REARFOOT BIOMECHANICS IN

ACHILLES TENDON FUNCTION

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Declaration

I declare that this thesis and the work presented in it is the result of my own efforts during my candidature for a Master of Philosophy degree at the University of Salford, United Kingdom. Where work involved other individuals I have identified my own contribution and that of others in the specific chapter. The data presented in chapter 3 was collected in collaboration with another post graduate student (Rachel Majumdar) but was all processed, statistically analysed and interpreted by the candidate. The data used in chapter 3 has been published as follows: Majumdar, R., Laxton, P., Thuesen, A., Richards, B., Liu, A., Arán-Ais, F., Parreño, E. M., Nester, C. J. (2013). Development and evaluation of prefabricated antipronation foot orthosis. *Journal of Rehabilitation Research and Development*, 50(10), 1331–42.

Abbreviations and symbols for unit of measurement

Abbreviation /Symbol	Unit of measurement
%	Percent
%•s^{-1}	strain loading rate expressed as percent per second
°	Degrees
°/s	angular velocity expressed in degrees per second
cm	centrimeter
cm^2	centimeter squared
g	Gram
Hz	Hertz
Kg	kilogram
MHz	megahertz
mm	millimeters
mm/°	millimeters per degree
mm^2	millimeters squared
MPa	mega pascal
mV	millivolts
N	load/force expressed in Newtons
N•s^{-1}	force loading rate expressed in Newtons per second
Nm/kg	unit of measurement for moments expressed in Newton meters per kilogram

Abbreviations and symbols for statistical, mathematical and mechanical engineering operations

Abbreviation /Symbol	Definition
±*SD*	standard deviation
ΔL	Absolute change in distance expressed in millimetre, $\Delta L = L - L_0$, where L_0 is the initial length and L is the final length
CI	confidence interval
d	Cohen's *d* effect size between two means
E	Young’s modulus also known as elastic modulus. Young modulus (E) = stress (σ)/strain (ε)
H_0	null hypothesis, hypothesis under test
H_1	alternative hypothesis
IQR	interquartile range IQR= 75th- 25th quartiles, where 25th is the lower quartile and 75th percentile is the upper quartile in relation to the 50th percentile (the median)
k	stiffness expressed as Newtons per millimetre, where stiffness (k)= N/mm
L_0	initial tissue length expressed in millimetre
LL	lower confidence interval limit
M	mean value
Mdn	median
n	number of subjects
p	*p*-value indicate the level of significance in given statistical test
r	the effect size *r* correlation between two means
R1	region of interest R1, where $R1(R1_x,R1_y)$ is a coordinate pair placed in the distal tendon structure medially or laterally
$\overrightarrow{R2R1}$	motion vector, where $R1(R1_x,R1_y)$ and $R2(R2_x,R2_y)$ are two coordinate pairs positioned in either the medial or lateral and distal or proximal tendon portion, then $\overrightarrow{R2R1} = \begin{pmatrix} R2_x - R1_x \\ R2_y - R1_y \end{pmatrix}$.
$\Vert \overrightarrow{R2R1} \Vert$	length of motion vector, where $R1(R1_x,R1_y)$ and $R2(R2_x,R2_y)$ are two coordinate pairs positioned in either the medial or lateral and distal or proximal tendon portion, then the length of the motion vector is $\Vert \overrightarrow{R2R1} \Vert = \sqrt{\left((R2_x - R1_x)^2 + (R2_y - R1_y)^2\right)}$.

Abbreviations and symbols for statistical, mathematical and mechanical engineering operations (continued)

Abbreviation /Symbol	Definition
R2	Region of interest R2, where R2($R2_x$,$R2_y$) is a coordinate pair placed in the proximal tendon structure medially or laterally
R^2	Pearson correlation coefficient
t	paired t-test critical value
Type I error	Type I error is rejecting of the null hypothesis H_0 (accepting a false alternative hypothesis H_1) when the null hypothesis H_0 is true.
Type II error	Type II error is rejecting of the alternative hypothesis H_1, (accepting a false null hypothesis H_0), when the alternative hypothesis H_1 in fact is true.
UL	upper confidence interval limit
x	the medial-lateral (sagittal) axis of the local coordinate system
XY	XY plane is formed by the intersection of the coronal (Y-axis) and the sagittal (X-axis) plane of the global reference systems' coordinate axis
x-y-z	Cardan/Euler rotation sequence. The sequence indicate the coordinate axis order of the first, second and third rotation
y	the anterior-posterior (frontal) axis of the local coordinate system
z	the vertical axis (transverse) axis of the local coordinate system
Z	Wilcoxon signed rank test critical value
α	Mean inclination angle
α	Alpha level is the probability of committing a type I error in statistical hypothesis testing
β	Mean deviation angle
β	The probability of committing a type II error in statistical hypothesis testing. The statistical power is denoted (1 -β).
ε	Tissue deformation (strain) expressed in millimetre due to stress (σ). Strain (ε) = ΔL/ Lo. Strain can also be expressed as a percentage (%).
ε max	Ultimate tendon strain
σ	Stress is force per tissue area expressed in Newtons per millimetre squared. Stress (σ) = N/mm^2
σ max	maximal stress

General Abbreviations and Symbols

Abbreviation /Symbol	Definition
2D	two dimensional
3D	three dimensional
AT	number of Achilles tendons or subjects with Achilles tendonpathy
BW	body weight
CSA	cross sectional area
EMG	electromyography
MRI	magnetic resonance imaging
n/a	n/a not available
PCSA	physiological cross sections area
ROI	region of interest
ROM	range of motion

Abstract

Pain and disorders of the human Achilles tendon can impact on the quality of life of those involved in leisure activities as well as professional sports, and also activities of daily living. Clinically, Achilles problems are challenging as no 'golden standard' exists for assessment of risk factors nor management. There is therefore a need to seek evidence for the risk factors associated with Achilles injury and evidence related to treatments.

The focus of the first part of this thesis is to outline the current state of scientific research on the Achilles tendon covering anatomical, tissue mechanics and biomechanical (kinematics and kinetics) factors. This is followed by a discussion of the theoretical background associated with causative factors and treatment options, with a specific focus on rearfoot function and foot orthoses. From this review research questions are defined and experimental studies proposed.

Based on the outcomes of the review, the overall aim of this thesis was to identify how (1) rearfoot movement and (2) foot orthoses, might affect Achilles tendon function. In the first experimental study changes in 3D kinematics due to foot orthosis were evaluated during walking and running. The objective was to gain a better understanding of how rearfoot angular position and movement change due to an orthosis since orthoses have been proposed as an effective treatment strategy. This involved thirty three symptom free subjects. The mean reduction in rearfoot eversion due to the foot orthosis was 3.91° for walking and 2.29° for running.

In the second experimental study changes in tension on the medial and lateral sides of the Achilles tendon where studied during inversion/eversion movement of the rearfoot. The aim was to gain a better understanding of how tissue displacement in the medial and lateral parts of the tendon relate to changes in the frontal plane position of the rearfoot

(such as changes that occur due to foot orthoses). This was conducted using ultrasound to measure tendon displacement in seventeen healthy subjects during passive pronation and supination of the foot. The study found that increasing rearfoot eversion increased displacement (stretch) on the medial side of the tendon, and reduced displacement on the lateral side. This was reversed for rearfoot inversion. The relationship between rearfoot position/motion and lengthening and shortening in the medial and lateral parts of the Achilles tendon was strong based on the mean data for the sample. The data allows the change in rearfoot position due to foot orthoses (study 1) to be put into a tendon displacement context.

This thesis is the first to report that different displacements occur in the medial and lateral parts of the Achilles tendon relative to frontal rearfoot position and movement in vivo. The results might have important clinical relevance for understanding how rearfoot movement could pose a risk to increases in medial Achilles strain and thereafter tissue damage. This thesis also indicates how the use of foot orthoses may affect strain in the medial and lateral parts of the Achilles tendon, and thus proves some insight into the biomechanical basis for orthotic use in cases of Achilles injury.

1 Introduction

1.1 Introduction

The Achilles tendon is largest and strongest tendon in the human body (Maffulli & Almekinders, 2007). During running gait forces transmitted by the Achilles have been estimated from 4.8 up to 12.5 times body weight (BW) (Komi, 1990; Scott & Winter, 1990; Sinclair, Isherwood, & Taylor, 2014). Despite its strength, this tendon is also highly prone to injuries involving inflammation, micro tears or even ruptures. A recent meta-analysis suggest that the incidence and prevalence of Achilles injuries ranges from 9.1 to 10.9 % and 6.2 to 9.5 % respectively of all running related injuries (Lopes, Hespanhol, Yeung, & Pena Costa, 2012).

Given its common nature but also disabling effects, a considerable amount of literature has been published on risk factors for Achilles tendinopathies. Some of the extrinsic causes have been cited to include inappropriate footwear, ground surface and poor training regimes (e.g. Cook & Purdam, 2012; Haglund-Åkerlind & Eriksson, 1993; McCrory et al., 1999; Smart, Taunton, & Clement, 1980). Proposed intrinsic factors include muscle imbalance, restricted joint flexibility (dorsiflexion), and also excessive, poorly timed or too fast rearfoot eversion (e.g. Clement, Taunton, & Smart, 1984; Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999; Maffulli & Almekinders, 2007; Smart et al., 1980; Wyndow, Cowan, Wrigley, & Crossley, 2010). It is thought that rearfoot eversion is coupled with changes in how stress is distributed within the Achilles tendon, thus posing a risk of injury. Specifically, eversion of the rearfoot, is hypothesised to be associated with stretching and thus stressing of the medial part of the Achilles tendon, and vice versa for inversion and the lateral part of the tendon. Clearly, if a disproportionate amount of the total force being applied to the tendon passes through a smaller area of tendon, that area might be at risk of tissue damage. However little is known about the relationship between rearfoot movement and distribution of stress in the Achilles tendon.

Based on an assumption that rearfoot movement is able to affect stress distribution in the Achilles tendon many people have suggested that foot orthoses be used as a mechanical treatment for Achilles injuries. However, several unanswered questions exist concerning the literature supporting the biomechanical effects of foot orthoses in the treatment of Achilles injuries. Perhaps most important of these is how does a change in rearfoot position due to a foot orthosis relate to a change in the stress distribution within the Achilles tendon? This is perhaps especially important since some reports indicate that the change in rearfoot eversion due to a foot orthosis can be quite small, perhaps $<2°$ (Mills, Blanch, Chapman, McPoil, & Vicenzino, 2010). We do not currently know what such a change in movement means for the stress inside the Achilles tendon.

The work contained in this thesis seeks to investigate gaps in the literature relevant to Achilles injuries. Specifically, how foot orthoses change rearfoot motion, and how rearfoot movement or position may relate to stress distribution in the Achilles tendon. This thesis includes two experimental studies and following this introduction there are four further chapters.

Chapter 2 covers the necessary background information and academic literature related to Achilles tendon anatomy, function, its injury and its treatment. In this review extrinsic and intrinsic factors associated with injury are identified and the proposed role of foot orthoses in the management of Achilles injury discussed. Gaps in the literature concerning the effect of foot orthotics on the rearfoot and how this may relate to Achilles loading are examined and the aims of the subsequent experimental chapters defined.

Chapter 3 reports an experimental study concerned with how foot orthoses affect rearfoot kinematics. This is a precursor to Chapter 4 in which the effects of changes in rearfoot kinematics on stress in the Achilles tendon are investigated. The relationship between rearfoot position and movement and stress in the separate medial and lateral regions of the Achilles tendon is investigated using measures of tendon stretch/shortening (tissue displacement expressed in millimetres (mm)) as a surrogate for measures of tissue stress. Chapter 5 presents a general discussion of the two experimental studies in the context of wider literature and suggests future directions for research.

2 Background and literature review

The aim of this chapter is to cover the necessary background information and academic literature related to Achilles tendon anatomy, function, its injury and its treatment**.** The purpose is to identify factors associated with injury and the role of foot orthoses in the management of Achilles injury by addressing these factors. This chapter thus includes a review of the Achilles tendon anatomical structures and biomechanics, and functional properties during weight bearing tasks. Concerning Achilles injuries, biomechanical mechanisms associated with clinical pathology and current mechanical treatment approaches are reviewed. Finally, gaps in the literature concerning the effect of foot orthotics on the rearfoot and how this may relate to Achilles loading are identified and the aims of the subsequent experimental chapters defined.

2.1 Anatomy of the Achilles tendon

The Achilles tendon (or calcaneal tendon) is the largest and strongest tendinous structure in the human body. Through aponeuroses the junction of gastrocnemius and soleus muscles (triceps surae) forms the tendon proximally and it inserts onto the posterior part of greater the tuberosity of calcaneus distally (Maffulli & Almekinders, 2007).

2.1.1 Proximal tendon formation

The gastrocnemius muscle consists of lateral and medial muscle compartments which originate proximally from the anatomically distinct posterior superior portions of the corresponding epicondyles of the femoral bone. This muscle groups form a complex integrated structure below the knee and later a separation of the lateral and medial muscle compartments can be observed. The entire muscle merges into a broad aponeurosis (tendinous lamina) on the anterior (deep) surface and later thinner in structure coalescing with the soleus aponeurosis into a tendinous structure (the Achilles tendon) (Blitz & Eliot, 2007; Cummins &

Anson, 1946; Dalmau-Pastor, Fargues-Polo, Casanova-Martínez, Vega, & Golanó, 2014; Schepsis, Jones, & Haas, 2002).

The medial head of the gastrocnemius muscle typically becomes part of the aponeurosis structure lower than the lateral side and is larger than its' lateral counterpart (Dalmau-Pastor et al., 2014; Edama et al., 2014; Elson et al., 2007). Edama and colleagues (2014) also observed that the medial head may have a unipennate structure while the lateral head can have more bipennate muscle fascicle formation (Schache et al., 2001).

The deeper and more anterior soleus muscle fibres originate from the upper part of the posterior shaft/head of the fibula and proximal and medial third aspect of the tibia bone (Dalmau-Pastor et al., 2014; Schepsis et al., 2002). The encasement of the soleus muscle on its aponeuroses is complex and the muscle architecture is often simplified in the literature. From anatomical cadaveric observations it can be seen as two independent muscle portions, ventral (bipennate) and dorsal (unipennate) muscle compartments. On the anterior surface both compartments are separated superiorly and inferiorly by the intramuscular aponeurosis. The majority of fibre bundles arise from the dorsal muscle and attach onto the superficial/posterior soleal aponeurosis (intersection lamina). This later fuses distally with the anterior surface of the aponeurosis of the ventral soleal portion. The superior part of the ventral bipennate composition is connected to the dorsal part of the muscle through the intramuscular aponeurosis medially and laterally. Both ventral portions coalesce into a median septum which later fuses anteriorly on to the posterior soleus aponeurosis (intersection lamina) (Agur, Ng-Thow-Hing, Ball, Fiume, & McKee, 2003; Dalmau-Pastor et al., 2014; Finni, Hodgson, Lai, Edgerton, & Sinha, 2003a). The soleus merges to the superficial gastrocnemius aponeurosis to form the Achilles tendon structure itself (e.g conjoint junction of aponeuroses) at approximately mid shaft of the tibia and fibula bones (Dalmau-Pastor et al., 2014; Tashjian, Appel, Banerjee, & DiGiovanni, 2003).

2.1.2 The muscle units attached to the Achilles tendon

The whole muscle-tendon unit spans over three joints of the lower limb, the knee, talocrucral and subtalar joints. However, because of their origins on different sides of the knee the muscles have some anatomical and functional distinction. The gastrocnemius muscle group is a tri-articular unit since it crosses knee and its activity contributes to control of knee flexion/extension, ankle plantarflexion/dorsiflexion and subtalar pronation/supination. Soleus only spans over the ankle and subtalar joint and thus contributes to control of the latter two movements (Perry & Burnfield, 2010). Thus, the activity and function of soleus will influence the ability of the gastrocnemius to exert force across the knee since the two structures are interdependent. For example, soleus action will change the distance between origin and insertion of the muscle/tendon complex and pre tension the Achilles, thus affecting the task of gastrocnemius.

The mass proportions of this plantarflexor muscle group seems be relatively equal. The mean (*M*) weight has been reported as 390, 221 and 129 gram (g) for the soleus and medial and lateral heads of the gastrocnemius respectively. These particular figures are based on an unknown sample size, gender, age distribution and literature sources (Pierrynowski, 1982). Others have not reported measures for all triceps surae components combined, and all have included small sample sizes less than nine subjects (Fukunaga et al., 1992).

More recently, Ward, Eng, Smallwood and Lieber (2009) dissected a number of lower limb muscles in order to map architectural properties (i.e. mass (g), muscle length in centimetres (cm), fiber muscle length (cm), muscle pennation angles (°) and physiological cross-sections area (PCSA), so a generalised definition of the individual muscles force-generating and excursion capacities could be made. These investigations were made on 21 geriatric cadaver legs using magnetic resonance imaging (MRI) and computed tomography images. The weight of triceps surae muscles was lower compared to the literature review by

Fukunaga et al. (1992). It seems that large variations (based on the reported *SD*) among elderly individuals (n=20 legs) exist (e.g. soleus muscle 275.8(±98.5) g, lateral gastrocnemius 113.5(±32.0) g and medial gastrocnemius 62.2(±24.6) g).

However, it is evident that these muscles differ in fibre composition. The soleus has a large portion ~70 percent (%) of low twitch fibres (type I fibres) while the gastrocnemius muscle contains a higher concentrations (~50 %) of fast twitch muscle fibres (type IIB fibres) (Benjamin, Theobald, Suzuki, & Toumi, 2007; Edgerton, Smith, & Simpson, 1975; Schepsis et al., 2002). Fibre type distribution may influence contractile properties during walking (Cronin, Avela, Finni, & Peltonen, 2013).

Estimates of muscle volume by means of PCSA of muscles are considered to important measures. They may aid in the prediction of functional capacity and relative contributions of the plantar flexors to muscle force, joint torques, and thus control of joint excursions (Fukunaga et al., 2001). Albracht and colleagues (2008) used MRI, ultrasound and three dimensional (3D) modelling techniques to estimate PSCA in nine young healthy individuals. The results revealed that the soleus muscle had the majority (62(±5) %) of the PSCA compared to gastrocnemius medialis, which accounted for 26(±3) %. On the other hand, the lateralis provided the least at 12(±2) %.

Furthermore, from anatomical studies it is also shown that these two muscle structures are architecturally different in terms of fascicle lengths, pennation angles, fibre length and volume. For example, soleus has a larger PCSA (51.8(±14.9) cm^2) compared to other lower limb muscles, similar length thus shorter fibre length compared to the medial (21.1(±5.7) cm^2) and lateral (9.7(±3.3) cm^2) gastrocnemius (Ward et al., 2009). Also, Blitz and Eliot (2008) observed that the medial muscle compartment of the gastrocnemius might be of similar dimensions or even shorter than the lateral head in some cases. Based on these reports it has been suggested that anterior and posterior parts of the soleus may have different functions during contraction (Agur et al., 2003).

2.1.3 **Anatomical muscle-tendon variations**

The muscle-tendon architecture complex varies among individuals, which could affect muscle-tendon length, arrangement and therefore function. Differential location of the muscular-tendon junctions as well as junction formation and aponeuroses shape are thought to be common (Blitz & Eliot, 2007, 2008; Van Sterkenburg, Kerkhoffs, Kleipool, & Niek Van Dijk, 2011). For example, in the gastrocnemius (medialis and lateralis) muscle, Blitz and Eliot (2007), observed that some specimens (n=20 of 66 cadavers) had a direct attachment (i.e. no aponeurosis) to the underlying soleus aponeurosis. There was also noted great variation in aponeurosis length among specimens.

Any variation in aponeuroses orientation and length would create deviations in the magnitude and direction of fibre muscle bundles, and could therefore influence the muscle pennation angles. Variations in the muscle fascicles–aponeurosis interface and muscle fibre attachment to the aponeurosis will likewise affect the direction of forces transmitted from the muscles to tendon. The muscle tendon complex can also involve accessory and absence of muscles proximally and distally. Accessory soleus or third gastrocnemius fascicle group/compartment have both been noted, as have distal tendon fusions with the superficial plantaris tendon (Bergman, Afifi, & Miyauchi, n.d.; Blitz & Eliot, 2007; Van Sterkenburg et al., 2011). All of these structural alterations would cause within-muscle variability and affect the internal arrangement of the Achilles tendon. It follows that the direction of forces transmitted from the muscles to tendon would also be affected.

The formation of the Achilles tendon occurs approximately half of the length of the tibial bone (Elson et al., 2007; Tashjian et al., 2003). However, anatomical cadavers studies by Elson et al. (2007) identified that the precise structures of the muscles just proximal to the formation of the Achilles tendon vary in shape, location and orientation. In the 21 specimens (19 paired), they found five different configurations of the combined muscular junction. These variations were present in 32 % of the paired legs, meaning that even between left and right of the same individual functionally relevant variations in structures may exist. They also noticed that the junction of the gastrocnemius may be located proximal or distally in relation to the combined intersection of both muscle groups. Such variations in structure could result in differences in the tendon-muscle fascicles and perhaps modify overall muscle and tendon function.

2.1.4 **Spiralling of Achilles fibres**

As the tendon of the muscular-tendon complex becomes completely tendinous distally, the tendon fascicles arising from specific portions of the triceps surae muscle are arranged in a twisted formation. This transverse plane rotation of fibres has been consistently observed in cadaver observational studies and ex vivo measures of Achilles tendon structure and function (Cummins & Anson, 1946; Edama et al., 2014; Kelikian & Sarrafian, 2011; van Gils, Steed, & Page, 1996). Some degree of rotation is present in all Achilles but to a varied magnitude. Cummins and colleagues (1946) found that the degree of rotation could be classified into three major types according to the contributing muscle components (Figure 2.1). In the majority of the 100 legs (52 %) the authors found that fibres originating from the medial gastrocnemius twisted and made a lateral attachment to the calcaneus, whilst those from the middle portion had a rectilinear lateral location on the insertion site. Laterally originating gastrocnemius fibres had a more ventral-lateral insertion site. The soleus part seems to display an external rotation (i.e. medial to lateral), as the anterior fibres proximally have a more posterior position distally at the calcaneal attachment site (Cummins et al., (1946)), cited in Kelikian & Sarrafian (2011)).

Van Gils et al. (1996) has largely confirmed this rotation pattern. The individual tendon fascicles from the soleus and gastrocnemius are thus entangled around each other along the course of the tendon and form a complex and varying structure through which to transmit load from the triceps surae muscles to the calcaneus.

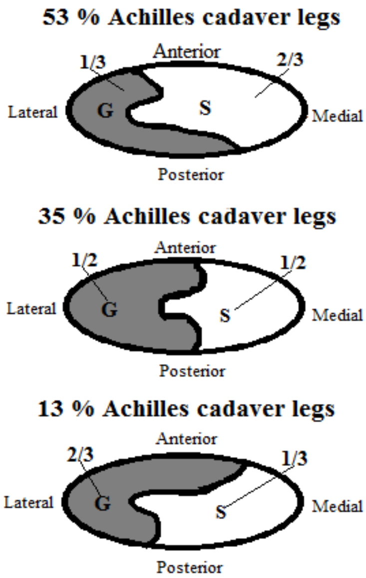


Figure 2.1. Twisting of the Achilles tendon transversal rotation (twisting) of the Soleus and the Gastrocnemius tendon fascicles' variations relatively to the calcaneal intersection. Three different configurations were observed. G=Gastrocnemius, S=Soleus, 100% =100 number of legs. Figure is adapted from (Cummins & Anson, 1946) with permission from the Journal of the American College of Surgeons, formerly Surgery Gynaecology & Obstetrics.

2.1.5 **Tendon composition**

The Achilles is a multi-layered connective structure, which mainly consists of collagen fibrils type I (65-80 %) and elastin (1-2 %) enclosed to a proteoglycan-water ground substance (Kannus, 2000). Each of these fibrils is arranged into collagen fibres, fibre bundles, and fascicles, which are covered by a paratendon to form the final structure (Comfort & Abrahamson, 2010; Maffulli & Almekinders, 2007; Nordin & Frankel, 2012) as seen in (Figure 2.2).

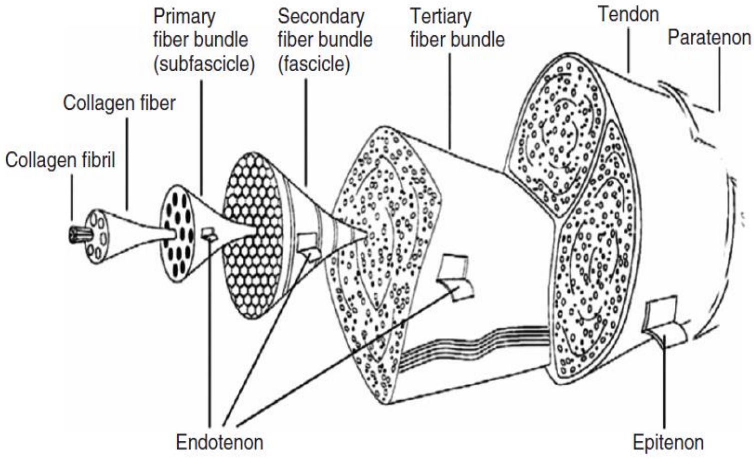


Figure 2.2. Illustration of the hierarchical organisation of the Achilles tendon. Picture taken from Smith et al. (2013) with permission Copyright © 2013, Wiley Periodicals, Inc.

This inner structure of collagen fibrils arrangement in general is also subjected to variations and at least five different twisting configurations (A-E) have been noticed (Jozsa & Kannus, 1997; Kannus, 2000), some fibrils running parallel while others twisting in variable patterns as illustrated in (Figure 2.3). This inner variations would likely also contribute to the overall twisting of the tendon fascicle of the Achilles.

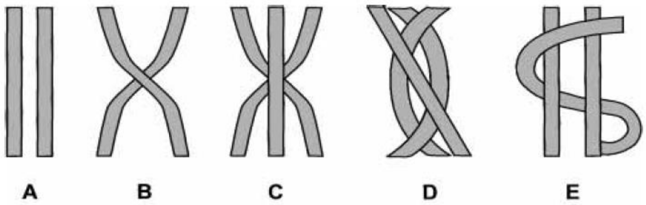


Figure 2.3. Collagen fibre-rotation variations. A = parallel structures, B = simple crossing fibres, C = crossing of two fibres, D = plait formation with three fibres, E = tying up of parallel fibres. From (Kannus, 2000) with permission Copyright © 2008, John Wiley and Sons.

The average area of fibre intersection onto the calcaneus has been reported to cover a region of 4.7 millimetres squared (mm^2) proximally and 5.1 mm^2 distally (Chao, Deland, Bates, & Kenneally, 1997; DeOrio & Easley, 2008). Variations regarding the transverse tendon fascicle formation/arrangement close to the calcaneus (approximately 1 cm) have only to some extent been described in anatomical dissections (Cummins & Anson, 1946; Edama et al., 2014; Szaro, Witkowski, Śmigielski, Krajewski, & Ciszek, 2009), thus little is known about their intersection point/areas on to the calcaneus (Ballal, Walker, & Molloy, 2014).

The average length of the tendon is approximately 15 cm (ranging from 11 to 26 cm), from its muscle-tendon junction to its distal intersection point (Apaydin et al., 2009; Del Buono, Chan, & Maffulli, 2013; O'Brien, 2005). Of this length, the gastrocnemius tendon 'portion' may extend from 11 to 26 cm whereas the tendious part from the soleus muscle is shorter from 3 up to 11 cm (Jozsa & Kannus, 1997). These numbers highlight large variations in structural features that are undoubtedly influential in load transmission during muscle contraction. For example, the ability of the tendon to elongate will be affected by its starting length and the demand on the gastrocnemius tendon component affected by the change in length in the soleus component, and vice versa.

Proximally, the tendon is wide and flat in structure with an average width and thickness of 6.8 cm and 7 mm respectively (Apaydin et al., 2009; Del Buono et al., 2013; Sadro & Dalinka, 2000). Some have suggested thickness in excess of 7 mm could indicate pathology (Mellado, Rosenberg, & Beltran, 1998; Weinstabl, Stiskal, Neuhold, Aamlid, & Hertz, 1991) although others have highlighted the effects of age, height and gender and to some extent bodyweight on tendons structures (Koivunen-Niemelä & Parkkola, 1995). The overall dimensions of the tendon decreases throughout its length distally and the tendon becomes more rounded in shape until about 4 cm above the calcaneus. Thereafter the tendon becomes flatter in form and takes on the shape of the insertion site on the calcaneus. Close to its proximal attachment to the calcaneal bone the tendon has been reported to be 3 cm wide and up to 3 mm thick (Apaydin et al., 2009; Koch & Tillmann, 1995; Sadro & Dalinka, 2000).

The Achilles tendon cross-sectional area (CSA) has been reported to vary from 0.8 to 1.4 cm^2 along the length of tendon (O'Brien, 1992, 2005), though this has also been reported to vary according to age and activity levels and types of physical exercise. Kongsgaard et al. (2005), for example, observed larger Achilles CSA in runners and volleyball players compared to a control group who practiced kayak sport. The thinnest part of the Achilles tendon, with a CSA of 0.4 to 1.4 cm^2 (Kvist, 1994; Magnusson & Kjaer, 2003) is located in the midportion of the tendon, 2 to 6 cm from the insertion of the tendon into the calcaneal bone.

2.2 Properties of the Achilles tendon

The primary role of a tendon is to join a muscle unit with its bony attachment and to transmit force generated by the physiological work of muscles to the skeleton. This load is able to create or control movements taking place and thus help co-ordination of complex motor tasks.

Tendons are primarily designed to transmit tensile loads through their passive stiffness in response to muscle and bone forces are either end of their structure. At a gross anatomical level they are thus strong in the direction of their longitudinal length. There are a very wide range of parameters of interest when describing tendon mechanical properties but the essential features are:

1. Force-Length relationship
2. Stiffness
3. Stress-Strain relationship
4. Elastic modulus
5. Maximal load capacity

2.2.1 **Force-Length relationship**

In general, this feature describes the tendons ability to stretch/lengthen (in millimetres) when the tissue is exposed to a continuous load expressed in Newtons (N). This force-length relationship provides us with information on how extensible tendon fibres are and what load the tendon can tolerate before its failure (tendon rupture). The relationship between tendon length and force is believed to be curvilinear (Figure 2.4). When load is initially applied the crimp or waviness of the collagen fibres in the fibrils starts to diminish and these become progressively straighter in structure (so called 'toe region'). Following this, a further increase in length occurs as fibres display a linear response to force (linear region). Near the upper part of the linear region the collagen might start to reach failure as some of the fibrils reach their tractable capacity. A complete failure occurs with further elongation of increased loading with irreversible changes in the tendon (Maganaris, Narici, Almekinders, & Maffulli, 2004; Nordin & Frankel, 2012).

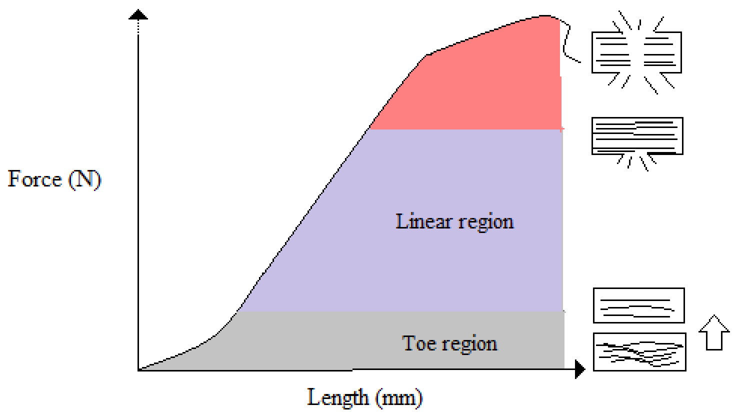


Figure 2.4. Force-Length relationship of tendons. Figure adapted from (Nordin & Frankel, 2012; Pearson, 2010).

2.2.2 **Stiffness**

Another important parameter can be derived from the force-length curve is stiffness (k) (Figure 2.4). Stiffness expressed as Newtons per millimetre (N/mm) is determined by the force and elongation ratio and describes the resistance of the tendon to a deformation in the linear region of the tendon. As load increases and the tendon becomes elongated, the tendon also becomes stiffer. For example, significantly higher Achilles stiffness rates have been observed at increased eccentric plantar flexion effort muscle contractions (Sugisaki, Kawakami, Kanehisa, & Fukunaga, 2011). Based on this work, (see Figure 2.5) the stiffness of the Achilles tendon increases by approximately two thirds during maximal eccentric contraction (eg. ECC_{max} in Figure 2.5) compared to stiffness values obtained during passive plantar-dorsiflexion movement (eg. PAS in Figure 2.5). However, others have suggested that stiffness might not be so dependent on loading rates ($N \bullet s^{-1}$) (Peltonen, Cronin, Stenroth, Finni, & Avela, 2013).

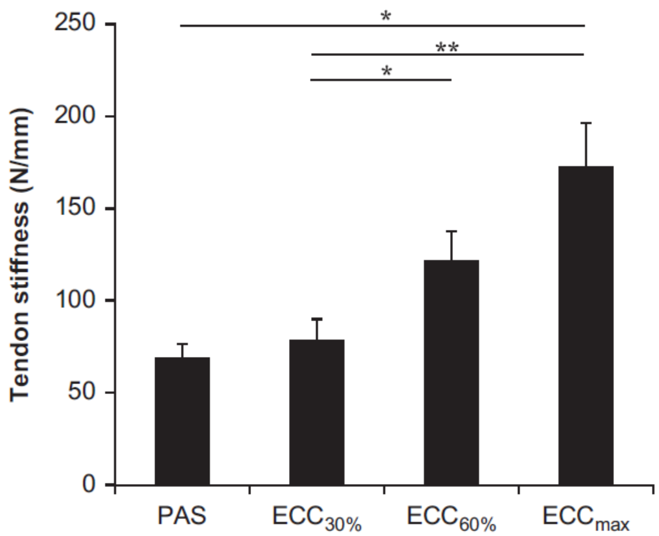


Figure 2.5. Stiffness comparisons during passive and eccentric loading at different intensities(30, 60 and 100 %). Reprinted from Sugisaki et al. (2011) with permission from Copyright © 2011 Elsevier Ltd. Positive joint ankle within this paper represent plantar flexion values.

2.2.3 **Stress-Strain relationship**

As the dimensions of the tendon are not uniform along its length nor the arrangement of its internal components consistent or uni-directional, the stress concentration transmitted will vary along the tendon length. Stress (σ) is the forces that exist within the tendon structure and is calculated as the force applied divided by the tendons CSA (expressed in N/mm^2 or N/cm^2). Stress relates closely with strain (ε), is the overall deformation of the tendon in millimetres due to the forces (stress) applied. Strain is calculated as the change in absolute length (ΔL) divided by the initial length of the tendon tissue (L_0), but is commonly expressed as a percentage (%) of length change (Nordin & Frankel, 2012). The stress-strain relationship offers information on the change in shape and resistance of the fibres when the tendon is exposed to tensile forces (Figure 2.6).

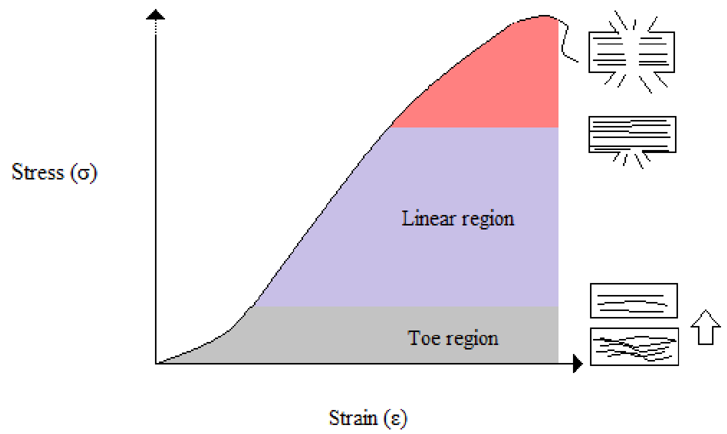


Figure 2.6. Stress-Strain relationship in tendons. Figure adapted from (Nordin & Frankel, 2012; Pearson, 2010).

2.2.4 **Elastic modulus**

The elastic or Young's modulus (E) measures the resistance (strength) to deformation and expressed in Giga Pascal or Mega Pascal (MPa). It describes the ratio of stress to the ratio of strain (as defined in Equation 2.1) within linear region of tendon deformation (Figure 2.6). It is thus a measure of the elastic properties of the tendon when in its linear response region of elongation.

Equation 2.1. Young's modulus (E) formula:

$$E = \frac{\text{Stress } (\sigma)}{\text{Strain } (\varepsilon)}$$

2.2.5 **Maximal load capacity**

In general, for all tendons, the classical values for maximal stress (σ max) (i.e. stress before complete rupture) have been reported to be 100 MPa with a corresponding tensile tissue strength (E-modulus) of 1500 MPa (Bennett, Ker, Imery, & Alexander, 1986; Butler, Grood, Noyes, Zernicke, & Brackett, 1984). Also, it is generally believed that an elongation (strain) of 8 to 10 % may lead to complete rupture of the tendon (Wang, Guo, & Li, 2012). However, these figures might be dependent on the pliancy and flexibility of tendon fibres as greater and lower ultimate tendon strain (ε max) ranges have been stated in the literature, with numbers between 4 up to 14% (Bennett et al., 1986; Butler, Grood, Noyes, & Zernicke, 1978; Devkota & Weinhold, 2003). Greater strain may induce degeneration or micro ruptures of the collagen fibres within the fibrils of the tendon. Thus, tensile tendon properties published in the literature have been suggested to vary across their anatomical function, location and structure (Ker, 2007; Wang, Iosifidis, & Fu, 2006).

Several studies (see Table 2.1) have investigated the in vitro mechanical behaviour of the Achilles tendon before tissue failure. The maximal tractive stresses (σ_{max}) tolerated by the Achilles tendon have been reported to range from 56 to 86 MPA based on in vitro cadaver material (Lewis & Shaw, 1997; Wren, Lindsey, Beaupré, & Carter, 2003; Wren, Yerby, Beaupré, & Carter, 2001). In addition to this, the two papers by Wren and colleagues (2003, 2001) highlighted that part of the tendon which had greatest potential to fail when exposed to repeated loading was located about 4 cm above the calcaneus, an area which also corresponds to the lowest CSA in this tendon (Kvist, 1994; Magnusson & Kjaer, 2003).

Table 2.1. Maximal load capacity in the Achilles tendon. Reported stress to tissue failure parameters for the human Achilles tendon in vitro cadavers during cyclic loading.

Paper	Modulus (E)	Stress (σ_{max})	Stiffness (k)	Strain (ε max)	Failure Load (N)	Subjects
Shaw and Lewis (1997)[a)]	345(±76) MPa	56 (±12) MPa	725±(253) N/mm	20(±7) %	N/A	n=16, 29 AT
Wren et al. (2001)[b)]	822(±211) MPa	86(±24) MPa	N/A	Distal tendon 16.1(±3.6) % Mid/proximal tendon 9.9(±1.9) %	5579(±1143) N	n=9 9 AT
Wren et al. (2003)[c)]	N/A	30-80 MPa range	N/A	< 10 % , n≈14 10-15 % , n≈11	N/A	n= 25 25 AT

Note. Strain loading parameter ($\%\cdot s^{-1}$) within the papers in review used [a)] 10 or $100\%\cdot s^{-1}$ [b)] 1 or $10\%\cdot s^{-1}$ [c)] $10\%\cdot s^{-1}$. All values in the table represent mean values and standard deviations *M*(±*SD*). AT= Achilles tendons in total; n =Number of subjects; N/A= Not Available data.

2.3 Forces in the Achilles during walking and running

Data from in vivo studies suggest that during gait the Achilles tendon experiences loads higher than the failure loads reported in vitro (Maganaris, Baltzopoulos, & Sargeant, 1999, 2002; Maganaris et al., 2004; Maganaris & Paul, 2002). These loads are believed to peak during late midstance phase (Novacheck, 1998).

There are three classical papers that are relied upon to describe Achilles forces during gait (Komi, Fukashiro, & Järvinen, 1992; Komi, 1990; Scott & Winter, 1990). Scott and Winter (1990) reported peak values from five running trials in three young healthy adults (two males and one female). Peak Achilles forces were 4412(±646) N (i.e. 7.2•bodyweight (BW)) were collected on the same male at different speed levels. The forces were slightly smaller for the female participant (6.3•BW). The authors reported peak stress (σ max) of 6200 N/cm^2 (e.g. 62 MPa) when the CSA was assumed to be 0.5 cm^2. These values were below the range of ~100 MPa expressed for other tendons than the Achilles (Bennett et al., 1986; D. L. Butler et al., 1984). However, this value contrasts with the failure stresses reported by Wren et al. (2001), 5098 N.

Komi (1990) measured Achilles forces on one male subject during various walking (1.2-1.8 m•s^{-1}) and running (3-9 m•s^{-1}) speeds. They used buckle transducer inserted under local anaesthesia into the tendon. Running at 6 m•s^{-1} induced the peak Achilles forces of 9000 N (e.g. 12.5•BW). With a cross sectional area 0.81 cm^2 (e.g. 8.1 mm^2) this was reported to create peak force per area (stress) of 11100 N/cm^2 (e.g. 1111 MPA in N/mm^2).

Recently, a study by Farris, Buckeridge, Trewartha and McGuigan (2012) reported lower mean peak force values of 2710(±830) N (i.e. 4.3(±7.83)•BW) in ten females of during barefoot running (3.1(±0.2) m•s^{-1}). These variations (±830 N), would result in stress differences of 166 MPa (considering small CSA=5 mm^2, 830N/5mm^2 = MPa).

Even more recently, Sinclair, Isherwood and Taylor (2014) investigated 12 young adult males running shod at 4.0(±5 %) m•s^{-1}, with and without foot orthoses. Normative peak values in this sample were reported to be 3594(±67) N (e.g. 4.8•BW). Greenhalgh and Sinclair (2014) investigated differences between genders (15 males and females, respectively) during running at similar speeds (4.0(±5 %) m•s^{-1}). They observed a significant 16 % greater peak forces among males compared to women.

During walking at slower cadences Finni, Komi and Lukkariniemi (1998) reported peak Achilles loads that were lower compared to running (five males and three females). An optical fibre was inserted through the subjects' skin approximately 2 to 3 cm above the tendon insertion. Measures were taken at three walking speeds ranges (1.1 to 1.8 m•s^{-1}) and peak forces ranged from 1320(±500) to 1490(±500) N respectively. These corresponded to peak stresses from 19 to 22 N/mm^2. In addition higher cadence increases tendon loading rates increase by 32 % (from 6570(±1810) to 9670(±3260) N•s^{-1}), but not peak forces.

That forces exerted on the Achilles tendon are greater during treadmill running than in walking have been confirmed recently by Wulf et al. (2015). They used in vivo ultrasound wave imaging in 27 healthy subjects. The principles underlying this measurement are a propagation velocity model, in which the speed of reflected ultrasound waves depends upon the mechanical properties and density of the tissue it penetrates through. This model assumes that the speed of sound is proportional to load formation in the tendon. The authors showed that the maximum velocity of ultrasound signals were significantly higher (26 m•s^{-1}) in the stance phase of running compared to walking, indicating higher forces. Indeed, they reported that the speed of transmission of ultrasound waves was correlated with higher cadence and greater ankle joint motion (which is itself associated with faster ambulation).

2.4 Force distribution within the Achilles tendon

Whilst force transmission and production in human skeletal muscle is generally relatively well understood there remains much uncertainty on how these forces are distributed throughout the tendons that transfer load to the skeleton. Haraldson et al. (2008) suggested that lateral (sideways) force distribution is negligible and that force is mainly transferred longitudinally within tendon fascicles, but independently within each fascicle. This suggests that different fascicles could have different mechanical properties and would allow for significant variation in stress distributions within specific tendons.

Indeed there are a range of structural features in the Achilles which strongly suggest regional variation in stress distribution. Firstly, as earlier sections have described, the elliptical shape of the Achilles results in variations of CSA along its length. It is therefore reasonable to assume that stress would differ along the tendon length as the transversal dimensions vary.

Furthermore, the Achilles tendon consists of three fascicles, one of each from a different triceps surae muscle portion, covered by a paratenon and thus the tendon as a whole has an organised hierarchical structure. However, as has been reviewed earlier, its internal structural arrangement and contribution of tendon fibres is variable (Cummins & Anson, 1946; Edama et al., 2014; van Gils et al., 1996). Because of such internal arrangements, it is difficult to know the exact contributions of the soleus and gastrocnemius to stress concentrations within the tendon and its sub components. As previously reported, the most frequent configuration is that medial tendon fibres arise from soleus, whereas those from gastrocnemius make a more lateral attachment on the calcaneus. An unanswered question related to this twist in the tendon fibres is whether mechanical properties differ within different tendon compartments (e.g. stress and strain within the tendon itself). In addition, the intersection of tendon fibre bundles into facets on the posterior part calcaneus may also be of clinical importance (Ballal et al., 2014) as it would influence the lever arms during ambulation. So far, only one paper has been successful in differentiating the individual tendon fascicles of the medial, lateral gastrocnemius and soleus muscles in vivo and classified these by the degree of fascicle rotation (Edama et al., (2014)) (see Figure 2.7). However, Szaro and colleagues (2009) have been able to dissect and chart the transversal proportions of the medial head of gastrocnemius into medial and lateral components (Figure 2.8 II). These two recent papers both report details of muscle-tendon fibre fascicle formation that contrast to the dated but classical ideas suggested by Cummins and Anson (1946) illustrated in (Figure 2.8 I).

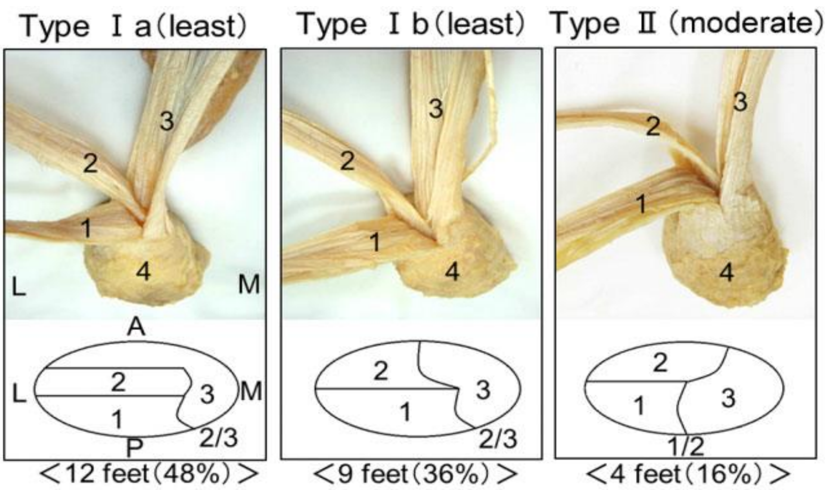


Figure 2.7. Three tendon-muscle classifications. Upper view displays the tendon fascile orientation in relation to the degree of torsion (Typ I-Type III) in posterior longitudinal direction. L=Lateral, M=Medial. Lower view shows the transversal cross section 1 cm above tubersitas calcanei of the tendon fasciles. 1: Gastrocnemius (medial) 2: Gastrocnemius (lateral). 3: Soleus. Picture taken from Edama et al. (2014), with permission Copyright © 2014 John Wiley & Sons A/S.

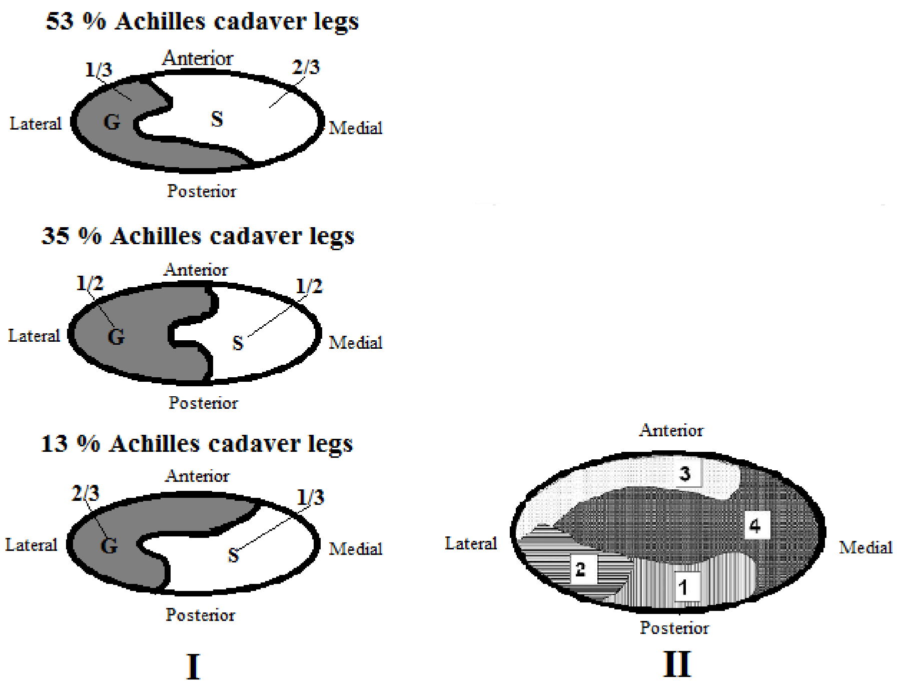


Figure 2.8. Transversal Achilles tendon cross sections1 cm above calcaneus intersection. I: From Cummins and Anson (1946) figure adapted with permission from the Journal of the American College of Surgeons, formerly Surgery Gynaecology & Obstetrics. S=Soleus. G=Gastrocnemius. II: From Szaro et al. (2009) reprinted with permission Copyright © 2015 Elsevier B.V. 1= Lateral portion of the gastrocnemius medialis, 2= Medial fibres from medial Gastrocnemius, 3= lateral Gastrocnemius fibres, 4=Soleus.

Whilst these differential distal arrangements will affect internal force distribution in the tendon, the complex aponeurosis attachments proximally, and known variations in these, will also likely affect within tendon force distribution (i.e. stress and strain). Indeed, much of the research in this area is based on the assumption that force contribution from the triceps surae is equal and force transmission from muscle fibres to the tendon via the myotendinous junction and aponeurosis is evenly distributed. However, it has been previously demonstrated by Arndt, Brüggemann, Koebke and Segesser (1999a) that non uniform stress exists in medial and lateral portions of a cadaver Achilles tendon when variable tractive forces are applied to the three isolated muscle segments of triceps surae. Medial tendon force concentrations were higher compared to forces measured in the lateral portion of the Achilles tendon when the medial head of gastrocnemius was exposed to tensile force alone. By contrast, forces were substantially higher in the lateral layer of the tendon when contractile load was applied only to both heads of gastrocnemius or the entire muscle complex (soleus, gastocnemicus medialis and lateralis) simultaneously. This strongly suggests a natural predisposition to load the lateral part of the tendon more than the medial. As tissue deformation of a tendon can be seen as a function of the muscle forces, this poses the question of whether differential loading patterns exist within the triceps surae so as to adapt to the stress distribution in the Achilles, with the objective of keeping stress distribution more equal.

Recent work by Obst, Renault, Newsham-West and Barrett (2014) using 3D ultrasound imaging during submaximal isometric plantar flexion conditions has shed further light on the regional and directional variations in stress and strain in the Achilles tendon. Three dimensional scans were obtained in eight healthy individuals with fully extended knees. The baseline condition was with the calf muscles relaxed and the ankle at 15° plantar flexion. A second condition involved a 70 % submaximal voluntary isometric plantar flexion contraction (with the ankle 15° plantarflexed). The free tendon shape was later subdivided into a percentage of the entire tendon length (0-100 %) and corresponding cross sectional representatives. Deformation parameters within the ventral/dorsal, medial/lateral of the tendon within each subsection were obtained using computed vector cross sections, allowing transversal rotations to be observed.

Their data suggest that the greatest variations and changes in CSA occur in the mid portion (±10%) of the tendon. It appears that contraction results in differential strain behaviour in this area, on average higher peak strain rates were observed in medial/lateral and ventral/dorsal portions (-14.6(±6.2) and 12.9(±6.3) %) respectively compared to more proximal and distal areas of the tendon. Also, the CSA reduced (-7.4(±7.5) %) leading to greater stress.

Overall, when loaded, the tendon becomes narrower in width in the medial and lateral direction and yet thicker in the anterior/posterior direction. Also, muscle contraction results in a smaller overall CSA. This could imply that the mid portion is a potential site for increased stress concentration due to a more narrow CSA. The results also indicate an effect of knee position on tendon behaviour, testament to the triarticular effects of the muscle-tendon complex.

2.5 Achilles Strain/Stress variations due to rearfoot movement

The analysis of strain distribution in the Achilles tendon has usually been limited to one plane of motion in vivo (e.g. sagittal plane) or by applying uniaxial force to specimens in isolated muscles or Achilles grafts during static conditions (e.g. Bojsen-Møller et al., 2004; Finni et al., 2003a; Lersch et al., 2012; Lyman, Weinhold, & Almekinders, 2004; Magnusson et al., 2003; Wren et al., 2003). Many studies have focused on tendon behaviour during plantar flexion contractions, as it is the primary plane, in which motion occurs at the ankle joint. However, the Achilles inserts medial and posterior to the centre of the ankle and rearfoot joints and thus force in the tendon creates an inversion moment. Furthermore, since inversion and plantarflexion of the heel are coupled with some adduction of the heel relative to the tibia/fibula, the heel and thus the Achilles tendon experience triplanar motion and forces during contraction of the calf muscles.

Indeed, during ambulation the foot does not move solely in one plane and motions in the frontal and transverse planes are widely recognised as normal features of rearfoot movement. Lundgren and co-workers (2008) provide bone pin derived rearfoot motion data for the heel relative to the tibia bone, perhaps the most valid description of the three separate rearfoot motions, albeit on five subjects. The reported mean (±*SD*) in total range of motion (ROM) between the heel and tibia was 17.0(±2.1), 11.3(±3.5) and 7.3(±2.4)º in the sagittal, frontal and transverse planes respectively. These movements are continuous throughout stance phase and only in some degree of synchronisation. Related work by Nester et al. (2014) on 100 participants reveals similar continuous movement patterns during stance (Figure 2.9). Also, changes in direction in different planes might lead to complex and non uniform stress in the Achilles. For example, changes from plantarflexion to dorsiflexion and abduction to adduction motion occur during 0-20 % stance, whereas in the frontal plane the heel everts continuously throughout this period.

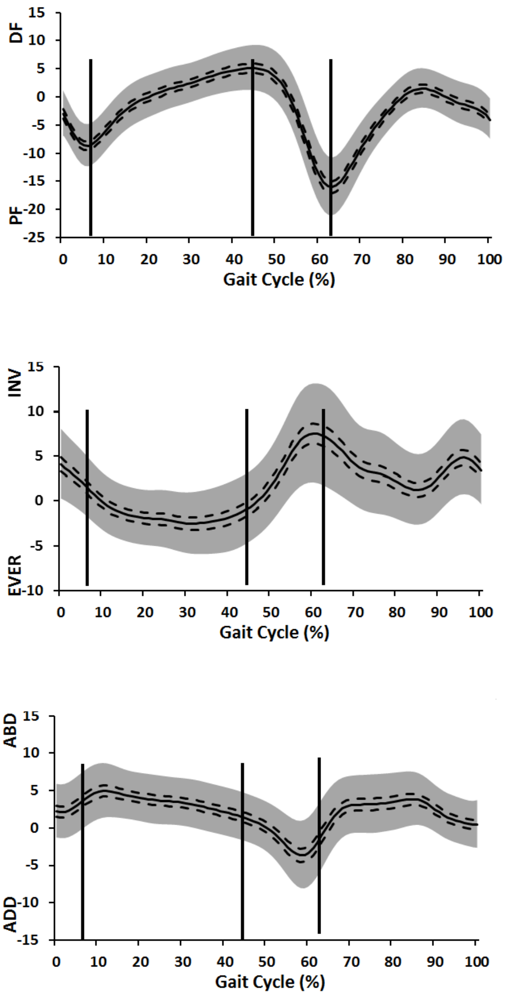


Figure 2.9. Movement of the rearfoot in 100 healthy individuals. Sagittal (top) frontal, and transverse (bottom) plane motion (°) between the heel and leg. ADD=Adduction, ABD=Abduction, EVER=Eversion, INV=inversion, PF=Plantar flexion, DF=Dorsiflexion. Figure from Nester et al. (2014).

Only Lersch et al. (2012) studied angular displacement of the calcaneus in relation to frontal plane Achilles tendon strain in vitro. This experimental evidence showed that inhomogeneous strain distribution exists in the distal/proximal and also medial/lateral regions of the tendon. Critically, these parameters were affected by different frontal plane calcaneal angular positions. Strain parameters seems to be more dependent on the angular heel position (7.5 or 15°) in the frontal than changes in loading intensities applied within triceps surae muscles. It suggests that larger eversion excursions will induce higher strain in the medial portion of the tendon which is associated mainly with the soleus rather than gastrocnemius muscle.

2.6 Summary of Achilles tendon structure and function.

The review thus far has established that the Achilles tendon has complex anatomical characteristics proximally, distally and in its mid portion. As a result the distribution of load within the tendon sub structures is complex and certainly not uniform. This leaves open the possibility that excessive stress might develop in small regions of the tendon, increasing risk of tissue damage and injury.

Both the proximal and distal attachments of the Achilles tendon and triceps surae muscles are constantly moving during gait and during periods when the muscles are active and tendon loaded. The rearfoot is closest to the tendon and has direct attachment to it, and exhibits large ranges of sagittal, frontal and transverse plane movement during gait. These triplanar movements perhaps further complicate the load distribution and stress within the tendon and could be implicated in the development of localized areas of excessive tendon stress. Likewise, modifications to rearfoot movement with footwear or orthotics might increase or decrease tendon stress. Having established these issues from the literature reviewed thus far, the Achilles injury literature will be reviewed to establish whether these facts may be relevant in Achilles injury and treatment.

2.7 Achilles injuries: structural and functional consequences

It has been suggested from a large meta analysis (n=3500) by Lopes, Hespanhol, Yeung and Pena Costa (2012) that the incidence and prevalence of Achilles injuries ranges from 9.1 to 10.9% and 6.2 to 9.5 % respectively of all running related injuries. Also, the incidence of Achilles tendinopathy or rupture seems to increase with the mode of running activity, with higher incidences rates (42 %) reported in former mid distance compared to long distance (3 %) elite runners (at least in Finland, (Kujala, Sarna, & Kaprio, 2005)). It also seems that males are more prone to injuries than females (Taunton et al., 2002).

The symptoms, proposed and suggested treatments for Achilles tendon injuries are very well described in the literature (Clement et al., 1984; Kader, Saxena, Movin, & Maffulli, 2002; Maffulli, Sharma, & Luscombe, 2004; Schepsis et al., 2002;Scott, Munteanu, & Menz, 2014; Sharma & Maffulli, 2005; Smart et al., 1980; van Dijk, van Sterkenburg, Wiegerinck, Karlsson, & Maffulli, 2011). In general, symptoms include pain, swelling and limited performance (Asplund & Best, 2013; van Dijk et al., 2011). Treatments include calf muscle eccentric loading regimes, taping, antiflammatory drug injections and ultrasound wave therapies (Grigg, Wearing, & Smeathers, 2012; Kader et al., 2002; Scott et al., 2014; Tan & Chan, 2008).

Several articles have sought to compare structural (i.e. mechanical and material properties) and functional (e.g. muscle power generation) between injured and healthy Achilles tendons (Arya & Kulig, 2010; Child, Bryant, Clark, & Crossley, 2010; Grigg et al., 2012; Kongsgaard et al., 2005). These help characterise the pathological status of the tendon structure and its functional consequences, and may point towards types of underlying etiological mechanisms.

Arya and Kulig (2010) showed the CSA of the Achilles tendon to be 65 % wider and more compliant (more flexible) in those with injured tendons. The tendons in the symptomatic group were significantly more extensible (18 %) at maximal tendon loading compared to a healthy control group (during maximum voluntary isometric contraction of the triceps surae). A reduced plantar force generation was also observed, evidence of accompanying functional deficit, although this could be due to protective mechanisms operating in the painful limbs rather than pre-existing weaker muscles. The authors proposed that stiffer tendons may explain why these individuals might be more predisposed to further injuries, although the stiffness might equally be a consequence of local inflammation or repair at the site of injury.

Child et al. (2010) also found injured Achilles tendons to be thicker (at the level of the medial malleolus) and reported strain rates that were double those in the uninjured control participants. On the other hand, no differences in plantar flexion force production was found between the two groups tested. This could be explained by compensation for the more compliant tendon by increased force production by the posterior calf muscles. Differences in force production between these two studies may also be due the different knee position used while testing (90° flexed in Child et al. (2010) versus 0° extended position in Arya and Kulig (2010)). Bojsen-Møller and colleagues (2004) have previously demonstrated the effect of knee position on soleus and gastrocnemicus function. Similarly, it has recently been shown that different knee flexion angles (30° versus 90°) induce different amounts of strain within different portions of the Achilles tendon during passive and active (eccentric) ankle plantar and dorsiflexion (Slane & Thelen, 2014).

Grigg et al. (2012) compared Achilles mid portion thickness in 11 male unilateral injured cases, to the contralateral limb of the same individuals and nine healthy participants. Not only was the injured Achilles almost twice as thick in the sagittal plane compared to the control legs, 9.4(±0.4) mm versus 5.0(±0.4) mm, the tendon was also thicker than the uninjured side of the injured participants (6.6(±0.4) mm).

In addition to structural changes in injured Achilles tendons, there are clear functional differences between those with and without AT injury. McCrory et al. (1999) studied 31 subjects with Achilles tendon tendinitis compared to a control group, and reported that they exhibited a significantly higher peak ankle dorsiflexor moment at isometric contractions at 60 degrees per second (°/s). Likewise plantarflexor moments were consistently lower at both angular velocities tested (60 and 180 °/s). They also observed a tendency for the injured subjects' plantar flexors muscles to produce less work and power. Mahieu et al. (2006) similarly identified the strength of the plantar flexors as a determining factor in the risk of the onset of Achilles tendon injury. Lower plantar flexion force strength during isokinetic muscle assessment (at 30 and 120 °/s) was found to correlate with greater incidences of the onset of developing Achilles tendonpathy in 69 military subjects.

Plantar flexion force production (power generation) in 42 Achilles tendon injured subjects was evaluated by Silbernagel, Gustavson, Thomeé and Karlsson (2006), using the uninjured limb of each participant as the control. The researchers used a more task orientated approach during weight bearing conditions rather than controlled isometric or concentric contractions (evaluated on healthy subjects (n=15) to test its validity). The test protocol consisted of jump, hopping, and single legged vertical jump tests. Furthermore, the subjects had to perform two different strength tests during unilateral heel raises with the use of a weight machine with increasing resistant modes. The researchers also noted the number of heel raises a subject was able to perform (related to endurance ability). Across all these measures a decrease in normal force generation capacity was observed. This could be a consequence of the presence of injury or lead to alternative muscle contractions strategies to combat these reductions.

2.8 Causes of Achilles injury

The underlying mechanisms by which Achilles injuries occur are not fully understood although there is general agreement that it is almost certainly multifactorial. A wide range of intrinsic and extrinsic factors have been associated with risk of injury, with varying degrees of evidence for the association.

Extrinsic factors that are thought to contribute to Achilles injury include poor running technique, inappropriate footwear, and changes in the duration or frequency of activity. Physical overload of the tendon, such as increased tensile, compression, shear forces during prolonged or unaccustomed exercise, may be contributing factors to overuse Achilles tendon injuries (Cook & Purdam, 2012). The intensity of training regimes has also been postulated as a cause of injury by several authors (Clement et al., 1984; Haglund-Åkerlind & Eriksson, 1993; McCrory et al., 1999). McCrory et al. (1999) postulated that running speed was correlated to higher injury incidence rather than the amount of running (hours) per week. However, Nielsen and colleagues found a positive association between the amount of training hours per week and risk of overuse injury (Nielsen, Buist, Sørensen, Lind, & Rasmussen, 2012).

Footwear choice may affect rearfoot motion and the external rearfoot moments that the Achilles tendon opposes. This may require the Achilles tendon to work outside its physiological range in terms of the mechanical work required. For example, Forghany et al. (2014) showed how a roll-over style shoe (with a curved heel and forefoot profile) can cause the calf muscle to activate earlier than when wearing standard footwear. This earlier activity in triceps surae would prolong the period over which the Achilles is loaded. It would also change the period instance when the muscle is loaded and thus the length of the muscle-tendon complex, because the rearfoot is in a different position when the Achilles is under tension.

Intrinsic factors are thought to include issues related to muscle function proximally (calf muscles) rearfoot motion distally, and the nature of the Achilles structure and its interface with the heel bone and calf muscles. Since the force in the Achilles is due to calf muscle contraction, differences in triceps surae activity during different activities has long been thought to be a causative factor. The external rearfoot moments that the Achilles opposes are also sensitive to gait style, specifically the foot strike patterns, most notably in runners. Runners who contact the ground with the midfoot or forefoot first apply an external dorsiflexion moment to their ankle compared to those using a heel strike pattern (which is more typical of walking) (Hasegawa, Yamauchi, & Kraemer, 2007; Kasmer, Liu, Roberts, & Valadao, 2013; Larson et al., 2011). This requires eccentric loading of the Achilles after initial contact and thereafter concentric calf muscle contraction later in stance. Compared to use of a heel contact pattern this requires the Achilles to operate across a wider range of functional situations, as the calf muscle action switches from eccentric to isometric to concentric.

It has previously been described within this thesis that the Achilles tendon consists of distinct fascicles arising from the soleus, medial gastrocnemius and lateral gastrocnemius muscles with variable configurations (Ballal et al., 2014; Cummins & Anson, 1946; Edama et al., 2014; Szaro et al., 2009). Haraldsson et al. (2008) suggested that the tendon fascicles are likely to behave as functionally independent structures in terms of force transmission. If differences in muscle fascicle behaviour exist this would likely lead to asymmetrical stress distribution along the length of the tendon but also non-uniform stress distribution within different tendon portions, resulting in regional differences in tissue compliance (strain). There is some evidence that variations in Achilles fascicle behaviour do occur. For example, Ishikawa et al. (2005) demonstrated that the medial fascicles of gastrocnemius contracted either isometrically or concentrically during early push off phase, whereas the soleus displayed largely eccentric behaviour. Both muscles appear to contract concentrically just before toe off. In contrast, however, others have reported more consistent isometric patterns in medial gastrocnemicus and soleus muscle fascicles during the midstance phase of walking (Cronin et al., 2013). Such different muscle behaviour will result in different tendon strains, and not surprisingly differences in the activation patterns within the triceps surae muscles have long been implicated in Achilles injuries. Electromyography (EMG) is the primary measure of muscle activity during gait and can give indications of the timing of triceps surae activity and with motion data relate these to knee or rearfoot movements. However, EMG signals are not directly related to muscle forces nor stress within the Achilles. The latter is an elastic response of the tendon to loads applied and dependent on many anatomical factors, as well and rearfoot position, movement and rates of loading too.

However, EMG still offers useful insight into calf muscle and Achilles function. Four studies have investigated muscle activity in cases of Achilles injury (Azevedo, Lambert, Vaughan, O'Connor, & Schwellnus, 2009; Baur et al., 2004, 2011; Wyndow, Cowan, Wrigley, & Crossley, 2013). Baur and colleagues (2004) used surface electrodes on tibialis anterior, peroneus, both heads of gastrocnemius and soleus in a comparative study between eight participants with Achilles pain and 14 controls. All subjects ran on a treadmill at a fixed velocity (3.3 m•s^{-1}) in barefoot and shod conditions. The EMG results showed no large nor systematic differences between Achilles injured and controls in the onset and offset timing events of all muscle groups, regardless of barefoot or shod conditions. The only statistically significant difference (p=0.00) was in the magnitude measured in millivolts (mV) of the lateral gastrocnemius activity during load acceptance phase (between initial contact and loading response). In the Achilles injured individuals there was a reduced amplitude (-25 %) compared to healthy controls. The authors suggested that these reductions in could be a possible strategy to relieve pain in the affected structures. However, the tendon fascicle of gastrocnemius lateralis inserts mostly on the lateral site of calcaneus and pain is most often experienced on the medial side (Edama et al., 2014; Szaro et al., 2009). Interesting, a related study by the same group found significant reductions (p=0.001) in gastrocnemius medialis, during similar test protocols but in a larger comparative study. The mean amplitude values in the symptomatic group were 11.5 % lower than in healthy individuals (Baur et al., (2011).

A recent meta-analysis by Munteanu and Barton (2011) confirms decreased activation of gastrocnemius lateralis (Cohens (*d*) effect size, d=-1.50) during initial stance of running. It also seems that when the original data by Baur et al. (2004) were converted to (Cohen's *d*) effect sizes and 95 % confidence interval (CI), that activity of the lateral and medial gastrocnemius when running in shoes is increased prior to propulsion in subjects with Achilles tendinopathic pain (d = 0.69 and d = 0.86 respectively). Also, the activation pattern of the gastrocnemicus lateralis is prolonged in symptomatic group throughout stance. Unfortunately, the later study by Baur et al (2011) wasnot included in the meta-analysis by Munteanu and Barton (2011). On the other hand, Azevedo et al. (2009) was included in the Munteanu and Barton (2011) review and they reported no significant differences for the amplitude of lateral gastrocnemius pre-and post-heel strike between those with and without Achilles injury.

Recently, Wyndow and colleagues (2013) similarly found that triceps surae activation was significantly altered in those with mid portion Achilles problems (n=15) during running compared to a symptom free control group (n=19). Larger timing differences between soleus and the lateral gastrocnemius were found in the symptomatic tendons. The EMG activity of lateralis occurred earlier (18 $m{\bullet}s^{-1}$) compared to simultaneous timing in healthy individuals.

Differences in action (contraction type due to foot strike patterns) and force (e.g. different timing of muscle activity) within the muscles could produce differences in the stress and strain within the Achilles tendon, increasing risk of injury. If this were coupled with a rearfoot position that further affected the distribution of stress and strain in the tendon, risk could be further elevated though a combination of factors. The contribution of rearfoot position and movement to risk of Achilles injury is now reviewed in more detail.

2.9 Rearfoot movement as a risk factor for Achilles injuries

Different factors and clinical concepts have been hypothesised to explain how rearfoot motion might cause Achilles injury. The published hypotheses and the experimental evidence supporting or refuting each is reviewed on the following sections.

2.9.1 Frontal plane rearfoot motion and Achilles injuries.

Over pronation (i.e. excessive eversion) of the subtalar joint and rearfoot complex has long been assumed to be a factor which can lead to excessive loading across the Achilles tendon. The implication is that rapid (too fast) or prolonged pronation (being everted when the rearfoot should not be) during midstance alters the load in the tendon as the foot moves into a supinated position during late stance. This is the period when the triceps surae are most active. Thus, the time at which the Achilles force is greatest might coincide with an abnormal rearfoot position and thus elevate risk of damage to the tendon.

This concept has also been termed “bowstringing” of the Achilles, or in the case of rapid rearfoot eversion, “whipping”. Both these hypotheses assume that the movement , speed or pattern leads to stress in the tendon being concentrated at specific locations (Clement et al., 1984; Maffulli et al., 2004; Schepsis et al., 2002). For example, if the heel is everted then tension would be increased on the medial side of the tendon since the distance between origin and insertion of the tendon would be increased. In contrast, the lateral side might experience a reduction in tensile forces. The “bowstringing” theory relates to this tension-slacken effect on the medial and lateral sides of the tendon. The “whipping” hypothesis relates to fast eversion, too rapidly increasing forces on the medial side of the tendon.

In 1979, Bates and colleagues published one of the first papers discussing kinematic variations in Achilles tendon injuries. Their early data has to some extent formed the basis of the hypothesis made by others (Clement et al., 1984; Smart et al., 1980). Based on these now old but still untested theories, much of the clinical literature continues to assume that excessive eversion is a determining factor in the onset of Achilles injuries (e.g. Hreljac, Marshall, & Hume, 2000; Hreljac, 2004; Huerta, Moreno, Kirby, Carmona, & García, 2009). This is despite only three studies investigating the frontal plane excursions in subjects with Achilles pathology compared to appropriate controls (Munteanu & Barton, 2011).

The first of these studies, McCrory et al. (1999), evaluated foot and ankle kinematics alongside exercise history, anthropometric data, muscle strength/endurance, and ground reaction forces during running to determine if any of these were linked to

Achilles injures in 31 runners. Data were compared to a non-injured control group (n=58). Two dimensional (2D) rearfoot kinematic analysis indicated that runners with Achilles injuries demonstrated significantly greater inversion at initial contact (+2(±0.8)°) and increased peak pronation (0.6(±0.2)°) and higher pronation velocity (2.2(±20.2)°/s) during running compared to healthy individuals. Greater velocity might affect the eccentric work by muscles and strain rates too.

That eversion is associated with an Achilles injury is supported to some degree by Ryan and co-workers (2009). They compared barefoot rearfoot kinematics in 27 subjects with a history of mid-portion Achilles injury to that of 21 healthy controls running at self-selected speeds. Individuals with injuries showed a significant increase (+2°) in the range of rearfoot eversion. Despite the poor effect sizes the authors also noted a tendency for injured subjects to have greater frontal motion excursion (+4°) compared to the control group. Although not measured the authors attributed some of the effects to the increased varus position at heel strike, assuming that this would increase the external eversion moment (Ryan et al., 2009;Clement et al 1984).

Donoghue, Harrison, Coffey and Hayes (2008) found that individuals with Achilles injuries (n=12) displayed 2-5° greater ankle dorsiflexion, knee flexion and eversion angles during treadmill running compared to a matched control group. Moreover, similar to McCrory et al. (1999) injured subjects had an increased inversion angle at initial contact. However, a major limitation of this study is that the investigators confined their criteria to those Achilles injured runners with demonstrable excessive pronation, whereas the asymptomatic control group was not matched for levels of over pronation (Donoghue, Harrison, Coffey, & Hayes, 2008).

Adding to these previous papers, Baur et al. (2011) have recently suggested that poor activity of the peroneal muscle (i.e. limited eversion control) found in cases of Achilles injuries may be associated with increased ranges of subtalar eversion during midstance.

Whilst these studies have reported changes in frontal plane kinematic that are supportive of the “bowstringing” and to a lesser extent “whipping” hypotheses (which is perhaps more concerned with velocity of pronation rather than range), the relationship between frontal plane heel position and stress distribution in the Achilles tendon is unproven. Thus, even if true, the importance of the frontal plane differences in cases of Achilles injury is unknown. Whilst it seems logical that eversion will increase tensile forces on the medial side of the tendon (Clement et al., 1984; Smart et al., 1980), this effect may not be consistent throughout the range of rearfoot movement. This is because the eversion of the heel is a consequence of complex movements at several joints. This might lead to changes in the centre of rotation and axes around which heel motion takes place during the eversion of the heel relative to the leg.

The hypothesis that subtalar joint axis position or orientation may affect Achilles stress has been presented in the clinical texts of Kirby (Kirby, 2001; Kirby, 2010). He suggested that abnormal deviations of the subtalar joint axis may increase and decrease the lever arm of tendons such as the Achilles. This may affect the forces produced by the muscle connected to the tendons and thus forces experienced by the tendons.

Applying this concept to Achilles injuries specifically, Reule and colleagues (2011) reported that the medial deviation (β) angle (i.e. from the frontal plane) of the subtalar axis was 6 to 10° greater in runners with Achilles injuries (n=95) compared to healthy and gender matched controls (n=212) displayed in Figure 2.10. A more medially deviated axis would reduce the moment arm for the Achilles and thus require greater muscle forces to produce the same rearfoot moment, thus increasing tendon forces too. It will also increase the contact area lateral to the axis and thus increase external pronation and dorsiflexion moments, which the Achilles helps to oppose.

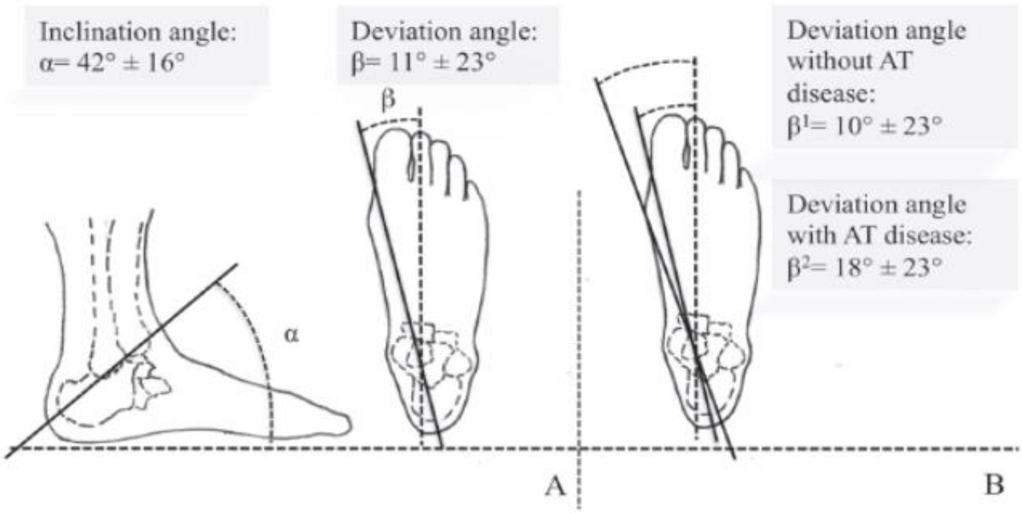


Figure 2.10. Subtalar axis orientation in Achilles injured and healthy individuals. The figure illustrates that the spatial orientation of the subtalar axis is different is different in subjects without and with Achilles tendon disorders (AT). A= mean inclination angle (*α*) and mean deviation angle (*β*), B= mean deviation angle difference between AT and control. Reprinted from Reule (2011) with permission Copyright © 2011, BMJ Publishing Group Ltd.

In general the evidence suggests that there are some differences in frontal plane position and movement during gait in those with Achilles injuries, but these are not conclusive. Also, what differences might exist are relatively small, in the region of <3°. Furthermore, whether this reflects causation or a consequence of symptoms is unclear. Some of the reports covered earlier indicate increased compliance (reduced stiffness) is also a consequence of Achilles tendon disorder and this might influence rearfoot kinematics and explain some of the proposed increased rearfoot eversion. Overall, whilst frontal plane rearfoot motion may be a contributing factor to risk of Achilles injury the evidence base for this remains weak.

2.9.2 Transverse plane rearfoot motion and Achilles injuries.

The range and asynchronous timing of foot pronation with tibial rotation in late midstance is thought to increase torsional Achilles forces and disrupt normal distribution of stress in the Achilles tendon. Transverse plane forces and movement between the tibia and heel is thought to create a *'wringing mechanism'*, which twists and strains the Achilles tendon (Clement et al., 1984). This would change the structural arrangement of the already twisted tendon fibres along the length of the tendon, and do so during a period of gait when it is bearing load. The proposed effect is to create localised areas of elevated stress and thus elevated risk of injury.

The foot and leg are mechanically coupled through bony and soft tissue mechanisms and changes in foot alignment and loading are part of a kinematic and kinetic interaction with the entire lower limb. During stance foot pronation, calcaneal eversion, and the corresponding transverse plane motion of the talus and thus tibia are all coupled (Dierks & Davis, 2007; Eslami, Begon, Farahpour, & Allard, 2007; Rodrigues, Chang, TenBroek, van Emmerik, & Hamill, 2015). Thus, when the foot is on the ground foot pronation occurs simultaneously with 'internal' tibial rotation. In the case of prolonged eversion into midstance and even propulsion, failure to resupinate the foot whilst the tibia externally rotates is thought to lead to altered transverse plane alignment of the Achilles tendon and therefore altered stress distribution.

Whilst the existence of coupling between the leg and rearfoot is a long standing concept, recent research suggests that there can be some independence in the motion patterns (Pohl, 2006; Pohl, Messenger, & Buckley, 2007). For example, during walking, the heel everts for a large part of the stance phase and during this period the tibia can begin to externally rotate independent of rearfoot inversion.

The transverse plane position of the foot and leg might also be relevant for the line of action of the calf muscles and thus the forces applied to the origins and insertion of the Achilles. Internal rotation of the leg relative to the abducted, everted and dorsiflexed (i.e. pronated) foot would position the medial head of the gastrocnemius in a relatively posterior position. This would theoretically change the line of action of this muscle component. As reported earlier, muscle orientation and action might itself be a strong contributor to risk of Achilles injury. Perhaps muscle action proximally and foot position distally are both interacting to influence the stress experienced in the Achilles.

Despite these concepts and their popularity there is limited evidence of differences in transverse plane motion between those with and without Achilles injuries. Two studies have measured transverse plane rearfoot motion and both found no differences in internal nor external rotation of the tibia relative to the foot (Donoghue, Harrison, Coffey, et al., 2008; Ryan et al., 2009). In a related issue, Williams III et al. (2008) investigated eight runners with a history of Achilles injury and compared these to runners without a history of Achilles injury. They reported significantly less internal knee rotation (-4(±1)º) and less external knee rotation moment (–0.3(±0.4) vs. –0.8(±0.3) Nm/kg). Altered knee rotation could affect tibial-foot position and force production within the medial and lateral components of gastrocnemius, which could result in uneven stress distribution along the length of the tendon. However, this is the only report to identify transverse plane differences in cases of Achilles injury.

2.9.3 **Sagittal plane rearfoot motion and Achilles injuries.**

Abnormal dorsiflexion motion at the rearfoot has also been reported as a factor associated with the onset of Achilles tendon injuries. This is the primary plane of motion for the rearfoot and the primary plane of action for the triceps surae, and is perhaps the most logical plane of motion to be associated with risk of injury.

Clement et al. (1982) was one of the first to suggest that a limited ROM at the talo-crural joint due to tightness of soleus-gastrocnemius complex may be responsible for increases in Achilles strain. A reduced ability to gain sufficient dorsiflexion throughout stance would therefore lead to compensatory movement at the knee (increased knee flexion) or/and greater subtalar eversion (Bates, Osternig, Mason, & James, 1979; Clement et al., 1984). The latter would then contribute to the 'whipping', 'bowstringing' and 'wringing' concepts and be indicative of a very stiff Achilles tendon. Despite the lack of evidence presented in the publications, these dated works continue to be cited in numerous reviews and used as evidence of ankle dorsiflexion as a risk factor for Achilles injury (e.g. Cook & Purdam, 2012; Kvist, 1991; Maffulli & Almekinders, 2007; Schepsis et al., 2002; Wyndow et al., 2010).

A prospective cohort study (n=449) showed that subjects who demonstrated lower static range of ankle dorsiflexion were at greater risk of Achilles tendonpathy compared to those with more ankle motion (Kaufman et al., 1999). This was supported by a more recent study by Rabin et al. (2014). They followed 70 healthy male military recruits during a six month training scheme and those developing Achilles injury (n=5) were found to exhibit greater limitation of ankle dorsiflexion (21.1(±6.1)°) compared to those who did not develop an Achilles injury (27.9(±5.6)°). It seems from these studies, with larger cohorts, that limited sagittal plane rearfoot motion, albeit measured during a static and non weight bearing, could be implicated in risk of Achilles injury. However, in contrast to the two studies above, Mathieu and co-workers (2006) observed in another prospective study (n=69) increased dorsiflexion was cited a factor in the onset of Achilles injuries.

2.10 Use of foot orthoses in Achilles injuries

Clinical management of Achilles tendon injuries in terms of orthoses remains controversial. This is because the data describing the biomechanical effects of orthoses has mainly focused on motion changes (i.e. kinematics) rather than changes in the Achilles tendon strain/stress or forces. Thus, changes in kinematics have not been put into context of the effects on the tendon tissue itself.

The conventional paradigm for orthotic use in cases of Achilles injury is based on the hypothesis that excessive frontal plane motion or misalignment of the rearfoot is causing the Achilles injury and thus requires correction. This is largely based on the texts from Clement et al. (1984) and Smart et al. (1980) which have already been described. Foot orthoses have therefore been designed with specific features in the heel and medial arch areas that limit foot pronation and alter frontal plane rearfoot movement. The changes in plantar pressure required to reduce foot pronation have been reported when wearing foot orthoses (Bonanno et al., 2012; Redmond, Landorf, & Keenan, 2009; Redmond, Lumb, & Landorf, 2000). In changing how external loads (e.g. plantar pressures) are applied to the sole of the foot, the orthoses should change the internal moments acting at the rearfoot joints, and thereafter change the internal moments that structures like the Achilles must contribute to. Since the moments acting at the rearfoot joints will change, the expectation is that rearfoot motion changes too.

The assumption that orthoses change foot motion, plus the assumption that incorrect rearfoot movement is implicated as a cause of Achilles injuries, has lead researchers to focus on changes in frontal plane motion as a primary measure of orthotic effect in Achilles injuries. However, there are remarkably few studies exploring this issue.

Donoghue, Harrison, Laxton and Jones (2008b) investigated the kinematic response to custom moulded foot orthoses during treadmill running in 12 individuals with chronic Achilles injuries. Surprisingly, with the orthoses, the investigators found significant increases in peak eversion angles, the range of eversion motion and a decrease in the eversion angle at initial contact. However, the authors did not explicitly define the orthoses used in terms of material nor shape, and thus the actual orthotic studied is unknown. Important information that is missing includes the amount of 'pronation control' in terms of any medial heel wedges used, amount of calcaneal alignment correction and arch support which were added but not specified. Interestingly, despite the adverse kinematic outcomes and the clinical results subjects reported an average of 92 % relief of symptoms with the orthoses, there was no control group against which to compare these improvements.

A later study also by Donoghue et al. (2008) examined the effects of foot orthoses (arch support in combination wih a medial wedge) in subjects who demonstrated over pronation during running and reported Achilles injuries. In this second study injured participants were compared to a control group of asymptomatic runners. The findings pointed to the orthoses having an effect in reducing peak dorsiflexion angle at the ankle as well the ability to reduce dorsiflexion ROM during midstance. This is one of the few reports of changes in sagittal plane motion, which seems relevant as an adjunct to evaluation of changes in frontal plane motion due to the orthoses. However, overall the motion profiles were similar to those of the control group. As in their previous work, wearing orthoses tended to increase the peak eversion angle and cause a greater rearfoot eversion position at touch down compared to no orthotic condition.

The kinematic effects of triplanar foot inserts (i.e combined heel lift and medial wedge) had earlier been investigated by some of these authors (Harrison, Laxton, & Bowden, 2001) on a very small sample of subjects with chronic Achilles tendonitis (n=6). They stated that orthoses had only significant effects on reducing the range of dorsiflexion motion (p=0.048) in the sagittal plane with no kinematic effects in the frontal plane.

To date only one study has investigated muscle activity in cases of Achilles injury and the effect of foot orthoses on this muscle activity. No differences were found in the onset or offset timings among the three triceps surae muscles while subjects were running with orthotics (Wyndow et al., 2013).

In the absence of other studies on orthotic effect in Achilles injured individuals, the wider literature on foot orthotic effect is relevant. In this respect the reviews and meta analyses of Mills et al. (2010) and Cheung, Chung and Ng (2011) are very pertinent. These both sought to pool data from the existing literature and perform meta analyses to arrive at an overall conclusion regarding whether foot orthoses affect frontal plane rearfoot movement. Both studies concluded that rearfoot eversion is reduced by foot orthoses, with mean reduction of 2.1° estimated by Mills et al. (2010), and reduction of 2.2° estimated by Cheung et al. (2011). Furthermore via their meta-analysis Mills et al. (2010) reported only 0.17° difference in the reduction in eversion due to orthoses in injured participants compared to non-injured participants. This could mean that data on samples of healthy participants is transferrable to populations with symptoms, though this likely assumes the underlying foot kinematics are not different between injured and uninjured participants.

Indeed, Rodrigues et al. (2013) have recently shown that orthotic response was not different between those with and without anterior knee pain.

Also, the literature concerning the effect of orthotics on peak forces or strain within the Achilles tendon has also been limited and only involved healthy individuals (Dixon & Kerwin, 2002, 1998; Farris, Buckeridge, et al., 2012; Reinschmidt & Nigg, 1995; Sinclair et al., 2014). For example, Sinclair et al. (2014) has recently reported on estimated tendon forces and changes in these due to orthoses. Kinematics and kinetics for the foot relative to the shank were collected for the sagittal plane. Ground reaction force data was used to compute internal and external moments around the ankle. Achilles tendon load was later estimated from the planar flexion moment and divided by an estimated Achilles moment arm. The moment arm was described as function of sagittal plane angle. The mean result was a reduction in peak Achilles forces of 0.4•BW (i.e. ~ 230.2(±67.4) N) due to the orthosis. The authors conclude that increased peak dorsiflexion angle in the orthoses condition may account for reduced forces in the tendon.

Early reports on the effect of heel lifts (range 7.5–15 mm, and 5-9.5° of elevation) suggested increased tendon loading (Dixon & Kerwin, 1998) or no substantial effect (Reinschmidt & Nigg, 1995) or subject specific effects reductions (Dixon & Kerwin, 2002). More recently heel lifts (e.g. 12 and 18 mm) have been reported to have no effect on muscular activations patterns while running barefoot in healthy samples (Farris, Buckeridge, et al., 2012).

2.11 Summary literature review on Achilles' injury and treatment.

The review thus far has established that when injured the Achilles tendon and the associated muscle units undergo a series of structural and functional changes that reduce its capacity for normal function. It is unclear whether some of these deficits pre-existed the injury or are entirely a consequence of the injury.

A precise or singular cause for Achilles injury seems unlikely since many factors affect the tendon behavior. Foot kinematics in the frontal, transverse and sagittal planes have been suggested as causative mechanisms, but there is limited evidence supporting these hypotheses. Regardless, foot orthoses are used as a mechanical treatment for Achilles injuries and have been shown to affect frontal plane rearfoot motion in ways that should influence the frontal plane stress in the Achilles tendon. However, how orthoses induce changes in frontal plane rearfoot motion which affect internal Achilles tendon factors has not been explained. Since frontal plane rearfoot motion is implicated in the cause of Achilles injuries, understanding the relationship between frontal plane rearfoot position, movement and Achilles stress/strain would be valuable.

2.12 Aims and Objectives of this MPhil thesis.

The literature review has revealed many important aspects of our understanding of Achilles injuries, the proposed causes and the potential role of foot orthotics in treatment of Achilles injuries. Of the areas covered, two interrelated gaps in our knowledge are of particular note:

- Rearfoot eversion is commonly implicated as a risk factor for Achilles injury, but how eversion/inversion affects distribution of load in the Achilles tendon is unknown.

- Evidence suggests that foot orthoses change the frontal plane movements of the rearfoot by ~<2.5°, but it is not clear how these changes will affect the distribution of load in the Achilles tendon.

The aims of this MPhil thesis was therefore to investigate the relationship between rearfoot eversion/inversion and stress distribution in the Achilles. The objective in meeting this aim was to identify implications for how (1) foot movement and (2) use of foot orthoses might affect Achilles injury.

To meet these aims two experimental studies were defined.

Study 1) This will measure the effect of a foot orthosis on frontal plane movements of the heel relative to the leg during walking and running.

Study 2) This will measure changes in the displacement experienced by the medial and lateral sides of the mid portion of the Achilles tendon during eversion and inversion of the foot. This will test the hypothesis that eversion increases displacement on the medial side of the Achilles tendon and therefore might be implicated as a risk factor for Achilles injury.

The changes in eversion angle due to the orthosis in study 1 will be combined with the data in study 2 to explore how important the change in eversion is for the distribution of strain in the Achilles. This study assumes there is a relationship between displacement of the tendon and strain and stress within the tendon.

3 Effect of foot orthoses on frontal plane rearfoot kinematics

3.1 Introduction

The literature review has indicated that rearfoot eversion is commonly implicated as a risk factor for Achilles injury, although how rearfoot eversion/inversion motion affects distribution of load in the Achilles tendon is not known. This chapter is therefore concerned with understanding the frontal plane changes in rearfoot kinematics due to a foot orthosis. This aims to add value to our understanding of how foot orthoses might affect the biomechanical work of the Achilles tendon and thus both risk of injury and any therapeutic effect of foot orthoses. The data from this chapter will be used with data from the subsequent chapter to put any change in frontal plane rearfoot position due to orthoses into an Achilles stress context.

3.2 Aim

The aim of this study was to quantify the effects of foot orthoses on frontal plane rearfoot motion during walking and running.

3.3 Hypothesis

H_0: Foot orthoses have no effect on rearfoot eversion during walking and running.

H_1: The foot orthoses reduce rearfoot eversion during walking and running.

3.4 Methods

A within subjects design was chosen in which the participants served as their own controls and thus walked and ran with and without the foot orthoses. This study had approval from the University of Salford ethical committee. All participants were given verbal information about the experiment and written consent was obtained.

3.4.1 Participants

Thirty three symptom free subjects ranging from 18 to 45 years of age were recruited from staff and the student population of the University (weight range 58-99 kg, height range 161-191 cm). Recruitment was made via advertisements within the department. The inclusion criteria were that subjects were involved in running activities on a weekly basis and were self reported as healthy. Exclusion criteria included history of systemic musculoskeletal disease (e.g. rheumatoid arthritis, osteoarthritis diabetes etc.), surgery that might affect foot posture and biomechanics, prior use of foot orthoses, and current lower limb pain.

3.4.2 Sample size considerations

A prior power analysis using a program (G*Power version 3.0.10) was conducted based on previously reported reductions in rearfoot eversion due to anti pronation orthoses. The angle data were extracted from the existing literature during walking and running where mean and standard deviation of difference was stated or could be calculated manually by the investigator (Faul, Erdfelder, Buchner, & Lang, 2009; Faul, Erdfelder, Lang, & Buchner, 2007).

Mündermann et al. (2003a) reported the mean and standard deviation (2.3(±0.4)º) of reductions in maximum foot eversion while running at a fixed speed (n=20 pronated feet). Maclean, Davis and Hamill (2008) looked at the immediate orthotic effects on rearfoot eversion angles (reduced by 1.56(±0.6)º) in twelve female runners. Nigg and colleagues (2003) found peak eversion of the shoe relative to leg to be significantly different (n=15) due to anti pronation orthoses. A full length medially posted orthosis reduced peak shoe eversion angle (by 1.5(±1.3)º). Based on this data, which is one of the smallest effects reported, the power analysis revealed that a sample size of nine subjects was needed for a statistical power of $\beta \geq 0.80$ significant alpha level $\alpha = 0.05$.

3.4.3 **Foot orthosis and footwear**

An anti pronation foot orthosis was chosen based on prior experience within the research group. This was a full length pre-fabricated orthotic (salfordinsole-FIRM (Shore A70)) comprising medial longitudinal arch support, heel cup but not plantar wedging under the heel or forefoot, and made in individual foot sizes (Figure 3.1). The orthotics were fitted by a qualified Orthotist (the candidate) to assure correct sizing.

Figure 3.1. Anti pronation foot orthoses. The foot orthoses had medial arch profile and heel cup, made in individual foot sizes.

All participants wore the same footwear. This was a New Balance neutral running shoe (Type M536SR New Balance sports shoes, UK) with an ethylene vinyl acetate mid-sole and blown rubber outer sole. Orthotic and shoe sizing was matched for each individual.

3.4.4 **Measurement of rearfoot motion**

Motion of the heel relative to the leg was recorded by attaching reflective markers to the leg and heel whilst subjects walked and ran with and without the foot orthosis. These markers were all attached by the same investigator. Kinematic data was collected using ten Qualisys ProReflex cameras (100 hertz (Hz) Qualisys AB, Gothenburg, Sweden).

To define the 3D position and movement of the shank four reflective markers were attached to a plastic plate on the lateral side of the leg. To define the 3D position and movement of the heel a triad of markers was screwed into a plastic disc attached to the lateral side of the participant's heel. This triad protruded through an aperture in the heel counter of the shoe (see Figure 3.2) so that the motion recorded was that of the heel bone not the shoe. The alignment of the triad of markers was kept consistent between shoe only and shoe plus orthotic conditions by keeping the plastic disc in situ throughout and using a locating pin to position the triad consistently.

Figure 3.2. Triad marker. This marker was attached to the heel through the heel counter of the shoe.

To provide anatomical references for the leg and heel local coordinate systems, additional anatomical markers were placed over the femoral condyles, malleoli, posterior heel and second metatarsal head during a static standing trial (Figure 3.3). The participant wore the shoe but no orthotic for this anatomical landmark calibration (static standing trial), this posture was maintained for one to two minutes and later used to define 0° in the kinematic data.

Figure 3.3. Marker placement. The figure illustrates the marker placement for heel and leg technical markers, and femoral, malleoli and foot anatomical markers viewed from the sagittal plane.

3.4.5 **Protocol**

Kinematic and kinetic data capture during walking and running was conducted at the Gait and Human Performance Laboratory, University of Salford, United Kingdom. Participants walked/ran over three AMTI force plates (500 Hz, AMTI, Watertown, MA 02472, USA) to detect initial contact and toe off and passed through infrared timing gates (10 meters apart, within a 20 meters long walkway) to measure speed.

Prior to commencement of any dynamic data capture, the subjects were instructed to walk or run on a pathway (running track) going through the capture volume of the motion capture system. They were instructed to walk and run at their comfortable self-selected speed for about from three to six minutes before their initial test condition. Timing gates (Brower

Timing Systems, USA) were used to monitor speed and an average was taken during this period as a representative of a target speed for walking and running. This time period was also used to adjust the participants' starting position to ensure that the left foot landed on the force plate first, without encouraging participants to target the force plates. During actual data collection subjects were verbally instructed to either increase/decrease their speed if they failed to meet the target speed (±5%).

A minimum of ten gait cycles were collected in each of the four conditions (walk shod, walk orthotic, run shod and run orthotic. A static standing trial was also collected during which technical and anatomical markers were in place on the legs, and the participant wore the shoe. On average, the preparation and data collection for a single subject took approximately 60 minutes to perform.

3.4.6 **Data processing**

Raw reflective marker coordinate data were labeled within Qualisys' software (Qualisys, ProReflex, Gothenburg, Sweden). Each motion trial was visually inspected to ensure that markers showed minimal movement outside the expected smooth trajectories. Furthermore, the vertical component (*Z*) of the force plate data was used to manually define heel strike (initial contact) and toe off for the left and right foot separately. The marker data were then exported as coordinate 3D files into Visual3D™ (C-Motion Inc.) to derive segmental angle data.

A third polynomial order interpolation procedure was performed, allowing ten frames to be the highest number which could be replaced with interpolated values. The raw kinematic data were then filtered with a fourth order low-pass bidirectional Butterworth filter, set at 30 Hz for running and 15 Hz for walking. Cut-off frequencies were set in order to keep 95 % of the raw signal frequency content.

The foot model adopted for the kinematic analysis consisted of two rigid body segments: (1) shank (tibia and fibula), and (2) rearfoot (calcaneus, incl. subtalar and talo-crucal (ankle) joints). The shank was modelled as a single rigid segment assuming no movement between fibula and tibia. The foot model assumed equal contributions from ankle and sub talar joints, or at least the individual contributions could not be derived.

The shank and heel segments were each represented by a local coordinate system. The vertical (*z*) axis of the local shank frame was defined using the knee and ankle centres (midpoints between femoral condyles and malleoli respectively). The anterior/posterior shank axis (*y*) was perpendicular to a plane defined by the femoral condyle and malleoli markers. The medial/lateral shank axis (*x*) was perpendicular to the other two shank axes.

The heel local frame orientation was set such that in relaxed standing the *z* (vertical) axis was perpendicular to the floor (*XY* of global coordinate system) and the *y* (anterior/posterior) axis was perpendicular to *x*-axis but parallel to the markers on heel and second metatarsal in the static standing trial. The *x*-axis was perpendicular to the other two axes. Joint rotations (heel relative to shank) were calculated in Visual3D™ (C-Motion Inc.) using Cardan angles sequence (x-y-z). The model had six degrees of freedom.

All data were normalised to 0-100 % of stance phase, 0 % being first contact, 100 % being toe off, and 0º was the joint position in relaxed standing in the shoe only condition. The ten gait cycles were then averaged to provide a single time series data set for each participant in each of the four experimental conditions:

1. walking shod
2. walking shod with the orthotic
3. running shod
4. running shod with the orthotic

3.4.6.1 Subject exclusion

Some subjects were excluded due to issues with data quality, such as missing markers or unseen errors in marker movement between conditions (i.e. not identified during testing). Two subjects were removed entirely (i.e. all data lost). Additionally, walking data for two participants, and running data for two different participants were also lost due to similar marker problems. Thus, the final data was from 29 participants for walking, and 29 subjects for running.

3.4.6.2 Kinematic parameters

An existing matrix laboratory programme script, was modified and implemented by the candidate, in order to extract discrete parameters from the data sets for each trial and subject during running and walking (Table 3.1).

The primary orthotic effects on frontal plane rearfoot motion during walking have been reported through meta-analysis by Mills et al. (2010) and Cheung et al. (2011). These are mainly peak heel eversion in midstance (F2) and total eversion excursion (F6) and have been reported extensively in prior research (e.g. Branthwaite, Payton, & Chockalingam, 2004; Eng & Pierrynowski, 1994; Maclean et al., 2008; MacLean, McClay Davis, & Hamill, 2006; McCulloch, Brunt, & Vander Linden, 1993; Mündermann, Nigg, Humble, & Stefanyshyn, 2003a, 2003b; Stacoff et al., 2000, 2007; Williams, Davis, & Baitch, 2003; Zifchock & Davis, 2008). There are far fewer reports of kinematic changes during running compared to walking, and only one paper reported any notable change in kinematics: the frontal plane position at heel strike (F1) (McCulloch et al., 1993). To these parameters others representing running equivalents of walking parameters, or maxima/minima values were also added. Table 3.1 lists all parameters extracted for statistical analysis.

Table 3.1. Frontal plane kinematic parameters.

Code	Definition
F1	Frontal plane angle at heel strike (°)
F2	Maximum eversion angle in stance phase (°)
TF2	Stance time (%) of F2
F3	Maximum eversion angle (°) within ≤ 30% of stance phase (walking only)
TF3	Stance time (%) at F3
F4	Angle position at toe off (°)
F5	Total frontal plane excursion angle (°)
F6	ROM between heel strike (°) (F1) and maximum eversion angle (°) (F2)
F7	ROM from heel strike (°) (F1) to maximum eversion angle (°) (F3) within ≤ 30% of stance phase (walking only)
F8	ROM from maximum eversion angle (°) (F2) to toe off angle (°) (T4)

3.4.6.3 Statistical analysis

Statistical tests were performed using IBM SPSS version 19.0 (SPSS, Inc., Chicago IL). Initially, normal (Gaussian) distribution of data was determined using the *Shapiro-Wilks test* for normality within the IBM SPSS software. If normal distribution was verified across the conditions/groups, a *Paired t-test* was used to compare angles in shod and orthotic conditions. Normal distributed data is presented as group means *M* and standard deviation (±*SD*). Results were also reported as differences between the two group's mean (*M* differences between groups) and standard deviation (*SD* of differences). Alpha level for statistical significance was set to α=0.05 and for the probability of committing a type I error. In addition *to p*-values, the data were supported with a 95% confidence interval (CI) for the difference between two conditions means (*M* differences) (Altman, Machin, & Bryant, 2000).

If normal distribution was violated in any of the parameters, a *Wilcoxon signed-rank* test was carried out using IBM SPSS software. In these non-parametric cases data each condition was also presented as sample mean *M* and standard deviation (±*SD*). Results were also shown as differences between two groups' means (*M* differences) and standard deviations (*SD* of differences). Thus, for statistical correctness median (*Mdn*), 25th and 75th quartiles and the interquartile range (*IQR*) were computed in excel for the between intervention conditions mean differences (Altman et al., 2000). In addition to this, 95 % CI was calculated for the difference between two group's mean scores (*M, reduction*) according to Altman et al. (2000). The level of significance was also coupled to alpha (α) level of 0.05.

In addition to basic statistical tests Cohen's (*d*) and effect size (*r*) were calculated to measure the magnitude of effect of the orthotic intervention between two groups/conditions (Becker, 2011). The effect size for Cohens (*d*) was considered to be small, $d = 0.2$, medium, $d = 0.5$ and large, $d = 0.8$ according to Cohen's (1998) definitions by (Becker, 2011).

In the case of some significant differences ($p<0.05$) a post hoc (retrospective) power analysis was obtained from G*Power (version 3.0.10). Computations were made on mean and standard deviation differences between interventions in order to estimate the power (1-β) (set at 0.80 and α=0.05, two-tailed) and therefore also the likelihood of accepting a false result (type II error). These numbers may be used as guidance for the sample size needed for future studies in order to obtain similar results.

3.5 Results

Results are presented for one side only (left) to ensure data are independent samples. Figure 3.4 and Figure 3.5 shows the kinematic data for the sample (n=29) in each of the two conditions. The interquartile range including the first and third quartile and median for the mean of differences between shod and orthotic conditions (non-parametric results) is listed in the Appendix 1 (Table A.1) and (Table A.2).

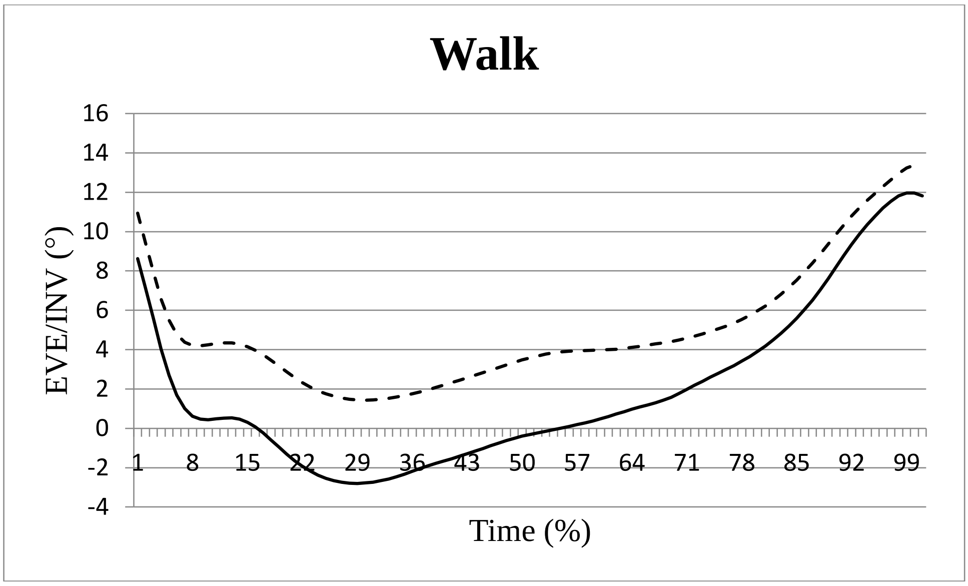


Figure 3.4. Frontal plane rearfoot movement during walking.The broken line represents the orthotic condition, and the solid line represents the shod condition. INV=inversion, EVE=Eversion.

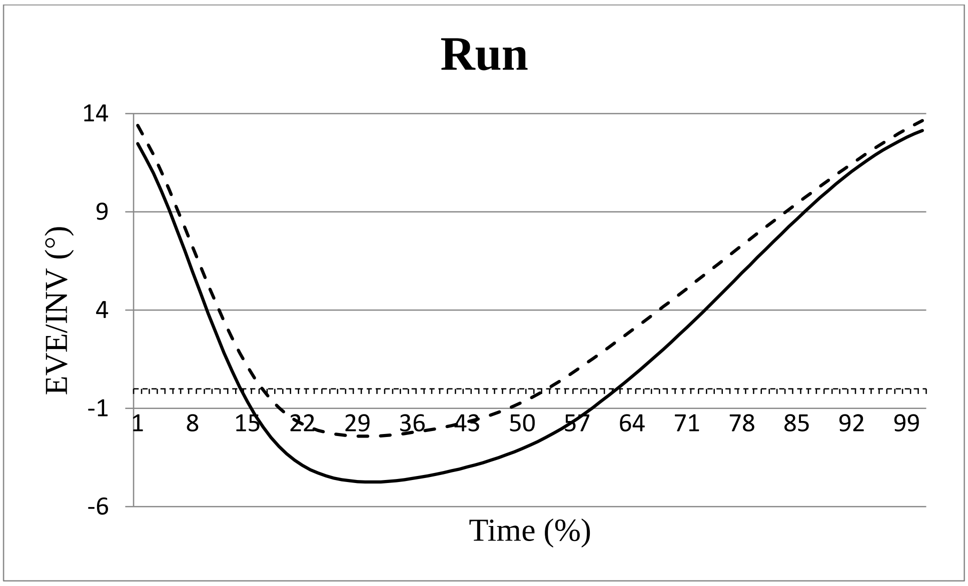


Figure 3.5. Frontal plane rearfoot movement during running. The broken line represents the orthotic condition, and the solid line represents the shod condition. INV=inversion, EVE=Eversion.

3.5.1 **Walking kinematics**

Rearfoot frontal plane motion showed significant differences in almost all parameters when walking with the orthoses (see Table 3.2). The position of the heel at initial contact was significantly different, $p < 0.001$, with the subjects showing a more inverted heel position with the orthosis (10.9(±6)°) compared to the shoe only condition (8.6(±4.8)°). On average, they were 2.3° more inverted at initial contact. The effect size for this comparison was medium ($d = 0.43$). Similar differences were observed at toe off, in which the orthotic exhibited greater inversion with the orthosis, Z (28)=-2.5°, ($p<0.05$).

Subjects demonstrated highly significant reductions in peak rearfoot eversion, mean reduction 3.9°, while wearing the orthosis ($p<0.001$). Cohen's effect size value ($d = 0.8$) suggested a large clinical significance. In addition, maximum eversion was found to occur 5.6 % later in stance with the orthosis ($p<0.01$). This prolonged eversion was present in the 74.7 % of the sample. Similar differences in subjects' response to the orthosis were found when evaluating peak eversion during the initial 30 % of stance, however these reductions were 8.2 % (0.32°) greater than the reduction in the maximum eversion taken during the entire stance phase. All frontal plane excursions were significantly reduced by the orthosis during stance ($p<0.001$).

A retrospective power analysis revealed power of (1-β) > 0.98 at an alpha level of 0.05 for sample size n=29 (for all statistically significant parameters). Poor post hoc test power (1-β) < 0.80 was found in the timing events for peak eversion in stance and rearfoot position at toe off.

Table 3.2. Frontal plane parameters for shod and orthotic conditions in walking, and outcomes of corresponding statistical tests.

WALK			**Shod**	**Orthotic**	*Orthotic effect*	95 % CI for difference		Statistical outcomes			
Code	Definition	Unit	*M*(±*SD*)	*M*(±*SD*)	*M*(±*SD*) reduction	Lower	Upper	Test	*p*-value	*d*	*r*
F1	HS position	°	8.6(±4.8)	10.9(±6)	2.3(±2.1)	1.5	3.1	t=5.9	p=0.000***[a]	0.43	0.21
F2	Peak eversion (MaxEv)	°	-4.4(±3.9)	-0.5(±5.7)	3.9(±3)	2.7[c]	5.1[c]	Z=-4.4	p=0.000***[b]	0.80	0.37
TF2	Peak eversion time	%	34(±13.6)	39.6(±19.8)	5.6(±15.1)	1.2 [c]	7.6 [c]	Z=-2.8	p=0.004**[b]	0.33	0.16
F3	Peak eversion <30%	°	-3.7(±4.2)	0.5(±5.7)	4.2(±2.8)	3.2	5.3	t=8.3	p=0.000***[a]	0.84	0.39
TF3	Peak eversion time <30%	%	24.7(±6.4)	24.8(±6.3)	0.1(±6.9)	-2.2 [c]	1.5 [c]	Z=-0.5	p=0.639[b]	0.01	0.01
F4	Toe off (TO) position	°	11.8(±6.4)	13.4(±7.8)	1.6(±3.1)	0.4 [c]	2.9 [c]	Z=-2.5	p=0.013*[b]	0.23	0.11
F5	ROM stance	°	18(±5.7)	15.6(±5.4)	-2.4(±3)	-3.5 [c]	-1.0 [c]	Z=-3.5	p=0.000***[b]	-0.44	-0.21
F6	ROM from HS to MaxEv	°	13(±3.6)	11.4(±4)	-1.6(±2.4)	-2.5	-0.7	t=-3.6	p=0.001**[a]	-0.42	-0.20
F7	ROM from HS to MaxEv <30%	°	12.4(±3.8)	10.4(±4.5)	-1.9±(2.3)	-2.8	-1.1	t=-4.6	p=0.000***[a]	-0.47	-0.23
F8	ROM from MaxEv to TO	°	16.2(±5.8)	13.9(±6)	-2.3(±2.7)	-3.2 [c]	-1.2 [c]	Z=-3.6	p=0.000***[b]	-0.39	-0.19

Note. *M* =Mean value; *SD*=Standard deviation; CI= Confidence interval; LL= Lower confidence interval limit; UL= Upper confidence interval limit; *d*= Cohen's d; [a] paired *t*-test; [b] Wilcoxon signed rank test; [c] CI computed based on Altman et al. (2000); *t*= Paired t-test critical value; Z= Wilcoxon signed rank test critical value; *p*= significance level *p< 0.05. ** p<0.01. **p<0.001. For further details, see Table 3.1 for abbreviations and definitions of the kinematics parameters.

3.5.2 **Running kinematics**

During running subjects were landing in a more inverted position at touch down with the orthosis (13.4(±6.6)°) compared to shoe condition only (12.5(±5.7))°, (*p*=0.028). Differences in maximum eversion during stance were highly significantly different (p<0.001). Peak eversion was reduced by 37.6% (2.3(±2.9)° of difference) while running with the orthosis. The effect size (*d*) for this parameter was medium. The total range of frontal plane motion was reduced by the orthosis (19.8(±6)°) compared to the shoe only condition (21.4(±5.9)°). There were no differences in the timing of peak eversion between the two conditions (see Table 3.3). The post hoc power calculation showed high statistical power (1-β) > 0.85 in all parameters with exception from the position of the heel at initial contact.

Table 3.3. Frontal plane parameters for shod and orthotic conditions in running, and outcomes of corresponding statistical tests.

RUN			**Shod**	**Orthotic**	*Orthotic effect*	95 % CI for difference		Statistical outcomes			
Code	Definition	Unit	*M*(±*SD*)	*M*(±*SD*)	*M*(±*SD*) reduction	Lower	Upper	Test	*p*-value	*d*	*r*
F1	HS position	°	12.5(±5.7)	13.4(±6.6)	0.9(±2.2)	0.1	1.8	t=2.3	*p*=0.028*[a]	0.15	0.08
F2	Peak eversion (MaxEv)	°	-6.1(±5.4)	-3.8(±5.8)	2.3(±2.9)	1.2	3.4	t=4.3	*p*=0.000***[a]	0.41	0.20
TF2	Peak eversion time	%	35.8(±9.9)	34.3(±10.8)	-1.5(±7.3)	-4.3	1.3	t=-1.1	*p*=0.288[a]	-0.14	-0.07
F4	Toe off (TO) position	°	13.2(±6.5)	13.7(±7.5)	0.5(±2.8)	-0.6	1.6	t=1.0	*p*=0.340[a]	0.07	0.04
F5	ROM stance	°	21.4(±5.9)	19.8(±6)	-1.6(±2.5)	-2.5[c]	-0.6[c]	Z=-3.1	*p*=0.001**[b]	-0.27	-0.13
F6	ROM from HS to MaxEv	°	18.5(±4.4)	17.2(±4.4)	-1.4(±2.4)	-2.3	-0.4	t=-3.0	*p*=0.005**[a]	-0.31	-0.15
F8	ROM from MaxEv to TO	°	19.2(±6.5)	17.4(±6.7)	-1.8(±3)	-2.5[c]	-0.7[c]	Z=-3.2	*p*=0.001**[b]	-0.27	-0.13

Note. *M* =Mean value; *SD*=Standard deviation; CI= Confidence interval; LL= Lower confidence interval limit; UL= Upper confidence interval limit; *d*= Cohen's d; [a] paired t-test; [b] Wilcoxon signed rank test; [c] CI computed based on Altman et al. (2000); *t*= paired t-test critical value; Z= Wilcoxon signed rank test critical value; *p*= significance level **p*< 0.05. ** *p*<0.01. ***p*<0.001. For further details, see Table 3.1 for abbreviations and definitions of the kinematics parameters.

3.6 Discussion

3.6.1 Kinematic changes due to the orthoses

The mean reduction in rearfoot eversion due to the anti pronation orthosis was 3.9° for walking and 2.3° for running. These effects are greater than mean differences estimated in the two meta analyses within the literature: 2.1° for running from Mills et al. (2010), and 2.2° for running and walking from Cheung et al. (2011). The latter review may have included studies using 2D kinematic data. McClay and Manal (1998b) have described how 2D measures might induce mathematical discrepancies within the frontal plane angles, e.g. abduction of the foot might create out of plane movements and distort the frontal plane measures. This may account for differences between the outcomes of the two reviews. The greater effect of the orthoses found here is perhaps due to differences in orthotic shape between studies, and the relatively harder material used in this study.

The mean reductions in maximum eversion were greater in walking than running. It seems reasonable to assume that the impact of the orthosis may be reduced during running as forces exerted on the body are greater (e.g. Farris, Buckeridge, et al., 2012; Finni et al., 1998; Greenhalgh & Sinclair, 2014; Komi, 1990; Scott & Winter, 1990) and the same orthosis might therefore have a more limited effect. However, the reductions found here were still higher than comparable reductions in running reported by Mills et al. (2010), and Cheung et al. (2011) for other orthoses.

The reduced range of eversion motion in walking and running are in line with earlier reports that orthoses reduce the range of frontal plane eversion (e.g. Eng & Pierrynowski, 1994; MacLean et al., 2006). Poor to medium effect sizes (*d*) were found with these variables, which is comparable to effect sizes reported by Mills et al. (2010).

On the basis of the results the hypothesis (H_0) investigated is accepted. Foot orthoses do reduce rearfoot eversion in terms of peak angles and the overall position of the rearfoot throughout stance. The effect occurs in walking and to a lesser extent in running.

3.6.2 Implications of the orthotic effect on hypothesised regional Achilles strain

If, as the literature review has indicated, there is a relationship between frontal plane rearfoot motion and Achilles injury, due to eversion increasing strain in the medial aspect of the Achilles tendon, then the results here would support the idea that orthoses could assist in reducing regional (medial) strain in the Achilles tendon. The reductions in peak eversion by 3-4° represent in the region of 20-30 % change in the total eversion movement during 0-20 % of stance (for example). If strain in the Achilles tendon changes in a linear way as the rearfoot moves from its maximum inversion at heel strike to a peak eversion position at 15-20 % of stance, the 3-4° reductions could lead to a corresponding significant reduction in medial tendon strain.

3.6.3 Limitations relevant to the interpretation of the results

3.6.3.1 Healthy versus symptomatic subjects

One limitation of this work is that symptom free individuals were investigated. Whilst, the association between foot motion, such as hindfoot eversion, and certain clinical manifestations are often assumed (e.g. Patella-femoral pain, Barton et al., (2011), Achilles injury, Munteanu et al., (2011), Plantar callus formation, Findlow et al., (2011)), the kinematic differences compared to healthy controls are often small and unsystematic. In a similar manner, heel eversion has been also implicated to play a role in the development of plantar fasciitis (e.g. Wearing, Smeathers, Urry, Hennig, & Hills, 2006). Although the kinematic features of those with plantar pain is notably different from those without any symptoms. In terms of the orthotic effect Mills et al. (2010) have pointed out that mean differences in kinematic changes due to foot orthoses between those with and without symptoms were small (0.17°). In addition, Rodrigues et al. (2013) have recently shown that orthotic response was not different between those with and without anterior knee pain.

There is some direct evidence that individuals with and without Achilles symptoms do not differ in terms of rearfoot kinematics. For example, McCrory et al. (1999) reported that peak eversion values in cases with Achilles (1.93° vs. 2.56°, respectively) were not different from their own healthy control sample. Also, there was only a slight (0.4°) and non statistically significant difference in eversion range of motion between healthy and Achilles subjects.

Qualitatively, the frontal plane rearfoot kinematics of the participants in this study are a good match to those with a range of foot symptoms in terms of both temporal characteristics of rearfoot motion and values of peak eversion (Rodrigues et al., 2013;Barton et al., 2011;Munteanu et al., 2011;Findlow et al., 2012).

3.6.3.2 The effect of foot type on rearfoot mechanics

A further potential limitation is that the participants foot type was not filtered. It could be argued that anti pronation foot orthoses are used in people whose feet show signs of over pronation. However, there is no clear definition of over pronation and there is no clear definition of the foot type associated with Achilles tendon injuries. As stated above, the rearfoot kinematics of those with Achilles injury do not differ significantly from other and symptom free feet.

However, some feet will exhibit a static or dynamic shape and kinematic patterns that means there is perhaps a limited capacity for a prefabricated orthosis to affect frontal plane motion. For example, if the medial arch of a foot is very high and unlikely to contact the arch geometry of the orthosis, the orthosis is unlikely to have any effect on foot motion. The sample investigated here might include individuals with such feet. However, if this is true the implication is that the results therefore underestimate the change in rearfoot motion. There is no evidence to suggest the relationship between frontal plane effects of the orthotic tested would be different in different foot types.

3.6.3.3 Gait evaluation of the immediate effects of orthotic treatment

This study only looked at immediate effects of an anti pronation orthotic during walking and running, and participants were not habitual orthotic users. This aspect of the research design is common to most of the research conducted on the kinematic and kinetic effects of foot orthoses (e.g. Branthwaite et al., 2004; Maclean et al., 2008; Williams et al., 2003; Zifchock & Davis, 2008). Few studies have addressed the long term effects, only a very limited number of studies have looked at prolonged kinematics effects of orthotics use including wear times from five to six weeks (Maclean et al., 2008; Mündermann et al., 2003a, 2003b). It is possible that use of orthoses on a long term basis induces further kinematic changes. For example, Woodburn et al. (2003) conducted a large controlled randomised study extending over a 2.5 year period in total. The authors examined the effect of custom made orthotics in the cases of painful rheumatoid arthritis (n=98). Rearfoot kinematic data was obtained for the transverse, frontal and sagittal planes and collected at 0, 12 and 30 months. They found that rearfoot kinematics showed immediate changes, but also that the reductions in rearfoot eversion got greater between 0 and 12, and between 12 and 30 months. Thus, immediate effects may not fully reveal the kinematic changes due to orthoses.

3.6.4 Wider clinical relevance of the kinematic changes reported

Whilst the kinematic effects reported here are in line with other studies they might be interpreted as small in the context of the overall ROM available at the rearfoot. Indeed, several authors have interpreted effect sizes similar to those reported here to be "small" and questioned whether such changes in kinematics have any clinical relevance (Butler, Davis, Laughton, & Hughes, 2003; Nigg, Khan, Fisher, & Stefanyshyn, 1998; Nigg et al., 2003; Stackhouse, Davis, & Hamill, 2004; Williams et al., 2003) and during walking (Branthwaite et al., 2004; Pascual Huerta et al., 2009; Zammit & Payne, 2007). The relatively small changes in angular position have led some to propose that changes in kinetics (forces) rather than kinematics due to orthoses are more important. Few research papers have investigated the effects of orthotics on both kinematics and kinetics, though certainly kinetics are affected by orthoses. Williams et al. (2003) found that whilst an orthosis had no effect on rearfoot kinematics during running (i.e. peak eversion and eversion excursion) the internal rearfoot inversion moment was significantly reduced. The decrease in internal inversion moment is supported by the study by Maclean et al. (2006) who investigated the effects of custom molded orthoses with five degree varus rearfoot posting (in healthy individuals during running). Nester et al. (2003) examined the effect of 10° medially and laterally wedged foot orthoses in a group of healthy individuals during walking in their own shoes (n=15). They observed that a medial wedge increased the maximum adduction moment at the ankle during mid-stance.

3.6.5 **Individual variation in changes in rearfoot kinematics**

It is interesting to note that four subjects during walking and seven in running exhibited little movement change or increased rearfoot eversion. These are sufficiently distinct from the mean effects to be worth discussion because these are the opposite of the expected biomechanical effect. Increases in peak eversion and eversion range of movement due to foot orthoses have been observed by others during running. For example, Williams et al. (2003) observed that about half of subjects (n=11) showed tendencies for greater rearfoot eversion while running with an inverted orthotic. The explanation for these nil or opposite effects is the same as the factors that affect any type of response, such as underlying foot shape and kinematic pattern. Thus, these observations might simply reflect a normal range of responses given the diversity of feet investigated and the use of one single orthotic design.

Furthermore, only one of the four participants showing a nil or opposite response in walking was the same as the participants showing this effect in running. Thus, a person might show different orthotic effects depending on whether they are walking or running. Differences in foot strike patterns or plantar loading might explain different responses in walking and running.

3.7 Conclusion

This chapter was concerned with understanding the frontal plane changes in rearfoot kinematics due to a foot orthosis. The orthoses tested produced a systematic shift in foot position, with the rearfoot becoming less everted throughout walking and running stance. If regional strain in the Achilles tendon is affected by frontal plane rearfoot position and movement, then foot orthoses that affect frontal plane rearfoot motion may influence regional strain in the Achilles tendon.

4 Effect of change in the frontal plane rearfoot position on regional displacement in the Achilles tendon

4.1 Introduction

The literature review in Chapter 2 identified that rearfoot eversion is commonly suggested as a risk factor for Achilles injury. The hypothesis for this is that rearfoot motion affects how load is distributed within the Achilles tendon. This, coupled with the anatomical factors that suggest there are regional variations in strain within the Achilles tendon, would increase strain in specific parts of the tendon and increase the risk of tendon tissue damage. However, the nature of the relationship between rearfoot eversion/inversion and the distribution of load in the Achilles tendon is unknown. Understanding this relationship is therefore important in the context of understanding whether rearfoot position and movement is relevant in the development of Achilles injury risk.

The literature reviewed in Chapter 2 also indicated that foot orthoses are commonly used to reduce rearfoot eversion in the belief that this will reduce regional strains within the Achilles tendon. Data collected and interpreted in Chapter 3 clarified the kinematic effects of foot orthoses, and revealed that foot orthoses:

- Produce a systematic shift in the frontal plane foot position throughout walking and running stance
- Reduce peak rearfoot eversion during walking and running
- Reduce the range of rearfoot eversion during walking and running

Based on the outcomes from the combined Chapters 2 and 3, if regional strain in the Achilles tendon is affected by frontal plane rearfoot position and movement, then foot orthoses should affect this regional strain.

Chapter 4 focuses on understanding whether regional strain in the Achilles tendon is affected by frontal plane rearfoot motion and position. This will help inform hypotheses related to Achilles injury and help us understand any potential biomechanical and clinical benefit from the use of orthoses.

4.2 Aim

The aim of the study was to investigate changes in displacement (strain) within the medial and lateral parts of the Achilles tendon during passive pronation and supination of the rearfoot. The purpose was to better understand regional stress in the Achilles tendon and how it changes with rearfoot position (using displacement within the tendon as an indirect measure of tendon stress). The purpose of this study was also to put the changes in kinematics due to foot orthoses (from Chapter 3) into a tissue stress context.

4.3 Hypothesis

H_0: Displacement on the medial side of the Achilles tendon will differ from that on the lateral side of the tendon during passive rearfoot pronation/supination motion.

H_1: Displacement on the medial and lateral side of the Achilles tendon will not differ during passive rearfoot pronation/supination motion.

4.4 Methods

Stress in the Achilles tendon was not directly measured since this would have involved invasive techniques. Instead the strain (displacement (stretch/shorten)) in the medial and lateral parts of the Achilles tendon was measured during passive rearfoot movement. This stretching (i.e. lengthening) and shortening displacement of tendon structures provides a surrogate measure of the stress experienced by the tendon tissue. The assumption was that lengthening (stretch) of the tendon was indicative of increased tensile stress in the tendon structure. Conversely, that shortening of the tendon due to elastic recoil after previous elongation would indicate a reduction in tensile stress in the tendon. The tendon displacements were measured using ultrasound and rearfoot motion induced and controlled using a motorised platform.

4.4.1 **Participants**

Experiments were conducted on the right leg of 22 males, ranging from 23 to 51 years of age (weight range 63.5-106 kg, height range 160-189.5 cm). Following ethical committee approval, participants were recruited from the staff and student communities of the University of Salford. Prior to testing the subjects were given an information sheet and signed a consent form. Participants were included if they were self-declared healthy and injury free, had no prior history of Achilles pain or symptoms, no prior rearfoot injuries, lower limb surgeries, and self reported as physically active.

4.4.2 **Experimental set up**

4.4.2.1 Dynamometer setup and foot position.

Tendon displacement data was collected whilst participants sat on a dynamometer (Type KC125AP; Kin Com, Chattanooga, TN) with their leg and foot attached to a motorised system that facilitates controlled passive motion of the foot through pronation and supination cycles. Prior to any data collection the dynamometer was set in its standard position, the person seated and adjustments made so the sole of the right foot could be attached to the footplate whilst the ankle was in a neutral position (i.e. 90° between leg and sole of the foot). All measurements were carried out barefoot.

To prevent reflective markers on the heel (used to measure rearfoot motion simultaneously with tendon displacement data) from hitting the dynamometer mechanism it was necessary to elevate the foot up from the standard dynamometer footplate. Multiple blocks of 1 cm thick ethylene vinyl acetate material were placed securely on top of the footplate. An insole was placed on top of the ethylene vinyl acetate sheets to provide some stabilization for the sole of the subject’s foot and help with consistent foot location (see Figure 4.1 below). The foot was secured onto the foot plate using velcro straps.

The seat and foot plate system were further adjusted to align the participants knee in 30° of flexion (similar knee positions have been used during passive and active ultrasound tendon examinations,(e.g. Slane & Thelen, 2014), whilst maintaining the 90° position of the ankle. A handheld goniometer was used to assure that proper alignment was achieved at the knee and ankle. This final position was then used as the start position for all data collection.

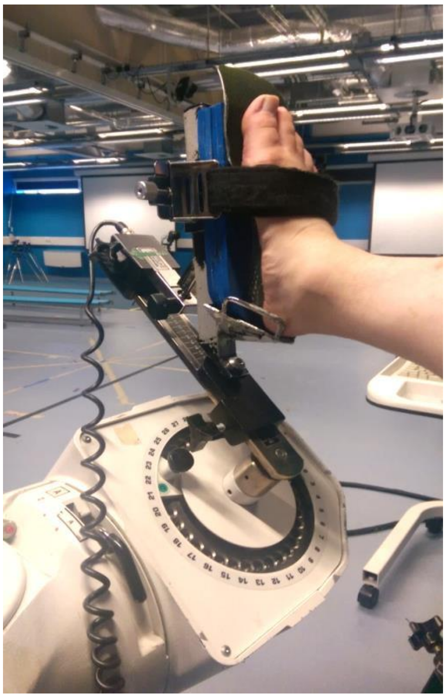


Figure 4.1. Foot positioned on foot plate. The foot is positioned on a simple insole and multiple layers of flat sheets of ethylene vinyl acetate to prevent the metal brace close to the heel from impinging on measurements. The foot plate is connected to the KinCom motor whose axis is in the centre of the circle beneath the foot plate and rotates in a plane parallel to the circular disc. Thus, the rotation creates pronation and supination of the foot.

4.4.2.2 Measurement of rearfoot kinematics.

The frontal plane movement of the rearfoot was measured simultaneously with the displacement within the tendon to allow for an accurate assessment of the relationship between rearfoot motion and tendon strain. Rearfoot motion data was collected using Qualisys ProReflex cameras (50 Hz Qualisys AB, Gothenburg, Sweden). The cameras were arranged around the dynamometer chair and foot plate (the mechanical structure of the dynamometer prevented collection of data from the left leg). The camera calibration for the global reference system was carried out using a wand and the L-shaped reference frame placed on a pedestal close to the dynamometer foot plate.

To measure the motion of the heel relative to the shank during foot pronation/supination four reflective makers were attached on a rigid plate to the lower aspect of the lateral side of the right shank. Rearfoot motion relative to the shank was collected using a triad of reflective markers on the lateral/posterior part of the heel. In addition, to enable the planes in which heel motion was described to be anatomically relevant, anatomical markers were added for a relaxed standing trial. These additional markers were placed on the medial/lateral femoral condyles, both malleoli and the second metatarsal head (these were removed for all data collection trials). The position of the foot in relaxed standing was used to set 0° in the data.

4.4.2.3 Measurement of Achilles tendon displacements using ultrasound.

The same examiner (the candidate) carried out all ultrasound measurements. Prior to any data capture exploratory free scanning of the Achilles was performed to explore any participant specific tendon features and the general interaction between tendon and probe as the rearfoot was pronated/supinated. All scanning was performed using a handheld 10-18 Megahertz (MHz) 40-mm linear array, B-mode ultrasound probe (Type LA 435, Mylab 70, Esaote Biomedica, Italy). All ultrasound images scans were captured at 25 Hz and these images were later exported as audio video interleave files.

For collection of data the probe was positioned longitudinally on the medial side of the Achilles tendon of the right leg. The distal end of the probe was placed about 2 cm above the calcaneal insertion site. Settings were adjusted on the ultrasound machine to optimise image quality and, allowing for anatomical variations, to find the optimal probe position. The primary goal of modifying the set up the ultrasound device was to achieve the most detailed visualisation (best possible resolution) of the entire tendon and sub structures possible for a specific participant (i.e. proper reflection of the waves from the tendon throughout the tissue). The return of the sound waves to the probe depends on an appropriate depth of the target tissues and the target in this case was the medial and lateral aspects of the Achilles tendon. In the scans, therefore, structures deep to the lateral portion of the tendon were always visible and the tendon/fat or skin boundary on the lateral side also visible (see later Figure 4.2 for an example image). Thus, the target areas were typically at the top/central to the ultrasound image. A related function is the frequency of the ultrasound probe. Higher frequencies result in shorter wave lengths and produce better image resolution, better for detailed identification of tendon substructures, but shallower penetration of tissues. These shorter waves limit the depth of view due to higher attenuation (heat absorption) of waves within tissues. Initially, the highest frequency was chosen for each of the subjects at the correct depth (2.5 to 6 cm, dependent on medial and lateral dimensions of the tendon). Where visual interpretation indicated improvement might be possible the frequency was lowered. Using lower frequencies (i.e. longer wavelengths) was pertinent for wider/thicker tendons enlarging the depth of interest (penetration) of ultrasonic waves, but at the expense of image resolution. Also, the focus of the ultrasound beam was adjusted at the level of the lower boundary of the medial tendon (in the middle of the image plane) to obtain the best resolution in that area for each participant.

In addition, adjusting the brightness (gain) on the ultrasound machine may alter the overall grey scale imaging of the tendon. Too bright or dark will limit tissue identification because tissue substructures becomes less distinguishable from each other. This was important since the focus was on substructure on the medial and lateral parts of the tendon. The gain controls were manually set for each individual. With increasing tendon depth the gain was lowered in the top part of the image and gradually increased in the bottom of the image. Thus, the overall criteria was that the gains were adjusted so the tendon speckles appeared not grainy nor too bright. Further modifications based on subjective judgement used the image contrast feature. For example, if the tendon consisted of fine, small (densely packed) tendon fibers, the image had tendency to appear very bright due to less distinguishable interface between tissue layers, and the contrast was therefore lowered to investigate whether a clearer image could be obtained.

The optimal tendon image produced by varying the scanning parameters above is partly dependent upon the ultrasound operator interpretation of the images during set up. It is also sensitive to the structural anatomy of the Achilles tendon of the subject being tested. For example, a wider and longer tendon might have more identifiable speckles than a short and narrow tendon. Tissue composition will affect image brightness and larger/smaller tendons with lie at different depths from the probe surface.

When a quality image and repeatable position was achieved the position of the distal and proximal ends of the probe were marked on the skin. A small spirit level was attached to the probe to minimise out of plane movements during data collection. Care was taken to apply minimal but constant pressure on the tendon tissue and this was checked visually on screen during practice movements. Multiple practice foot movements and visual inspection of scan data were performed to optimise the operators' position relative to the participant, the operators hand position and posture, and create a stable alignment between ultrasound probe and the foot/leg.

4.4.3 Protocol

Kinematic and ultrasound data capture was carried out at the Gait and Human Performance Laboratory, University of Salford, United Kingdom. Prior to the data collection the subjects' right foot and footplate was moved manually by the investigator (the candidate) to identify the maximum everted and inverted foot plate angles that could be tolerated. Due to the nature of the dynamometer mechanism, the foot plate moved in a triplanar manner. Thus, in producing rearfoot inversion using the dynamometer, the foot actually experienced inversion, adduction and plantarflexion, and vica versa for rearfoot eversion. These maximum positions were identified by feedback from the subjects and the dynamometer software adjusted to prevent movement beyond these maximum positions. The footplate was then set to move at a constant rate of 5 °/s and perform two movement sequences:

(1) from maximum foot eversion through to maximum inversion, and

(2) from maximum foot inversion back to maximum eversion.

Similar angular velocities have been used for monitoring passive movement in the sagittal plane in order to minimise the risk of hysteresis and increased stiffness values (Edama et al., 2014; Kubo, Kanehisa, & Fukunaga, 2002), also others have used similar velocities for passive inversion and eversion movements of the foot (Barbanera, Araujo, Fernandes, & Hernandez, 2012). The tests were repeated in a cyclic manner and therefore not randomised.

A trigger generated a voltage signal from the Qualisys motion capture system to the ultrasound machine and enabled motion data and ultrasound video data to be synchronised. A minimum of ten iterations of the eversion to inversion and inversion to eversion cycles were performed. The participants were instructed to relax during the movement of the foot plate and given short rest periods in between movement cycles. The entire data collection session with each participant took approximately 60 minutes.

4.4.4 **Data analysis**

4.4.4.1 Kinematic analysis

Raw marker coordinates were labelled within Qualisys' software (Qualisys, ProReflex, Gothenburg, Sweden). Visual inspection was carried out to ensure that smooth trajectories were presented within each motion file. These files were later transferred as coordinate 3D files into Visual3D™ (C-Motion Inc.) to compute and derive segmental angle data. Any missing markers were interpolated using a third polynomial order method, allowing a maximum of 10 missing frames to be interpolated. The data was later filtered with a fourth order low-pass bidirectional Butterworth filter, set at 6 Hz.

The foot model adopted for the kinematic analysis consisted of two rigid body segments: (1) shank (tibia and fibula), and (2) rearfoot (calcaneus, incl. subtalar and talo-crucal (ankle) joints). The shank and heel segments were each represented by a local coordinative system. The vertical (*z*) axis of the local coordinate system of the shank was defined using the knee and ankle centres (midpoints between femoral condyles and malleoli respectively). The anterior/posterior shank axis (*y*) was perpendicular to a plane defined by the femoral condyle and malleoli markers. The medial/lateral shank axis (*x*) was perpendicular to the other two shank axes.

The heel local coordinate frame orientation was set such that in relaxed standing the *z*-axis was perpendicular to the floor (XY plane of global reference system) and the y (anterior/posterior) axis was perpendicular to *z*-axis but parallel to the markers on heel and second metatarsal in the static standing trial. The *x*-axis was perpendicular to the other two axes. Joint rotations (heel relative to shank) were computed using Cardan angle rotation sequence (x-y-z), in which rotations around the *y*-axis of the shank represented frontal plane movements. The number of degrees of freedom was six for the model.

4.4.4.2 *Ultrasound image analysis*

Ultrasound images were processed to derive displacement data for tendon structures on the medial and lateral sides of the Achilles tendon during the eversion to inversion and inversion to eversion foot movement cycles. The image files were coded to blind the person analysing the audio video interleave files. Ultrasound images were processed using in house software developed in a separate PhD project (School of Computing, Science and Engineering, Salford University, by PhD candidate Ahmad S.A. Mohamed). The analysis using this software involves changing programming code to set up, run and quality check the analysis of ultrasonic images. This requires knowledge of the coding and the mathematical algorithms used to track the speckle details in each ultrasound image, and these were in development by the PhD candidate (Azlan S.A. Mohamed) when this MPhil project was performed. A. Mohamed therefore had to conduct the image analysis. The method for tracking the intra tendon structures and associated algorithms have subsequently been published (Pearson, Ritchings, & Mohamed, 2013, 2014).

The candidate visually inspected the videos of the ultrasound images in order to identify any trials where probe movement, a loss of contact between probe and skin, or excess load on the probe had produced gross errors in image quality and data integrity. The target was to identify a minimum of five data sets for each movement (eversion to inversion, and separately inversion to eversion) and this was achieved for all participants.

The ultrasound video files were loaded into the custom matrix laboratory programme written by PhD candidate A. Mohamed. The video files were split into individual image frames to enable the bespoke algorithm (Pearson et al., 2013, 2014) to calculate the displacement of four intra tendon structural features relative to each other, and thus the stretch/shorten behaviour on the medial and lateral sides of the tendon. These four structural features, or regions of interest (ROI), were two areas of at least (15•15) pixels within the medial tendon, and two similar areas within the lateral tendon. The distance between the two medial ROI (and two lateral ROI) was calculated for each image frame and thus displacement (lengthen/shorten) between the two ROI could be calculated over successive image frames. Further technical details are provided below.

The image analysis followed a four-step process that was developed collaboratively by the candidate and A. Mohamed, who then modified and implemented the mathematical algorithm and software code, as follows:

1) The image of the tendon was split into two equal layers, with the top layer representing the medial side of the tendon and the bottom layer the lateral side. The analyser (A. Mohamed) was blind to which portion of the tendon was top/bottom in the image and which movement direction was being viewed.
2) In the first frame from the ultrasound video, four (15•15) pixels ROI were manually identified using the speckle patterns in the image, two in the top layer and two in the bottom layer of the image. One ROI (R2) was identified on the left of the image (i.e. proximal tendon side) while another ROI (R1) was identified on the right side of the image (i.e. distal tendon). The line connecting the two ROI represented a motion vector $\overrightarrow{R2R1} = \begin{pmatrix} R2_x - R1_x \\ R2_y - R1_y \end{pmatrix}$ for each tendon layer (Figure 4.2) and the initial distance between the two tendon structures.

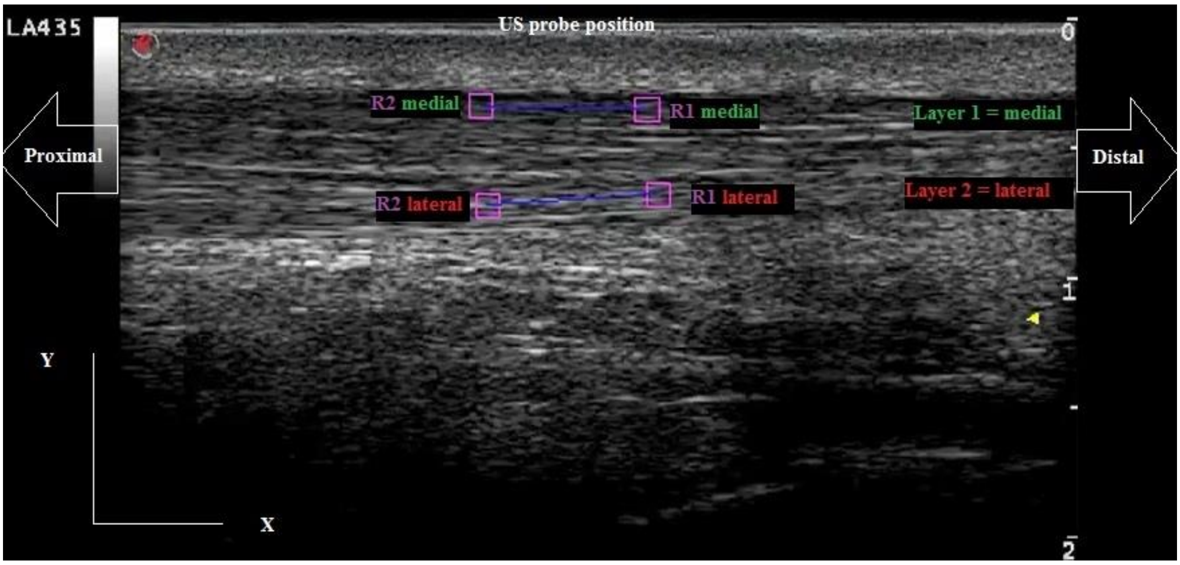


Figure 4.2. Definition of layers and placement of ROI within the tendon.

3) The second frame from the ultrasound video was then loaded. An automatic block-matching algorithm using a normalized cross correlation function was used to explore a predefined search area in the second frame. The search areas were the areas around the four ROI identified in the first frame. The normalized cross correlation algorithm searched to identify the best speckle pattern match in the second frame for each of the four ROI in the first frame. The search window was set at a fixed 2•ROI size in X-direction (width) and 1•ROI size in the Y-direction (height) of the image plane (see Figure 4.2 left corner). This identified the speckle pattern close to each ROI that was the best match to the ROI speckle pattern in the first frame. The distance between the two new ROI (i.e. in the second frame) was then derived for layer 1 and 2 (i.e. medial and lateral sides of the tendon). A more detailed description of the mathematical basis for the ROI tracking is presented (S. J. Pearson et al., 2013, 2014).
4) This process was then repeated for frames number 3, 4, 5 and so on, to provide a distance measure between R1 and R2 (medial layer see Figure 4.2 top view) and between R1 and R2 (lateral layer see Figure 4.2 bottom view) during the eversion to inversion, and inversion to eversion movement of the foot.

Absolute change in distance length (ΔL) between R1-R2 (medial layer) and R1-R2 (lateral layer) was expressed in millimetres (mm) from the initial frame to the final frame of the ultrasound video images and computed using the method taken from Arndt et al. (2012) and adopted by Pearson et al. (2014). The overall tendon excursion, thus the displacement, was the length of the movement vector $\| \overrightarrow{R2R1} \|$ in the first frame (F_1, start video) subtracted from the final length of the movement vector $\| \overrightarrow{R2R1} \|$ in last frame (F_n, end video) (see Equation 4.1).

In order to compute strain, the displacement (as defined in Equation 4.1), must be divided by the initial distance (L_0), between R1 and R2 (i.e. length of the movement vector $\| \overrightarrow{R2R1} \|$ in the first frame F_1).

Equation 4.1. Tissue displacement formula:

$$\text{Displacement layer 1 (medial)} = \sqrt{((x_{R2med} - x_{R1med})^2 + (y_{R2med} - y_{R1med})^2)}_{\text{Initial frame (start video)F1}} - \sqrt{((x_{R2med} - x_{R1med})^2 + (y_{R2med} - y_{R1med})^2)}_{\text{Final frame (end video)Fn}}$$

$$\text{Displacement layer 2 (lateral)} = \sqrt{((x_{R2lat} - x_{R1lat})^2 + (y_{R2lat} - y_{R1lat})^2)}_{\text{Initial frame (start video)F1}} - \sqrt{((x_{R2lat} - x_{R1lat})^2 + (y_{R2lat} - y_{R1lat})^2)}_{\text{Final frame (end video)Fn}}$$

The normalized cross correlation algorithm was not always able to follow the ROI's throughout the entire movement of the foot. This was especially an issue at the extremes of movement because changes in tendon shape and maintaining consistent probe contact was more challenging. A minimum number of image frames and therefore range of foot movement was identified to ensure the data would have functional relevance compared to the ranges of motion used during walking and running. In all cases a minimum of 9.0° of frontal plane rearfoot movement was targeted in the everted to inverted and inverted to everted movements and the corresponding tendon displacement data used. This is approximately 80-90 % of the total frontal plane excursion used during walking and running in Chapter 3, and was typically exceeded by most participants (see Results 3.5. section). However, where the 9.0° (or more) motion occurred within the total range of movement that each foot experienced varied between the everted to inverted, and inverted to everted movement sequences. Typically, the data came from early in each movement sequence, since this was when probe placement had been established and was most secure. As foot movement occurred, it was increasingly difficult to maintain probe contact.

Subjects for whom it was impossible to track the equivalent of 9.0° of rearfoot motion were excluded. Four subjects were lost due to this issue and a further one data set was unusable due to errors in the synchronisation of data. Thus, the data presented here is from 17 subjects.

4.4.4.3 Compound analysis

All final computations were carried out in Microsoft Excel. Medial and lateral tendon displacement values from each trial were aligned with the corresponding frontal rearfoot motion data and then each data set were then averaged to produce single tendon displacement and single rearfoot motion data set for each participant. The seventeen tendon displacement data sets and motion data sets were then averaged for each of the two movement sequences separately (i.e. all everted to inverted data was averaged, and then all inverted to everted data was separately averaged). This provided group data for the everted to inverted sequence, and separately for the inverted to everted movement sequence. Data is presented as mean (*M*) and standard deviation (±*SD*).

Graphical methods were used to illustrate tendon displacement (mm) in relation to angular rearfoot data (°). Excel was used to compute Pearson correlations coefficients (R^2). The interpretations are made according to the recommendations by (Taylor, 1990) and are given in Table 4.1.

Table 4.1. Pearson Correlation (R^2) Values Interpretations

R^2 value	Interpretation
<0.35	Low or weak correlations
0.36 to 0.67	Modest or moderate correlations
0.68 to 1.0	Strong to high correlations

4.5 Results

4.5.1 Kinematic data

The mean total range of frontal plane rearfoot motion during tests was 24.3(±7.3)° for the everted to inverted movement, and 22.6(±6.7)° for the inverted to everted movement (Table 4.2). The amount of this motion used with the corresponding tendon displacement data was 12.8(±1.9)° for the everted to inverted movement, and 11.5(±1.1)° for the inverted to everted movement. Thus, approximately 50 % of the total frontal plane rearfoot movement was used in the correlation analysis between frontal plane motion and tendon displacement.

For the group averages, 7.1(±4.0)° of frontal plane motion was used for the everted to inverted movement, and 6.7(±2.3)° of frontal plane motion was used for inverted to everted movement sequence. These are smaller than the ranges for the individual participants because they can only be calculated when the position of the ROM overlaps for all 17 participants (see Figure 4.3 in section 4.5.3).

4.5.2 Tendon Displacement

When the rearfoot was moved from everted to inverted, there was an average of 1.1(±0.5) mm elongation on the lateral side of the tendon, and 1.1(±0.8) mm shortening on the medial side of the tendon. When the rearfoot was moved from an inverted to everted position, there was an average of 0.9(±0.8) mm shortening on the lateral side of the tendon, and 1.8(±0.9) mm elongation on the medial side of the tendon (Table 4.2).

Table 4.2. Frontal ROM during test conditions and displacement ROM from two extreme positions for the right leg.

	EVERTED TO INVERTED				INVERTED TO EVERTED			
Subject number	TOTAL ROM (°)	ROM used in Ultrasound data (°)	Lateral Displacement (mm)	Medial Displacement (mm)	TOTAL ROM (°)	ROM used in Ultrasound data (°)	Lateral Displacement (mm)	Medial Displacement (mm)
1	34.8	13.8	1.2	-1.1	32.0	9.9	0.0	2.8
2	31.9	13.1	0.6	-0.2	25.9	12.7	0.0	2.0
3	23.7	13.4	1.3	-1.4	21.3	12.0	-1.0	0.6
4	21.2	16.2	2.5	-1.8	18.8	11.0	-1.1	0.4
5	22.7	12.6	1.4	-0.8	17.8	12.2	-1.2	2.6
6	24.4	11.7	0.8	-1.5	22.8	11.7	-2.1	1.8
7	34.5	15.8	1.1	-0.1	32.2	12.9	0.7	0.9
8	33.5	16.4	0.0	-0.3	30.4	11.3	-1.6	2.4
9	18.3	11.6	1.3	-2.1	17.7	10.6	-1.3	2.3
10	16.5	9.2	0.8	-1.3	15.6	10.3	-2.0	1.3
11	18.5	12.0	1.4	-0.7	21.0	9.0	-0.4	2.9
12	16.9	12.2	0.8	-2.1	16.6	12.0	-0.6	1.6
13	20.6	11.5	1.3	-1.1	19.4	13.0	-2.0	3.5
14	16.4	11.6	1.7	-0.6	14.2	11.2	-0.8	1.5
15	33.2	12.2	1.0	-0.2	32.3	12.1	-0.7	1.3
16	15.5	11.5	1.4	-2.8	14.8	12.1	-1.7	1.6
17	31.5	12.2	0.8	-1.2	31.2	12.1	-0.9	1.7
***M*(±*SD*)**	**24.3(±7.3)**	**12.8(±1.9)**	**1.1(±0.5)**	**-1.1(±0.8)**	**22.6(±6.7)**	**11.5(±1.1)**	**-0.9(±0.8)**	**1.88(±0.9)**

Note. ROM= rearfoot range of movement in relation to shank within the frontal plane. Negative displacement (-) indicates tissue compression whereas positive displacement (+) suggests tissue elongation.

4.5.3 Relationship between tendon displacement and rearfoot motion

According to the mean data for the sample (see Figure 4.3), when the foot was moved from an everted to an inverted position (mean, 7.1°) the medial portion of the Achilles tendon underwent negative displacement, i.e. got shorter (by 0.5(±0.3) mm), whereas the lateral tendon experienced positive displacement, i.e. got longer (stretched, by 0.4(±0.6) mm). The opposite tissue behavior occurred when the foot was moved from inverted to everted (mean, 6.7°). In this case the lateral portion of the tendon became shorter (i.e. negative displacement, by 0.6(±0.7) mm) and the medial layer was lengthened (i.e. positive displacement, by 1.1(±0.8) mm).

The relationship between mean tendon displacement and mean rearfoot motion data for the sample was close to linear, R^2 being >0.97. The R^2 was slightly lower for everted to inverted movement (0.97) compared to inverted to everted movement (0.99). These relationships are illustrated in Figure 4.3.

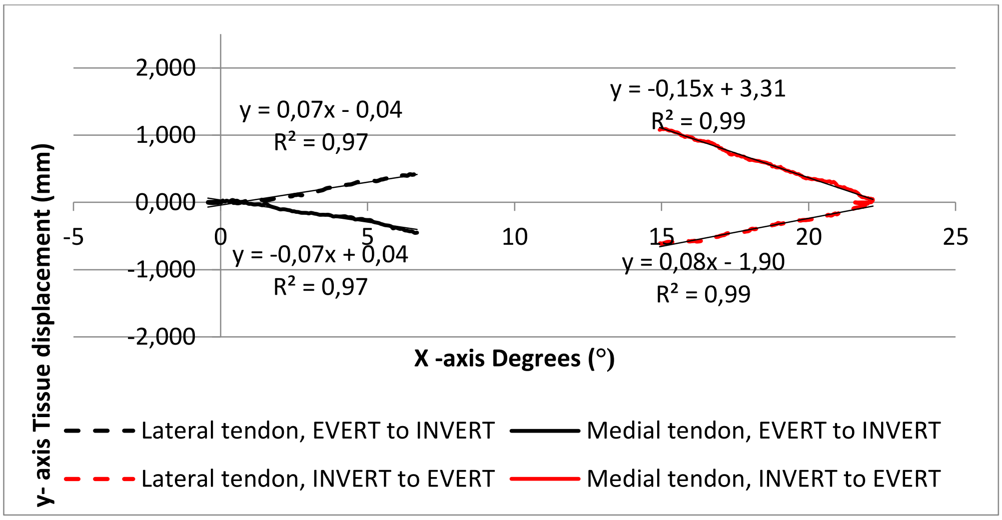
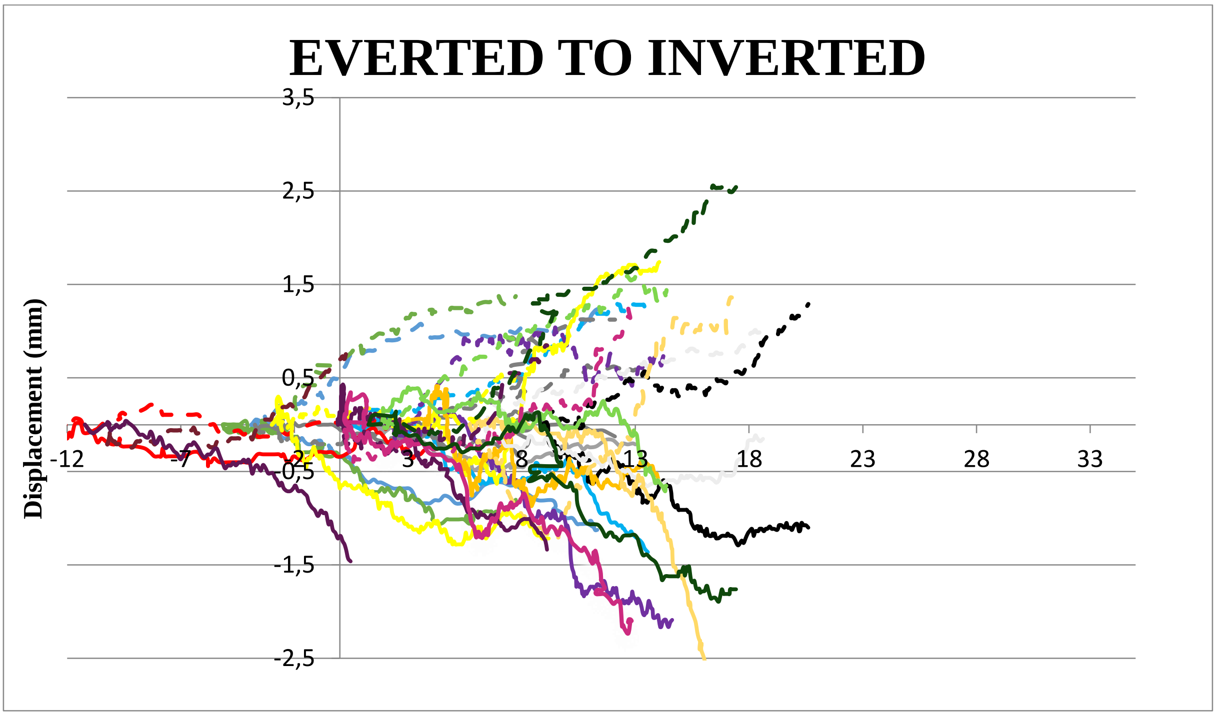


Figure 4.3. Group mean (*M*) ensemble angular and tissue displacements values for both test conditions. Y-axis displays tissue displacement in (mm). Negative displacement (-) indicates tissue shorting whereas positive displacement (+) shows tissue stretch. X-axis indicates frontal plane rearfoot position in degrees (°), where negative (-) values are eversion and positive (+) angles are inversion, and 0° was the position of the foot in relaxed standing. Pearson correlation coefficient (R^2).

Pearsons correlation (R^2) values for the 17 individual participants are provided in Table 4.3. Mean correlations were moderate to strong (0.64–0.81 R^2) for the entire sample. The seventeen individual motion versus displacement data is illustrated in Figure 4.4.

Table 4.3. Correlations between angle excursions and displacement during both test sessions.

	EVERTED TO INVERTED		**INVERTED TO EVERTED**	
	Pearson R^2		Pearson R^2	
Subject	Lateral	Medial	Lateral	Medial
1	0.83	0.90	0.11	0.90
2	0.95	0.16	0.12	0.97
3	0.95	0.81	0.90	0.63
4	0.83	0.84	0.74	0.69
5	0.93	0.93	0.97	0.97
6	0.67	0.89	0.90	0.59
7	0.78	0.04	0.77	0.07
8	0.11	0.07	0.87	0.93
9	0.59	0.91	0.80	0.97
10	0.34	0.95	0.91	0.74
11	0.95	0.49	0.20	0.93
12	0.24	0.93	0.21	0.80
13	0.83	0.87	0.90	0.97
14	0.45	0.75	0.34	0.90
15	0.97	0.39	0.97	0.94
16	0.61	0.78	0.93	0.83
17	0.43	0.79	0.21	0.89
***M*(±*SD*)**	**0.67(±0.27)**	**0.68(±0.32)**	**0.64(±0.34)**	**0.81(±0.23)**



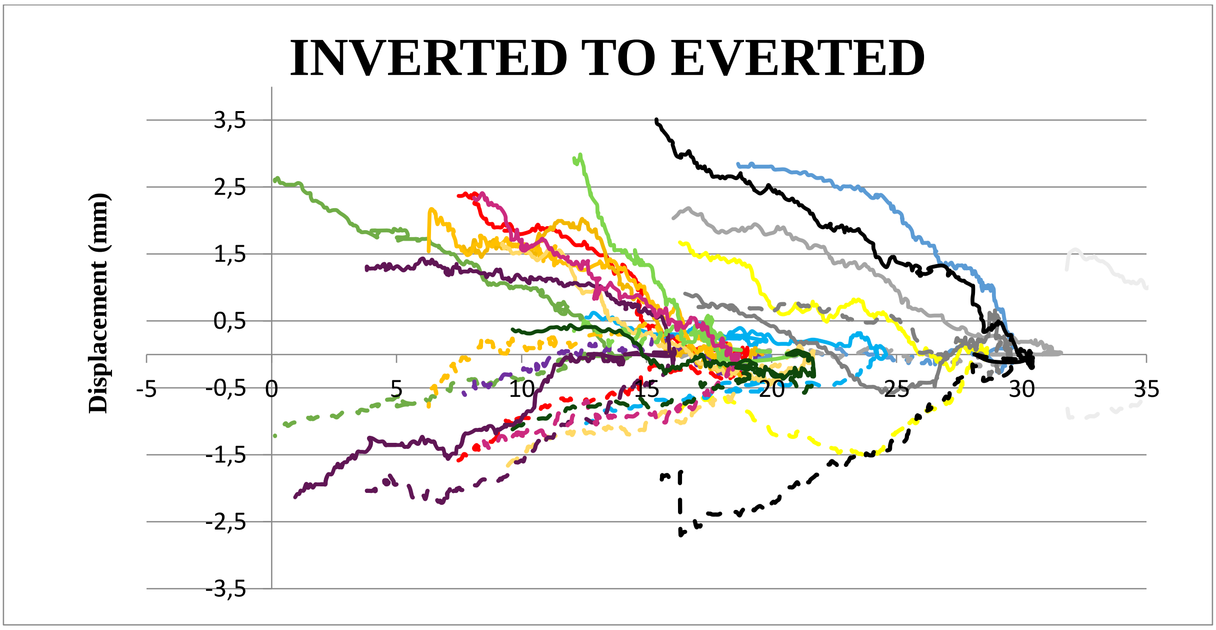


Figure 4.4. Individual averages (*M*) for the medial and lateral Achilles layerfor each of the Seventeen Subjects. X-axis indicates frontal plane rearfoot position in degrees (°), where negative (-) values are eversion and positive (+) angles are inversion, and 0° represents relaxed standing position. Negative displacement (-) indicates tissue compression. Positive displacement (+) indicates tissue elongation. Y-axis displays tissue displacement in (mm). Broken lines represent lateral Achilles tendon. Solid lines show the medial tendon

Twelve of the 17 subjects showed modest to strong correlations between tissue displacement and angle position in both medial and lateral tendon parts during movement from everted to inverted rearfoot position. In these the correlations for the lateral side ranged from 0.43 to 0.97 R^2 and from 0.39 to 0.97 R^2 for the medial side. Eight of these 12 subjects showed medium to strong correlations during movement from an inverted to everted rearfoot position, with values of 0.74 to 0.97 R^2 for the lateral side, and 0.59 to 0.97 R^2 for the medial. Only two subjects showed very weak correlations across both test sessions. There were similar numbers of weak/moderate/strong correlations for medial and lateral sides of the tendons, and in everted to inverted and inverted to everted movement sequences (see Table 4.3).

4.6 Discussion

This study aimed to investigate if differences exist between displacement in the medial and lateral parts of the Achilles tendon during passive eversion and inversion excursions of the foot. Differences in medial and lateral tendon displacement were observed for both eversion to inversion and inversion to eversion foot movements. The relationship between rearfoot motion and lengthening and shortening in the tendon was strong based on the mean data for the sample, and the majority of the individual participant data. The difference between the results at sample and individual participant level is explained mainly due to the use of different amounts of rearfoot frontal motion data in the two calculations, with reduced motion data used in the calculation of the sample data (e.g. 7.1° compared to 12.8° for the individual participants, see Figure 4.3 and Table 4.2).

This is the first time that ultrasound has been used to explore the frontal plane behaviour of Achilles tendon sub structures in vivo. The observation that the medial part of the Achilles tendon experiences greater strain as the rearfoot is everted, and less strain during inversion, leads us to accept the null hypothesis being investigated. This result also concurs with studies that also investigated this hypothesis. Lersch et al. (2012) studied Achilles strain in vitro rather than in vivo, but also observed that eversion was associated with larger strain rates in the medial tendon. Conversely, greater rearfoot inversion was associated with larger strain rates within the lateral tendon. Direct comparisons with this study are difficult due to differences in the methods used. This current study used B-Mode Ultrasonography probe to observe in vivo how tendon structures displace during foot motion. By contrast, Lersch et al. (2012) used surface mounted pin markers attached to Achilles tendon fascicles, thus observing changes in the Achilles surface rather than within tendon structures. They also investigated only a small number of cadavers (n=4). Also, the results presented here were carried out during passive frontal plane rearfoot movements and without any purposeful activity in the triceps surae muscles, and thus loading of the Achilles. Also, the authors were able to load the tendon structures whilst they concurrently observed the effects of inversion and eversion movements. Finally, the range of frontal rearfoot motion over which both motion and ultrasound data was available was less than the ROM observed by Lersch et al. (~30°, compared to ~12° achieved in this study).

The actual tendon displacements reported here are much lower than the peak values observed in the sagittal plane during fast walking (range 5.9-7.6 mm) in Franz et al. (2015) and during passive eccentric loading (>4.0 mm) in Slane and Thelen (2014). The displacements observed (0.9-1.8 mm) are however in line with those observed during slower walking (0.4-1.2 mm) (Franz et al., 2015). Lower displacement values may also be explained by the smaller range of rearfoot motion over which tendon displacement was measured. For example, Slane and Thelen (2014) measured tendon displacement over 30° of ankle motion, and Franz et al. (2015) observed tendon displacement over 13.6-18.3° of motion. Both are larger than the frontal plane motions in this current study (11.5°-12.8°).

The results of this study, when interpreted in the context of the within tendon sub structures identified in the literature review, may suggest some degree of sliding occurs between tendon sub structures during rearfoot eversion and inversion (Haraldsson et al., 2008). Indeed, non-uniform tendon displacement within different calf and Achilles muscle tendon compartments has also been recently reported during walking (Franz et al., 2015), passive and eccentric loading conditions (Slane & Thelen, 2014) and passive conditions (Arndt et al., 2012). However, all of these reports have focused on sagittal plane longitudinal displacement whereas this study focused on frontal plane displacements.

Weak correlations between movement and displacement occurred twice as often on the lateral as the medial sides of the tendon. There were eight weak correlations (i.e. R^2 <0.38) on the lateral side compared to 3 on the medial side (see Table 4.3 (earlier)). This totals 11 weak correlations out of 34, suggestive that typically a correlation does exist between rearfoot motion and tendon displacements. Weak correlations may occur if the rearfoot motion and position leads to either no more tendon displacement (i.e. it is maximally elongated, very stiff, and liable to fracture) or because the tendon is under no strain at all. Weak correlations may also occur when either the tendon displacement or motion data is subject to noise or error. This would occur if the ROM used to calculate the tendon displacement was very small and thus tendon displacements also very small. Indeed, in all 11 instances of a weak correlation the tendon displacements were below the mean for the sample, ranging from 0.0 to 0.9 mm, and in 5 of the 11 cases tendon displacement was equal to or less than 0.4 mm. Thus, smaller displacements, which are likely more susceptible to noise in the data, seem to be associated with lower correlation values.

The absolute displacements were sub mm in some cases and thus even an acceptable error in the location in the ROI could lead to a significant % change in displacement from one ultrasound image to the next. This could lead to real tendon displacements being missed and thus lower correlations. The opposite (i.e. error in ROI location creating strong correlations that do not in fact exist) seems less likely because the errors would have to be highly systematic in order for them to artificially generate a moderate to strong correlation over the range of rearfoot motion observed.

4.6.1 **Potential clinical implications**

These results support the hypothesis (H_0) that a greater range of eversion movement might induce greater strain within the medial Achilles tendon and therefore that eversion could be a risk factor for Achilles injury. However, it is not clear whether more eversion would lead to the strain being outside the physiological limits for a specific person. It may be that as long as the tendon structures have adapted to the strains experienced that no injury occurs. However, there would be some physiological limit to this and some individuals might use their Achilles closer to the physiological limit than others. Then, in some circumstances, such as increase in physical activity, the risk of the Achilles experiencing strain outside the physiological limit could be greater. It might equally be that if there are circumstances where eversion increases beyond the normal ranges for a person, perhaps due to a different choice of footwear, or fatigue, then this might temporarily increase tendon strain. There is evidence that running related fatigue affects foot movement and posture and could increase rearfoot eversion (Cowley & Marsden, 2013).

The displacement in the medial tendon was larger when the foot was moved from an inverted to an everted position. Despite slightly lower ROM experienced by the foot from inverted to everted position, the medial tendon displacement was 1.8 mm compared to 1.1 mm for the opposite movement. This may be explained by the fact that the maximum inversion positions and the inversion positions achieved when moving from an everted to inverted position would not always coincide. However, the movement from inverted to everted position reflects the initial contact phase movement pattern in walking and running, a period hypothesised to relate to excessive strain in the Achilles and risk of injury (Chapter 2). Greater displacements in the inverted to everted sequence could therefore be functionally relevant if it is indeed movement and tendon strain in that early contact phase that is associated with injury. It is worth noting however that because the feet of some participants did not pass through 0° the position of the foot when tendon displacement data was recorded was likely more inverted than might be used during gait. In fact it could be questioned whether the rearfoot position in some participants represented a functional position if it did not pass through 0°. However, since the relationship between rearfoot position and tendon displacement was close to linear and very similar in both movement directions, it is unlikely that the relationship would simply not exist in a different rearfoot position.

In terms of orthotic effect, the data here allow the effects of the change in eversion reported in Chapter 3 to be put into a tendon displacement and perhaps strain context. For example, the mean reduction in rearfoot peak eversion due to the use of a foot orthotic was ~3.9°. Prior to this current study it was not possible to extrapolate what this reduction might mean for Achilles tendon displacement. However, using the figures for the sample (Table 4.2), if 1.8 mm displacement was observed for the medial tendon portion for 11.5° of motion (mean for sample, during movement from inverted to everted position) then this equates to ~0.2 mm/° displacement per degree of eversion movement. The corresponding figure for everted to inverted movement is ~0.1 mm/° displacement per degree of inversion movement. Thus, based on the data reported here it can be estimated that a 3.9° reduction in eversion would mean a potential reduction of 0.3-0.6 mm in tendon displacement. This will relate to a reduction in tendon strain, although the precise value cannot be calculated because the initial resting length of the tendon at the start and end of the rearfoot movements was not measured. However, 0.3-0.6 mm represents 50 % of the mean total displacement observed and it might be assumed to be of significance. Assuming strain in the Achilles in walking represents 100 % strain, and adopting displacement values reported by Franz et al. (2015) (i.e. 0.4-1.2 mm during slow walking), then a 0.3-0.6 mm reduction in strain due to the 3.9° reduction in eversion could relate to a 50-100 % reduction in total medial tendon displacement experienced in walking. Assuming tissue stress and tendon displacement are closely correlated then these general estimations suggest that foot orthoses would reduce medial Achilles tendon strain a considerable degree. However, great caution is required with these values and extrapolations. Strain in the tendon occurs in all three directions and displacement could not be reduced by 100% (i.e. resulting in no displacement at all). Also, the displacement values reported here were measured in Achilles tendons that were not under any load and thus real displacements due to frontal plane rearfoot movements will be different than those reported here. Regardless, the overall impression is that a 3.9° change in rearfoot position due to an orthosis could lead to a non-negligible change in medial Achilles displacement, and thereafter it is assumed strain and stress.

4.6.2 Limitations relevant to the interpretation of the results

The operator dependent reliability of ultrasound imaging is well recognised and the ability to track tendon sub structures during rearfoot movement will be affected by how the probe is held and positioned (Cronin & Lichtwark, 2013; Seynnes et al., 2015). Slipping of the probe due to low friction gel interface, out of plane movement of the probe or tendon underneath the stationary probe, and variation in the tissues beneath the probe might all affect the ability to track tendon features during movement sequences used. Several features of the protocol were designed to reduce risk of error. The final protocol was the result of several months of testing, including some experimental designs discarded because they were too difficult. The same operator became expert in the approach and did all the scanning. Many pre data collection practices and scans were taken in all cases. Some feet and participants were rejected at recruitment stage because initial tests on their feet identified unusual anatomy around the Achilles preventing a good quality fit between probe and tendon. This was in part the reason for studying young males who tended to have less fatty tissue surrounding the tendon, and larger tendons against which it was easier to locate the 40 mm probe. The candidate also screened all the ultrasound video to identify evidence for probe related problems with image quality, discarding data from four participants. Despite these efforts, some out of plane and in plane probe movements would be inevitable. This would have affected the ability of the ROI tracking algorithm to identify and track suitable areas on the ultrasound images. The effect of the size of recorded displacement on the correlation values (risk of weak correlations) has already been discussed.

It is also the case that 2D ultrasound imaging is vulnerable to errors due to movements within planes other than that plane being imaged. For example, the 2D image cannot detect the movement of a speckle on an image (i.e. a feature of the tendon structure) if it moves out of the plane of view, but might mistake a new speckle entering the viewing plane as that previously being tracked. Obst and colleagues (2014) used 3D ultrasound in vivo to suggest that the midportion of the tendon undergoes large dimensional changes in the medial and lateral but also anterior and posterior directions during isometric plantar flexion contractions. For this reason speckles on an image may overlap and the initial tendon speckle in one frame may be substituted for another in the following frames.

The current study has only examined tendon behavior within healthy individuals. Others have suggested that a symptomatic tendon may behave differently because of pathological changes, and it might be easier to identify and thereafter track sub structures using these pathological changes (Arya & Kulig, 2010; Child et al., 2010). It is also of note that the data presented relates only to males. As stated above, this was a pragmatic choice to reduce the risk of error in the image data, but may have limited the external validity of the data (i.e. how it relates to the wider population).

The range of rearfoot motion used to investigate the relationship between tendon displacement and rearfoot movement was 11.5-12.8°. This is much smaller than the total ROM available at the rearfoot and the total ROM through which the feet investigated moved on the dynamometer. Furthermore, whilst it is close to the typical ranges of rearfoot eversion/inversion reported in the literature. For example, Nester et al. (2014), reported rearfoot excursion of 12.9º in n=100), where the 11.5-12.8° of motion used here occurs within the total available ROM was variable between participants, and unknown. Thus, the foot position during the 11.5-12.8° of motion may not correspond to the same rearfoot position used in walking. However, in establishing a comfortable starting position for the participants in the dynamometer chair, the foot was not placed in an extreme or atypical position. Indeed, a comfortable resting position was targeted for the benefits of participants. It is likely that this broadly corresponds to rearfoot positions used in standing and walking, since these probably affect the concept of a “comfortable” or natural resting position for the foot for each participant. It was not possible to recreate the relaxed standing position of the rearfoot in the dynamometer set up. However, though it seems likely, whether any part of the 11.5-12.8° of motion used in this study that corresponds to the motion used during walking/running is not known. As stated earlier, because the relationship between rearfoot position and tendon displacement was close to linear and very similar in both movement directions, it is unlikely that the relationship would simply not exist in the rearfoot position actually used during gait.

Finally, EMG activity of the gastrocnemius complex or tibialis anterior was not monitored and measures of passive torque and muscle force were not obtained using the dynamometer. These parameters would have provided assurances that muscle activity was not interfering with the kinematic measures nor especially the strain measures in the Achilles tendon. Participants were asked to and visually appeared to relax during the experiments. Also, the speed of motion 5 °/s was easily tolerable and unlikely to create a stretch reflex from muscle tissues.

4.7 Conclusion

This chapter was concerned with understanding whether regional strain in the Achilles tendon is affected by frontal plane rearfoot motion and position. The results show that displacement in the medial Achilles tendon increases as the rearfoot is everted, and decreases as the rearfoot is inverted. The opposite occurs on the lateral side of the tendon. Thus, rearfoot eversion and inversion is strongly coupled with increases and decreases in displacement in the medial and lateral sides of the Achilles tendon, and it is assumed this relates to strain and stress in each part of the tendon too. This suggests that rearfoot eversion position and movement could be a factor associated with increased risk of medial Achilles strain, and thereafter stress and injury risk. Also, combining these results with those in Chapter 3, foot orthoses that reduce rearfoot eversion would likely reduce medial Achilles tendon displacement and therefore perhaps strain.

5 General discussion

The aim of this chapter is to discuss the findings of the two experimental studies in the context of other literature, to discuss limitations of the work, and potential future research that would add to our understanding.

5.1 Intra Achilles tendon displacement and stress

The results within this thesis provide further support for the hypothesis that eversion movement induces greater strain within the mid portion of the medial Achilles tendon and that this likely leads to increase stress in that region. This observation supports the idea that fascicles may act as distinct extensible units. The regional stretch and shortening observed on the ultrasound images may be evidence of the differential sliding between fascicles previously suggested by Haraldsson et al. (2008). This also points to the fact that different parts of the tendon sub-structures may be loaded differently during gait and that loading distribution through the tendon due to the triceps surae action may be more complex than previously assumed and in line with early suggestions from Arndt et al. (1999b).

Recent attempts to define internal tendon dynamics for the Achilles has been progressed by Franz et al. (2015). Ten healthy subjects walked barefoot on a treadmill instrumented with force plates. Walking was performed at three different speeds (0.8, 1.0, and 1.3 m•s^{-1}). A 2D ultrasound probe (length, 3.8 mm) incorporated into a leg orthotic was placed over an area 6 cm above the calcaneus. Achilles tendon displacement was measured using an elastographic speckle tracking algorithm obtained from two measurement sites either (1) distal lateral gastrocnemius muscle-tendon junction or (2) the Achilles tendon. The tendon was divided by the investigators into medial gastrocnemius and deeper soleus portions in accordance to the findings by Szaro et al. (2009). Sagittal ankle angle and probe position were also measured. The results showed that there was a strong dependency between walking speed and tendon displacement (0.9(±0.3) mm at 0.8 m•s^{-1}, 1.2(±0.3) mm at 1.0 m•s^{-1} and 1.7(±0) mm at 1.3 m•s^{-1}). However, the superficial layer was consistently more elongated than the deep layer, by 26-33 % across the different walking speeds. The greatest elongation (7.6(±2.6) mm) was in superficial layers, compared to 5.9(±2.6) mm in the deep layer, and occurred during the fastest walking velocity 1.3 m•s^{-1}. The EMG activity of the gastrocnemius also increased proportional to walking speed and tendon displacement. This paper provides novel information on tissue behaviour within the Achilles tendon during gait and supports the assumption that non-uniformity exists between tendon fascicles. It would have been of greater interest if further details on the deformations and EMG magnitudes varied in different sub phases of stance, especially when peak dorsiflexion occurs, due to its' relationship with internal plantar flexor torque.

That displacement differs within different regions of the Achilles tendon during eccentric and passive tendon loading has also been suggested by Slane and Thelen (2014). They also used 2D ultrasound elastography to measure displacement and elongation of the tendon, with an ultrasound transducer positioned slightly over the superior part of the calcaneus, on nine healthy individuals. In contrast to Franz et al. (2015), the authors divided the tendon into three sagittal plane portions: superficial, middle and anterior. The trials consisted of eccentric loading and passive angle excursions during cyclic dorsiflexion and plantar flexion movements, with the knee at 30 and 90° of flexion. The results revealed that the superficial, middle and anterior tendon layers show different displacements. In common with Franz et al. (2015), the least displacement was observed in the deep anterior layer during ankle dorsiflexion, with greatest displacement in the superficial layer. There was significantly greater displacements in the middle and deep tendon layers when the knee was less flexed (30° of flexion), highlighting the role of gastrocnemius in Achilles loading and displacement. Furthermore, passive movement induced similar (in fact greater) asymmetrical longitudinal displacements. That passive movement results in differential displacements within tendon layers has been demonstrated by Arndt et al. (2012). These results support the use of passive foot movements to study tendon behaviour, as was the case in this current thesis, suggesting the results presented in this thesis should have some validity in terms of differential displacements during muscle induced loading of the tendon. From an anatomical and mechanical perspective ankle dorsiflexion places the soleus and gastrocnemius muscles under greater stretch as they crosses the ankle. However, for gastrocnemius, this could be combined with even greater tension if the dorsiflexion coincided with extension of the knee, as it spans over both the knee and ankle. For example, Szaro et al. (2009) looked at which ankle and knee position increased the stretch experienced by the medial and lateral head of the gastrocnemius muscle. Ankle dorsiflexion and full knee extension created the greatest tendon stretch. Since gastrocnemius and soleus have different integration with tendon formation proximally (chapter 2), differential tension in these two muscles should relate strongly to the forces experienced by different sub structures within the Achilles tendon. Accordingly, ankle and knee position is likely a further factor affecting the distribution of stress in the Achilles.

It has also been established that greater longitudinal displacement occurs within the free mid portion of the tendon compared to more proximal tendon structures, meaning that the free and mid portion of the Achilles tendon is more compliant (Bojsen-Møller et al., 2004; Farris, Trewartha, McGuigan, & Lichtwark, 2012; Finni, Hodgson, Lai, Edgerton, & Sinha, 2003b a; Magnusson et al., 2003)**.** For example, significantly less strain (ε, 1.4(±0.4) %) was reported within the proximal tendon structures (i.e. close to the muscle) than within the distal free tendon (ε, 8.0(± 1.2) %) (Magnusson et al., 2003). Greatest strain rates and tendon displacement are also reported to occur within the free tendon during active contractions (Finni et al., 2003b). These outcomes have been confirmed with 3D ultrasound image reconstruction techniques in Farris, Trewartha et al. (2012).

In summary, these studies highlight that several mechanisms co-exist and affect the tensile forces experienced by the Achilles tendon and its sub structures, and thus lead to differential displacements and stress with the tendon. Speed and direction of motion, ROM, and joint position all result in changes in passive (e.g. fascia, tendon) and active (muscle) forces applied to the Achilles tendon and thereafter the displacement of sub structures within the tendon and internal stress experienced by different regions within the tendon.

5.2 Anatomical factors in distal tendon formation and tendon stress distribution

The second study of this thesis investigated displacement within the medial and lateral parts of the Achilles tendon during passive rearfoot inversion and eversion excursions. It was not the primary goal of this thesis to differentiate the relative contribution of the soleus or the gastrocnemius muscle-tendon portions to the distal tendon fascicle formation. However, based on anatomical studies (chapter 2) the probability is that the medial side of the tendon may be the soleal tendon portion. However, twisting of the tendon along its length and variation between individuals in muscle-tendon junction anatomy limit the certainty with which ultrasound can be used to investigate specific distal tendon parts associated with the gastrocnemius and soleus muscles (Cummins & Anson, 1946; Edama et al., 2014; Elson et al., 2007; Szaro et al., 2009; van Gils et al., 1996).

Classically fibre orientations from the both heads of gastrocnemius are likely to be located in the posterior (superficial) and lateral aspects of the Achilles tendon (Edama et al., 2014; Szaro et al., 2009), whereas the portions from soleus lies more anteriorly and medially (Cummins et al., 1946). However, recently Edama et al. (2014) noted that in only half the Achilles cadaver legs investigated (16 cadavers, 25 legs in total) the lateral gastrocnemius fibres were situated in the deep middle portion of the tendon (1 cm above calcaneus intersection). This could imply three rather than two fascicle layers (posterior superficial (medial gastrocnemius), middle and medially (soleus) and anterior (lateral gastrocnemius). These differences might also highlight ethic variation since the studies were on Japanese versus Gaussian cadavers. Given these variations, which part of the tendon is under investigation during walking (Franz et al., 2015), eccentric contractions and passive movement (Slane & Thelen, 2014), or during passive movement (Arndt et al., 2012) is perhaps not as clear as first thought.

Together with the assumption that different muscle portions exert non-uniform force (thus also stress) along each tendon-fascicle component observed in vitro (Arndt et al., 1999a, 1999b), any anatomical variations might have important implications in Achilles tendon injury development. The smallest cross sectional area of the tendon network is located within the midportion of the tendon (Kvist, 1994; Magnusson & Kjaer, 2003; Obst et al., 2014), and thus fascicle formations would result in differential force distributions within this smallest area. These regional anatomical differences in combination with elevated stress and strain rates could indicate why some people have increased risk of injury while others have not.

Recently, Lersch et al. (2012) pointed out that the twist observed within the Achilles tendon may be important in balancing strain distribution in different parts of the tendon. Furthermore, the authors have speculated whether higher increments of eversion result in less longitudinal twisting of the tendon, leading to higher stretch/strain within the medial tendon compartment. The degree of twisting was not measured within that study, nor within this study, so this assumption can not be confirmed. However, the authors suggested that this twisting mechanism, or lack of it, in combination with larger frontal plane calcaneal excursion might play an important role in the onset of injuries of the Achilles tendon.

A final issue with the differential force distribution is how it may affect joint lever arm length and therefore affect muscle and force requirements to generate suitable joint moments. It has been theorised that given muscle contraction could also affect the line of action of the muscle-tendon complex, also that the position and orientation of the tendon structure will change in relation to the ankle and subtalar joint axes of rotations (Maganaris, Baltzopoulos, & Sargeant, 1998; Maganaris et al., 1999). Greater tensile forces in the deeper layers would reduce the effective moment arm and may increase the forces required from muscle contractions to generate the same joint moments.

5.3 Kinematics in those with and without Achilles injuries does not differ.

It is a widely held view that in cases of Achilles tendon pain greater peak eversion and greater range of eversion motion are risk factors in the clinical presentation(Clement et al., 1984; Hreljac et al., 2000; Maffulli et al., 2004; Schepsis et al., 2002; Smart et al., 1980). Indeed, this view underpins the principle that in using a foot orthosis to change rearfoot eversion the cause of the Achilles injury is being addressed.

However, the data reported in the three kinematic studies comparing those with and without Achilles injuries (Donoghue, Harrison, Coffey, et al., 2008; McCrory et al., 1999; Ryan et al., 2009) suggest that rearfoot motion is not always markedly different. For example, McCrory et al. (1999) reported peak eversion values in cases of Achilles pain of 1.9° vs. 2.6° in symptom free individuals. In the same study there was only a slight and non-significant difference in eversion excursion, 10.2° vs. 9.9°. By comparison Ryan and colleagues (2009) found significant differences in eversion excursion of 2° at the ankle joint of those with mid portion Achilles pain (13° vs. 11°) but no significant increase (~1°) in peak eversion (11° vs. 12°). Donoghue et al. (2008a) studied 11 individuals with injured Achilles and reported a non-significant 2.2° increase in peak eversion (~16.1° vs. ~13.8°) compared to symptom free controls. However, greater eversion excursion was reported in cases of Achilles injury, ~21.1° vs. ~16.4° for shod running (no *p*-value stated). The picture is therefore far from clear as to the kinematic differences in cases of Achilles injury.

Furthermore, based on wider literature that reports rearfoot kinematics in pain free individuals for running and walking, the data from Ryan et al. (2009) and McCrory et al. (1999) matches the normal (i.e. pain free subjects) ranges of eversion during running and walking (range 5.5°-12.9°) and peak eversion too (range 2.2° to 11.2°) (Cornwall & McPoil, 1999; McClay & Manal, 1998a; Moseley, Smith, Hunt, & Gant, 1996; Nester et al., 2014). Also, a recent multi segmental kinematic study by Nester et al. (2014) identified that large individual variations in foot kinematic data exist within pain free populations (n=100) and data for those with Achilles tendonpathy general fits into these patterns.

Thus, data for Achilles sufferers in the literature generally falls within the ranges of motion reported for pain free individuals. The frontal plane data reported by Donoghue and colleagues (2008a) is the only study which conflict with data representing ‘normal rearfoot motion’ and could be considered to be evidence of excessive amounts of eversion (peak eversion and excursion) during shod running (goes for both groups). However, the individuals included in their Achilles injured group were selected based on clinical presentation of excessive frontal plane movement. Their control group was not selected nor matched in the same way.

This undermines the theory that rearfoot motion is always, or even often, an important risk factor for Achilles injury. This is important because in this thesis whilst the hypotheses investigated the relationship between Achilles function and injury, the participants did not have Achilles injury. It could be argued therefore that the conclusions of this thesis are likely transferable to people with Achilles injuries.

The evidence for variation between people in rearfoot kinematics and its association with tendon displacement may predispose some people to Achilles injury more than others. Based on the results of chapter 4, on average, the medial Achilles tendon tissue layer underwent ~0.2 displacement per degree of eversion motion (mm/°). Putting this in a pathological context, it might be the case that some individuals display a greater ratio of tendon displacement per degree of eversion. This might indicate a predisposition to greater strain and possibility also localised stress within the tendon.

5.4 Achilles tendon biomechanics and foot orthoses

The clinical significance of the reductions in frontal plane rearfoot angle due to foot orthoses is not known. Although extensive research has been carried out on the effect of orthotics on healthy individuals (Cheung et al., 2011; Mills et al., 2010), this does not seem to strongly correlate with the few studies of the effects on Achilles pain (Wyndow et al., 2010).

The outcomes of chapter 3 indicate that the orthosis tested reduced peak eversion in walking and running by 3.9° and 2.3° respectively, and decreased eversion ROM by 1.6° and 1.4°. This effect is greater than earlier reports in the literature(Cheung et al., 2011; Mills et al., 2010) and thus if a relationship does exist between rearfoot motion and Achilles injury due to rearfoot eversion, this particular orthotic might offer more effect than others. However, whilst the outcomes of chapter 4 allow an estimation of how medial tendon displacement might change due to the reduction in eversion observed in chapter 3, no assumption can be made in terms of the actual change in tendon strain and risk of tissue injury.

It has also been highlighted within chapter 3 that individual kinematic responses to foot orthoses occur. In a similar way, data in chapter 4 reveal person specific relationships between rearfoot position/movement and medial and lateral tendon displacement. Therefore it is also possible that a reduction by one degree introduced by an orthotic (considered by some in the literature to be without clinical significance) may have a greater implication in some subjects compared to others. For example, using data from two subjects from chapter 4, it can be seen that one subject (e.g. subject 11) might display medial tissue displacement of 0.3 mm while another (e.g. subject 7) only 0.1 mm per degree of eversion movement. In this example, angular changes imposed by orthotics could lead to perhaps 3-4 greater reductions in medial tissue displacement in some subjects compared to others.

The findings of this thesis cannot conclude if orthotics are efficient in reducing stress exerted on the tendon, since only tendon displacement was measured. So far only one study has sought to report the direct effects of a foot orthosis on Achilles tendon forces, using inverse dynamics to predict tendon forces. Sinclair et al. (2014) found peak Achilles forces were reduced by 230.2(±67.4) N (i.e. 0.3•BW) when running with an orthosis. Such a change in force (230 N), would result in average stress changes of 46(±13) MPA (assuming small CSA= 5 mm^2, 230/5 N/mm^2 = MPA). Interestingly the authors suggested that these force reductions could be explained by changes in dorsiflexion angles and the effect of frontal plane angles on the tendon forces was not explored.

5.5 Future directions

There are a number of new paths that could follow on from the research conducted in this thesis. These would help further advance the knowledge in the area of the biomechanics of the Achilles tendon and its relationship with rearfoot function, and the role orthotics might have in this relationship.

5.5.1 Moving from static to dynamic studies of Achilles tendon function.

One important area is to transfer the concepts measured in this thesis, i.e. internal displacement in the Achilles and tendon strain and how foot orthoses affect these, from static to dynamic (i.e. gait) studies. Measures of Achilles tendon behaviour during gait have been reported by Franz et al. (2015) and allow tendon behaviour to be investigated in ‘close-to natural’ loaded/dynamic walking conditions. This is more realistic than those loading situations simulated in most static studies and in this thesis. To date the ultrasound tracking approach to Achilles biomechanics has been limited to a small 2D image and sagittal plane analysis of the Achilles. However, this highlights the forward step taken in this thesis, as the frontal plane has not previously been addressed. At present, the assessment of regional strain in the frontal plane might not be transferable into in vivo gait and loaded conditions due to ultrasound probe constraints (i.e. maintaining skin probe contact). However, this points to an opportunity for research into new measurement approaches. Indeed, the work described in Chapter 4 was the outcome of many pilot tests to perfect the measurement approach, and comparable, or indeed longer duration pilot work, is likely required to transfer the frontal plane measurement of tendon behaviour to studies of gait.

If a suitable methodology was developed, studies of orthotic effect on the tendon in vivo could include the effect of anti pronation devices, such as those tested in chapter 3, but also related designs such as heel lifts and footwear adaptations. This may allow measures of rearfoot motion, other data such as joint moments, and intra tendon behaviour to be coupled together. This would provide a more complete picture of the various effects of orthotic and footwear designs on different elements of rearfoot biomechanics.

5.5.2 **Investigating how transverse plane twisting of the tendon occurs. and relate to foot position.**

A further avenue of research could be greater understanding of the transverse plane twisting of tendon fibres and how foot position and motion affect these. This would allow any differences between those with or without Achilles injury and the effect of foot orthoses to be put into a 3D tendon context. This would be a step forward away from a single plane model of Achilles injury (sagittal or frontal plane thus far) and allow a full perspective on Achilles anatomy and function to be created. Before this can be achieved, future studies must focus on development and implementations of 3D measurement approaches and ideally make these suitable for gait. Current 3D ultrasonic image reconstruction techniques have been used by Obst et al. (2014)and Farris, Trewartha et al. (2012), but these only allows mapping of transverse rotation and strains along the length of the entire free tendon component (paratendon) in the sagittal plane as well as longitudinal strains.There are other techniques which may also be useful for imaging transverse and thus the third dimension of tendon behaviour. A recently developed 3D magnetic resonance imaging (MRI) method by Clarke et al. (2015) enables dynamic registration of ankle bone segments, but also allows visualisation of the anterior and posterior curvature of the Achilles tendon fibres during joint motion. This new method would allow dynamic scans of both the Achilles and the distal joints affecting its loading. It is not clear whether this would allow sufficient tracking of specific tendon sub structures, such as fascicles. This method is also promising as it could enable improved accuracy for inverse dynamics calculation in vivo (i.e. tendon-muscle moment arm, line of force action, in vivo mapping of joint rotation axes), which is critical for estimations of tendon stress. Thus, a further overarching theme for future research is the integration of various data types to provide a more complete picture of the Achilles behaviour and how orthotics may relate to the data describing internal tissue behaviour.

5.5.3 **Understanding the independent contributions of the various calf muscle structures to Achilles tendon loading.**

Anatomical and functional studies are needed to explain how the parts of soleus and gastrocnemius muscles contribute to Achilles tendon strain, and how these might change in athletes and whether they relate to injury risk. Throughout this thesis it has been pointed out that there are large anatomical differences in proximal tendon formation in the triceps surae muscle (e.g. Agur et al., 2003; Dalmau-Pastor et al., 2014; Finni et al., 2003a). Whilst this thesis focussed on how distal tendon factors, rearfoot angle for example, affected intra tendon displacement, considering proximal factors would provide a more comprehensive approach to understanding risk of tendon injury. A good starting point would be to conduct cadaver studies and identify anatomical factors and variations in these (e.g. nature of muscle tendon junctions) on a large number of cadavers. These studies might also provide an example of how to accurately define and segment individual tendon fascicles using ultrasound or provide an experimental basis for computer modelling of tendons, since such models rely on valid tissue geometry data. Mapping the proximal calf muscle and tendon arrangement is important as it provides a greater understanding how mechanical stress (force per fascicle area) might be transmitted from the muscles throughout the tendon. The suggestion for a future focus on proximal structures is not to suggest that all the distal factors are now understood, indeed this thesis looked at only one distal factor, frontal plane rearfoot position. Indeed, more research on individual tendon fascicle contact and intersection with the calcaneus is also very relevant as this is an area of high stress concentration.

5.6 Overall conclusion

In past studies the Achilles tendon has been regarded as a single in-series elastic structure originating proximally from several distinct muscle groups and inserting distally on a single site. The literature reviewed in this thesis and the results of the experimental work reveal that the displacement of the tendon and its internal structures, and thus the likely stress within these structures, is far more complex than an in-series elastic element. It is a three dimensional structure, with complex and dynamic responses under load, and these responses change as proximal and distal factors change. In this thesis frontal plane rearfoot position was shown to affect the displacement of the medial and lateral parts of the mid portion of the Achilles tendon. A position and movement of rearfoot eversion induced stretch on the medial side of the tendon, vice versa for inversion position and movement.

Furthermore, interventions to affect Achilles loading and stress often focus on changing rearfoot motion, and in this thesis it has been demonstrated that frontal plane rearfoot position is affected by foot orthoses. Putting the results of the two experiments together, there is clear potential for foot orthoses to reduce the internal stretch experienced by the medial part of the Achilles tendon during rearfoot eversion. This may be a factor contributing to the observation that foot orthoses are clinically beneficial in cases of Achilles injury.

Appendix 1

Table A1.1. Frontal plane parameters for shod and orthotic conditions in walking, and outcomes of additional statistical tests.

WALK			**Shod**	**Orthotic**	*Orthotic effect*	*Orthotic effect*	Quartiles			Statistical outcomes
Code	Definition	Unit	*M*(±*SD*)	*M*(±*SD*)	*M*(±*SD*) reduction	*Mdn* reduction	25^{th}	75^{th}	*IQR*	*p*-value
F2	Peak eversion (MaxEv)	°	-4.4(±3.9)	-0.5(±5.7)	3.9(±3)	4.1	1.5	6.1	4.7	p=0.000***[b]
TF2	Peak eversion time	%	34(±13.6)	39.6(±19.8)	5.6(±15.1)	2.9	0.2	7.7	7.4	p=0.004**[b]
TF3	Peak eversion time <30%	%	24.7(±6.4)	24.8(±6.3)	0.1(±6.9)	0.7	-2.6	1.7	4.4	p=0.639[b]
F4	Toe off (TO) position	°	11.8(±6.4)	13.4(±7.8)	1.6(±3.1)	1.9	-0.8	4.0	4.8	p=0.013*[b]
F5	ROM stance	°	18(±5.7)	15.6(±5.4)	-2.4(±3)	-1.6	-4.7	0.0	4.7	p=0.000***[b]
F8	ROM from MaxEv to TO	°	16.2(±5.8)	13.9(±6)	-2.3(±2.7)	-1.7	-4.0	-0.4	3.6	p=0.000***[b]

Note. *M* =Mean value; *SD*=Standard deviation; *Mdn*= median; 25^{th}= Lower quartile of the median; 75^{th}= Upper quartile of the median; *IQR*=interquartile range, where *IQR*= 75^{th}- 25^{th} quartiles; [b] Wilcoxon signed rank test; *p*= significance level **p*< 0.05. ** *p*<0.01. ***p*<0.001. For further details, see Table 3.1 in chapter 3 for abbreviations and definitions of the kinematics parameters.

Table A1.2. Frontal plane parameters for shod and orthotic conditions in running, and outcomes of additional statistical tests

RUN			**Shod**	**Orthotic**	*Orthotic effect*	*Orthotic effect*	Quartiles			Statistical outcomes
Code	Definition	Unit	*M*(±*SD*)	*M*(±*SD*)	*M*(±*SD*) reduction	*Mdn* reduction	25^{th}	75^{th}	*IQR*	*p*-value
F5	ROM stance	°	21.4(±5.9)	19.8(±6)	-1.6(±2.5)	-1.8	-3.3	0.2	3.5	p=0.001***[b]
F8	ROM from MaxEv to TO	°	19.2(±6.5)	17.4(±6.7)	-1.8(±3)	-1.5	-3.2	0.0	3.6	p=0.001***[b]

Note. *M* =Mean value; *SD*=Standard deviation; *Mdn*= median; 25^{th}= Lower quartile of the median; 75^{th}= Upper quartile of the median; *IQR*=interquartile range, where *IQR*= 75^{th}- 25^{th} quartiles; [b] Wilcoxon signed rank test; *p*= significance level **p*< 0.05. ** *p*<0.01. ***p*<0.001. For further details, see Table 3.1 in chapter 3 for abbreviations and definitions of the kinematics parameters

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