BIOMECHANICAL EVALUATION OF OVERUSE RISK FACTORS DURING RUNNING

Illustrated by the example of Achilles tendon injuries

DISSERTATION

Von der Fakultät für Wirtschafts- und Sozialwissenschaften der Universität Stuttgart zur Erlangung der Würde eines Doktors der Philosophie (Dr. Phil.) genehmigte Abhandlung

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ABSTRACT

INTRODUCTION Long distance runners often experience chronic overuse injuries and Achilles tendon injuries are known to be one of the most common physical complaints within this group. However, while several injury inducing factors were discussed in the literature, no clear proof of cause and effect could be settled. Excessive foot motion in the frontal plane and increased running speed are frequently contemplated as injury inducing factors in runners. Motion control shoes aim at limiting immoderate pronation and supination. During running repeated impacts are transferred in the axial direction along the lower leg, therefore possibly affecting the oscillation behavior of the Achilles tendon. Also, variations in neuromuscular control may lead to increased stresses and loadings at the Achilles tendon. Until now, no simultaneous measurements of plantar pressure as an impact load, muscle activity as a modulating system, foot kinematics as a load distributor and oscillations at the Achilles tendon as a loading case were performed in runners. The purpose of the thesis at hand was the development and implementation of a complex measurement system to assess parameter constellations, which may be associated with overuse injuries at the Achilles tendon in runners and which allow the differentiation between runners with Achilles tendon complaints and their healthy counterparts.

METHODS A series of measurements consisting of four separate studies was performed. Biomechanical parameters from plantar pressure, muscle activity, foot kinematics and oscillations at the Achilles tendon were evaluated while several external factors were varied. During the first two studies, the effects of different footwear modifications, variations in running speed and in ground condi-

tions (over ground versus treadmill) on the biomechanical parameters were investigated. During study 3, changes in these parameters were evaluated during a fatiguing one-hour run, representing a common long distance run. In the final study, a comparison between runners with Achilles tendon complaints and their healthy counterparts was performed. Plantar pressure was recorded using an in-shoe measurement insole (F-Scan) during the first two studies and using an instrumented treadmill during the last two studies. During all experiments, muscle activity was measured with a Noraxon Telemyo installation. An inertial measurement unit was utilized to determine foot kinematics in the frontal plane and oscillations at the Achilles tendon were recorded with two accelerometers.

RESULTS The selected sensor combination was found to be useful and reasonable for recording the biomechanical data described above. Running shoe modifications led to significant changes in kinematic variables, peak forces underneath the foot, vibrations at the Achilles tendon and muscle activity. Also, running speed had an influence on muscle activity, foot kinematics, and parameters describing oscillations at the Achilles tendon, even if small changes of only 0.6 m/s were applied. Changes in ground conditions (treadmill versus over ground) influenced peak forces underneath the foot and average muscle activity of M. Gastrocnemius lateralis in interaction with footwear modifications. However, foot kinematics as well as oscillations at the Achilles tendon did not differ if subjects ran over ground as compared to treadmill running. No specific changes in the investigated parameters could be found between distinct time instants during a one-hour run. But gait parameters, which were analyzed to control the state of exhaustion of the participating runners showed increasing fatigue over the course of the run. In the final study, very clear effects of Achilles tendon complaints were found in the evaluated data. In summary, study participants with Achilles tendon complaints showed higher total plantar force, decreased average muscle activity, increased maximum muscle activity, larger and faster trajectories of the foot in the frontal plane as well as clear differences in oscillations at the Achilles tendon in time and frequency space when compared to healthy runners.

DISCUSSION First insight was gained in the oscillation behavior occurring at the Achilles tendon and the simultaneous adaptations or causations of plantar pressure, muscle activity and foot kinematics in runners. External factors could be detected, which influence the investigated parameters. They may be applied to modify the investigated biomechanics. The chosen parameters met the demands to allow a differentiation between subjects with Achilles tendon complaints and healthy runners. An important part of the complex issue of overuse injuries at the Achilles tendon may be explained by the proposed parameter constellation, which seems to be superior to individual parameters. Future work should focus on early identification of runners who are at risk for Achilles tendon complaints to prevent injuries of this structure. Also, changes of the investigated external factors should be implemented in runners with Achilles tendon complaints in order to achieve a parameter constellation closer to that shown by healthy subjects. The achievement of a rehabilitative effect may be possible but requires confirmation through further prospective studies. No distinct conclusion can currently be drawn with regard to the cause-and-effect chain due to the retrospective design of the cross sectional study.

ZUSAMMENFASSUNG

EINLEITUNG Im Langstreckenlauf treten oftmals chronische Überlastungsfolgen auf, bei denen Achillessehnenverletzungen eine der häufigsten Beschwerden der Läufer darstellen. Während in der Vergangenheit eine Vielzahl verletzungsinduzierender Ursachen diskutiert wurde, fehlt weiterhin die eindeutige Darlegung einer schlüssigen Ursachen-Wirkungs-Kette. Übermäßige Bewegungen des Fußes in der Frontalebene sowie eine hohe Laufgeschwindigkeit werden oftmals als Ursachen von Laufverletzungen in Betracht gezogen. Daraufhin wurden spezielle Laufschuhe eingesetzt, welche darauf abzielen, Pronations- und Supinationsbewegungen auf einen gemäßigten Bewegungsumfang zu limitieren. Während des Laufens kommt es beim Aufsatz des Fußes auf dem Boden wiederholt zu Stoßkräften, welche in axialer Richtung entlang des Unterschenkels weitergeleitet werden. Diese Stoßkräfte haben möglicher Weise auch einen Einfluss auf das Schwingungsverhalten der Achillessehne. Darüber hinaus können Änderungen der neuromuskulären Kontrolle zu einer Zunahme der Belastungen und Beanspruchungen der Achillessehne führen. Bisher wurden noch keine simultanen Messungen der plantaren Druckverteilung als Maß der Stoßbelastung, der Muskelaktivität als modulierendes System, der Fußkinematik als Lastverteiler und der Schwingungen an der Achillessehne als Belastungsfall bei Läufern durchgeführt. Das Ziel der vorliegenden Arbeit war die Entwicklung und Anwendung eines komplexen Messsystems zur Untersuchung von Parameterkonstellationen, die im Zusammenhang mit Überlastungsfolgen an der Achillessehne im Laufsport stehen könnten und welche eine Differenzierung zwischen Läufern mit und ohne Achillessehnenbeschwerden zulassen.

METHODIK Zur Erreichung der oben beschriebenen Ziele wurden vier einzelne Studien durchgeführt. Hierbei wurden biomechanische Parameter der plantaren Druckverteilung, der Muskelaktivität und Fußkinematik sowie Oszillationen an der Achillessehne unter der Variation diverser externer Faktoren beurteilt. Während der ersten beiden Studien wurden dazu die Effekte von Laufschuhmodifikationen, Variationen der Laufgeschwindigkeit und des Untergrundes (Laufband versus über Grund) auf die biomechanischen Parameter untersucht. In der dritten Studie wurden die Veränderungen dieser Parameter während eines einstündigen, ermüdenden Laufs beobachtet. Mit diesem Setup sollte ein üblicher Ausdauerlauf simuliert werden. In der der abschließenden Studie wurde dann ein Vergleich zwischen Läufern mit Achillessehnenbeschwerden und einer gesunden Vergleichsgruppe durchgeführt. Die plantare Druckverteilung wurde in den ersten beiden Studien mittels eines einlagenbasierten Messsystems (F-Scan) im Schuh aufgezeichnet und während der letzten beiden Studien mittels eines instrumentalisierten Laufbandes quantifiziert. Die Muskelaktivität wurde in allen Studien mit Hilfe einer Noraxon Telemyo Anlage erfasst und Bewegungen des Fußmittels eines Inertialsensors bestimmt. Für die Ermittlung der Schwingungen an der Achillessehne wurden zwei Beschleunigungsaufnehmer genutzt.

ERGEBNISSE Die ausgewählte Sensorkombination erwies sich als zweckmäßig und sinnvoll zur Erhebung der oben beschriebenen biomechanischen Daten. Die Modifikation von Laufschuhen führte zu signifikanten Veränderungen kinematischer Variablen, wirkender Maximalkräfte unter dem Fuß, von Vibrationen an der Achillessehne sowie der muskulären Aktivität. Darüber hinaus konnte auch ein Einfluss der Laufgeschwindigkeit auf muskuläre Aktivität, Fußkinematik und Vibrationsparameter an der Achillessehne nachgewiesen werden. Dieser Einfluss konnte selbst für geringe Geschwindigkeitsänderungen von 0.6 m/s nachgewiesen werden. Variationen des Laufuntergrundes (Laufband versus über Grund) beeinflussten die wirkenden Maximalkräfte unter dem Fuß sowie die durchschnittliche Aktivität des M. Gastrocnemius lateralis in Interaktion mit Laufschuhmodifikationen. Jedoch veränderte sich weder die Fußkinematik, noch die Schwingungen and der Achillessehne wenn die Probanden über Grund im Vergleich zum Laufband liefen. Es konnten keine eindeutigen Änderungen der untersuchten Parameter zwischen einzelnen Zeitpunkten eines einstündigen Laufs ermittelt werden. Allerdings konnte anhand spezifischer Gangparameter, die evaluiert wurden um den Ermüdungsstatus der Versuchspersonen zu kontrollieren, festgestellt werden, dass entsprechend der Intention eine zunehmende Ermüdung während dem Lauf auftrat. Durch die zuletzt durchgeführte Studie zum Vergleich von Probanden mit Achillessehnenbeschwerden und gesunden Läufern wurden sehr klare Effekte der Achillessehnenbeschwerden auf die untersuchten Daten nachgewiesen. Zusammenfassend kann gesagt werden, dass die Beschwerdeläufer höhere Maximalkräfte unter dem Fuß, reduzierte durchschnittliche und erhöhte maximale Muskelaktivität, größere und schnellere Bewegungsabläufe des Fußes sowie deutliche Unterschiede der Schwingungen an der Achillessehne im Zeit- wie auch im Frequenzbereich aufwiesen im Vergleich zu gesunden Läufern.

DISKUSSION Im Rahmen der vorliegenden Arbeit konnten erste Einblicke erlangt werden in das Schwingungsverhalten an der Achillessehne sowie die simultanen Adaptationen oder Kausalitäten von plantarer Druckverteilung, Muskelaktivität und Fußkinematik bei Läufern. Im Rahmen dieser Arbeit ist es gelungen externe Faktoren auszumachen, welche die untersuchten biomechanischen Parameter beeinflussen. Diese können ebenfalls zur Modifikation der besprochenen Daten dienen. Die untersuchten Parameter zeigten sich darüber hinaus als geeignet zur Differenzierung zwischen Probanden mit Beschwerden im Bereich der Achillessehne und gesunden Läufern. Ein wichtiger Teil der komplexen Problematik bezüglich Verletzungen der Achillessehne kann mit der vorgeschlagenen Parameterkonstellation erklärt werden, welche der Betrachtung einzelner Parameter überlegen zu sein scheint. Weiterführende Arbeiten sollten auf eine frühzeitige Erkennung von Läufern mit erhöhtem Risiko für Überlastungen an der Achillessehne fokussiert sein, um dadurch gegebenenfalls Verletzungen dieser Struktur zu vermeiden. Abänderungen der untersuchten externen Faktoren könnten bei Läufern mit Achillessehnenbeschwerden Anwendung finden. Hierüber sollte eine Annäherung an die Parameterkonstellation erzielt werden, die bei gesunden Läufern detektiert wurde. Das Erreichen eines rehabilitativen Effekts ist ebenfalls denkbar, bedarf jedoch der Bestätigung durch weiterführende, prospektive wissenschaftliche Studien. Eine Aussage bezüglich der Ursache-Wirkungs-Kette kann zum aktuellen Zeitpunkt aufgrund des retrospektiven Designs der Querschnittstudie nicht getroffen werden.

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LIST OF ABBREVIATIONS

d_s .... effect size
$Mean_c$ .... mean value in the complaint group
$Mean_h$ .... mean value in healthy subjects
n_c .... number of subjects in the complaint group
n_h .... number of healthy subjects
SD_h .... standard deviation in healthy group
SD_p .... pooled standard deviation
VO_2max ...... maximal oxygen uptake
AKP .... Anterior knee pain
ANOVA .... Analysis of variance
ASTM .... American Society of the International Association for Testing and Materials
AvgAcc .... Average acceleration
CI .... Confidence interval
Con1 .... Shoe configuration 1 with high arch support, medial wedges and soft damping material
Con2 .... Shoe configuration 2 with low arch support, no medial wedges and hard damping material
DFT .... Discrete Fourier Transformation
EMG .... Electromyography
ffc .... First foot contact
FFCP .... Forefoot contact phase
FFP .... Foot flat phase
FFPOP .... Forefoot push off phase
FFT .... Fast Fourier Transformation
Hz .... Hertz
ICP .... Initial contact phase
IMU .... Inertial measurement unit

ISEK .... International Society of Electrophysiology and Kinesiology
KMO-index ... Kaiser-Meyer-Olkin index
M. Gastr. ...... Musculus Gastrocnemius
M. Gastr. lat. .. Musculus Gastrocnemius lateralis
M. Gastr. med. Musculus Gastrocnemius medialis
M. Peroneus long. Musculus peroneus longus
M. tib. ant. .... Musculus tibialis anterior
max force ..... maximum force
max force slope maximum force slope
MaxPro .... Maximum pronation
MaxProVel .... Maximum pronation velocity
MEMS .... micro-electro-mechanical sensors
NS .... Neutral all-purpose shoe
PA .... pre-activation
PCA .... Principal component analysis
PeakAcc .... Peak acceleration
ROM .... Range of motion
SD .... standard deviation
TCT .... Total contact time
TFPro .... Time to final pronation
ttpeak .... Time to peak acceleration

Part I

INTRODUCTION

1 PRELIMINARY REMARKS TO THE THESIS AT HAND

The results presented here were worked out in the context of the research project *Sensor Controlled Running - Untersuchung zur Integrationsmöglichkeit moderner Sensorik am Schuh am Beispiel der Vermeidung von Überlastungsfolgen bei Sport- und Arbeitsschuhen durch schuhtechnische Maßnahmen*. The project was funded by the Federal Ministry of Economics and Technology with grant number 17515N. The organization and implementation of the project took place in cooperation with the Prüf- und Forschungsinstitut Pirmasens. The project's objectives included the development of a sensor supported measurement shoe to quantify impact forces, muscle activity, foot kinematics and oscillations at the Achilles tendon. These requirements limited the utilized measurement tools to sensors that could also be integrated into the shoe and to measurement locations that were close to or at the foot. Therefore, the employed setup does not necessarily represent the gold standard but resembles a reasonable compromise between measurement accuracy and the ease of integration into a shoe. A theoretical introduction is given at the beginning of this work. Developmental work is performed in the sensor selection and algorithm design. Finally, empirical tasks were solved during the conduction of the measurement series.

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Part II

CURRENT STATE OF RESEARCH

2

RUNNING RELATED INJURIES

2.1 THEORETICAL BACKGROUND

In research concerning overuse risk factors in sports, the injury sequence model of van Mechelen (1992) is often applied. It represents a theoretical model, aiming at reducing the incidence of sports related injuries. The first step of this model establishes the extend of the injury by describing incidence and severity of the problem. The second step implies the investigation of its etiology and the understanding of mechanisms leading to the concern before effective preventive measures can be introduced in step 3. The fourth step then allows a feedback-mechanism to assess the prevention's effectiveness by repeating step 1 of the model. In the thesis at hand, a biomechanical evaluation of overuse risk factors during running is performed. Particular attention is given to individual risk factors for Achilles tendon complaints.

2.2 EPIDEMIOLOGY

Running is one of the most common physical activities spent during leisure time. Over the past years long distance runs have gained popularity not only for professional athletes but also for novices. The number of participants in marathon runs has increased from 25,000 in 1976 to 541,000 in 2013 in the US alone (Lamppa, 2014). Besides its positive effects on the incidence of hypertension, obesity, depression and other major health issues (Kruisdijk, Hendriksen, Tak, Beekman & Hopman-Rock, 2012; Williams & Thompson, 2014) running may also cause acute or chronic injuries (Schueller-Weidekamm, 2010; Hreljac, 2005). In an extensive review of the epi-

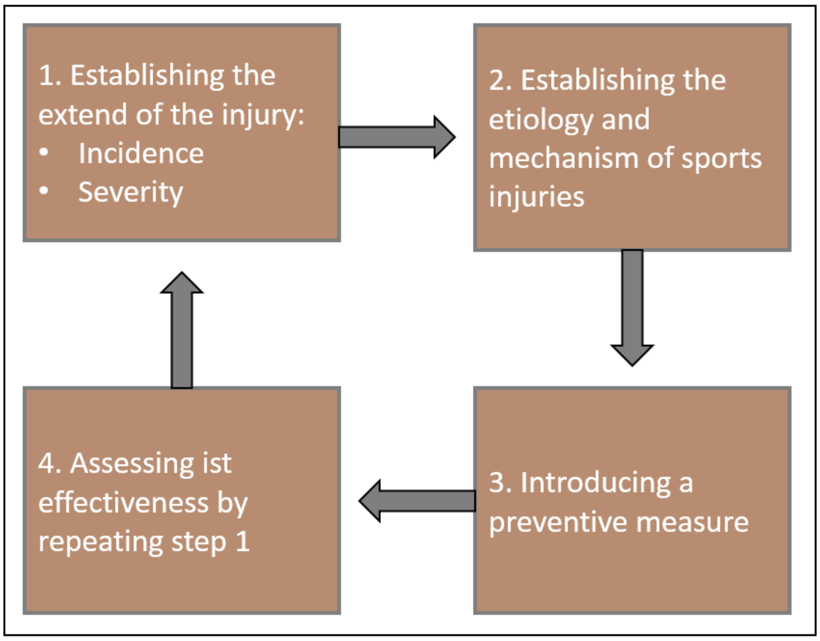

Figure 1: The injury sequence model representing the chronology of the prevention of sports injuries. Figure modified according to van Mechelen (1992).

demiological literature, van Mechelen (1992) observed an incidence rate of 37% to 56% while van Gent, Siem, van Middelkoop, van Os, Bierma-Zeinstra & Koes (2007) concluded an incidence of 19.4% to 79.3% for lower extremity injuries alone. Mayer, Grau, Maiwald, Ploog, Bäurle, Beck, Baur & Müller (1999) found a shift of injury location from the knee down towards the Achilles tendon.

For many years, the most common injury site in runners was the knee joint (Mayer et al., 1999) while in recent years, the Achilles tendon has become affected more frequently. Table 1 sums up the frequency distribution of running injuries at the lower extremity over the past 40 years. Common chronic injuries at the knee joint include patellofemoral pain syndrom, insertion tendinosis of the patella tendon as well as tendinitis of the patellar tendon itself and inflammations of the Tractus Iliotibialis' insertion, which are often caused by friction of the Tractus Iliotibialis at the lateral femoral condyle (Mayer, Grau, Beck, Krauss, Maiwald & Baur, 2000). The increase in Achilles tendon injuries in recent years could not solely be caused by an actual rise in its incidence rate but may also be determined by improvements in diagnostic procedures. Advancements in ultrasound and magnetic resonance imaging procedures could allow more precise medical diagnoses, which may not have been referred to the Achilles tendon in the past.

Insufficient shock absorption while running on solid floor, misalignment of joint axes as well as training habits are indicated as injury inducing factors in several studies (Reule, Alt, Lohrer & Hochwald, 2011; Clement & Tauton, 1981; Cowan, Jones, Frykman, Polly, Harman, Rosenstein & Rosenstein, 1996; Macera, 1992; Ross, 1993; van Mechelen, 1992). The authors consistently concluded sufficient damping through adequate running shoe modifications as well as frequent running training on softer surfaces were needed to reduce knee joint complaints. The shoe industry reacted to this demand by implementing several damping concepts, therefore often increasing the distance between the runner's heel and the ground. Several changes in running training were implemented, including runs on uneven grounds such as forest floors or gravel as well as functional training of core muscles (Mayer et al., 2000; Ferber, Hreljac & Kendall, 2009). From the 1990s onwards, an increase in Achilles tendon complaints in runners can be noted with a decreasing tendency in knee joint injuries (see table 1). Interestingly, most knee joint injuries were detected during routine examinations with runners not seeing a doctor specifically for that complaint. Runners with Achilles tendon injuries however, were seriously impaired in their training and were seeing a doctor explicitly because of their complaints at the tendon (Mayer et al., 2000). Running related overuse injuries of the foot and ankle have an incidence rate of 250 in 1000 athletes per season and Achilles tendinopathy, plantar fasciitis and stress fractures are the most commonly studied injuries (Sobhani, Dekker, Postema & Dijkstra, 2013). Kaufman, Brodine, Shaffer, Johnson & Cullison (1999) found Achilles tendinitis to be one of the most common overuse injuries in a 2-year prospective investigation of Navy SEAL candidates with an injury rate of

30 ± 6.7 % and Lopes, Hespanhol Júnior, Yeung & Costa (2012) concluded it to be the most frequent injury location in ultramarathon runners.

2.3 CAUSE ANALYSIS OF RUNNING RELATED INJURIES

Many biomechanical parameters are discussed as injury inducing factors in habitual runners. Excessive stresses and loads at biological structures may result from a repetitive incidence of high impact forces. Increased kinetics may not only limit the comfort of running but plantar flexion peak torque was also found to be a significant discriminator between noninjured runners and injured runners with Achilles tendinitis (McCrory, Martin, Lowery, Cannon, Curl, Read, Hunter, Craven & Messier, 1999). Measurements in this study were performed with a force platform while subjects ran over ground. Injured runners also showed a tendency to higher peak ground reaction forces than their healthy counterparts. In a literature review on running biomechanics of individuals with Achilles tendinopathy, Munteanu & Barton (2011) found differences in dynamic plantar pressure and ground reaction forces. However, other research groups did not find a difference in impact forces when comparing runners with Achilles tendinopathy to healthy runners (Azevedo, Lambert, Vaughan, O'Connor & Schwellnus, 2009). A new paradigm, postulated by Nigg (2001) suggests the requirement of decreased damping of impact forces to allow adequate reactions through muscle contraction. If, however, impact forces are highly damped by large cushioning complexes in running shoes, inadequate reactions of the body may occur, leading to harmful loadings of certain structures. The shoe industry recently reacted to the paradigm as well as other results of the research by introducing minimalist footwear, also known as barefoot shoes. These shoes are characterized by very little shock absorption, which promotes a midfoot or a forefoot running stride while shoes with a cushioned heel seem to facilitate a rearfoot stride (Lieberman, Venkade-

Table 1: Frequency distribution of running injuries over the past 40 years.

Study	Frequency distributions of injury locations
James, Bates & Osternig (1978)	20% knee, 11% Achilles tendon, 13% shint
Krissoff & Ferris (1979)	25% knee, 18% Achilles tendon, 15% shin
Cavanagh & Lafortune (1980)	23% knee, 20% Achilles tendon, 10% shin
Gudas (1980)	31% knee, 7% Achilles tendon, 10% shin
Clement & Tauton (1981)	42% knee, 6% Achilles tendon
Newell & Bramwell (1984)	50% knee
Marti, Vader, Minder & Abelin (1988)	28% knee, 12% Achilles tendon
Fallon (1996)	31% knee, 19% Achilles tendon
Fischer (1998)	25% knee, 40% Achilles tendon
Mayer et al. (1999)	25% knee, 32% Achilles tendon
Taunton, Ryan, Clement, McKenzie, Lloyd-Smith & Zumbo (2002)	42% knee, 6% Achilles tendon
Lopes et al. (2012)	16% knee, 19% Achilles tendon, 11% shin

san, Werbel, Daoud, D'Andrea, Davis, Mang'Eni & Pitsiladis, 2010). Therefore, if runners wear minimalist shoes, the first ground contact is usually made either with the mid- or the forefoot. Forefoot- and mid-to rearfoot runners differ not only in the first contact point of the foot with the ground but also in foot kinematics and plantar kinetics as shown in figure 2. Especially for runners with Achilles tendon injuries, these extreme changes in footwear should be seen as critical. Almonroeder, Willson & Kernozek (2013) showed that peak Achilles tendon forces occur earlier and impulse as well as loading rate in the tendon increase in barefoot running. The researchers transferred their findings to a resulting additional 47.7 bodyweight for each mile run with mid- or forefoot strike pattern. Other researchers confirm these results by finding increased tensile stresses at the Achilles tendon due to augmented muscular activation (Goss & Gross, 2012).

As mentioned above, activation of the calf muscles may play an important role in the development of Achilles tendon injuries. The muscle complex of Mm. Gastrocnemii and M. Soleus allows the Achilles tendon to assimilate an enormous amount of energy before any injury due to strain occurs (Gallo, Plakke & Silvis, 2012). While uneven force production of the two heads of the M. Gastr. and the resulting unbalanced strain at the tendon may overload particular areas of the Achilles tendon (Smart, Taunton & Clement, 1980; James et al., 1978), runners suffering from Achilles tendinopathy also show lower electromyographic activity of the M. Gastr. during stance phase compared to healthy runners (Azevedo et al., 2009). The authors therefore concluded that footwear, specifically affecting muscle activity of the Mm. Gastr. may be beneficial in the rehabilitation of Achilles tendinopathy in runners. While fatigue occurs during prolonged runs, very limited to no binding statements can be made with regard to its explicit effect as risk factor for overuse injuries in running. Past research shows an association between fatigue and a decreased tolerance of impacts in biological structures (Clansey, Hanlon, Wallace & Lake, 2012). Running

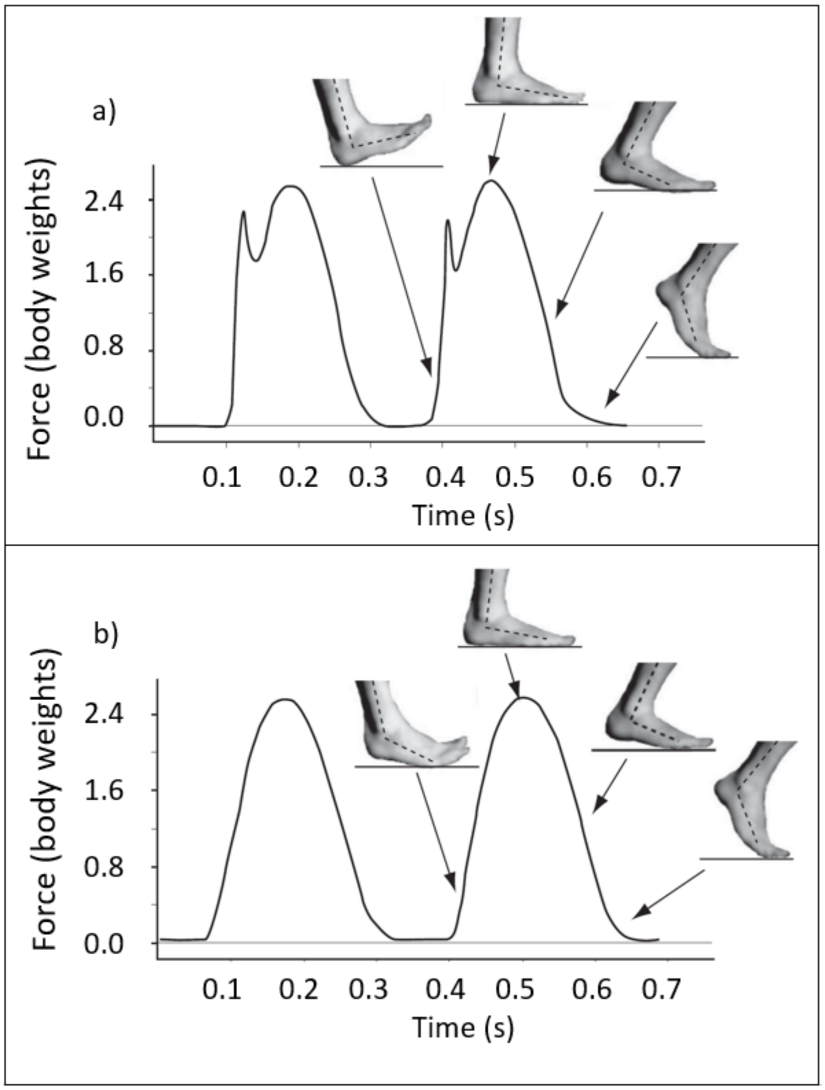

Figure 2: Foot kinematics and vertical ground reaction forces during rearfoot (a) and forefoot running (b). Figure modified according to Lieberman et al. (2010).

biomechanics were found to deteriorate near exhaustion, possibly facilitating harmful motions or loadings of the system (Fourchet, Girard, Kelly, Horobeanu & Millet, 2015). However, no distinct mechanisms of fatigue as a risk factor for running injuries were yet defined.

Especially excessive motion of the rearfoot in the frontal plane is frequently suggested to be potentially harmful (Messier & Pittala, 1988; Ferber et al., 2009). Stresses and Strains at the Achilles tendon may reach up to 7000 N and can be increased through excessive pronation at the stance phase during running (Clement, Taunton & Smart, 1984; Hennig, 1993; Hintermann, 1998; Lohrer, 1991). Lersch, Grötsch, Segesser, Koebke, Brüggemann & Potthast (2012) found Achilles tendon strain to be highly influenced by foot kinematics and recommend the consideration of rearfoot inversion in clinical movement analysis. Also, asymmetric pulls at the tendon due to an imbalance of medial and lateral head of the M. Gastrocnemius (M. Gastr.) or due to a dysbalanced position of the calcaneus may increase loading of the tendon enormously. Another proposed trauma-causing mechanism is the whipping or bowstring motion of the Achilles tendon following pronation (James et al., 1978). This motion is intensified if runners display excessive rearfoot motion (McCrory et al., 1999). Clement et al. (1984) describes the whipping motion following augmented rear foot motion after initial foot contact with the ground, which is accompanied by internal tibial rotation. Micro traumata at the paratenon as well as the tendon itself are provoked by this disadvantageous movement and may lead to inflammations. McCrory et al. (1999) were able to discriminate healthy runners and runners with Achilles tendon complaints based on rearfoot kinematics. The injured group was characterized by larger pronation values, a shorter time to maximum pronation, a greater maximum pronation velocity (MaxProVel) and more inversion at initial foot contact. Therefore motions of the rearfoot in the frontal plane seem to play an important role in the genesis of

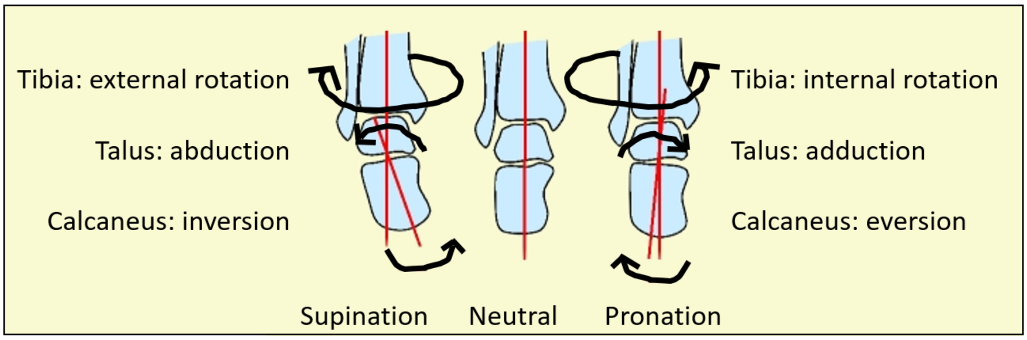

Figure 3: Schematic representation of the coupling mechanism of foot eversion and knee rotation. Figure modified according to Brookbush (2015).

Achilles tendon complaints in runners, although different mechanisms are discussed to lead to a harmful effect.

Foot eversion describes the rotary motion of the calcaneus around the subtalar axis in lateral direction. It is coupled with internal tibial rotation as shown in figure 3 (DeLeo, Dierks, Ferber & Davis, 2004). Therefore, increased eversion may not only influence the foot, ankle and Achilles tendon but also lead to augmented tibial rotation. Overloading of other structures at the lower extremity such as the knee joint may be a consequence of excessive motion of the foot in the frontal plane (Hintermann, 1998). Rodrigues, Chang, TenBroek & Hamill (2013) found runners with anterior knee pain (AKP) to use a greater amount of their passive pronation range of motion (ROM) compared to healthy subjects. Also, pronation change during the first 10% of stance is known to be a discriminating factor between healthy subjects and those with AKP (Duffey, Martin, Cannon, Craven & Messier, 2000). In this study however, injured subjects showed smaller changes than their healthy counterparts. No consensus exists about the influence of rear foot motion on running injuries. While many injured athletes show excessive calcaneal movements during locomotion, numerous healthy counterparts demonstrate similar or even larger pronation angles (Mayer et al., 2000).

An important factor, which is often discussed to be relevant for the development of running related injuries is the type of footwear a runner uses. Soft, unstable shoes are expected to allow excessive pronation of the foot, therefore leading to overuse injuries (Bahlsen, Denoth, Luethi, Nigg & Stacoff, 1986). Others state shoes with soft heel caps or stiff outer soles to increase stresses at the Achilles tendon (Subotnick, 1989). Specific shoes were designed to prevent overuse injuries with particular attention to Achilles tendon injuries (Grau & Horstmann, 2007). Large spring and pitch of a shoe were associated with complaints at this structure (Frey & Shereff, 1988). Also, very stiff heel caps may increase friction at the Achilles tendon to a harmful level. External mechanical stimuli lead to swelling of the peritendineum with a secondary change of the tendon itself (Mayer et al., 2000). The running shoe is the only factor described here, which may be directly adjusted and modulated. More recent developments include the integration of miniaturized measurement systems in shoes and sports wear, so called smart clothes. Smart clothes are also commercially available and include sensor integration into running shoes. Available products include Adidas miCoach® or Nike+®, which inform the runner about step and position data but do not include complex biomechanical parameters. Also, no systems are yet available which offer valuable insight into gait parameters or allow running style optimization for their user. Running shoe modifications were found to influence many of the above mentioned injury inducing factors like foot kinematics, kinetics and muscle activity (Mündermann, Nigg, Humble & Stefanyshyn, 2003). It therefore suggests itself to change this external component within the complex biomechanical system of a runner. However, careful evaluation of these adjustments is required to assure beneficial effects without harmful side effects.

Although many biomechanical data were evaluated in the past to reveal underlying overuse mechanisms, the complex cause-and-effect chain has not been explained. The biological system includes an enormous variety of structures and functions. However, most researchers have focused on discrete small parts instead of the whole system. Including data of all these structures and the Achilles tendon during continuous motion may be a clear advantage as it allows a more extensive picture of the runner. The contributing risk factors running speed, footwear, ground condition and fatigue were yet investigated on a separate basis but no synchronized evaluation of these parameters was performed including their modifying effect on biomechanical data of plantar pressure, foot kinematics, muscle activity and the oscillation behavior at the Achilles tendon. The underlying mechanisms leading to injuries during running is yet not understood to a sufficient and fully explanatory degree. If only single factors are investigated, the interaction between these factors remains unknown. However, the combined behavior of these factors on numerous biomechanical parameters might play a key role in the predisposition for injuries. Pattee (1973) stated "The problem is precisely at the interface between the detail structure and the abstraction of function", showing the difficulty to chose an appropriate number of parameters to explain a system's behavior and therefore to reveal the mechanism of overuse injuries. The number of parameters evaluated in past studies seems to be too limited to understand the underlying mechanism of Achilles tendon complaints. Therefore, a wider perspective should be taken, considering not only data of several biomechanical structures at the same time but also taking their interactions into account. The deduction of overuse injury mechanisms will only be possible if we understand how the biomechanics of running are composed.

2.4 AIMS AND OBJECTIVES

The essential goals of this thesis are:

1. Implementation of a simultaneous use of several measurement systems

2. Detection of disadvantageous parameter constellations, which lead to harm or injury in runners

The background information given above shows the uncertainty about injury inducing factors in runners and leads to the conclusion that predisposing factors for chronic overuse injuries in individuals as well as the running population altogether still remain unknown. A high likelihood of multifactorial causes can be assumed. Several personal factors (for example age, gender) and physical factors (for example foot posture, blood circulation within the tendon) have been shown to be related to complaints at the Achilles tendon. However, single cause-and-effect chains do not seem to be sufficient to differentiate between healthy and injured runners. Few researchers have looked at several parameters in the investigation of injury inducing factors. However, none of these parameter combinations seems to fully explain the issue as many uncertainties remain. Therefore, within this thesis at hand, measurement methods were derived from the literature and applied in a simultaneous matter. A number of measurement instruments were identified as suitable to allow acquisition and analysis of complex biomechanical data in runners. We chose plantar pressure, EMG, foot kinematics and oscillations at the Achilles tendon as these data either showed promising results in previous studies concerning overuse risk factors (plantar pressure, EMG, foot kinematics) or they were not included as possibly predisposing factors in past research (oscillation behavior). In general, no simultaneous recordings of these data were yet conducted during running. Small, lightweight sensors were preferred to be included in the measurements of this work and attention was turned to the possibility of integrating the selected sensors into a running shoe. An important goal was the implementation and the evaluation of the simultaneous use of these measurement systems. The synchronized collection of the specified data has not been performed before and detailed analysis of plantar pressure, electromyography (EMG) and kinematic data as well as of oscillations at the Achilles tendon are an essential objective of this work at hand. Particularly the detection of oscillations at the Achilles tendon and their modification due to changes of external factors can be considered as fundamental research and gives new insight into the tendon's behavior during running. Following the study of modifying factors on the investigated biomechanical parameters, a first outlook is given on the comparison of runners with Achilles tendon complaints and healthy runners. Through this comparison, a detection of disadvantageous parameter constellations should be achieved and a variety of factors possibly characterizing runners with Achilles tendon pain should be identified.

Part III

METHODOLOGICAL BACKGROUND

3 PLANTAR PRESSURE DISTRIBUTION

3.1 THEORY AND METHODS OF PRESSURE QUANTIFICATION

External forces acting at the human body during running occur as air resistance and as ground reaction forces when the runner's foot is in contact with the floor. Kinetics underneath the foot can be recorded and assessed using pedobarometry, a useful tool in the evaluation of load distributions in clinical and sports biomechanics. Ground reaction forces are quantified and pressure values can be determined if the contact surface is known. Ground reaction forces are defined as the sum of a person's body weight and all acting forces, caused by accelerations or decelerations of the body (Belli, Kyröläinen & Komi, 2002). Typical time series of the forces detected underneath a rear foot runner's foot are shown in figure 4.

To quantify reaction forces on the foot during floor contact, force plates, pressure mats or in-shoe pressure measurement systems can be used. Force plates are utilized to assess ground reaction forces during over ground locomotion. The systems record up to three measurement directions, whereas the vertical force component accounts for about 80% of the total ground reaction force (Hegewald, 1999). Force plates commonly record at very high sampling rates, which allow very precise measurements under laboratory conditions. A disadvantage of these platforms is that they are stationary, can hardly be used in the field and subjects have to time their steps to hit the plate. The results may be distorted through this non-natural step sequence. Pressure mats use a large number of pressure sensors to detect both, quantity and position of the applied forces. Pressure mats have recently been integrated into treadmills to allow continuous measurements during treadmill running

(Kalron & Achiron, 2014; Macellari, Groeneveld, Torre & Giacomozzi, 1994). In-shoe measurement systems consist of an inserted sole, into which pressure sensors are integrated. A big advantage of these systems is the continuous recording of a large number of consecutive steps during over-ground locomotion without any limitations to laboratory conditions. Therefore, these in-shoe pressure soles can be worn in almost any kind of shoe, at any kind of surface, outdoors as well as indoors. A critical point of the insoles is their durability, though. Especially during high impact loadings, stresses can exceed the insoles' capabilities and they may brake (Hegewald, 1999). While measurements with force plates can be performed in three dimensions, no differentiation of the force vector into its components can be made with either pressure mats or in-shoe pressure measurement systems.

3.2 F-SCAN MEASUREMENT SYSTEM

In the series of measurements described in this thesis (chapter 8 and 9), an insole pressure system was used (F-Scan®, Tekscan Inc., South Boston, MA, USA). The extremely thin soles are constructed as a matrix of 954 pressure sensing elements and can be trimmed down to any shoe size from 14 (US mens) downwards. The devices take up very little room in the shoe as their total thickness is 0.2 mm. Measurements up to 862 kPa can be performed with the F-Scan measurement system at a sampling rate of up to 100 Hertz (Hz) (wireless) or up to 750 Hz (datalogger). A graphical illustration of the insole's construction is shown in figure 5 with its data specifications in table 2.

Ahroni, Boyko & Forsberg (1998) tested the reliability of pressure measurements using F-Scan insoles. Several sections of the footprints were analyzed as well as the total area underneath the foot. Non of the investigated tests showed poor reliability and the researchers concluded the measurement system to be generally reliable. Hsiao, Guan & Weatherly (2002) evaluated the accuracy and

Table 2: General dimensions of F-scan insoles. Nomenclature corresponds to figure 5.

General Dimensions		Sensing Region Dimensions	
Overall Length	Overall Width	Column Width	Column Spacing
327.2 mm	313.7 mm	2.5 mm	5.1 mm
Matrix Width	Matrix Height	Row Spacing	Row Width
106.7 mm	304.8 mm	5.1 mm	2.5 mm

precision of the F-Scan pressure measurement system and found the measurement error to be in an acceptable range between 1.3% and 5.8% if proper calibration of the system was performed prior to measurements. It should be noted however, that external devices may alter the movement characteristics of a runner. In a study on six participants, Kong & De Heer (2009) found shorter stride lengths and higher stride frequencies while wearing an F-Scan measurement system compared to running without the system. Even though significant differences were found in this study, the magnitudes of these differences were very small when compared to the repeatability of the investigated parameters. The authors therefore do not address any clinical implications to the findings. In conclusion, the F-Scan measurement system can be described as a reliable tool to detect plantar pressure during running without heavy limitations of the runner, which may lead to relevant changes in their movement patterns. The parameters derived from this system included total force and peak forces. Total force implies the entirety of all forces quantified underneath the foot during stance phase while peak forces describe the maximum force in any region of the foot print.

Other available in-shoe pressure measurement systems include Pedar® (Novel GmbH, Munich, Germany), Medilogic® (T&T medilogic Medizintechnik GmbH, Schönfeld, Germany) and ORTHO-Control® (Cosinos, Fürstenfeldbruck, Germany). While the Pedar®

system seems to be the most frequently used plantar pressure system in scientific research, we decided againt this product as it comes at a relatively high market price, does not allow individual cutting of the insoles and therefore requires separate insoles for each shoe size. The Pedar® insoles come at a price which is about 20 times higher than that of the F-Scan insoles. Therefore economical reasons as well as the disadvantage of fixed sizes excluded these insoles from our decision. The Medilogic® measurement system allows direct synchronization with the Noraxon Telemyo 2400 G2, which we used in our studies. However, similar to the Pedar® insoles, no individual cropping can be performed and insoles come at a relatively high price. Also, higher follow-up costs were expected as the software needs professional recalibration after 5 000 steps. ORTHOControl®was not the measurement system of choice as it is designed as a closed system. Therefore no data export and no synchronization with other measurement devices is possible. We therefore chose the F-Scan® plantar pressure measurement system for the studies performed for this thesis as it allows individual cutting of relatively low-cost, ultra-thin sensor insoles with a high sensor resolution as well as the scientific results confirming its reliability and accuracy.

3.3 INSTRUMENTED ZEBRIS FDM-T TREADMILL

In the series of measurements described in chapters 10 and 11, an instrumented treadmill (Zebris FDM-T, Zebris®, Isny, Germany) was used to assess plantar pressure during running. The measurement system is made up of a treadmill with an integrated, calibrated measurement matrix. The matrix has a size of 150 x 50 cm and consists of more that 5000 capacitive pressure sensing elements. Pressure data can be synchronized with other biomechanical recordings, such as EMG or acceleration data, using Noraxon MR3 software (Noraxon Corporate, Scottsdale, AZ, USA). The software automatically compensates the treadmill's movements to allow a stable analysis of pressure patterns. As described in section 3.1, pressure insoles are not sufficiently durable for high impact loading as well as highly repetitive loading, while an instrumented treadmill is considered more suitable for recordings during prolonged runs. In the literature, differences in pressure measurements were found when recordings were performed at an instrumented walkway compared to an instrumented Zebris treadmill (Wearing, Reed & Urry, 2013). It is however likely, that these differences were caused by variations in the walking pattern during treadmill walking compared to walking over ground. This assumption is supported by a study of Braun, Veith, Hell, Döbele, Roland, Rollmann, Holstein & Pohlemann (2015), who found high correlations and marginal mean differences when comparing data obtained with a Zebris treadmill to data from a pressure sensing insole. Other researchers studied the reliability and sensitivity of the instrumented treadmill and concluded the Zebris treadmill to be suitable to detect changes in spatio-temporal parameters as well as ground reaction forces (Reed, Urry & Wearing, 2013). It should be noticed that, while high comparability exists between center of pressure measurements with in-shoe measurement systems and external force plates (Dyer & Bamberg, 2011; Chesnin, Selby-Silverstein & Besser, 2000), differences between the two measurement methods were found for force magnitudes (Barnett, Cunningham & West, 2001). Pressure sensing insoles show a tendency to detect lower peak forces, shorter step durations and therefore lower force integrals than external devices. Barnett et al. (2001) quantified the differences between the two systems to be 1.8% for temporal parameters, 6.3% for impulse data and 13.4% for force data. Also, in-shoe measurements were found to be influenced by the type of shoe worn. The researchers conclude both measurement procedures to be valid and reliable but do not encourage comparisons between data obtained with an insole and data obtained with an external force plate or an instrumented treadmill.

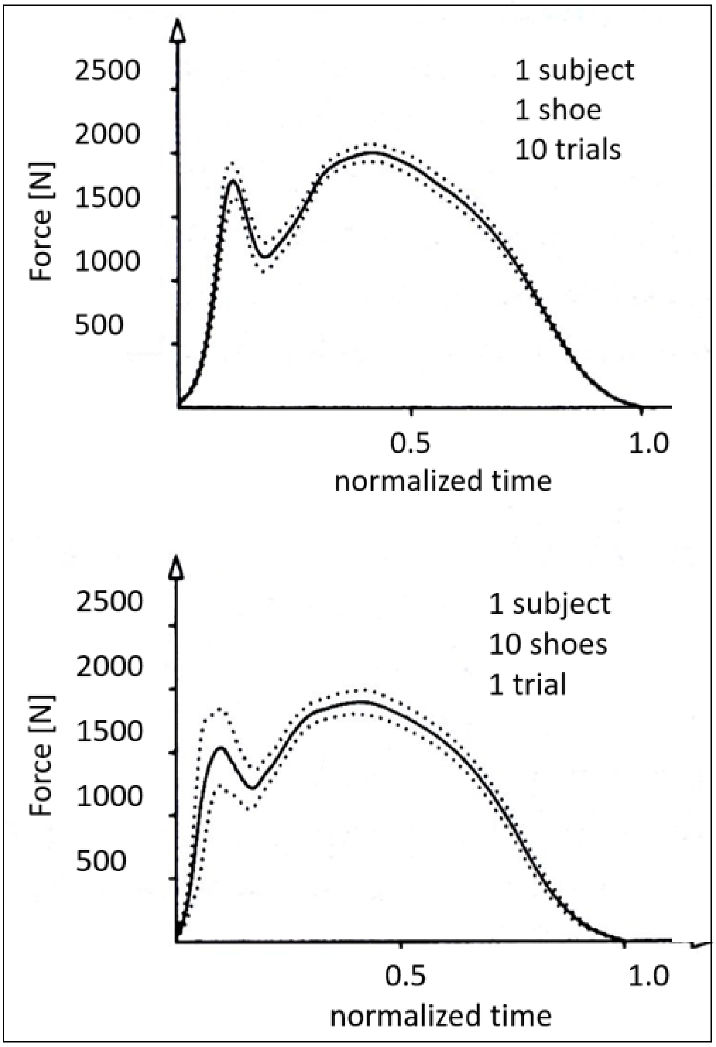

Figure 4: Characteristic ground reaction forces in vertical direction during rear foot running. The upper graph shows the reliability of this measurement technique with 10 recorded trials of one subject while the bottom graph shows the variability within one subject while wearing 10 different types of shoes. Figure modified according to Bahlsen et al. (1986)

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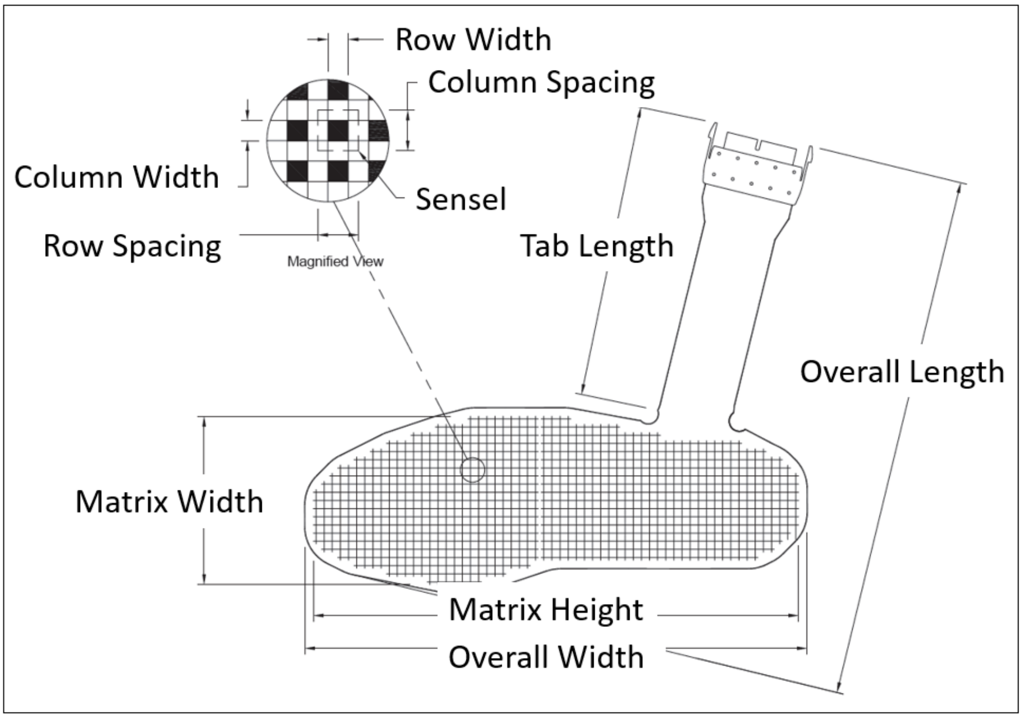

Figure 5: Detail drawing of an F-Scan pressure measurement insole. Figure modified according to Tekscan (2015).

4

MUSCLE ACTIVITY

4.1 THEORY AND METHODS OF ELECTROMYOGRAPHY

EMG is a measurement method of objectively quantifying muscle activity through voltage fluctuations, which occur in the stages of rest and activation of skeletal muscles. However, no direct conclusion can be drawn from the recorded EMG signal to the force generated by this muscle (De Luca, 1997). The EMG signal is the sum of several action potentials from various motor units laying underneath the recording electrode and always includes system noise caused by measuring instruments, cables and the like (see figure 6) (Bartlett, 2007). Other factors influencing the obtained signal are physiological crosstalk between neighboring muscles, electrode locations and tissue characteristics such as the type and thickness of tissue overlaying the muscle, temperature and humidity as well as the proximity of electrode adhesion (De Luca, 1997). In order to reduce these interference factors, EMG measurements of the studies described in chapter 8 - 11 were performed following the European guidelines of Surface Electromyographie for the Non-Invasive Assessment of Muscles (Hermens, Freriks, Merletti, Stegeman, Blok, Rau, Disselhorst-Klug & Hägg, 1999).

4.2 PRACTICAL APPLICATION OF ELECTROMYOGRAPHY

In sports biomechanics, EMG is frequently used to collect information on the timing of muscle activation following a certain event or the sequencing of various muscles within one muscle group (Bartlett, 2007). The overlap of agonist and antagonist activity has been related to skill during specific motor tasks or sports move-

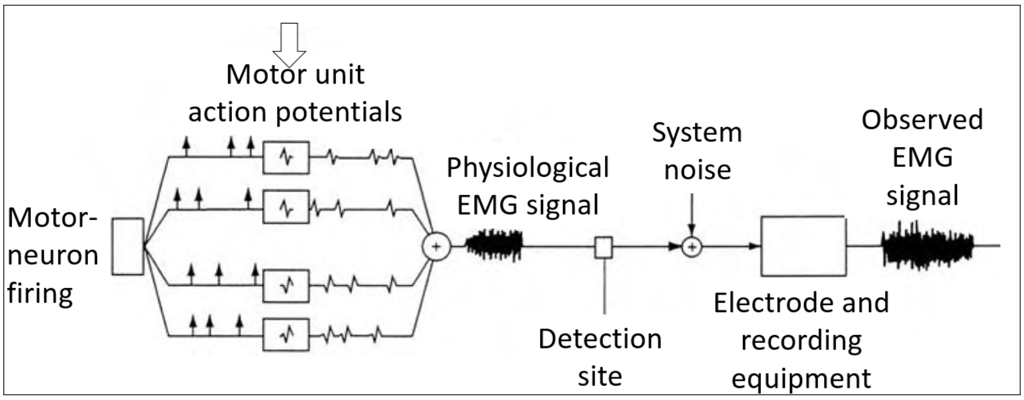

Figure 6: Schematic flowchart representing the generation and recording of EMG signals.Figure modified according to Bartlett (2007).

ments. EMG also allows the study of changes in muscle activation magnitude as a result of training or to differentiate between injured and healthy athletes (Cholewicki, Greene, Polzhofer, Galloway & Radebold, 2002). EMG results are also required when calculating internal forces in the musculoskeletal system as muscles may provide substantial strength, acting at bones and joints (Hanley & Bissas, 2013). Besides the noninvasive surface EMG an invasive measurement version exists: needle or fine wire EMG, which is inserted into the muscle and allows the recording of deep muscles and a more precise discharge location. However, fine wire EMG is not suitable for measurements during sporting activities as it may increase the likelihood of injuries at the insertion site or inside the muscle. Therefore, surface EMG is commonly used during sporting activities, which limits the analysis to superficial muscles (De Luca, 1997). However, surface EMG shows advantages over fine wire EMG when researching average muscle activity. Data collected with surface electrodes showed a better reproducibility than fine wire EMG data (Kadaba, Wootten, Gainey & Cochran, 1985).

In a repetitive motion like running, it is beneficial to record several cycles of the motion of interest to increase reliability of the obtained results (Winter & Yack, 1987). Previous studies showed, that a minimum of 6 steps is required to obtain a representative data set (Shiavi, Frigo & Pedotti, 1998). In the test series described in the thesis at hand, all recordings included $\geqslant 6$ consecutive steps. EMG-signals were rectified (Winter & Yack, 1987) before the envelope of the amplitude signal was divided into time segments, each including one stance phase. These segments were then normalized to time to allow averaging over stance phases (Shiavi et al., 1998). A large variety of analysis procedures exist for EMG data, ranging from simple to exceedingly complex. In the measurement series described in this work, explicit analytical methods where chosen, which lead to clear interpretations of the results. For data acquisition, Noraxon Telemyo 2400 G2 (Noraxon Corporate, Scottsdale, AZ, USA) was used while the corresponding Noraxon MR3 software (Noraxon Corporate, Scottsdale, AZ, USA) was utilized for EMG data analysis. Fundamental validity and reliability of surface EMG is well accepted (Morrish, 1999; Pullman, Goodin, Marquinez, Tabbal & Rubin, 2000; Türker, 1993) and the afore mentioned hard- and software were proven to be valid measurement systems (Walters, Kaschinske, Strath, Swartz & Keenan, 2013).

5

FOOT KINEMATICS

5.1 THEORY AND METHODS OF INERTIAL MEASUREMENT UNITS

Inertial sensors are a group of measurement devices including inertial system assemblies, inertial measurement units, vertical reference units and others as shown in figure 7. Inertial measurement units (IMU) are a specific type of inertial sensors, which include an accelerometer as well as a gyroscope. The sensor's output is given in m/s^2 for accelerations and in rad/s for angular velocities. The operating mode of gyroscopes is based on the conservation of angular momentum, meaning that the total angular momentum remains constant within a system unless external forces act on the system. There are several types of gyroscopes like rotor gyroscopes, monolithic silicon gyroscopes or optical gyroscopes. A mechanical gyroscope is typically made up of a disk, which rotates around a spin axis and a frame to which it is affixed. The frame itself rotates around one or more axes, offering the spinning disk additional degrees-of-rotation and therefore defining the degrees-of-freedom of the sensor. Due to the conservation of angular momentum, the spinning disk remains robust to dislocations in space and maintains a constant orientation. The rate of rotation of the sensor's spin axis around its output axis is equivalent to the applied torque or rotation of the system (Fraden, 2004). The application of external forces will cause tiliting of the spin axis, resulting in a momentum of torque. The spin axis will tilt orthogonal to the applied force to conserve the total angular momentum. This phenomenon is called precession (Fraden, 2004). Precession is directly related to the external forces applied to the system and change of position becomes

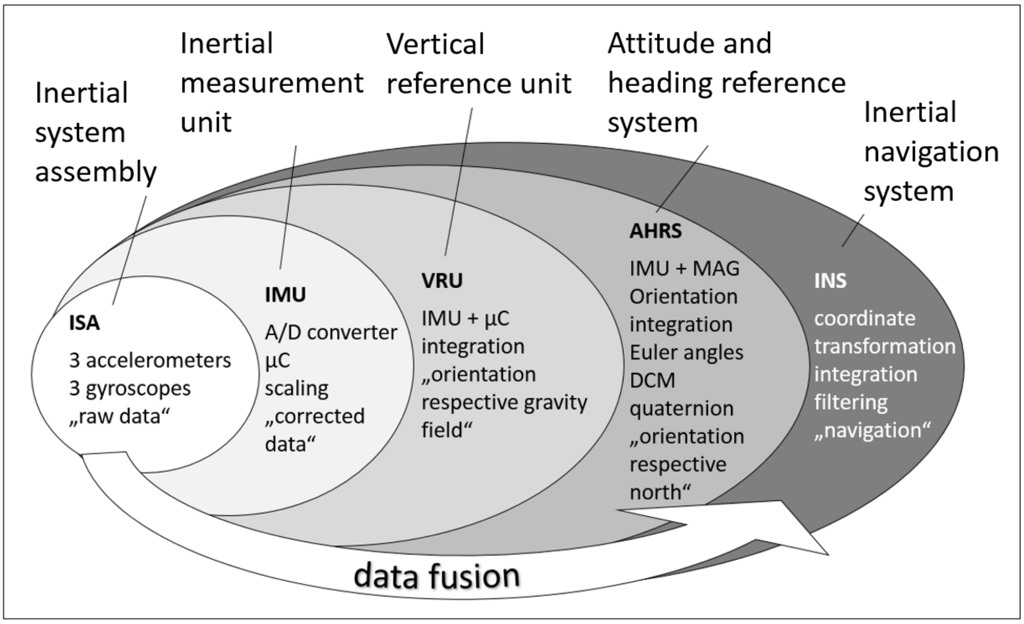

Figure 7: Definitions of inertial sensors according to Dorobantu (1999). Figure modified according to Schäfer & Wild-Pfeiffer (2015)

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measurable. A graphical illustration of a rotor gyroscope is shown in figure 8 including depictions of the axes mentioned above.

The small devices are easy to synchronize with other sensors, measurement axes can be aligned to the given object or segment, they are light weight and cost-efficient (Schäfer & Wild-Pfeiffer, 2015). IMUs include both, accelerometers and gyroscopes and can therefore benefit from data fusion of these two sensors. A technique to integrate both sensor data is the application of so called complementary filters, which will be described in section 5.2.

5.2 COMPLEMENTARY FILTER

A complementary filter is a mathematical method to fuse data of an accelerometer and a gyroscope as included in an IMU. It combines the raw accelerometer data and the integrated gyroscope data by applying a first order low pass filter to the former and a first order high pass filter to the latter. Finally, both outputs are added

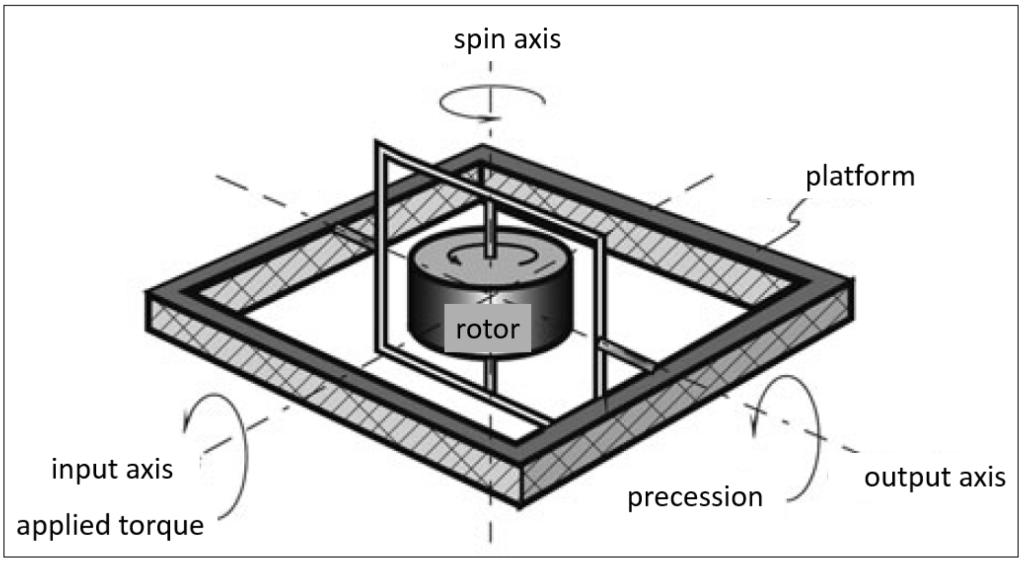

Figure 8: Mechanical gyroscope with a single degree of freedom. Figure modified according to Fraden (2004)

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to estimate movement angles (Garcia, Escareno & Rosas, 2010). A schematic illustration of a complementary filter is shown in figure 9. Equation 9 explains the calculation of movement angles, where f_1 = 0.98 and f_2 = 1- f_1. Both values represent the complementary factors and were identified experimentally. The starting angle α_1 was determined using equation 2. In this equation α represents the movement angle while ω stands for the gyroscope data and therefore angular velocity and a represents accelerations and therefore accelerometer data.

$$\alpha_{i+1} = \sum_{i=1}^{n-1} (f_1 * (\alpha_i + \omega_i * dt) + f_2 * a_i \qquad i \in (1, n) \tag{1}$$

$$\alpha_1 = f_1 * (\omega_1 * dt) + f_2 * 1 \tag{2}$$

In a complementary filter approach, accelerations are used as a damper of the calculated angle. High accelerations have an amplifying effect, low accelerations attenuate the angle. The complemen-

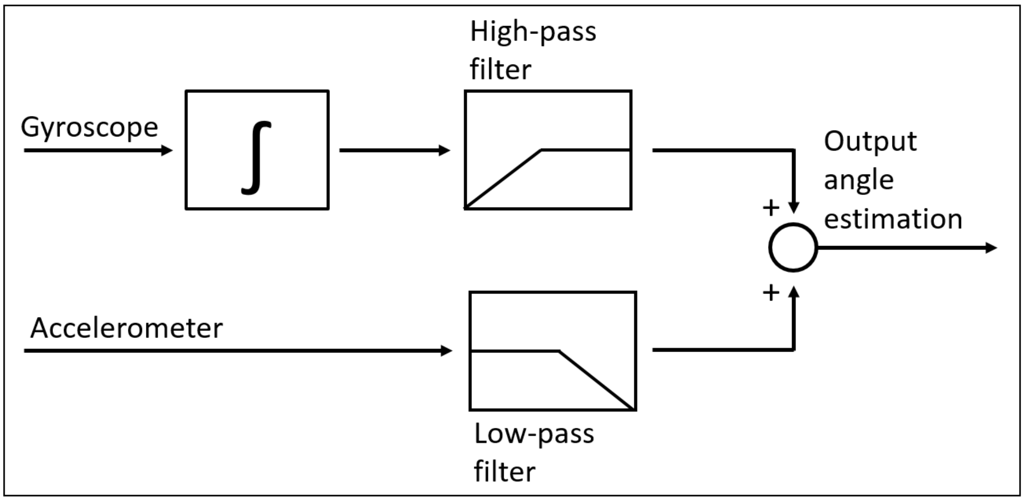

Figure 9: A complementary filter adds low pass filtered accelerometer data to high pass filtered integrated gyroscope data to estimate movement angles

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tary factors used in the studies of this thesis were maintained experimentally by evaluating the obtained movement angles and validating them with a well-established commercial IMU movement analysis system (Xsens MTw®, Xsens, Enschede, The Netherlands). Foot motions were recorded during gait and the IMU as well as the complementary filter used during the series of measurements were found to be highly comparable to those of the Xsens (see figure 10).

5.3 APPLICATION OF INERTIAL MEASUREMENT UNITS IN MOTION ANALYSIS

Motion capturing often involves complex, time consuming measurements and is frequently bound to lab environments with its corresponding constraints. They include the application of reflective markers on the body's segments, the localization of bony landmarks to set up local coordinate systems and the capturing of a reference posture. The reference posture is used to define a neutral joint position which serves as a zero-point and from which motions can then be calculated. IMUs are frequently used for ana-

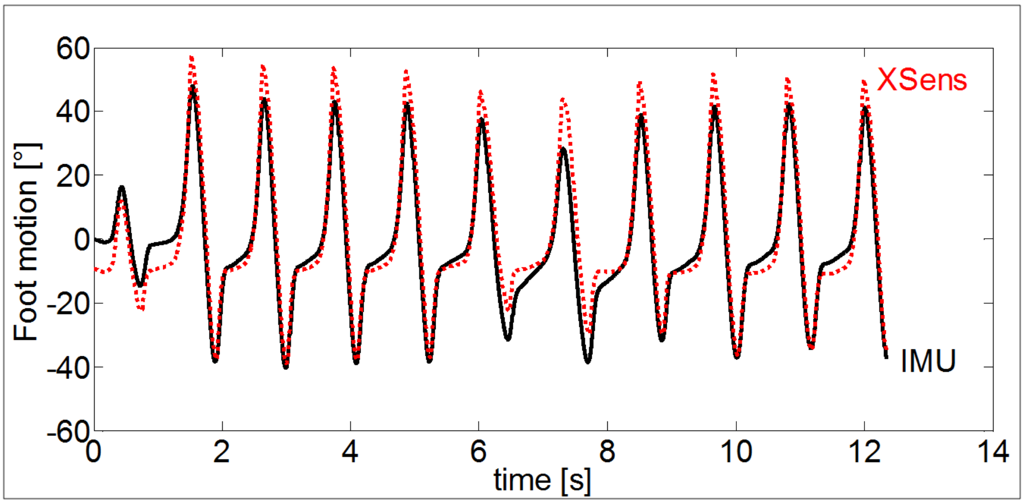

Figure 10: Comparison of data obtained with a custom-made IMU and a complementary filter used to apply to the data and the commercially available Xsens MTw showed high conformity.

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lyzing kinematics during human movement. They are low cost, can be used indoors and outdoors, wireless versions do not constrain movements due to cables and they do not require highly specialized personnel to perform measurements. The sensors have been validated with optical motion capture systems, which are currently considered the gold standard when recording motion trajectories (Hu & Soh, 2014; Leardini, Lullini, Giannini, Berti, Ortolani & Caravaggi, 2014). Errors were determined to be as small as 1° when evaluating joint angles (Seel, Raisch & Schauer, 2014). Even single IMUs were found suitable for application in gait analysis and showed good results when detecting temporal gait parameters like stride and step time (Trojaniello, Cereatti & Della Croce, 2014). Ultra light-weight IMUs with a weight of 2.9 grams were developed to record data over long time spans without limiting the user in his or her daily routine (Zecca, Saito, Sessa, Bartolomeo, Lin, Cosentino, Ishii, Ikai & Takanishi, 2013). It can therefore be concluded that IMUs are very useful sensors for biomchenical analysis with high potential in the field of kinematic research.

In the series of measurements described in chapter 8 - 11, a custom-made IMU was used to evaluate foot motions in the frontal plane. A single IMU was intentionally chosen to check the suitability of this measurement technique for possible in-shoe integration in future research. We limited the motion analysis to the frontal plane as past research has indicated this direction of movement to be most profound for the analysis of overuse risk factors. The sensor was attached to the heel cap of the right shoe using a metal clamp as shown in figure 11. The fixation of the sensor to the clamp was accomplished using double sided tape (tesafix® 51960, tesa SE, Norderstedt, Germany) in addition to two zip ties. The clamp was attached to the shoe using double sided tape underneath the clamp and adhesive tape straps (tesa extra Power®, tesa SE, Norderstedt, Germany) to secure the mounting.

Figure 11: The IMU was affixed to a metal clamp and mounted onto the shoe's heel cap to allow recordings during running.

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6

OSCILLATIONS AT THE ACHILLES TENDON

6.1 THEORY AND METHODS OF ACCELEROMETRY

Sensors can be described as transducers which convert physical or chemical properties into electrical signals. Accelerometers are sensors which detect accelerations and transcribe them into processable data types. Several classes of accelerometers are available, such as piezoelectric, strain gauge, piezoresistive, capacitive, reluctive, servo and magnetic accelerometers (Meydan, 1997). Most of today's accelerometers can be classified as micro-electro-mechanical sensors, also known as MEMS . The underlying principle of these sensors is based on a mass-spring model, taking Hooke's law and Newton's second axiom into account. Hooke's law is described in equation (3), with F representing the agent, D representing the spring rate and ΔL representing the change in spring length:

$$F = D * \Delta L \tag{3}$$

Newton's second axiom is one of the three basic principles of motion. It is described in equation (4), where m resembles the object's mass and a is the respective acceleration.

$$F = m * a \tag{4}$$

When a force is applied to a spring-mass model, the spring will generate a counteracting force proportional to the compressing or stretching force. The two parameters *mass* and *stiffness* of the spring can be controlled. Therefore, the resultant acceleration of the mass can be determined as shown in equation (5).

$$a = \frac{D * \Delta L}{m} \tag{5}$$

The accelerometers used in the studies described in the thesis at hand are based on piezoelectric effects to sense displacements of the proof mass, which is proportional to the applied acceleration. Piezoelectricity relies on the linear electromechanical relationship between the electrical and the mechanical state in certain crystals (Gautschi, 2002). A deformation of piezoelectric material causes a proportional electric polarization, which differs on opposite sides of the crystal. Therefore, electric charge of the material appears with mechanical stress. In a piezoelectric accelerometer, this electric charge is translated into voltage as output signal. The number of axes along which accelerations are measured in a sensor may vary between one and three. To collect accelerations in multiple directions, several layers of crystalline material are used, one for each direction. Further specifications of the device may include an output range, sensitivity and bandwidth capacities. The output specifies the measurement range of the accelerometer and may be as close as one g or as wide as several thousand g. The sensitivity of a sensor indicates the "input parameter change required to produce a standardized output change" (Du, 2014). The higher the sensitivity, the more accurate the measurement will be. The bandwidth of a sensor informs about the number of reliable samples taken per second. For highly accurate readings of impact forces for example, a bandwidth may have to be in the range of hundreds of Hz.

6.2 QUANTIFYING OSCILLATIONS USING ACCELEROMETERS

Accelerometers are widely used in biomechanical measurements when classifying gait parameters (Kavanagh & Menz, 2008), to assess physical strains during prosthetic gait (Bussmann, Damen & Stam, 2000), to detect heel contact and toe off events (Jasiewicz, Allum, Middleton, Barriskill, Condie, Purcell & Li, 2006), to monitor mobility (Schutz, Weinsier, Terrier & Durrer, 2002) or to quantify soft tissue oscillations (Boyer & Nigg, 2006b). According to Meydan (1997), accelerometers are the “preferred technique of measuring and/or monitoring shock and vibration“. In the present work, accelerometers were used to quantify oscillations at the Achilles tendon during running.

6.2.1 *Oscillations in Time Domain*

In the time domain, the accelerometer signal is displayed as a function of time. It gives a quickly comprehensible display option and allows the researcher first insight into the data. Oscillations in time domain are characterized by their amplitude and variables such as max amplitude or mean amplitude can be determined. Complex systems' behavior in the time domain can be explained in terms of differential equations (Schneider, 2008).

6.2.2 *Oscillations in Frequency Domain*

Every periodic time function can be displayed by an infinite number of sine or cosine waves with varying amplitudes and frequencies. Figure 12 shows an example of a random signals and its cosine components. Therefore, a signal can be decomposed into these cosine waves. Certain analysis techniques can transfer a signal from time domain to frequency domain. After transforming the signal it is displayed in the frequency domain and broken down into its frequency components. In the series of measurements described in chapters 8-11, a finite signal with discrete, equidistant data points is analyzed. This type of signal comes from sampling the data at a certain frequency. The continuous analog signal in real life is therefore transformed into a discrete signal. Thus, the Fast Fourier Transformation (FFT) as an optimization of the Discrete Fourier Transformation (DFT) is applied and a discrete, finite frequency spectrum is obtained. The FFT uses a reduced number of steps when calculating the Fourier coefficients compared to the DFT. After the application of a FFT information about the frequency components which make up the original signal are obtained. By transforming the signal into the frequency domain it changes its number range from real to complex numbers. The frequency spectrum can then be graphically illustrated in a bar diagram with the amplitude of the cosine wave on the vertical axis and the frequency of the oscillation on the horizontal axis. The resulting diagram shows a discrete line spectrum of the oscillation and information about amplitude and frequency distribution can be obtained from these graphs. It should be noted that, to detect a frequency in a signal, at least twice the sampling frequency of that frequency is required. Therefore the highest detectable frequency is limited to be half the sampling frequency.

Signal energy is indicated by the area under the squared curve of the frequency spectrum as shown in equation (6). However, it would become infinitely large with a longer signal. Therefore, the average power of a signal is the area under the squared curve, divided by the number of periods over which the signal goes, as shown in equation (7).

$$E_x = \sum_{n=0}^{N-1} (x_n)^2 \tag{6}$$

$$P_x = \frac{E_x}{N} \tag{7}$$

Oscillations at the Achilles tendon are forced oscillations, meaning the system is exposed to external actions. External actions, in this case force inputs which are forwarded from the foot in longitudinal direction of the leg as well as foot position and orientation and muscle activation, define the period of oscillations. If the input frequency coincides with the natural frequency of the system, resonance occurs at the so-called dominant frequency (Aladjev &

Bogdevicius, 2006). At dominant frequencies, the oscillation amplitude (peak displacement), its velocity and therefore its acceleration will increase and the system will get into an unstable state. The dominant frequency, at which resonance occurs at the Achilles tendon may appear if the frequency of the input signal (impact at heel strike) is close to the natural frequency of the Achilles tendon. The dominant frequency appears as a very high peak or peaks on the frequency spectrum, plotted after FFT was applied. There is evidence in the literature, that if a biological system is excited with an input frequency close to its natural frequency, the power dissipated by soft tissues is increased (Boyer & Nigg, 2004; Wakeling, Liphardt & Nigg, 2003). Also, muscle activity is modulated at this input frequency just before and following the onset of an externally applied oscillation. The resonance frequency of soft tissue changes with running speed (Boyer & Nigg, 2004). However, only frequency not amplitude varies when changing the input signal. Soft tissue compartments (muscle compartments) were found not to vibrate like a rigid body but rather as a continuum (Wakeling, Pascual & Nigg, 2002; Pain & Challis, 2002). However, no research was performed on oscillations at the Achilles tendon. Results from soft tissue compartments may not be transferred to the Achilles tendon as muscle tissue and tendon tissue vary considerably. The amount of energy dissipated by the Achilles tendon as well as its resonance frequency and signal power can be estimated from acceleration measurements.

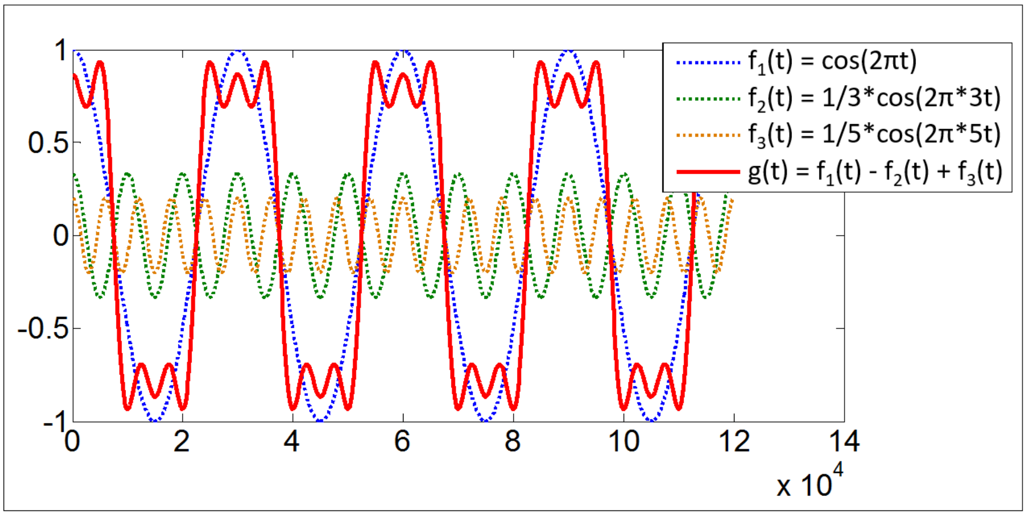

Figure 12: Random signal (solid) and its cosine components (dotted)

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7

STATISTICAL ANALYSIS

7.1 ANALYSIS OF VARIANCE

Analysis of variance (ANOVA) is a statistical procedure to test for differences between group means. Field (2009) describes ANOVA as "statistical procedure that uses the F-ratio to test the overall fit of a linear model". It is performed if more than two groups or more than two conditions should be compared or if more than one independent variable is included in the data set. It is not recommended to use multiple t-tests in the aforementioned cases as it would increase the chance of finding a statistically significant result by chance (O'Donoghue, 2012). The probability of type 1 errors is usually accepted at a 5% level. Therefore, one out of 20 t-tests randomly shows statistically significant differences while ANOVA corrects for the likelihood of significant results due to numerous comparisons. For example, if comparing three groups with multiple t-tests instead of an ANOVA, the likelihood of a type 1 error would increase from 5% to 14.3% because type 1 error = $1 - (0.95)^n = 1 - (0.95)^3 = 0.143$.

An important outcome measure of ANOVAs is the so called F-value, which is based on an F-test and informs about the main effects of the investigated factor on a specific parameter. It allows the differentiation of two groups with regard to their variance and informs about treatment effects in clinical or biomechanical studies. It resembles the amount of variability, which is due to a treatment. If the F-value is $\leqslant 1$, no treatment effect was found. In scientific documents, the F-value is usually specified together with the degrees of freedom of the test. The first degree of freedom gives information about the number of groups to be compared and is calculated as the difference between the number of groups (or comparisons) and 1. The second degree of freedom resembles the difference between the number of observations or subjects and the number of groups or comparisons (Lawner-Weinberg & Knapp-Abramowitz, 2008). While the F-test informs about whether or not statistical differences between the groups exist or not, it does not include disclosure about which comparison shows differences. To find differences between the investigated groups or comparisons, follow-up tests need to be made. These are usually performed through modified t-tests with Bonferonni-correction. Bonferonni-correction drops the significance level of a test by dividing the p-value by the number of groups or comparisons made. Field (2009) recommends Bonferonni-correction if aiming at tight control over type 1 error.

The results of an ANOVA may not only include main effects but also interaction effects. If an interaction effect is found to be statistically significant, the effect of the first factor differs depending on the occurrence of the second factor and vice versa (Quinn & Kenough, 2002). The difference between the variables' means therefore do not just reflect the effects of each factor. Thus, the detection of interaction effects is another advantage of ANOVA over multiple t-tests, which do not have the power to reveal interactions.

7.2 PRINCIPAL COMPONENT ANALYSIS

At the starting point of explorative research, many variables of a biological system may seem to be of concern. A large number of variables hinder the interpretation of statistical results and may disguise the actual outcome. Principal component analysis (PCA) can be described as a statistical tool to filter out redundant information in a data set, which is not required for its final outcome. PCA identifies groups or clusters of variables and is applied to reduce a data set to a manageable number of parameters while retaining most of the information originally included (Field, 2009). The result therefore becomes clearly understandable and can be interpreted. PCA reports correlations or discrepancies between variables through quantitative descriptions. To do so, statistical variables are approximated by a lesser or equal number of linear combinations, which are represented by linearly uncorrelated principal components. These principal components are orthogonal, represent the eigenvectors of the covariance matrix and can therefore be graphically illustrated if the number of components is $\leqslant 3$. The first principal component always accounts for the largest amount of variance in the data set while subsequent components explain a decreasing amount of variance.

There are two main outcome measures in PCA: factor scores and factor loadings. Factor scores are the transformed values of a variable. They represent the coordinates of this variable, projected in a factor coordinate system. Factor scores can be graphically represented in so called score plots, which are characterized by each axis representing a factor, with measurement variables plotted along these axes. They are a useful tool to quickly inspect the strength of relationships between each factor and specific variables as well as to find clusters or groups of variables. Factor loadings represent the loading of each factor on a variable, illustrating the correlation of that variable with the factor. The variable is multiplied by its factor loading to obtain the corresponding principal component score.

If numerous factors are found in a data set, not all of them are retained in subsequent analysis. A factor's eigenvalue contains information about the substantive importance of this specific factor. It describes the variance explained by this factor (Kaiser, 1960). Therefore, factors with large eigenvalues are retained in subsequent analysis. Cartell (1966) suggested a visualization through so called scree plots. In scree plots, each factor is plotted against its eigenvalue and the importance of each factor can be read from the graph. Typically these graphs have a sharp descent on the left while they even out towards the right. Cartell (1966) recommended the inflection point of the graph as a cut-off point when selecting relevant factors. Factors to the left of the inflection point should be considered in further analysis. Although scree plots are a useful tool, they may turn out ambiguously with no clear inflection point. Kaiser (1960) recommended all factors with an eigenvalue $\geqslant 1$ to be retained. An eigenvalue $\geqslant 1$ represents a substantial amount of variation, which is explained by this factor. In the present manuscript, PCA was used for data reduction in chapter 10 and score plots were created to separate different stages of fatigue in biomechanical data. Kaiser's criterion was applied to decide about a factor's importance as it provides a clear quantitative measure. Additionally, scree plots were created in each PCA to control the decision on factor retainment.

Part IV

APPLICATION OF THE INTEGRATED MEASUREMENT SYSTEM

8

EFFECT OF RUNNING SPEED AND SHOE MODIFICATIONS ON THE COLLECTED DATA

Table 3: Details of study 1, comparing shoe modifications while running at different speeds

Details of the Study	
N	20 ♂
Age [years]	22.5 ± 1.4
Height [meters]	1.83 ± 5.72
Weight [kg]	81.9 ± 9.0
Weekly running distance [km]	19.4 ± 10.5
Measurement systems	3D IMU
	Two 3D accelerometers
	F-Scan plantar pressure system
	EMG
Conditions	Three types of footwear
	(Con1, Con2, NS)
	Three running speeds
	(2.9 m/s, 3.5 m/s, 4.2 m/s)

8.1 INTRODUCTION

Shoe modifications are frequently used to influence foot kinematics or impact forces, aiming at injury prevention (Shih, Wen & Chen, 2011; Richards, Magin & Callister, 2009; Hreljac, 2005). Common modifications to control foot pronation are medially posted insoles, modified heel counters or variations in the cushioning material of the sole. These modifications were shown to influence foot kinematics by reducing peak eversion (Rodrigues et al., 2013), rearfoot angle (Perry & Lafortune, 1995), peak eversion velocity and ROM (Rodrigues et al., 2013). In a large meta-analysis, Cheung, Chung & Ng (2011) found motion control shoes to significantly reduce rearfoot motion. The authors found running shoes with medial wedges or heel flare to be more effective in controlling eversion than those with dual midsole materials. Nevertheless, other researchers found high inter-subject variability in rearfoot kinematics and therefore diverse effects of different shoe sole constructions on tibiocalcaneal motion (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth & Stüssi, 2001; Kersting & Brüggemann, 2006). Similarly, diverse conclusions have been drawn with regard to the clinical effect of motion control shoes. Some authors found motion control shoes to have positive effects on AKP and Achilles tendon loading rate (Shih et al., 2011; Sinclair, Taylor & Atkins, 2014) while others found no controlled trials or systematic reviews proving an effect of these shoes in reducing injury rates in runners (Richards et al., 2009).

Running speed is frequently discussed as an injury inducing factor. de David, Carpes & Stefanyshyn (2014) found joint loading of the ankle and knee to increase with increasing running speed. Higher peak moments may lead to harmful loading of these structures. Also, torques and power at the ankle as well as the work done at this joint increase when changing running speed from 3.5 to 5.02 m/s but do not vary significantly with further increase in running speed (Schache, Blanch, Dorn, Brown, Rosemond & Pandy, 2011). Other authors, however, found injured runners to choose a lower running speed compared to healthy runners (Peng, Seay, Montero, Barnes, Vincent, Conrad, Chen & Vincent, 2014). Yet, this may rather be an effect than a cause of the injury. To our knowledge, no study has yet investigated the dependency of shoe modification effects on different running speeds. Even though there are multiple studies suggesting a connection between rear foot motion and the incidence of running injuries, some authors found no association between the two. Nielsen, Buist, Parner, Nohr, Sorens, Lind & Rasmussen (2014) could not detect overuse injuries related to foot pronation when studying a large cohort of novice runners who recently took up this sport. It should be noted though, that the researchers used the Foot Posture Index to categorize study participants into groups according to their static foot posture. Therefore, these findings may not be controversial to those mentioned earlier as the dynamic motion in the frontal plane was not examined.

Foot kinematics as well as runners' reactions to shoe modifications seem to be highly individual concerns. An easy to use tool is needed to assess kinematic responses to differences in footwear not only in a lab environment but also in the field. A modular running shoe system with a multitude of possible configurations may be a useful appliance when assessing an individual's response to variations in footwear. Therefore, the goal of the present study was to determine foot kinematics at different running speeds using an IMU while wearing two differently configured running shoes as well as a neutral all-purpose shoe. Decreased ROM and MaxProVel were hypothesized to occur when running in shoes that were equipped with motion control components like medial wedges. As these shoes aim at limiting foot kinematics in the frontal plane, it was hypothesized that the final pronation following initial contact would occur earlier than in the neutral all-purpose shoe. Previous research showed increased ankle joint excursions with increasing running speed in barefoot runners (Bishop, Fiolkowski, Conrad, Brunt & Horodyski, 2006). We therefore hypothesized to find an increase in ROM and MaxProVel at higher running speeds.

During running, repeated impacts are transferred in axial direction along the lower leg, possibly affecting the vibration behavior of the Achilles tendon. Large oscillations may impair the musculoskeletal system and are minimized by muscle tuning in order to prevent damage (Wakeling & Nigg, 2001). Boyer & Nigg (2004) found a correlation between impact loading rate and muscle activation. The authors proposed that muscle activity is adapted to impact forces to control soft tissue vibrations. Another study showed that variations in muscle activity are associated with different impact characteristics, resulting in modifications of the acceleration transmissibility of soft tissues (Boyer & Nigg, 2007). Increased ground reaction forces and increased muscle activity are expected to occur at high running speeds (Schache & Dorn, 2014). Impact attenuation is an important function of running shoes. Therefore, shoe modifications may influence plantar pressure distributions as well as muscle activation characteristics (Nigg, Stefanyshyn, Cole, Stergiou & Miller, 2003). The present study aimed at exploring the effects of different running speeds and shoe modifications, as well as their interaction, on impact forces underneath the foot and on the activity characteristics of selected lower leg muscles. Changes in these parameters may lead to variations in the vibration behavior of the Achilles tendon. Achilles tendon elastic strain energy is known to increase with higher running speeds (Lai, Schache, Lin & Pandy, 2014), possibly leading to an augmented risk of sustaining harmful stresses at this structure. Therefore, the oscillation behavior of the Achilles tendon was explored at different running speeds, while wearing different types of footwear and between three measurement directions. Mercer, Vance, Hreljac & Hamill (2002) found shock attenuation along the runner's body to increase linearly with running speed. Higher accelerations were expected to occur at the distal end of the tendon due to impact attenuation. Maximum accelerations were hypothesized to increase with running speed as plantar pressure increases. The longitudinal axis of the lower leg represents the main direction of impact transmission. We therefore expected accelerations in longitudinal direction to be higher compared to those in the two orthogonal directions. Therefore, the overall purpose of the present study was to test the effects of different footwear modifications and different running speeds on plantar pressure data, muscle activity, foot kinematics and oscillations at the Achilles tendon.

8.2 METHODS

Twenty male rearfoot runners participated in this study (see table 3). Subjects were given a six minute warm up period at a self-selected running speed to get accustomed to the treadmill (Woodway ERGO XELG 90®, Woodway USA Inc., Waukesha, WI, USA). Past research has shown no significant differences in kinematics of the lower extremity and therefore a successful familiarization after running on a treadmill for six minutes (Lavcanska, Taylor & Schache, 2005). Subjects were asked to run at three different running speeds (2.9, 3.5, 4.2 m/s) wearing two different running shoe configurations as well as a neutral all-purpose shoe (NS; Adidas Gazelle®, Adidas, Herzogenaurach, Germany). A modular running shoe (Runaissance 3.0®, Newline, Vodskov, Denmark) was used to provide two different running shoe configurations. This system allows the modification of three components of the shoe: footbed, wedges and cushioning (see figure 13). Three different footbed insoles with varying arch supports are provided by the manufacturer as well as three different medial wedges, which are integrated in separate insoles. The wedges come in 2, 3 and 4 mm thickness. Another customization can be performed by changing the cushioning characteristics of the shoe. Therefore, cushioning inserts at the first metatarsophalangeal joints and at the postero-lateral heel can be exchanged. Again, three different cushioning inserts are provided by the producer. In the present study, we used the modular running shoe system to provide two different configurations of this shoe: one with high arch support, medial wedges (4 mm) and soft damping material (Con1), the other with low arch support, no medial wedges and hard damping material (Con2). All shoes were tested according to American Society of the International Association for Testing and Materials (ASTM) F-1976 standards. The results of these tests are shown in table 4.

An insole plantar pressure system (F-Scan®, Tekscan Inc., South Boston, MA, USA) was used for step detection. Recordings were

Table 4: ASTM F-1976 test results of the shoe configurations used in this study

ASTM F-1976 test results			
Shoe	Maximum Force [N]	G-Score (Peak)	Peak-to-peak ratio [%]
Con1	867.3	14.3	55.6
Con2	914.9	15.6	55.5
NS	1546.5	24.1	50.7

performed at 100 Hz and synchronized with records of the other biomechanical parameters described below. De Cock, De Clercq, Willems & Witvrouw (2005) introduced four distinct phases of the stance during running which are described in detail in table 5. In the current study, these step phases were also determined and data was analyzed within these phases. The results of these analysis were used to verify the data obtained from EMG and kinematics and will not be discussed in detail. Results of the examined step phases can be found in the appendix of this manuscript.

During measurements subjects were equipped with an IMU which was firmly attached to the heel cap of their right shoe. Foot kinematics were evaluated in the frontal plane during stance phase. A static measurement was performed before the trials in each shoe to define the neutral position as a reference. Steps were identified through heel contact and toe off in the foot pressure data. Acceleration data from the IMU were low pass filtered to remove gravitational influence. Then, acceleration and gyroscope data were combined with a complementary filter and integrated to get movement angles of the IMU. Motions in the frontal plane will further be denoted as pronation and supination although it should be noted that in this study they do not describe motions of the rearfoot relative to the tibia. Maximum Pronation (MaxPro), MaxProVel, ROM in the frontal plane as well as time to final pronation following foot con-

Figure 13: The modular running shoe was used to set up two different configurations of this shoe. Damping material and insoles can be exchanged in this shoe to obtain individual configurations.

tact (TFPro) were used as dependent variables for statistical analysis. These parameters were calculated per step and subsequently averaged over all steps per trial.

Vibrations at the Achilles tendon were measured in 20 male runners using two skin mounted triaxial accelerometers (Noraxon®3D inline accelerometer, Model 317A). These sensors were chosen as they have a relatively small weight of 2.8 g which reduces the self-oscillations of the device. Double sided tape was used to attach the accelerometers to the skin. Additionaly, blue Kinesio sports tape (Atex Medical®, Gyeonggi-do, Korea) secured and preloaded the device. Congruence of motion with the underlying skin was therefore improved. A study by Rosso, Schuetz, Polzer, Weisskopf, Studler & Valderrabano (2012) revealed a significant correlation between the length of the Achilles tendon and the length of the tibia. In order to standardize sensor locations between subjects, tibia length of each participant was determined from ground to tibial plateau. The distal accelerometer was attached at 26% of tib-

Table 5: Temporal parameters to describe step phases of total foot contact

Step Phases		
Stance Phase	Start	End
Initial contact (ICP)	1st foot contact	1st metatarsal contact
Forefoot contact (FFCP)	1st metatarsal contact	Forefoot flat
Foot flat (FFP)	Forefoot flat	Heel off
Forefoot push off (FFPOP)	Heel off	Last foot contact
Total contact time (TCT)	1st foot contact	Last foot contact

ial length (from the ground up) and the proximal accelerometer was attached at 36% of the tibia. The distal location was chosen as it reflects accelerations occurring about 5 cm above the calcaneal insertion point. This area is highly susceptible to inflammations of the tendon which may be caused by poor blood supply (Clain & Baxter, 1992). It was not possible to select an application location further distal as the sensor would have interacted with the shoe during running. The proximal location was chosen at 36% as it represents the most proximal fixation point that did not interact with the cuff used to secure the plantar pressure measurement system (see figure 14). The devices were aligned with each other and parallel to the Achilles tendon.

During running, EMG of the following muscles of the right leg was recorded: M. Gastrocnemius medialis (M. Gastr. med.), M. Gastr. lat., M. Tib. ant. and M. Peroneus long. Skin preparation and electrode placement was performed according to the recommendations of the *International Society of Electrophysiology and Kinesiology (ISEK)*.

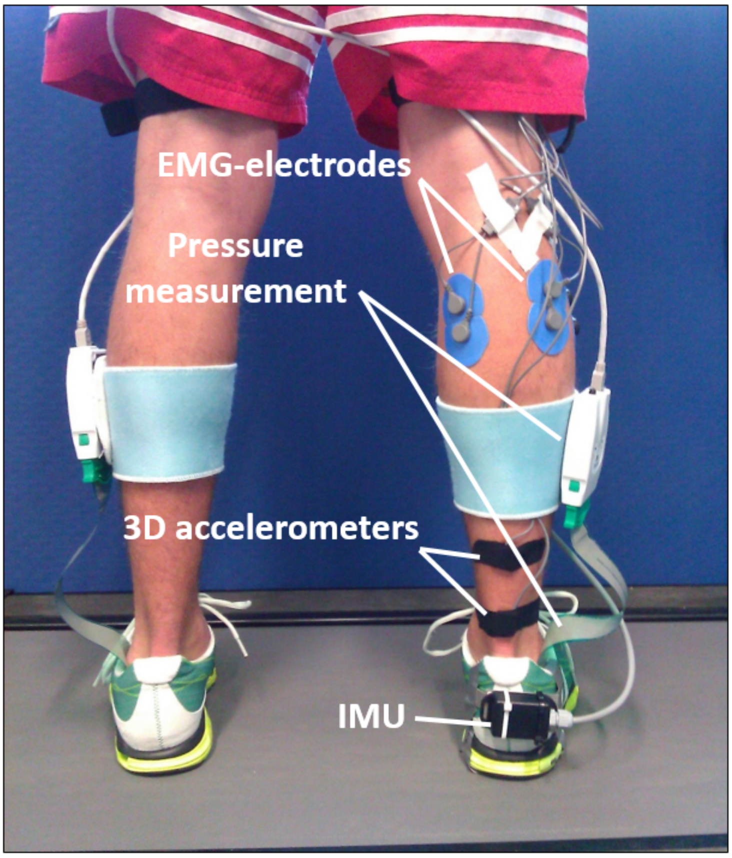

Figure 14: Subjects were equipped with an insole plantar pressure measurement system, EMG measurement devices, two accelerometers at the Achilles tendon and an IMU at their right heel cap.

Electrode locations were chosen at the middle of each muscle belly to prevent sliding off of the muscle underneath the electrode. The skin was then shaved at this location, danders were removed through sanding and sebum was removed using an antiseptic (Softasept® N, B. Braun Melsungen AG, Melsungen, Germany). Cables of the electrodes were taped to the subjects skin to prevent artifacts. Kendall® ECG electrodes (Covidien Ilc, Mansfield, MA, USA) were used in this study. Electrodes were cut on one side each to assure the recommended inter electrode distance of 2 cm. This way unstable recordings, which may occur due to crosstalk between different muscles are avoided. Also, oast research has shown that interelectrode distance influences EMG results. Larger distances lead to increases in average amplitude of the EMG signal (Zedka, Kumar

& Narayan, 1997). Therefore, standardization of interelectrode distance is of high importance for the results. For each muscle average activation during stance phase was determined for each step and averaged per trial. Data of the IMU, the two accelerometers at the Achilles tendon and EMG data were synchronously recorded at a sampling frequency of 3000 Hz using Noraxon Telemyo 2400 G2 (Noraxon Corporate, Scottsdale, AZ, USA).

To determine if there were significant differences in the dependent variables between shoe conditions or running speeds two-way repeated measures ANOVAs were performed. Separate ANOVAs were run for each dependent variable. The assumption of normality was checked and the data were found to meet the criteria. Violations of sphericity were controlled using Maulchy's test of sphericity and either Greenhouse-Geisser or Huynh-Feldt corrections were applied according to Girden (1992). If a Greenhouse-Geisser epsilon of > 0.75 was found, the Huynth-Feldt corrected value was used for that parameter. Otherwise the Greenhouse-Geisser corrected value was used. Post hoc tests were performed using modified t-tests with Bonferroni correction. All statistical calculations were completed using SPSS (SPSS 21, IBM, Armonk, NY, USA) and the alpha-level was set at 0.05.

8.3 RESULTS

8.3.1 *Plantar Pressure Distribution*

TOTAL FORCE Running speed had a significant main effect on total force detected underneath the foot during stance, $F(2, 38) = 17.06$, $p < 0.01$. A significant increase in total force was detected when running at 3.5 m/s (1676 ± 501 N) compared to 2.9 m/s (1583 ± 450 N, $p < 0.01$) and when running at 4.2 m/s (1732 ± 542 N) compared to 2.9 m/s ($p < 0.01$). However, no difference could be proven between running trials at 3.5 m/s and 4.2 m/s ($p = 0.11$).

PEAK FORCES A significant main effect was found for configuration, $F(1.66, 31.58) = 4.19$, $p = 0.03$. Average peak forces were significantly lower when running in Con1 (45 ± 13 N) compared to running in NS (51 ± 15 N). Further, running speed had a main effect on average peak forces, $F(2, 38) = 14.64$, $p < 0.01$. Similar to the results of overall forces, peak forces differed between running at 2.9 m/s (44 ± 13 N) and running at 3.5 m/s (47 ± 14 N) as well as between running at 2.9 m/s and running at 4.2 m/s (49 ± 15 N). Again, no difference could be detected between running at 3.5 m/s and running at 4.2 m/s ($p = 0.27$).

8.3.2 *Muscle Activity*

AVERAGE ACTIVATION OF M. GASTR. LAT. Running speed had a main effect on average activation of the M. Gastr. Lat. ($F(1.256, 23.75) = 44.36$, $p < 0.01$). The results obtained for this muscle differed significantly between all running speeds ($p < 0.01$). In agreement with the expectations, average activation was lowest when running at 2.9 m/s (107 ± 56 μV), medium when running at 3.5 m/s (131 ± 67 μV) and highest at 4.2 m/s (164 ± 88 μV).

AVERAGE ACTIVATION OF M. GASTR. MED. Configuration and running speed had a main effect on average activation of the M. Gastr. Med. ($F(1.90, 36.16) = 4.76$, $p = 0.02$ and $F(1.28, 25.32) = 27.40$, $p < 0.01$ respectively). The results obtained for this muscle showed a significantly lower average activation when running in NS (122 ± 59 μV) compared to running in Con2 (142 ± 79 μV; figure 16). Also, increasing muscle activity was found with increasing running speed ($p < 0.01$; figure 15).

AVERAGE ACTIVATION OF M. TIB. ANT. For the analysis of M. Tib. ant. only 19 subjects were included as measurement difficulties occurred for subject number 20 with this muscle, probably due to a damaged cable. Therefore, no data were recorded for this sub-

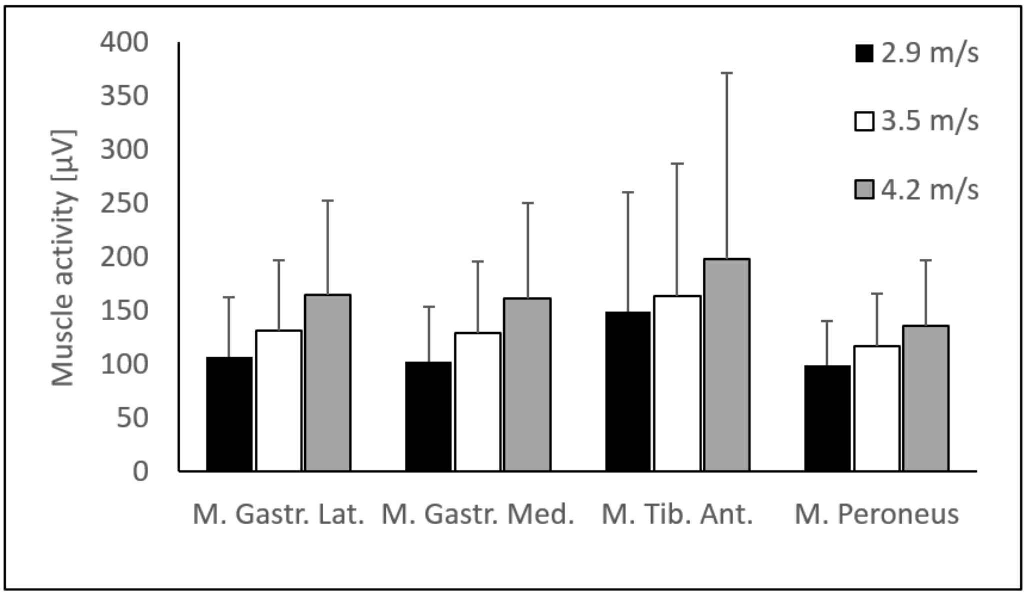

Figure 15: Average muscle activity of all four muscles when running at different running speeds. For each of the muscles, activation levels differed between all running speeds. For reasons of clearity it is not marked separately in the figure.

ject. The following results were obtained from data including the remaining 19 subjects. Running speed had a main effect on average activation of the M. Tib. Ant. ($F(1.33, 23.84) = 15.50$, $p < 0.01$). Activation levels were increasing significantly with increasing running speed ($p < 0.04$). When running at 2.9 m/s activation was lowest (149 ± 111 µV), while it increased when running at 3.5 m/s (164 ± 123 µV) and 4.2 m/s (198 ± 174 µV).

AVERAGE ACTIVATION OF M. PERONEUS LONG. A significant main effect was found for running speed, $F(1.27, 24.20) = 28.67$, $p < 0.01$. Pairwise comparison showed increasing muscle activity with increasing running speed ($p < 0.01$). Highest activation was found at 4.2 m/s (136 ± 61 µV) while lowest was detected at 2.9 m/s (100 ± 41 µV). Figure 15 shows average activation levels for all muscles examined in this study at different running speeds. In summary, all muscles increased activity with increasing running speed.

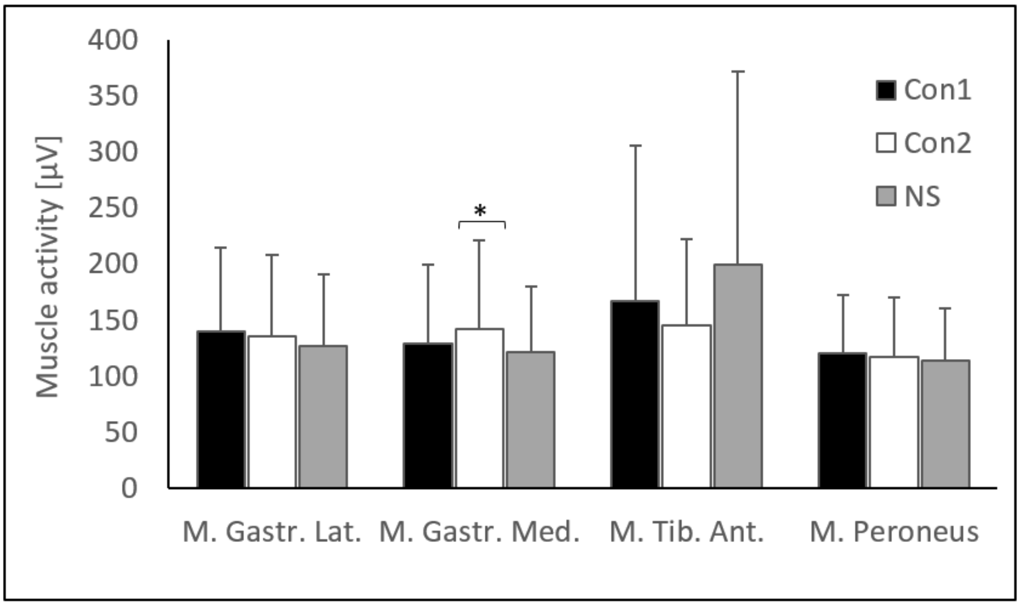

Figure 16: Average muscle activity of all four muscles when running in different running shoes. Significant differences are marked by a star.

8.3.3 *Foot Kinematics*

Two examples of foot motion in the frontal plane are shown in figure 17. Subjects were either wearing Con1 or NS. The upper two graphs (figure 17a) show a subject reacting to different running shoes by variation of foot motions while the subject to whom the bottom graphs (figure 17b) belong, shows very constant trajectories, not reacting to differences in footwear.

maximum pronation MaxPro did not differ when varying shoe conditions (p = 0.86) or running speed (p = 0.63).

maximum pronation velocity Significant main effects were found for configuration, F(1.55, 29.37) = 4.00, p = 0.04 and running speed, F(1.84, 34.9) = 24.68, p < 0.01. A mean increase of 30.5 °/s in MaxProVel was found while running in the NS (132.9 ± 140.3 °/s) compared to Con1 (102.4 ± 108.5 °/s; p = 0.03). MaxProVel also differed significantly between running at 2.9 m/s (102.2 ± 108.1 °/s)

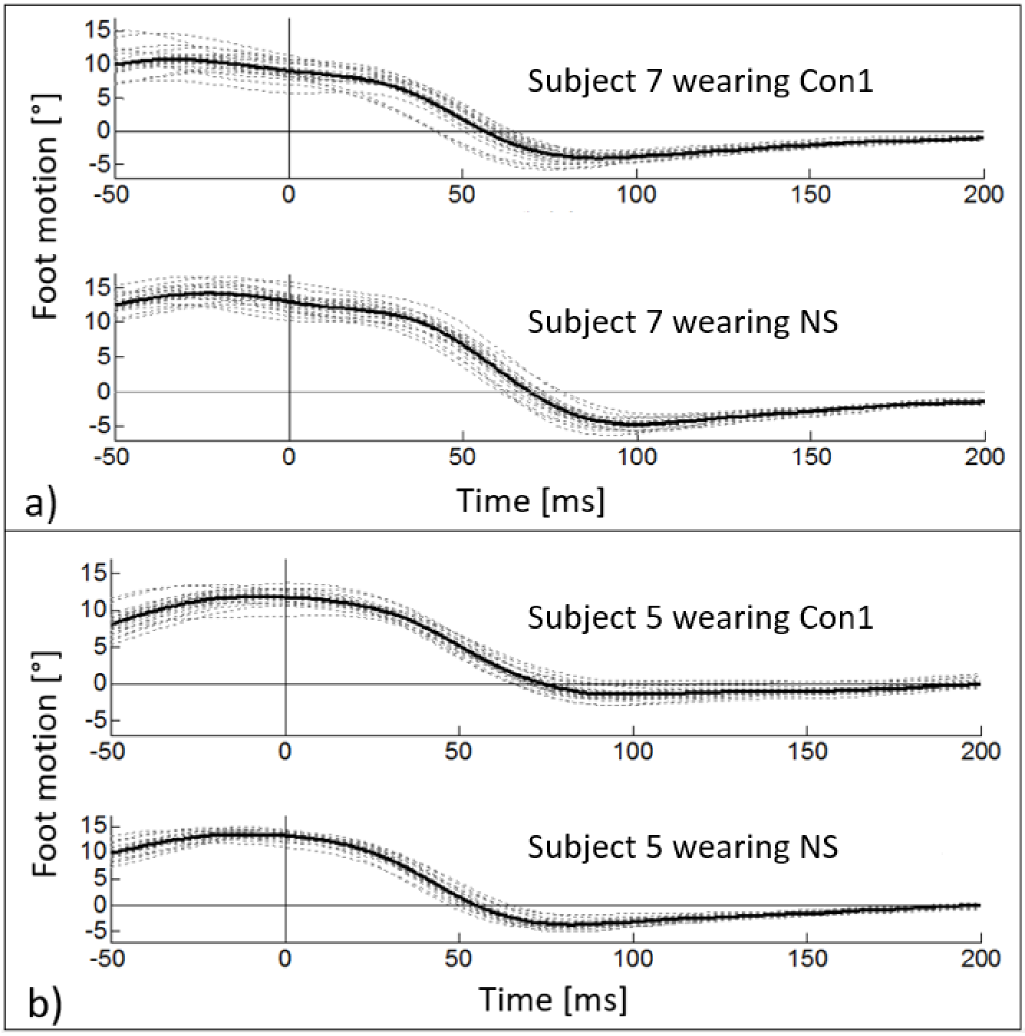

Figure 17: Foot motion in the frontal plane of two exemplary subjects. Solid lines show mean data while dashed lines represent single steps. First foot contact occurs at time = 0 ms. Subject 7 reacts to differences in footwear (figure 17a) while subject 5 shows constant trajectories, not responding to different shoes (figure 17b).

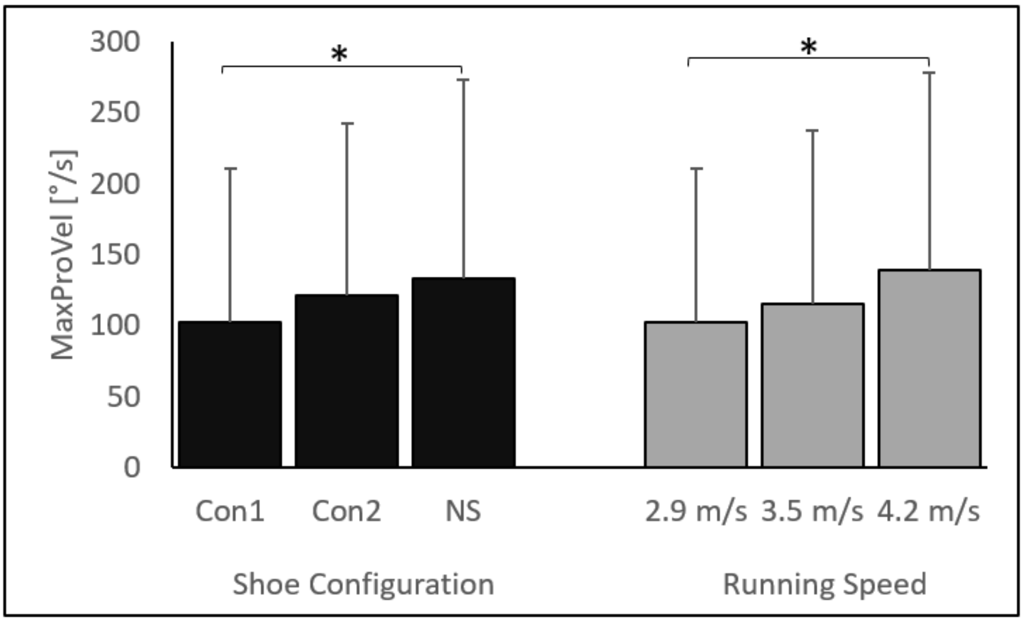

Figure 18: MaxProVel averaged over shoe configurations and running speeds. Significant differences are indicated by a star.

and running at 4.2 m/s (138.7 ± 139.6 °/s; $p < 0.01$). A graphical illustration of pairwise comparisons of all main effects can be seen in figure 18. No significant interaction effects between the examined parameters were found.

RANGE OF MOTION Significant main effects were found for configuration $F(1.94, 36.94) = 13.48$, $p < 0.01$ and running speed $F(1.46, 27.72) = 8.35$, $p < 0.01$. Subjects showed a significantly higher ROM while running in NS (5.6 ± 4.9°) compared to running in Con1 (4.3 ± 4.3°; $p < 0.01$) or Con2 (4.8 ± 4.7°; $p = 0.04$). An increase in ROM could also be observed at 4.2 m/s (5.2 ± 5.3°) compared to 2.9 m/s (4.5 ± 4.4°, figure 19).

TIME TO FINAL PRONATION Non of the investigated comparisons revealed a significant difference between shoe configurations ($p = 0.4$) or running speeds ($p = 0.4$) in TFpro. Also, no interaction effect was found for configuration*speed ($p = 0.8$).

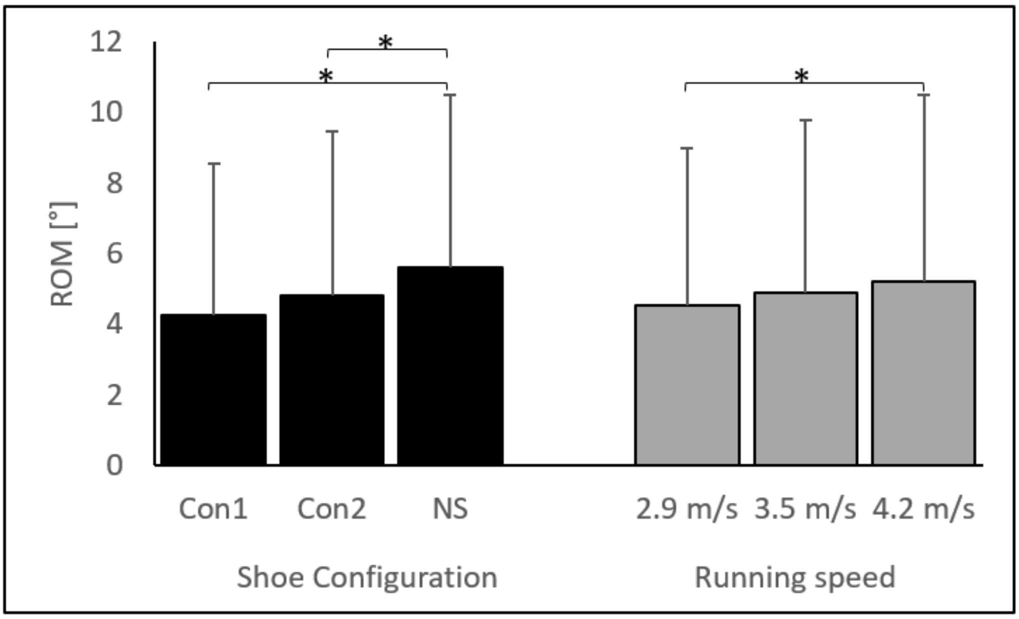

Figure 19: ROM averaged over shoe configurations and running speeds. Significant differences are indicated by a star.

8.3.4 *Oscillations at the Achilles tendon*

8.3.4.1 *Proximal accelerometer - time space*

PEAK ACCELERATIONS Significant main effects on peak accelerations (PeakAcc) were found for running speed, $F(1.80, 34.14) = 130.87$, $p < 0.01$, and for measurement direction, $F(1.87, 35.43) = 11.86$, $p < 0.01$. PeakAcc measured while subjects ran at 2.9 m/s (38 ± 13 m/s^2) were significantly lower than these measured at 3.5 m/s (44 ± 13 m/s^2) which were again significantly lower than these detected at 4.2 m/s (52 ± 13 m/s^2), as can be seen in figure 20. Higher PeakAcc were found in medio-lateral direction (53 ± 13 m/s^2) compared to cranio-caudal direction (43 ± 15 m/s^2) and in medio-lateral direction compared to anterior-posterior direction (38 ± 10 m/s^2), (see figure 20). No difference was found between either of the three shoe conditions, $F(1.81, 34.36) = 0.24$, $p = 0.77$ (see figure 20).

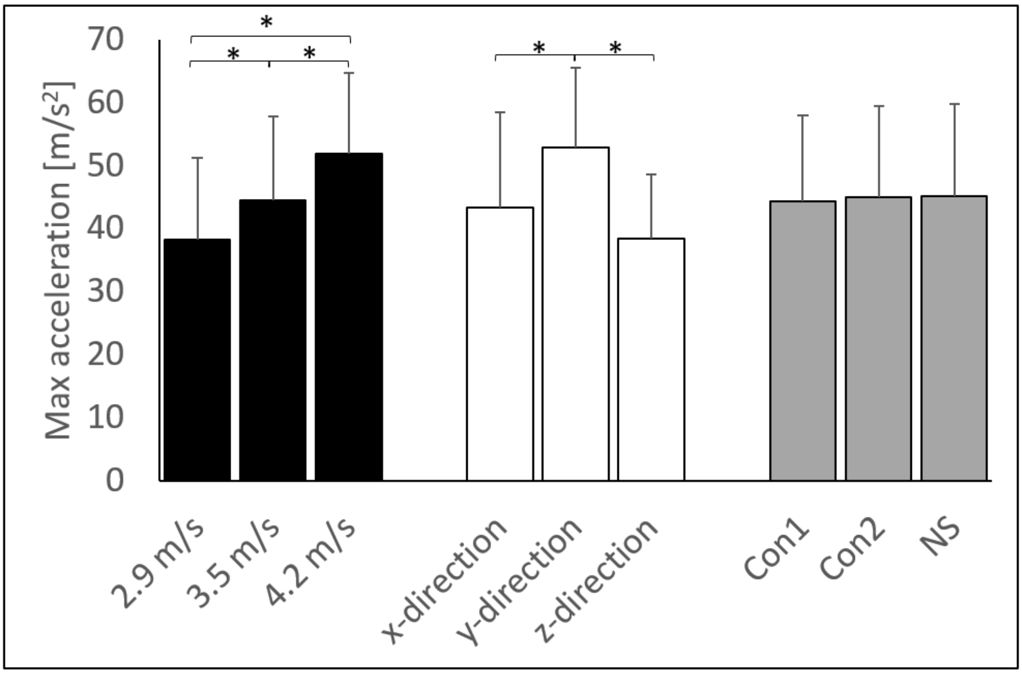

Figure 20: PeakAcc detected with the proximal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

AVERAGE ACCELERATIONS Measurement direction and running speed had main effects on average accelerations (AvgAcc) detected by the proximal sensor, $F(1.31, 24.86) = 64.72$, $p < 0.01$ and $F(1.92, 36.55) = 28.73$, $p < 0.01$ respectively. AvgAcc in anterior-posterior z-direction were highest (8 ± 5 m/s^2), those in cranio-caudal x-direction were medium (6 ± 2 m/s^2) and these measured in medio-lateral y-direction were lowest (2 ± 2 m/s^2). Accelerations also increased with increasing running speed as can be seen in figure 21.

A significant two-way interaction effect was found for Direction* Speed, $F(2.07, 39.40) = 7.74$, $p < 0.01$. While average accelerations were always higher with higher running speeds, the speed variations mainly influenced average accelerations at the x-axis while they stayed relatively constant on y-axis and z-axis (figure 22).

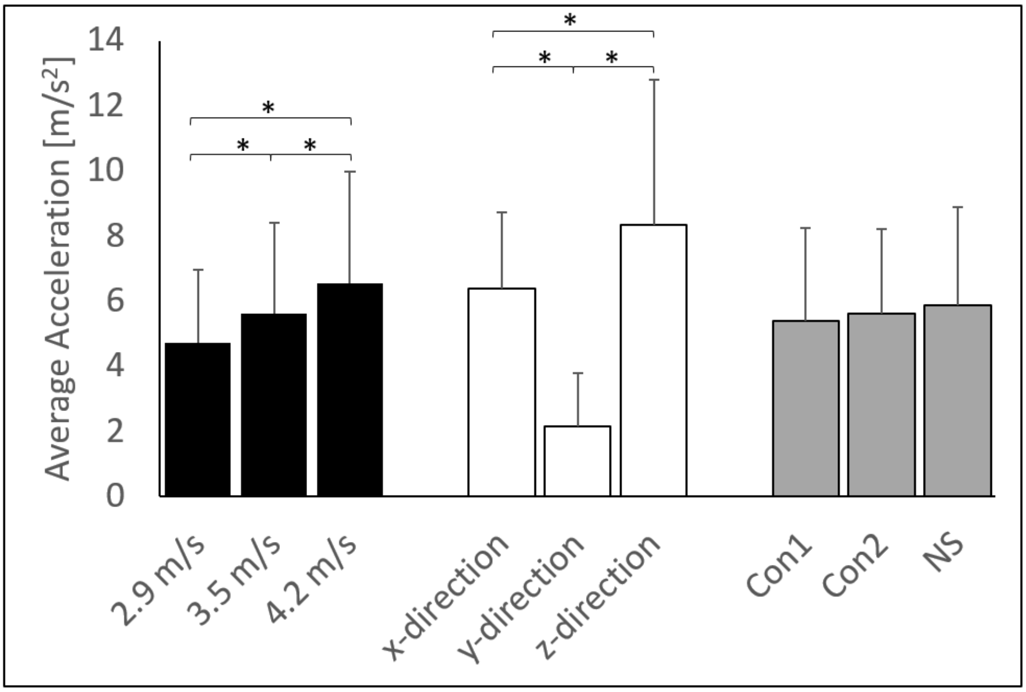

Figure 21: Average accelerations detected with the proximal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

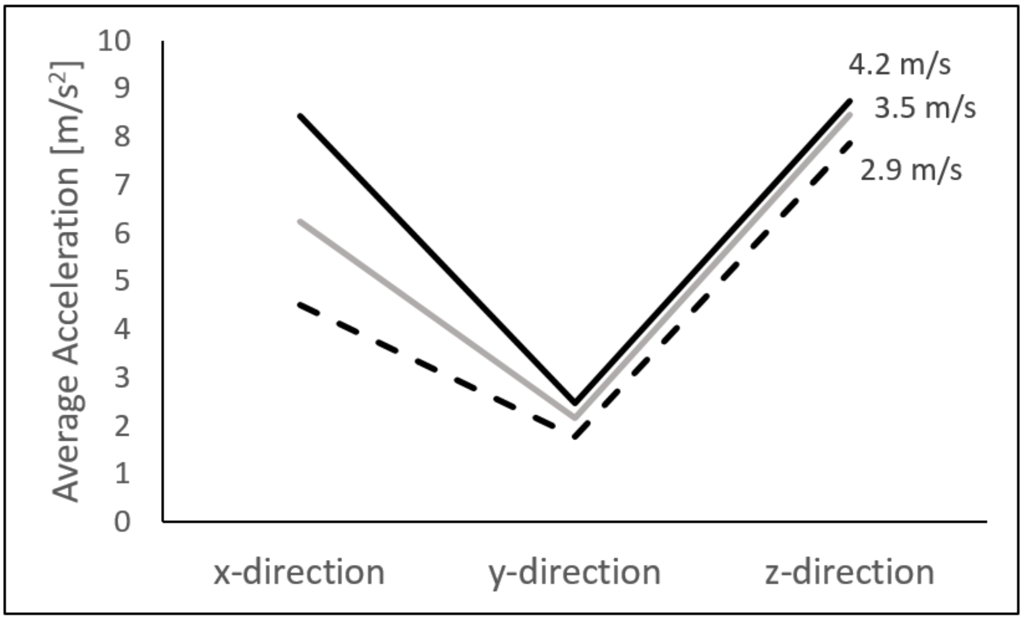

Figure 22: Interaction effect of direction*speed at average accelerations at the Achilles tendon, measured with the proximal accelerometer.

8.3.4.2 *Distal accelerometer - time space*

PEAK ACCELERATIONS Significant main effects were found for direction, $F(1.49, 28.23) = 9.36$, $p < 0.01$ and speed, $F(1.65, 31.33) = 114.08$, $p < 0.01$. Significant interaction effects were identified for direction*configuration, $F(2.52, 47.79) = 7.76$, $p < 0.01$ as well as for direction*speed, $F(2,54, 48.19) = 7.51$, $p < 0.01$ and for direction*configuration*speed, $F(5.91, 112.34) = 3.62$, $p < 0.01$. PeakAcc in cranio-caudal x-direction (50 ± 16 m/s^2) were 8 m/s^2 (95% confidence interval (CI) [0.63, 16.04]) higher than these in anterior-posterior z-direction (42 ± 10 m/s^2). A significant increase of 11 m/s^2 (95% CI [6.57, 15.67]) also occurred in medio-lateral y-direction (53 ± 12 m/s^2) compared to anterior-posterior z-direction. PeakAcc at the Achilles tendon increased continuously with increasing running speed. Accelerations measured at 4 m/s (56 ± 12 m/s^2) were significantly higher than these at 3.5 m/s (48 ± 13 m/s^2) and 2.9 m/s (41 ± 13 m/s^2). A difference of 7 m/s^2, 95% CI [5.29, 9.25], was also proven between peak accelerations when running at 2.9 m/s and 3.5 m/s. Figure 23 visualizes comparisons of all measurement conditions.

As stated above, an interaction effect of direction*configuration was found, meaning that configuration had not the same effect on peak accelerations at all levels of the factor *direction* as can be seen from figure 24. Large differences between the configurations were found especially in cranio-caudal x-direction and medio-lateral y-direction. While peak accelerations in cranio-caudal x-direction were found to be highest while running in Con2 (54 ± 15 m/s^2), they were lowest in this direction when running in NS (47 ± 15 m/s^2). A different order appears in medio-lateral y-direction, where highest accelerations were found in NS (56 ± 10 m/s^2) and lowest accelerations in Con1 (50 ± 13 m/s^2).

The interaction effect of direction*speed showed different effects of running speed on the measured accelerations, depending on the measurement direction of the accelerometer (figure 25). However, the effect is only based on a slight decrease in Achilles ten-

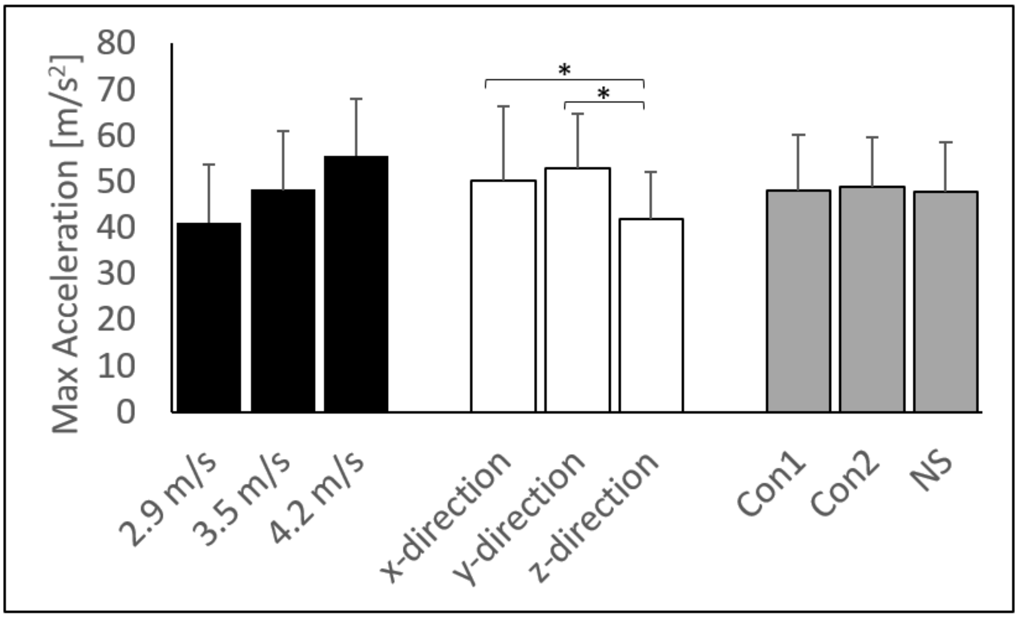

Figure 23: Peak accelerations detected with the distal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

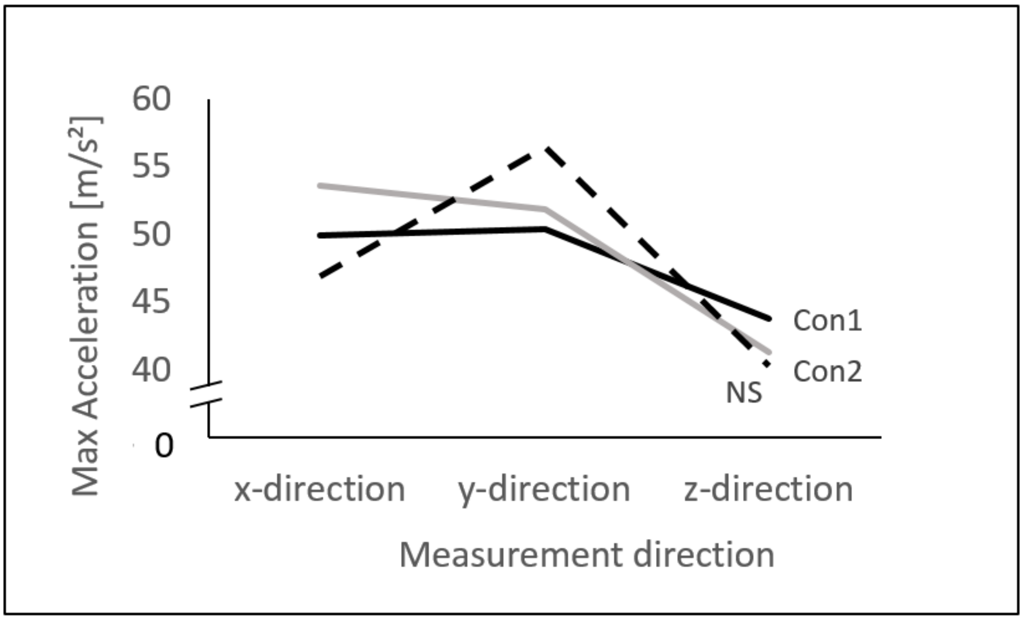

Figure 24: Interaction effect of direction*configuration on peak accelerations at the Achilles tendon, measured with the distal accelerometer.

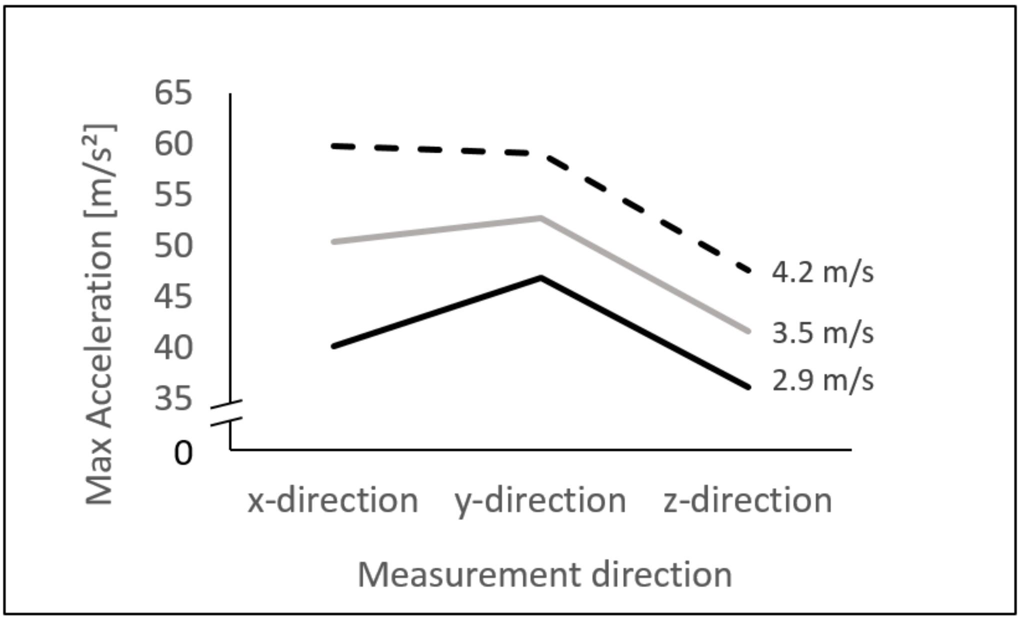

Figure 25: Interaction effect of direction*speed on peak accelerations at the Achilles tendon, measured with the distal accelerometer.

don accelerations at 4.2 m/s in cranio-caudal x-direction (60 ± 14 m/s^2) compared to those found in medio-lateral y-direction (59 ± 9 m/s^2) . During all other running speeds accelerations are higher in medio-lateral y-direction compared to cranio-caudal x-direction. All together the highest increases in acceleration with increasing running speed appears in cranio-caudal x-direction.

AVERAGE ACCELERATIONS Measurement direction and running speed had a significant main effect on AvgAcc detected at the distal sensor, $F(1.44, 27.26) = 75.15$, $p < 0.01$ and $F(1.79, 33.91) = 36.71$, $p < 0.01$. AvgAcc detected in medio-lateral y-direction (2.43 ± 2.03 m/s^2) were significantly lower than those in cranio-caudal x-direction (7 ± 2 m/s^2) or anterior-posterior z-direction (8 ± 5 m/s^2). Again, AvgAcc increased with increasing running speed. A comparison of all measurement conditions can be seen in figure 26 Similar to the results of PeakAcc, a significant main effect was found for Direction*Speed, $F(2.23, 42.41) = 9.94$, $p < 0.01$, which is shown in the figure 27.

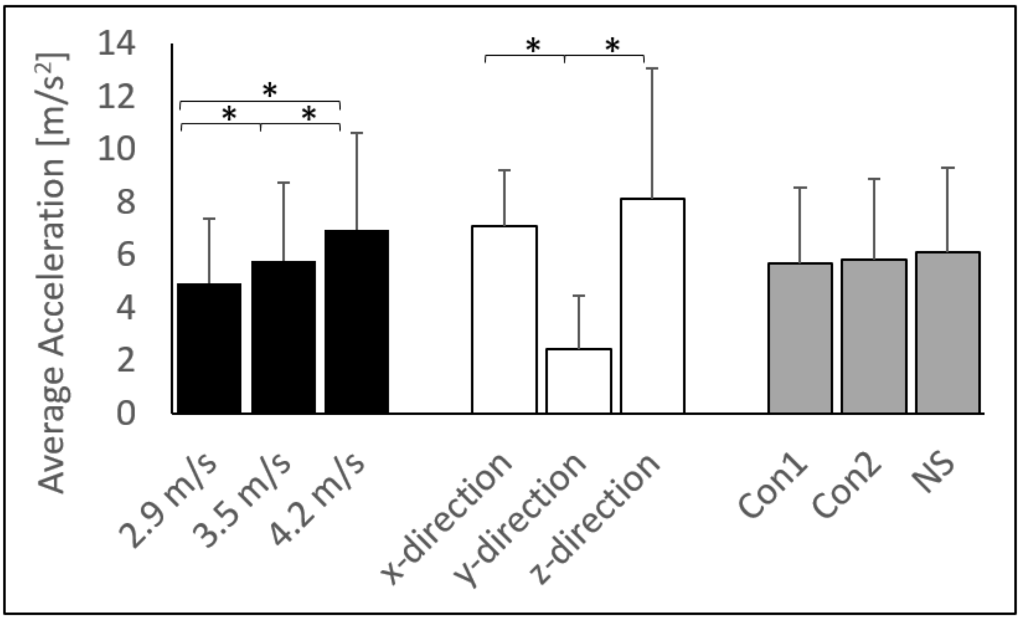

Figure 26: Average accelerations detected with the distal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

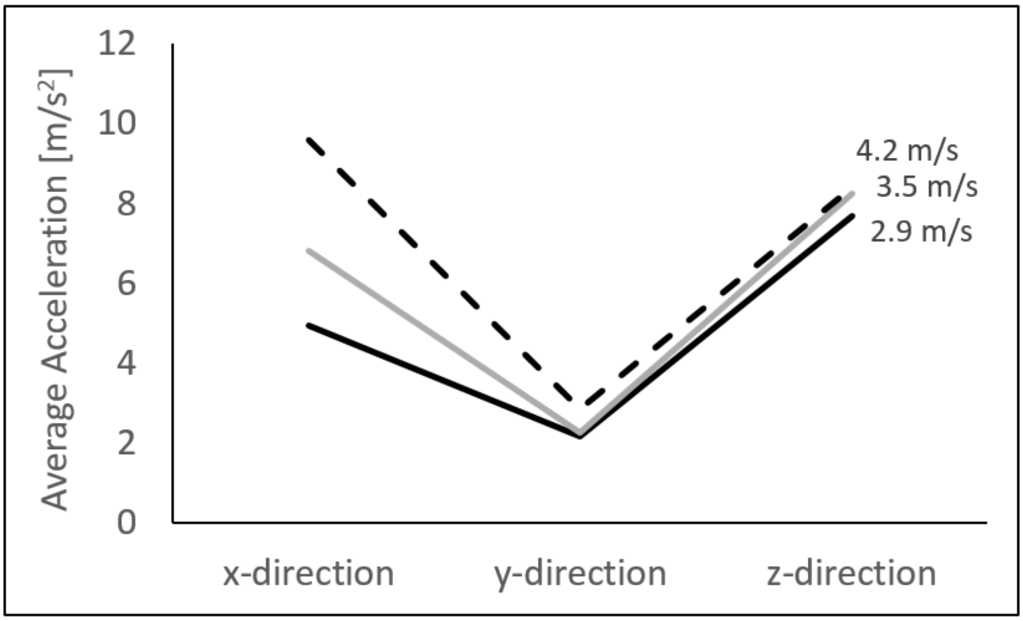

Figure 27: Interaction effect of direction*speed on average accelerations at the Achilles tendon, measured with the distal accelerometer.

8.3.4.3 *Proximal accelerometer - frequency space*

MAXIMUM OSCILLATION FREQUENCIES A significant main effect was found for direction, $F(1.5, 26.87) = 23.28$, $p < 0.01$, as well as for configuration, $F(1.66, 31.6) = 10.48$, $p < 0.01$, and running speed, $F(2, 38) = 65.87$, $p < 0.01$. No significant interaction effects were detected. An increase of 12 Hz, 95% CI [8.44, 16.02], in maximum oscillation frequency was found in in the anterior-posterior z-direction (53 $\pm$ 10 Hz) compared to the cranio-caudal x-direction (41 $\pm$ 9 Hz), and of 10 Hz, 95% CI [3.82, 16.71] in anterior-posterior z-direction compared to the medio-lateral y-direction (43 $\pm$ 14 Hz). No significant difference in oscillation frequencies was found between x- and y-direction. With regard to the different shoe configurations, an increase in frequency of 4 Hz, 95% CI [1.79, 6.4], appeared in NS (48 $\pm$ 13 Hz) compared to Con2 (44 $\pm$ 12 Hz; $p <$ 0.01) and of 5 Hz, 95% CI [1.03, 8.18], in NS compared to Con1 (44 $\pm$ 12 Hz; $p = 0.01$). No significant difference appeared between the two differently configured shoes. All three running speeds showed different Achilles tendon frequencies. At 3.5 m/s (46 $\pm$ 12 Hz) a 7 Hz, 95% CI [4.28, 8.79], higher frequency appeared than at 2.9 m/s (40 $\pm$ 10 Hz; $p < 0.01$). A further increase in frequency of 11 Hz, 95% CI [8.34, 14.13], could be proven at 4.2 m/s (51 $\pm$ 12 Hz) compared to 2.9 m/s ($p < 0.01$). A significant difference was also present between running trials at 3.5 m/s and 4.2 m/s ($p < 0.01$). A graphical comparison of all measurement conditions is shown in figure 28.

AVERAGE OSCILLATION FREQUENCIES Significant main effects were found for direction, $F(1.89, 35.94) = 137.68$, $p < 0.01$, and for speed, $F(2, 38) = 7.43$, $p < 0.01$, while a significant interaction effect could be detected for direction*speed, $F(4, 76) = 27.13$, $p < 0.01$. Mean oscillation frequencies differed in all three directions as can be seen from figure 29. An increase in oscillation frequency was found when running at 2.9 m/s (24 $\pm$ 4 Hz) compared to trials during which subjects ran at 4.2 m/s (23 $\pm$ 6 Hz; $p = 0.03$).

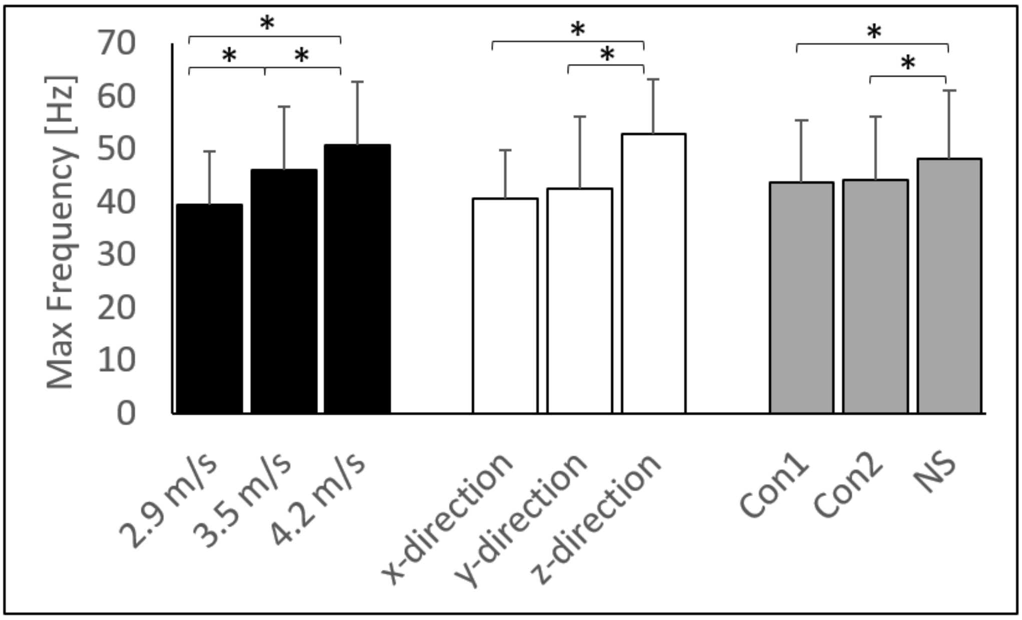

Figure 28: Maximum oscillation frequencies detected with the proximal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

The interaction effect of direction*speed is shown in figure 30. While oscillation frequencies during running at 4.2 m/s were lowest in cranio-lateral x-direction (19 $\pm$ 4 Hz respectively), they were among the highest in anterior-posterior z-direction (30 $\pm$ 3 Hz). Therefore frequencies at 4.2 m/s showed a much steeper increase from medio-lateral y-direction (21 $\pm$ 4 Hz) to anterior-posterior z-direction compared to trials at the two other running speeds.

8.3.4.4 *Distal accelerometer - frequency space*

MAXIMUM OSCILLATION FREQUENCIES Significant main effects were found for measurement direction, $F(1.7, 32.13) = 16.36$, $p < 0.01$, shoe configuration, $F(1.9, 35.25) = 8.11$, $p < 0.01$ and running speed, $F(2, 38) = 103.23$, $p < 0.01$. Max oscillation frequencies were higher in anterior-posterior z-direction (55.78 $\pm$ 11.07 Hz) compared to those measured in cranio-caudal x-direction (47 $\pm$ 11 Hz) and in medio-lateral y-direction (50 $\pm$ 11 Hz). Also, maximum oscillation frequencies were significantly higher when subjects ran in

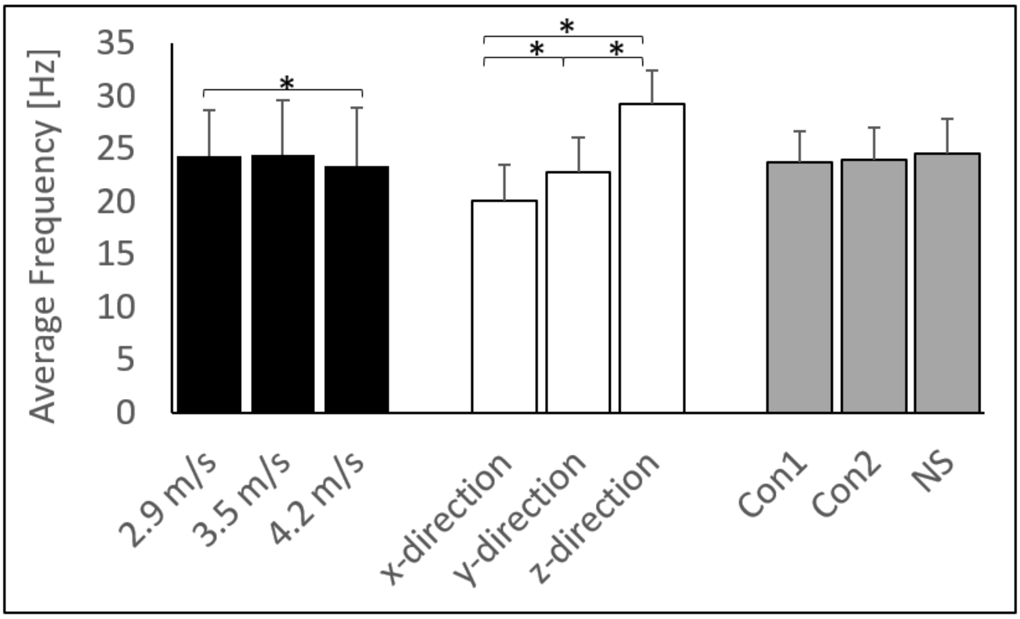

Figure 29: Average oscillation frequencies detected with the proximal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

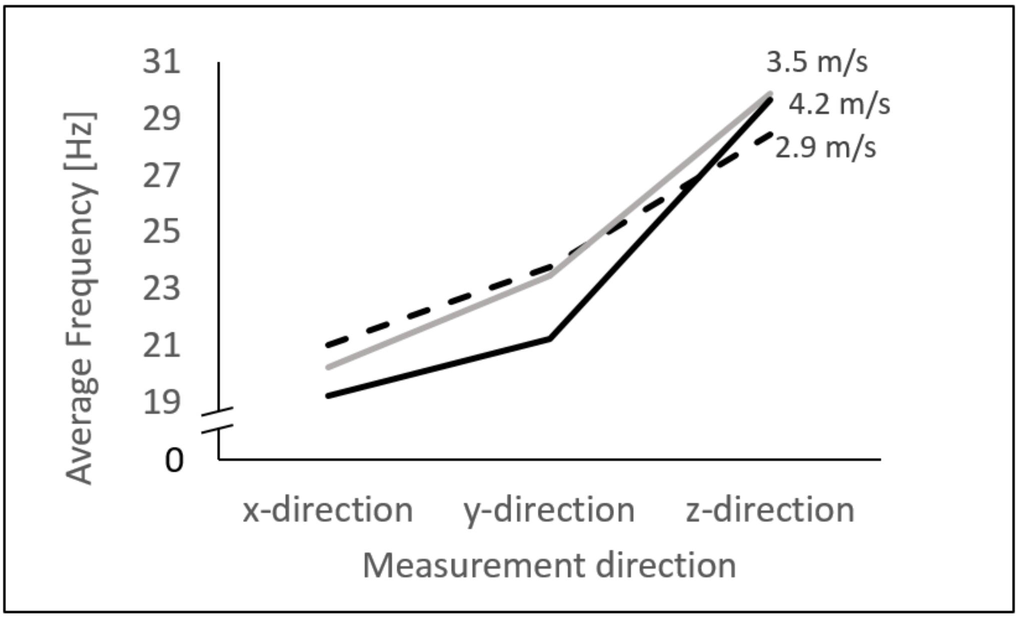

Figure 30: Interaction effect of direction*speed on average accelerations at the Achilles tendon, measured with the proximal accelerometer.

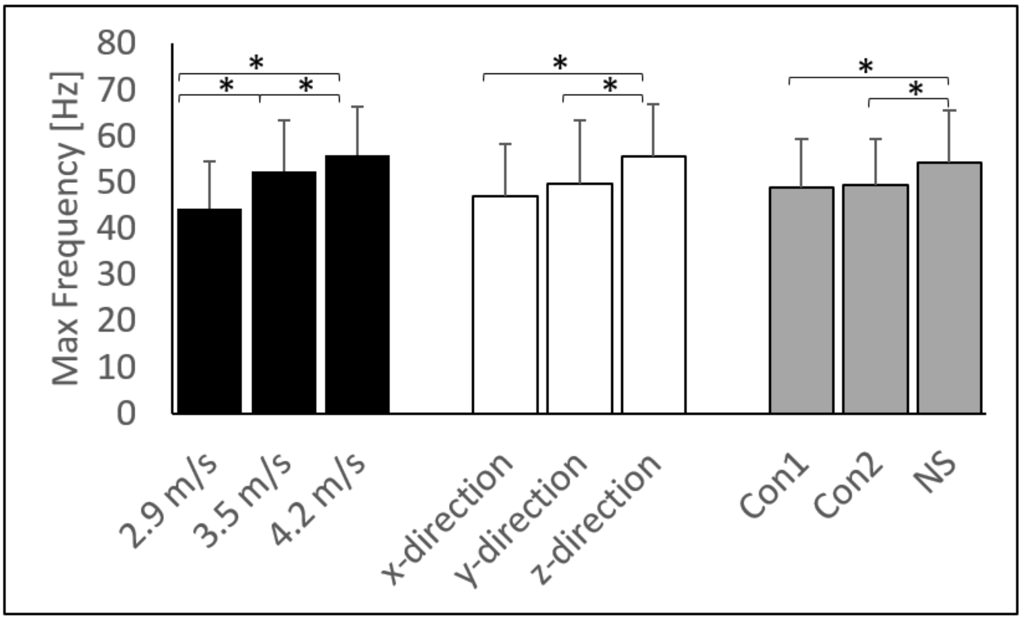

Figure 31: Maximum oscillation frequencies detected with the distal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

NS (54 ± 11 Hz) compared to the two other running shoe configurations (Con1: 49 ± 11 Hz; Con2: 49 ± 10 Hz). Maximum oscillation frequencies increased with increasing running speed as shown in figure 31.

AVERAGE OSCILLATION FREQUENCIES Significant main effects were found for measurement direction, $F(1.26, 23.84) = 136.15$, $p < 0.01$, and for shoe configuration, $F(2, 38) = 3.84$, $p = 0.03$. Significant interaction effects were be detected for direction*configuration, $F(3.9, 73.73) = 3.84$, $p = 0.01$, as well as for direction*speed, $F(3.9, 74.06) = 24.9$, $p < 0.01$. All three measurement directions of the accelerometers showed different results. Oscillation frequencies in anterior-posterior z-direction (31 ± 3 Hz) were significantly higher than these measured in cranio-caudal x- direction (24 ± 3 Hz) and in medio-lateral y-direction (25 ± 4 Hz; $p < 0.01$ and $p = 0.01$ respectively). A significant difference also exists between frequencies in medio-lateral y-direction and in anterior-posterior z-direction ($p <$

0.01) as can be seen in figure 32. A significant increase in oscillation frequency measured with the distal accelerometer was found in NS (27 ± 5 Hz) compared to Con1 (26 ± 4 Hz; p = 0.02). However, no difference could be detected between NS and Con 2 (26 ± 5 Hz; p = 0.23) or the two configurations of the modular shoe system (p = 1; figure 32).

The significant interaction effect of direction*configuration is visualized in figure 33. In cranio-caudal x-direction Con1 (23 ± 3 Hz) and Con2 (23 ± 3 Hz) show identical results but different from NS (24 ± 3 Hz) while in medio-lateral y-direction comparable frequencies were found in all shoes (24 ± 3 Hz; 24 ± 3 Hz; 25 ± 4 Hz respectively). However, different oscillation frequencies appear in all three shoes in anterior-posterior z-direction (Con1: 30 ± 3 Hz, Con2: 31 ± 3 Hz, NS: 32 ± 3 Hz). The significant interaction effect of direction*speed is visualized in figure 34. Highest mean frequencies are found in anterior-posterior z-direction for all running speeds. While the lowest values are found in cranio-caudal x-direction and in y-direction at 4.2 m/s, they show to be among the highest frequencies at this running speed in anterior-posterior z-direction.

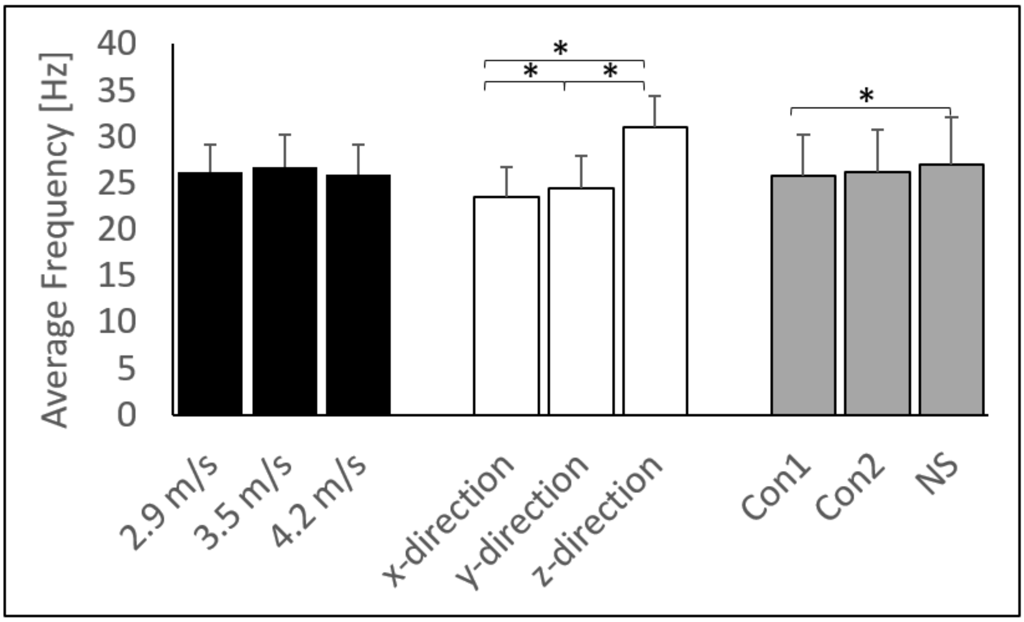

Figure 32: Average oscillation frequencies detected with the distal accelerometer at different measurement conditions. Significant differences between conditions are indicated by a star.

8.4 discussion

The purpose of the present study was to test the effects of different footwear modifications and different running speeds on plantar pressure data, muscle activity, foot kinematics and oscillations at the Achilles tendon. The analysis of plantar pressure gives insight into the input signal which is then forwarded to the system. Plantar force peaks increased with increasing running speed, which is in agreement with the hypothesis as well as with previous findings (Rosenbaum, Hautmann, Gold & Claes, 1994). An effect of footwear variations could only be proven for peak forces, which were found to be higher while running in NS compared to Con1. The two shoes represent the most diverse types of footwear used in the present study. While NS shows much higher values in the damping characteristics (see table 4), Con1 represents a typical motion control shoe with soft damping material, medial wedges and arch support. Therefore, increased plantar force peaks are generated by less

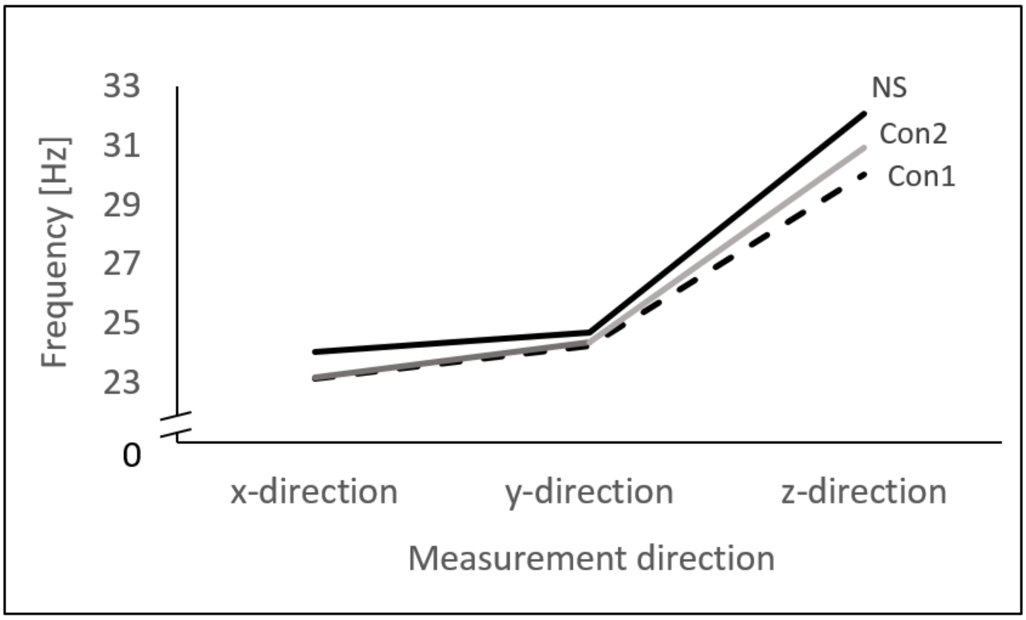

Figure 33: Interaction effect of direction*configuration on average accelerations at the Achilles tendon, measured with the distal accelerometer.

damped shoes. Our findings are in agreement with those of Schuh, Trnka, Sabo, Reichel & Kristen (2011) who found decreased force peaks when using insoles made from soft material compared to those made from stiff material. O'Leary, Vorpahl & Heiderscheit (2008) observed the effects of cushioned insoles on ground reaction forces in runners at a self-selected speed. These researchers confirmed the outcome of decreased peak forces if additional cushioning is used but also found lower average ground reaction forces. The latter is in contrast to the results of the present study. A more recent study by Baltich, Maurer & Nigg (2015) came to a controversial conclusion, when the researchers found increased impact forces with softer midsole shoes. The findings of the present study follow neither of these outcomes as no difference in total force was found between shoe configurations. A comparison of the results of these studies may be subjected to error though, as only the present study provides precise information of the damping characteristics of the shoes used. An overall description of hard or soft cushioning material is not sufficient and does not allow implica-

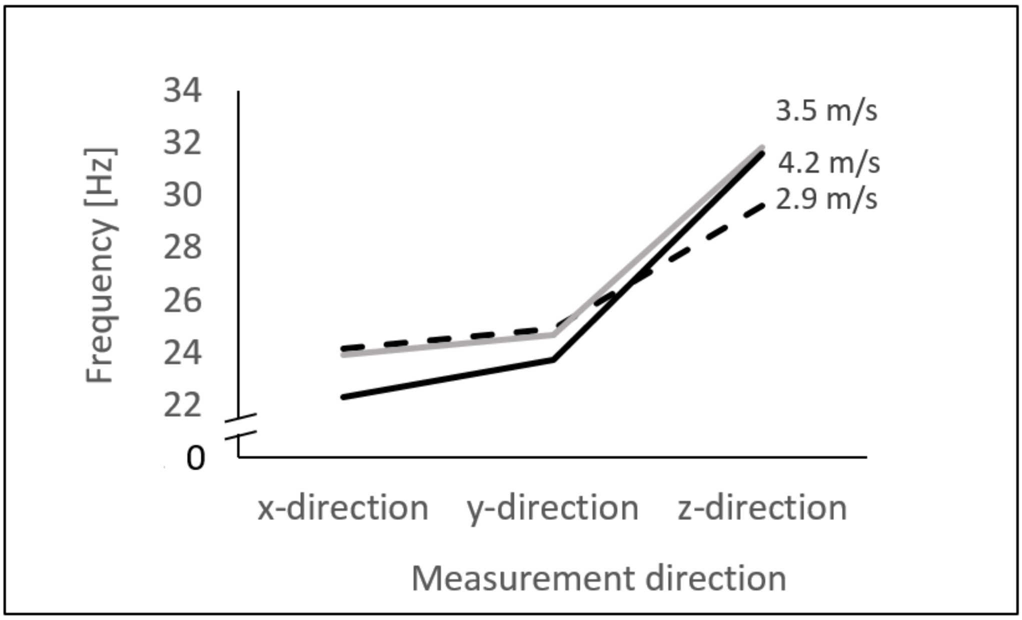

Figure 34: Interaction effect of direction*speed on average accelerations at the Achilles tendon, measured with the distal accelerometer.

tions to other shoes. A precise testing procedure like the ASTM F-1976 should become standard in reporting experimental results. The shoes used by O'Leary et al. (2008) may have had a different range of the cushioning properties compared to those investigated by Baltich et al. (2015). The footwear which was worn by runners in the present study may have been in between those used by the two afore mentioned research groups, therefore resulting in no significant differences in total force. However, this explanation remains hypothetical as no specific quantitative measure of the damping material is given in the studies of O'Leary et al. (2008) and Baltich et al. (2015).

In agreement with the expectations, muscle activity increased with increasing running speed. During stance phase, an important function of the lower leg muscles is to increase joint stiffness to lead to a robust system (Reeves, Narendra & Cholewicki, 2007). Total force as well as force peaks were shown to increase with running speed. As these parameters indicate a more intense input signal to the system, increased muscle activity represents the appropriate reaction during the landing phase. Schache & Dorn (2014) examined a variety of muscle groups at different running speeds and found M. Soleus and Mm. Gastrocnemii to contribute to a large portion of the ground reaction force during push off, therefore increasing with running speed. The effect of different types of footwear on a variety of muscles has been investigated in the past (Murley & Landorf, 2013). Several researchers found no effect of shoe variations on muscle activity (Boyer & Nigg, 2004; Komi, Gollhofer, Schmidtbleicher & Frick, 1987; Nigg et al., 2003; O'Connor & Hamill, 2004; Roy & Stefanyshyn, 2006) while other studies state an effect of shoe modifications on the activation of specific muscles. O'Connor, Price & Hamill (2006) found a decrease in EMG activity of the M. Tib. Ant. with a neutral shoe compared to a shoe with medial wedges. Wakeling et al. (2002) described a significantly altered total EMG intensity with different midsole hardnesses for M. Gastr. Med. and M. Tib. Ant. Baur, Hirschmüller, Müller & Mayer (2003) found an increase in EMG activity of the M. Peroneus long. when wearing shoes with arch support. In the present study, we found average muscle activity of the M. Gastr. Med. to differ when running in NS compared to Con1 while non of the other investigated muscles showed an effect of footwear modifications. Higher activation levels of the M. Gastr. Med. were found when subjects wore Con1, a shoe with medial wedges, soft damping material and arch support. The only difference in kinematics (MaxProVel) was also found between these two types of footwear. However, while MaxProVel occurs during the first half of stance phase, the main function of M. Gastr. Med. is to push the runner off ground during the second half of stance phase. Although these two findings may not be linked directly, they show that variations in footwear may lead to both, changes in kinematics and in muscle activity.

The ROM of the foot in the frontal plane found in the present study is comparable to previous findings (Peltz, Haladik, Hoffman, McDonald, Ramo, Divine, Nurse & Bey, 2014). It should be noted however, that in the present study foot motions were not measured relative to the tibia but a quantification of the roll dynamics of the foot was performed. This methodological difference may explain the slightly lower ROM found in the present study. If motions are determined relative to the tibia, increased ROM may be detected due to movements of the lower leg during the stance phase with the foot itself being stationary.

Reduced ROM was found when subjects wore Con1 or Con2 compared to NS. These findings show that shoes with higher G-scores and therefore harder damping material resulted in increased ROM of the foot during stance phase. Together with knee flexion, pronation is known to function as a damper for impact forces acting at the lower extremity. Shoes with higher G-scores will forward higher impact loading to the body. Therefore, increased pronation-supination ROM may display a technique to limit impact forces to a bearable, non-harmful amount. This finding goes along with the paradigm proposed by Nigg (2001) who describes impact forces as input signals which stimulate muscle tuning to reduce joint and tendon loading. However, as can be seen from the large variation found in our data, the response to changes in footwear remains a highly subject specific issue (see also figure 17).

Shoes with medial wedges only had a limited influence on ROM in the frontal plane as no significant differences were found between trials during which the runners wore Con1 and those during which they wore Con2. This is in contrast to the results of Cheung et al. (2011) who found shoes with medial wedges to be more effective in controlling foot pronation than those with dual midsole material. Other authors who also proofed an effect of medially posted insoles on foot kinematics in the frontal plane quantified motions around the subtalar joint by using a photogrammetric method. Rodrigues et al. (2013) affixed their markers directly to the foot and lower leg while in the present study kinematics were recorded with an IMU attached to the outside of the runners' shoes. Therefore, relative movement of the foot inside the shoe cannot be ruled out and may not have been detected. The actual ROM of the foot may therefore be larger than the values attained during measurements at the outer shoe.

A significant decrease in MaxProVel could be proven when running in Con1 compared to NS, which is in agreement with the findings of Brown, Donatelli & Catlin (1995) who stated MaxProVel to be lower while walking with arch supports (119.9 $\pm$ 40.6) compared to walking with shoes only (138.7 $\pm$ 50.2). Although these authors used a photogrammetric measurement system with markers attached directly to the subjects' feet, MaxProVel obtained in the present study is highly comparable to those values. It should be noted though, that we found a dependency of MaxProVel on running velocity. Therefore MaxProVel is expected to be lower during walking compared to running. The comparability of these results may have been caused by the neglected relative motion of the foot inside the shoe as described earlier.

In agreement with our hypothesis, accelerated running speed led to increases in ROM and MaxProVel. The simultaneous gain in these parameters is in agreement with the findings of Shih, Ho & Shiang (2014). Running speed therefore not only influences joint loading (de David et al., 2014) but also joint kinematics as shown for motions of the foot in the frontal plane. While we found significant differences in the kinematic variables between trials at 2.9 m/s and 4.2 m/s others found torques and power at the ankle to increase when running at 3.5 m/s compared to 5.02 m/s (Schache et al., 2011). The unequal running speeds between which differences were found may have been generated by incomparable measurement conditions. Schache et al. (2011) performed their measurements with subjects running over ground while in the present study, subjects ran on a treadmill.

No interaction effect of speed*configuration was found in the present study. Therefore running speed did not influence the effects of shoe modifications on foot kinematics. Findings from previous studies on shoe modifications which were conducted at different running speeds may consequently be compared. However, it remains to be explored whether this also applies for studies at varying fatigue levels or the comparison of studies performed on the treadmill and those performed over ground.

To detect stance phases an in-shoe plantar pressure system was used with a sampling frequency of 100 Hz. At higher running speeds this sampling frequency may have been too low to flawlessly distinguish initial and final contact of the foot with the ground. Therefore the calculations of TFPro may have been inaccurate as they are based on the total contact time of the foot with the ground. Interpretations of these results should therefore be made with caution.

Motions of the foot were analyzed in the frontal plane using a single IMU, which was found to be a useful device when estimating foot kinematics during running. Whether these findings can be compared to data from photogrammetric measurement systems or results obtained from models accounting for motions of the foot segment relative the tibia remains to be analyzed in more detail. Also, relative motions of the foot inside the shoe take place during running. It should therefore be considered that movements of the foot might be different than the data obtained on the outside of the shoe.

Running speed has been discussed as an injury inducing factor in the past (McCrory et al., 1999). The current study showed that running speed has an effect on accelerations and oscillation frequencies at the Achilles tendon. Running speed alone as well as in interaction with measurement direction is a factor which leads to changes in the vibration behavior at the tendon. As mentioned above, plantar pressure as well as EMG activity increased when running at higher speeds. Therefore, the input signal sent to the system (quantified via plantar pressure) was varied as well as the muscular strains which are evoked through contractions of the calf muscles. Even though the muscular system may be described as a useful instrument to control soft tissue oscillations (Wakeling & Nigg, 2001; Boyer & Nigg, 2004), an increase in max frequency as well as in accelerations measured at the Achilles tendon occurred in the present study. Mercer et al. (2002) studied the effect of running speed on shock attenuation within the runner's body. The authors quantified accelerations at the leg and the forehead of their subjects and found an increase in shock attenuation with increasing running speed. While shock attenuation seems to occur throughout the entire body, the Achilles tendon is located distally and spans a relatively short distance compared to a whole body. It may therefore not experience a highly effective damping of vibrations, resulting in differences between running speeds. Despite the fact that the oscillation characteristics varied with running speed, they may not have exceeded an acceptable, non-harmful range. Therefore the changes in muscle control may have been sufficient to allow for an economic running style and keep the tendon from damaging oscillations. The attenuation of vibrations along the Achilles tendon should be analyzed in more detail to clarify the signal damping in proximal direction. Hence, a quantification of the signal power attenuation along the Achilles tendon was performed in the following test series (chapter 9.

Shoe modifications are frequently recommended for the treatment of Achilles tendon complaints, such as proper shock absorption or specific sole structures (Hess, Cappiello & Hunter, 1989; Sandmeier & Renström, 1997). Other studies proved the influence of soft tissue vibrations at the thigh by using different types of footwear during drop jumps (Fu, Liu & Zhang, 2013). Since different shoes influenced oscillation frequencies in the current study, the prevention or the treatment of Achilles tendon injuries may, in the future, be assisted by appropriate footwear. No differences were found between the two configurations of the modular running shoe but only between NS and either of the two configurations (Con1 and Con2). Therefore, using a modular running shoe system may not allow sufficient variations in the oscillation behavior to cause a significant difference. It remains to be investigated whether the system's modifications are adequate to influence Achilles tendon vibrations on an individual level to prevent or treat complaints at this structure. In order to further investigate effects of shoe modifications on general samples, the variations of the modular running shoe system were reinforced in the test series described in chapter 9.

LIMITATIONS Plantar pressure distribution was recorded at a sampling rate of 100 Hz while all other data were collected at 3000 Hz. To define gait events (heel contact and toe off) of a step, the particular frame numbers of the pressure data had to be multiplied by 30 to correspond to the frame numbers of EMG, kinematics and oscillation data. It is, however, uncertain where within those 30 frames the real event took place. Therefore, step detection took place with an uncertainty of 30 Hz or 10 ms. In the current study, oscillation frequencies were determined by counting the number of changes in direction of the acceleration signal per second. While this approach does, in theory, represent the frequency of the acceleration signal, a more precise analysis method may be the performance of a frequency analysis. Therefore, an improvement in the analysis took place in the second test series described in this thesis by using a Fourier transformation to analyze the data. While standardized procedures were used to prepare the skin and attach EMG electrodes, sudor is known to affect the conductibility of the electrodes. Some subjects showed major perspiration, especially while running at higher speeds. This may have affected the recorded data and untruly pronounced the differences found in muscle activity between running speeds. A general limitation of the quantification of foot kinematics using a single IMU is the absence of a second segment around which rotations are performed. For the calculations presented here, a stable tibia segment is assumed. However, if runners performed motions of the tibia relative to the foot, these trajectories could not be accounted for. Also, rearfoot motion in the frontal plane occurs around the subtalar axis. The location and orientation of this axis may vary considerably but were not incorporated in our kinematic analysis. An inclusion of these specifications into the kinematic calculations would result in a model, which considers not only global foot motions but also motions around anatomical axes. The measurement of oscillations at the Achilles tendon comprises skin movement artifacts. These are well known from marker based kinematic analysis and may vary depending on subcutaneous fatty tissue or conflicting tendons and muscular pulls. A large advantage of the Achilles tendon is its location directly underneath the skin. Therefore, skin movement artifacts should mainly be of concern in the vertical movement directions as the tendon may slide underneath the skin without the possibility of detection through skin mounted accelerometers.

CONCLUSION In conclusion, first insight was gained in plantar pressure data, muscle activity, foot kinematics and the oscillation behavior at the Achilles tendon while running at different speeds and in different types of footwear. The IMU which was used as a measurement tool and the algorithm described for data analysis proofed to be a useful method when assessing foot kinematics during running. The modular running shoe system may provide information on how individuals react to shoe modifications as differences in footwear led to significant changes in kinematic variables, peak forces underneath the foot, vibrations at the Achilles tendon and EMG activity. Also, running speed had an influence on the measured data. Future studies should therefore consider these factors as possibly influencing the mechanisms which may lead to overuse injuries. Long distance runners most commonly perform their sport running over ground, not on a treadmill. It remains unknown whether the results obtained during treadmill running can be transferred to over ground running. Further research is required to clarify whether the findings of this study are applicable for running on asphalt roads, running tracks and forest tracks.

9

EFFECT OF GROUND CONDITIONS AND SHOE MODIFICATIONS ON THE COLLECTED DATA

Table 6: Details of study 2, comparing shoe modifications while running at different ground conditions (treadmill vs. over ground)

Details of the Study	
N	20 ♂
Age [years]	20.7 ± 2.9
Height [meters]	1.83 ± 0.06
Weight [kg]	79.9 ± 7.5
Weekly running distance [km]	20.4 ± 8.1
Running experience [years]	5.4 ± 3.2
Measurement systems	3D IMU
	Two 3D accelerometers
	F-Scan plantar pressure system
	EMG
Conditions	Three types of footwear
	(Con1, Con2, NS)
	Three running speeds
	(2.9 m/s, 3.5 m/s, 4.2 m/s)

9.1 INTRODUCTION

In the previous study (chapter 8), differences in the investigated biomechanical parameters were found between types of footwear and between running speeds. While these results were obtained from trials on a treadmill, long distance runners most commonly perform their sport running over ground. Even though treadmills are not used as frequently for training, they provide means for researchers and clinicians to investigate athletes or patients in a controlled setting. However, several discrepancies were found between treadmill running and running over ground. Authors comparing the two types of locomotion found structural differences in the variability of stride timing (Lindsay, Noakes & McGregor, 2014), deviations in plantar pressure (Hong, Wang & Zhou, 2012), ground contact time (McKenna & Riches, 2007), overall running performance, pacing strategy and even thermoregulation (Heesch & Slivka, 2015) as well as better running economy when running over ground than on a treadmill (Mooses, Tippi, Mooses, Durussel & Mäestu, 2015). However, other researchers concluded kinematic and kinetic parameters to be comparable, even if not fully equivalent (Riley, Dicharry, Franz, Croce, Wilder & Kerrigan, 2008) and found no significant differences in stride frequency, step length, support time and flight time (Frishberg, 1983). In a literature review, Williams (1985) summarized that significant differences between running over ground or at a treadmill only occur at running speeds above 5 m/s. It remains unknown whether the results obtained during treadmill running in the previous study (chapter 8) can be transferred to over ground running. If deviations between the two ground conditions exist, quantification of those differences needs to be specified.

Impact forces occur during running at every step with the foot hitting the ground. These forces are then transferred along the lower leg in longitudinal direction and cause oscillations of soft tissues surrounding bony structures, including muscles and tendons (Nokes, Fairclough & Mintowt-Czyz, 1984). As shown in chapter 8, as well as by other researchers, the magnitude of impact forces can be modified by changes in running speed (Nigg, Bahlsen, Luethi & Stokes, 1987) and through the use of different running shoes (Logan, Hunter, Hopkins, Feland & Parcell, 2010). Shock absorbing materials are used in running shoes, aiming to limit these forces (Jorgensen & Ekstrand, 1988). Differences in loading rate and impact magnitude also occur when running over ground compared to running on a treadmill (Hong et al., 2012; Ki-Kwang, Lafortune & Valiant, 2005). Therefore every running shoe may lead to specific input characteristics to the system when running at different running speeds as well as on different grounds.

Due to its location at the distal end of the lower leg, the Achilles tendon is strongly affected by impact forces. Loading of this tendon can reach up to 12.5 times body weight during running as described by Komi (1990); Komi, Fukashiro & Järvinen (1992). However, these values should be considered with care. The researchers measured in vivo Achilles tendon loadings with implanted transducers. The results may therefore be influenced by motions of the transducer in the synovial sheath of the tendon as well as by forced applied through contact with ambient structures. Elite male distance runners have a 52% risk of Achilles tendinopathy (Zafar, Mahmood & Maffulli, 2009). According to Lohrer (2006), overuse injuries of the Achilles tendon are the most common reason for dropouts in athletic careers, especially in track and field. Impact forces and loading rates are often suggested to affect injury rates at the lower extremity (Zadpoor & Nikooyan, 2011; Gerlach, White, Burton, Dorn, Leddy & Horvath, 2005). The influence of these forces and the resulting oscillations of soft tissue compartments were recently studied (Boyer & Nigg, 2004, 2006b; Enders, von Tscharner & Nigg, 2014). However, their effects on sports-related injuries seem to be poorly understood. In the past, high impact forces were thought to cause harm to the musculoskeletal system. Newer findings suggest positive effects of undamped impacts, serving as input signals to the system. In mechanical components, resonant oscillations are expected to destabilize a system, possibly resulting in catastrophic damage. In biological systems, however, the mechanical properties of soft tissues (viscoelastic component), including the Achilles tendon, may be altered through neuromuscular adaptations (Boyer & Nigg, 2007; Wakeling & Nigg, 2001). Even though specific resonance frequencies exist for tendons, they can be altered through changes in joint angles and muscular contractions (Wang, Hsiao, Wang & Shau, 2007). Kinematics as well as muscular activity of the lower extremity differ substantially when running on a treadmill compared to running over ground (Nigg, De Boer & Fisher, 1995; Wang, Hong & Xian Li, 2014). These differences in running technique may also lead to alterations in the vibration behavior at the Achilles tendon. To further explore the effects of these parameters, a more detailed analysis method is used in this study compared to the study described in chapter 8, including a more detailed evaluation in frequency space.

The input signal to the Achilles tendon results from ground reaction forces which may be altered through changes in running shoes. Shoe modifications lead to changes in the input frequency during walking (Wakeling et al., 2003). Also, changes in transmissibility of soft tissue vibrations were found during drop jumps when wearing different shoes (Fu et al., 2013). Therefore, alterations in cushioning and stabilizing components of the shoe as well as alterations of ground conditions were expected to result in changes of the analyzed biomechanical parameters. During treadmill running, an decrease in plantar pressure was expected due to the damping characteristics of the treadmill compared to concrete floor. Variations in ground reaction forces are known to have an influence on muscle activation in the lower leg (Wakeling & Nigg, 2001). Therefore, differences in EMG activity were hypothesized to occur when running on a treadmill compared to over ground. It is known from the literature that kinematics change, depending on ground condition (McKenna & Riches, 2007) and that pronation movements are used as damping mechanism to reduce joint loading (Hintermann, 1998; Nigg, 2001). As we expected to find lower plantar pressure values when running on a treadmill, decreased ROM was hypothesized for this ground condition. The line of arguments outlined above also leads to the expectation of changes in the vibration behavior at the Achilles tendon between running conditions. Therefore, the purpose of this study was to explore the effects of different running shoes and different surface conditions (treadmill versus over ground) on plantar pressure, muscle activity, foot kinematics and oscillations at the Achilles tendon during running.

9.2 METHODS

Twenty male rearfoot runners participated in this study. Subject characteristics are presented in table 6. Part of the measurement performed in this study are identical to those described in chapter 8 and will therefore not be specified in detail. Subjects were given a six minute warm up period at a self-selected running speed to get accustomed to the treadmill (Woodway ERGO XELG 90®, Woodway USA Inc., Waukesha, WI, USA). Subsequently, study participants were asked to run at a running speed of 2.9 m/s on the treadmill and over ground while wearing two differently configured running shoes as well as one neutral all-purpose shoe as a reference (NS; Adidas Gazelle®, Adidas, Herzogenaurach, Germany). The order in which the shoes were worn as well as the ground conditions were randomized. A modular running shoe (Runaissance 3.0®, Newline, Vodskov, Denmark) was used to provide two different running shoe configurations. This system is similar to the one used in study 1 (see 8.2). Two different configurations of this shoe were set up: one with high arch support, medial wedges (4 mm), heel wedges (2cm) and soft damping material (Con1), the other with low arch support, no medial wedges and hard damping material (Con2). Therefore, Con1 was further modified compared to the study described in chapter 8 by adding heel wedges.

Subjects were given a 1 minute familiarization period while running in each shoe condition, followed by a 10 second measurement period. Over ground running was performed on a 30 meter runway on concrete floor. Trials were accepted if the time needed to cover this distance was between 9 sec (3.3 m/s) and 11 sec (2.7 m/s).

Step detection took place using an in shoe plantar pressure system (F-Scan®, Tekscan Inc., South Boston, MA, USA). Data included in further analysis were limited to stance phases of the right foot. Plantar pressure data were analyzed in an analogous matter as in study 1 (see 8.2) and included the identification of total force and peak forces during stance phase. Identical to study 1 (chapter 8.2), an IMU was attached to the right heel cap of the subjects' shoes to allow acquisition of kinematic data of the foot. Muscle activity was again captured of M. Gastr. med., M. Gastr. lat., M. Tib. ant. and M. Peroneus long. In addition to the analysis of average activity during TCT, an evaluation of average activity during pre-activation (PA) was performed.

Two tri-axial accelerometers (Noraxon Corporate, Scottsdale, AZ, USA) were attached to the skin overlaying the right Achilles tendon using double sided tape and kinesio tape. Locations of accelerometer placement were identical to study 1. Data of each accelerometer were analyzed individually and separately in the vertical x-direction as well as in the yz-plane (resulting vector). Absolute peak accelerations and the time to peak acceleration following first foot contact (ffc) were determined as dependent variables. Data were high pass filtered with a cut off frequency of 10 Hz as frequencies below 10 Hz contain movement artifacts (Boyer & Nigg, 2004). Spectral analysis of the acceleration data was performed according to Boyer & Nigg (2006b). In short, the data were zero-padded and transformed to the frequency domain using FFT (Matlab R2013b®, Mathworks, Natick, MA, USA). Average signal power was calculated for each accelerometer and in each measurement direction (vertical x -direction and horizontal yz-plane resultant). Power was then normalized to stance phase and normalized by peak power from both accelerometers of each trial. Normalized power will further be denoted as P(f) and was analyzed in a low-frequency interval (10 - 25 Hz), medium-frequency interval (25 - 50 Hz), high-frequency interval (50 - 100 Hz) and a highest-frequency interval (> 100 Hz). The dominant frequency in each trial was determined as the highest peak in each power spectrum and was used as dependent variable.

Transfer functions resemble mathematical relations between input and output signals of a dynamic system in the frequency domain and are a frequently used engineering term in systems theory. In the present study the transfer function H(f) resembles the ratio between normalized power of proximal and distal accelerometer signals. It quantifies the power attenuation between the two accelerometers.

$$H(f) \quad = \quad 10log[\frac{P(f)_{proximal}}{P(f)_{distal}}] \tag{8}$$

Acceleration data at the Achilles tendon, EMG and IMU data were sampled with a frequency of 1500 Hz using Noraxon Telemyo 2400 G2 (Noraxon Corporate, Scottsdale, AZ, USA) while pressure data were sampled at 100 Hz.

Repeated measures ANOVAs were performed to analyze differences in the dependent variables between running shoes, ground conditions, accelerometer locations, and frequency intervals. The assumption of normality was checked and the data were found to meet the criteria. Violations of sphericity were controlled using Maulchy's test of sphericity and either Greenhouse-Geisser or Huynh-Feldt corrections were applied according to Girden (1992). If a Greenhouse-Geisser epsilon of > 0.75 was found, the Huynth-Feldt corrected value was used for that parameter. Otherwise the Greenhouse-Geisser corrected value was used. Post hoc tests were performed using modified t-tests with Bonferroni correction. All statistical calculations were completed using SPSS (SPSS 21®, IBM, Armonk, NY, USA) and the alpha-level was set at 0.05.

9.3 RESULTS

9.3.1 *Plantar Pressure Distribution*

TOTAL FORCE Non of the investigated parameters had a main effect on total force and no interaction effect could be proven.

PEAK FORCES A significant interaction effect of Configuration *GroundCondition could be proven, $F(1.99, 37.79) = 4.53$, $p = 0.2$. When wearing Con1 or NS, average force peaks were higher while running on a treadmill compared to over ground running whereas they were lower during treadmill running when wearing Con2 (figure 35).

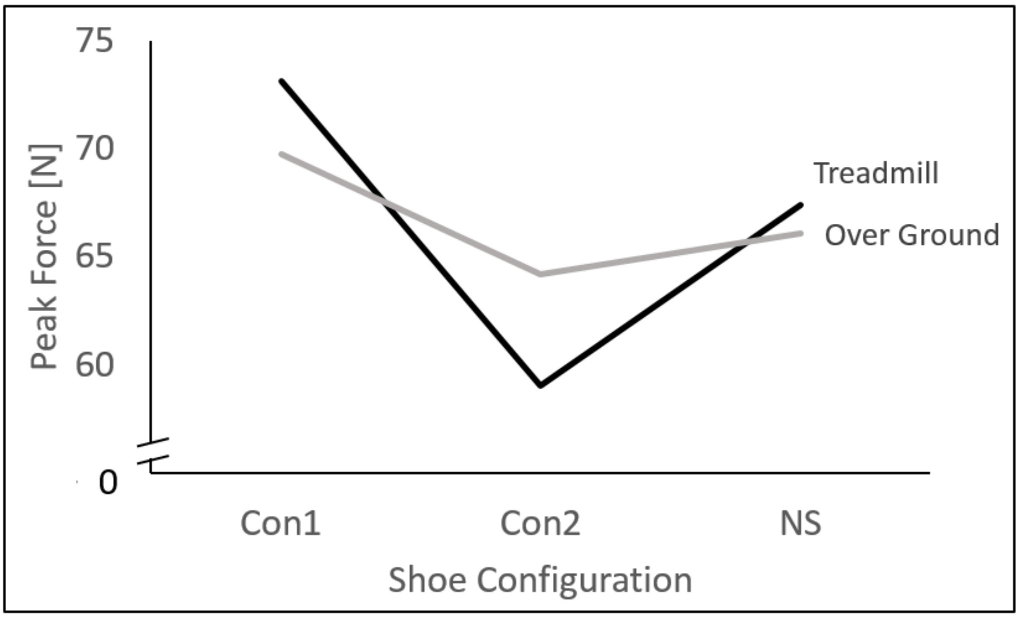

Figure 35: Interaction effect of Configuration*GroundCondition on average peak forces underneath the foot.

9.3.2 *Muscle Activity*

AVERAGE ACTIVATION OF M. GASTR. LAT. GroundCondition and TimeInstant were found to have a main effect on average activity of the M. Peroneus longus (F(1, 19) = 5.84, p = 0.03 and F(1, 19) = 89.05, p < 0.01, respectively). An increase in activation was seen while running over ground (107 ± 43 µV) compared to running on a treadmill (94 ± 31 µV; p = 0.03). Also, average activation was higher during TCT (129 ± 34 µV) than during PA (73 ± 40 µV; p < 0.01). A significant interaction effect could be proven for Configuration*GroundCondition, F(1.74, 33.03) = 3.88, p = 0.04. During over ground running, highest activation levels were found when running in Con2 while during treadmill running, lowest activation levels were found when running in this shoe. Also, during over ground running, lowest activation levels were found when running in NS while during treadmill running they were highest in this footwear (figure 36).

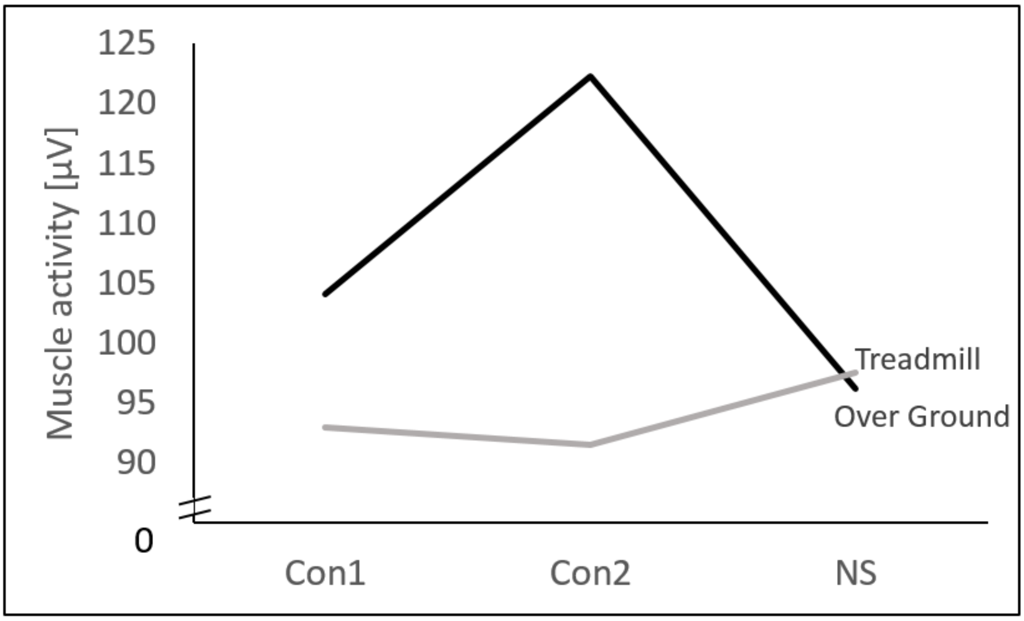

Figure 36: Interaction effect of Configuration*GroundCondition on average activation of M. Gastr. lat.

average activation of m. gastr. med. A significant main effect could be proven for TimeInstant, F(1,19) = 23.71, p < 0.01. Average activity of M. Gastr. med. showed a 52 µV (95% CI [29.45, 73.84]) higher activation during TCT (161 ± 65 μV) compared to PA (110 ± 88 μV).

AVERAGE ACTIVATION OF M. TIB. ANT. A significant main effect could be proven for TimeInstant, $F(1,19) = 33.80$, $p < 0.01$. Average activity of M. Tib. ant showed a 99 μV (95% CI [63.29, 134.49]) higher activation during PA (200 ± 111 μV) compared to TCT (101 ± 106 μV).

AVERAGE ACTIVATION OF M. PERONEUS LONG. Configuration and TimeInstant had main effects on the average activation of M. Peroneus long., $F(1.73, 32.90) = 3.78$, $p = 0.04$ and $F(1, 19) = 5.76$, $p = 0.03$ respectively. Subjects showed higher activation levels when running in Con1 (145 ± 63 μV) compared to running in NS (125 ± 59 μV). Also, activation levels were on average 35 μV (95% CI [4.42, 64.73]) higher during TCT (151 ± 42 μV) than during PA (117 ± 86 μV). Figure 37 shows average muscle activity for all investigated muscles during both TimeInstants: PA and TCT.

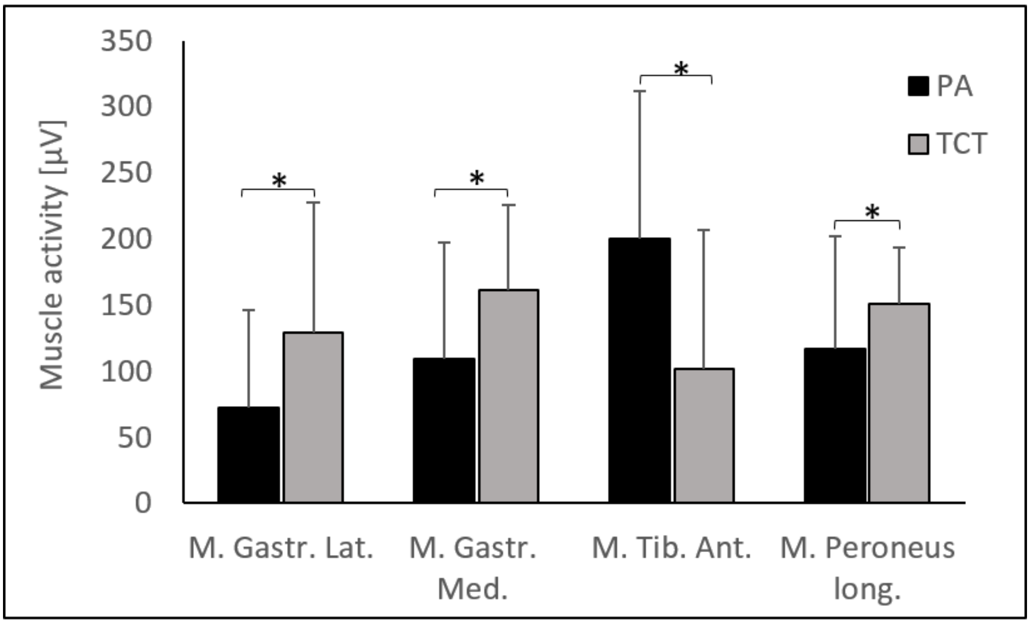

Figure 37: Average activity of all four muscles during PA and TCT. Each muscle showed significant differences between the investigated TimeInstants as indicated by a star.

9.3.3 *Foot Kinematics*

MAXIMUM PRONATION Shoe configuration was found to have a main effect on MaxPro, $F(1.40, 26.68) = 6.12$, $p = 0.01$. Pairwise comparison showed significant differences between trials during which subjects wore Con1 ($0.6 \pm 1.9°$) compared to Con2 ($2.7 \pm 3.9°$) or NS ($2.2 \pm 2.9°$).

MAXIMUM PRONATION VELOCITY A main effect could be proven for Configuration, $F(1.89, 35.78) = 41.36$, $p < 0.01$. MaxProVel differed at a significance level of $p < 0.01$ when comparing either of the running shoe configurations (Con1: 62.7 ± 36.5 °/s, Con2: 88.5 ± 42.0 °/s, NS: 133.1 ± 45.7 °/s).

RANGE OF MOTION Similar to the results of MaxPro and MaxProVel, a significant main effect of shoe configuration was found for ROM in the frontal plane, $F(2.00, 36.00) = 5.76$, $p = 0.01$. ROM was increased by 3.1° (95% CI [0.63:5.58]) when running in NS ($11.0 \pm 3.9°$) compared to Con1 ($7.9 \pm 3.6°$; $p = 0.01$).

TIME TO FINAL PRONATION Likewise, shoe configurations had a main effect on TFpro, $F(2.00, 38.00) = 12.74$, $p < 0.01$. Final pronation occurred later when running in Con1 (92 ± 10 ms) compared to NS (77.1 ± 17.0 ms, $p < 0.01$) or Con2 (79.9 ± 12.7 ms, $p < 0.01$).

9.3.4 *Oscillations at the Achilles tendon*

TIME SPACE The acceleration signal of one representative subject in the vertical x-direction as well as the resulting acceleration in the horizontal yz-plane are depicted in figure 38. The acceleration data shown in the graphs were recorded with the subject running on a treadmill. A time frame of 200 ms before and 300 ms after ffc is depicted.

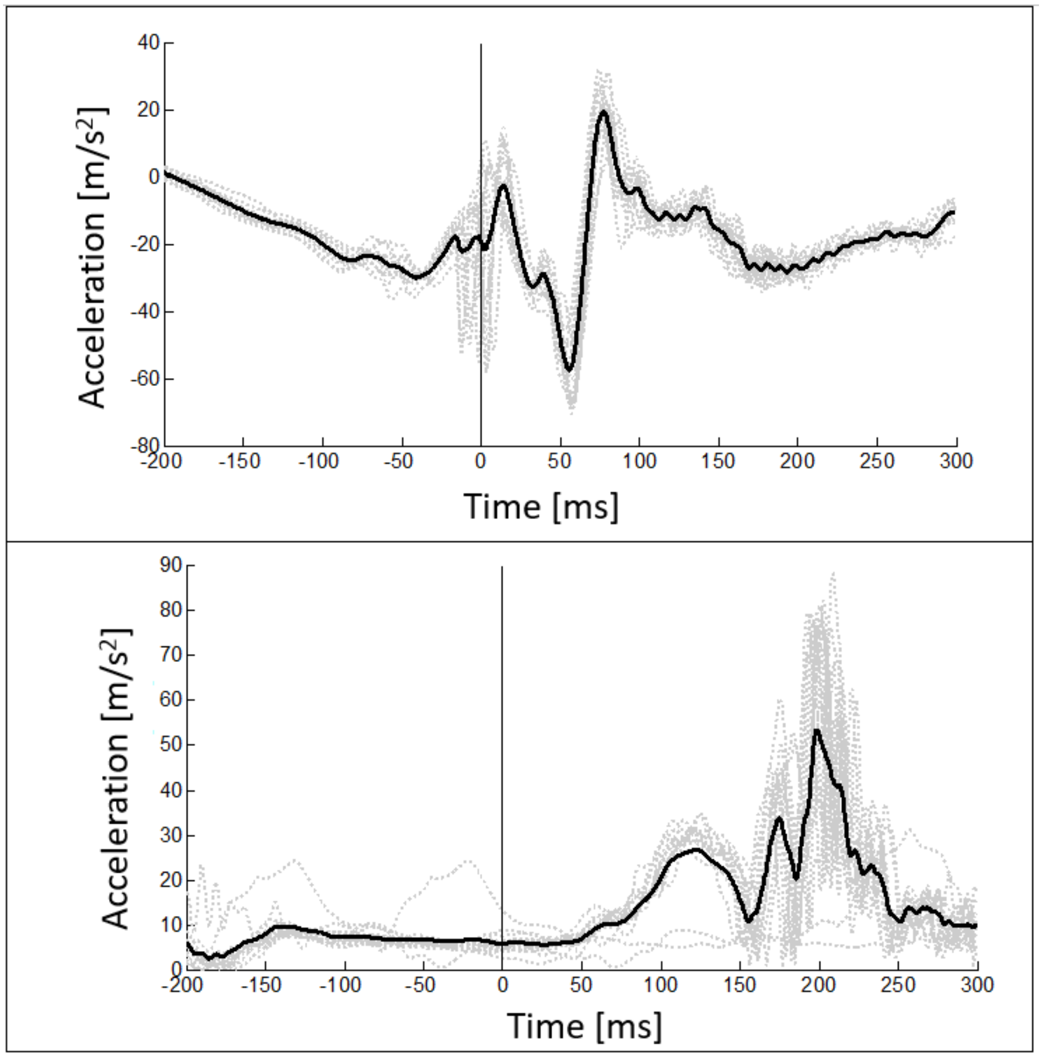

Figure 38: Characteristic acceleration signals: Acceleration signals recorded at the Achilles tendon while running on a treadmill shown for one representative subject. Grey lines show the data obtained from each step while black lines depict mean curves, averaged over all steps. ffc occurred at time = 0 ms.

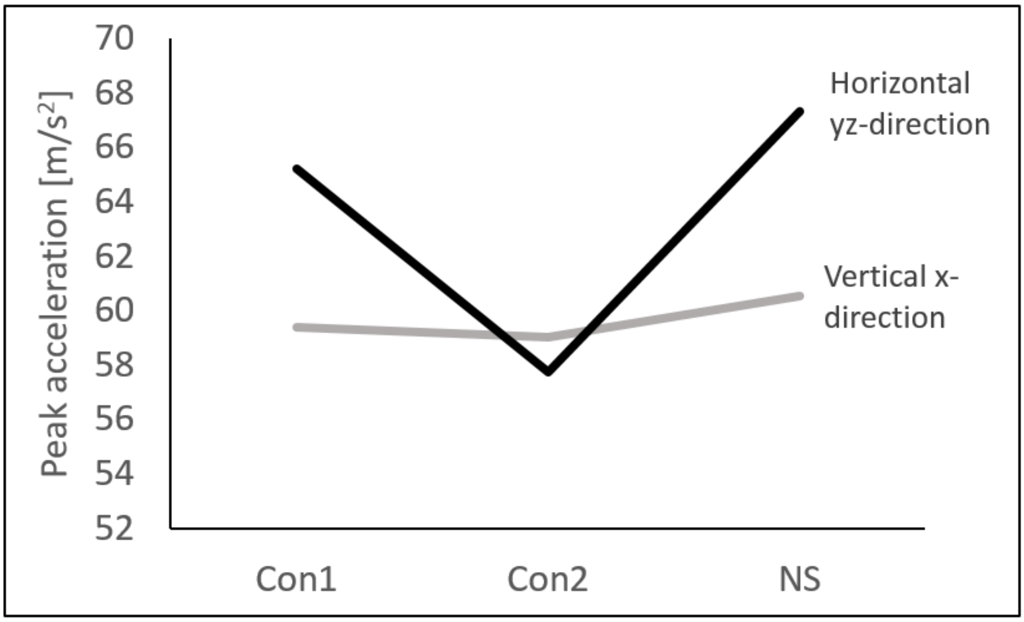

Figure 39: Interaction effect of Direction*Configuration. Peak accelerations measured at the Achilles tendon in two different directions (vertical x-direction and horizontal yz-plane). Measurements were performed with subjects running in three different types of footwear.

Direction was found to have a significant main effect on PeakAcc, F(1, 19) = 6.62, p = 0.02. PeakAcc in the vertical x-direction (60 $\pm$ 13 m/s^2) were significantly lower than those in horizontal yz-plane (63 $\pm$ 17 m/s^2), p = 0.02. A significant interaction effect was detected for Direction*Configuration, F(2, 38) = 8.27, p = 0.01. PeakAcc in vertical x-direction were barely influenced by the shoe condition while those in horizontal yz-plane were decreased when running in Con2 (58 $\pm$ 19 m/s^2) compared to Con1 (65 $\pm$ 14 m/s^2) or NS (67 $\pm$ 18 m/s^2) as shown in figure 39.

Accelerometer location was found to have a significant main effect on time to peak acceleration following foot contact (ttpeak), F(1, 19) = 8.00, p = 0.1. ttpeak was shorter at the distal (112 $\pm$ 75 ms) compared to the proximal accelerometer (124 $\pm$ 75 ms), p = 0.1. Therefore, peak accelerations occurred earlier at the distal accelerometer than at the proximal sensor. A significant interaction effect could be proven for Location*Direction, F(1, 19) = 16.60, p

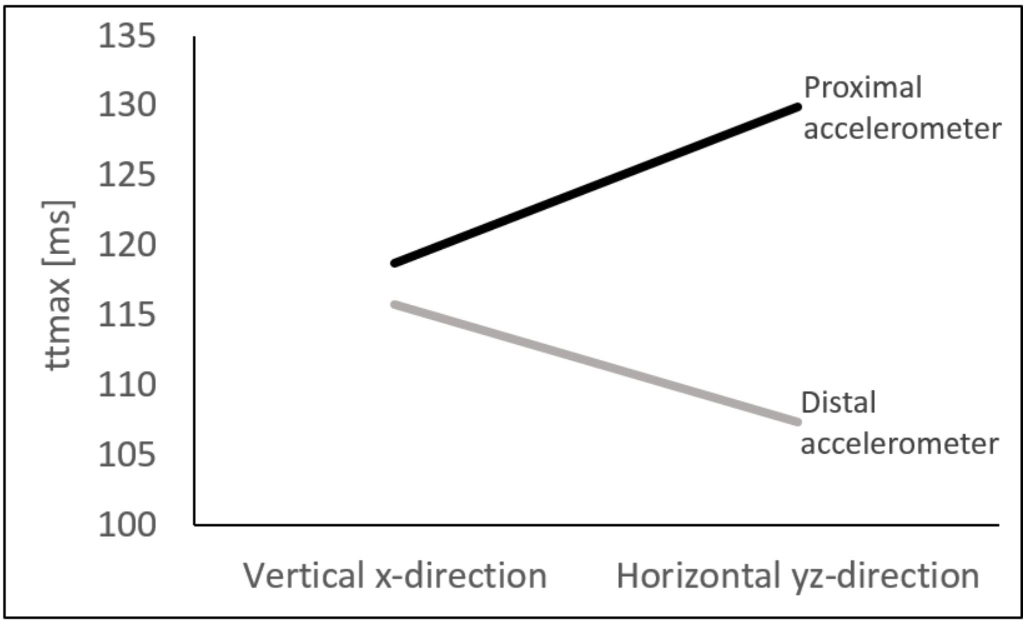

Figure 40: Interaction effect of Location*Direction. ttpeak measured at the Achilles tendon using two different sensor locations. Measurements were performed in two different directions (vertical x-direction and horizontal yz-plane).

< 0.01. ttpeak in vertical x-direction was comparable between the two accelerometers (distal: 116 ± 79 ms; proximal: 119 ± 74 ms). However, for the proximal accelerometer a profound elongation of ttpeak was found in the horizontal yz-plane (130 ± 76 ms) while a reduction of ttpeak occurred at the distal accelerometer (107 ± 71 ms, figure 40).

FREQUENCY SPACE Figures 41 shows mean normalized power spectra of the acceleration signals obtained while running at two different ground conditions (over ground and treadmill), each in the horizontal yz-plane and in the vertical x-direction. While data in the horizontal yz-plane show a more even power distribution with a moderate peak in the medium-frequency interval, the power spectra obtained from data in the vertical x-direction show two characteristic peaks, one in the middle-frequency interval and the other in the high-frequency interval. When comparing the power spectra obtained during treadmill running to those obtained during over ground running, no distinct differences in their characteristics can be found. Figure 42 shows mean normalized power spectra of the distal accelerometer in vertical x-direction while wearing different shoe configurations.

The average dominant frequency of oscillations at the Achilles tendon was found to be 31 ± 18 Hz. A difference in the dominant frequency was only found between measurement directions, $F(1, 19) = 20.89$, $p < 0.01$, not between oscillations induced by running at different grounds, in different shoes or between the two accelerometer locations. The dominant frequency of oscillations in vertical x-direction was 26 ± 12 Hz while it was 35 ± 20 Hz in the horizontal yz-plane. Significant main effects on normalized power were found for configuration, $F(2, 38) = 25.82$, $p < 0.01$, ground, $F(1, 19) = 7.30$, $p = 0.01$, frequency intervals, $F(1.99, 42.13) = 977.63$, $p < 0.01$, and accelerometer location, $F(1, 19) = 34.54$, $p < 0.01$.

Average normalized power was significantly higher when running in Con3 (0.7 ± 0.2) compared to Con1 (0.6 ± 0.1; $p < 0.01$) and when running in Con2 (0.6 ± 0.2) compared to running in Con1 ($p < 0.01$). A significant mean difference of 0.02 (95% CI [0.04:0.01]) in normalized power was found between running trials performed on a treadmill ($p = 0.01$) and those performed over ground with signal power being slightly lower when running over ground. Normalized power was highest in the high-frequency interval (0.9 ± 0.2) and lowest in the medium-frequency interval (0.3 ± 0.2). Figure 43 illustrates the differences between frequency intervals, all significant differences are indicated by p-values < 0.01. Average normalized power was 0.3 (95% CI [0.25:0.29]) lower in the signal of the distal accelerometer (0.5 ± 0.1) compared to the signal of the proximal accelerometer (0.8 ± 0.2).

In figure 44 the mean transfer function in both measurement directions is shown in logarithmical plotting for better clarity. In both measurement directions only minimal power attenuation can be seen in the graphs. While barely any damping characteristics can be observed in the vertical x-direction, the power of the proximal

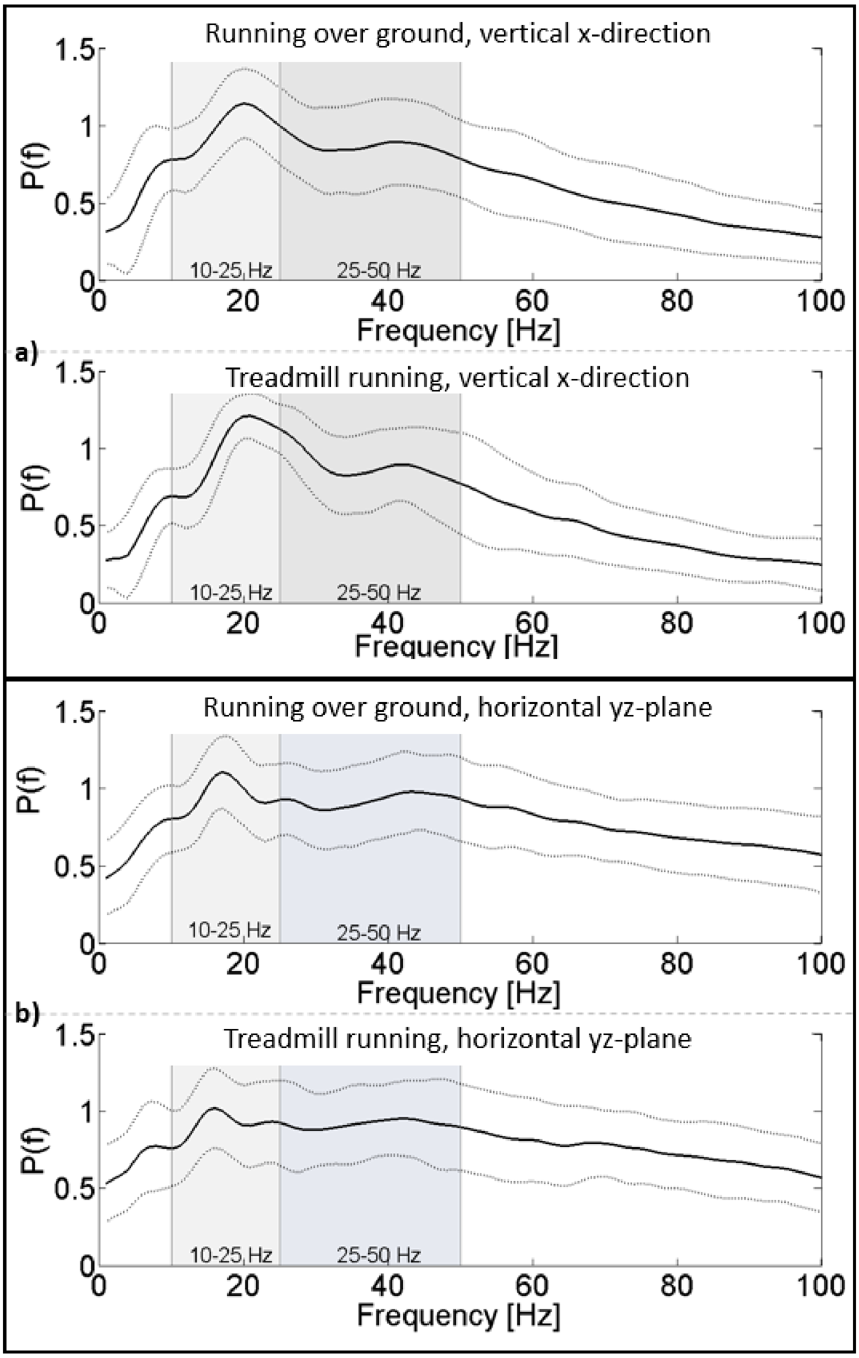

Figure 41: Normalized power spectra of the distal accelerometer: average power spectrum for running over ground and running on a treadmill, separated in two measurement directions. Graph a) shows data in the vertical x-direction while graph b) shows data in the horizontal yz-plane. Solid lines indicate mean values of all study participants, dashed lines indicate standard deviations (SD). Low-frequency and medium-frequency intervals are depicted in the graphs.

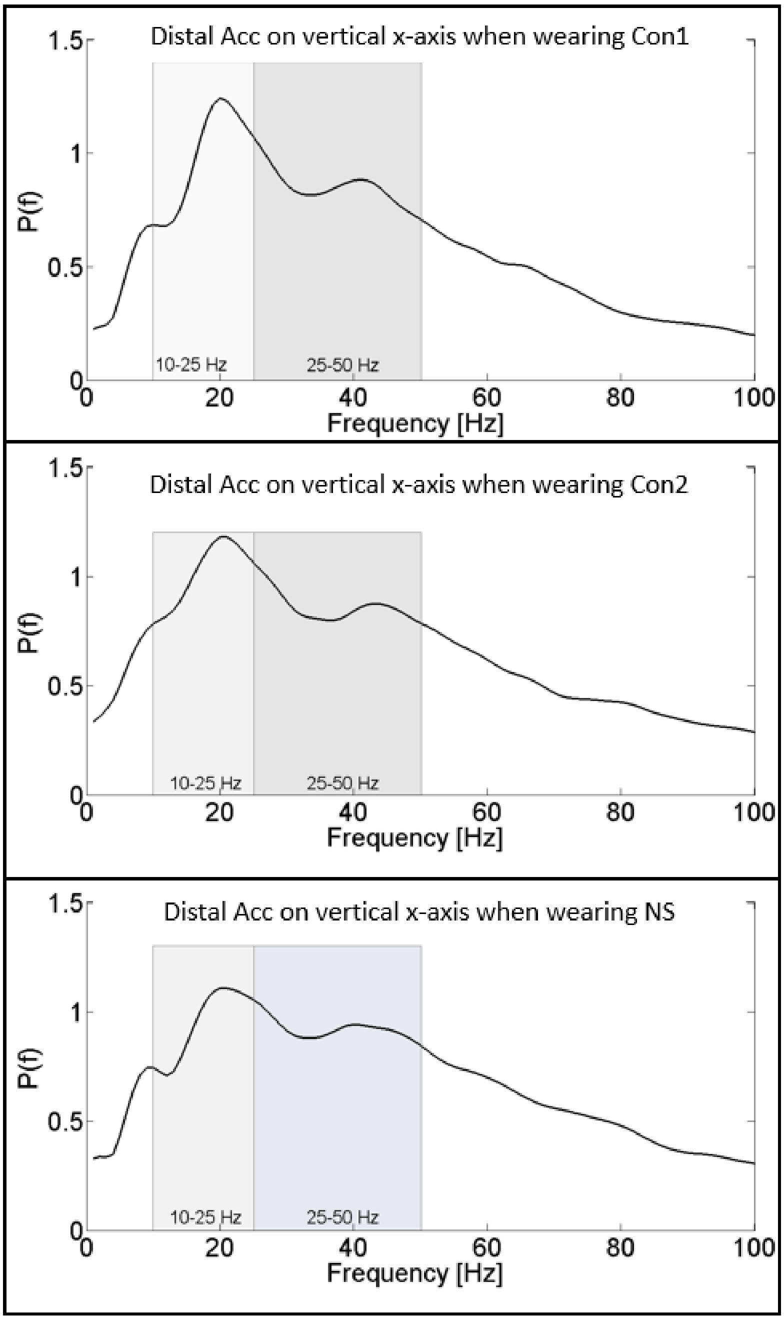

Figure 42: Normalized power spectrum of the distal acceleration signal in vertical x-direction: Power is averaged over ground conditions (treadmill and over ground) and separately shown for three different running shoe configurations. Low-frequency and medium-frequency intervals are depicted in the graphs.

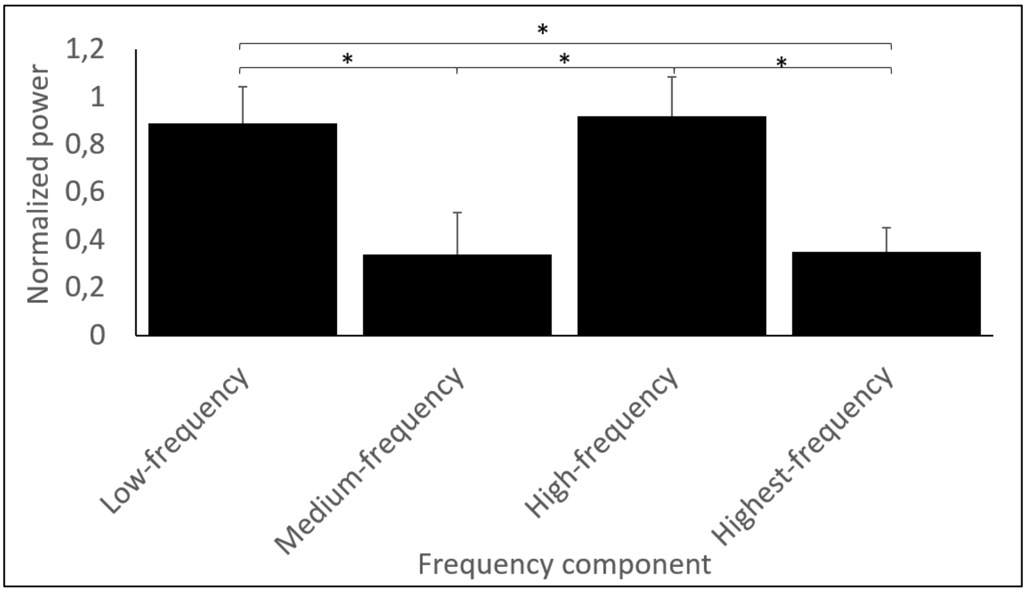

Figure 43: Average normalized power in different frequency intervals: Power was averaged over all subjects and all trials, both accelerometers are included. Bars represent mean values while error bars show SDs. Stars indicate significant differences.

accelerometer is very slightly attenuated in the proximal accelerometer data in the horizontal yz-plane in frequency components above 50 Hz.

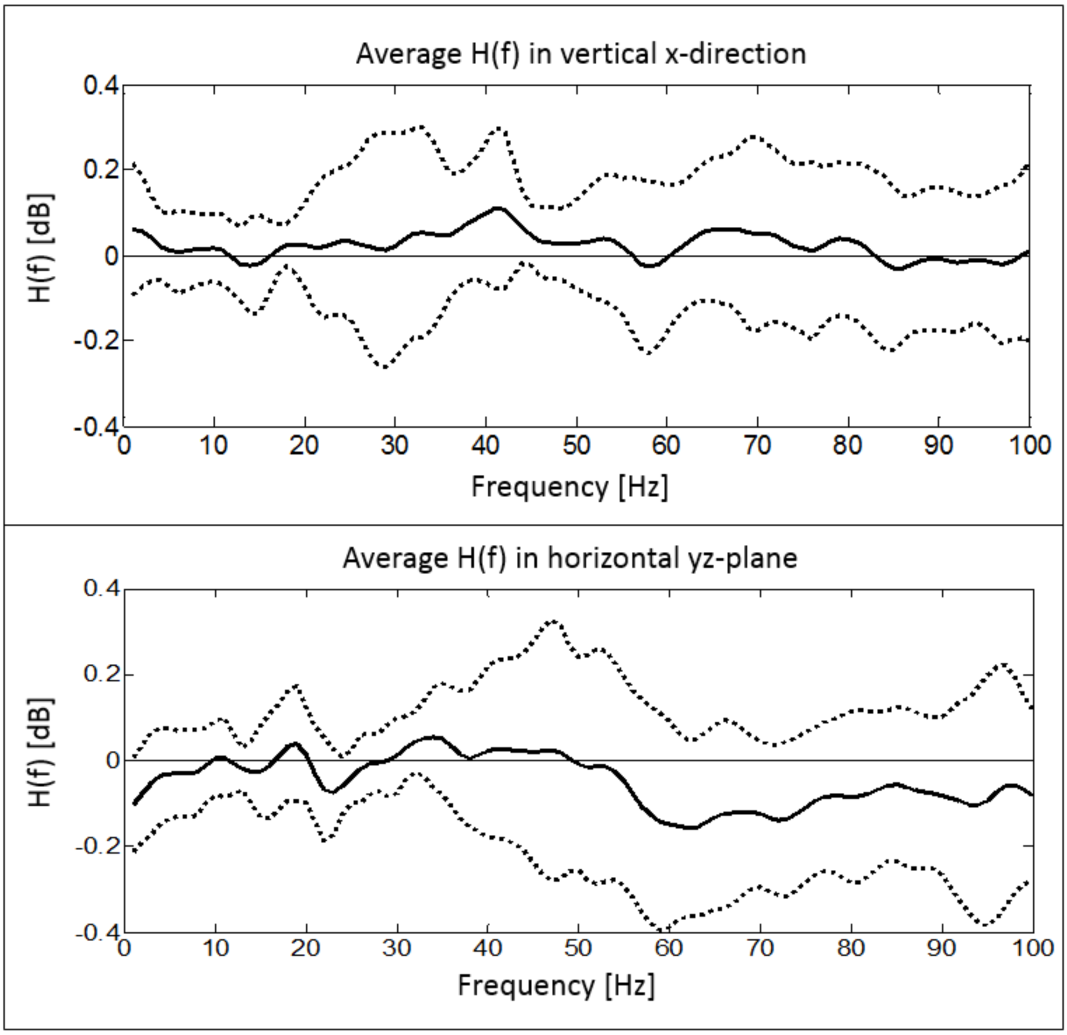

Figure 44: Average transfer function: Power attenuation between distal and proximal accelerometer data. Solid lines show mean data in both measurement directions while dashed lines represent SDs.

9.4 DISCUSSION

The purpose of this study was to explore the effects of different running shoes and different surface conditions (treadmill versus over ground) on plantar pressure, muscle activity, foot kinematics and oscillations at the Achilles tendon during running. Contrary to the hypothesized decrease in total force underneath the foot during treadmill trials, no effect of ground condition on this parameter was found. A damping effect of the treadmill was expected to occur at every step. This damping effect was not quantified during measurements though and must therefore be denied based on the total force outcome. Contrary to other researchers (García-Pérez, Pérez-Soriano, Llana, Martínez-Nova & Sánchez-Zuriaga, 2013), no main effect of ground condition on peak pressure was proven in the present study. This diverse finding may be explained by differences in the experimental protocol. In the present study, participants ran for short durations after a warm up and familiarization period. In the study by (García-Pérez et al., 2013), subjects performed a fatiguing 30 min run. This suggests the occurrence of differences in plantar pressure with prolonged running, which also better reflects real life practice of this sport. Therefore, future studies should take possible effects of prolonged runs and therefore increased fatigue on the investigated parameters into consideration. Even though, no main effect of ground condition could be proven for peak forces, an interaction effect of Configuration*GroundCondition was found. Overall, peak forces differed more distinctly when running on a treadmill. Variations between shoe conditions were congruent in both ground conditions, while runners showed more exceeding values when running on a treadmill. It should therefore be noted, that the effect of running shoes on peak forces is more pronounced during treadmill running compared to running on a concrete floor.

Overall muscle activity differed between the investigated gait phases (PA and TCT). Average activity during TCT was higher than 50 ms before heel contact (at PA) for M. Gastr. Lat., M. Gastr.

Med. and M. Peroneus long. but not for M. Tib. Ant. Accoring to Götz-Neumann (2011), the M. Tib. Ant. performs excentric muscle work during PA to decelerate the plantarflexion of the foot, which is caused by gravity. Additionally, contraction of this muscle leads to a forward pulling of the tibia, resulting in knee flexion of about 15°. The other investigated muscles, however, perform their main functions during TCT, therefore resulting in higher activation levels during this gait period. Average activity of M. Peroneus long. was affected by shoe configuration. Subjects included in this study showed higher EMG activity of the M. Peroneus long. when running in a shoe configuration with medial wedges, soft damping material and an arch support (Con1) compared to the reference shoe (NS). This effect could not be proven in the first study (chapter 8) and may therefore be caused by the additional heel wedge used in the present study. Besides the heel wedge, which was added to Con1, no differences in shoe configurations exist between the two test series. The results of the present study are in contrast to the findings of Cheung & Ng (2010) who found a 9.6% higher activity in a neutral shoe compared to a motion control shoe. Other researchers found differences in the timing of activation when comparing different shoe modifications, but not in the amplitude of the EMG activity (Cheung & Ng, 2009). Even though, a large variety of insoles and shoe modifications was tested by this group, a comparison to the conditions used in the present study is hardly possible. The higher activation level of M. Peroneus long. may be explained as a contraction counteracting the control elements used in Con1. Baur et al. (2003) came to a similar conclusion when they found increased activity of the M. Peroneus long. after equipping shoes with arch support insoles. Higher activation levels of this muscles may be used to counteract the effects of medial wedges and arch support in the shoe, which are aiming at decreasing pronation movements. The function of M. Peroneus long. however is to pronate the foot, therefore counteracting the effects of medial wedges.

Muscle activity did not vary between ground conditions, which is in agreement with the studies of Wang et al. (2014) and Wank, Frick & Schmidtbleicher (1998). A recent study by Wang et al. (2014) looked at differences in muscle activity when running on a treadmill and on different types of ground surfaces. The group studied two muscles at the upper leg as well as M. Tibialis ant. and M. Gastrocnemius. They found differences in upper leg muscle activity but not in either of the two lower leg muscles. Wang et al. (2014) explained the differences in muscle activity of the upper leg muscles by kinematic adjustments when running over ground or on a treadmill. In line with this argumentation, no kinematic differences were found in the present study, linking up with consistent muscle activation levels throughout ground conditions. Although no main effects of ground condition on muscle activity could be proven, an interaction effect of Configuration*GroundCondition on average activity of the M. Gastr. lat. was found. Running in NS led to similar EMG activity, irrespective of the ground condition. During treadmill trials, a large increase in muscle activity was found when running in Con2 compared to Con1 while a small decrease was found for this comparison during over ground running. The subjects included in the present study were habitual road runners who usually run over ground, not on a treadmill. The interaction effect described above shows a rather consistent activation level of the muscle when running over ground. Performing this motion on a treadmill may have challenged their motor control more, leading to different findings in this ground condition. Dolenec, Stirn & Strojnik (2015) found differences in muscle activity during PA of M. Tib. ant. and M. Peroneus brevis when running on asphalt and on grass surface. It therefore remains uncertain, whether the results obtained for running on concrete floor in the current study fully reflect the biomechanics of road running. Also, many athletes do not only train on roads but also on forest tracks, gravel roads or running tracks. It can therefore not be excluded, that results might differ when running on these surfaces.

Contrary to our expectations, foot kinematics did not change significantly between ground conditions. This hypothesis was based on the assumption of damping characteristics of the treadmill. A damping mechanism of the treadmill would influence the need of pronation motions, which are known to act as a damper of forces applied to the foot in vertical direction. As described above, no indication could be found for such damping mechanism of the treadmill as total force underneath the foot did not differ between ground conditions. Therefore, no adaption of foot kinematics in the frontal plane was required to compensate differences in ground reaction forces. McKenna & Riches (2007) found changes in runners' kinematics while running on a treadmill compared to over ground running, while Fellin, Manal & Davis (2010) found no variations in foot motions. It should be noted though, that the investigated running speed in the study by McKenna & Riches (2007) was much higher (7.0 m/s) than in the current study. But in the study of Fellin et al. (2010), where no kinematic differences were present, a running speed of 3.35 m/s was chosen, which is comparable to the speeds investigated in the present study. It can therefore be concluded, that kinematic differences between the two ground conditions may only occur at higher running speeds. These speeds cause higher ground reaction forces and may therefore lead to an actual damping effect of the treadmill. If a damping effect of the treadmill is present at higher speeds, it demands changes in the damping characteristics of the biological system. Pronation is known to damp forces striking the foot (Hintermann, 1998; Nigg, 2001) and may therefore be adapted to lower ground reaction forces at the treadmill.

In contrast to the expectations, ground conditions (treadmill versus over ground) did not have an effect on oscillations in time space and produced only slight variations in frequency space. Even though a significant modification of signal power could be detected depending on the ground condition, a change of 0.02 seems extremely small. It will therefore most likely not be of biomechanical relevance. We originally expected differences in ground conditions to lead to variations in running technique and therefore to alterations in the vibration behavior at the Achilles tendon. However, movement pattern fixation is known to occur in experienced runners, especially in running style parameters (Lees & Bouracier, 1994). Therefore, the experienced runners included in the present study may have held on to their running style, independent of ground condition. This assumption is also supported by the kinematic results presented in this study. No effect of ground condition on foot kinematics was found. Results of Achilles tendon oscillations obtained from treadmill experiments may likely be transferred to over ground running.

The power spectra of the present study show a large amount of power in the frequency band of 0 - 10 Hz, while a high pass filter with a cut off frequency of 10 Hz was applied. This filter was used to minimize accelerations that are due to motions of the lower leg during running, rather than accelerations due to vibrations at the Achilles tendon. An ideal filter would have resulted in zero power at frequencies $\leqslant$ 10 Hz. In reality, however, filters do not cut off data abruptly. A frequency attenuation falls off rather slowly with increasing frequency, which can be inspected by the signal dropping off beneath the cut off frequency, as desired (Cimbala, 2013; Ray & Acharya, 2004).

While differences between the data obtained from the two accelerometers were found in the time domain, no differences could be proven in the frequency domain. In contrast to the results of Boyer & Nigg (2006b) we did not detect an attenuation of power between distal and proximal accelerometer locations. Therefore, the signal power of the two accelerometers did not differ. This effect may have been influenced by the relatively close locations of the two accelerometers in the present study. Step detection was performed using an in-shoe pressure system, which included the mounting of a cuff on the lower leg. This cuff restricted the accelerometer locations at the Achilles tendon. Also, tendon length was estimated by measuring tibial length, which may not be a highly accurate appraisal. The use of an ultrasound system could provide a more accurate definition of the Achilles tendon length and therefore allow a more precise placement of the accelerometers (Farris, Trewartha & McGuigan, 2012).

The stiffness of the underlying material may have influenced the detected attenuation. Boyer & Nigg (2006b) studied oscillations at the M. Quadriceps femoris while we collected data at the Achilles tendon. Stiffness of muscle tissue highly depends on its activation level while tendons are known as a relatively stiff material (Wang, De Vito, Ditroilo, Fong & Delahunt, 2015; Lichtwark & Wilson, 2008). They may therefore transfer the input signal without significant attenuation. Future research should investigate the power attenuation at a farther distance between two accelerometers at the Achilles tendon. Increasing the distance between the two accelerometers may also lead to larger differences in ttpeak of the two sensors. In the present study ttpeak was significantly shorter at the distal accelerometer, which is in agreement with the expected impact transmission along the lower leg following heel contact.

Overall, peak normalized power occurred in the medium-frequency interval (25 - 50 Hz) while average normalized power was lowest in this interval. The normalized power spectra do not show different characteristics when comparing over ground running to treadmill running (figure 41). As mentioned above, a minor decrease in signal power was detected when running over ground. Ground reaction forces are known to differ when running on a treadmill compared to over ground running, therefore leading to altered input to the system (Riley et al., 2008; AlGheshyan, 2012). In the current study, differences in peak forces were proven, while no changes were found in total force underneath the foot. These differences in the system's input signal may have been encountered with specific muscle activity to ensure a stable oscillation behavior at the Achilles tendon, independent of ground condition.

Oscillations in the vertical x-direction caused a more distinct peak in the medium-frequency interval compared to those in the horizontal yz-plane. The input, possibly characterized by more pronounced first impacts of the foot to the ground, may have led to differences between the power spectra properties of the two measurement directions. In the horizontal measurement direction, a resulting acceleration vector was used for calculations while in the vertical measurement direction, a single vector was used. This may have led to changes in the signal amplitude but not in its frequency. Although the power spectrum obtained in the horizontal yz-plane is more evenly spread over a larger variety of frequencies compared to that in vertical x-direction, the maximum peaks can be found in the medium-frequency interval for both measurement directions. It should be noted that past research has either focused on the vertical direction when analyzing soft tissue oscillations or did not specify measurement directions, although accelerometers with multiple dimensions were used (Boyer & Nigg, 2006b; Wakeling & Nigg, 2001; Fu et al., 2013). Comparisons with values from previous studies should therefore be made with caution.

Variations in shoe configurations led to changes in average normalized power. Overall, signal power decreased with increased damping of the shoe. Due to the modification of both, shock absorption and foot position (wedges), it is impossible to determine a single component that caused the changes in average power. Therefore, the Achilles tendon was possibly under different tension, experienced different impact frequencies and was positioned differently. It can therefore only be determined that the combination of those modifications affected the power at the tendon. A single source causing the difference cannot be specified with the present setup. Nor can the possibility that only the combined modifications leads to a change be eliminated.

No effect of shoe modifications on resonance frequency was proven in the present study. Our findings are in agreement with those of Fu et al. (2013) who did not find differences in the resonance frequency when wearing a basketball or a control shoe during drop jumps. Anticipatory muscle activity of the runners may have prevented an effect of shoe modifications on resonance frequency as well as power attenuation. Subjects participating in the present study were experienced runners. Therefore precise motor control during this well-known movement can be assumed, possibly damping oscillations at the Achilles tendon more effectively than differences in the damping characteristics of a running shoe.

It should be noted that subjects were given a familiarization period in each running shoe before the measurements were conducted. This may have led to an optimized muscle tuning while running in each footwear. The results may have been different if no familiarization period was given, leading to an unexpected change in shoe conditions which could have led to less effective neuromuscular control. A lack of muscle adaptation to the unexpected shoe modifications may lead to changed impact forces, resulting in increased soft tissue oscillations and increased oscillations at the Achilles tendon (Boyer & Nigg, 2006a).

Even though skin movement artifacts are known to be a common source of error when analyzing structures which lay underneath the skin, it is currently unknown how much influence the relative movement between the Achilles tendon and the skin had on the accelerations measured in the present study. Further research is needed in order to clarify the extent to which accelerations obtained through skin-mounted accelerometers correspond to accelerations of the Achilles tendon itself. It also remains unknown whether our findings can be transferred to less experienced runners or subjects with a less precise motor control.

LIMITATIONS During measurements of over ground running, running speed was monitored using a customary stop watch and no speed check throughout the runway was performed. Hence, running speed variations during over ground running can not be ruled out while a constant running speed at the treadmill is confirmed.

Therefore, the results comparing over ground and treadmill running may have been affected by running speed variations during over ground running.

CONCLUSION First insight was gained in the oscillation behavior occurring at the Achilles tendon during running. In conclusion, oscillations were found to vary between measurement directions, sensor locations and if different footwear was worn. Only very slight variations could be detected if running over ground compared to treadmill running. While peak accelerations reach the distal end of the Achilles tendon faster compared to the proximal end, no clear attenuation of oscillations could be proven along the tendon. It can therefore be summarized, that the Achilles tendon is homogeneously affected by the magnitude and frequency of oscillations but varies in the timing of these vibrations.

10

EFFECT OF FATIGUE ON THE COLLECTED DATA

Table 7: Details of study 3, comparing biomechanical data during different time points of an hour-long endurance run

Details of the Study	
N	31 ♂
Age [years]	31.3 ± 10.71
Height [cm]	179.10 ± 5.86
Weight [kg]	75.3 ± 6.2
Weekly running distance [km]	29.6 ± 20.0
Running experience [years]	8.3 ± 8.3
Measurement systems	3D IMU
	Two 3D accelerometers
	Force plate instrumented treadmill
	EMG
Conditions	One hour endurance run
	Data collection at seven distinct time instants

10.1 INTRODUCTION

In amateur or professional running, long distance runs are performed over distances of at least three kilometers. Therefore, road runners experience a highly repetitive load pattern which may alter with increasing fatigue after a prolonged period of running. Fatigue may therefore have relevant influence on running biomechanics. Achilles tendinopathy is the second most common musculoskeletal injury in runners with its prevalence reaching up to 18.5% in ultra marathon runners (Lopes et al., 2012). Friesenbichler, Stirling, Federolf & Nigg (2011) found increased vibration intensities at higher fatigue stages when measuring oscillations at the calf muscles in medio-lateral direction but not in axial direction. The authors could also detect an effect of fatigue on maximum vibration intensity, which occurred later if subjects were fatigued. However, those results are based on measurements undertaken at a relatively small sample size of 10 participants and runs were performed on an outdoor course including elevation gain on each lap. Also, while runs are described as *fatiguing*, no information is given about the duration of the runs. In a simulation study, Nikooyan & Zadpoor (2012) concluded that an increase in vibration amplitude of the lower body soft tissue packages occurs with increasing fatigue. An experimental confirmation of this simulation can be found in the study by Khassetarash, Hassannejad, Ettefagh & Sari-Sarraf (2015) who studied vibrations of the M. Gastrocnemius during a prolonged run. They concluded decreased muscle function to occur with increasing fatigue, leading to larger vibration amplitudes. If the calf muscles' vibration behavior changes with increasing fatigue, muscle tuning may be impaired leading to variations in oscillations at the Achilles tendon. Farris et al. (2012) studied Achilles tendon mechanics before and after a 30-min run. They found the properties of the Achilles tendon not to vary after the run. Therefore, changes in vibrations at the Achilles tendon may be attributed to variations in muscle tuning as well as in foot kinematics. Until now, no study has has considered the effect of fatigue on oscillations of the Achilles tendon, though.

Past research suggests the regulation of ground reaction forces as well as of soft-tissue vibrations to be realized through muscle activation (Nigg, 2001). With muscle activity as an important factor in shock attenuation, both impact shock as well as resulting vibrations, it is important to investigate how plantar pressure, EMG activity and oscillations at the Achilles tendon change over a prolonged run. Variations in impact attenuation as well as in oscillation damping may be of relevance for injury development and therefore injury prevention in runners. Up until now, only little is known about the effects of fatigue on soft tissue vibrations, although it is known that substantially higher impact accelerations occur at the tibia in a fatigued state (Mizrahi, Verbitsky, Isakov & Daily, 2000). An inconsistent pattern is seen in the literature with regard to the effect of fatigue on ground reaction forces. Some researchers found a decrease in ground reaction force with increasing fatigue (Girard, Millet, Slawinski, Racinais & Micallef, 2010; Nummela, Rusko & Mero, 1994; Rabita, Slawinski, Girard, Bignet & Hausswirth, 2011) while others found either no effect of fatigue (Gerlach et al., 2005; Nikooyan & Zadpoor, 2012) or even an increase in ground reaction forces (Fourchet et al., 2015; Christina, White & Gilchrist, 2001; Nyland, Shapiro, Stine, Horn & Ireland, 1994). In static measurements, Escamilla-Martínez, Martínez-Nova, Gómez-Martín, Sánchez-Rodríguez & Fernández-Seguín (2013) identified pressure variations under the second metatarsal head and the medial heel, which were encompassed by a general tendency towards pronation after a 60 min run. The outcome on foot kinematics is in agreement with the results of Dierks, Davis & Hamill (2010), who looked at kinematic changes during a fatiguing training run. The researchers found most notably effects of fatigue on foot eversion. Komi (2000) suggested angular displacements to be controlled in large by muscle activity. If a runner becomes exerted, neuromuscular functions alter and may lose the ability to maintain the desired foot displacements, leading to increased trajectories of this segment.

The goal of the present study was to obtain parameter constellations of biomechanical data, which allow differentiation between stages of fatigue during prolonged running. According to the literature mentioned above, plantar pressure and EMG data, kinematic data of the foot and data describing oscillations at the Achilles tendon were included in the analysis. The identification of principal components was hypothesized to allow the differentiation between distinct time instants of a one hour run. It was expected to find a suitable variable constellation to distinguish between early and late phases of the run.

10.2 METHODS

Thirty-one male rearfoot runners participated in this study. Subject characteristics are presented in table 7. After a six minute warm up period on the treadmill, a flying start to a one hour fatiguing run was performed. Subjects were instructed to choose a speed close to their one hour competitive running speed. In the first study described in this manuscript (see chapter 8), differences in the collected data were found between running speeds. Therefore, running speed variations within the one hour run were only allowed in a range of 0.28 m/s around the participant's starting speed. This constraint was applied to limit the confounding effect of different running speeds within one subject. Thoughout the run, data were collected at 7 distinct time instants: 00 min, 15 min, 30 min, 45 min, 50 min, 55 min, 59 min. Closer intervals were chosen towards the end of the run as fatiguing effects were expected to be more pronounced in the final 15 minutes compared to earlier time spans.

Measurements and data analysis of EMG and kinematics data as well as of oscillations at the Achilles tendon were identical to the study described before (see Chapter 9). Plantar pressure data was obtained through a treadmill, which is equipped with an integrated measuring platform (Zebris®, Isny, Germany). The platform matrix consists of about 5000 pressure sensors and measures 150 x 50 cm. Data collection was performed in a synchronized fashion using Noraxon Telemyo 2400 G2 (Noraxon Corporate, Scottsdale, AZ, USA) at a sampling frequency of 3000 Hz. Pressure data was analyzed in Noraxon MR3 software (Noraxon Corporate, Scottsdale, AZ, USA) with the following output parameters: maximum (max force), force integral and maximum force slope (max force slope). Force integrals were calculated as the product of force [N] and stance time [% of stride] while max force slopes were obtained by division of force [N] through stance time [% of stride]. A total of 33 variables was determined at each of the seven time instants. Therefore, 231 distinct variables were collected of each participant. Sporadically, missing values occurred in the data set. These were due to difficulties with the measurement devices like failure of plantar pressure recordings or loss of single accelerometer axis. These values were only missing for single trials of a subject, not for entire recordings. As missing data hinders research as most statistical methods assume full data sets, missing values were estimated via multiple imputation. Multiple imputation is a sophisticated missing-value analysis, which calculates a multitude of estimates for each missing data point (Rubin, 1987). Missing values are therefore replaced by plausible data and statistical procedures can be performed on the imputed data set. Several steps of imputation are run to incorporate missing-data uncertainty. Each missing value is replaced by a list of imputed, plausible values, representing the uncertainty about the true value (Zhen, 2008). Subsequently an average value is selected to be inserted in the data sheet.

In the studies presented in chapter 8 and 9, differences between shoes, running speeds and ground conditions were tested. Therefore, distinct variables were analysed and checked for differences between the investigated conditions. In the present study, the data structure was analyzed exploratory and not tested for differences. Data structures were sought, which allow a differentiation between time instants and therefore fatigue stages during the one-hour run. Therefore, data were separately submitted to explorative PCAs to find principal components consisting of different variables. Therefore, discrete PCAs were run for pressure, EMG and kinematics data as well as for data describing oscillations at the Achilles tendon. Data were automatically scaled to allow comparisons of values, which are represented in different units (e.g. accelerations [m/s^2] and frequencies [Hz]). Variables which showed loading factors of $\leqslant 0.4$ on the first four principal components were excluded if more than three components were found for this data set. Subsequently, another PCA was run on the reduced data, leading to a specification of the results and a reduction of the components to a graphically presentable number. The number of components was determined through eigenvalue theory with relevant components obtaining an eigenvalue $\geqslant 1$. Also, scree plots were created for all PCA results and checked visually. Kaiser-Meyer-Olkin index (KMO-index) and Bartlett-test of sphericity were calculated for each data set to check suitability of the input parameters. All PCAs were performed using SSPSS (SPSS 21®, IBM, Armonk, NY, USA). Subsequently, score plots of the principal components were created using either Microsoft Excel (Microsoft Office Professional Plus 2013®, Microsoft, Redmond, WA, USA) for one and two dimensional plots or Matlab (Matlab R2013b®, Mathworks, Natick, MA, USA) for three dimensional plots. Within the score plots, the different time instants of measurement were color coded to allow for differentiation between stages of fatigue. Those scree plots were then checked for data clusters of each time instant.

10.3 RESULTS

Due to complication of measurements 14 subjects had to be excluded from data analysis. For three subjects, the reference posture was not recorded correctly and could therefore not be used to analyze foot kinematics. For six subjects, pressure data was either totally or partially missing and could not be used for step detection. For another 5 subjects, the IMU showed inconsistent data. Therefore, results presented in this chapter were obtained from data of 17 subjects who were included in the final analysis.

Step frequency and flight time were recorded throughout the investigated time instants. Step frequency was found to decrease with increasing running time while flight time showed a steady increase over time (see figures 45 and 46). Graphs of the descriptive statistics of all investigated parameters can be found in the appendix. Most of the investigated data showed high correlations, therefore representing suitable input arguments for a PCA. Despite the relatively small sample size, most correlations were also found to be significant. These findings are also resembled by the KMO-index and the Bartlett-test of sphericity. The KMO-index was consistently > 0.58, namely 0.58 for pressure data, 0.62 for EMG data, 0.65 for vibration data and 0.72 for kinematics data. Bartlett-tests of sphericity were significant for all data. Therefore, interrelations between the variables were proven to exist. Altogether, the data was found to meet the criteria and was therefore suitable for PCA.

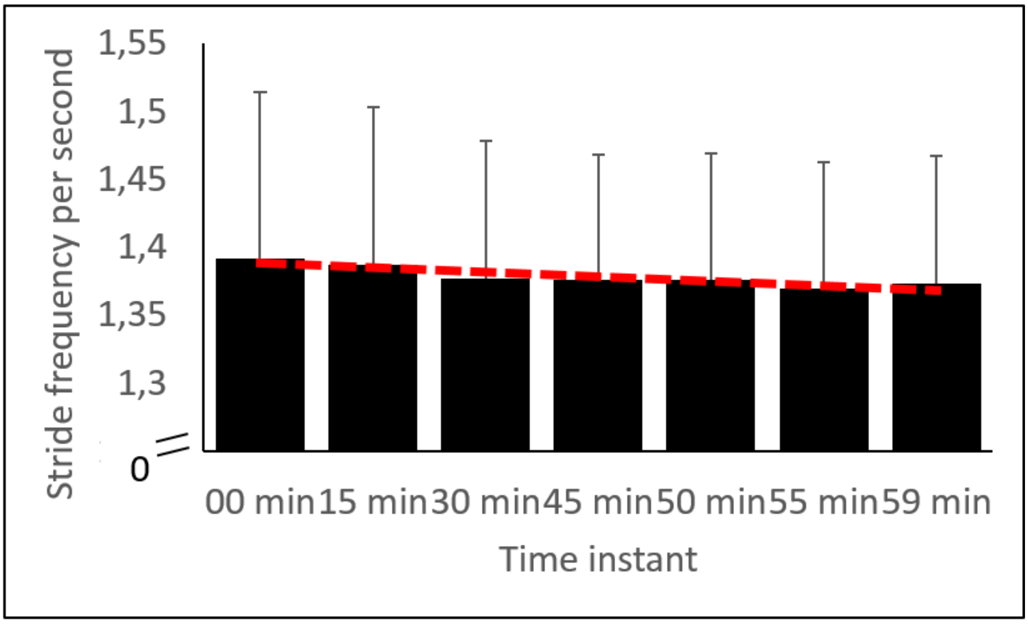

Figure 45: Step frequency of steps of the right foot per second depicted for each investigated time point

Pressure data could be reduced to a single component with an eigenvalue of 1.95, explaining 64.89% of the total variance. EMG data was reduced to two components, explaining 86.18% of the total variance. The eigenvalue of component 1 was 3.42 and 1.75 of component 2. Kinematic data were categorized in two components, explaining 75.02% of the total variance with eigenvalues of 2.99

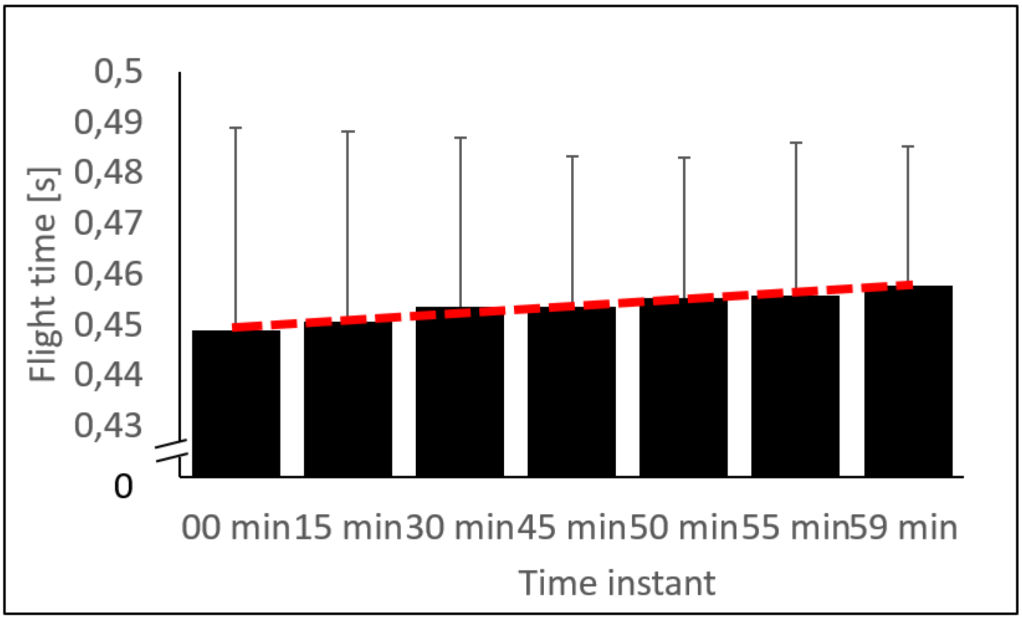

Figure 46: Flight time of steps of the right foot depicted for each investigated time point

for component 1 and 1.51 for component 2. Vibration data were summed up in 5 components after running the first PCA. After two more PCAs on the reduced data, three components were obtained, explaining 70.04% of the total variance. Component 1 had an eigenvalue of 5.32, component 2 of 2.64 and omponent 3 of 1.85. The component matrix shows the loading of each variable on the components (see tables 8 and 9).

Score plots were created to help interpret the effect of fatigue on the components obtained through PCA. An example of a score plot is shown in figure 47, representing the values obtained for EMG data. Similar plots were obtained for the other investigated data types. No clustering of score values was found in either of the plots. Therefore, no differentiation could be made between the seven time instants of measurements. Also, no score clustering could be identified in plots which only included the first and last measurement of the prolonged run (00 min and 59 min, respectively). A detailed presentation of all score plots can be found in the appendix of this manuscript (see appendix A).

Table 8: Component matrix showing the loadings of variables on a component. Results obtained through PCA of pressure, EMG and kinematics data collected during a prolonged run.

Pressure data		
	Component 1	
Max force	0.9	
Force integral	0.88	
Max force slope	0.61	
EMG data		
	Component 1	Component 2
Mean M. Gastr. med.	0.92	0.05
Mean M. Gastr. lat.	0.90	-0.02
Mean M. Soleus	0.89	0.19
Max M. Gastr. Med.	0.85	0.19
Max M. Gastr. Lat.	0.06	0.98
Max M. Soleus	0.16	0.96

	Component 1	Component 2
Stride frequency	-0.96	0.11
Flight time	0.95	-0.11
MaxPro	0.92	0.12
MaxProVel	-0.49	0.12
ROM	0.03	-0.88
TFPro	-0.11	0.86

Table 9: Component matrix showing the loadings of variables on a component. Results obtained through PCA of oscillation data collected at the Achilles tendon during a prolonged run.

Oscillations at the Achilles tendon			
	Component 1	Component 2	Component 3
High power component, distal, x-direction	0.93	-0.06	0.06
Dominant frequency, distal, x-direction	0.90	-0.08	0.09
High power component, proximal, x-direction	0.89	-0.11	0.04
Dominant frequency, proximal, x-direction	0.85	-0.05	0.03
Medium power component, distal, x-direction	0.80	0.09	0.35
Medium power component, proximal, x-direction	0.80	-0.02	0.47
ttmax, distal, yz-direction	0.16	0.87	-0.06
ttmax proximal, yz-direction	-0.08	0.83	0.06
ttmax, distal, x-direction	-0.03	0.76	-0.13
ttmax proximal, x-direction	-0.21	0.72	0.25
Low power component, proximal, x-direction	0.18	0.03	0.87
Low power component, distal, x-direction	0.30	0.03	0.85
Max acc, proximal, x-direction	0.08	0.06	-0.64
Dominant frequency, distal, yz-direction	0.29	0.11	0.37

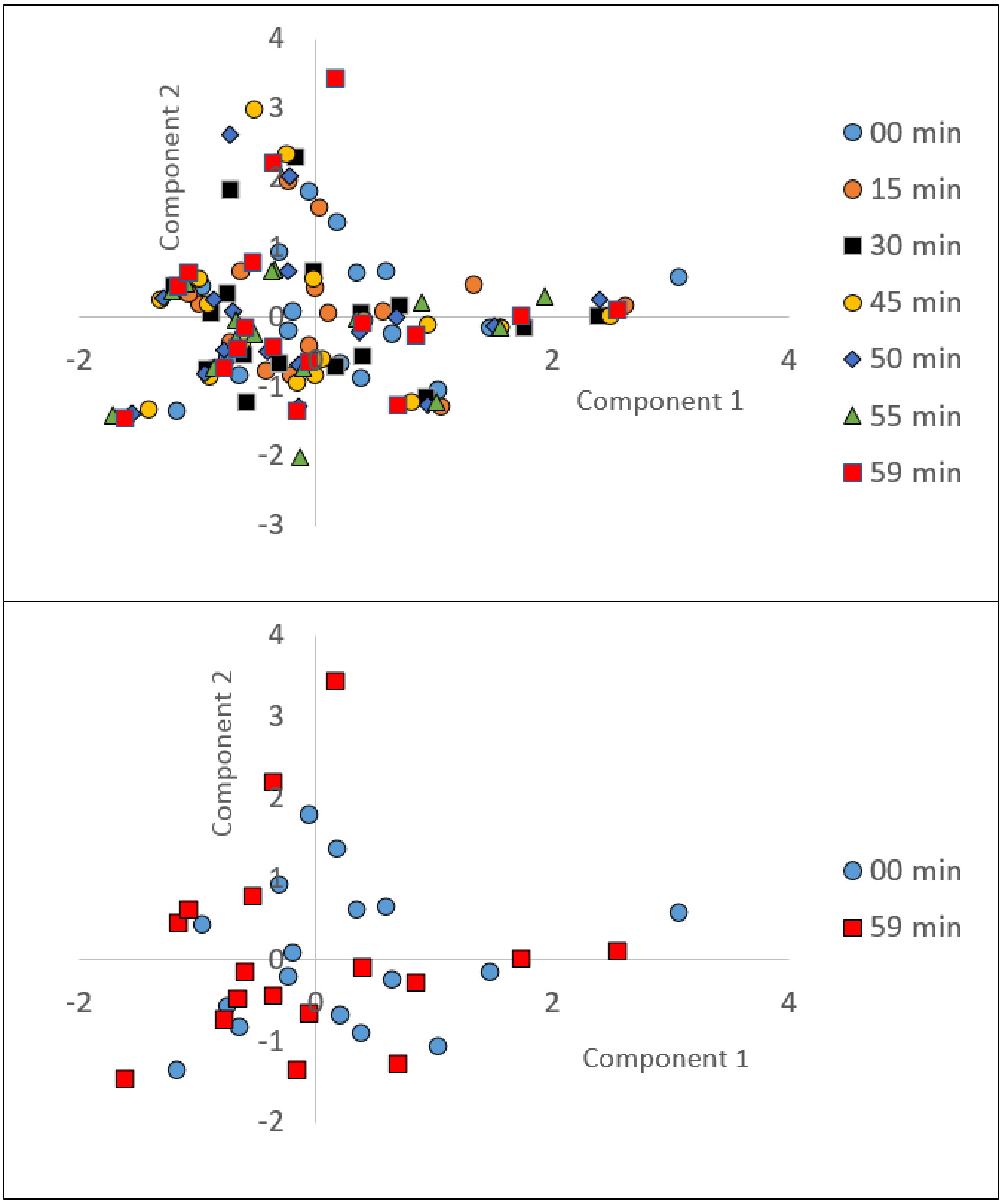

Figure 47: Representative score plot of the findings obtained through PCA. The plot shows the two components of EMG data with color coded score values.

10.4 DISCUSSION

The present study investigated exhaustion induced biomechanical changes during a one hour endurance run. The goal of the present study was to obtain parameter constellations of biomechanical data, which allow differentiation between stages of fatigue during prolonged running. The data was found to be suitable for PCA and principal components could be extracted from the multitude of variables. Therefore, PCA was found to be a useful tool to reduce the number of variables to a maximum of three components. Contrary to the expectations, no suitable variable constellation was determined to distinguish between early and late phases of the run. Visual inspection of the descriptive statistics of all variables shows little to no difference between time instants (see appendix A). Subjects may not have been sufficiently exhausted to allow a clear separation of the investigated time instants. All participants included in this study were experienced runners whose self-assessment of a fatiguing one-hour run should be a valid representation of a typical exhaustive training run. Flight time and step frequency of the participants were analyzed. Past research showed a decrease in stride frequency (Mizrahi et al., 2000) as well as an increase in flight time (Hanley & Mohan, 2014). Both of these findings were present in the collected data, therefore indicating increased fatigue during the one-hour run performed in this study (see figures 45 and 46). While runners participating in this study showed signs of fatigue, a more pronounced level of exhaustion may have been required in order to separate time instants on the basis of principal components. No quantitative measure was applied to objectively control the state of fatigue. Different levels of exhaustion may have occurred in the individual participants, possibly confounding the result. Future research should therefore include an objective measure of fatigue to ensure comparable stages of exhaustion in all study participants.

Runners may increase their flight time during a fatiguing treadmill run to let the treadmill pass underneath them while airborne.

This finding only seems to be present during runs performed over a longer time period. Fourchet et al. (2015) researched flight time during an exhausting run on a treadmill. They found an increase in time spent airborne. However, fatigue was not induced through a long duration of running but rather through high exercise intensity as participants ran at 95% of the velocity associated with their maximal oxygen uptake (VO_2max). Future research should consider the investigation of fatiguing endurance runs over ground. If prolongation of flight time is due to a more economic running style of letting the treadmill pass underneath the runner, different results might be expected during over ground running.

In the present study, data was found to be suitable for a PCA and met all inclusion criteria. However, no differentiation between time instants could be accomplished. Therefore, PCA may have been suitable for variable reduction but not to separate principal components between time instants. Whether this statistical tool remains futile when collecting data in a setting outside the lab during over ground running remains to be explored. It should be considered, that PCA, like many other statistical procedures, is recommended for large sample sizes. In the literature, references are either given as minimum total sample size, as ratio of subjects to number of items or as a combination of both. Comfrey & Lee (1992) describe sample sizes of up to 50 participants as *poor* while only sample sizes $\geqslant$ 1000 are expressed as *excellent*. Others found samples of $n = 50$ to be sufficient (Barrett & Kline, 1981). Osborne & Costello (2004) describe the disadvantages of small sample sizes with the possibility of extracting incorrect factors or assigning wrong items to a factor. Therefore, analysis of a larger cohort may achieve more distinct findings in the separation of time instants in score plots.

LIMITATIONS Runners with different proficiency levels were included in the present study and measurements took place during off season. Although study participants were asked to run at a speed which leads to exertion after one hour, the two above mentioned facts may have contributed to vague or diverse feelings of perceived exertion. Many study participants had difficulties choosing a speed and may have selected a relatively low running speed as they were limited in its variation. Prospective studies should be performed during running season and a current competition time should be used as a reference to set the expected running speed. Also, most participants were habitual road runners who have difficulties evaluating treadmill running speeds. A challenging test run on a treadmill several days before the actual measurements may help overcome this limitation. This way, a comparable fatigue level could be assumed for all participants and still allow a steady running speed.

CONCLUSION In conclusion, plantar pressure data, muscle activity, foot kinematics and the oscillation behavior at the Achilles tendon were studied during a fatiguing one-hour run. Signs of exhaustion could be identified in the collected data. Up to three principal components were found in the investigated variables of each data set. However, no differentiation between time instants could be made through visual inspection of score plots. Therefore, no characteristic factor loading occurred at the investigated time instants. It remains unknown whether the results obtained during prolonged treadmill running can be transferred to over ground running. Further research is required to clarify whether the findings of this study are applicable for road running or whether time instants may be differentiated in score plots if data is collected while running over ground as well as in larger cohorts.

11

COMPARISON OF RUNNERS WITH ACHILLES TENDON COMPLAINTS AND HEALTHY RUNNERS.

Table 10: Details of study 4, comparing runners with Achilles tendon complaints to healthy runners

Details of the Study	
N	8 ♂
Age [years]	34.1 ± 9.1
Height [cm]	178.75 ± 4.95
Weight [kg]	75.8 ± 10.7
Weekly running distance [km]	15.6 ± 15.7
Running experience [years]	7.0 ± 8.3
Persistence of complaints [years]	4.6 ± 7.6
Measurement systems	3D IMU
	Two 3D accelerometers
	Instrumented treadmill
	EMG
Conditions	Three types of footwear
	(Con1, Con2, NS)
	Comparison to healthy runners

11.1 INTRODUCTION

Achilles tendons are considered the thickest and strongest tendons in the human body (Scioli, 1994), yet they are one of the most common injury sights and male elite distance runners have a lifetime risk of 52% of sustaining Achilles tendinitis (Zafar et al., 2009). The cause of these overuse injuries is multifactorial, combining anatomic and biomechanical criteria as well as training habits. Several injury inducing factors were discussed in the literature (see chapter 2.2 and 2.3). While many researchers focused on individual outcome measures, only few considered a multitude of parameters and their interactions when studying injury inducing determinants (McCrory et al., 1999; Clement & Tauton, 1981; Di Caprio, Buda, Mosca, Calabrò & Giannini, 2010). However, no understanding of the etiological mechanisms of overuse injuries at the Achilles tendon was achieved. In a recent literature review Lorimer & Hume (2014) came to the conclusion that most of the proposed biomechanical parameters showed unclear results in their association with Achilles tendon injuries. The exploration of factors influencing the incidence of chronic overuse injuries at the Achilles tendon therefore still poses a challenge to medical and biomechanical sciences.

The biomechanical parameters described in this manuscript's previous chapters were found to be modifiable through extrinsic factors like running speed or type of footwear. They therefore seem to be characteristic for certain situations or parameter constellations. They might also differ in parameter constellations, to which the variabe *Achilles tendon complaints* is added. Biomechanics were found to differ in runners with Achilles tendon complaints compared to their healthy counterparts (McCrory et al., 1999; Smart et al., 1980; Clement & Tauton, 1981). If biomechanics of plantar pressure, EMG, foot kinematics and vibrations at the Achilles tendon are influenced by the runner's setting, they may also be influenced by physical complaints, such as pain. The goal of the present study was to investigate typical variations in biomechanical data, which can be associated with subjects who complain about Achilles tendon pain. To check for modifiability of the investigated parameters in a group of runners with Achilles tendon complaints, different shoe modifications were tested. A comparison of healthy runners and those with Achilles tendon complaints was performed using effect sizes. Therefore, the purpose of this study was to explore the effects of Achilles tendon complaints on plantar pressure, muscle activity, foot kinematics and oscillations at the Achilles tendon during running and to compare these data between runners with Achilles tendon complaints and their healthy counterparts.

11.2 METHODS

Eight runners with Achilles tendon complaints participated in this study. Subject characteristics are shown in table 10. Study participants were currently not under medical treatment due to their complaints and were therefore not considered patients. The experimental setup, data acquisition and data analysis were identical to those of study 3 and will therefore not be described in detail (see chapter 10.2). Participants were asked to run at the instrumented treadmill at a speed of 2.9 m/s while wearing the three shoe conditions described in chapter 9.2. In short, three types of footwear were investigated: Con1 with high arch support, medial wedges, heel wedges and soft damping material, Con2 with low arch support, no medial wedges and hard damping material and NS as a neutral reference shoe. The obtained data was then compared to data collected in healthy subjects while wearing identical footwear. For that purpose, the data sets acquired in study 2 with subjects running on a treadmill were used for comparison.

Effect sizes are a a useful tool to describe the magnitude of an effect and to help decide whether this effect is of meaning (Cohen, 1988). It's a measurement of the magnitude of differences between two mean values. Contrary to many other statistical tools, effect sizes are independent of sample size and can be estimated for groups of different sizes. The effect size d_s is defined as the difference between the two group means, divided by the pooled standard deviation, as described by Rosenthal & Rubin (1986). To calculate d_s the following equation was used in the present study:

$$d_s = \frac{Mean_c - Mean_h}{SD_p} \tag{9}$$

where $Mean_c$ is the mean value of the investigated parameter in the complaint group, $Mean_h$ is the mean value in healthy subjects and SD_p is the pooled SD. SD_p is determined as the square root of the average SD:

$$SD_p = \sqrt{\frac{(n_c - 1) * SD_c^2 + (n_h - 1) * SD_h^2}{n_c + n_h - 2}} \tag{10}$$

with n_c describing the number of subjects in the complaint group, SD_c the standard deviation in that group, n_h representing the number of healthy subjects and SD_h quantifying the standard deviation of the investigated parameter in the healthy group. In this study, Cohen's d_s values are specified as measures of effect sizes. Small effect sizes are represented by $0.2 \leqslant d_s \leqslant 0.5$, moderate effect sizes by $0.5 \leqslant d_s \leqslant 0.8$ and large effect sizes by $d_s \geqslant 0.8$ (Cohen, 1988). The 90% confidence interval (CI) around effect size d_s was chosen as it represents 100*(1-2α)% = 90%. The traditional α-level in biomechanics when testing for significant differences is set at 0.5. If the 90% CI excludes the value zero, the null hypothesis can be rejected. According to Steiger (2004), this hypothesis test is equivalent to the F-test performed during ANOVA.

11.3 RESULTS

11.3.1 *Plantar Pressure Distribution*

Achilles tendon complaints had large effects on total plantar force detected either in shoe (healthy subjects) or between outer sole and treadmill surface (subjects with Achilles tendon complaints). All d_s values were well above one. 90% CI indicate significant F-values between the two groups as non of the intervals span zero. On average, the complaint group showed lower total force values compared to the healthy group. Results are shown in table 11.

Table 11: Comparison of average total force between heal-thy runners and subjects with Achilles tendon complaints.

Total Force [N]				
	Healthy	Complaint	d_s	90% CI
Con1	1635 ± 269	1174 ± 166	1.9	[1.08:2.68]
Con2	1589 ± 354	1268 ± 158	1.0	[0.30:1.75]
NS	1645 ± 356	1196 ± 171	1.4	[0.66:2.17]

Table 12: Comparison of average activity of the Mm. Gastrocnemii between healthy runners and subjects with Achilles tendon complaints.

Average Activity M. Gastrocnemius medialis [µV]				
	Healthy	Complaint	d_s	90% CI
Con1	165 ± 57	86 ± 24	1.6	[0.79:2.33]
Con2	158 ± 50	92 ± 24	1.5	[0.71:2.23]
NS	152 ± 46	91 ± 27	1.5	[0.71:2.23]
Average Activity M. Gastrocnemius lateralis [µV]				
	Healthy	Complaint	d_s	90% CI
Con1	122 ± 28	76 ± 18	1.6	[0.87:2.42]
Con2	121 ± 24	82 ± 20	1.7	[0.90:2.47]
NS	121 ± 32	80 ± 18	1.4	[0.66:2.17]

11.3.2 *Muscle Activity*

Large effects were found for Achilles tendon complaints on average and maximum muscle activity of the Mm. Gastrocnemii. 90%CI of all d_s do not include zero. Therefore, F-tests between the two groups are statistically significant. The groups show differences in average and maximum activity of the Mm. Gastrocnemii in all three shoe conditions (see tables 12 and 13). While subjects with Achilles tendon complaints showed lower levels of average activation levels than their healthy counterparts, maximum activity was higher in the complaint group.

A comparison between subjects with Achilles tendon complaints and healthy subjects could not be performed for the M. Soleus. This muscle was not included in data acquisitions of healthy runners. The average activity of the M. Soleus in subjects with Achilles tendon complaints was 66.86 ± 19.60 when running in Con1, 68.70 ± 18.48 in Con2 and 69.19 ± 22.56 in NS. Maximum activity of the M. Soleus in this group was 1405.13 ± 376.95 in Con1, 1319.13 ± 354.05 in Con2 and 1288.00 ± 408.10 in NS.

Table 13: Comparison of maximum activity of the Mm. Gastrocnemii between healthy runners and subjects with Achilles tendon complaints.

Maximum Activity M. Gastrocnemius medialis [μV]				
	Healthy	Complaint	d_s	90% CI
Con1	867 ± 279	1640 ± 421	-2.4	[-3.26:-1.52]
Con2	800 ± 269	1653 ± 325	-3.0	[-3.94:-2.04]
NS	850 ± 344	1698 ± 453	-2.3	[-3.10:-1.40]
Maximum Activity M. Gastrocnemius lateralis [μV]				
	Healthy	Complaint	d_s	90% CI
Con1	646 ± 230	1623 ± 709	-2.4	[-3.21:-1.49]
Con2	675 ± 236	1494 ± 251	-3.4	[-4.42:-2.39]
NS	618 ± 250	1433 ± 362	-2.9	[-3.79:-1.93]

11.3.3 *Foot Kinematics*

Medium to large effect sizes could be proven for Achilles tendon complaints on foot kinematics. Effect sizes were large for MaxPro, MaxProVel and ROM in all three types of footwear used in this study (see table 14). Healthy subjects showed decreased MaxPro, MaxProVel and ROM compared to the complaint group. Achilles tendon complaints had medium effects on TFPro while running in either Con2 or NS. These comparisons also showed 90% CIs which span zero. Therefore, no significant F-values between the two groups can be assumed. However, a large effect size was found for this parameter when running in Con1 with a 90% CI excluding the value zero, indicating significant F-values. TFPro was higher in the group of healthy subjects compared to their counterparts with Achilles tendon complaints. Therefore, pronation was finalized at a later time instant in healthy subjects.

Table 14: Comparison of foot kinematics in the frontal plane between healthy runners and subjects with Achilles tendon complaints.

Average MaxPro [°]				
	Healthy	Complaint	d_s	90% CI
Con1	-0.1 ± 1.5	-6.4 ± 4.5	2.4	[1.50:3.23]
Con2	-3.4 ± 6.1	-8.6 ± 4.3	0.9	[0.20:1.63]
NS	-2.3 ± 2.8	-10.4 ± 4.6	2.4	[1.53:3.26]
Average MaxProVel [°/s]]				
	Healthy	Complaint	d_s	90% CI
Con1	-65.9 ± 34.6	-197.6 ± 61.8	3.0	[2.06:3.98]
Con2	-87.1 ± 43.1	-355.3 ± 238.0	2.1	[1.26:2.91]
NS	-123.6 ± 43.8	-339.8 ± 204.5	1.9	[1.12:2.72]
Average ROM [°]				
	Healthy	Complaint	d_s	90% CI
Con1	7.5 ± 15.2	27.1 ± 6.4	-1.5	[-2.22:-0.70]
Con2	9.6 ± 4.7	30.5 ± 7.8	-3.7	[-4.72:-2.61]
NS	10.2 ± 3.9	31.1 ± 7.8	-3.0	[-3.95:-2.05]
Average TFPro [ms]				
	Healthy	Complaint	d_s	90% CI
Con1	90.6 ± 12.6	76.0 ± 14.5	1.1	[0.38:1.84]
Con2	77.5 ± 19.3	68.4 ± 7.8	0.5	[-0.16:1.23]
NS	75.2 ± 14.0	66.9 ± 6.8	0.7	[-0.04:1.37]

11.3.4 *Oscillations at the Achilles tendon*

TIME SPACE Small to medium effect sizes could be proven for Achilles tendon complaints on MaxAcc at the distal accelerometer while no effects or small effects were found for these parameters at the proximal sensor (see tables 15 and 16). Besides one exception (horizontal horizontal yz-plane, Con1, distal accelerometer), all 90% CIs span zero and therefore reflect non-significant F-values. For ttmax, large effect sizes were found for Achilles tendon complaints in all measurement directions at the proximal accelerometer while they were medium to large at the distal sensor. At the proximal accelerometer, non of the 90% CIs span zero, therefore proving significant F-values. However, the 90% CIs for ttmax at the distal accelerometer reflect non-significant F-values when running in Con2 only.

FREQUENCY SPACE The dominant frequency of oscillations at the Achilles tendon was lower in healthy subjects than in those with Achilles tendon complaints. Large effect sizes were found for Achilles tendon complaints on the dominant frequency for all conditions besides measurements performed with the distal accelerometer in vertical x-direction whith subjects wearing NS. For this condition, a medium effect size was proven with a 90% CI spanning zero. All comparisons of dominant frequencies are presented in table 17. Signal power in the low frequency interval was decreased in the complaint group compared to the healthy group besides measurements in the horizontal yz-plane performed with the distal accelerometer. In this measurement condition, small to medium effect sizes were found. In Con1 and NS, healthy subjects showed lower signal power while it was higher when running in Con2. This finding indicates different effects of footwear in healthy runners compared to subjects with Achilles tendon complaints. A detailed description of the results of signal power in the low frequency interval can be seen in table 18. In the medium and high frequency intervals a different picture was apparent compared to the low frequency interval. In all conditions, subjects with Achilles tendon complaints experienced considerably higher signal power than their healthy counterparts. This is represented not only by large mean differences but also by large effect sizes and 90% CIs which do not span zero and therefore indicate significant F-values between the two groups. Details of the statistical results are given in table 19 and table 20. An inverse behavior can be seen in the signal power of the highest frequency interval. Like in the low frequency interval, healthy subjects show lower signal power than those in the complaint group. One exception is found for measurements performed with the distal accelerometer in vertical x-direction. Here, healthy subjects experienced lower signal power. It should be noted though, that all effect sizes indicated small to intermediate effects in the highest frequency interval and that all of the 90% CIs span zero. Therefore, F-values between the two groups are not significant. All results of the signal power analysis in the highest frequency interval can be seen from table 21.

Table 15: Comparison of oscillation parameters in time space between healthy runners and subjects with Achilles tendon complaints. Data was obtained with the proximal accelerometer.

MaxAcc vertical vertical x-direction [m/s^2]				
	Healthy	Complaint	d_s	90% CI
Con1	62 ± 11	63 ± 7	-0.1	[-0.80:0.57]
Con2	64 ± 13	65 ± 7	-0.1	[-0.79:0.58]
NS	62 ± 17	65 ± 7	-0.2	[-0.91:0.47]
MaxAcc horizontal horizontal yz-plane [m/s^2]				
	Healthy	Complaint	d_s	90% CI
Con1	65 ± 17	59 ± 10	0.4	[-0.34:1.05]
Con2	61 ± 20	62 ± 13	-0.1	[-0.74:0.64]
NS	64 ± 21	64 ± 11	0.0	[-0.68:0.69]
ttmax vertical vertical x-direction [ms]				
	Healthy	Complaint	d_s	90% CI
Con1	104 ± 54	56 ± 20	1.0	[0.28:1.73]
Con2	124 ± 97	47 ± 22	0.9	[0.20:1.63]
NS	126 ± 72	47 ± 22	1.3	[0.53:2.01]
ttmax horizontal horizontal yz-plane [ms]				
	Healthy	Complaint	d_s	90% CI
Con1	121 ± 68	61 ± 21	1.0	[0.29:1.73]
Con2	123 ± 94	56 ± 24	0.8	[0.11:1.54]
NS	123 ± 64	50 ± 17	1.3	[0.59:2.08]

Table 16: Comparison of oscillation parameters in time space between healthy runners and subjects with Achilles tendon complaints. Data was obtained with the distal accelerometer.

MaxAcc vertical vertical x-direction [m/s^2]				
	Healthy	Complaint	d_s	90% CI
Con1	62 ± 9	64 ± 6	-0.3	[-0.95:0.43]
Con2	59 ± 11	64 ± 6	-0.5	[-1.19:0.20]
NS	59 ± 13	64 ± 6	-0.5	[-1.20:0.20]
MaxAcc horizontal horizontal yz-plane [m/s^2]				
	Healthy	Complaint	d_s	90% CI
Con1	68 ± 10	56 ± 14	1.1	[0.39:1.85]
Con2	61 ± 15	56 ± 16	0.4	[-0.34:1.05]
NS	66 ± 17	60 ± 13	0.4	[-0.32:1.06]
ttmax vertical vertical x-direction [ms]				
	Healthy	Complaint	d_s	90% CI
Con1	98 ± 42	56 ± 21	1.1	[0.38:1.84]
Con2	113 ± 110	48 ± 22	0.7	[-0.02:1.39]
NS	128 ± 86	46 ± 15	1.1	[0.37:1.83]
ttmax horizontal horizontal yz-plane [ms]				
	Healthy	Complaint	d_s	90% CI
Con1	96 ± 43	41 ± 11	1.5	[0.71:2.23]
Con2	99 ± 100	42 ± 15	0.7	[-0.04:1.37]
NS	105 ± 58	43 ± 13	1.3	[0.51:2.00]

Table 17: Comparison of dominant frequencies between heal-thy runners and subjects with Achilles tendon complaints.

Dominant frequency vertical x-direction, distal accelerometer [Hz]				
	Healthy	Complaint	d_s	90% CI
Con1	25 ± 9	39 ± 15	-1.3	[-1.99:-0.51]
Con2	22 ± 12	32 ± 5	-0.9	[-1.60:-0.17]
NS	29 ± 15	36 ± 11	-0.5	[-1.16:0.24]
Dominant frequency horizontal yz-plane, distal accelerometer [Hz]				
	Healthy	Complaint	d_s	90% CI
Con1	33 ± 16	72 ± 23	-2.2	[-3.07:-1.38]
Con2	31 ± 20	75 ± 19	-2.3	[-3.15:-1.44]
NS	39 ± 28	57 ± 29	-0.6	[-1.34:0.07]
Dominant frequency vertical x-direction, proximal accelerometer [Hz]				
	Healthy	Complaint	d_s	90% CI
Con1	21 ± 8	31 ± 14	-1.0	[-1.75:-0.30]
Con2	23 ± 12	36 ± 11	-1.1	[-1.80:-0.34]
NS	28 ± 14	41 ± 17	-0.9	[-1.62:-0.18]
Dominant frequency horizontal yz-plane, proximal accelerometer [Hz]				
	Healthy	Complaint	d_s	90% CI
Con1	32 ± 14	57 ± 29	-1.3	[-2.06:-0.57]
Con2	29 ± 21	62 ± 25	-1.5	[-2.27:-0.74]
NS	40 ± 26	69 ± 22	-1.2	[-1.89:-0.42]

Table 18: Comparison of power in the low frequency interval between healthy runners and subjects with Achilles tendon complaints.

Low frequency interval vertical x-direction, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	1.0 ± 0.1	0.9 ± 0.2	0.7	[0.02:1.44]
Con2	1.0 ± 0.1	0.9 ± 0.2	0.7	[-0.02:1.39]
NS	0.9 ± 0.2	0.9 ± 0.2	0.2	[-0.47:0.91]
Low frequency interval horizontal yz-plane, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.9 ± 0.2	1.0 ± 0.3	-0.4	[-1.07:0.32]
Con2	0.9 ± 0.3	0.9 ± 0.2	0.1	[-0.57:0.81]
NS	0.9 ± 0.3	1.0 ± 0.2	-0.2	[-0.85:0.53]
Low frequency interval vertical x-direction, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	1.0 ± 0.1	0.9 ± 0.2	1.2	[0.44:1.91]
Con2	1.0 ± 0.1	0.9 ± 0.2	1.6	[0.82:2.36]
NS	1.0 ± 0.1	0.8 ± 0.2	0.8	[0.11:1.53]
Low frequency interval horizontal yz-plane, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.9 ± 0.2	0.9 ± 0.1	0.0	[-0.69:0.69]
Con2	0.9 ± 0.2	0.8 ± 0.2	0.3	[-0.36:1.03]
NS	0.8 ± 0.2	0.8 ± 0.2	0.2	[-0.52:0.86]

Table 19: Comparison of power in the medium frequency interval between healthy runners and subjects with Achilles tendon complaints.

Medium frequency interval vertical x-direction, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.9 ± 0.2	1.7 ± 0.6	-2.6	[-3.44:-1.67]
Con2	0.8 ± 0.2	1.8 ± 0.63	-2.7	[-3.65:-1.82]
NS	0.9 ± 0.3	1.8 ± 0.5	-2.5	[-3.40:-1.63]
Medium frequency interval horizontal yz-plane, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	1.0 ± 0.2	1.4 ± 0.6	-1.4	[-2.14:-0.64]
Con2	0.9 ± 0.2	1.5 ± 0.5	-2.2	[-2.98:-1.31]
NS	0.9 ± 0.2	1.5 ± 0.6	-2.1	[-2.87:-1.22]
Medium frequency interval vertical x-direction, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.8 ± 0.2	1.6 ± 0.5	-2.1	[-2.87:-1.22]
Con2	0.9 ± 0.2	1.7 ± 0.6	-2.7	[-3.55:-1.74]
NS	0.9 ± 0.2	1.7 ± 0.5	-2.4	[-3.31:-1.56]
Medium frequency interval horizontal yz-plane, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	1.0 ± 0.1	1.6 ± 0.5	-2.3	[-3.13:-1.43]
Con2	0.8 ± 0.2	1.6 ± 0.5	-2.6	[-3.54:-1.74]
NS	0.8 ± 0.2	1.5 ± 0.6	-2.4	[-3.27:-1.53]

Table 20: Comparison of power in the high frequency interval between healthy runners and subjects with Achilles tendon complaints.

	Healthy	Complaint	d_s	90% CI
High frequency interval vertical x-direction, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.4 ± 0.2	1.1 ± 0.6	-2.1	[-2.95:-1.29]
Con2	0.5 ± 0.2	1.1 ± 0.5	-2.1	[-2.92:-1.26]
NS	0.6 ± 0.2	1.2 ± 0.5	-2.1	[-2.93:-1.27]
High frequency interval horizontal yz-plane, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.7 ± 0.1	1.6 ± 0.5	-3.4	[-4.42:-2.29]
Con2	0.7 ± 0.2	1.5 ± 0.4	-2.7	[-3.64:-1.81]
NS	0.8 ± 0.2	1.5 ± 0.5	-2.5	[-3.42:-1.65]
High frequency interval vertical x-direction, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.4 ± 0.1	1.2 ± 0.6	-2.3	[-3.19:-1.48]
Con2	0.4 ± 0.2	1.1 ± 0.5	-2.3	[-3.14:-1.44]
NS	0.5 ± 0.2	1.2 ± 0.6	-2.0	[-2.78:-1.15]
High frequency interval horizontal yz-plane, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.7 ± 0.1	1.5 ± 0.5	-3.0	[-3.93:-2.03]
Con2	0.7 ± 0.2	1.4 ± 0.5	-2.6	[-3.46:-1.68]
NS	0.7 ± 0.2	1.5 ± 0.4	-2.9	[-3.85:-1.97]

Table 21: Comparison of power in the highest frequency interval between healthy runners and subjects with Achilles tendon complaints.

Highest frequency interval vertical x-direction, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.1 ± 0.0	0.1 ± 0.0	0.0	[-0.69:0.69]
Con2	0.1 ± 0.0	0.1 ± 0.0	0.3	[-0.42:0.96]
NS	0.1 ± 0.0	0.1 ± 0.0	0.7	[-0.04:1.37]
Highest frequency interval horizontal yz-plane, distal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.1 ± 0.0	0.1 ± 0.0	-0.3	[-0.99:0.39]
Con2	0.1 ± 0.1	0.1 ± 0.0	0.1	[-0.59:0.79]
NS	0.1 ± 0.2	0.1 ± 0.0	0.1	[-0.62:0.76]
Highest frequency interval vertical x-direction, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.1 ± 0.2	0.1 ± 0.0	0.3	[-0.42:0.96]
Con2	0.1 ± 0.1	0.1 ± 0.0	0.3	[-0.44:0.94]
NS	0.1 ± 0.1	0.1 ± 0.0	0.2	[-0.51:0.87]
Highest frequency interval horizontal yz-plane, proximal accelerometer				
	Healthy	Complaint	d_s	90% CI
Con1	0.1 ± 0.1	0.1 ± 0.0	0.3	[-0.44:0.94]
Con2	0.1 ± 0.1	0.1 ± 0.0	0.3	[-0.43:0.96]
NS	0.2 ± 0.2	0.1 ± 0.0	0.4	[-0.34:1.05]

11.4 DISCUSSION

The purpose of this study was to explore the effects of Achilles tendon complaints on plantar pressure, muscle activity, foot kinematics and oscillations at the Achilles tendon during running and to compare these data between runners with Achilles tendon complaints and their healthy counterparts. Large effects of Achilles tendon complaints were found on plantar force, muscle activity, foot kinematics and vibrations at the Achilles tendon. Total force underneath the foot was lower in the complaint group compared to healthy study participants. This result was verified in all shoe conditions. However, effect sizes differed between footwear types. While McCrory et al. (1999) analyzed peak ground reaction forces and found them to be higher in injured runners, no peak forces were included in the analysis of the present study. Calculations of peak forces in the F-Scan measurement system clearly differ from those obtained using the instrumented treadmill. It was therefore decided not to include this parameter. Total force may have been higher in healthy runners due to longer durations of stance phase or differences in body weight. The latter was higher in healthy runners. Body weight has been discussed in the literature as an injury inducing factor for Achilles tendon complaints. However, diverse findings exist. When comparing the groups studied by Donoghue, Harrison, Laxton & Jones (2008) an effect size of 0.62 $\pm$ 4.31 can be found for body weight on Achilles tendon complaints while in the groups studied by Ryan, Grau, Krauss, Maiwald, Taunton & Horstmann (2009), a diverse effect of -0.62 $\pm$ 2.96 was present. The results on total force presented in this study should be considered with caution as different measurement devices were used to collect them. In healthy subjects, an in-shoe measurement system was used while an instrumented treadmill was utilized for participants with Achilles tendon complaints. Therefore, the effects found in total force may have been confounded by unequal measurement devices.

Lersch et al. (2012) studied the influence of calf muscle forces and calcaneus positions on Achilles tendon straints in cadaveric lower leg congeries. They found calcaneus position in the frontal plane to be determining up to 15% of intratendinous strain differences, therefore emphasizing the importance of rearfoot eversion for Achilles tendon loading. Our findings are in agreement with those results from Lersch et al. (2012) as healthy runners demonstrated lower MaxPro, MaxProVel and ROM in the frontal plane. Average TFPro was shorter in subjects with Achilles tendon complaints. Other research groups have stated similar results when comparing runners with Achilles tendon injuries to their symptom-free counterparts. Donoghue et al. (2008) reasoned calcaneal angle and calcaneal ROM to be higher in runners with Achilles tendon complaints when running either barefoot or shod. Similar conclusions were drawn by Ryan et al. (2009) who found ankle eversion-inversion ROM and ankle eversion velocity to be lower in healthy runners. Donoghue, Harrison, Coffey & Hayes (2008) were able to differentiate healthy runners and those with chronic Achilles tendon complaints based on the kinematics of their lower extremity using PCA. They found runners with Achilles tendon injuries to experience greater foot eversion compared to healthy runners. The different types of footwear examined in the present study seem to meet their intended goals. For both groups, differences can be seen in all kinematic parameters between Con1, Con2 and NS. Con1 can be described as a shoe configuration aiming at motion control. MaxPro, MaxProVel and average ROM were lowest in trials during which subjects ran in Con1. Also, TFPro was prolonged when wearing this shoe. It can therefore be concluded, that running shoes influence kinematic variables in the intended direction not only in healthy runners but also in subjects with Achilles tendon complaints.

On average, healthy runners activated their Mm. Gastrocnemii more than subjects with Achilles tendon complaints while they showed much smaller maximum activation. These findings are in agreement with the results of Azevedo et al. (2009) who found higher integrated EMG values of the M. Gastrocnemius in runners with Achilles tendinopathy compared to healthy runners. If runners experience pain at the Achilles tendon, they might lower their average activity of calf muscles to reduce traction at the impaired structure. However, higher maximum activation levels might be required in this group to influence oscillation of the tendon. Runners with Achilles tendon complaints do not just show higher maximum activity of the Mm. Gastrocnemii but they also show a much shorter ttmax. To decrease the time period between heel contact and maximum accelerations at the Achilles tendon, higher levels of muscle activity may be required while the muscle is in a more relaxed state during the rest of stance, leading to smaller average activity.

If considering the dominant frequency of oscillations at the Achilles tendon, differences in effect sizes can be noticed between shoe conditions. While the dominant frequency is always higher in the complaint group, smallest group differences are found when running in NS compared to the two other shoe conditions. Especially at the distal accelerometer, the dominant frequency was highest in healthy runners when running in NS while it was medium or lowest compared to the other shoe conditions in the complaint group. In runners with Achilles tendon complaints a higher dominant frequency at the distal part of the tendon was determined when running in shoes specifically constructed for running (Con1 and Con2) compared to a non-specific, all purpose shoe (NS). The results were vice versa at the proximal measurement site with higher dominant frequencies in NS compared to the other two shoe conditions. An implication of this result may be, that runners with complaints at the distal Achilles tendon should wear a shoe without arch support, no medial wedges and a very hard damping material like NS. This shoe resulted in the lowest average dominant frequency, therefore achieving results that are closest to those of healthy runners. However if runners experience complaints at the distal Achilles tendon, shoes with high arch support, medial wedges, heel wedges and soft damping material (Con1) should be worn as they resulted in lowest dominant oscillation frequencies.

For most conditions, Achilles tendon complaints showed small to medium effects on average signal power in the investigated frequency intervals. Signal power in the highest frequency band was rather small in both groups with the 90% CI overlapping the value zero. It strongly increased in the high frequency interval of the complaint group. Effect sizes of up to -3.4 were found in this frequency band with healthy subjects consistently experiencing lower power at their Achilles tendon. In general, average power was higher in the complaint group in all frequency intervals except for low frequency. It can therefore be concluded, that a shift in signal power towards higher oscillation frequencies is present in runners with Achilles tendon complaints.

While differences between healthy runners and those with Achilles tendon complaints could be proven, limitations of the methods used when performing this study should be considered. While data collection and data analysis were identical for EMG, kinematic data and oscillations at the Achilles tendon, different measurement and analysis procedures were applied when evaluating plantar pressure. These deviations may have contributed to the large effect sizes found between the two groups. EMG was performed in a similar manner but no normalization took place as only within-subject comparisons were originally intended. The decision of between-subject comparison was taken post-hoc. Subsequent normalization of the data to its mean or peak values was refused as these methods are highly under debate (Halaki & Ginn, 2012). While normalization of EMG data to peak or mean activation levels obtained during the investigated task has been used in the past (Burden, 2010; Yang & Winter, 1984), more critical considerations were made recently. Although these methods decrease variability in the data and therefore have beneficial effects on statistical outcomes, different individuals may always display diverse muscle activation patterns and control strategies to achieve the same motion. Also, reliability of this method between days was shown to be low even within the same individuals (Knutson, Soderberg, Ballantyne & Clarke, 1994). Therefore, large variations may be present in the activation level during a reference contraction. In a review article, Halaki & Ginn (2012) conclude these methods to be invalid for normalization of EMG data and state the utilization of these normalizations only to be acceptable if patterns of muscle activation are to be compared.

The presence of pain and discomfort at the Achilles tendon was an inclusion criterion for study participants in the complaint group. All subjects had seen a physician because of their complaints but were not transferred with a distinct clinical diagnosis. Therefore, most of the participants had been diagnosed with tendinopathy in the past without further specification. However, differentiations of these symptoms can be classified further depending on the affected region. Mayer et al. (2000) distinguish between Achillotendinopathy at four different locations: clinical changes at the Achilles tendon itself, pathologies of the peritendineum, complaints at the calcaneal insertion point or pain at the bursa between tendon and calcaneus. Besides classifications based on the location, differentiations can be made with regard to the type of injury: inflammations at the Achilles tendon (tendinitis), degenerative changes of the tendon (tendinosis) and inflammations or degenerative changes of the peritendineum (peritendinitis/peritendinosis). These specifications were not considered in the current study as a broader approach was chosen. Future studies examining this question should group subjects according to injury location and type of injury. Close cooperations with local physicians or sports clinics should be facilitated to ensure specific clinical diagnoses of the subject's current condition.

LIMITATIONS Plantar pressure data of healthy subjects and those with Achilles tendon complaints were performed not only with two different measurement systems but strictly speaking with two different measurement techniques. In healthy runners, plantar pressure was quantified in-shoe while it was recorded outside the shoe in study participants with Achilles tendon complaints. Comparison of EMG data between subjects is a precarious procedure if no normalization of the measurements took place. Maximum voluntary contractions are frequently used for that purpose but come with limitations regarding the ability of study participants to accomplish maximum contractions. Future research should yet include this normalization procedure including a superimposed electronic twitch to avoid issues regarding the voluntary ability to contact a muscle. Running shoe characteristics may change over time as cushioning material may fatigue and stabilizing components may wear out. We compared trials of two different measurement time points, which were conducted at an interval of about one year using identical test shoes. It can not be excluded that changes in running shoe characteristics appeared over that period of time. These changes may have increased the effect sizes found between healthy runners and those with Achilles tendon complaints.

CONCLUSION In conclusion, first insight was gained in differences between healthy runners and study participants with Achilles tendon complaints. Medium to large effect sizes were found for most biomechanical parameters when comparing the two groups. Therefore, the two groups could be differentiated based on these results. Future studies should include larger sample sizes to allow better transfer of these results to the general population. Future work should also consider the timing of the M. Gastrocnemius lateralis and the M. Gastrocnemius medialis as unbalanced contractions are discussed to increase stresses at the Achilles tendon. Also, as the parameters analyzed in this study showed promising results when differentiating between healthy runners and the complaint group, it should be investigated whether a distinction can be made before an Achilles tendon injury occurs. Therefore, runners who are at risk of sustaining such an injury might be identified before hand and preventive measures may be applied at early stages.

Part V

SUMMARY AND CONCLUSION

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SUMMARY AND CONCLUSION

The purpose of the thesis at hand was the implementation of a simultaneous use of several measurement systems and the detection of disadvantageous parameter constellations, which allow the differentiation between runners with Achilles tendon complaints and their healthy counterparts. Both aims were achieved and fundamental knowledge regarding oscillations at the Achilles tendon could be generated. A continuous advancement in data analysis was reached throughout the series of measurements to obtain clear and comprehensible results. This development specifically becomes apparent in the comparison of the results obtained during the first and the last study (chapter 8 versus chapter 11). With regard to oscillations presented in this thesis, an important concern is the limited transferability of the obtained results to the vibration behavior of the Achilles tendon itself. Data was collected on the skin overlaying the Achilles tendon, not at the tendon itself. Similar limitations are known from marker based kinematic analysis for which reflective markers are mounted on the skin. Skin movement artifacts influence the results and may even cause misleading interpretations. The only experimental method, which could quantify the reliability of oscillations measured with accelerometers on the runner's skin, is a comparison to oscillations measured on the tendon itself. Similar comparisons were made in kinematic analysis with skin-mounted markers in comparison to bone-mounted markers. In measurements concerning oscillations of the Achilles tendon, only a comparison between tendon-mounted accelerometers to skin-mounted accelerometers would allow conclusions regarding the validity of the measurement procedures presented here. However, the option of such a validation was abandoned as it would require highly complex cadaver studies and computer simulations to confirm the findings. It was therefore decided to exclude these specific validations and follow similar procedures as in all measurements performed on soft tissue vibrations in the past.

The variation of external factors like running speed, footwear and running ground led to modifications of biomechanical parameters assumed to play an important role in the development of chronic overuse injuries. Finding opportunities to influence these biomechanical variables is of crucial importance as it allows non-invasive, unproblematic and implementable corrective action and control of these parameters. In the present work, most experimental settings required subjects to run on a treadmill. However, long distance runners almost solely train outdoors on a variety of grounds. Especially uneven terrain like forest floor may lead to muscle activation patterns deviant from those of treadmill running. Motor control may be challenged differently as stepping on roots or rocks causes an unstable base of support. Therefore, fatiguing effects may also differ or cause more pronounced changes in running biomechanics. A measurement system, which is integrated into the running shoe would allow data acquisition in real life situations without the constraints of a laboratory setting. Therefore, an integration of the sensors applied in the series of measurements of this thesis would resemble an important improvement of the experimental conditions. Deviating results can not be excluded if runners are studied while running outdoors and an effect of different terrains on the evaluated parameters can not be ruled out.

Subjects suffering from complaints at the Achilles tendon varied clearly in the investigated parameters from healthy runners. These differences were either more or less pronounced depending on the type of footwear in which the study participants ran. However, no distinct conclusion can yet be drawn as to which shoe type modifies the parameters in the complaint group to a comparable quantity of healthy runners. Running shoe research should therefore consider the above mentioned findings as well as individual differences in runners. Different shoe types might be required for subjects suffering from Achilles tendon pain and those who are free of complaints at this structure. Longitudinal studies are required to observe the long-term effects of permanent use of the modified shoes in a large cohort. Implications may change if runners frequently wear the shoes. Also, wearing comfort should be considered in future work as uncomfortable shoes may lead to other complaints and are not expected to be gladly worn by the runner. However, at the current state of knowledge, no explicit recommendations can be made with regard to the construction of these shoes.

Also, further research should focus on clarifying weather the differences found between the investigated complaint group and healthy runners are a cause or an effect of Achilles tendon complaints. While a differentiation between the complaint group and healthy runners was possible with the parameter constellation compiled in the thesis at hand, it currently remains unknown if subjects who are at risk for overuse injuries at the Achilles tendon can be discriminated from other runners through this parameter constellation. Therefore, a larger prospective study should be conducted on a cohort of habitual runners without any physical complaints. The parameter constellation proposed in the present work should be collected from the study participants and a follow up regarding Achilles tendon complaints should be executed. Thus, the predictive capacity of the parameters at hand could be evaluated.

Part VI

APPENDIX

A DETAILED STATISTICAL RESULTS

A.1 Descriptive statistics chapter 8

A.1.1 *Statistical comparisons between step phases*

MUSCLE ACTIVITY M. Gstr. Lat.: StancePhase had a significant main effect on average activation of the M. Gastr. Lat. ($F(2.37, 44.95) = 38.26$, $p < 0.01$). Activation differed significantly between all StancePhases ($p < 0.05$) besides the comparisons between BCP and FF-POP ($p = 0.14$) and between TCT and FFCP ($p = 0.19$). Figure 48 demonstrates the comparisons between average activation levels of the M. Gastr. Lat. during different StancePhases. A significant interaction effect could be proven for StancePhase*Speed, $F(2.5, 47.48) = 4.83$, $p = 0.01$, as shown in figure 49.

M. Gastr. Med.: StancePhase had a main effect on average activation of the M. Gastr. Med. ($F(1.90, 30.03) = 46.56$, $p < 0.01$). Activation differed significantly between all StancePhases ($p < 0.05$). Figure 50 demonstrates the comparisons between average activation levels of M. Gastr. lat. during different StancePhases. A significant interaction effect was found for StancePhase*Speed, $F(1.84, 34.94) = 6.05$, $p = 0.01$, which is depicted in figure 51. A very clear similarity can be seen with figure 49.

M. Tib. Ant.: StancePhase had a main effect on average activation of the M. Tib. Ant. ($F(1.33, 25.15) = 9.62$, $p < 0.01$). Differences in average activation of this muscle between StancePhases are shown in figure 52.

M. Peroneus long.: A significant main effect was found for StancePhase, $F(2.93, 55.73) = 35.38$, $p < 0.01$. Pairwise comparison showed

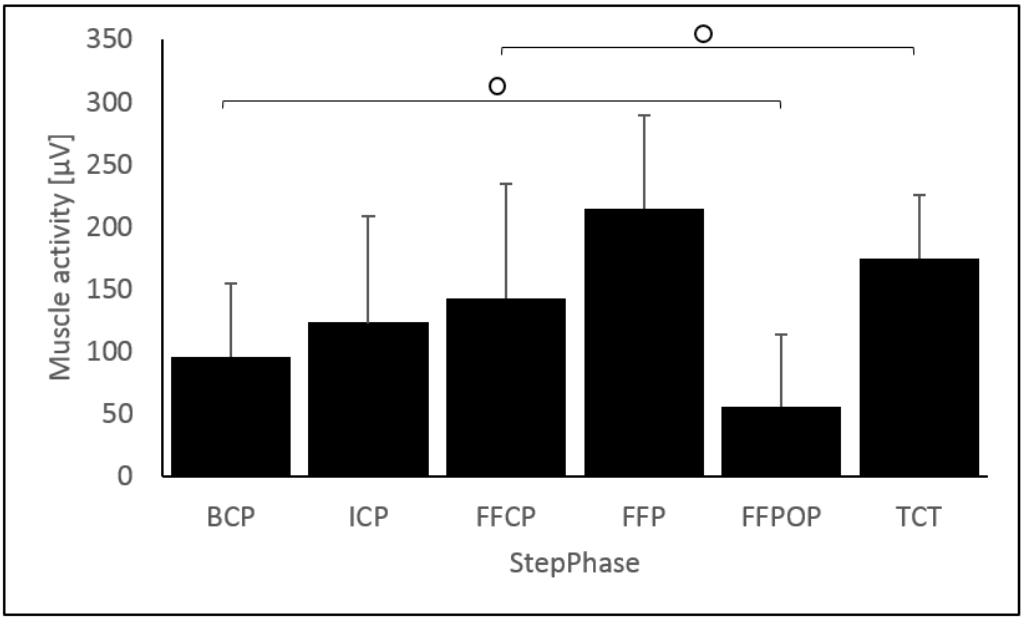

Figure 48: Average activation of M. Gastr. lat. during the investigated StancePhases. All comparisons of the StancePhases showed significant difference except for the ones indicated with a circle. Bars show mean values while error bars indicate SDs

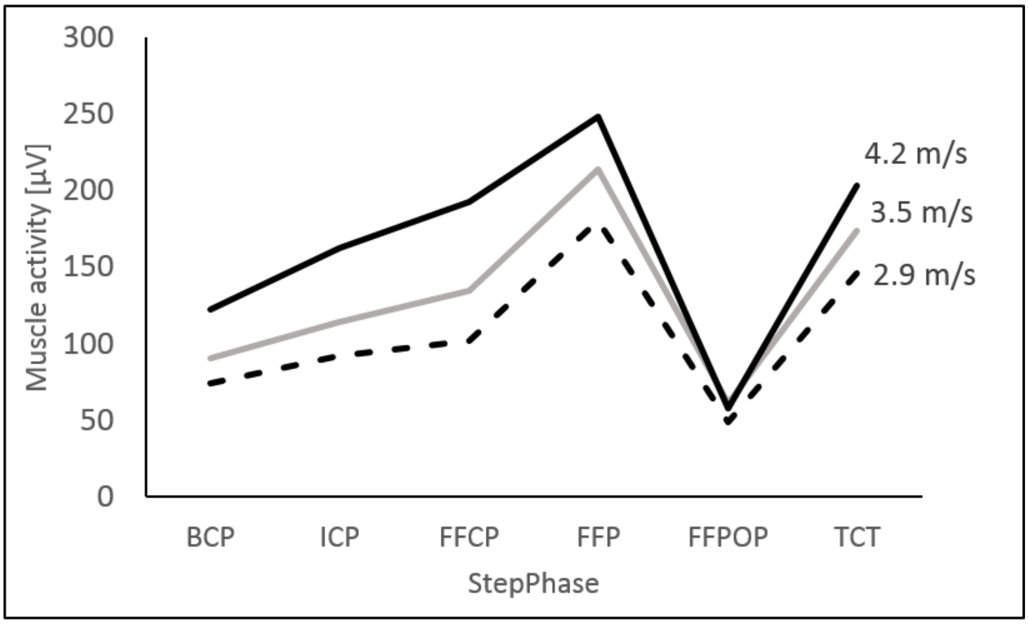

Figure 49: Interaction effect of StancePhase and running speed on average activation of M. Gastr. lat.

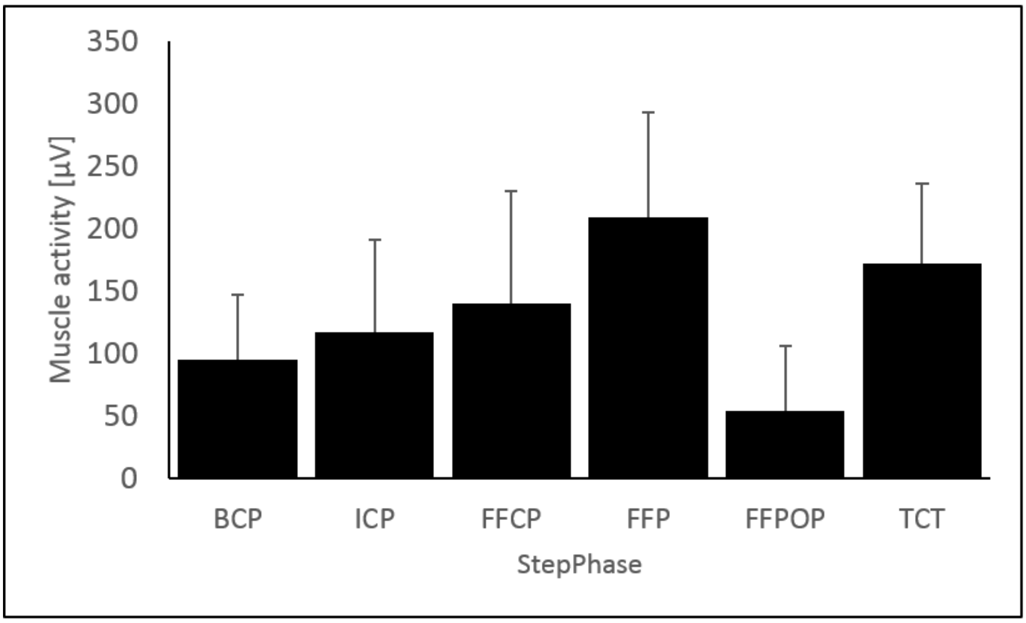

Figure 50: Average muscle activity of M. Gastr. med. at the investigated StancePhases. All comparisons showed significant differences between the StancePhases and are therefore not marked any further.

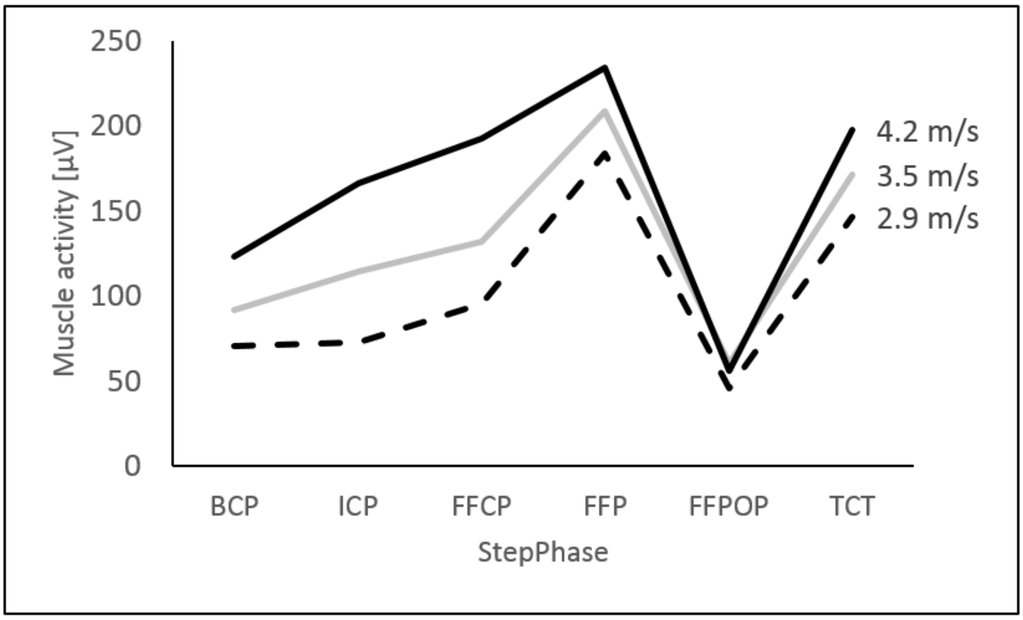

Figure 51: Interaction effect of StancePhase and running speed on average activation of M. Gastr. med.

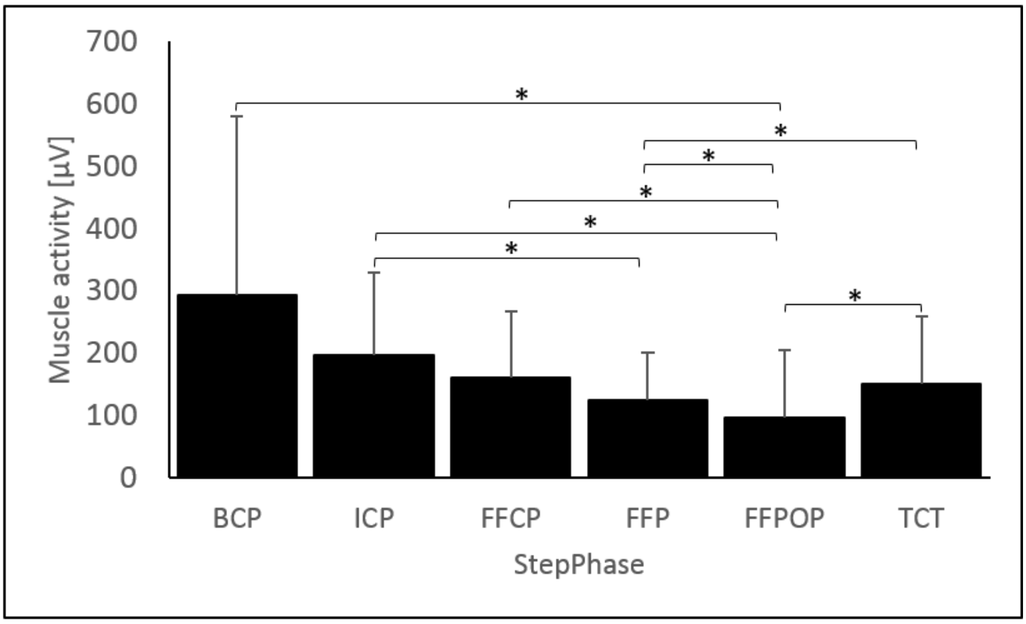

Figure 52: Average muscle activity of M. Tib. ant. at the investigated StancePhases. Significant differences between phases are indicated by a star.

differences between average activation of the M. Peroneus between most StancePhases as shown in figure 53.

FOOT KINEMATICS MaxPro: A significant main effect was found for StancePhase, F(1.48, 28.04) = 89.12, $p < 0.01$. MaxPro was found to differ between all step phases ($p < 0.01$) except for the comparison between FFCP (1.97 ± 4.9°) and FFPOP (0.54 ± 2.27°) as well as between FFP (-4.18 ± 1.67°) and TCT (-4.23 ± 1.68°) as can be seen in figure 54. Negative values resemble the foot to be positioned in pronation while positive values resemble the foot to be supinated. No other significant main effects or interaction effects could be proven.

MaxProVel: A signifiant main effect was found for StancePhase, F(2.39, 45.46) = 103.83, $p < 0.01$. MaxProVel differed between all StancePhases ($p < 0.05$). Results from the pairwise comparison are shown in figure 55.

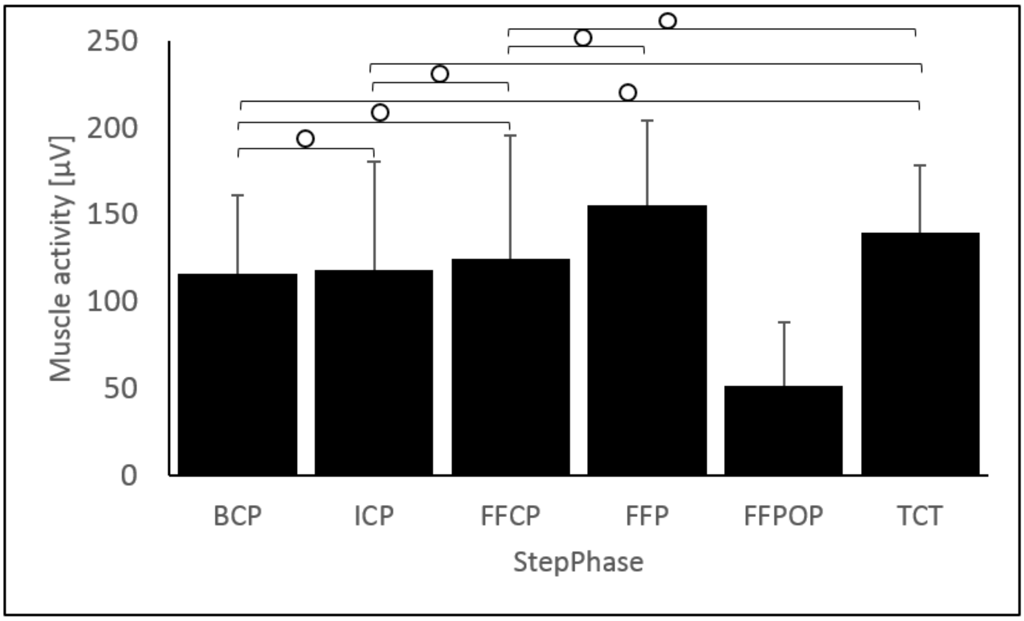

Figure 53: Average muscle activity of M. Peroneus long. at the investigated StancePhases. Non significant differences between phases are indicated by a circle. All other comparisons showed statisticly significant differences.

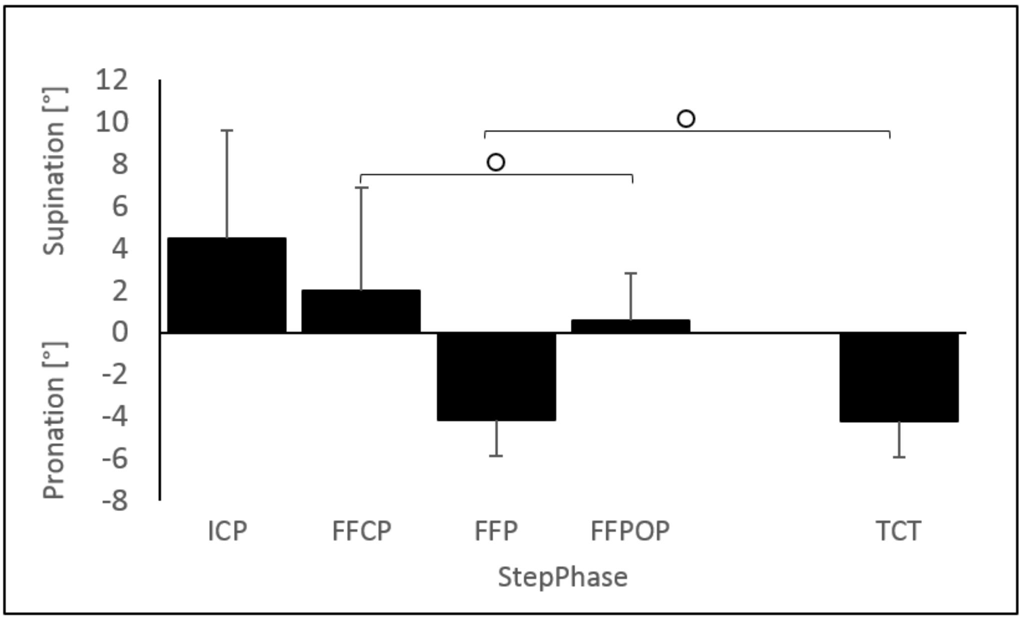

Figure 54: Average MaxPro at the investigated StancePhases. Non significant differences between phases are indicated by a circle. All other comparisons showed statisticly significant differences.

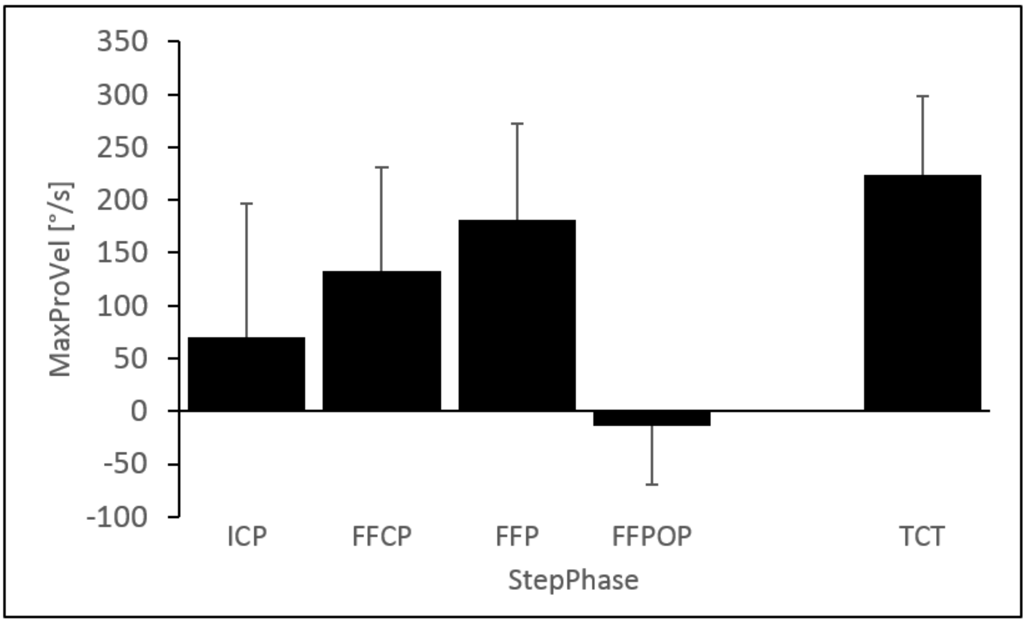

Figure 55: MaxProVel averaged over StancePhases. Comparisons between all StancePhases were showed significant differences. For reasons of clarity these are not specifically depicted in this figure.

ROM: StancePhase had a significant main effect on ROM, F(1.95, 37.12) = 261.74, p < 0.01. All StancePhases were found to differ in their ROM, p < 0.05 respectively (figure 56). A significant interaction effect could be proven for Configuration*StancePhase, F(2.95, 56.53) = 4.44, p < 0.01. During all StancePhases an increase in ROM can be seen in NS compared to Con2 except for ICP, which shows a decrease in ROM of 0,6°, see figure 57. Another interaction effect was found for Speed*StancePhase, F(3.39, 64.31) = 4.51, p < 0.01 (figure 58).

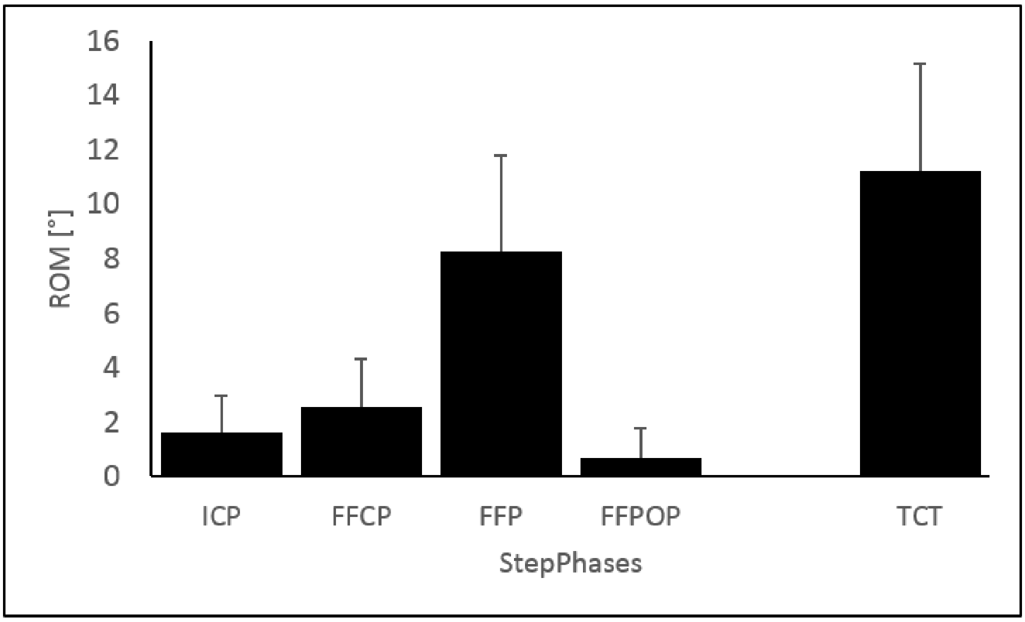

Figure 56: ROM averaged over StancePhases. All comparisons between StancePhases were showed significantly different results. For reasons of clarity these are not marked separately in the graph.

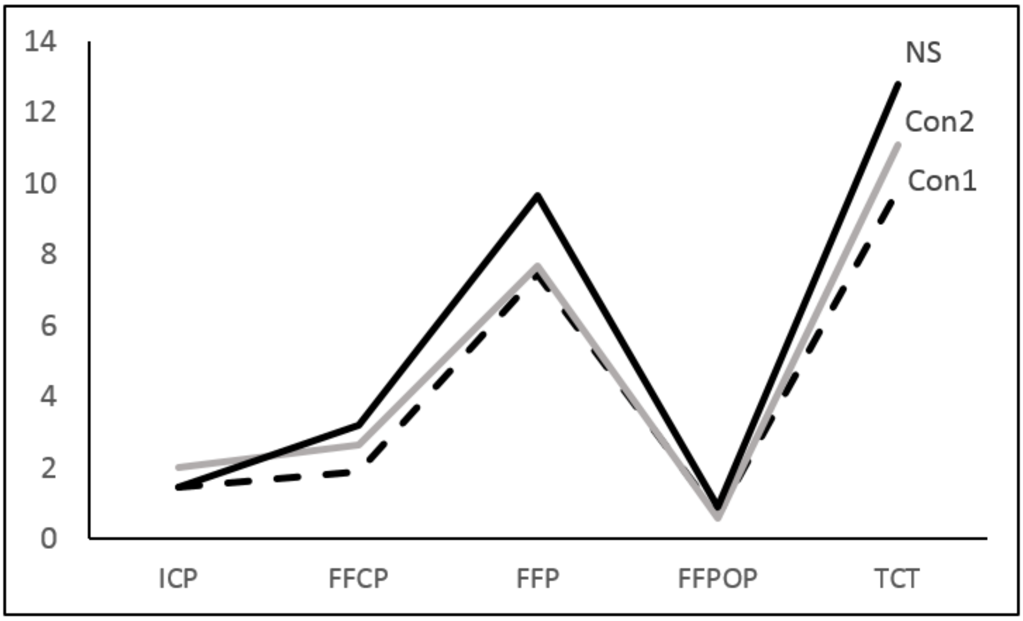

Figure 57: Interaction effect of configuration*StancePhase on ROM.

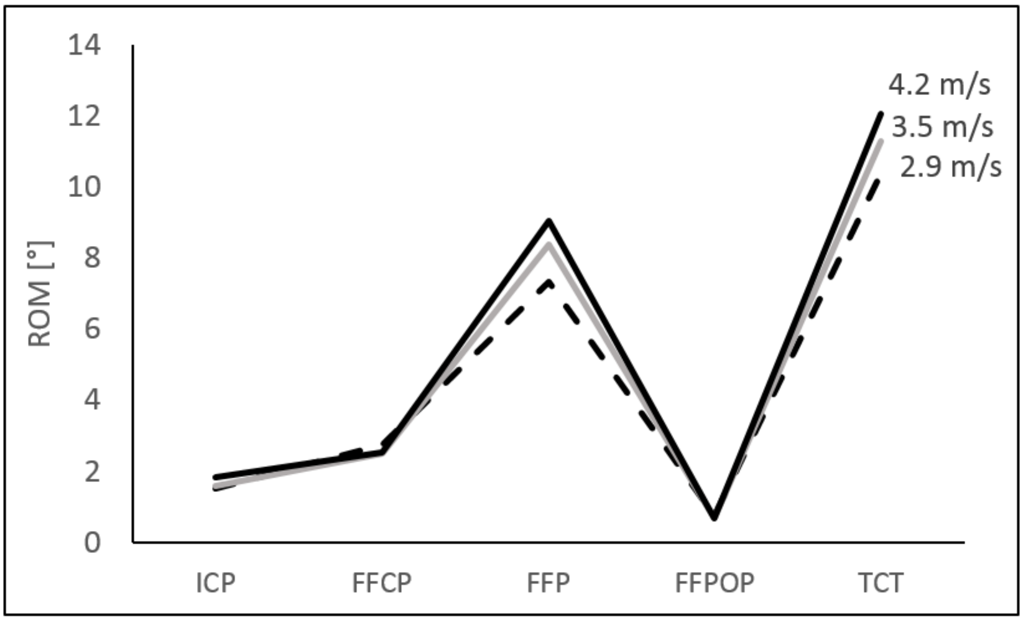

Figure 58: Interaction effect of Speed*StancePhase on ROM.

Table 22: Descriptives of Plantar Pressure

	Force Peaks	Average Force
Con1	44.78 ± 13.38	1689.58 ± 492.43
Con2	45.09 ± 13.84	1694.31 ± 520.34
NS	50.89 ± 14.65	1607.00 ± 480.26
2.9 m/s	44.25 ± 13.13	1583.19 ± 450.47
3.5 m/s	47.39 ± 13.88	1675.50 ± 500.85
4.2 m/s	49.12 ± 14.86	1732.19 ± 541.72

Table 23: Descriptives of Average Muscle Activity

	M. Tib. Ant.	M. Peroneus long.
Con1	134.00 ± 90.37	147.61 ± 44.95
Con2	119.47 ± 47.55	144.24 ± 48.15
NS	123.69 ± 51.86	142.85 ± 46.35
2.9 m/s	112.26 ± 61.03	127.96 ± 38.85
3.5 m/s	120.80 ± 55.93	145.55 ± 41.07
4.2 m/s	144.10 ± 76.33	161.18 ± 52.36
	M. Gastr. Med.	M. Gastr. Lat.
Con1	182.00 ± 75.80	197.89 ± 71.21
Con2	198.47 ± 81.44	187.37 ± 65.95
NS	186.55 ± 78.64	188.18 ± 61.53
2.9 m/s	161.78 ± 67.49	159.66 ± 53.52
3.5 m/s	189.17 ± 76.33	189.48 ± 60.13
4.2 m/s	216.07 ± 82.66	224.30 ± 68.31

Table 24: Descriptives of PeakAcc

	Proximal Accelerometer	Distal Accelerometer
Con1	44.39 ± 13.59	48.02 ± 12.32
Con2	45.03 ± 14.48	48.94 ± 10.74
NS	45.15 ± 14.59	47.88 ± 10.67
2.9 m/s	38.24 ± 12.99	41.08 ± 12.60
3.5 m/s	44.44 ± 13.42	48.25 ± 12.76
4.2 m/s	51.89 ± 12.83	55.52 ± 12.38
x-direction	43.34 ± 15.23	50.13 ± 16.20
y-direction	52.83 ± 12.73	52.91 ± 11.98
z-direction	38.40 ± 10.27	41.80 ± 10.30

Table 25: Descriptives of Average Oscillation Frequency

	Proximal Accelerometer	Distal Accelerometer
Con1	23.72 ± 2.99	25.82 ± 3.15
Con2	24.06 ± 3.03	26.16 ± 3.09
NS	24.54 ± 3.33	26.96 ± 3.44
2.9 m/s	24.41 ± 2.95	26.25 ± 2.96
3.5 m/s	24.52 ± 3.15	26.81 ± 3.40
4.2 m/s	23.39 ± 3.26	25.88 ± 3.31
x-direction	20.16 ± 3.26	23.47 ± 3.17
y-direction	22.82 ± 3.09	24.45 ± 3.42
z-direction	29.33 ± 3.00	31.02 ± 3.08

Table 26: Descriptives of Maximum Oscillation Frequen-cy

	Proximal Accelerometer	Distal Accelerometer
Con1	43.73 ± 8.85	48.97 ± 10.51
Con2	44.24 ± 9.21	49.40 ± 10.06
NS	48.33 ± 10.78	54.27 ± 11.22
2.9 m/s	39.51 ± 8.56	44.44 ± 10.20
3.5 m/s	46.04 ± 10.16	52.32 ± 11.26
4.2 m/s	50.74 ± 10.12	55.88 ± 10.49
x-direction	40.7 ± 8.13	47.09 ± 10.15
y-direction	42.67 ± 12.05	49.77 ± 12.59
z-direction	52.93 ± 8.65	55.78 ± 9.05

Table 27: Descriptives of Foot Kinematics

	MaxPro	MaxProVel	ROM	TFPro
Con1	3.89 ± 1.53	102.36 ± 108.45	4.25 ± 4.30	77.31 ± 14.79
Con2	4.02 ± 1.51	120.90 ± 121.57	4.79 ± 4.65	80.58 ± 13.92
NS	4.78 ± 1.73	132.90 ± 140.25	5.59 ± 4.89	79.20 ± 12.76
2.0 m/s	3.77 ± 1.46	102.23 ± 108.14	4.53 ± 4.44	77.77 ± 13.68
3.5 m/s	4.26 ± 1.66	115.24 ± 121.79	4.88 ± 4.88	79.69 ± 13.76
4.2 m/s	4.65 ± 1.66	138.68 ± 139.55	5.22 ± 5.25	79.63 ± 14.04

A.2 DESCRIPTIVE STATISTICS CHAPTER 9

Table 28: Descriptives of Plantar Pressure

	Total Force		Peak Forces	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	1653.29 ± 322.86	1635.32 ± 268.74	73.28 ± 23.81	70.07 ± 21.30
Con2	1552.75 ± 405.74	1588.92 ± 353.85	58.02 ± 28.58	63.91 ± 29.72
NS	1641.65 ± 368.46	1645.07 ± 355.76	66.78 ± 24.08	64.85 ± 21.38

Table 29: Descriptives of Average Muscle Activity

	M. Gastr. Lat.		M. Gastr. Med.	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	134.22 ± 38.88	121.65 ± 28.20	164.89 ± 57.08	185.61 ± 103.78
Con2	149.02 ± 51.50	120.47 ± 24.06	157.67 ± 49.99	153.03 ± 79.45
NS	127.40 ± 31.37	120.91 ± 32.01	151.95 ± 45.98	154.08 ± 52.61
	M. Tib. Ant.		M. Peroneus long.	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	131.32 ± 144.14	141.88 ± 306.11	152.69 ± 27.50	164.00 ± 45.86
Con2	79.99 ± 42.01	100.34 ± 69.94	150.77 ± 40.82	147.76 ± 69.18
NS	72.65 ± 35.85	82.34 ± 35.40	142.18 ± 29.28	149.11 ± 41.42

Table 30: Descriptives of Foot Kinematics

	MaxPro		MaxProVel	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.13 ± 1.50	1.00 ± 2.19	56.86 ± 34.55	68.48 ± 38.39
Con2	3.37 ± 6.10	1.93 ± 1.65	87.06 ± 43.09	89.89 ± 40.92
NS	2.30 ± 2.75	2.55 ± 3.03	123.56 ± 43.82	142.56 ± 47.57
	ROM		TFPro	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	7.77 ± 3.82	7.97 ± 3.33	90.56 ± 12.61	93.08 ± 7.36
Con2	9.71 ± 4.76	8.93 ± 3.83	77.48 ± 19.28	82.22 ± 14.64
NS	10.31 ± 3.97	11.65 ± 3.85	75.19 ± 13.98	79.08 ± 11.34

Table 31: Descriptives of oscillations in time space detected with the distal accelerometer

	PeakAcc x-direction		PeakAcc yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	1.17 ± 1,34	0.81 ± 0.59	3.66 ± 2.04	3.46 ± 1.57
Con2	1.65 ± 1.51	2.31 ± 1.84	5.03 ± 2.15	4.74 ± 1.85
NS	2.05 ± 1.85	0.76 ± 0.79	5.24 ± 2.73	3.41 ± 1.40
	ttpeak x-direction		ttpeak yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	276.89 ± 136.67	313.31 ± 88.85	285.19 ± 164.16	251.08 ± 114.02
Con2	266.10 ± 137.32	210.75 ± 97.16	251.82 ± 133.29	236.41 ± 124.65
NS	265.20 ± 138.53	267.39 ± 92.96	238.43 ± 141.92	287.85 ± 100.60

Table 32: Descriptives of oscillations in time space detected with the proximal accelerometer

	PeakAcc x-direction		PeakAcc yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	62.19 ± 11.00	56.97 ± 11.09	64.17 ± 17.36	63.89 ± 15.10
Con2	64.01 ± 13.02	58.41 ± 17.36	60.74 ± 19.80	55.35 ± 20.46
NS	61.72 ± 16.62	62.04 ± 12.04	64.11 ± 21.16	68.57 ± 17.39
	ttpeak x-direction		ttpeak yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	103.98 ± 54.28	119.51 ± 51.60	120.53 ± 67.54	131.54 ± 64.28
Con2	123.47 ± 96.87	128.15 ± 108.37	122.52 ± 94.04	149.96 ± 105.63
NS	125.96 ± 71.76	111.59 ± 58.49	122.08 ± 63.46	132.03 ± 63.56

Table 33: Descriptives of dominant oscillation frequencies detected with both accelerometer

	Distal Accelerometer			
	Dominant Frequency x-direction		Dominant Frequency yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	23.30 ± 6.61	24.80 ± 9.06	38.75 ± 18.48	32.70 ± 15.53
Con2	28.65 ± 13.86	22.25 ± 12.20	36.70 ± 19.30	30.50 ± 19.62
NS	27.55 ± 12.96	29.20 ± 14.94	36.00 ± 21.40	38.95 ± 27.81
	Proximal Accelerometer			
	Dominant Frequency x-direction		Dominant Frequency yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	24.20 ± 10.36	20.80 ± 7.56	39.95 ± 16.55	31.75 ± 13.74
Con2	25.65 ± 13.51	23.20 ± 11.96	29.90 ± 21.75	29.05 ± 20.81
NS	30.25 ± 12.90	27.65 ± 13.95	30.25 ± 12.90	40.40 ± 25.45

Table 34: Descriptives of average power in the investigated power components detected with the distal accelerometer

	Low power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.98 ± 0.07	0.85 ± 0.20	0.85 ± 0.18	0.91 ± 0.15
Con2	1.02 ± 0.09	1.01 ± 0.12	0.93 ± 0.14	0.93 ± 0.26
NS	0.95 ± 0.13	0.94 ± 0.18	0.92 ± 0.27	0.91 ± 0.26
	Medium power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.85 ± 0.17	0.99 ± 0.09	0.94 ± 0.17	0.97 ± 0.18
Con2	0.90 ± 0.16	0.83 ± 0.15	0.92 ± 0.19	0.90 ± 0.16
NS	0.92 ± 0.17	0.93 ± 0.26	0.87 ± 0.13	0.89 ± 0.15
	High power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.35 ± 0.11	0.44 ± 0.16	0.71 ± 0.12	0.71 ± 0.09
Con2	0.49 ± 0.18	0.47 ± 0.17	0.77 ± 0.15	0.73 ± 0.19
NS	0.49 ± 0.16	0.57 ± 0.22	0.76 ± 0.15	0.79 ± 0.19
	Highest power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.09 ± 0.03	0.09 ± 0.03	0.09 ± 0.02	0.09 ± 0.03
Con2	0.10 ± 0.06	0.09 ± 0.04	0.11 ± 0.06	0.11 ± 0.11
NS	0.08 ± 0.02	0.09 ± 0.03	0.12 ± 0.10	0.13 ± 0.16

Table 35: Descriptives of average power in the investigated power components detected with the proximal accelerometer

	Low power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.99 ± 0.08	0.98 ± 0.07	0.76 ± 0.17	0.88 ± 0.15
Con2	1.01 ± 0.11	1.03 ± 0.09	0.93 ± 0.14	0.92 ± 0.24
NS	0.94 ± 0.09	0.97 ± 0.13	0.90 ± 0.24	0.84 ± 0.24
	Medium power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.81 ± 0.19	0.82 ± 0.22	0.89 ± 0.16	0.96 ± 0.13
Con2	0.86 ± 0.16	0.85 ± 0.15	0.77 ± 0.19	0.82 ± 0.15
NS	0.89 ± 0.17	0.92 ± 0.22	0.81 ± 0.21	0.76 ± 0.16
	High power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.36 ± 0.12	0.41 ± 0.13	0.71 ± 0.13	0.69 ± 0.10
Con2	0.39 ± 0.17	0.40 ± 0.18	0.68 ± 0.18	0.67 ± 0.22
NS	0.40 ± 0.16	0.52 ± 0.23	0.75 ± 0.21	0.70 ± 0.20
	Highest power component			
	x-direction		yz-direction	
	Treadmill	Over Ground	Treadmill	Over Ground
Con1	0.10 ± 0.14	0.10 ± 0.15	0.12 ± 0.13	0.12 ± 0.14
Con2	0.13 ± 0.13	0.11 ± 0.14	0.14 ± 0.15	0.13 ± 0.13
NS	0.11 ± 0.15	0.10 ± 0.13	0.14 ± 0.16	0.17 ± 0.23

A.3 SCORE PLOTS OF CHAPTER 10

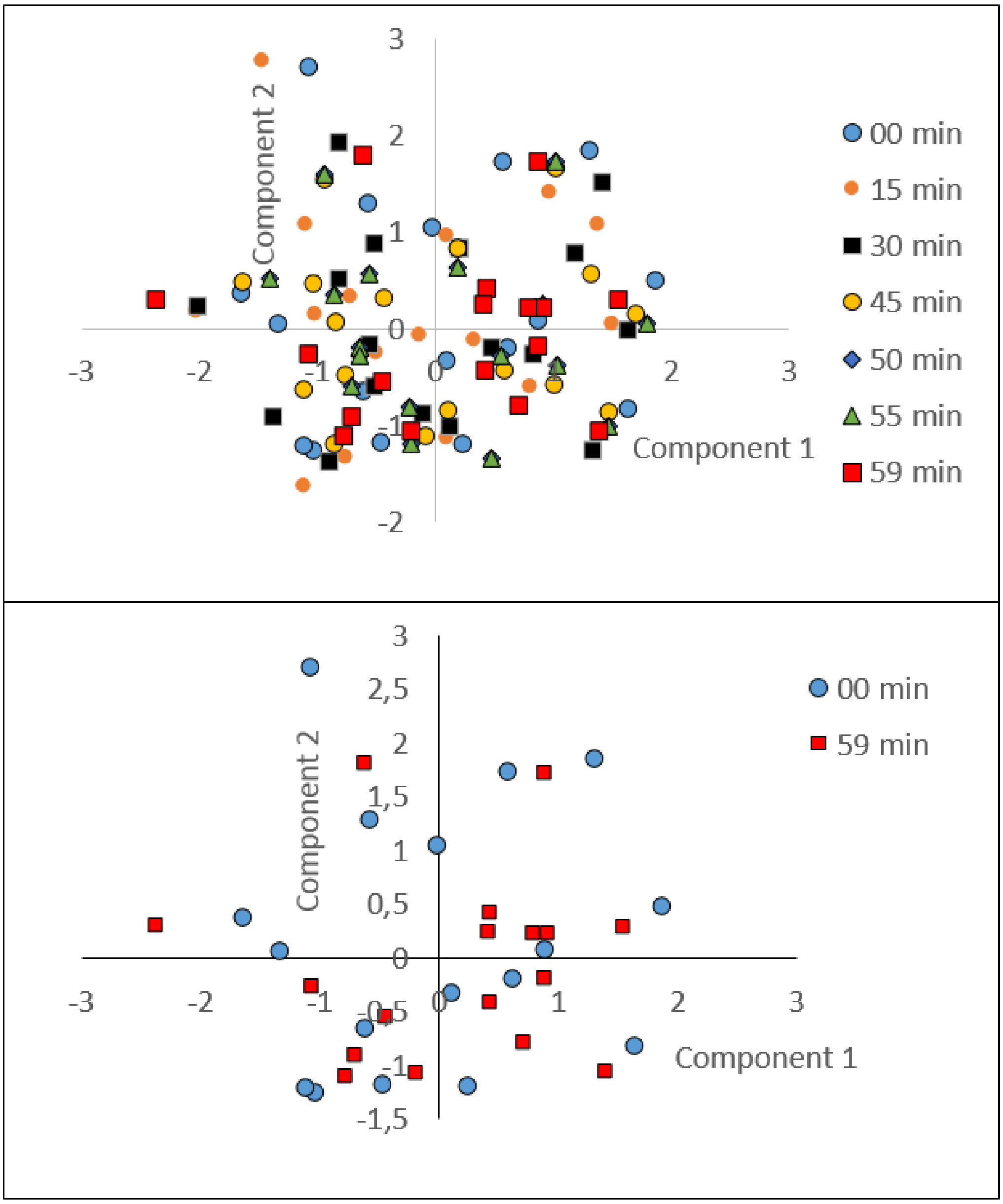

Figure 59: Score plot showing the factor loadings on the two components of kinematic data with color coded score values.

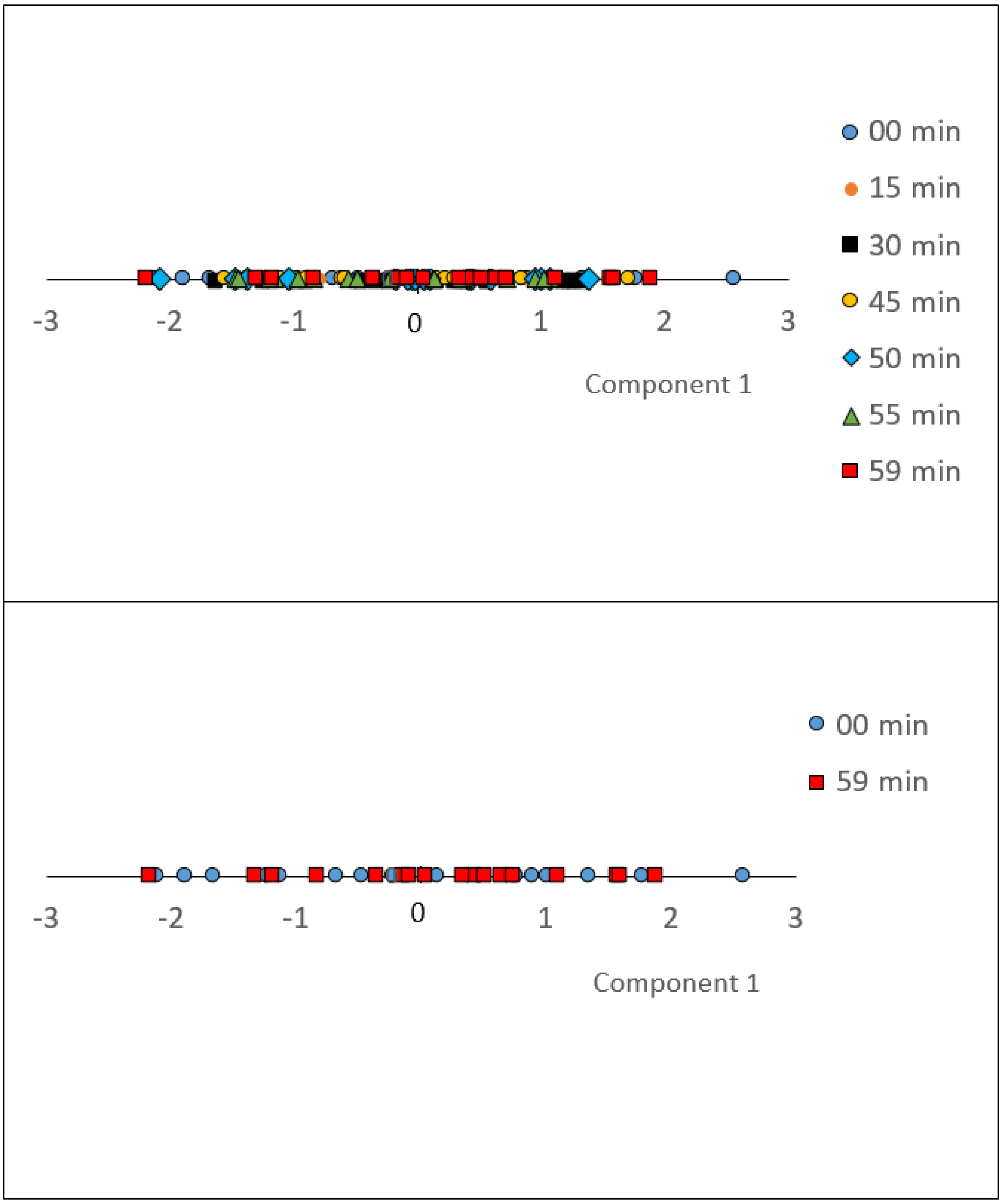

Figure 60: Score plot showing the factor loadings on the component of pressure data with color coded score values.

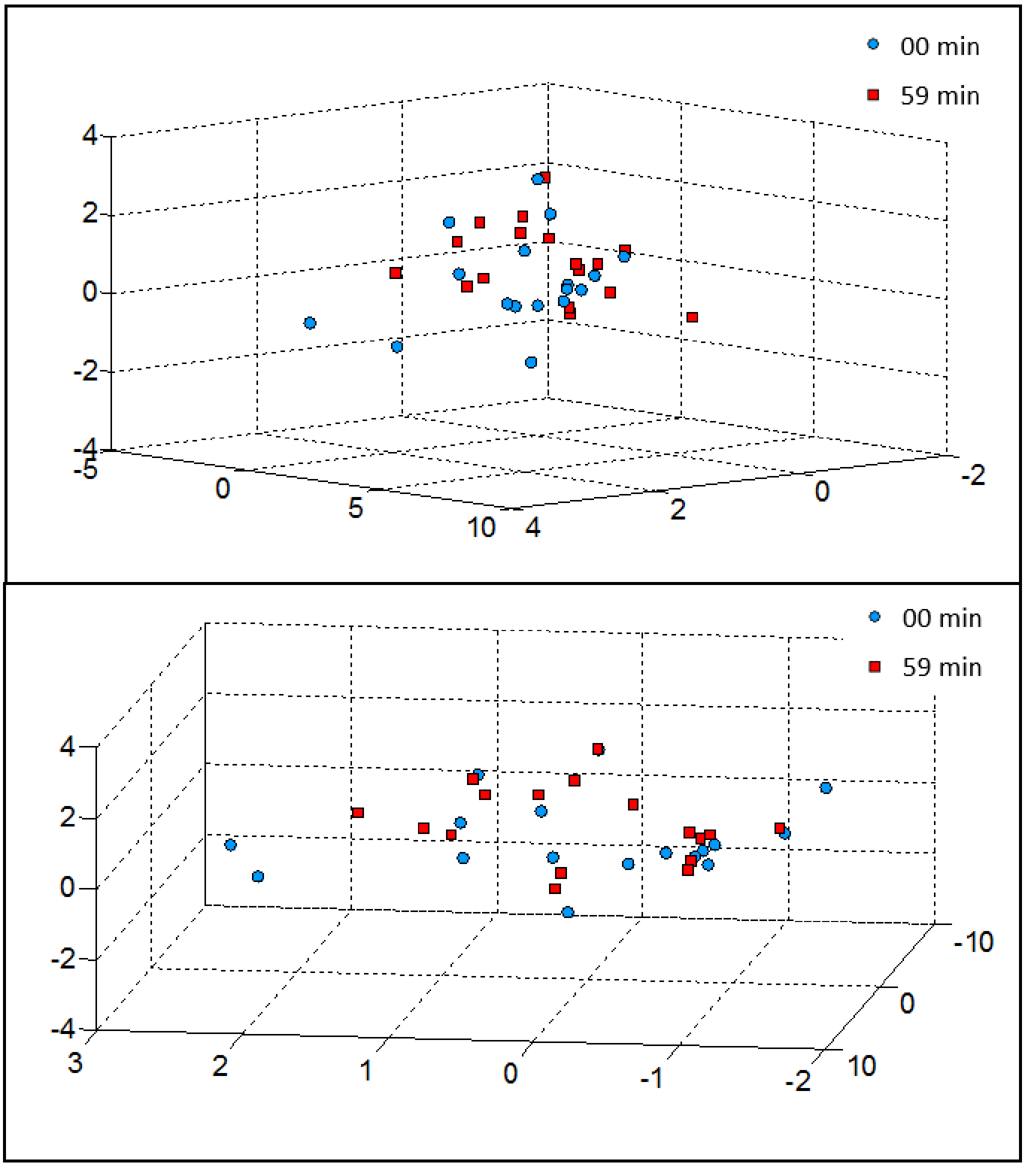

Figure 61: Score plot showing the factor loadings on the component of oscillations at the Achilles tendon with color coded score values. Two different perspectives of a three dimensional plot are shown.

A.4 DESCRIPTIVE STATISTICS CHAPTER 11

Table 36: Descriptives of average total plantar force in subjects with Achilles tendon complaints.

	Total Force
Con1	1174.13 ± 165.89
Con2	1267.63 ± 157.87
NS	1196.25 ± 170.92

Table 37: Descriptives of average and peak muscle activity in subjects with Achilles tendon complaints.

	Average Activity		
	M. Gastr. Lat.	M. Gastr. Med.	M. Soleus
Con1	76.06 ± 18.36	86.19 ± 24.39	66.86 ± 19.60
Con2	81.76 ± 19.81	92.36 ± 23.50	68.70 ± 18.48
NS	79.99 ± 17.96	90.65 ± 27.26	69.19 ± 22.56
	Peak Activity		
	M. Gastr. Lat.	M. Gastr. Med.	M. Soleus
Con1	1622.88 ± 708.75	1640.38 ± 420.62	1405.13 ± 376.95
Con2	1493.75 ± 251.23	1652.50 ± 325.25	1319.13 ± 354.05
NS	1432.50 ± 362.18	1697.75 ± 453.16	1288.00 ± 408.10

Table 38: Descriptives of foot kinematics in subjects with Achilles tendon complaints.

	MaxPro	MaxProVel	ROM	TFPro
Con1	6.39 ± 4.45	197.57 ± 61.76	27.06 ± 6.37	75.99 ± 14.48
Con2	8.61 ± 4.26	355.34 ± 238.01	30.54 ± 7.81	68.36 ± 7.83
NS	10.43 ± 4.61	339.80 ± 204.47	31.10 ± 7.76	66.88 ± 6.77

Table 39: Descriptives of dominant oscillation frequencies detected with both accelerometers in subjects with Achilles tendon complaints.

	Distal accelerometer		Proximal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	38.50 ± 14.87	72.25 ± 22.71	30.63 ± 13.63	56.88 ± 29.11
Con2	31.75 ± 5.26	74.75 ± 18.26	35.75 ± 11.18	62.13 ± 24.90
NS	35.63 ± 10.85	56.88 ± 28.98	40.88 ± 16.60	68.88 ± 22.25

Table 40: Descriptives of oscillations in time space detected with the both accelerometer in subjects with Achilles tendon complaints.

	PeakAcc proximal accelerometer		PeakAcc distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	63.34 ± 6.96	59.10 ± 9.92	64.06 ± 6.04	55.73 ± 13.49
Con2	65.27 ± 7.18	61.63 ± 12.63	64.17 ± 5.99	55.89 ± 15.92
NS	64.89 ± 6.98	64.01 ± 10.62	64.28 ± 5.86	60.21 ± 13.25
	ttpeak proximal accelerometer		ttpeak distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	56.25 ± 20.35	61.25 ± 20.73	56.38 ± 20.50	41.00 ± 11.34
Con2	47.13 ± 21.74	55.50 ± 23.55	48.13 ± 21.79	41.63 ± 15.32
NS	46.63 ± 13.38	49.75 ± 16.58	45.88 ± 15.16	42.88 ± 13.25

Table 41: Descriptives of average power in the investigated power component detected with the both accelerometer in subjects with Achilles tendon complaints.

	Low power component			
	Proximal accelerometer		Distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	0.86 ± 0.16	0.88 ± 0.12	0.91 ± 0.15	0.98 ± 0.26
Con2	0.85 ± 0.16	0.84 ± 0.24	0.92 ± 0.16	0.90 ± 0.21
NS	0.84 ± 0.22	0.80 ± 0.22	0.90 ± 0.20	0.95 ± 0.20
	Medium power component			
	Proximal accelerometer		Distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	1.60 ± 0.52	1.57 ± 0.47	1.70 ± 0.55	1.44 ± 0.58
Con2	1.69 ± 0.56	1.56 ± 0.48	1.79 ± 0.63	1.53 ± 0.50
NS	1.68 ± 0.48	1.52 ± 0.55	1.78 ± 0.49	1.53 ± 0.55
	High power component			
	Proximal accelerometer		Distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	1.17 ± 0.59	1.49 ± 0.49	1.11 ± 0.55	1.58 ± 0.47
Con2	1.10 ± 0.51	1.44 ± 0.45	1.11 ± 0.52	1.47 ± 0.42
NS	1.20 ± 0.56	1.46 ± 0.38	1.20 ± 0.45	1.51 ± 0.45
	Highest power component			
	Proximal accelerometer		Distal accelerometer	
	x-direction	yz-direction	x-direction	yz-direction
Con1	0.09 ± 0.03	0.09 ± 0.04	0.09 ± 0.03	0.10 ± 0.04
Con2	0.08 ± 0.03	0.10 ± 0.04	0.08 ± 0.03	0.10 ± 0.04
NS	0.08 ± 0.03	0.10 ± 0.03	0.07 ± 0.03	0.12 ± 0.04

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DECLARATION

I hereby declare, that this dissertation is solely my original work. I have used only the sources and materials indicated and have not received any unauthorized assistance from others. All quotations from other works as well as paraphrases or summaries of other works have been identified as such and properly acknowledged in the dissertation. This dissertation or parts thereof have not been submitted to an educational institution in Germany or abroad as part of an examination or degree qualification. I fully understand the meaning of this affidavit as well as the criminal penalties for submitting a false or incomplete statement. I hereby confirm, that to the best of my knowledge the above statements are true, correct and complete.

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