinated from a separate PC running Vicon Nexus™. Data gathered from the pedar®-x system was synchronized by using a transistor–transistor logic signal.

4.3 Experimental protocol

A detailed description of the experimental procedure is provided in the following.

4.3.1 Positioning

With body weight, height and age ascertained, participants were introduced to the testing cycle. After taking a seat on the CSMi NORM™ system's reclining chair, posture was adjusted. Sitting upright the chair's back (for lumbar support) was extended to meet the participant's back with about a handbreadth gap remaining between knee pit and seat edge.

At that point volunteers were asked to extend their left leg while the left foot was stabilized on the footplate of the ankle adapter. Input arm penetration and footplate penetration were adjusted to align ankle axis of rotation with the dynamometer axis. The ankle joint was defined as the midpoint of the line connecting both medial and lateral malleolus (see Section 2.2.1).

With the foot resting on the plate, ankle in neutral position (approximately $90°$) and the knee fully extended, the reclining chair was moved forward and then secured at the closest distance to the dynamometer. These adjustments served as the reference setting (position O) for the testing cycle. Since intact human gastrocnemius muscles operate on the ascending limb of the force–length relation (Herzog, Read, & Ter Keurs, 1991) the knee joint needed to be fully extended, in order to potentially capture peak torque values.

4.3.2 Marker Placement

The marker set consisted of 11 passive retroreflective spherical markers. In addition to the four mentioned markers fixed on dynamometer and insole, remaining markers were applied on the participants at the following positions: C7, trochanter major, lateral and medial malleolus, the most prominent points of the lateral and medial femoral condyles and lastly the tuber calcanei (Figure 4.5).

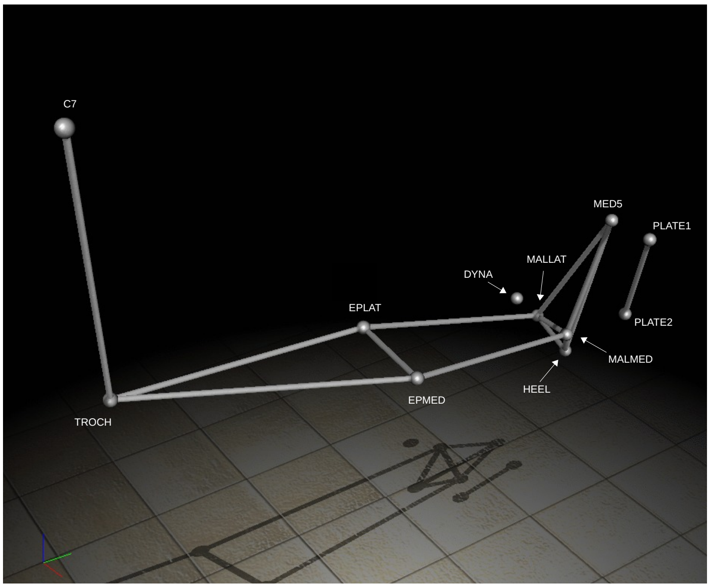


Figure 4.5: Illustration of the used marker set.
Note. Produced using Vicon Polygon™4.2

4.3.3 Maximal voluntary contractions

Prior to marker application a warm up phase of about five minutes was completed. Participants performed multiple submaximal as well as a few maximal isometric plantar flexion efforts. For the actual testing all subjects performed isometric maximal voluntary contractions at 4 different dynamometer reclining chair positions. The mentioned reference point is referred to as position O. Position I was defined at a $3\,\mathrm{cm}$ closer distance to the dynamometer axis. At Position II the dynamometer chair was moved $6\,\mathrm{cm}$ closer to the dynamometer axis compared to the reference setting. Position III was defined as the closest tolerable position participants where still able to fully extend their left knees and perform MVCs. At the first appearance of any feeling of pain, the trial would have been aborted.

Precise adjustment for moving the reclining chair forward or backward was ensured by using the lever mechanism on the chair's pedestal. Participants were asked to fold their arms across the chest during testing. Measurement sequence of the different positions was randomized. Yet at a specific position testing followed the same procedure:

1. A static measurement of the knee angle with legs bent but foot remaining adjusted on the footplate as during MVC (see Figure 4.6).
2. One trial of fastest possible voluntary contraction plantar flexion efforts consisting of two contractions with a $20\,\mathrm{s}$ pause in between.
3. At least three trials of isometric MVC plantar flexions with left knee fully extended

and the inextensible strap fastened around the involved thigh. Participants increased the force to maximal over 3 s and were strongly verbally encouraged to exert maximal force for approximately 5 s.

Figure 4.6: Participant's adjustment on the dynamometer at one of the closer positions, while leg at rest.

In between trials the thigh strap was loosened and participants given at least one minute of rest (prevention of muscle fatigue). While MVC trials, visual feedback was provided in form of a torque vs. time line graph scrolling left-to-right across the software's GUI (HUMAC®2015v15.000.0103). Both, verbal encouragement and visual feedback were provided for participants during trails in order to reach their "true" MVC levels (Bickers, 1993; Toumi, Jakobi, & Simoneau-Buessinger, 2016).

4.4 Data processing and statistical analysis

The analysis of the all experimental data was performed with the MATLAB® R2014a software package (MathWorks, Natick, MA, United States).The 3D trajectories of the markers were filtered by a fourth-order zero lag Butterworth filter with cut-off frequency of 12 Hz. Analogue torque data and pressure data were filtered with a fourth order zero lag Butterworth filter with a cut-off frequency of 10 and 6 Hz respectively. A common frequency (2000 Hz) for all measured data (torque, kinematic, pressure) was achieved by means of cubic spline interpolation. Trials were processed using Vicon Nexus™ 2.5 (OMG plc, Oxford, UK), exported data analyzed with IBM® SPSS® Statistics v23 (IBM Corp., Armonk, New York, USA). The outcomes of the four different chair configurations were compared to assess whether there was any difference of these positions on the examined parameters. To do this, a one-way analysis of variance with repeated measures and a post-hoc test with a Bonferroni correction was performed to evaluate the differences between the four position groups (O, I, II & III). The level of statistical significance was set at $\alpha = .05$ for all the tests applied.

5 Results

Measurements taken at the four different chair positions are presented in the following Chapter. Table 5.1 lists the mean distance alterations at which trials were executed. Position O was set as reference point. Participants placed their left foot on the dynamometer footplate with leg fully extended. While maintaining a straight knee angle (approximately $180°$) the chair was locked in place at the closest distance to the dynamometer axis. Position I means a $3\,\mathrm{cm}$ closer distance to the dynamometer axis. At Position II the dynamometer chair was moved $6\,\mathrm{cm}$ closer to the dynamometer axis compared to position O. Position III was defined as the closest tolerable position participants were still able to fully extend their left leg and perform MVC.

Table 5.1: Measuring positions via horizontal distances (mean values ± SD, $n = 14$).

Position	Altered distance to dynamometer axis [cm]
O	0.00 ± 0.00
I	-3.00 ± 0.00
II	-6.00 ± 0.00
III	-8.25 ± 0.85

5.1 Hip angle measurements

Statistically significant differences between most groups were found for hip angle deviation at the onset of contractions (Figure 5.1). Solely comparing position II and III, no significant difference could be discovered. At MVC, significant differences in hip angle deviation were only found between positions closer to the dynamometer (I, II, III) compared to referece position O (Figure 5.7). For both conditions (onset of contraction and MVC) deviations were smallest at O and systematically decreasing for closer (tighter) participant positioning.

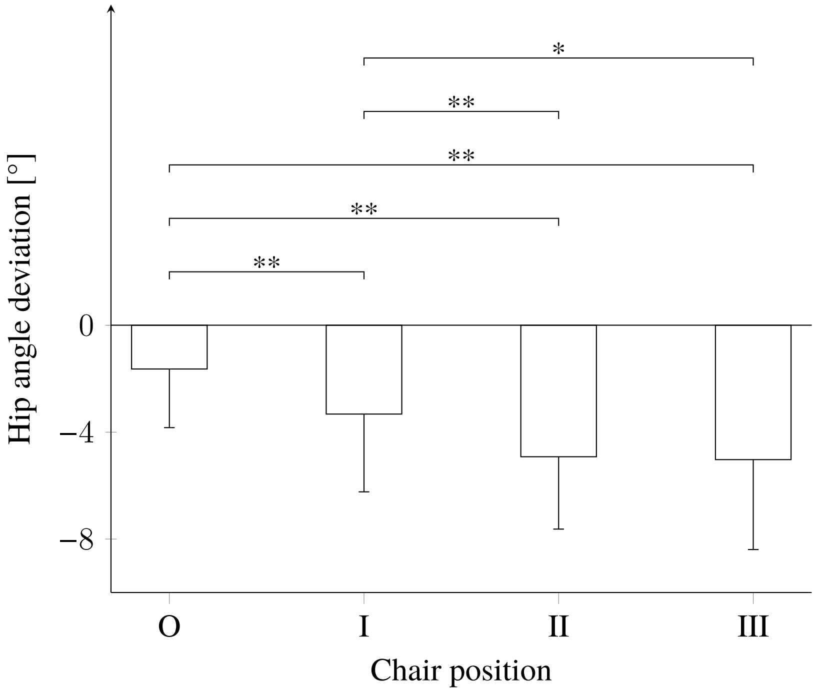


Figure 5.1: Hip angle deviation at the onset of contractions (mean ± SD, n = 41).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .389, $F_{3,38} = 19.934$, $p < .000$ and $\eta^2 = .611$.
*: Statistical significance level of $p < .05$.
**: Statistical significance level of $p < .01$.

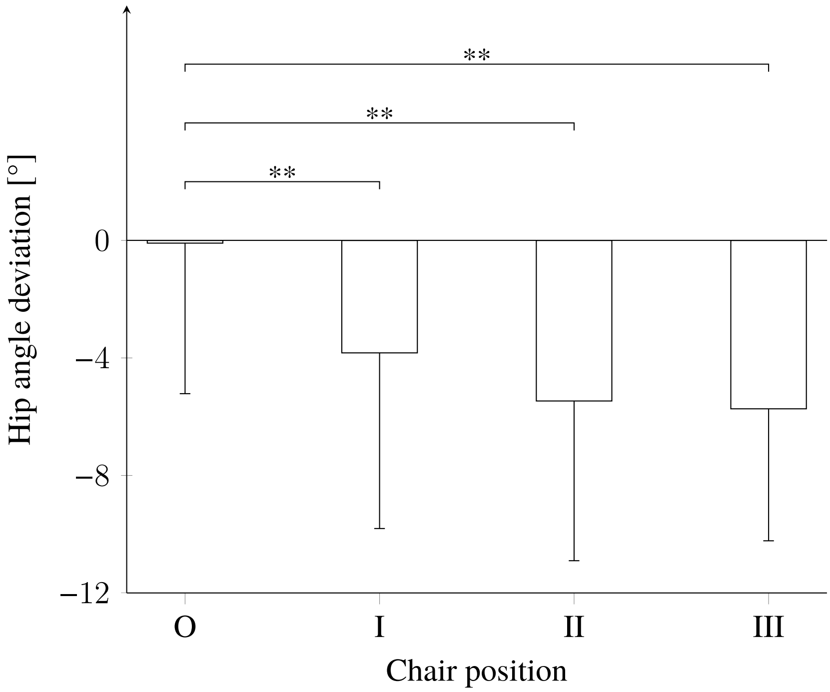


Figure 5.2: Hip angle deviation at MVC (mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .268, $F_{3,39} = 35.418$, $p < .000$ and $\eta^2 = .732$.
**: Statistical significance level of $p < .01$.

5.2 Knee angle measurements

At every measuring position on the dynamometer chair, bent-knee angles with legs at rest were gathered as well. Figure 5.3 shows the mentioned knee angles, specific values are listed in Table 5.2.

Altered knee angels at MVC would affect peak torque development (as mentioned in Section 4.3.1). No statistically significant differences between groups were found at the onset of contraction (Figure 5.4). At MVC knee angels measured at neutral position O significantly differed compared to all other positions (Figure 5.5). Yet no statistically significant differences found between the other groups.

Table 5.2: Measuring positions via knee angles (mean values ± SD, n = 13).

Position	Bent-knee angle before MVC [°]
O	180.00 ± 0.00
I	153.10 ± 7.36
II	146.00 ± 7.01
III	141.34 ± 4.41

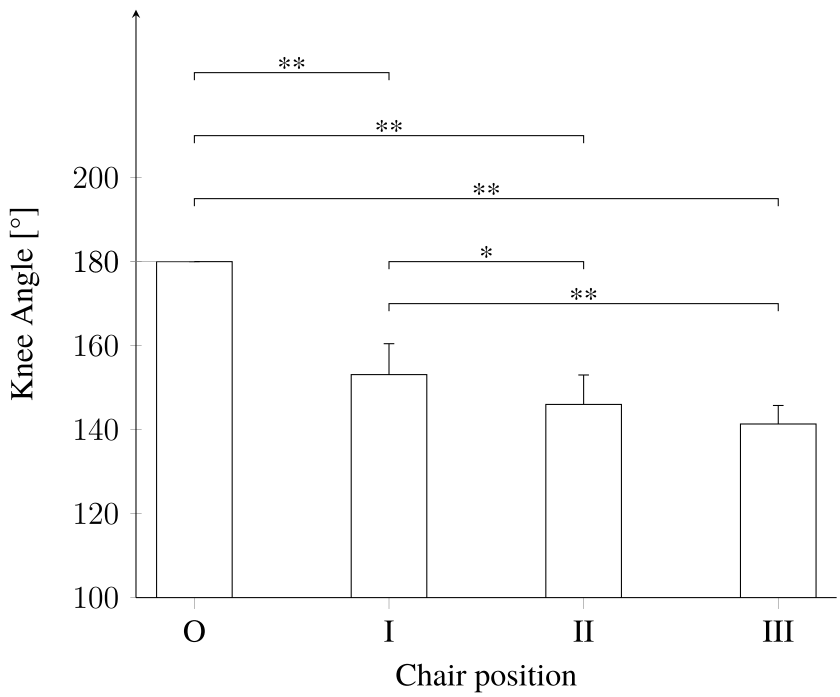


Figure 5.3: Angles measured at the different chair positions with bent knee before MVC were performed (mean ± SD, n = 13).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .009, $F_{3,10}$ = 368.976, p < .000 and η^2 = .991.
*: Statistical significance level of p < .05.
**: Statistical significance level of p < .01.

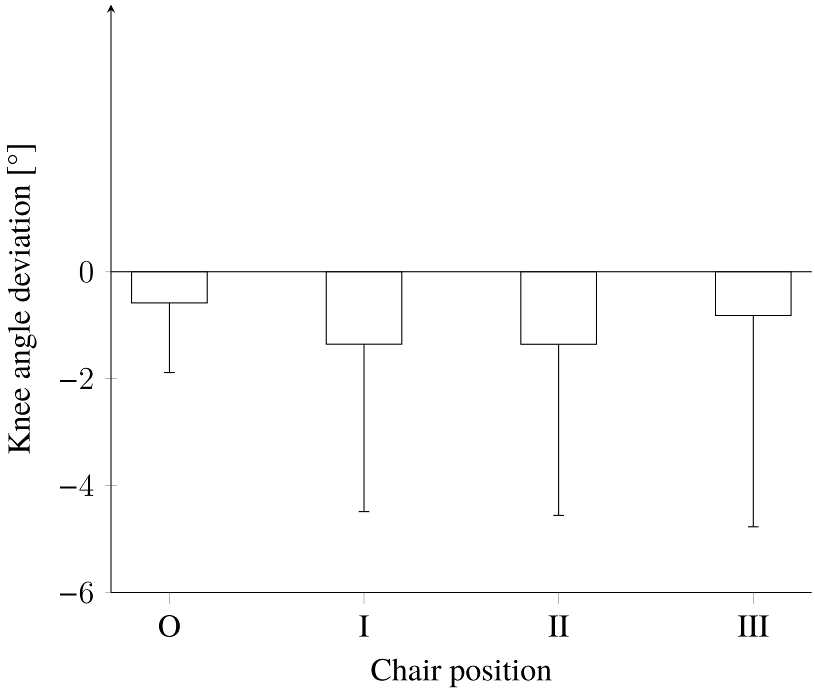


Figure 5.4: Knee angle deviation at onset of contractions (mean ± SD, $n = 41$).
Note. On the examined parameter no statistically significant effect for position was found.

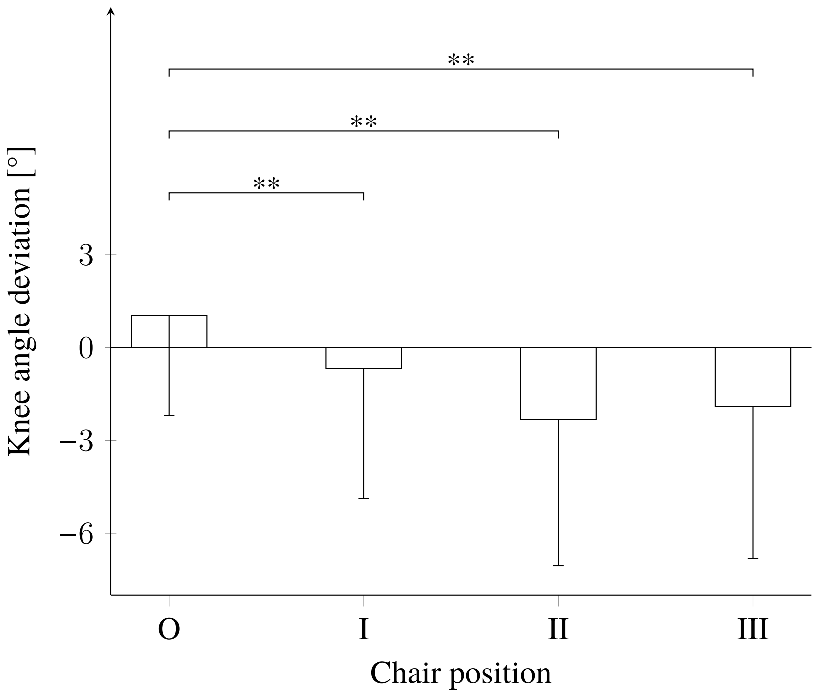


Figure 5.5: Knee angle deviation at MVC (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .666, $F_{3,39} = 6.527$, $p < .001$ and $\eta^2 = .334$.
**: Statistical significance level of $p < .01$.

5.3 Ankle angle measurements

Ankle angle deviation at the onset of contractions was marginal (Figure 5.6). Yet between position O and III a statistically significant difference appeared. At MVC significant differences in ankle angle deviation were found for every position compared to all other groups. As presumed, ankle angle deviation is lowest for configuration III and significantly higher for each position further away from the dynamometer axis (Figure 5.7).

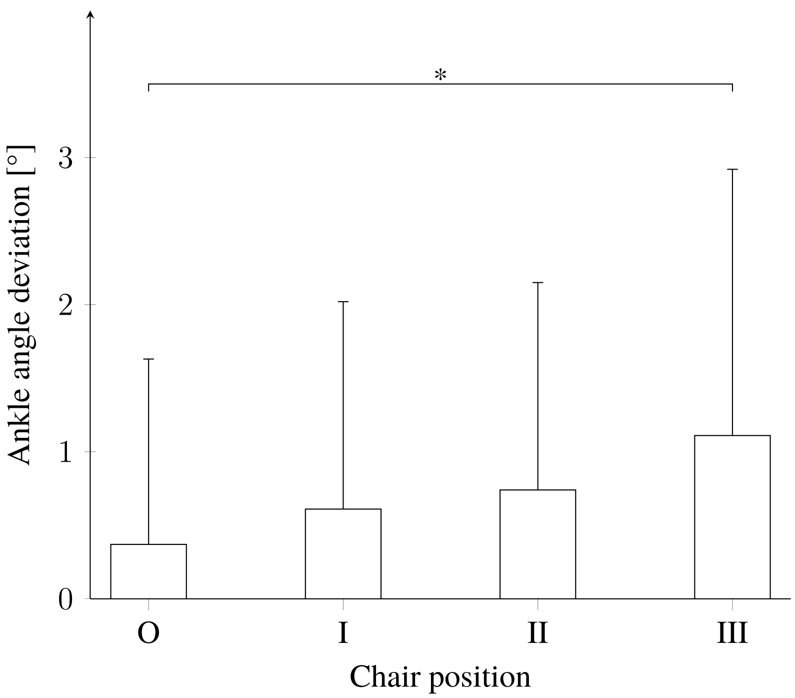


Figure 5.6: Ankle angle deviation at the onset of contractions (mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .790, $F_{3,39}$ = 3.463, p < .025 and η^2 = .210.
*: Statistical significance level of p < .05.

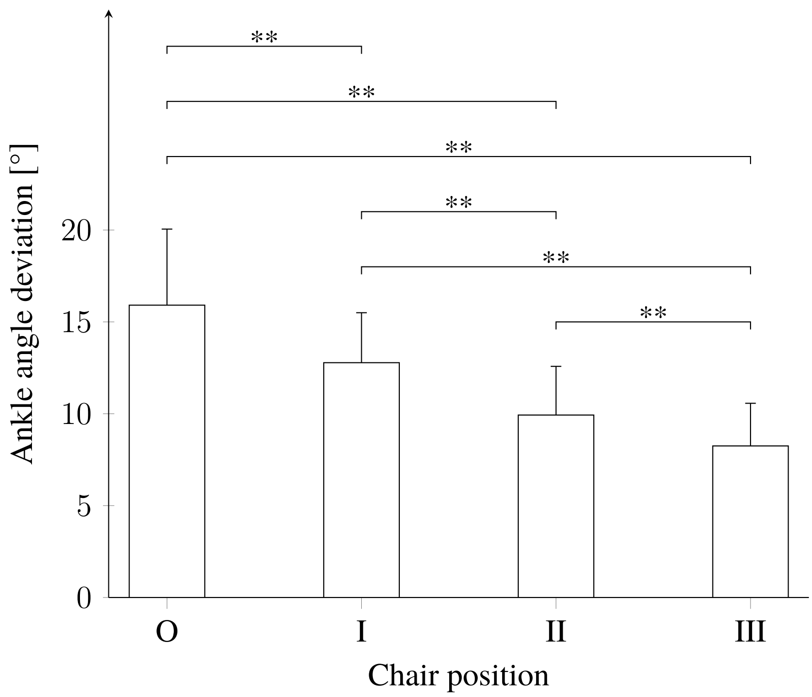


Figure 5.7: Ankle angle deviation at MVC (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .086, $F_{3,39} = 138.675$, $p < .000$ and $\eta^2 = .914$.
**: Statistical significance level of $p < .01$.

5.4 Peak torque measurements

In accordance with Figure 5.8, maximum peak torque was achieved at position III. Yet no significant differences were found compared to position II. Only at positions further away from the dynamometer axis, measured peak torques were significantly lower (I, O).

Dynamometer footplate rotation seems to correlate with peak torques. Also here (Figure 5.9) statistically significant differences were found in between all groups except for position II and III, since no significantly higher torque was exerted on the dynamometer.

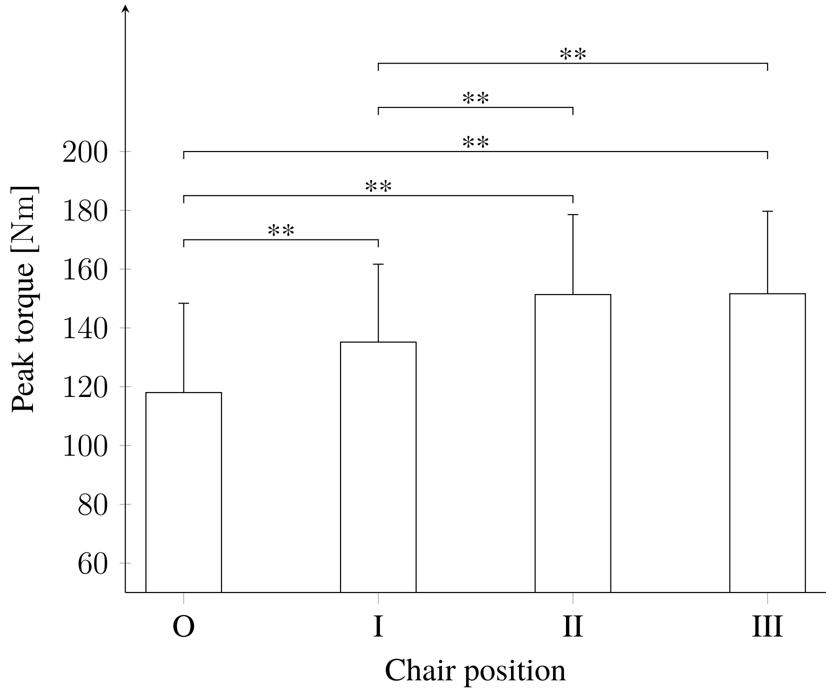


Figure 5.8: Measured peak torques at the ankle joint at maximal voluntary contractions (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .176, $F_{3,39} = 60.683$, $p < .000$ and $\eta^2 = .824$.
**: Statistical significance level of $p < .01$.

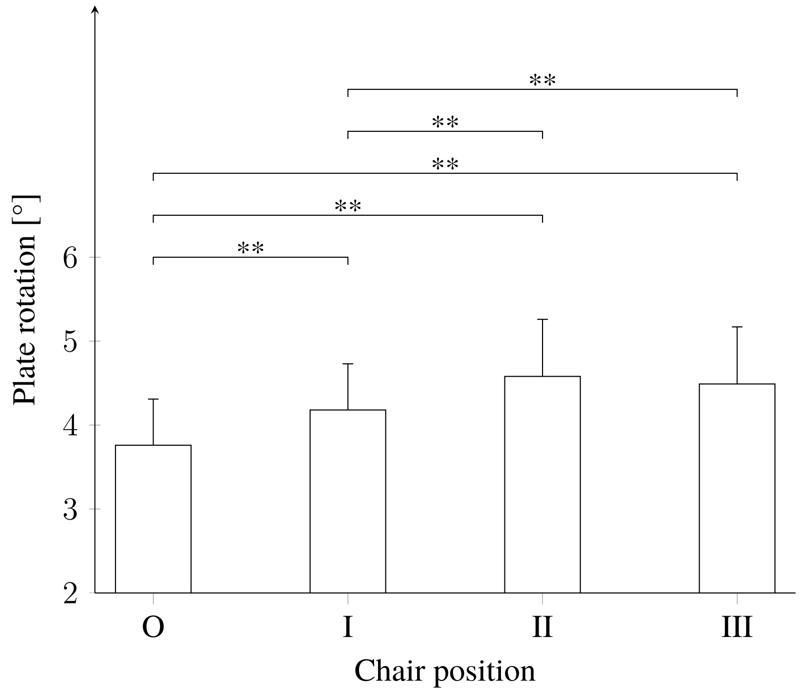


Figure 5.9: Dynamometer foot adapter plate rotation at MVC (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .273, $F_{3,39} = 34.687$, $p < .000$ and $\eta^2 = .727$.
**: Statistical significance level of $p < .01$.

5.5 Point of force application measurements

Points of force application were determined from data recorded by the pressure distribution measuring insole. At the onset of contractions statistically significant differences were found at position O compared to all other groups (Figure 5.10). While at MVC significant differences were found between position III compared to all others as well as between position O and II (Figure 5.11).

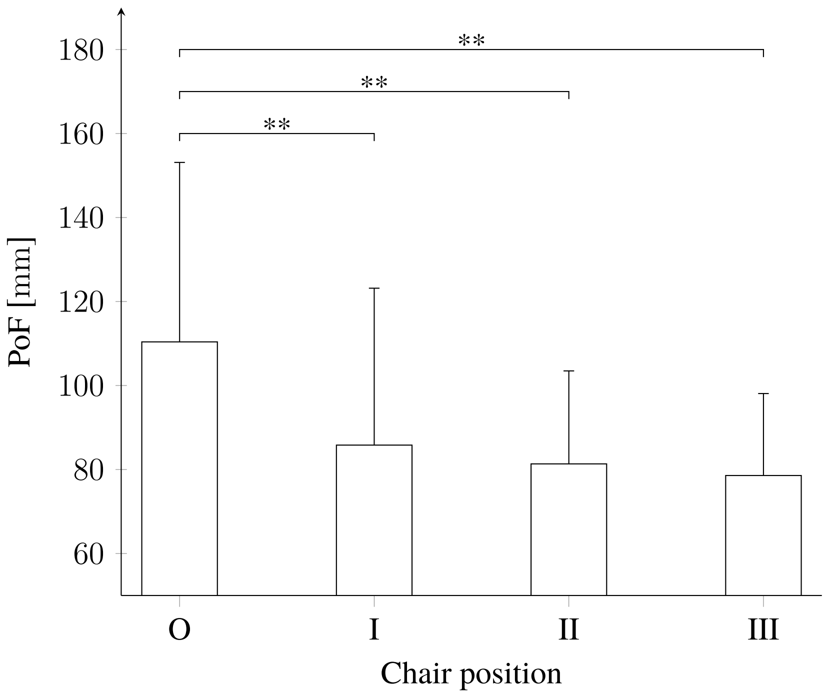


Figure 5.10: Point of force application on the dynamometer foot adapter plate at the onset of contractions (mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .500, $F_{3,39} = 13.012$, $p < .000$ and $\eta^2 = .500$.
**: Statistical significance level of $p < .01$.

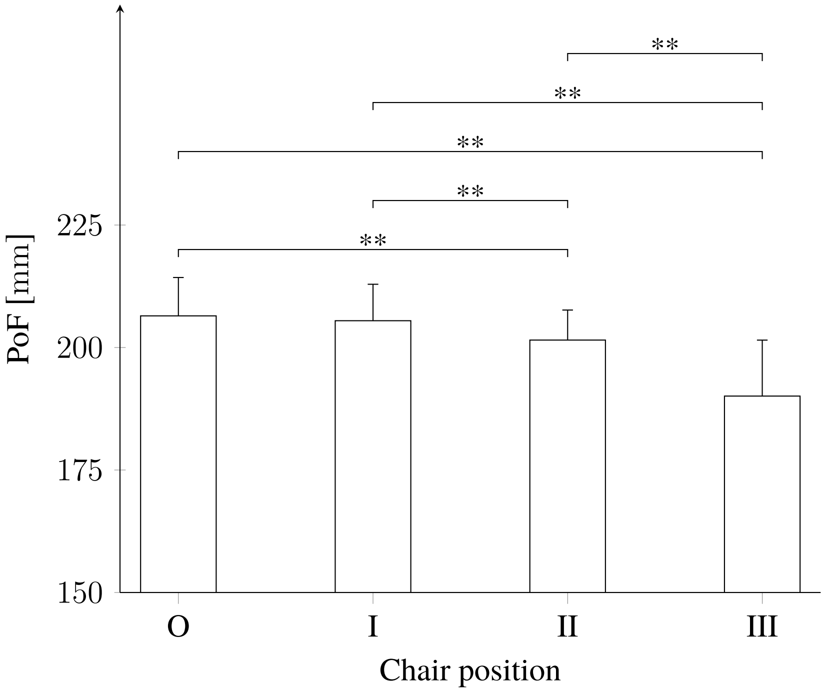


Figure 5.11: Point of force application at MVC
(mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position:
Wilk's Lamda = .355, $F_{3,39} = 23.636$, $p < .000$ and $\eta^2 = .645$.
**: Statistical significance level of $p < .01$.

5.6 Pressure measurements

In addition to point of force application, the peak pressure exerted on the dynamometer ankle adapter was measured with the insole placed on the footplate. As shown in Figure 5.12, significant differences were found in between all compared position groups, with values increasing for every position closer to the dynamometer axis. The biggest difference appeared comparing II with III where peak pressure increased about 120 kPa.

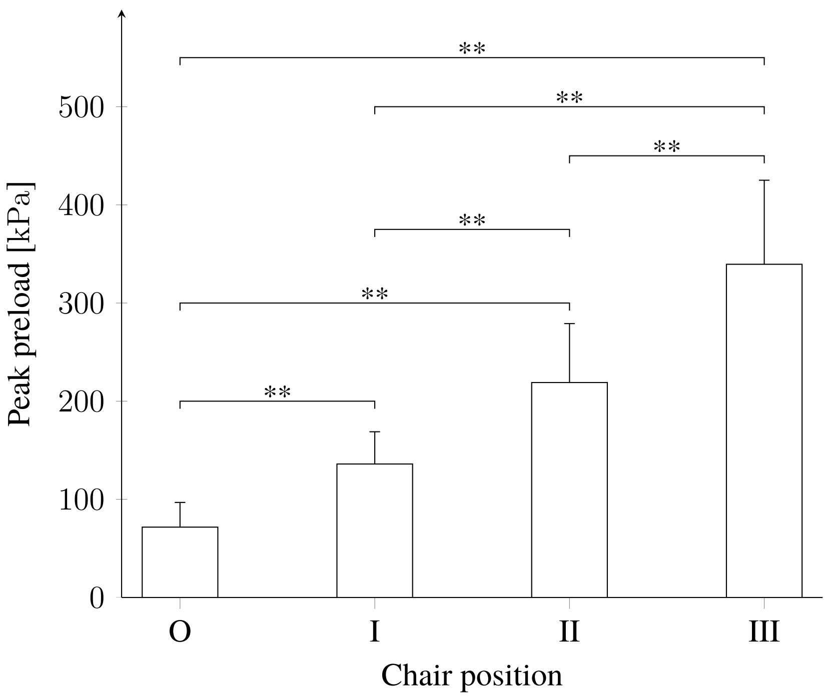


Figure 5.12: Peak preload applied on the dynamometer footplate at onset of contractions (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .062, $F_{3,39} = 196.005$, $p < .000$ and $\eta^2 = .938$.
**: Statistical significance level of $p < .01$.

5.7 Moment arm measurements

With knowledge of the point of force application together with the line defined by the 2 footplate markers, moment arm lengths of the dynamometer and ankle could be determined from the kinematic data.

At the onset of contractions significant differences were found at position O compared to all other groups (Figure 5.13). While at MVC significant differences were found between position III compared to all other chair positions, as illustrated in Figure 5.14.

Also for the ankle moment arm length, chair position dependent differences were found to be statistically significant only for position O compared to the other positions (Figure 5.15). As for MVC Figure 5.16 shows a slightly different result that for then observed dynamometer moment arm. In addition to significant differences at position III compared to the other group a statistically significant difference was found between chair position O and I.

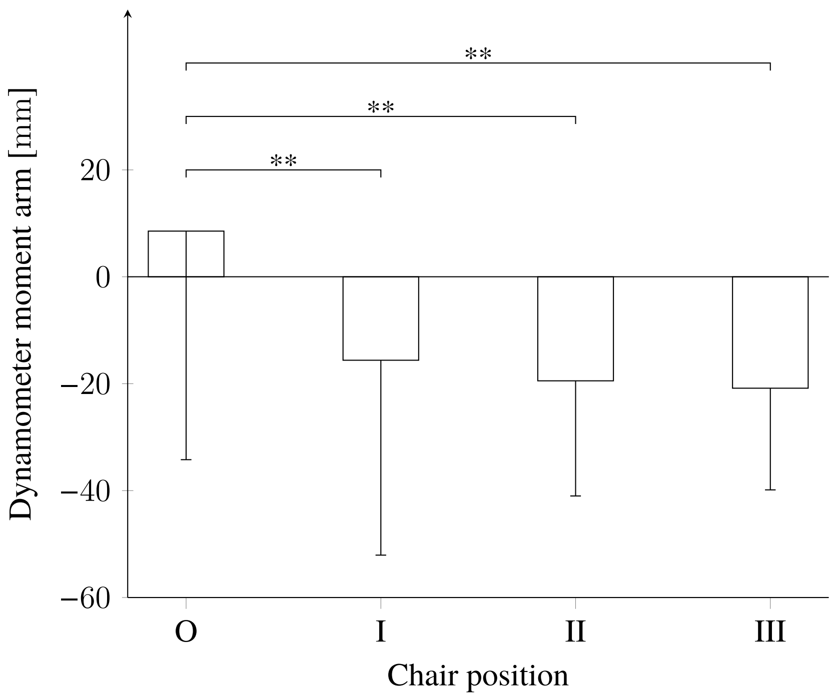


Figure 5.13: Dynamometer moment arm length at the onset of contractions (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .525, $F_{3,39} = 11.763$, $p < .000$ and $\eta^2 = .475$.
**: Statistical significance level of $p < .01$.

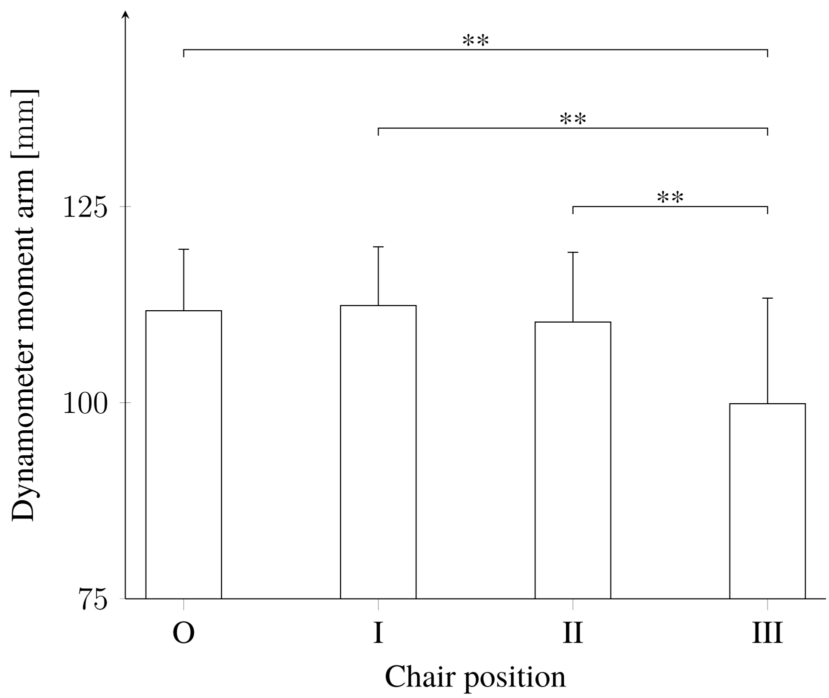


Figure 5.14: Dynamometer moment arm length at MVC (mean ± SD, $n = 42$).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .546, $F_{3,39} = 10.822$, $p < .000$ and $\eta^2 = .454$.
**: Statistical significance level of $p < .01$.

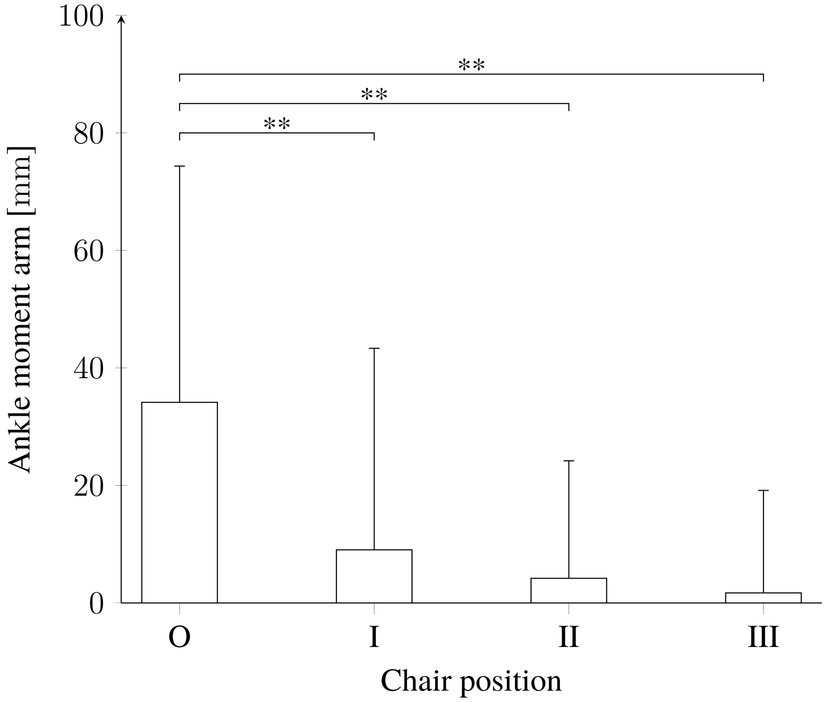


Figure 5.15: Ankle moment arm length at the onset of contractions (mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .486, $F_{3,39} = 13,730$, $p < .000$ and $\eta^2 = .514$.
**: Statistical significance level of $p < .01$.

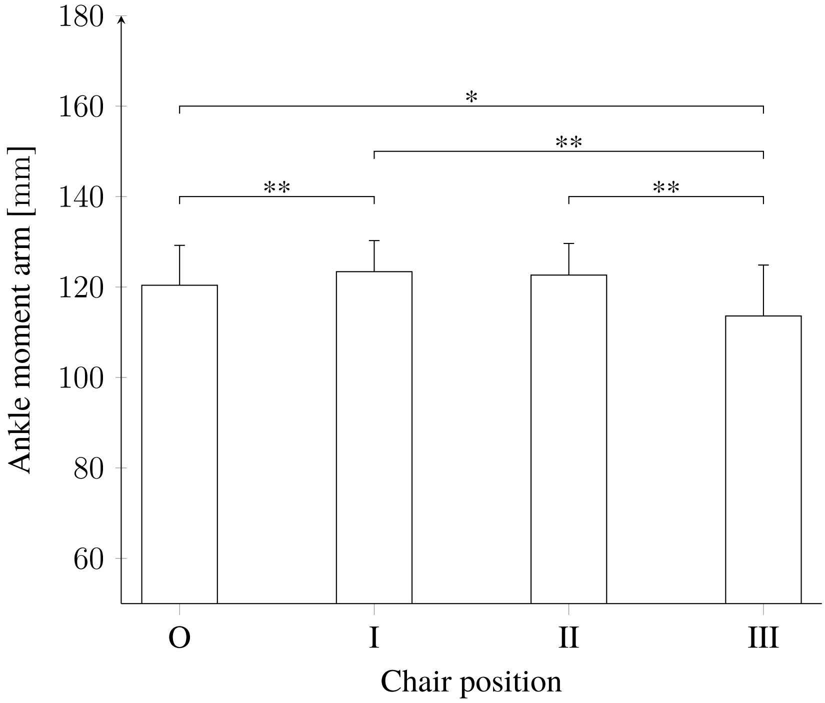


Figure 5.16: Moment arm of the reaction force at the foot to the ankle joint during a maximal voluntary plantar flexion contractions (mean ± SD, n = 42).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .468, $F_{3,39} = 14.753$, $p < .000$ and $\eta^2 = .532$.
*: Statistical significance level of $p < .05$.
**: Statistical significance level of $p < .01$.

5.8 Peak rate of torque development measurements

The results for peak rate of torque development (RTD_{max}) are shown in Figure 5.17. As mentioned in Section 4.3, data was acquired from the additional trials of fastest possible voluntary contraction plantar flexion efforts. Statistically significant differences were only found between position O and II. It is at the latter position the highest RTD_{max} levels were achieved. Measuring at the closest position to the dynamometer axis (III) did not further improve but reduce values, not significantly though.

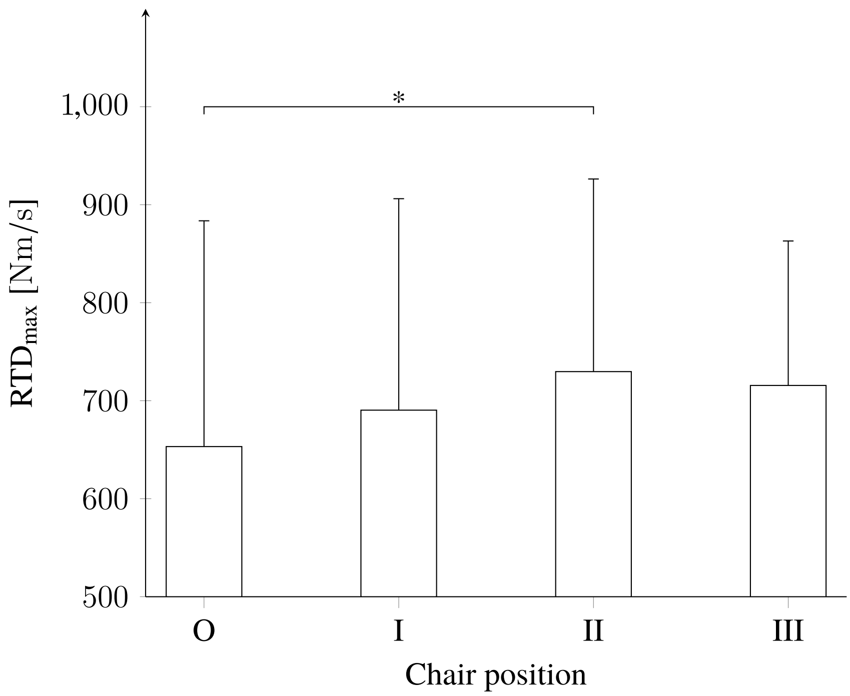


Figure 5.17: RTD_{max} during trials of fastest possible voluntary contractions (mean ± SD, n = 28).
Note. On the examined parameter there was a statistically significant effect for position: Wilk's Lamda = .709, $F_{3,25} = 3.419$, $p < .033$ and $\eta^2 = .291$.
*: Statistical significance level of $p < .05$.

6 Discussion

6.1 Main findings

The main finding in this study is the significant effect of subject's positioning on the exerted torque as well as other mechanical parameters. Specifically the exerted torque at position II and III ($6\,\mathrm{cm}$ and $8.3\,\mathrm{cm}$ respectively) was significantly greater (about $28\,\%$) than at the reference position (O). Also at position I the torque increase was $\approx 14\,\%$ compared to position O, indicating that adequate pressure on the dynamometer footplate is needed in order to acquire more accurate torque results. Another important aspect emerges from the fact that no significant difference was found between exerted torque at position II and III. Positioning the subject at the closest tolerable distance (III) might be therefore unnecessary and Position II (with $6\,\mathrm{cm}$) sufficient.

It was reported (Arampatzis, Morey-Klapsing, et al., 2005) that due to the ankle rotation and the misalignment of the joint and dynamometer axis the corrected torque could reach $6\,\%$ to $10\,\%$ higher values. It appears that subject positioning would have a bigger impact on the examined parameter irrespective of the axis misalignment. In the present study, due to the tightened positioning of the subject, ankle joint rotation at reference position $15.5° \pm 4.0°$ reduced compared to position II and III by $\approx 40\,\%$ and $\approx 55\,\%$ respectively. Nevertheless, ankle joint rotation established at reference position is in accordance to data ($13.8° \pm 5.8°$) previously reported via an almost similar study design (Karamanidis et al., 2005). In comparison, results from another study Rosager et al. (2002) showed considerably less ankle joint rotation with $\approx 3°$ during maximal plantar flexion effort. These difference between studies can be attributed to the different devices utilized to first measure force and joint rotation. Rosager et al. (2002) used a rigid steel frame to position the subject and monitored joint angular change with an electrical goniometer. A custom made frame may have more rigid parts compared to isokinetic dynamometers worldwide available since these also regard comfort. In the present study the non rigidity of the isokinetic dynamometers could be also explained by the dynamometer's plate rotation during maximal contraction which ranged from $3.7°$ to $4.6°$. This result indicates that the necessary corrections for joint rotation could start at approximately $4°$ onwards, depending on the device in use.

Additionally it is reasonable to assume that less movement of the ankle joint would actuate less misalignment of both axes and therefore the $6\,\%$ to $10\,\%$ torque increase would no longer be realistic. This outcome must be investigated in future studies and analyze the portion of every correction to the total exerted torque. However, the lesser joint rotation has more important implication for estimation of other mechanical parameters like the fascicle length, fascicle pennation angle and the tendon or aponeurosis displacement. In many studies (Albracht & Arampatzis, 2013; Bayliss et al., 2016) the aim of the researchers was to examine an intervention (training, detraining) regime in the mechanical and morphological properties of the muscle tendon unit. It has been reported (Karamanidis et al., 2005) that due to the ankle joint rotation, fascicle length was overestimated by $1.53\,\mathrm{cm}$ and the pennation angle by about $5.5°$. Moreover the authors performed a correction consisting of measured fascicle lengths and pennation angles acquired by passive joint rotation and concluded that not accounting for ankle joint rotation could result to erroneous findings. In the present study the joint rotation was reduced by $\approx 50\,\%$ due to the tightened positioning of the subjects and therefore it can be expected that also the overestimation of the fascicle length and pennation angle might be reduced by an almost similar amount. However, this hypothesis needs to be examined in future studies and assert the possible alterations of the muscle fascicles architecture and its implication on the force production capability of the muscle. In a similar manner the tendon displacement during a maximal voluntary contraction is in part due to the joint rotation and in part due to the forces acting on the tendon. The less ankle joint rotation in combination with the higher exerted torque could increase the calculated stiffness of the tendon/aponeurosis and possibly affect its energy dissipation in the unload phase. Again such hypotheses need to be experimental validated in order to determine the possible error in the estimation of the tendon's mechanical properties.

6.2 Positioning

In the literature (Karamanidis et al., 2005; Arampatzis, Morey-Klapsing, et al., 2005) it is often mentioned that the positioning of the subject on the dynamometer chair is performed subjectively and in some cases the ankle joint is secured with inextensible straps. The latter should prevent the ankle from rotating but will inevitably restrain the ankle motion and therefore the exerted torque will not be the actual one. Other strategies are needed to overcome that issue. In the present study ankle joint securing was avoided and it is therefore most likely that the real exerted torque was acquired, despite other probable influencing factors. In order to restrain the joint motion the pressure of the subject's foot sole on the dynamometer's plate (Figure 5.12) was increased by a factor of ≈ 5 at closest position (Factor $1.9, 3.1, 4.7$; position I, II, III respectively) measured as peak pressure from the pedar pressure measuring system. This adjustment significantly ($p < .05$) changed the point of force application at the onset of the maximal plantar flexion contraction by moving it more posterior (from $110.4\,\mathrm{mm}$ to $85.8\,\mathrm{mm}$, $81.3\,\mathrm{mm}$ and $78.6\,\mathrm{mm}$ for the position O, I, II, III respectively, see Figure 5.10) again measured with the pressure insole. Such an outcome would be ankle joint angle dependent suggesting that at more dorsiflexed angles the point of force application would be moved anteriorly and vice versa. This result appears to have no effect on the torque outcome but nevertheless is an indication of alterations made by the different subject positioning. On the opposite, the point of force application at the instance of maximal torque (Figure 5.11) showed a significant ($p < .001$) decrease for the more compressed ($\approx 8\,\%$ reduction, from $206.5\,\mathrm{mm}$ to $190.1\,\mathrm{mm}$ for O and III respectively) position.

6.3 Moment arm lengths

It is noteworthy that the closest (smallest) point of force application was shown in the more compressed position. This result combined with the ankle moment arm length (Figure 5.16) allows to calculate the actual force acting on the tendon or the force exerted by the triceps surae muscles. Since the same torque levels were reached for position II and III, and due to the smaller ankle moment arm at position III; one could assume an even greater force potential of the plantar flexor muscles at MVCs. For example, the reaction force in the third (III) position with the maximal exerted torque ($151.6\,\mathrm{N\,m}$) would be $1334.8\,\mathrm{N}$ with a moment arm of $113.6\,\mathrm{mm}$ or $1259.4\,\mathrm{N}$ with a moment arm of $120.4\,\mathrm{mm}$. Yet actual causes for those differences most probably arise from various effects combined.

Force-length or force-velocity relationships are prominent and often aimed for muscle properties (Maganaris, 2003; Herzog et al., 1991). But also the muscle's very own moment arms influence the resultant exerted torque. In the study of Maganaris, Baltzopoulos, and Sargeant (1998), Achilles tendon moment arm length systematically increased (from $5.4\,\mathrm{cm}$ to $7\,\mathrm{cm}$) with increasing ankle angle (from $-15°$ to $30°$, with neutral ankle position defined at $0°$), both at rest and during MVC. In that case, force development at position III would have been even greater, because of the less plantar flexion and the corresponding smaller Achilles tendon moment arm.

Concerning the force output of the triceps surae muscle Nourbakhsh and Kukulka (2004) came to the result that while plantar flexion of the ankle across their experimental conditions, gastrocnemii length did not have a significant impact on the pattern of the TS EMG activity, no matter if muscle length was held constant or shortened. Changes in the Achilles tendon moment arm, on the contrary, seemed to predominate over the muscle length variation. “Despite a constant gastrocnemii muscle length, the maximum plantar flexion torque developed in the TS decreased by about $50\,\%$ as the Achilles tendon MA decreased from its longest to shortest length”(Nourbakhsh & Kukulka, 2004, p. 271). Although in that case a contribution due to the change in Achilles tendon MA length would be marginal because of the small difference in ankle angle between position II and III. Even those findings considered, the influencing impact is not sufficient to explain potential differences in force potential between measurements the compressed Position (II and III).

6.4 Joint axes misalignment

The method proposed by Karamanidis et al. (2005) was used for correcting the measured torque developed of the subjects. Although at position II and III the exerted torque showed no significant ($p > .05$) differences. Due to the different dynamometer and ankle joint moment arms the corrected torque developed from $118.0, 135.2, 151.3$ and $151.6\,\mathrm{N\,m}$ to $127.2, 148.4, 168.3$ and $172.5\,\mathrm{N\,m}$ for the O, I, II, III position respectively. The percentage increase of torque after the correction was $7.8, 9.8, 11.2$ and $13.7\,\%$ for the 4 positions respectively. This finding is in contrast to the one by Arampatzis, Morey-Klapsing, et al. (2005) who reported a lower corrected torque. This discrepancy can be attributed to the different subject positioning at the beginning of the contractions in the present study. A closer look at the moment arm values reveals a systematic misalignment of ankle joint and dynamometer axis, despite careful considerations. As indicated by the negative sign, participant’s ankles had been constantly placed about $24\,\mathrm{mm}$ posterior to the dynamometer axis. Despite non-negligible ankle joint rotation this resulted in a greater MA at the ankle joint than at the dynamometer axis while MVC. Consequently this led to an underestimation of the “true” peak torque potential. Although care was taken to align the ankle joint axis with the axis of the dynamometer, this was evidently not possible.

A different, yet affordable method would be necessary to spot any misalignment at participant’s positioning instantly. In the present study changes in muscle-tendon architecture or inner joint motion throughout the contractions could not be monitored. Future work might reveal promising findings with medical imaging devices to additionally monitor maximal plantar flexion efforts. For one example Tsaopoulos, Baltzopoulos, Richards, and Maganaris (2011) used the rather expensive method of real-time X-ray videos to compare errors when moments were calculated using measurements from external anatomical surface markers or obtained from the isokinetic dynamometer. Investigating internal knee angle rotation from rest to maximal contraction, nonnegligible differences could been established. This method of videofluoroscopy is also often used for in vivo estimation of total ankle arthroplasties kinematics (List et al., 2012). If system integration with torque measuring devices could be feasible, it would provide kinematic data at high precision without being limited by skin movement artifacts.

Another known shortcoming of the applied study design lies in the assumption that the direction of the reaction force is perpendicular to the dynamometer lever. Yet it seems that potential influences of shear forces contribution on the measured and resultant moment are negligible Arampatzis, Morey-Klapsing, et al. (2005), there is new studies tackling this issue with help of a custom designed dynamometer. Results from Toumi et al. (2015) confirmed that proper capacity of plantar flexion torque production is significantly underestimated in a 1 dimensional consideration compared to measurements done with a tridimensional torque sensor. With their novel dynamometer design Toumi et al. (2015) also compared measurements between a locked-unit (participant is restrained within the unit) and open-unit (participant's position is independent of the ankle dynamometer) configuration. In their conclusion the authors suggest to measure at the latter setting in order to establish plantar flexion torque that is exclusive of accessory muscle but inclusive of all ankle joint movements.

Nevertheless misalignment of both axes, although constant, confirms the suggestions of previous studies (Rosager et al., 2002; Karamanidis et al., 2005; Arampatzis, Morey-Klapsing, et al., 2005) for implementing corrections for all (moment arm, exerted torque) examined parameters. Nonetheless this shortfall would not affect the main outcome of this study which is the increased torque output with tightener subject position.

6.5 RTD, hip and knee angel variations

It was hypothesized that with more rigid dynamometer, due to tightener positioning, the rate of torque development would also be increased in the more tense position (III). Our hypothesis could not be confirmed since only at position II a significant (p = .017) greater rate was found compared to the reference position (O). One can only speculate about that result by indicating the significant different ankle joint deviations among positions. It appears that it could be an optimum in ankle joint motion in order to achieve higher rates of torque development. But again this is a speculation and that issue needs further attention and should be examined in future studies.

The hip joint angles in the present study showed a significant difference not only in the onset of every position (Figure 5.1) but at maximum contraction also (Figure 5.2). The difference at the onset although small in magnitude can be attributed to the step wise posterior displacement of trochanter marker due to the repositioning of the subject. The increased pressure on the dynamometer's footplate also pushed the trochanter in a more posterior position which inevitably altered the hip joint angle. The latter can be seen at dynamic contraction likewise, where hip joint angle was altered by a similar amount. Although significant differences between positions were found it was presumed that $\approx 3°$ hip angle deviation between positions would not affect the main outcome of this study. Similar to the hip joint the knee joint was also minor but significantly affected by the different joint positions. Again the absolute differences ($\approx 1°$) were too small to change the architectural characteristics of both gastrocnemii and further affect their force potential.

The aim of this study was to examine the effect of subject positioning on the exerted torque and probably find a method that would quantitatively and objectively standardize the subject positioning. This was achieved via the pressure insoles which clearly showed that considerable preload pressure is needed for maximizing the exerted torque and to minimize joint angle motion. However, such apparatus is costly, complicated and not commonly available. By examining the knee joint angle (Figure 5.3) at rest, an alternative approach was conceptualized. It appears that a rest knee joint angle of 146° to 141° would suffice in order to increase the pressure on the dynamometer plate and remove the majority of the dynamometer-arm-foot-system's elasticity. Nonetheless, such a suggestion is limited to the dynamometers with similar elasticity as the one used in the present study. This is crucial to be validated in future research with other systems that have more or less elasticity than the system used in the present study.

6.6 Participant's sensibility for positioning comfort

Participant's verbal feedback about their sensation at the different measurement positions was examined throughout the study. It seems that for potion O, I, & II, no considerable discomfort was noticed. Yet the individually closest tolerable position (III) was described as rather unusual and a little discomforting. However, no feeling of pain was mentioned at any given moment. For some participants, keeping their left foot placed on the dynamometer ankle adapter for too long resulted in a feeling numbness. Taken into consideration, volunteers therefore walked a few steps on the laboratory floor in between position changes. Most of the time this was enough to prevent the participant's leg to fall asleep.

6.7 Limitations and future direction

In our study we did not account the probable activation of the antagonist tibialis anterior in the resulted exerted torque since it could have a considerable portion (Arampatzis, Morey-Klapsing, et al., 2005) in the development of the overall torque. It has been shown that the influence of the antagonist muscles in the torque development is in average $4.3\,\%$ (Arampatzis, Morey-Klapsing, et al., 2005) but the influence of the subject positioning in the present study was significantly higher, almost the 6.5-fold compared to the coactivation. Nevertheless it can not be excluded that the subjects positioning would affect not only the activation of the agonist but also of the antagonist due to the less ankle joint rotation or due to the higher exerted torque. It is therefore advisable to examined in future studies their activation levels in order to establish a more accurate insight of their working mechanisms. We also did not account for any inertial impact on the resulted torque since at isometric contractions the influence of the inertial component would be marginal. Another important aspect is the direction of the point of force application. For present study it was assumed that the force vector was perpendicular to the plate surface since there was no other technical means for accurate estimation. It is possible that the force vector was not perpendicular but it had an angle to the surface and therefore the estimation of the actual ankle joint moment arm could be under- or overestimated. This could be apparent in more dorsiflexed or plantar flexed joint angles, where greater anterior-posterior forces could be acting. Nevertheless this issue should be examined in future studies by means of a small force plate attached to the dynamometer plate. This device could parallel measure the anterior-posterior and medio-lateral force components in order to more precisely estimate ankle joint moment arms at various stages of the voluntary plantar flexion contraction.

7 Conclusion

It can be concluded that tightener subject positioning may significantly increase the exerted torque, alter the ankle joint moment arm and reduce joint rotation during MVC. For measuring true peak torque values there is evidence to suggest a dynamometer configuration with participant's knee at rest enclosing an angle of about $143°$. Further investigations are needed to establish misalignment prevention methods and to examine the influence of the aforementioned alterations on the electro-mechanical and morphological properties of the human muscle-tendon unit. Nonetheless this study might contribute to an improved plantar flexion torque determination in the future.

Bibliography

Albracht, K. & Arampatzis, A. (2013). Exercise-induced changes in triceps surae tendon stiffness and muscle strength affect running economy in humans. *Eur J Appl Physiol*, *113*(6), 1605–1615.

Arampatzis, A., Karamanidis, K., Stafilidis, S., Morey-Klapsing, G., DeMonte, G., & Brüggemann, G.-P. (2006). Effect of different ankle- and knee-joint positions on gastrocnemius medialis fascicle length and EMG activity during isometric plantar flexion. *J Biomech*, *39*(10), 1891–1902.

Arampatzis, A., Morey-Klapsing, G., Karamanidis, K., DeMonte, G., Stafilidis, S., & Brüggemann, G.-P. (2005). Differences between measured and resultant joint moments during isometric contractions at the ankle joint. *J Biomech*, *38*(4), 885–892.

Arampatzis, A., Stafilidis, S., DeMonte, G., Karamanidis, K., Morey-Klapsing, G., & Brüggemann, G. P. (2005). Strain and elongation of the human gastrocnemius tendon and aponeurosis during maximal plantarflexion effort. *J Biomech*, *38*(4), 833–841.

Barnett, C. H. & Napier, J. R. (1952). The axis of rotation at the ankle joint in man. Its influence upon the form of the talus and the mobility of the fibula. *J Anat*, *86*(1).

Bayliss, A. J., Weatherholt, A. M., Crandall, T. T., Farmer, D. L., McConnell, J. C., Crossley, K. M., & Warden, S. J. (2016). Achilles tendon material properties are greater in the jump leg of jumping athletes. *J Musculoskelet Neuronal Interact*, *16*(2), 105.

Bickers, M. J. (1993). Does verbal encouragement work? The effect of verbal encouragement on a muscular endurance task. *Clinical Rehabilitation*, *7*(3), 196–200.

Brockett, C. L. & Chapman, G. J. (2016). Biomechanics of the ankle. *Orthop Trauma*, *30*(3), 232–238.

Butler, D. L., Grood, E. S., Noyes, F. R., & Zernicke, R. F. (1978). Biomechanics of ligaments and tendons. *Exerc Sport Sci Rev*, *6*, 125–181.

Cheung, J. T.-M. (2008). Biomechanical modelling and simulation of foot and ankle. In Y. Hong & R. Bartlett (Eds.), *Routledge Handbook of Biomechanics and Human Movement Science* (pp. 65–80). Abingdon, Oxon: Routledge.

Chèze, L. (2014). *Kinematic Analysis of Human Movement*. Hoboken, NJ: John Wiley & Sons, Inc.

Dao, T. T. & Tho, M.-C. H. B. (2014). *Biomechanics of the Musculoskeletal System: Modeling of Data Uncertainty and Knowledge*. Hoboken, NJ: John Wiley & Sons, Ltd.

Donatelli, R. (Ed.). (1990). *The Biomechanics of the Foot and Ankle*. Philadelphia, PA: F. A. Davis Company, 3. Contemporary Perspectives in Rehabilitation.

Haskell, A. & Mann, R. A. (2008). Biomechanics of the Foot. In J. D. Hsu, J. W. Michael, & J. R. Fisk (Eds.), *AAOS Atlas of Orthoses and Assistive Devices* (4th ed.). Philadelphia, PA: Mosby.

Haskell, A. & Mann, R. A. (2014). Biomechanics of the Foot and Ankle. In M. J. Coughlin, C. L. Saltzman, & R. B. Anderson (Eds.), *Mann's Surgery of the Foot and Ankle* (9th ed., pp. 3–36). Philadelphia, PA: Elsevier Saunders.

Hennig, E. M. & Lafortune, M. A. (1997). Technology and application of force, acceleration and pressure distribution measurements in biomechanics. In P. Allard, A. Cappozzo, A. Lundberg, & C. L. Vaughan (Eds.), *Three-dimensional Analysis of Human Locomotion* (2, pp. 109–127). International Society of Biomechanics Series. Chichester, West Sussex: John Wiley & Sons, Ltd.

Herzog, W. (1988). The relation between the resultant moments at a joint and the moments measured by an isokinetic dynamometer. *J Biomech*, *21*(1), 5–12.

Herzog, W., Read, L., & Ter Keurs, H. (1991). Experimental determination of force—length relations of intact human gastrocnemius muscles. *Clin Biomech*, *6*(4), 230–238.

Hicks, J. H. (1953). The mechanics of the foot: I. The joints. *J Anat*, *87*(4), 345–357.

Hoy, M. G., Zajac, F. E., & Gordon, M. E. (1990). A musculoskeletal model of the human lower extremity: The effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle. *J Biomech*, *23*(2), 157–169.

Isman, R. E. & Inman, V. T. (1969). Anthropometric studies of the human foot and ankle. *Bull Prosthet Res*, *10-11*, 97–129.

Jastifer, J. R. & Gustafson, P. A. (2014). The subtalar joint: Biomechanics and functional representations in the literature. *Foot*, *24*(4), 203–209.

Kapandji, I. A. (2006). *Funktionelle Anatomie der Gelenke. Schematisierte und kommentierte Zeichnungen zur menschlichen Biomechanik. Bd. 2: Untere Extremität.* (4., unveränd. Aufl.; einbd. Ausg). Stuttgart: Thieme.

Karamanidis, K., Stafilidis, S., DeMonte, G., Morey-Klapsing, G., Brüggemann, G.-P., & Arampatzis, A. (2005). Inevitable joint angular rotation affects muscle architecture during isometric contraction. *J Electromyogr Kinesiol*, *15*(6), 608–616.

Klenerman, L. & Wood, B. (2006). *The Human Foot: A Companion to Clinical Studies*. London: Springer-Verlag London Ltd.

Knudson, D. (2007). *Fundamentals of Biomechanics* (2nd ed.). New York: Springer Science+Business Media LLC.

Knuttgen, H. G. & Komi, P. V. (2003). Basic Considerations for Exercise. In P. V. Komi (Ed.), *Strength and Power in Sport* (pp. 3–7). Oxford, UK: Blackwell Science Ltd.

Kubo, K., Kanehisa, H., & Fukunaga, T. (2002). Effect of stretching training on the viscoelastic properties of human tendon structures in vivo. *J Appl Physiol*, *92*(2), 595–601.

Kubo, K., Kanehisa, H., & Fukunaga, T. (2005). Comparison of elasticity of human tendon and aponeurosis in knee extensors and ankle plantar flexors in vivo. *J Appl Biomech*, *21*(2), 129–142.

Lang, J. & Wachsmuth, W. (Eds.). (1972). *Bein und Statik*. Springer Berlin Heidelberg, *4*. Praktische Anatomie.

List, R., Foresti, M., Gerber, H., Goldhahn, J., Rippstein, P., & Stüssi, E. (2012). Three-dimensional kinematics of an unconstrained ankle arthroplasty: A preliminary *in vivo* videofluoroscopic feasibility study. *Foot Ankle Int*, *33*(10), 883–892.

Lundberg, A. (1997). Functional Anatomy. In P. Allard, A. Cappozzo, A. Lundberg, & C. L. Vaughan (Eds.), *Three-dimensional Analysis of Human Locomotion* (2, pp. 27–47). International Society of Biomechanics Series. Chichester, West Sussex: John Wiley & Sons, Ltd.

Lundberg, A., Svensson, O., Nemeth, G., & Selvik, G. (1989). The axis of rotation of the ankle joint. *J Bone Joint Surg Br. 71*(1), 94–99.

Maganaris, C. N. (2002). Tensile properties of in vivo human tendinous tissue. *J Biomech*, *35*(8), 1019–1027.

Maganaris, C. N. (2003). Force-length characteristics of the in vivo human gastrocnemius muscle. *Clin Anat*, *16*(3), 215–223.

Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1998). Changes in Achilles tendon moment arm from rest to maximum isometric plantarflexion: *In vivo* observations in man. *J Physiol*, *510*(3), 977–985.

Magnusson, S. P., Aagaard, P., Rosager, S., Dyhre-Poulsen, P., & Kjaer, M. (2001). Load-displacement properties of the human triceps surae aponeurosis *in vivo*. *J Physiol*, *531*(1), 277–288.

Martin, R. L. (2011). The Ankle and Foot Complex. In P. K. Levangie & C. C. Norkin (Eds.), *Joint Structure and Function: A Comprehensive Analysis* (5th ed., pp. 440–481). Philadelphia, PA: F. A. Davis Company.

Menegaldo, L. L. & Oliveira, L. F. d. (2009). Effect of muscle model parameter scaling for isometric plantar flexion torque prediction. *J Biomech*, *42*(15), 2597–2601.

Nigg, B. M. & Herzog, W. (2007). Joints. In B. M. Nigg & W. Herzog (Eds.), *Biomechanics of the Musculo-skeletal System* (3rd ed., pp. 244–259). Chichester: John Wiley & Sons, Ltd.

Nourbakhsh, M. R. & Kukulka, C. G. (2004). Relationship between muscle length and moment arm on EMG activity of human triceps surae muscle. *J Electromyogr Kinesiol*, *14*(2), 263–273.

Rodgers, M. M. & Cavanagh, P. R. (1984). Glossary of Biomechanical Terms, Concepts, and Units. *Phys Ther*, *64*(12), 1886–1902.

Rosager, S., Aagaard, P., Dyhre-Poulsen, P., Neergaard, K., Kjaer, M., & Magnusson, S. P. (2002). Load-displacement properties of the human triceps surae aponeurosis and tendon in runners and non-runners. *Scand J Med Sci Sports*, *12*(2), 90–98.

Schünke, M., Schulte, E., & Schumacher, U. (Eds.). (2014). *Allgemeine Anatomie und Bewegungssystem: LernAtlas der Anatomie*. Stuttgart: Georg Thieme Verlag.

Siegler, S., Chen, J., & Schneck, C. D. (1988). The three-dimensional kinematics and flexibility characteristics of the human ankle and subtalar joints—Part I: Kinematics. *J Biomech Eng*, *110*(4), 364–373.

Simoneau, E. M., Longo, S., Seynnes, O. R., & Narici, M. V. (2012). Human muscle fascicle behavior in agonist and antagonist isometric contractions. *Muscle Nerve*, *45*(1), 92–99.

Stafilidis, S. & Arampatzis, A. (2007). Muscle-tendon unit mechanical and morphological properties and sprint performance. *J Sports Sci*, *25*(9), 1035–1046.

Toumi, A., Jakobi, J. M., & Simoneau-Buessinger, E. (2016). Differential impact of visual feedback on plantar- and dorsi-flexion maximal torque output. *Appl Physiol Nutr Metab*, *41*(5), 557–559.

Toumi, A., Leteneur, S., Gillet, C., Debril, J.-F., Decoufour, N., Barbier, F., . . . Simoneau-Buessinger, E. (2015). Enhanced precision of ankle torque measure with an open-unit dynamometer mounted with a 3D force-torque sensor. *Eur J Appl Physiol*, *115*(11), 2303–2310.

Tsaopoulos, D. E., Baltzopoulos, V., Richards, P. J., & Maganaris, C. N. (2011). Mechanical correction of dynamometer moment for the effects of segment motion during isometric knee-extension tests. *J Appl Physiol*, *111*(1), 68–74.

Turpin, N. A., Costes, A., Villeger, D., & Watier, B. (2014). Selective muscle contraction during plantarflexion is incompatible with maximal voluntary torque assessment. *Eur J Appl Physiol*, *114*(8), 1667–1677.

Winter, D. A. (2009). *Biomechanics and Motor Control of Human Movement* (4th ed.). Hoboken, NJ: John Wiley & Sons, Inc.

Appendices

Erklärung

„Ich erkläre, dass ich die vorliegende Arbeit selbstständig verfasst habe und nur die ausgewiesenen Hilfsmittel verwendet habe. Diese Arbeit wurde weder an einer anderen Stelle eingereicht noch von anderen Personen vorgelegt.“

Wien, am ______________________ ______________________________

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Fremdsprachen

Englisch, Italienisch (Grundkenntnisse)

EDV

Betriebssysteme	UNIX (Linux Ubuntu), Windows
Sprachen	Pascal, TeX, Lua
Anwendungen	LibreOffice, QtiPlot, TeXworks, Vicon Nexus

Interessen Fahrrad, Fotografie, Philosophie, Schifahren

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Löffler, D. & Sickinger, C. (2014). Bewegung und Sport in der Volksschule und Kindergarten. *Bewegung und Sport, 68*(5), 25-27.

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