

A thesis submitted for the degree of Doctor of Philosophy

**Methodological and Anatomical Modifiers of  
Achilles Tendon Moment Arm Estimates:  
Implications for Biomechanical Modelling**

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This thesis is dedicated to my wife Annick and my son Etienne.

## **Abstract**

Moment arms are important in many contexts. Various methods have been used to estimate moment arms. It has been shown that a moment arm changes as a function of joint angle and contraction state. However, besides the influence of these anatomical factors, results from recent studies suggest that the estimation of moment arm is also dependent on the methods employed.

The overall goal of this thesis was to explore the interaction between the methodological and anatomical influences on moment arm and their effect on estimates of muscle-tendon forces during biomechanical modelling. The first experiment was a direct comparison between two different moment arm methods that have been previously used for the estimation of Achilles tendon moment arm. The results of this experiment revealed a significant difference in Achilles tendon moment arm length dependent on the moment arm method employed. However, besides the differences found, results from both methods were well correlated. Based on these results, methodological differences between these two methods were compared across different joint angles and contraction states in study two. Results of experiment two revealed that Achilles tendon moment arms obtained using both methods change in a similar way as a function of joint angle and contraction state. In the third experiment, results from the first two experiments were used to determine how methodological and anatomical influences on Achilles tendon moment arm would change muscle-tendon forces during the task of submaximal cycling. Results of the third experiment showed the importance of taking the method, ankle angle and contraction state dependence of Achilles tendon moment arm into account when using biomechanical modelling techniques.

Together, these findings emphasize the importance of carefully considering methodological and anatomical modifiers when estimating Achilles tendon moment arm.

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# Contents

<b>I</b>	<b>Chapter 1</b>	<b>1</b>
<b>1</b>	<b>General introduction</b>	<b>2</b>
<b>II</b>	<b>Chapter 2</b>	<b>5</b>
<b>2</b>	<b>Review of literature</b>	<b>6</b>
2.1	General significance of moment arm measurements . . . . .	6
2.2	Significance of moment arm within the context of biomechanical modelling .	7
2.3	Methodological considerations . . . . .	9
2.3.1	Derivation of the tendon excursion method . . . . .	9
2.3.2	Derivation of the centre of rotation method . . . . .	14
2.4	Anatomical considerations . . . . .	19
2.4.1	Achilles tendon moment arm changes as a function of ankle angle . .	19
2.4.2	Achilles tendon moment arm changes as a function of contraction state . . . . .	20
2.5	Overall summary . . . . .	21
2.6	Thesis aims and research hypotheses . . . . .	22
<b>III</b>	<b>Chapter 3</b>	<b>24</b>
<b>3</b>	<b>Direct comparison of in vivo Achilles tendon moment arms obtained from ten- don excursion and centre of rotation methods</b>	<b>25</b>
3.1	Introduction . . . . .	25
3.2	Methods . . . . .	27
3.2.1	Participants . . . . .	27
3.2.2	Experimental protocol: MR imaging . . . . .	28
3.2.3	Experimental protocol: Ultrasonography . . . . .	29
3.2.4	Moment arm calculation using the tendon excursion method and determination of torques about the ankle joint . . . . .	31

3.2.5	Statistical analysis . . . . .	32
3.3	Results . . . . .	33
3.4	Discussion . . . . .	38
<b>IV</b>	<b>Chapter 4</b>	<b>42</b>
<b>4</b>	<b>Interactive effects of joint angle, contraction state and method on estimates of Achilles tendon moment arms</b>	<b>43</b>
4.1	Introduction . . . . .	43
4.2	Methods . . . . .	44
4.3	Results . . . . .	45
4.4	Discussion . . . . .	45
<b>V</b>	<b>Chapter 5</b>	<b>48</b>
<b>5</b>	<b>The impact of Achilles tendon moment arm methods and anatomical modifiers on Achilles tendon forces during an inverse dynamic simulation of submaximal cycling</b>	<b>49</b>
5.1	Introduction . . . . .	49
5.2	Methods . . . . .	51
5.2.1	Participants . . . . .	51
5.2.2	Experimental protocol: Kinematic and kinetic analysis of cycling (protocol 1) . . . . .	52
5.2.3	Calculation of ankle torque during the cycling tasks . . . . .	52
5.2.4	Experimental protocol: Maximal isometric plantarflexion torque (protocol 2) . . . . .	53
5.2.5	Derivation of dependent variables . . . . .	54
5.2.6	Experimental protocol: Achilles tendon moment arm (protocol 3) . . . . .	54
5.2.7	Derivation of Achilles tendon moment arm dependent variables . . . . .	55
5.2.8	Five methods of deriving Achilles tendon moment arm . . . . .	55
5.2.9	Statistical treatment . . . . .	56
5.3	Results . . . . .	56
5.4	Discussion . . . . .	57
<b>VI</b>	<b>Chapter 6</b>	<b>62</b>
<b>6</b>	<b>General discussion</b>	<b>63</b>
6.1	Summary of findings . . . . .	63

6.2	Effect of moment arm method on Achilles tendon moment arm estimates . . . . .	64
6.3	Effect of anatomical modifiers on Achilles tendon moment arm and Achilles tendon force . . . . .	65
6.4	Significance of findings and suggestions for future research . . . . .	66
6.5	Hypotheses . . . . .	67
6.6	Limitations . . . . .	68
6.7	Summary . . . . .	69
6.8	Conclusions . . . . .	70
<b>VII References</b>		<b>71</b>
<b>References</b>		<b>72</b>
<b>VIII Appendices</b>		<b>83</b>
<b>A Appendix</b>		<b>84</b>
A.1	Additional information . . . . .	84
A.1.1	Influence of knee angle on Achilles tendon moment arm during contraction state . . . . .	84
A.1.2	Residual analysis . . . . .	85
A.1.3	Free-body diagram of the foot segment during cycling . . . . .	87
A.2	Ethical approval . . . . .	88
A.3	Participants health questionnaire . . . . .	92
A.4	Consent form . . . . .	93
A.5	Conference abstract . . . . .	94
A.6	Peer-reviewed publication . . . . .	96

# List of Tables

2.1	Summary of Achilles tendon moment arm reported at rest over a full range of motion (20° dorsiflexion to 30° plantarflexion) . . . . .	17
3.1	Pearson’s product moment correlations between Achilles tendon moment arm obtained from COR and TE methods at a neutral ankle angle, as well as CV for interexperimental set-ups . . . . .	35
4.1	Pearson’s product moment correlations between Achilles tendon moment arm obtained from the COR and the TE methods at four different ankle angles . . . . .	46
5.1	Descriptive kinematic results during submaximal cycling at two resistances (mean ± SD). . . . .	57
5.2	The impact of five Achilles tendon moment arm modifications on the maximal resulting Achilles tendon force and the Achilles tendon force profiles (RAD) during submaximal cycling. . . . .	58

# List of Figures

2.1	Schematic illustration of the centre of rotation method (COR) using Reuleaux' geometrical method. To determine the moment arm at $0^\circ$ (ankle is perpendicular to the tibia), scans at $\pm 15^\circ$ were used to determine the COR of the ankle joint. The tibia was assumed to be constant throughout the rotation and the talus was the rotating segment. The moment arm at $0^\circ$ was defined as the perpendicular distance from the COR of the ankle joint to the line of force, represented by the Achilles tendon. . . . .	15
3.1	Schematic illustration of the relationship between ankle rotation and movement of the muscle-tendon junction (MTJ) of the gastrocnemius medialis and the Achilles tendon. The ankle was rotated from a dorsiflexed position of a minimum of $-15^\circ$ to a plantarflexed position of a minimum of $+30^\circ$ (plantar) and vice versa (dorsi) at $10^\circ \cdot s^{-1}$ (five consecutive rotations resulting in three plantar- and two dorsiflexion rotations). The MTJ shortened or lengthened according to the ankle rotation. . . . .	30
3.2	Schematic illustration of the tendon excursion method. Achilles tendon moment arm was calculated as the first derivative of the ratio of the change in muscle-tendon length represented by the displacement of the muscle-tendon junction (MTJ) of the gastrocnemius medialis and the Achilles tendon ( $\Delta$ MTJ) to the change in ankle angle ( $\Delta$ angle). The derivative was calculated for $0^\circ$ (ankle is perpendicular to the tibia) over five linear intervals ( $\pm 1^\circ$ , $\pm 2^\circ$ , $\pm 5^\circ$ , $\pm 10^\circ$ , $\pm 15^\circ$ numbered as 1 to 5, respectively, in the figure). In addition, a second- and third-order polynomial was fitted to the ratio of the change in muscle-tendon length to the change in angle and differentiated at $0^\circ$ (numbered 6 and 7). . . . .	32

3.3	Comparison of Achilles tendon moment arm results between the center of rotation (COR; MR imaging) and the tendon excursion (TE; ultrasound imaging) methods as the mean of three consecutive plantarflexion rotations for the neutral ankle angle. The moment arm calculated using the COR method was determined over a $\pm 15^\circ$ angular interval using a geometrical approach. The moment arm for the TE method was calculated over five different angular differentiation intervals ( $\pm 15^\circ$ , $\pm 10^\circ$ , $\pm 5^\circ$ , $\pm 2^\circ$ , $\pm 1^\circ$ ) and two polynomial fitting procedures ( $2^{nd}$ - and $3^{rd}$ -order polynomials) as the ratio of tendon to ankle displacement. COR: centre of rotation method; $\pm 15^\circ$ , $\pm 10^\circ$ , $\pm 5^\circ$ , $\pm 2^\circ$ , $\pm 1^\circ$ : angular intervals for differentiation; $2^\circ$ poly: $2^{nd}$ -order polynomial fit; $3^\circ$ poly: $3^{rd}$ -order polynomial fit. * indicates a significant difference between the COR and the TE methods ( $p < .01$ ) . . .	34
3.4	Highest correlation between Achilles tendon moment arms calculated using the centre of rotation method (COR) using MR imaging and the tendon excursion method (TE) using ultrasonography. For the COR method, the moment arm was determined over a $\pm 15^\circ$ angular interval using a geometrical approach. For the TE method, tendon displacement of the musculotendon junction of the gastrocnemius medialis and the Achilles tendon was differentiated over an angular interval of $\pm 10^\circ$ (i.e. $20^\circ$ range) for three consecutive plantarflexion rotations. . . . .	36
3.5	Loading and unloading curves of torques about the ankle joint measured during passive rotations. The ankle joint was rotated from a plantarflexed position of $30^\circ$ to a dorsiflexed position of $15^\circ$ three times at $10^\circ \cdot s^{-1}$ and two times vice versa. * indicates a significantly smaller torque about the ankle joint during plantar- compared to dorsiflexion rotations ( $p < .01$ , two-tailed, Bonferroni corrected). . . . .	37
4.1	Achilles tendon moment arm measurements (mean $\pm$ SD) at four different ankle angles obtained from the tendon excursion method at rest ( $TE_{rest}$ ) and from the centre of rotations method at both rest ( $COR_{rest}$ ) and during a maximum isometric contraction ( $COR_{MVC}$ ). $0^\circ$ ankle angle refers to the foot being perpendicular to the tibia. . . . .	46
A.1	Graph showing the relationship between knee angle and the percentage increase in Achilles tendon moment arm from rest to a maximal voluntary contraction. . . . .	84
A.2	Plot of the residual between a filtered and an unfiltered torque signal which changes as a function of the $4^{th}$ -order Butterworth filter cutoff frequencies (until 500 Hz) . . . . .	86

A.3	Illustration of the estimation of the cutoff frequency as shown in Figure A.2 (zoomed in). Line <i>d</i> to <i>e</i> represents the residual of noise; <i>a</i> represents the intersect of line <i>de</i> with 0 Hz; <i>b</i> represents the intersect of a horizontal line from <i>a</i> to the true signal which rises from line <i>de</i> . The x-coordinate of point <i>b</i> represents the final cutoff frequency of the filter. . . . .	86
A.4	Free-body diagram of the foot segment constraint to the pedal showing reaction and gravitational forces, ankle joint net torque and all linear and angular accelerations. The right foot was assumed to be a rigid segment as well as being stationary with respect to the pedal. . . . .	87

# Chapter 1

# 1 General introduction

In order to describe and analyse human movement, one has to understand the underlying movement mechanics. Movement occurs when skeletal muscles produce forces that are transferred to the joints via tendons (Zajac, 1989). Tendons insert at the joints at a distance resulting in a rotational force (i.e., torque or moment) causing a particular segment to accelerate. The perpendicular distance between the centre of the joint and the muscle-tendon unit is called the moment arm (Pandy, 1999). Moment arms are of fundamental importance in many contexts. For example, strength is often measured as a rotational force using isokinetic dynamometry (Aagaard, Simonsen, Andersen, Magnusson, & Dyhre-Poulsen, 2002; Andersen & Aagaard, 2006). To understand how muscle and tendon interact to produce the individual's strength, the moment arm length is needed for the calculation of muscle forces from joint torques to perform a detailed analysis of muscle function. Thus, moment arm is an important determinant to explore the underlying mechanism that create strength. Another example demonstrating the importance of muscle moment arms is its relevance within the context of movement economy. For instance, running economy (the energy cost for a given running speed) has been shown to strongly correlate with Achilles tendon moment arm length (Raichlen, Armstrong, & Lieberman, 2011; Scholz, Bobbert, van Soest, Clark, & van Heerden, 2008). A smaller moment arm increases the amount of stored elastic energy in the tendon. As a consequence, the increased storage and reutilisation of the elastic energy in the tendon can reduce the metabolic costs of running and therefore increase the running economy (Scholz et al., 2008). In addition to being an important determinant of movement economy, muscle moment arms are also of relevance for transferring muscular forces to the environment effectively. Within this context, it has been shown that the ratio of Achilles tendon moment arm to the distance between the ankle joint and point of ground reaction force application is important (gear ratio). Although a high gear ratio (i.e. small Achilles tendon moment arm in combination with a long foot and toe) places the muscle at a mechanical disadvantage in terms of its moment arm, higher muscular forces can be generated based on the force-velocity relationship as the muscle-fibres have to shorten less for a given joint rotation (Baxter, Novack, Van Werkhoven, Pennell, & Piazza, 2011; Lee & Piazza, 2009). In this context, the higher forces can compensate for the mechanical disadvantage in terms of moment arm. The significance of this gear ratio in terms of running performance is controversially discussed within the scientific community (Baxter et al., 2011; Karamanidis et al.,

2011; Lee & Piazza, 2009; Scholz et al., 2008). While Lee and Piazza (2009) as well as Baxter et al. (2011) reported a smaller Achilles tendon moment arm for sprinters as a possible explanation for their sprinting performance, Karamanidis et al. (2011) did not find a difference. A smaller moment arm for sprinters in combination with a high gear ratio would enhance the muscle force generation and therefore be advantageous for their sprinting performance. Thus, moment arm might be an important determinant to determine sprinting performance. Finally, within a developmental context, moment arms have been used to better understand the underlying mechanisms that cause age-related changes in human movement. O'Brien, Reeves, Baltzopoulos, Jones, and Maganaris (2010) and Waugh, Blazeovich, Fath, and Korff (2012) explored the mechanical properties of the patellar and Achilles tendons in prepubertal children. In order to determine the mechanical properties of the tendons, the authors estimate the individual moment arms for the children that were significantly smaller compared to adults (O'Brien, Reeves, Baltzopoulos, Jones, & Maganaris, 2009; Waugh, Blazeovich, Fath, & Korff, 2011). Morse, Thom, Mian, Birch, and Narici (2007) and Wood, Dixon, Grant, and Armstrong (2006) also reported smaller moment arms for children in comparison to adults, when looking at the strength of children. Thus, moment arm is important to enable exploration of age-related changes in human movement within a developmental context.

From a methodological perspective, moment arms are important because they allow the calculation of muscular forces from torques. As direct measurements of muscular forces are difficult to perform and highly invasive (Finni, Komi, & Lukkariniemi, 1998; Komi, Salonen, Järvinen, & Kokko, 1987), an accurate and reliable determination of moment arms is of fundamental importance to calculate muscular forces from torques. Various methods exist to estimate moment arms at different joints such as the tendon excursion method which has been applied for example at the finger (An, Takahashi, Harrigan, & Chao, 1984; Storace & Wolf, 1979), shoulder (Ackland, Pak, Richardson, & Pandy, 2008), and ankle joints (Spoor, van Leeuwen, Meskers, Titulaer, & Huson, 1990; Ito, Akima, & Fukunaga, 2000); the centre of rotation method (Reuleaux, 1875) which has been applied for example at the elbow (Wood et al., 2006) and the ankle joints (Maganaris, Baltzopoulos, & Sargeant, 1998a; Rugg, Gregor, Mandelbaum, & Chiu, 1990); a method using malleoli markers for the ankle joint (Manal, Cowder, & Buchanan, 2010); the instant centre of rotation (Smidt, 1973); the tibiofemoral contact point (Baltzopoulos, 1995; Kellis & Baltzopoulos, 1999; Tsaopoulos, Baltzopoulos, Richards, & Maganaris, 2007) and the geometric centre of the posterior femoral condyles (Tsaopoulos, Baltzopoulos, Richards, & Maganaris, 2009) for the knee.

In addition to such methodological differences, various anatomical factors also influence the moment arm of a muscle or tendon about a particular joint. The joint angle is a modifier for moment arm. Because the origin and insertion of the muscle-tendon unit

are fixed to the corresponding bones that create the joint, the moment arm is directly influenced by joint angle changes due to the bone geometry (Maganaris et al., 1998a; Spoor et al., 1990). Another factor that may affect the magnitude of muscle or tendon moment arm is the contraction state of the corresponding muscle. When the muscle contracts, the muscle thickness increases and therefore it changes the orientation of the muscle-tendon unit with respect to the centre of rotation of the joint (Baxter et al., 2011; Maganaris et al., 1998a; Maganaris, Baltzopoulos, & Sargeant, 1999; Sheehan, 2007; Wilson & Sheehan, 2009). As the moment arm is defined as the perpendicular distance from the muscle-tendon unit to the centre of rotation, the contraction has a direct influence on the estimation of moment arm.

It becomes clear that both methodological and anatomical factors have dramatic effects on estimates of muscle-tendon moment arms. Not always are such effects taken into consideration when forces are estimated from torques. For example, many musculoskeletal models take the angle dependence of moment arm for their calculation into account, but the dependence of contraction state is not included (Buchanan, Lloyd, Manal, & Besier, 2004; Manal, Gravare-Silbernagel, & Buchanan, 2011).

The overall goal of this thesis was to investigate the interaction between methodological and anatomical influences on the Achilles tendon moment arm and their effect on estimates of muscle-tendon forces during biomechanical modelling. The Achilles tendon was chosen because it is the most important force carrying structure for the lower limb, since it connects the plantarflexor muscles of the triceps surae (soleus, gastrocnemius medialis and lateralis) to the ankle joint at the calcaneus. Thus, there is a high practical application for the Achilles tendon and consequently the Achilles tendon moment arm as it is involved in any human movement including the lower limb (i.e., walking, cycling).

Three experimental chapters were performed. In the first experiment, two different methods of determining Achilles tendon moment arm were analysed. In the second experiment, methodological differences between these two methods were compared across different joint angles and contraction states. Results from the first two experiments were then used to determine how methodological and anatomical influences on Achilles tendon moment arm would alter the quantification of muscle-tendon forces during a complex lower limb motor task (submaximal cycling).

# Chapter 2

## 2 Review of literature

### 2.1 General significance of moment arm measurements

To understand the importance of moment arms for the human body to produce movement and consequently for biomechanical modelling, one has to explore how the underlying structures that cause movement are linked together and how they interact. In order to produce any motion or movement, the production of forces is needed. It is well documented for skeletal muscles how the contractile elements of the muscles fibres produce muscular forces based on the cross-bridge theory as well as the force-length and force-velocity relationship (Delp et al., 1990; Herzog, Read, & Keurs, 1991; Huxley, 1957; Zajac, 1989). In order to transfer the muscular forces to the joints to produce the desired movement, the skeletal muscles are in series with tendons that insert onto the skeleton (Buchanan et al., 2004; Zajac, 1989). To cause a segmental rotation and therefore a torque about the joint, the tendon does not insert at the centre of the joint but inserts at a distance acting as a lever (principle of Newton's second law of motion). The perpendicular distance between the centre of the joint and the muscle-tendon unit is defined as the moment arm (Pandy, 1999; Spoor et al., 1990). As the rotational torque can be calculated as the product of the muscular forces times the corresponding moment arms, the importance of moment arm length becomes apparent. For example, a given amount of muscular force will produce a greater rotational torque if the moment arm is greater. Therefore, the length of the moment arm has a direct influence on joint torque, power and work output but also on movement economy (Miller & Melissa Gross, 1998; Nagano, Komura, Himeno, & Fukashiro, 2003; Raichlen et al., 2011; Scholz et al., 2008; Voigt, Bojsen-Møller, Simonsen, & Dyhre-Poulsen, 1995). Recently, the anatomical advantage of the gear ratio of long distance runners and sprinters is discussed within the scientific community (Raichlen et al., 2011; Scholz et al., 2008). As an example of the importance of moment arms, their results show that the running economy is strongly related to the Achilles tendon moment arm ( $MA_{AT}$ ) which transfers the plantarflexor forces to the ankle joint. Scholz et al. (2008) reported shorter  $MA_{AT}$  for long distance runners measured by 2d photographic images causing a greater storage and release of elastic strain energy (Raichlen et al., 2011). However, these findings are in contrast to the  $MA_{AT}$ s results determined by Baxter et al. (2011) and Lee and Piazza (2009) using magnet resonance imaging. The authors demonstrated that the  $MA_{AT}$ s for sprinters were shorter in comparison to non-sprinters. Interestingly,

in a recent study by Karamanidis et al. (2011) the tendon excursion method (An, Ueba, Chao, Cooney, & Linscheid, 1983) using ultrasonography was applied. The authors found no significant correlation between sprint performance and  $MA_{AT}$ . Within this context, as all three studies used different methods to estimate  $MA_{AT}$ , the question arises of whether the method used can influence the  $MA_{AT}$  length. Regardless of the potential influence of methods on moment arm estimates, the knowledge of moment arms helps us to better understand the underlying mechanism of human movement such as muscular forces as the moment arm is needed to determine muscular forces from angular torques.

## 2.2 Significance of moment arm within the context of biomechanical modelling

The aim of biomechanical modelling techniques is to replicate and analyse human movement in order to explore the underlying mechanisms that cause the movement (Winter, 2009). Direct measurements of *in vivo* muscular forces can be performed using for example a buckle-type force transducer which has to be surgically implanted (Fukashiro, Komi, Järvinen, & Miyashita, 1993; Gregor, Komi, Browning, & Järvinen, 1991; Komi et al., 1987) or an optic fibre technique which has to be inserted through the force carrying structure (i.e. tendons) (Finni et al., 1998) The use of these invasive techniques has to be performed under local anaesthesia. While these techniques provide valuable information about the force production of muscles, their practical application is limited due to their invasive nature.

As an alternative to the invasive direct measurements, researchers have estimated muscular forces or muscular torques indirectly using alternative biomechanical modelling techniques, such as the inverse or forward dynamics approaches in order to analyse human movement (Winter, 2009). Studies using inverse and forward dynamics to estimate muscular forces or joint torques include, for example, walking (Anderson & Pandy, 2001; Piazza & Delp, 1996; Thelen & Anderson, 2006; Zajac, Neptune, & Kautz, 2002), jumping (Pandy & Zajac, 1991) and cycling (Fregly & Zajac, 1996; Gregor et al., 1991; Hull & Jorge, 1985; Martin & Spirduso, 2001; Neptune & Hull, 1998; Raasch, Zajac, Ma, & Levine, 1997).

Within this context, in order to compute muscular torques and consequently muscular forces, inverse dynamics uses kinematic measures such as the position, velocity and acceleration of body segments obtained using motion analysis as well as kinetic measures that act on the human body such as ground reaction forces obtained from force platforms (Winter, 2009). In addition to the kinematic and kinetic measures, information about the segmental masses, moment of inertia and the centre of mass are needed which have to be scaled to the participant's anthropometrics using regression equations (Jensen, 1989;

Winter, 2005). To compute the net muscular forces that produce the muscular torques, information about the corresponding musculoskeletal geometry to estimate the moment arms of the muscles are needed. Ericson, Ekholm, Svensson, and Nisell (1985), for example, divided the plantarflexor torque during cycling by a constant moment arm. However, it has been shown that moment arm changes as a function of joint angle (Delp et al., 1990; Grieve, Pheasant, & Cavanagh, 1978; Klein, 1996; Maganaris et al., 1998a). By assuming a constant moment arm throughout the range of motion of the ankle joint, the resulting muscular forces are over- or underestimated. When using musculoskeletal models such as OpenSim or SIMM which include information about the geometry of the muscle-tendon unit as well as the bony segments (Delp et al., 1990; Wickiewicz, Roy, Powell, & Edger-ton, 1983; Yamaguchi, Sawa, Moran, Fessler, & Winters, 1990); the angle dependence of moment arms can be taken into account using the tendon excursion method (An et al., 1984). The tendon excursion method computes the moment arm as the ratio of changes in muscle-tendon length to changes in joint angle (Buchanan et al., 2004). The origin and insertion of the muscles and tendons as well as the paths of the muscle-tendon units have to be well defined in the model as they define the length of the muscle-tendon unit (Erdemir, McLean, Herzog, & van den Bogert, 2007).

Alternative to the application of moment arms within the context of multi-joint movements using inverse or forward dynamics, researcher determined muscular torques and muscle moment arms to calculate muscular forces at single joint movements (Blazevich, Coleman, Horne, & Cannavan, 2009; Maganaris, 2003a; Morse et al., 2008). To estimate the individual moment arm, researchers have used different methods such as the tendon excursion method (An et al., 1984; Grieve et al., 1978; Spoor et al., 1990) and the centre of rotation method (Reuleaux, 1875; Rugg et al., 1990) *in vitro* (Hintermann, Nigg, & Sommer, 1994; Klein, 1996; Spoor et al., 1990; Visser, Hoogkamer, Bobbert, & Huijing, 1990) as well as *in vivo* using magnet resonance imaging (MRI) (Maganaris et al., 1998a; Rugg et al., 1990; Spoor & van Leeuwen, 1992; Wood et al., 2006) and ultrasonography (Fukunaga, Ito, et al., 1996; Ito et al., 2000; Lee & Piazza, 2009; Maganaris, 2003a), an approach combining ultrasonography and motion analysis (Manal et al., 2010) for the ankle joint as well as methods specifically for the knee such as instant centre of rotation (Smidt, 1973), the tibiofemoral contact point (Baltzopoulos, 1995; Kellis & Baltzopoulos, 1999) and the geometric centre of the posterior femoral condyles (Tsaopoulos et al., 2009).

Within this context, measurements of *in vivo* moment arms using MR imaging have shown that the moment arms not only change as a function of angle but also as a function of contraction as it significantly increased from rest to measurements taken during a maximal voluntary contraction (MVC) (Maganaris et al., 1998a, 1999). In these studies, the researcher used the centre of rotation methods (Reuleaux, 1875; Rugg et al., 1990) to estimate the Achilles and tibialis anterior moment arm as the perpendicular distance from

the line of force (i.e. Achilles and tibialis anterior tendons) to the centre of rotation of the ankle joint. In contrast, when using the tendon excursion method to estimate the moment arms from rest and MVC, the researcher did not find an increase in moment arm length (Maganaris, Baltzopoulos, & Sargeant, 2000; Maganaris, 2000). As estimates of moment arm as part of the model parameter greatly influence the prediction of muscular forces and torque (Maganaris, Baltzopoulos, Ball, & Sargeant, 2001; Menegaldo & Oliveira, 2009), the increase in moment arm as a function of contraction state should be included in the musculoskeletal models.

## 2.3 Methodological considerations

For lower limb models, the Achilles tendon is the most important force carrying structure, as it connects the plantarflexor muscles of the triceps surae to the ankle joint. To date, there have been two methods to determine the  $MA_{AT}$  *in vitro* and *in vivo* which have become popular within the scientific community: the tendon excursion (TE) and the centre of rotation (COR) method which will be explained in more detail in the context of this thesis.

### 2.3.1 Derivation of the tendon excursion method

The TE method is based on the relationship between changes in muscle-tendon length with respect to joint rotation. The resulting moment arm is defined as the derivative of this relationship (An et al., 1984; Grieve et al., 1978). In comparison to the COR method, no knowledge of the centre of the joint is needed. Over the years, researchers have used different approaches to derive equations to calculate moment arms. In an *in vitro* study, Grieve et al. (1978) quantified the length changes of the gastrocnemius medialis muscle-tendon unit by rotating the specimens' ankle and knee joint through their range of motion. While one of the joints stayed constant, the other joint was rotated in increments of 10°. The resulting muscle-tendon displacement was manually measured for each joint rotation. The authors reported an equation to predict muscle-tendon length ( $\Delta L$ , expressed in terms of percent of segment length) as a function of ankle and knee angles ( $\theta_i$ , expressed in degrees)

$$\Delta L = A_0 + A_1 \cdot \theta_i + A_2 \cdot (\theta_i)^2 \quad (2.1)$$

which enables researchers to predict muscle-tendon length without direct measurements. Following Grieve et al. (1978), Bobbert, Huijing, and van Ingen Schenau (1986) rearranged equation 2.1 with respect to time. The aim was to calculate the moment arm ( $d$ ) of the triceps surae at the ankle joint by calculating the first derivative of the shortening and

lengthening velocity ( $V$ ):

$$\begin{aligned} V &= A_1 \cdot \dot{\theta}_i + 2 \cdot A_2 \cdot \theta_i \cdot \dot{\theta}_i \\ &= \dot{\theta}_i \cdot d \end{aligned} \quad (2.2)$$

In order to determine the resulting moment arm, the two terms in equation 2.2 are equalised and solved for the moment arm ( $d$ )

$$d = \frac{180}{\pi} \cdot (A_1 + 2 \cdot A_2 \cdot \theta_i) \quad (2.3)$$

Although the original equation 2.1 to determine the length of the gastrocnemius (Grieve et al., 1978) was defined as a function of ankle and knee rotation, the influence of the knee angle was neglected for the resulting moment arm ( $d$ ) of the ankle (equation 2.3) (Bobbert et al., 1986).

A different approach to calculate muscle moment arm was based on the equilibrium equation and the principle of virtual work (An et al., 1984; Storace & Wolf, 1979). In this context, Storace and Wolf (1979) applied the virtual work principle to derive the force equilibrium equation. The virtual work principle states, that if an external torque ( $T$ ) is being applied to a distal segment of a joint resulting in forces being generated in muscles and tendons ( $F_i$ ), the displacement of the tendons ( $\Delta L_i$ ) is a function of the joint displacement ( $\Delta\theta$ ). In this context, ligaments are treated as inextensible and bony contact points of joints as frictionless.

$$\sum_{i=1}^n F_i \cdot \left( \frac{\Delta L_i}{\Delta\theta} \right) + T = 0 \quad (2.4)$$

In the above equation (2.4), the term  $\left( \frac{\Delta L_i}{\Delta\theta} \right)$  represents the moment arm for every muscle-tendon unit ( $T_i$ ) (An et al., 1983, 1984). In equation 2.4 an external torque ( $T$ ) was applied to the joint. In contrast to An et al. (1984), Spoor et al. (1990) developed their moment arm equation not by applying an external torque to the joint but by focusing on how muscular forces ( $F$ ) change tendon length ( $\Delta L$ ) and, as a consequence, causing a joint rotation ( $\Delta\theta$ ) and joint torque ( $T$ )(see equation 2.5).

$$F \cdot \Delta L = T \cdot \Delta\theta \quad (2.5)$$

It is obvious that when rearranging equation 2.5 for torque ( $T$ ),

$$T = \frac{\Delta L}{\Delta\theta} \cdot F \quad (2.6)$$

equation 2.6 represents the moment arm as described in equation 2.4. The advantage of

equation 2.6 compared to 2.4 lies in the fact, that the moment arm from Spoor et al. (1990) takes changes in the orientation of the force vector into account. When a force acts perpendicular to a joint axis, the distance between the centre of rotation of the axis and the line of force is defined as the moment arm. However, any deviation  $\alpha$  from the perpendicular angle would eventually change the corresponding moment arm as well as the resulting joint torque. This deviation has to be taken into account. When applying equation 2.6, changes in  $\alpha$  can be taken into account by multiplying the moment arm with  $\sin \alpha$ . In this context, the moment arm is called the effective moment arm (Spoor et al., 1990).

### **Previous studies and findings using the tendon excursion method**

The TE method has been used to estimate the  $MA_{AT}$  *in vitro* (Grieve et al., 1978; Hintermann et al., 1994; Klein, 1996; Spoor et al., 1990; Spoor & van Leeuwen, 1992), as well as *in vivo* using MR imaging (Maganaris et al., 2000) and ultrasonography (Fath, Blazeovich, Waugh, Miller, & Korff, 2010; Karamanidis et al., 2011; Lee, Lewis, & Piazza, 2008; Lee & Piazza, 2009; Maganaris, 2003a; Waugh et al., 2011) (see Table 2.1).

For the *in vitro* testing, the authors used a micrometry experimental measurement technique to determine muscle-tendon length changes as a function of joint rotation (Grieve et al., 1978; Klein, 1996). More specifically, these studies demonstrated that  $MA_{AT}$  increased from dorsiflexion to plantarflexion across the range of motion of the ankle joint (20° dorsi to 30° plantar). The exception of the *in vitro* studies is the study by Spoor et al. (1990) who reported on increase in  $MA_{AT}$  from dorsiflexion to the neutral angle followed by a decrease in moment arm to the most plantarflexed angle.

Maganaris et al. (2000) were the first to apply the tendon excursion method *in vivo* using MR imaging during rest and maximal voluntary contraction (MVC). To estimate the  $MA_{AT}$  using MR imaging, sagittal images of the ankle joint were taken every 15° over the full range of motion. To define the change in tendon length, the Achilles tendon insertion point on the calcaneus was tracked over two images by superimposing the shape of the calcaneus. As a reference point, the tibia is considered the rotating segment (Maganaris et al., 2000). The authors found no significant difference for  $MA_{AT}$  using the tendon excursion method at rest and during contraction state.

Since the use of real time ultrasonography has been shown to produce reliable and valid measurements (Fukunaga, Ito, et al., 1996; Fukashiro, Rob, Ichinose, Kawakami, & Fukunaga, 1995; Ito et al., 2000; Narici et al., 1996), the TE method using ultrasonography during plantarflexion has been applied in a few studies (Karamanidis et al., 2011; Lee et al., 2008; Lee & Piazza, 2009; Maganaris, 2003a). Although all studies were looking at the Achilles tendon, absolute moment arms lengths differed significantly in size (see Table 2.1). These differences can only partially be explained by the testing protocol. Lee and

Piazza (2009) manually rotated the foot from 20° dorsiflexion to 30° plantarflexion in three seconds while performing a MVC. The displacement data were analysed by an automatic tracking algorithm based on speckle tracking (Lee et al., 2008). In order to calculate  $MA_{AT}$ , a polynomial function was fitted through the tendon and angular displacement data and differentiated at the angle of interest. In contrast, Maganaris (2003a) calculated their  $MA_{AT}$  over a range of 10° taking ultrasound still images  $\pm 5^\circ$  from the ankle angle of interest during rest. Interestingly, in a recent study by Karamanidis et al. (2011), the authors applied the same technique as described by Maganaris (2003a). However, their  $MA_{AT}$  length results are shorter and are in line with those reported by Maganaris (2003a) (see Table 2.1).

### **Limitations of the tendon excursion method**

Errors may have been introduced in various ways for the moment arm estimation using the TE method. First, in all *in vitro* and *in vivo* studies using the TE method, it is assumed that the ankle joint axis is perpendicular to the scanning plane (i.e. sagittal plane) (Hintermann et al., 1994; Klein, 1996; Lee & Piazza, 2009; Maganaris et al., 2000; Maganaris, 2003a; Spoor et al., 1990). However, a deviation from this assumption can lead to an underestimation of  $MA_{AT}$  (Klein, 1996; Spoor et al., 1990) (see equation 2.6). A similar effect on  $MA_{AT}$  is the assumption that the muscle-tendon unit acts in a straight line. This may be true in *in vitro* experiments using pulling devices, but it has been shown that during *in vivo* conditions the line of force changes its orientation due to a significant increase in muscle thickness (Maganaris, Baltzopoulos, & Sargeant, 1998c). However, care has to be taken when applying the TE method during muscular contraction. The TE method is based on the principle of virtual work (An et al., 1984; Storace & Wolf, 1979) and therefore assumes that the work done by an external torque is equivalent to the virtual work done by the muscles and tendons. Implicit in this is the assumption that no energy is lost during a muscle contraction (ligaments are treated as inextensible and bony contact points of joints as frictionless). Given that muscles and tendons store, release and dissipate elastic energy during muscle contractions, the principle of virtual work is violated, and this violation is likely to be more significant when large muscle forces are produced. Additional errors may be introduced when measuring the displacement of the muscle-tendon unit ( $\Delta L$ ). The displacement of the muscle-tendon complex is dependent on the conditioning of the tendon due to the tendon's viscoelastic nature (Maganaris, 2003b). Prior to *in vivo* experiments tracking the muscle-tendon junction using ultrasonography (Lee et al., 2008), the tendon should be stretched a few times using maximum contraction as the tendon properties are history and load dependent (Maganaris, 2003b). Within this context, if the TE method is applied during submaximal or maximum contractions without conditioning (Lee et al., 2008; Lee & Piazza, 2009), it has been reported that the

displacement of the muscle-tendon complex measured at the muscle-tendon junction has to be corrected for heel movement (Karamanidis et al., 2005; Maganaris, 2005). As the heel lifts off the measuring device during the contraction, the calcaneus is shifted proximodistally by about 13 mm (Maganaris, 2005). When only measuring the displacement at the muscle-tendon junction, this will lead to an underestimation of the muscle-tendon displacement for a given angular rotation.

When calculating the moment arm based on the principle of virtual work and the equilibrium equation (see equation 2.4), the external torque  $T$  must be equal to the sum of all internal muscle and tendon forces  $F$  (An et al., 1984; Storage & Wolf, 1979). Therefore, the presence of active or passive forces during passive ankle joint rotation would violate the principle of virtual work and could affect the accuracy of the moment arm estimate. In addition, Spoor et al. (1990) indicated that moment arm measurements might be dependent on movement direction of the ankle joint (dorsiflexion-plantarflexion and vice versa) due to hysteresis. This is in line with other studies looking at the loading and unloading of the tendon (Kay & Blazevich, 2008). Differences in the magnitude of forces during loading and unloading could result in differences in the reliability of moment arm measures obtained during different movement directions.

Within the TE literature, researchers have modelled the ratio of tendon and angular displacement ( $\Delta L/\Delta\theta$ ) with a first (Grieve et al., 1978; Ito et al., 2000; Maganaris et al., 2000; Maganaris, 2003a), a second or third (Lee & Piazza, 2009; Murray, Delp, & Buchanan, 1995) as well as a fifth polynomial fit or quintic spline (Spoor et al., 1990). A higher polynomial fit accounts for specific changes in moment arm, especially at both ends of the range of motion. However, the choice of how to best represent the ratio of tendon and angular displacement seems to be unclear although it directly influences the  $MA_{AT}$  length especially at both ends of the range of motion.

## Summary

Previous studies have demonstrated that  $MA_{AT}$  can be determined using the TE method *in vitro* and *in vivo* using different testing protocols such as micrometry, MR imaging and ultrasonography. As a result, these studies have demonstrated that  $MA_{AT}$  increases as a function of ankle angle from dorsiflexion to plantarflexion. Additionally, it was shown that the TE method cannot be used to determine changes in  $MA_{AT}$  during muscular contraction. For all approaches, the absolute length of the  $MA_{AT}$  is similar with the exception of MR imaging. When comparing these results to studies using alternative methods such as a combination of ultrasonography and motion analysis (Manal et al., 2010) or direct 3D moment arm estimates using MR imaging (Hashizume et al., 2011), moment arm lengths are similar. However, when comparing the results of the TE method using ultrasonography to results obtained using the centre of rotation method, the moment

arm length is smaller for the TE method. Therefore, there seems to be a lack of research for directly comparing different methods to determine  $MA_{AT}$  to understand if these differences in length are present.

### 2.3.2 Derivation of the centre of rotation method

As an alternative to the TE method, the  $MA_{AT}$  can also be estimated using the COR method. The calculation of the  $MA_{AT}$  using this approach is based on the definition that the moment arm is defined as the perpendicular distance of the COR of a joint to the muscle-tendon line of force. In the particular case of the ankle joint, the Achilles tendon is considered to be the line of force as it transfers the force of the triceps surae to the calcaneus to create a plantarflexion movement of the ankle. In order to estimate the  $MA_{AT}$  based on this definition, the centre of the ankle joint as well as the path of the Achilles tendon have to be located. In order to locate the COR of the ankle joint, a geometric approach is used (Reuleaux, 1875). According to this approach, the rotation axis or temporary centre of two bodies can be calculated if one of the segments is fixed in space and the relative movement between the segments is measured. For the ankle, one of the segments is the rotation axis of the ankle joint which passes through the talus bone in a medial to lateral direction. The second segment is the tibia and fibula representing the shank which rotates around the rotation axis of the ankle for dorsi- or plantarflexion movement (van den Bogert, Smith, & Nigg, 1994). It is not important which segment is defined as the fixed segment. Rugg et al. (1990), for example, chose the talus whereas Maganaris et al. (1998a) the tibia as the fixed segment. However, in both cases the researcher used MR imaging for their *in vivo* testing. The advantage of MR imaging, in contrast to X-ray for example, is that MR imaging not only visualises bony structures but also human tissue such as tendons as the contrast of the images is based on redirecting water atoms. To determine the centre of rotation between the two segments at a specific joint angle, MR images have to be taken, for example,  $15^\circ$  before and after the joint angle of interest (Maganaris et al., 1998a, 2000; Rugg et al., 1990). It has been reported that the range of angle ( $\pm 15^\circ$ ) greatly influences the accuracy of the COR calculation (Panjabi, 1979; Panjabi, Goel, & Walter, 1982).

In order to define the two segments for the COR calculation, two identical anatomical reference points are chosen on the rotating segment at the first angle (A, B; i.e.  $-15^\circ$ ) and at the second after the rotation (A', B'; i.e.  $+15^\circ$ ). The COR is defined as the intersection of the perpendicular bisector lines of line  $\overline{AA'}$  and line  $\overline{BB'}$  by superimposing the two images ( $\pm 15^\circ$ ) on to the image of the joint angle of interest (see Figure 2.1). The resulting COR is only one discrete point. The Achilles tendon line of action can be represented as a straight line from the insertion point of the Achilles tendon (calcaneus) along the centre of the tendon. Consequently, the  $MA_{AT}$  can be measured as the perpendicular distance

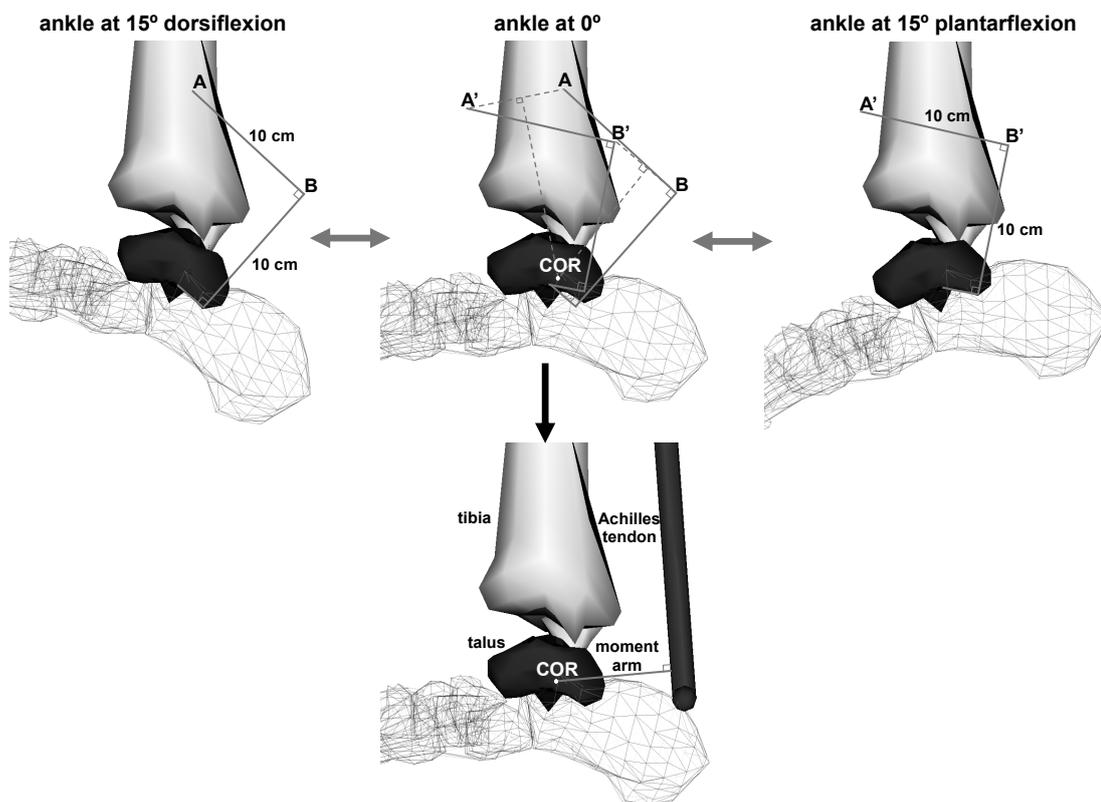


Figure 2.1. Schematic illustration of the centre of rotation method (COR) using Reuleaux' geometrical method. To determine the moment arm at 0° (ankle is perpendicular to the tibia), scans at  $\pm 15^\circ$  were used to determine the COR of the ankle joint. The tibia was assumed to be constant throughout the rotation and the talus was the rotating segment. The moment arm at 0° was defined as the perpendicular distance from the COR of the ankle joint to the line of force, represented by the Achilles tendon.

from the Achilles tendon to the estimated COR of the ankle joint at the angle of interest (Figure 2.1).

### **Previous studies and findings using the centre of rotation method**

*In vivo*  $MA_{AT}$  using MR imaging was first determined by Rugg et al. (1990). The authors used a modification of Reuleaux' graphical approach in order to determine the influence of a fixed and a moving COR on the Achilles and tibialis anterior moment arm estimation (Reuleaux, 1875; Sammarco, Burstein, & Frankel, 1973). To determine the fixed COR, the ankle joint was rotated through its maximum range of motion from dorsi- to plantarflexion while performing a submaximal contraction. The resulting fixed COR was defined as the average location across the range of motion. In contrast, the moving COR was determined multiple times over smaller ankle joint rotations ( $20^\circ$ ) across the full range of motion. As a result, the authors found an increase of approximately 20% in  $MA_{AT}$  from maximal dorsi- to maximal plantarflexion. When comparing the influence of a fixed versus moving centre of rotation on the resulting  $MA_{AT}$ , there was no significant difference (although moment arm was 3.1% greater when using a fixed COR) (Gregor et al., 1991; Rugg et al., 1990).

Maganaris et al. (1998a, 2000) extended these findings by determining the influence of contraction state on  $MA_{AT}$  estimates (i.e. rest vs. MVC). In their previous work, the authors had shown using ultrasound, that the orientation of the triceps surae and therefore the line of force for the COR method changed throughout a MVC due to a significant increase in muscle thickness (Maganaris et al., 1998c). As a consequence,  $MA_{AT}$  length was 22% - 27% greater during MVC than during rest (Maganaris et al., 1998a). In contrast, a recently published study by Baxter et al. (2011) did not find a difference in  $MA_{AT}$  length between rest and MVC using the same experimental setup. According to the authors, the differences between the studies could be caused by the ankle strapping technique as well as the level of co-contraction during the plantarflexion contraction. In addition,  $MA_{AT}$  also increased as a function of ankle angle from dorsi- to plantarflexion (about 25%) which is in line with Rugg et al. (1990) and later studies (Maganaris et al., 2000; Maganaris, 2004; Maganaris, Baltzopoulos, & Sargeant, 2006; Magnusson, Aagaard, Rosager, Dyhre-Poulsen, & Kjaer, 2001)(see Table 2.1). The length of  $MA_{AT}$  is comparable across studies as the sample size and anthropometric measures of the participants were similar.

### **Limitations of the centre of rotation method**

When using the COR method to estimate  $MA_{AT}$ , researchers also have to consider the limitations of this approach. In general, errors can be introduced by not correctly representing the anatomy of the ankle joint and its rotational axis and by the influence of measuring errors due to human errors. Starting with the anatomical representation, there are major limitations of the COR method using MR imaging. The first assumption is

Table 2.1. Summary of Achilles tendon moment arm reported at rest over a full range of motion (20° dorsiflexion to 30° plantarflexion)

Study	MA length (mm)	Type	Method
Spoor et al. (1990)	27 - 55	in vitro	$TE^a$
Rugg et al. (1990)	51 - 60	in vivo	$COR^1$
Rugg et al. (1990)	49 - 59	in vivo	$COR^2$
Gregor et al. (1991)	47 - 53	in vivo	$COR^2$
Hintermann et al. (1994)	29 - 54	in vitro	$TE^a$
Klein (1996)	43 - 62	in vitro	$TE^a$
Maganaris et al. (1998a, 2000)	44 - 55	in vivo	$COR^2$
Maganaris et al. (2000)	43 - 56	in vivo	$TE^b$
Magnusson et al. (2001)	51*	in vivo	$COR^2$
Maganaris (2003a)	51 - 70	in vivo	$TE^c$
Maganaris et al. (2006)	47 - 61	in vivo	$COR^2$
Lee and Piazza (2009)	42*	in vivo	$TE^c$
Fath et al. (2010)	46 - 57	in vivo	$COR^2$
Hashizume et al. (2011)	46 - 56	in vivo	$COR^2$
Hashizume et al. (2011)	35 - 41	in vivo	$3D^d$
Karamanidis et al. (2011)	42*	in vivo	$TE^c$
Baxter et al. (2011)	58.5*	in vivo	$COR^2$
Sheehan (2012)	53*	in vivo	$3D^d$

Note:  $TE^a$  = tendon excursion method applied in cadaver specimens where the Achilles is directly fixed in a pulling apparatus to measure tendon displacement;  $TE^b$  = tendon excursion method applied using magnet resonance imaging;  $TE^c$  = tendon excursion method applied using ultrasonography;  $3D^d$  = three dimensional direct measurements using magnet resonance imaging (only -10° to 20°); COR = centre of rotation method based on graphical approach with a fixed ( $COR^1$ ) and a moving centre of rotation ( $COR^2$ ) (Reuleaux, 1875; Rugg et al., 1990); \* = measurements only taken at 0°.

based on the anatomical design of the ankle joint. The talar bone is part of two joints. On the one hand the tibio-talar joint allowing plantarflexion-dorsiflexion movement and on the other hand the talo-calcaneo-navicular joint allowing inversion-eversion movement (Donatelli, 1996; Hicks, 1953; Lundberg & Svensson, 1993; Lundberg, Svensson, Németh, & Selvik, 1989; van den Bogert et al., 1994). It is often assumed that the plantarflexion or dorsiflexion movement only occurs in 2D in a sagittal plane. This way the ankle joint is treated as a planar mechanism (Ito et al., 2000; Maganaris et al., 1998a, 2000; Maganaris & Paul, 1999; Rugg et al., 1990; Sammarco et al., 1973). While the tibio-talar joint only has one degree of freedom, its axis of rotation is not perpendicular to the scanning plane and changes as a function of ankle rotation (Isman & Inman, 1969; Leardini, OConnor, Catani, & Giannini, 1999; Lundberg & Svensson, 1993; Lundberg et al., 1989; van den Bogert et al., 1994). The misalignment of the sagittal scanning plane to the actual anatomical rotation axis can be up to  $10^\circ$  and can be corrected by using trigonometry as described in equation 2.5 (Isman & Inman, 1969; Spoor et al., 1990; van den Bogert et al., 1994). Besides the error from the misalignment of the tibio-talar axis to the scanning axis, a shift of the talo-calcaneo-navicular axis can also occur during muscular contraction (MVC) creating an inversion-eversion movement due to ligament compliance (van den Bogert et al., 1994). In addition, when comparing the orientation of the talo-calcaneo-navicular joint to the insertion point of the Achilles tendon on the calcaneus, a lateral deviation can be noted. During loading of the Achilles tendon, this can cause an eversion of the calcaneus (Donatelli, 1996; Maganaris et al., 2000). However, measurement errors due to misalignments of the tibio-talar and talo-calcaneo-navicular joint compared to the scanning plane cannot be taken into account when using the 2D geometrical approach to determine the COR (Reuleaux, 1875). In a recent three dimensional approach to determine the  $MA_{AT}$ , a misalignment between the sagittal MR scan to the talocrural joint axis of  $21.4^\circ$  on the transverse and  $14.8^\circ$  and the coronal plane was reported (Hashizume et al., 2011). As a result, the length of 3D versus 2D  $MA_{AT}$ s was significant smaller ( $p < .01$ ) (see Table 2.1).

In order to determine the COR for the moment arm estimation, images of the stationary and rotating segments (i.e. tibia and talus) have to be taken over a deviation of angles (Reuleaux, 1875). This approach incorporates two limitations. First, if images at  $\pm 15^\circ$  are taken, it is not possible to determine the COR and therefore the moment arm for angles at the end of the range of motion. Second, if the range of  $30^\circ$  is reduced, errors in the calculation of locating the COR rotation will be introduced (Panjabi et al., 1982; Panjabi, 1979; Spiegelman & Woo, 1987).

In addition, the multiple steps of applying Reuleaux's method to determine the COR can introduce errors. By not correctly superimposing overhead transparencies or not accurately measuring the required distance of 10 cm and the  $90^\circ$  angles, the resulting COR can deviate from the actual position (Maganaris et al., 1998a). Maganaris et al. (1998a)

reported a coefficient of variation for 15 repeated COR measures on one image of 4.8% deviation in horizontal and of 4.1% deviation in vertical direction from a constant point (Klein, 1996). This resulted in a 5.1% deviation of  $MA_{AT}$  for intraday measures. Three repeated scans for one participant showed a 7.9% coefficient of variation in moment arm measurements. These studies demonstrate the importance of multiple COR measurements for a given joint angle.

Finally, when applying the COR method during muscular contractions using MR imaging, it is not possible to test the validity and reliability of participants' maximal ankle torque. While direct measurements on an isokinetic dynamometer are possible for familiarisation purposes before the testing, a comparison to the MR testing cannot be made. In addition, it has been shown that during maximal isometric contractions, it is difficult to prevent ankle rotation by external strap fixation (Arampatzis et al., 2005; Magnusson et al., 2001; Rosager et al., 2002). As the MR images are taken at the end of the contraction, it is not possible to measure the angular displacement. For example, assuming that the ankle rotation was similar for  $\pm 15^\circ$  and for the neutral ankle angle ( $0^\circ$ ). Consequently, the COR and therefore the moment arm would be not calculated at  $0^\circ$  but at a more plantarflexed angle.

## Summary

Previous studies have demonstrated that  $MA_{AT}$  obtained using the COR method increases as a function of ankle angle from dorsiflexion to plantarflexion as well as a function of muscular contraction. The length of  $MA_{AT}$  is similar in all studies using this method. However, when comparing the MA length to results using the TE method, findings are greater in almost all cases. Also, when comparing the  $MA_{AT}$  length of this 2D to a recently published study using 3D MR imaging (Hashizume et al., 2011), 3D  $MA_{AT}$  results were significantly smaller and more in line with results of the TE method (see Table 2.1).

## 2.4 Anatomical considerations

### 2.4.1 Achilles tendon moment arm changes as a function of ankle angle

Previous research has shown that  $MA_{AT}$  changes as a function of ankle angle (Fukunaga, Roy, Shellock, Hodgson, & Edgerton, 1996; Grieve et al., 1978; Klein, 1996; Maganaris et al., 1998a, 2000). More specifically,  $MA_{AT}$  was found to gradually increase as the ankle joint is rotated from a dorsi- to a plantarflexed position. The only exception is an *in vitro* study by Spoor et al. (1990) who reported first an increase in  $MA_{AT}$  until  $0^\circ$  of ankle angle (sole of the foot is perpendicular to the tibia), following a decrease. This might be due to participant specific property of the ankle joint ligaments (Maganaris et al., 2000). When comparing  $MA_{AT}$  results obtained using the TE and the COR methods from Maganaris

et al. (2000), the results indicate that  $MA_{AT}$  changes in a similar way as a function of ankle angles. However, the researcher did not statistically test this assumption.

On an anatomical level, the increase in  $MA_{AT}$  from dorsi- to plantarflexion can be explained by the shape of the calcaneus bone. During a plantarflexion rotation of the ankle joint, the calcaneus bone shifts distally with respect to the COR of the ankle joint. As the Achilles tendon inserts onto the distal end of the calcaneus, the distance between the line of force and the COR of the ankle joint increases during plantarflexion. Therefore, the  $MA_{AT}$  increases during plantarflexion. On a physiological and biomechanical level, the calculation at the end-ranges of the range of motion for the TE method is problematic because of the hysteresis effect of the tendon during the loading and unloading phase of the tendon (Kay & Blazevich, 2008). As a consequence during the loading of the tendon, a reduction in tendon displacement can be found for a given joint rotation which results in shorter  $MA_{AT}$  as reported during dorsiflexion (Rugg et al., 1990).

#### **2.4.2 Achilles tendon moment arm changes as a function of contraction state**

Maganaris et al. (1998a, 2000) were the first researchers to explore the effect of contraction state (MVC) on the length of  $MA_{AT}$  estimates. As a result of the maximal isometric contraction, they found that  $MA_{AT}$  length increased significantly from rest to MVC (22% - 27%) across the range of motion. According to the authors, an explanation for the increase in  $MA_{AT}$  length can be the significant increase in muscle thickness during a MVC (Maganaris et al., 1998c; Maganaris & Paul, 1999; Maganaris, 2004). As the contractile elements of the muscle fibres are shortened during muscular contraction based on cross-bridge theory (Herzog et al., 1991; Huxley, 1957), a direct result of the shortening is an increase in the pennation angle of the muscle fibres (Buchanan et al., 2004; Maganaris & Paul, 1999; Narici et al., 1996; Zajac, 1989) and consequently an increase in the thickness of the muscle. For the  $MA_{AT}$  calculation using the COR method, the Achilles tendon is considered as the line of force which transfers the net plantarflexor forces to the ankle joint (i.e. calcaneus). Therefore, any deviation in the orientation of the line of force for a given COR of the joint results in an increase or decrease of the respective  $MA_{AT}$  length. As the Achilles tendon is in series with the plantarflexor muscles, changes in the pennation angle of the muscles and therefore in muscle thickness affects directly the orientation of the Achilles tendon. The increase in pennation angle during muscular contraction causes the Achilles tendon to deviate distally with respect to the tibia. As the distance between the COR of the ankle joint and the line of force (i.e. the moment arm) is increased, the  $MA_{AT}$  is therefore increased (Maganaris et al., 1998a).

In contrast to the reported increase in  $MA_{AT}$ , Baxter et al. (2011) did not find a difference between rest and MVC estimates of  $MA_{AT}$ . Besides using the same experimental setup as Maganaris et al. (1998a), the authors explained their findings by a different

strapping technique for the the ankle joint as well as the use of co-contraction during the plantarflexion contraction. In addition, Maganaris et al. (2000) did not find a significant difference of  $MA_{AT}$  between rest and contraction state (MVC) for the TE method. Using MR imaging, the displacement of the muscle-tendon unit was quantified as a change in the orientation from the tendon insertion point on the calcaneus by superimposing the MR images. According to Maganaris et al. (2000), the reason for the nonsignificant difference was that the ratio and therefore the magnitude of the  $MA_{AT}$  remained constant, although  $\Delta L$  and  $\Delta\theta$  changed throughout the muscular contraction. Besides the aforementioned study, other researchers applied the TE method during submaximal or maximal muscular contractions (Fukunaga, Ito, et al., 1996; Ito et al., 2000; Lee & Piazza, 2009). Lee and Piazza (2009) used a maximal muscle contraction in order to reduce the influence of tendon tension while manually rotating the ankle joint through its range of motion (Lee & Piazza, 2009, p. 3702). However, it has been shown that a preconditioning of the tendon before the testing increases the reliability of muscle-tendon measurements (Hawkins, Lum, Gaydos, & Dunning, 2009; Maganaris & Paul, 2000; Schatzmann, Brunner, & Stäubli, 1998). Following these recommendations, a muscular contraction during ankle rotation as described by Lee and Piazza (2009) for  $MA_{AT}$  measurements is not needed. Besides these experimental approaches to reduce the creep of the tendon, the use of the TE method during maximal contractions has to be taken with care as the method is based on the principle of virtual work (An et al., 1984; Storace & Wolf, 1979). The principle of virtual work assumes that if an external torque is applied to a joint, the sum of all muscle and tendon forces must be equivalent (see equation 2.4). Within this context, the ligaments are assumed to be inextensible and bony contact points frictionless. This assumption, however, ignores the energy storage capacity as well as the viscoelastic nature of the tendon and soft tissue during the contraction (Fukashiro, Hay, & Nagano, 2006) which violates the virtual work principle. This violation is likely to be more significant when larger muscles are involved as during a muscular contraction.

## 2.5 Overall summary

Accurate estimates of  $MA_{AT}$  play an important role when determining muscular forces from joint torques or vice versa. Although the TE and the COR methods have been frequently used, recent work suggests that  $MA_{AT}$  length seems to be dependent on the method used. For the TE method, different polynomial fittings have been used to model the ratio of tendon and angular displacement ( $\Delta L/\Delta\theta$ ). However, it remains unknown what effect the use of the different methods and polynomials have on the length and reliability of the  $MA_{AT}$  estimates. It is possible, that although the results of both methods differ in length, their relationship across ankle angles might be similar and therefore

comparable even during muscular contraction.

## 2.6 Thesis aims and research hypotheses

In view of the present literature, the aim of the present thesis was to perform a direct comparison between the TE and the COR methods *in vivo* and to document the effects on the resulting  $MA_{AT}$ . As moment arms have direct implications for musculoskeletal modelling, the effect of  $MA_{AT}$  obtained using the two methods and anatomical modifiers (angle- and contraction-state-dependence) should be determined during submaximal cycling. Three empirical studies were completed to answer the following aims and research hypothesis.

### Study 1: Aims

1. To directly compare *in vivo*  $MA_{AT}$ s obtained by the tendon excursion and the centre of rotation methods using ultrasound and MR imaging, respectively.
2. To test the reliability of  $MA_{AT}$  measurements for the tendon excursion method when using different polynomial fitting for the  $MA_{AT}$  calculation.
3. To test the influence of movement direction of the ankle on the reliability of  $MA_{AT}$ s measures derived from the TE method.

### Study 1: Research hypothesis

$H_0$ : *In vivo*  $MA_{AT}$  obtained using the tendon excursion method will be significantly smaller than the  $MA_{AT}$  obtained using the centre of rotation method.

$H_0$ : Larger intervals or higher polynomial fittings for the ratio of tendon and angular displacement data will result in more reliable measurements of  $MA_{AT}$ .

$H_0$ : Passive dorsiflexion in comparison to plantarflexion rotations of the ankle will result in a higher reliability of  $MA_{AT}$  measures.

### Study 2: Aims

1. To directly compare  $MA_{AT}$  estimated using the TE method at rest and the COR method during MVC.
2. To test for an interaction effect between moment arm methods, joint angle and contraction state.

### Study 2: Research hypothesis

$H_0$ : *In vivo*  $MA_{AT}$  obtained using the tendon excursion method at rest will be highly correlated to those obtained from the centre of rotation method during a maximal voluntary contraction.

$H_0$ : There will be no significant interaction effect on  $MA_{AT}$  between moment arm method, joint angle and contraction state.

### **Study 3: Aims**

1. To determine the influence of moment arm methods on the Achilles tendon force during submaximal cycling using an inverse dynamic.
2. To determine the influence of  $MA_{AT}$  which changes as a function of ankle angle on the Achilles tendon force during submaximal cycling.
3. To determine the influence of  $MA_{AT}$  which changes as a function of ankle angle and contraction state on the Achilles tendon force during submaximal cycling.
4. To determine the influence of  $MA_{AT}$  which changes as a function of ankle angle, knee angle and contraction state on the Achilles tendon force during submaximal cycling.

### **Study 3: Research hypothesis**

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  obtained using the tendon excursion and centre of rotation methods during submaximal cycling.

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  which changes as a function of ankle angle during submaximal cycling.

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  which changes as a function of ankle angle and contraction state during submaximal cycling.

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  that changes as a function of ankle angle, knee angle and contraction state during submaximal cycling.

# Chapter 3

# 3 Direct comparison of *in vivo* Achilles tendon moment arms obtained from tendon excursion and centre of rotation methods

## 3.1 Introduction

Quantifying human muscle forces *in vivo* is difficult because direct measurements are highly invasive (Finni et al., 1998; Fukashiro et al., 1993). Therefore, muscle forces are typically estimated indirectly using biomechanical modelling techniques. One method used to estimate muscle forces is to calculate the ratio of the muscular torque about a joint and the moment arm of the muscle or tendon of interest. A number of techniques exist to obtain muscle moment arms, including cadaver dissection (Grieve et al., 1978; Hintermann et al., 1994; Klein, 1996; Spoor et al., 1990), magnetic resonance (MR) imaging (Maganaris et al., 1998a; Rugg et al., 1990) and ultrasound imaging techniques (Ito et al., 2000; Lee et al., 2008; Lee & Piazza, 2009; Maganaris, 2003a).

Two of these methods in particular have become popular within the scientific community. The first technique uses sagittal plane 2D MR images to estimate the centre of rotation of a joint (COR) (Reuleaux, 1875; Rugg et al., 1990). The perpendicular distance between the centre of the joint to the line of action of the muscle or tendon of interest is then measured directly (Maganaris et al., 1998a, 2006; Rugg et al., 1990). The second technique is the tendon excursion (TE) method. Using the principle of virtual work (An et al., 1984; Spoor et al., 1990), the moment arm is computed as the ratio between the linear displacement of the tendon and the angular excursion of the corresponding joint (Ito et al., 2000; Maganaris et al., 2000; Spoor et al., 1990). Thus, it does not require knowledge of the location of the COR.

Both, the COR and the TE methods have advantages and disadvantages. One limitation of the COR method is that the multiple steps of manual MR image processing required to determine the COR can introduce errors in the COR calculation and therefore in the moment arm estimation (Maganaris et al., 1998a). Another disadvantage of the COR method is the limited accessibility and relatively high costs involved in using MR scanners. A ma-

major advantage of MR imaging, however, is the high visibility of the underlying anatomical structures about the joint. In particular, the line of force can be easily identified. The major advantage of the TE method is that it does not require knowledge of the COR or the line of action of muscle or tendon force. In addition, ultrasonography can be used for the TE method, which is often easier to access and more time and cost efficient compared to MR scanning. The main limitation of the tendon excursion method is the assumption that no energy is lost as ligaments are assumed to be inextensible and bony contact points frictionless (principle of virtual work) (An et al., 1984; Storace & Wolf, 1979). Thus, it is assumed that both active and passive forces within the joint are negligible and that all muscles spanning the joint of interest are inactive. As muscles and tendons store, release and dissipate elastic energy during muscle contractions, the principle of virtual work is violated.

Lee and Piazza (2009) used the TE method to determine in vivo values of Achilles tendon moment arms  $MA_{AT}$ . Their  $MA_{AT}$ s were smaller than those reported by Maganaris et al. (1998a) who used the COR method. Whilst these differences can have multiple explanations (including the use of participants with different anthropometric characteristics), these findings raise the question as to whether  $MA_{AT}$  obtained using these two methods are comparable. However, to our knowledge a direct comparison between these two methods has not been made. Therefore, differences in the moment arms reported in the literature cannot incontrovertibly be attributed to methodological differences. The first aim of this study was to compare  $MA_{AT}$  measures using the COR method (MR imaging) and the TE method (ultrasound imaging).

When evaluating scientific measurement techniques, the reliability of the dependent measure is another important consideration. Coefficients of variation (CVs) of  $MA_{AT}$ s obtained using the COR method have been reported to be 7.9% (Maganaris et al., 1998a, 2000). The mean between-day difference in moment arms obtained from the TE method has been reported to be approximately 5% (Lee & Piazza, 2009). Together, these results suggest that moment arms obtained using the TE method are potentially more reliable than those obtained using the COR method. Therefore, the second aim of this study was to compare the reliabilities of the moment arm measures obtained from the COR and the TE methods.

When using the TE method, the moment arm is obtained by mathematically differentiating the tendon displacement with respect to the corresponding joint angle. Two methods have been used to perform this differentiation. The first method is to fit a straight line between two tendon displacement values over a given angular displacement (Grieve et al., 1978; Ito et al., 2000; Maganaris, 2003a; Maganaris et al., 1998a). The second method is to approximate the relationship between linear tendon and angular joint displacements by means of a  $2^{nd}$ - or  $3^{rd}$ -order polynomials and perform an analytical differentiation (Lee &

Piazza, 2009; Murray et al., 1995). Within the context of the present investigation, it was sought to understand how different differentiation methods would influence the relationship between the moment arms obtained using the COR and the TE methods and affect the reliability. Therefore, we used seven differentiation techniques to estimate the  $MA_{AT}$  using the TE method.

Finally, there is some evidence that  $MA_{AT}$  obtained from the TE method might differ depending on the movement direction of the passive rotation during which tendon excursion and angular displacement are measured (Spoor et al., 1990). It has been speculated that the presence of greater active or passive forces during loading of the tendon (dorsiflexion rotations) compared to unloading (plantarflexion rotations) could contribute to these differences. Understanding this potential discrepancy is important for two reasons. First, the presence of active or passive forces violates the principle of virtual work and could affect the accuracy of the moment arm estimate. Second, differences in the magnitude of forces during loading and unloading could result in differences in the reliability of moment arm measures obtained during different movement directions. Therefore, the third aim of this study was to quantify the reliabilities of  $MA_{AT}$ s derived from the TE method for two rotational directions (dorsiflexion vs. plantarflexion rotations) and their relationship with the  $MA_{AT}$ s obtained from the COR method. To explain potential effects of movement direction on  $MA_{AT}$ s, the torques about the ankle joint arising during passive ankle rotation were quantified (since the torque is the direct effect of all passive and active forces) and compared it across rotational positions and movement directions.

## 3.2 Methods

### 3.2.1 Participants

To compare  $MA_{AT}$ s derived from the COR and the TE methods (Aim 1), nine healthy adults (7 male, 2 female) volunteered to participate in this experiment. Mean values and standard deviations for age, stature and body mass were  $31 (\pm 5)$  years,  $180 (\pm 10)$  cm and  $81 (\pm 15)$  kg, respectively. Seven out of the nine participants (age =  $30 \pm 5$  years, stature =  $183 \pm 5$  cm, body mass =  $85 \pm 13$  kg) were tested three times for both the COR and TE methods to quantify the reliability of each method (Aim 2). All participants were physically active on a recreational basis and reported no recent lower limb musculoskeletal injuries. The study was approved by the Human Research Ethics Committee of the School of Sport and Education at Brunel University. The participants provided written informed consent, and they were made aware of their right to withdraw from the study at any time without penalty.

### 3.2.2 Experimental protocol: MR imaging

Participants were asked to lie supine on a wooden board (183 x 49 cm) placed inside a 3-Tesla MR scanner (Siemens Magnetom Trio syngo MR 2004A, Erlangen, Germany) with their knee straight and the sole of their foot positioned against a custom-made wooden block, which was attached to the board by six pins and two inelastic Velcro straps. This wooden block determined the angular position of the ankle joint. Alterations of the ankle position were achieved by attaching blocks of different shapes to the wooden board. MR images (sagittal scans, TR = 600 ms, TE = 12 ms, 3 excitations, 300 mm field of view, 2 mm slice thickness) were taken at three ankle positions (expressed as the absolute foot angle): 0° (neutral ankle position; the foot being in a vertical position), +15° and -15° (plantar- and dorsiflexed position, respectively). Two inelastic Velcro straps were used to secure the foot to the wooden block in its transverse plane. One additional Velcro strap was secured around the distal part of the thigh.

Participants were instructed to remain relaxed whilst in the MR scanner. Before the testing, to pre-condition the muscle-tendon complex, participants were asked to perform three ramped isometric contractions to up to 50% of their perceived maximum at the neutral ankle angle as well as three maximal voluntary contractions (Maganaris & Paul, 1999). All participants had been familiarised with the testing devices as well as the testing protocol one day prior to their testing at the laboratories at Brunel University. During this familiarisation, participants were introduced to perform maximal isometric contractions as well as ramped contractions on an isokinetic dynamometer and were asked to replicate those on the MRI testing device. For the testing, sagittal localiser pre-scans were used at the level of the malleoli to ensure that the orientation of the talus and the Achilles tendon were similar throughout the three ankle angles of interest (Maganaris et al., 1998a). The duration of each scan was 147 s, and 25 slices were obtained. After each scan, the next wooden ankle block was inserted for the scanning of the foot at a different ankle angle. The order in which the different blocks were presented to the participants was from a plantar- to a dorsiflexed ankle position. Seven out of the nine participants were scanned three times at each angle. Once they were scanned at each of the three angles, [1] the participants were removed from the scanner, [2] the wooden blocks were detached from the board, [3] the board was taken out of the scanner, [4] the setup was re-assembled, and [5] the participant was re-tested. This procedure was repeated two times (resulting in three independent scans at each angle) to determine the inter-experiment reliability of the scanning method (Aim 2).

### Moment arm calculation using the centre of rotation method

MR images were processed using the iPACS viewer (version 4.230, San Bruno, CA, USA). For every participant, the COR of the ankle joint, the line of action of the force (rep-

resented by the orientation of the Achilles tendon) and the  $MA_{AT}$  was determined for the neutral foot position (ankle angle of  $0^\circ$ ). A detailed description of this technique has been published previously (Maganaris et al., 1998a). Briefly, using Reuleaux' method (Reuleaux, 1875), the COR at an ankle angle of  $0^\circ$  was assessed using the MR scans taken at ankle angles of  $-15^\circ$  and  $+15^\circ$  (see Figure 2.1).

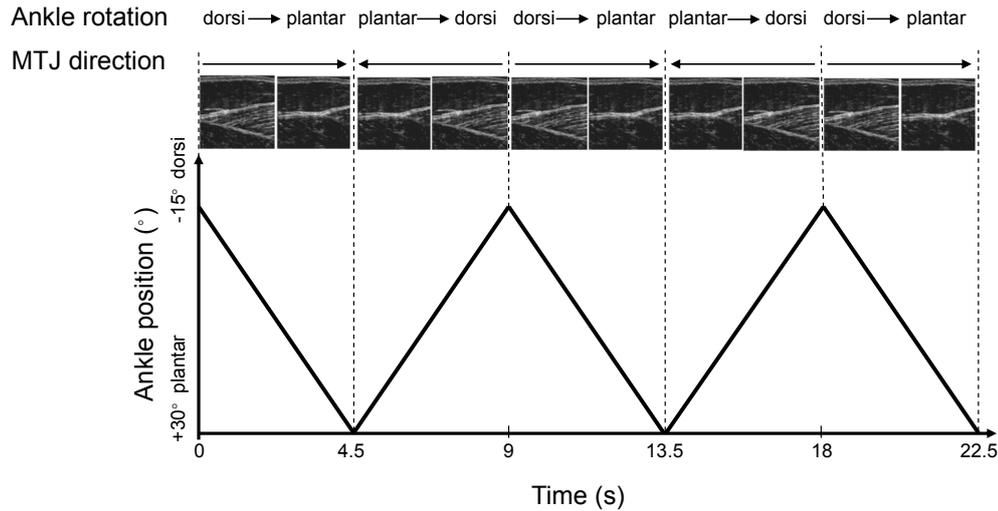
The tibia was assumed to be fixed and the talus was assumed to be the rotating segment. The outline of the tibia and talus were drawn onto an overhead transparency with the foot in the dorsiflexed position (ankle angle of  $-15^\circ$ ). Two well defined anatomical points on the talus (A, B) were chosen. Based on the small distance between the two points, they were extended by drawing two straight lines subtending at a right angle 10 cm proximal to the talus (Maganaris, 2004; Maganaris et al., 1998a). The shape of the tibia was then superimposed onto the image taken at the plantarflexed position (ankle angle of  $+15^\circ$ ) and the shape of the talus was drawn onto the same transparency. The same two anatomical points of the talus (A' and B') in this second position were marked on the initial transparency. Again, they were extended and drawn 10 cm proximal to the talus using two straight lines subtending at a right angle. The COR was geometrically constructed as the intersection of the perpendicular bisector of lines AA' and BB' as illustrated in Figure 2.1. The line of force was defined as the line connecting the point of insertion of the Achilles tendon into the calcaneus and the point 6 cm more proximal on the midline of the Achilles tendon. For the definition of this line, the MR image obtained at the neutral foot position (ankle angle of  $0^\circ$ ) was used. The  $MA_{AT}$  was then measured as the perpendicular distance from the line of force to the COR.

### **Data processing**

All morphometric measurements were analysed three times by the same investigator for each MR image, and mean values across the three moment arm measures were taken for further analysis. The mean CV across all nine participants for analysing an image three times was  $3.3 \pm 1.5\%$  ( $\pm$  SD). To quantify the reliability of  $MA_{AT}$ s derived from the COR methods between inter-experiment setups (Aim 2), the CVs were calculated using the  $MA_{AT}$  values obtained from repeated testing as described in the previous section ( $n = 7$ ).

### **3.2.3 Experimental protocol: Ultrasonography**

Ultrasound imaging was conducted on a separate day to the MR imaging, but at the same time of day. Participants were seated on an isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, Inc., NY) with their right ankle securely fixed with Velcro straps to a footplate to prevent any heel movement. The knee was fully extended and the relative hip angle was set to  $85^\circ$ . The centre of the lateral malleolus was aligned with



*Figure 3.1.* Schematic illustration of the relationship between ankle rotation and movement of the muscle-tendon junction (MTJ) of the gastrocnemius medialis and the Achilles tendon. The ankle was rotated from a dorsiflexed position of a minimum of  $-15^\circ$  to a plantarflexed position of a minimum of  $+30^\circ$  (plantar) and vice versa (dorsi) at  $10^\circ \cdot s^{-1}$  (five consecutive rotations resulting in three plantar- and two dorsiflexion rotations). The MTJ shortened or lengthened according to the ankle rotation.

the rotation axis of the dynamometer. The thigh was securely fixed to the dynamometer seat with inelastic straps to prevent any medial or lateral changes in the orientation of the ankle alignment. To pre-condition the muscle-tendon complex, participants performed five ramped isometric plantarflexions to up to 50% of their perceived maximal, as well as three maximal isometric voluntary contractions (Maganaris & Paul, 1999). The maximal range of motion of the ankle joint was then determined for each participant by manually rotating the dynamometer until the participant felt discomfort; for all participants, the limits of the range of motion were greater than  $-15^\circ$  and  $30^\circ$  (dorsi- and plantarflexed position, respectively).

Before the testing, the ankle was passively rotated through its range of motion at a constant velocity of  $10^\circ \cdot s^{-1}$  to familiarise the participant with the rotation. For the actual test, the ankle was rotated passively through its range of motion five consecutive times: three times from a dorsi- to plantarflexed position and two times from plantar- to a dorsiflexed position (see Figure 3.1).

The mechanical concept underlying the TE method is the principle of virtual work (An et al., 1984; Storace & Wolf, 1979). To minimise internal forces, we asked our participants to relax their muscles throughout the range of motion. During the passive rotations, the displacement of the muscle-tendon junction (MTJ) was recorded using B-mode ultrasound with a 10 MHz, 40 mm linear probe (Esaote Megas GPX, Genova, Italy). The probe was fixed in a custom-built foam cast, which was aligned smoothly with the skin. A water-

based transmission gel was placed between the skin surface and the ultrasound probe to aid acoustic coupling and to avoid skin deformation. Care was taken to align the probe with the movement direction of the MTJ in the sagittal plane. The ultrasound probe was taped to the shank with zinc oxide tape to prevent probe movement.

The same subset of participants that was tested for the reliability of the COR method was tested for the reliability of the TE method. For these participants, the probe was removed and reattached. Before reattaching the ultrasound probe, any marks left by the gel and the probe were removed to avoid bias in the replacement of the probe. This change in setup was repeated two times for the purpose of quantifying the reliability of the TE method between three experimental setups.

### **Data processing**

All data were collected using Cortex 1.1.4 software (MotionAnalysis, Santa Rosa, CA) onto a personal computer. Raw position and torque data from the isokinetic dynamometer were analogue-digital converted at 1000 Hz using a 12 bit A/D card (NI PCI-6071E, National Instruments, Austin, TX). Analogue ultrasound video data, sampled at 25 Hz, were digitally converted with a digital video recorder (ADVC55, Grass Valley, Conflans St. Honorine, France) connected via the SVHS port and saved onto a personal computer as an uncompressed AVI (720 x 576) file. Ultrasound, position and torque data were synchronised using an electrical trigger (Stimulator DS7A, Digitimer Ltd., Hertfordshire, UK).

The MTJ of the gastrocnemius medialis was manually digitised using specialized imaging software (ImageJ, 1.42q, National Institutes of Health, USA) for all three probe placements and five full rotations (three plantar- and two dorsiflexion rotations). Raw tendon excursion, raw angular position and raw torque data were processed in Matlab (version 7.4, The MathWorks, Inc., Natick, MA). The raw angular position and torque data were filtered using a low-pass digital 4<sup>th</sup>-order, zero-lag Butterworth filter with cut-off frequencies of 3.75 Hz for angular position and 12.47 Hz for torque, respectively. The filtered position data were downsampled to 25 Hz to match the ultrasound data. Pixel coordinates of the raw tendon excursion data were converted to millimeters with a conversion factor of 9.2 pixels per millimeter and filtered with a digital low-pass, fourth-order zero-lag Butterworth filter with a cut-off frequency of 2.62 Hz. The cut-off frequencies for the low-pass filters were determined by means of residual analyses (Winter, 2005) (see Appendix ??).

#### **3.2.4 Moment arm calculation using the tendon excursion method and determination of torques about the ankle joint**

The  $MA_{AT}$  at an ankle angle of 0° was calculated as the first derivative of tendon displacement with respect to ankle angle (Figure 3.2). It was obtained using seven different

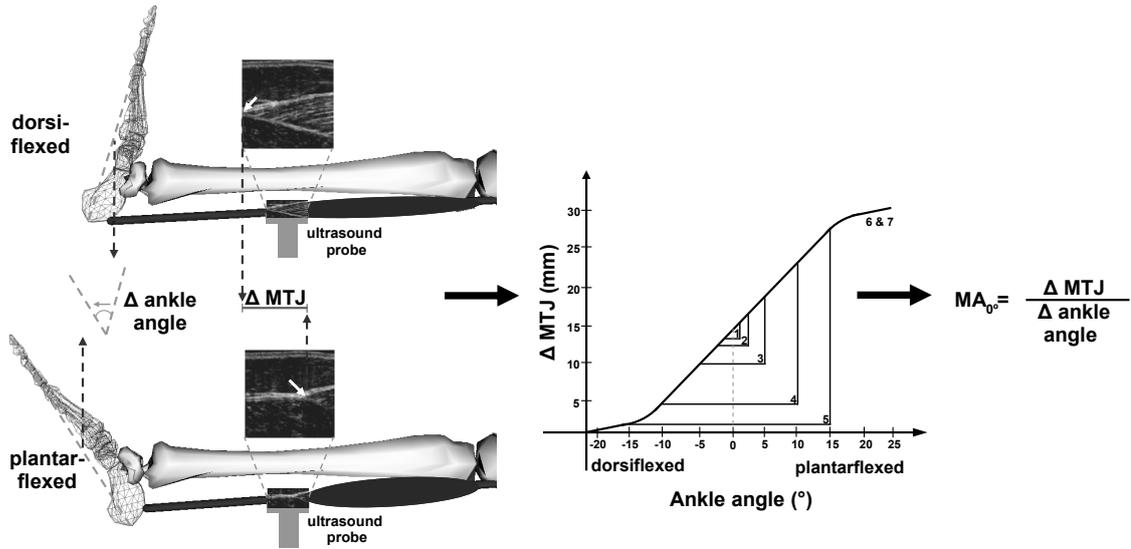


Figure 3.2. Schematic illustration of the tendon excursion method. Achilles tendon moment arm was calculated as the first derivative of the ratio of the change in muscle-tendon length represented by the displacement of the muscle-tendon junction (MTJ) of the gastrocnemius medialis and the Achilles tendon ( $\Delta$  MTJ) to the change in ankle angle ( $\Delta$  angle). The derivative was calculated for  $0^\circ$  (ankle is perpendicular to the tibia) over five linear intervals ( $\pm 1^\circ$ ,  $\pm 2^\circ$ ,  $\pm 5^\circ$ ,  $\pm 10^\circ$ ,  $\pm 15^\circ$  numbered as 1 to 5, respectively, in the figure). In addition, a second- and third-order polynomial was fitted to the ratio of the change in muscle-tendon length to the change in angle and differentiated at  $0^\circ$  (numbered 6 and 7).

differentiation methods: The slope of the lines connecting the tendon displacement data over five different angular intervals ( $\pm 1^\circ$ ,  $\pm 2^\circ$ ,  $\pm 5^\circ$ ,  $\pm 10^\circ$ ,  $\pm 15^\circ$ ) was calculated. Due to the relatively low resolution of the tendon displacement data, they were spline fitted so that the exact values at the ankle angles of interest could be found. In addition, second- and third-order polynomials were fitted to the tendon displacement data. These polynomials were then analytically differentiated at an ankle angle of  $0^\circ$ . These seven different differentiation techniques were performed for each of the three passive plantar- and the two passive dorsiflexion rotations.  $MA_{ATS}$  were taken as the average across the plantar- and dorsiflexion rotations, respectively, resulting in 14 different conditions (seven differentiation methods two movement directions). These torques measured during the passive rotations were reported for ankle angles of  $+25^\circ$ ,  $+15^\circ$ ,  $+5^\circ$  (plantarflexed position) and  $-5^\circ$ ,  $-15^\circ$  (dorsiflexed position) for both movement directions and averaged across rotations and participants.

### 3.2.5 Statistical analysis

To determine the absolute difference between the  $MA_{ATS}$  obtained from the two methods, two-tailed paired  $t$ -tests were performed (Bonferroni corrected). To determine the rela-

relationship between the  $MA_{AT}$ s obtained using the COR and the TE methods, 14 Pearson's product moment correlations and CVs were calculated. These represented the relationship between the  $MA_{AT}$ s obtained using the COR method and those obtained using each of the 14 variations of the TE method. To quantify the reliability of the  $MA_{AT}$  measures between experimental setups, the CV for each participant for each method of determining the moment arm was calculated. These CVs were then averaged across participants within each condition. To compare the torques between movement directions during the TE method, a repeated measures MANOVA was performed. Dependent measures were the torques measured about the ankle joint at the aforementioned ankle angles ( $+25^\circ$ ,  $+15^\circ$ ,  $+5^\circ$ ,  $-5^\circ$ ,  $-15^\circ$ ). Post hoc paired  $t$ -tests (Bonferroni corrected) were then performed at each angle. All analyses were performed using SPSS statistical software (v.15.0; LEAD Technologies Inc., USA). Statistical significance was accepted at  $p < .05$ .

### 3.3 Results

Moment arms obtained using the TE method were significantly smaller for all differentiation methods and both rotational directions when compared to those obtained from the COR method ( $p < .01$ , Table 3.1 and Figure 3.3).

The mean percentage differences across all participants and differentiation methods were  $29.3 \pm 5.7\%$  for plantarflexion rotations and  $26.2 \pm 6.0\%$  for dorsiflexion rotations, respectively (mean  $\pm$  SD). In spite of the absolute differences between methods, the  $MA_{AT}$ s obtained using the COR method correlated well with those obtained using the TE method (Table 3.1). The strength of this relationship was dependent upon the mathematical differentiation technique and the movement direction of the ankle used for the TE method. For the passive plantarflexion rotations, the relationship between  $MA_{AT}$ s obtained from the TE and the COR methods were stronger than for the passive dorsiflexion rotations ( $.64 \leq R^2 \leq .94$  and  $.40 \leq R^2 \leq .60$  for plantarflexion and dorsiflexion rotations, respectively; see Table 3.1).

With the exception of two conditions ( $\pm 5^\circ$ , 3<sup>rd</sup>-order polynomial during the dorsiflexion rotation) all correlations were statistically significant ( $p < .05$ ) (Table 3.1). In general, the use of polynomial differentiation and linear differentiation over larger intervals ( $\pm 15^\circ$ ,  $\pm 10^\circ$ ) resulted in stronger correlations between both methods. The strongest relationship between the two methods was found when tendon displacement was differentiated over an angular interval of  $\pm 10^\circ$  ( $R^2 = .94$ ) during plantarflexion rotations (Figure 3.4).

The inter-experiment CV for the  $MA_{AT}$ s obtained using the COR method was 3.9%. For the  $MA_{AT}$ s obtained using the TE method, inter-experiment CVs ranged from 4.5% to 9.7%. These CVs were dependent upon the differentiation technique and the direction of the passive foot rotation. Passive plantarflexion rotations tended to produce  $MA_{AT}$ s more

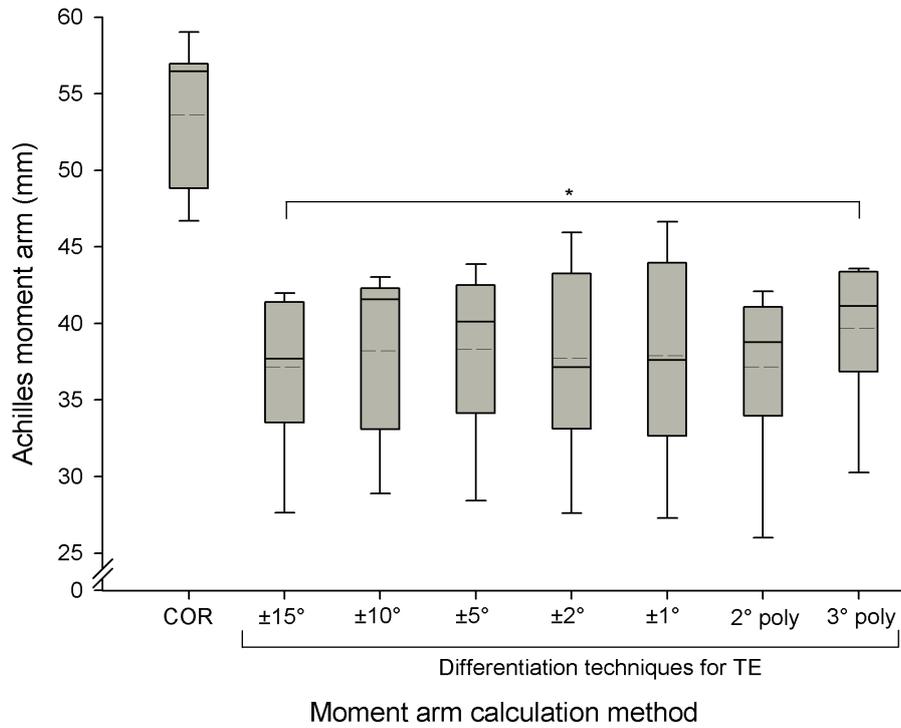


Figure 3.3. Comparison of Achilles tendon moment arm results between the center of rotation (COR; MR imaging) and the tendon excursion (TE; ultrasound imaging) methods as the mean of three consecutive plantarflexion rotations for the neutral ankle angle. The moment arm calculated using the COR method was determined over a  $\pm 15^\circ$  angular interval using a geometrical approach. The moment arm for the TE method was calculated over five different angular differentiation intervals ( $\pm 15^\circ$ ,  $\pm 10^\circ$ ,  $\pm 5^\circ$ ,  $\pm 2^\circ$ ,  $\pm 1^\circ$ ) and two polynomial fitting procedures ( $2^{nd}$ - and  $3^{rd}$ -order polynomials) as the ratio of tendon to ankle displacement. COR: centre of rotation method;  $\pm 15^\circ$ ,  $\pm 10^\circ$ ,  $\pm 5^\circ$ ,  $\pm 2^\circ$ ,  $\pm 1^\circ$ : angular intervals for differentiation;  $2^\circ$ poly:  $2^{nd}$ -order polynomial fit;  $3^\circ$ poly:  $3^{rd}$ -order polynomial fit. \* indicates a significant difference between the COR and the TE methods ( $p < .01$ )

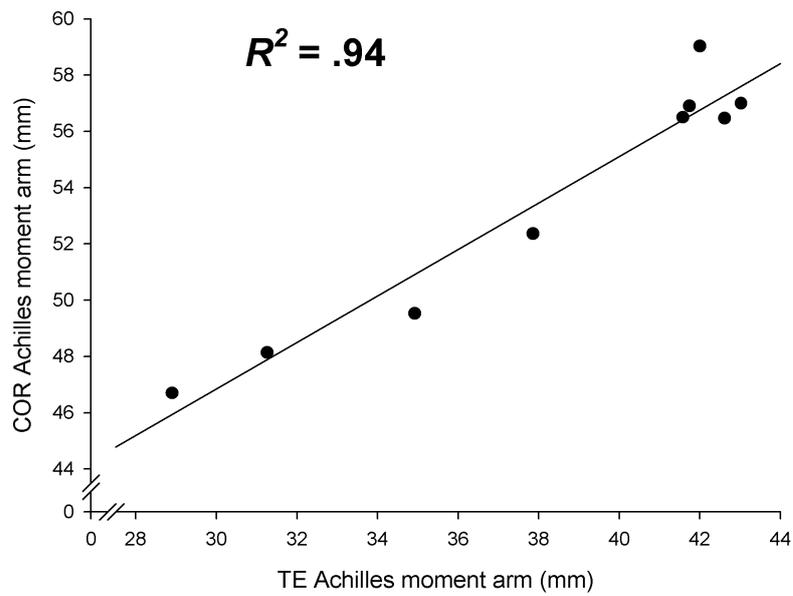
Table 3.1. Pearson's product moment correlations between Achilles tendon moment arm obtained from COR and TE methods at a neutral ankle angle, as well as CV for interexperimental set-ups

Differentiation techniques	Moment arm (mm) (mean $\pm$ SD)	$t(8)$	r	$R^2$	CV(%)
<i>Plantarflexion rotations</i>					
Angular interval					
$\pm 15^\circ$	37.2 $\pm$ 4.9	25.6**	.92*	.85	4.9
$\pm 10^\circ$	38.2 $\pm$ 5.3	32.0**	.97*	.94	5.1
$\pm 5^\circ$	38.3 $\pm$ 5.2	21.0**	.91*	.83	7.0
$\pm 2^\circ$	37.7 $\pm$ 6.1	12.9**	.80*	.64	8.5
$\pm 1^\circ$	37.9 $\pm$ 6.6	12.1**	.82*	.68	9.7
Polynomial fit					
2 <sup>nd</sup> order	37.2 $\pm$ 5.3	20.9**	.90*	.80	4.5
3 <sup>rd</sup> order	39.7 $\pm$ 4.5	24.4**	.93*	.86	4.8
<i>Dorsiflexion rotations</i>					
Angular interval					
$\pm 15^\circ$	37.6 $\pm$ 3.3	16.7**	.77*	.60	8.5
$\pm 10^\circ$	39.2 $\pm$ 4.1	12.6**	.69*	.47	7.8
$\pm 5^\circ$	40.3 $\pm$ 4.8	9.9**	.62	.39	7.5
$\pm 2^\circ$	40.9 $\pm$ 5.3	10.0**	.71*	.50	7.3
$\pm 1^\circ$	41.0 $\pm$ 5.5	10.1**	.73*	.53	9.0
Polynomial fit					
2 <sup>nd</sup> order	37.6 $\pm$ 4.8	13.6**	.71*	.50	7.0
3 <sup>rd</sup> order	40.5 $\pm$ 4.0	10.6**	.63	.40	8.0

Note: Values are mean  $\pm$  SD for moment arm; n = 9 for Pearson's product-moment correlations between Achilles tendon moment arm obtained from centre of rotation (COR; magnetic resonance imaging) and tendon excursion (TE; ultrasound imaging) methods, n = 7 for coefficient of variation (CV) for 3 interexperiment set-ups.

\* indicates a significant correlation of Achilles tendon moment arm using the TE and the COR method ( $p < .05$ )

\*\* indicates a significantly smaller Achilles tendon moment arm using the TE method compared to the COR method ( $p < .01$ , two-tailed, Bonferroni corrected)



*Figure 3.4.* Highest correlation between Achilles tendon moment arms calculated using the centre of rotation method (COR) using MR imaging and the tendon excursion method (TE) using ultrasonography. For the COR method, the moment arm was determined over a  $\pm 15^\circ$  angular interval using a geometrical approach. For the TE method, tendon displacement of the muscle-tendon junction of the gastrocnemius medialis and the Achilles tendon was differentiated over an angular interval of  $\pm 10^\circ$  (i.e.  $20^\circ$  range) for three consecutive plantarflexion rotations.

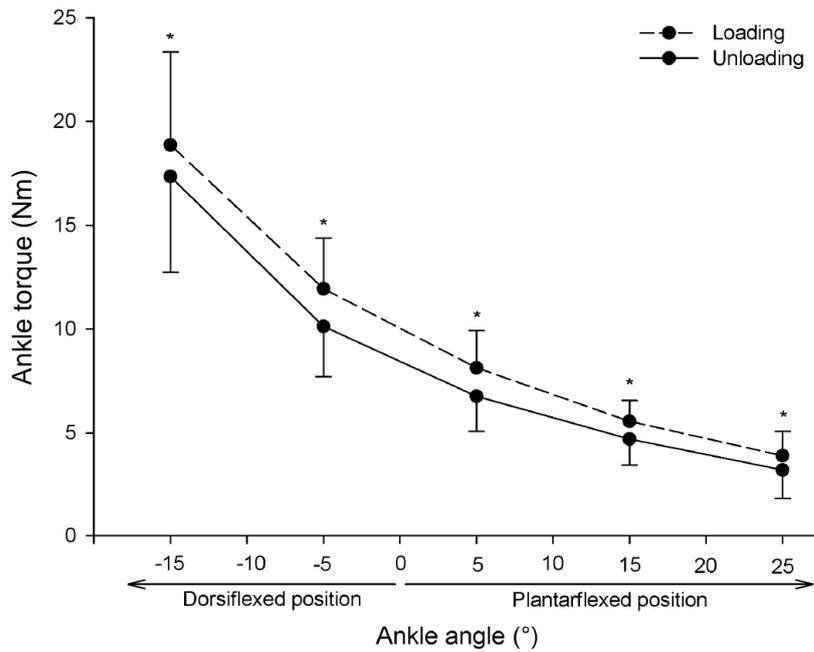


Figure 3.5. Loading and unloading curves of torques about the ankle joint measured during passive rotations. The ankle joint was rotated from a plantarflexed position of 30° to a dorsiflexed position of 15° three times at 10°·s<sup>-1</sup> and two times vice versa. \* indicates a significantly smaller torque about the ankle joint during plantar- compared to dorsiflexion rotations ( $p < .01$ , two-tailed, Bonferroni corrected).

reliably when compared to passive dorsiflexion rotations (Table 3.1). The smallest CV when using the TE method was found when a 2<sup>nd</sup>-order polynomial was fitted to the ratio of tendon and angular displacement data during plantarflexion and differentiated at 0° (CV = 4.5%). Furthermore, it was found that linear differentiation over the smaller angular intervals produced less reliable MA<sub>AT</sub>s when compared to polynomial differentiation or linear differentiation over larger intervals. The CV for the differentiation method with the strongest relationship between the two methods (linear differentiation over ±10° ankle angle during plantarflexion rotations) was 5.1% (Table 3.1).

The repeated measures MANOVA revealed a significant main effect of rotational direction on torques about the ankle joint (Wilks'  $\Delta = 0.036$ ,  $F(5, 4) = 21.36$ ,  $p < .05$ ). Post-hoc  $t$ -tests revealed that torques about the ankle joint derived during the plantarflexion rotations were significantly smaller than those during dorsiflexion rotations at all foot angles (see Figure 3.5).

### 3.4 Discussion

The first aim of this study was to compare  $MA_{AT}$  using the COR method (MR imaging) and the TE method (ultrasound imaging). Previous research suggests that the use of different methodologies of determining  $MA_{AT}$  may lead to differences in moment arm estimates. However, this speculation has not been confirmed conclusively. Our results demonstrate that the  $MA_{AT}$  magnitude depends on the method employed. Absolute  $MA_{AT}$  values obtained from the COR method were more than 25% greater than those obtained from the TE method. Possible explanations for these differences in  $MA_{AT}$ s lie in the assumptions of each method. When using the COR method, it is assumed that the ankle is a hinge joint and that the talus, as the rotating segment, moves in the sagittal plane. The fact that ankle plantar- and dorsiflexion rotations are typically accompanied by inversion or eversion at the subtalar joint (Donatelli, 1996; Isman & Inman, 1969) is therefore neglected. Any inversion or eversion could change the shape and orientation of the talus and could therefore cause errors in the determination of the two anatomical reference points needed to calculate the COR. Furthermore, the repeated need for the manual drawing of lines to geometrically construct the COR may result in an erroneous estimation of its position (Panjabi, 1979; Panjabi et al., 1982). The summation of these errors could lead to an overestimation of the  $MA_{AT}$ , which is a potential explanation for  $MA_{AT}$ s being greater when derived from the COR method compared to the TE method. When using the TE method, it is assumed that no elastic energy is stored or dissipated during the passive rotation (principle of virtual work) (An et al., 1984; Storace & Wolf, 1979). However, during dorsiflexion rotations significant increases in the torques measured about the ankle joint were found, suggesting the storage and release of elastic energy of the muscles and tendons which has an affect on the length of the muscle-tendon unit and consequently on the position of the MTJ. Such elastic energy could result in an underestimation of tendon displacement for a given joint angular displacement. A possible consequence is an underestimation of the  $MA_{AT}$ , which is a potential explanation for the smaller  $MA_{AT}$ s derived from the TE method when compared to the COR method.

Our observed dependence of  $MA_{AT}$ s on the method employed has vast implications regarding the development and interpretation of musculoskeletal models. First, knowledge of the method used to determine Achilles tendon moment arm is crucial to accurately interpret results derived from musculoskeletal models. Second, these results need to be taken into consideration when developing such models. When developing a musculoskeletal model, it is important to test the sensitivity of its outputs to variations in the estimation of its parameters. Ranges for such sensitivity analysis are typically equal to or less than 10% of the relevant estimated parameter (Redl, Gfoehler, & Pandy, 2007; Xiao & Higginson, 2010). Our results suggest that sensitivity analyses should cover a wide range (i.e., up to 30%) when a model's sensitivity to variations in  $MA_{AT}$  is evaluated.

Our findings raise the question as to which method is the most appropriate to use. Using Reuleaux' geometrical approach (Reuleaux, 1875), the determination of the COR coordinates are susceptible to errors due to the numerous manual processing steps (Panjabi, 1979; Panjabi et al., 1982). A recent study used motion analysis in combination with ultrasonography to estimate  $MA_{AT}$  with fewer processing steps and therefore less potential for errors (Manal et al., 2010). The authors assumed the COR to be the midpoint of the line connecting the medial and lateral malleoli and measured the distance between this point and the line of action of the force. The resulting  $MA_{AT}$ s were similar to those obtained from the TE method in the present investigation.  $MA_{AT}$ s obtained from cadavers (Spoor et al., 1990) are also more similar to our  $MA_{AT}$ s derived the TE method when compared to the COR method. Therefore, the  $MA_{AT}$ s obtained from the TE method are potentially closer to the true  $MA_{AT}$  of the Achilles tendon than those obtained from the COR method.

Despite the differences in absolute  $MA_{AT}$  values, an important finding of the present study was that there was a strong relationship between the  $MA_{AT}$ s obtained from the TE and the COR methods when measured at the neutral ankle angle. The strength of this relationship was dependent on the differentiation method used and the direction of the passive foot movement. Stronger relationships between the COR and the TE method were found when the foot was passively plantarflexed. A possible explanation for the dependence of the strength of the relationship on movement direction is the difference in torques about the ankle joint between plantar- and dorsiflexion rotations (Kay & Blazeovich, 2008). The greater torque observed during dorsi- compared to plantarflexion rotations could potentially result in a reduction in tendon elongation for a given ankle rotation, and therefore a less accurate  $MA_{AT}$  estimate and weaker relationship with the  $MA_{AT}$ s derived from the COR method. Furthermore, the  $MA_{AT}$ s derived from the TE method were more strongly correlated to those obtained from the COR method when linear differentiation was performed over larger angular intervals. When using polynomials to differentiate the tendon excursion data, the correlations were similar to those obtained from the linear differentiation over the large intervals. The result that both methods correlated well across participants is important as it implies that  $MA_{AT}$  differences and changes across participants (e.g. during growth) are independent of the method being used.

The second aim of the present study was to determine the reliability of repeated  $MA_{AT}$  measurements for both the COR and the TE methods. The mean CV describing the reliability of moment arms obtained from the COR method between experimental set-ups (3.9%) is similar to previously published values (Lee & Piazza, 2009; Maganaris et al., 2000) and is demonstrative of good reliability of the COR method. The CVs describing reliability between experimental setups for  $MA_{AT}$ s obtained from the TE method ranged from 4.5% to 9.7% for plantarflexion and 7.0% to 9.0% for dorsiflexion rotations, depending

on the differentiation method used. The magnitude of this variability is also comparable to that reported in previous studies (Fukunaga, Ito, et al., 1996; Ito et al., 2000; Lee & Piazza, 2009). Our findings extend previous results by demonstrating that the TE method is most reliable when using polynomial fitting to differentiate tendon displacement with respect to joint rotation. Furthermore,  $MA_{ATs}$  were more reliable when tendon displacement was differentiated over larger angular intervals ( $\pm 15^\circ$ ,  $\pm 10^\circ$ ) than over smaller intervals ( $\pm 1^\circ$ ,  $\pm 2^\circ$ ) during plantarflexion rotations. The likely reason for this finding is that minor deviations in tendon excursion resulting from errors in digitising the position of the MTJ are magnified when tendon length is differentiated over smaller angular intervals.

The third aim of the present study was to explore the influence of movement direction on the reliability of  $MA_{ATs}$  measures derived from the TE method. As a result, the variability of the  $MA_{ATs}$  between experimental set-ups was considerably greater during dorsiflexion than during plantarflexion rotations, independent of the differentiation method (Table 3.1). The higher CVs observed during dorsiflexion rotations can potentially be explained by the significantly higher torques about the ankle joint for this movement direction. These are indicative of active or passive forces, which violate the assumption of virtual work, and could thereby reduce the reliability of the  $MA_{AT}$  estimates. Our results have implications for researchers who wish to use the TE method to derive  $MA_{AT}$ ; based on both the smaller CVs and the stronger correlations with MRI-derived  $MA_{ATs}$ , using passive plantarflexion rotations and polynomial or large interval differentiation can be recommended. An additional advantage of using a polynomial fitting to differentiate tendon excursion with respect to ankle angle is that the moment arm can be determined over the recorded range of angles (although the validity and reliability of such a procedure was not subject of this study).

In summary, significant differences in absolute  $MA_{AT}$  derived from the COR (MR imaging) and the TE (ultrasound imaging) methods were found. This result has important implications for musculoskeletal modelling, as the method used to determine  $MA_{AT}$  can substantially influence the output of such models. Our results make it clear that when using  $MA_{ATs}$  (e.g., for musculoskeletal modelling), serious consideration needs to be given to the limitations associated with the method used to derive them. This could be done by choosing appropriate limits for performing sensitivity analyses on  $MA_{AT}$  to account for the significant between-method differences (up to 30%). Furthermore, the present results allow us to retrospectively compare published tendon force data more accurately as considerable absolute differences but strong relationships between the two methods were shown. These findings are useful for researchers who are interested in examining moment arm differences between populations (e.g., training-, gender- or growth-related) as population differences are likely to be consistent, independent of the method used. Finally, a good reliability of both methods was demonstrated, which suggests that they are robust

against their limitations. Collectively, the present results will help researchers to make more informed decisions about which method to use when determining  $MA_{AT}$  *in vivo*.

# Chapter 4

# 4 Interactive effects of joint angle, contraction state and method on estimates of Achilles tendon moment arms

## 4.1 Introduction

Muscle-tendon moment arms ( $MA_{AT}$ ) are important input parameters for musculoskeletal models.  $MA_{AT}$  has been estimated in vivo using both the tendon excursion (TE) (Lee et al., 2008; Lee & Piazza, 2009; Maganaris, 2003a) and the centre of rotation (COR) methods (Maganaris et al., 1998a, 2000). Recently,  $MA_{AT}$  magnitude was reported to be significantly smaller when the TE method was used compared to the COR method. However, both methods correlated well across participants for a range of joint angles (see chapter 3) (Fath et al., 2010). Thus, the possibility exists that while  $MA_{AT}$  estimates might differ between the methods, the joint-angle dependent changes in  $MA_{AT}$  are relatively consistent. Furthermore,  $MA_{AT}$  changes with the level of muscle contraction (Maganaris et al., 1998a). However, contrary to the COR method, the measurement of  $MA_{AT}$  using the TE method during muscular contraction (MVC) is associated with severe limitations. The TE method is based on the principle of virtual work (An et al., 1984; Storace & Wolf, 1979) and therefore assumes that the work done by an external torque is equivalent to the virtual work done by the muscles and tendons. Implicit in this is the assumption that no energy is lost during a muscle contraction. Given that muscles and tendons store, release and dissipate elastic energy during muscle contractions, the principle of virtual work is violated, and this violation is likely to be more significant when large muscle forces are produced. Using the relationship between moment arms obtained from the TE method at rest and those obtained using the COR method during a MVC might therefore be a more meaningful way of accounting for contraction-dependent changes in moment arms derived from the TE method.

Therefore, the overall purpose of this experiment was to investigate the interactive effects of method, joint angle and contraction level on  $MA_{AT}$  estimates. The specific aims were: (1) to test the assumption that the  $MA_{AT}$  estimated using the TE method at rest ( $TE_{rest}$ ) would be related to estimates obtained using the COR method during MVC ( $COR_{MVC}$ ), and (2) to test the assumption that joint angle-related changes in  $MA_{AT}$

would be independent of method and contraction state.

## 4.2 Methods

With institutional ethical approval and after providing written informed consent, six healthy adults (4 men and 2 women) participated in this study (age =  $30 \pm 6$  years, stature =  $1.76 \pm 0.11$  m, mass =  $74 \pm 14$  kg).

$MA_{AT}$  about the right ankle joint was obtained using both the COR (at rest and during MVC) and TE (at rest) methods. For the COR method, participants were asked to lie supine in a 3-Tesla magnetic resonance (MR) imaging scanner (Siemens Magnetom Trio syngo MR 2004A) with their leg straight (i.e. knee angle =  $180^\circ$ ). The foot was securely fastened with two inelastic Velcro straps. MR images were taken during rest and MVC (sagittal scans, TR = 600 [20] ms, TE = 12 [5] ms, 3 [1] excitations, 300-mm field of view, 2 [3]-mm slice thickness for rest and MVC [ ]) at six different ankle positions ( $-30^\circ$  to  $+45^\circ$ , in  $15^\circ$  increments;  $0^\circ$  = ankle perpendicular to tibia). Using the MR images, the COR of the ankle joint, the line of action (of the Achilles tendon) and consequently the  $MA_{AT}$  were determined during rest and MVC at  $-15^\circ$ ,  $0^\circ$ ,  $15^\circ$  and  $30^\circ$  ankle angles using the Reuleaux method as previously described by others in detail (Maganaris et al., 1998a; Reuleaux, 1875) (Figure 2.1). One day prior to the testing, participants had been familiarised with the testing device and the protocol at Brunel University. Since torque measurements during the MRI testing were not possible, participants had been introduced to perform maximal isometric contractions using an isometric dynamometer. During the MR testing, participants were asked to replicate these contractions.

For the TE method, participants were seated on an isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, Inc., NY) with their right knee straight ( $180^\circ$ ) and a relative hip angle of  $85^\circ$ . The right foot was secured firmly to the dynamometer's footplate with the lateral malleolus aligned with the COR of the dynamometer. The ankle was passively rotated through its range of motion by the dynamometer five consecutive times at  $10^\circ \cdot s^{-1}$ . For all participants, the range of motion was greater than  $-15^\circ$  (dorsiflexion) and  $30^\circ$  (plantarflexion). To determine tendon displacement, a 10-MHz B-mode, 40-mm linear ultrasound probe (Esoate Megas GPX, Genova, Italy) was placed over the muscle-tendon junction (MTJ) of the gastrocnemius medialis (GM). Raw position data (sampled at 1000 Hz) from the isokinetic dynamometer were low-pass filtered ( $4^{th}$ -order butterworth, zero-lag, 3.75 Hz cut-off). Analogue ultrasound video data were sampled at 25 Hz. The positions of the MTJ and the Achilles tendon were manually digitised, low-pass filtered ( $4^{th}$ -order butterworth, zero-lag, 2.63 Hz cut-off) and downsampled to 25 Hz. The tendon and joint angular displacement data were plotted against joint angular displacement over the interval of  $-15^\circ$  and  $30^\circ$ , and approximated by fitting a  $2^{nd}$ -order polynomial (mean ( $\pm$

SD) coefficient of determination  $0.996 \pm 0.002$ ). To calculate the  $MA_{AT}$ , the polynomial was analytically differentiated at the four ankle angles of interest.  $MA_{AT}$  measurements were analysed three times at each angle for each method. The coefficient of variation was smaller than 5% for all conditions. More detailed descriptions of the experimental protocol and the derivation of  $MA_{AT}$  are described in chapter 3.

To determine the relationship between  $TE_{rest}$  and  $COR_{rest}$  as well as  $TE_{rest}$  and  $COR_{MVC}$ , eight Pearson's product moment correlations were performed (one at each angle for each comparison). To test if  $MA_{AT}$  obtained using  $TE_{rest}$ ,  $COR_{rest}$  and  $COR_{MVC}$  would change similarly as a function of ankle angle, and, to determine if this change was independent of muscular contraction level, a repeated measures ANOVA ( $3 \times 3$ ,  $TE_{rest}$ ,  $COR_{rest}$ ,  $COR_{MVC}$  at ankle angles of  $-15^\circ$ ,  $0^\circ$  and  $15^\circ$ ) was performed. Here, a method  $\times$  angle interaction was tested. To test this effect independent of differences in  $MA_{AT}$  magnitude, all  $MA_{AT}$  values were normalised by the  $MA_{AT}$  obtained at  $120^\circ$  for the corresponding condition. To further illustrate the changes of  $MA_{AT}$  across angles and method, the correlations between  $MA_{AT}$  and ankle angle for each participant and each method was quantified. To provide more specific information about the  $COR_{rest}$  -  $COR_{MVC}$  comparison, the percentage differences for all angles were reported. Statistical significance was accepted at an alpha of .05.

### 4.3 Results

The correlations coefficients between  $TE_{rest}$  -  $COR_{rest}$  and  $TE_{rest}$  -  $COR_{MVC}$  ranged between .63 to .92 and .72 to .93, respectively (Table 4.1).

The repeated measures ANOVA revealed no significant method  $\times$  angle interaction ( $F(4, 20) = 0.769$ ;  $p = 0.558$ ) (Figure 4.1). The mean correlations between  $MA_{AT}$  and ankle angle were  $1 \pm .00$ ,  $.91 \pm .10$  and  $.95 \pm .08$  for  $TE_{rest}$ ,  $COR_{rest}$  and  $COR_{MVC}$ , respectively.  $MA_{AT}$  magnitudes were larger at the  $30^\circ$  than the  $-15^\circ$  ankle angle with mean differences of  $24.5 \pm 12.2\%$ ,  $19.9 \pm 6.3\%$  and  $24.3 \pm 7.3\%$  for  $TE_{rest}$ ,  $COR_{rest}$  and  $COR_{MVC}$ , respectively. When comparing  $COR_{rest}$  and  $COR_{MVC}$ , the percentage differences in  $MA_{AT}$  ( $\pm$  SD) were  $0.8 \pm 6.5\%$ ,  $3.7 \pm 2.8\%$ ,  $5.5 \pm 6.4\%$  and  $7.9 \pm 6\%$  for  $-15^\circ$ ,  $0^\circ$ ,  $15^\circ$  and  $30^\circ$ , respectively.

### 4.4 Discussion

The first aim of this study was to directly compare  $MA_{AT}$  obtained from  $TE_{rest}$  and  $COR_{MVC}$ . A strong correlations between  $MA_{AT}$  obtained from  $TE_{rest}$  and  $COR_{rest}$  as well as  $COR_{MVC}$  across a range of ankle angles with  $TE_{rest}$   $MA_{AT}$  being significantly smaller was found. These findings extend the results of chapter 3 by demonstrating that  $MA_{AT}$  obtained using the COR method at rest and MVC and TE method at rest are

Table 4.1. Pearson's product moment correlations between Achilles tendon moment arm obtained from the COR and the TE methods at four different ankle angles

angle	Moment arm (mm) (mean $\pm$ SD)			TE <sub>rest</sub> - COR <sub>rest</sub>		TE <sub>rest</sub> - COR <sub>MVC</sub>	
	TE <sub>rest</sub>	COR <sub>rest</sub>	COR <sub>MVC</sub>	r	R <sup>2</sup>	r	R <sup>2</sup>
-15°	30.8 $\pm$ 6.7	46.1 $\pm$ 3.3	46.5 $\pm$ 3.4	.63	.39	.72	.51
0°	34.4 $\pm$ 4.5	51.7 $\pm$ 4.3	53.8 $\pm$ 5.5	.92**	.84	.93**	.86
15°	37.9 $\pm$ 5.4	55.4 $\pm$ 1.5	58.9 $\pm$ 5.5	.69	.48	.77*	.59
30°	41.4 $\pm$ 6.9	56.7 $\pm$ 2.4	61.8 $\pm$ 5.2	.75*	.56	.78*	.61

Note: Values are mean  $\pm$ SD for moment arm; n = 6 for Pearson's product-moment correlations between Achilles tendon moment arm obtained from the centre of rotation (COR; magnetic resonance imaging) and the tendon excursion (TE; ultrasound imaging) methods) at four ankle angles at rest (TE<sub>rest</sub>, COR<sub>rest</sub>) and a maximal voluntary contraction (COR<sub>MVC</sub>); \* = indicates a correlation of Achilles tendon moment arm using the TE and the COR methods ( $p < .10$ ); \*\* = indicates a significant correlation of Achilles tendon moment arm using the TE and the COR methods ( $p < .05$ )

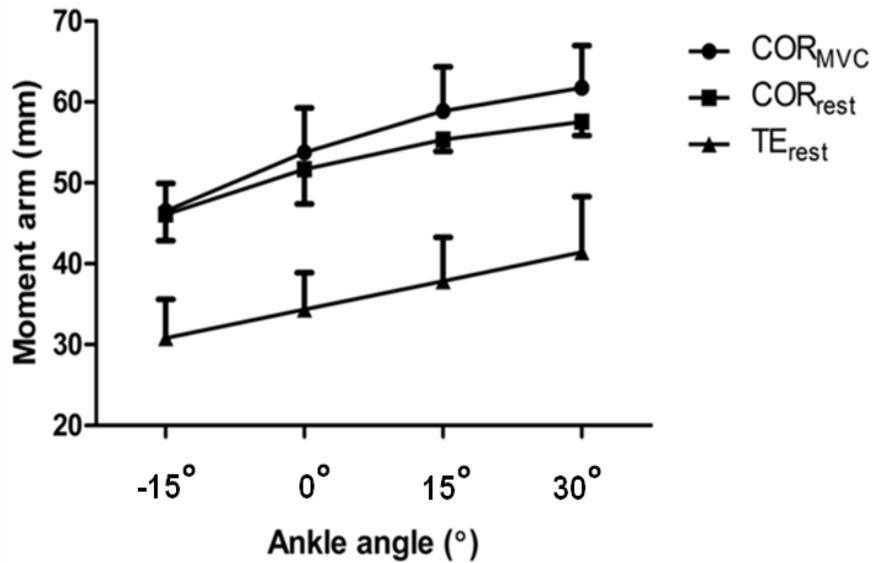


Figure 4.1. Achilles tendon moment arm measurements (mean  $\pm$  SD) at four different ankle angles obtained from the tendon excursion method at rest (TE<sub>rest</sub>) and from the centre of rotations method at both rest (COR<sub>rest</sub>) and during a maximum isometric contraction (COR<sub>MVC</sub>). 0° ankle angle refers to the foot being perpendicular to the tibia.

well correlated and therefore independent of contraction state. The significantly smaller  $MA_{AT}$ 's obtained from the TE method can be explained by the viscoelastic nature of the tendon. As the Achilles tendon relaxes during plantarflexion rotation, the displacement of MTJ for a given joint rotation is reduced which leads to an underestimation of  $MA_{AT}$ . The second aim was to test the hypothesis that  $MA_{AT}$  would change as a function of ankle angle independently of the method of  $MA_{AT}$  estimation. In conformity with this hypothesis, it was found: (1) no angle  $\times$  method interaction, and (2) similar moment arm joint angle correlations for all experimental conditions (Figure 4.1). Our results extend previous findings by demonstrating that the relationship between  $MA_{AT}$  and joint angle is not only independent of muscular contraction level but also of the method used.

Another interesting aspect of our data is the difference in  $MA_{AT}$  between  $COR_{rest}$  and  $COR_{MVC}$ . This difference ranged between 1 - 8%, which is considerably smaller than that reported by Maganaris et al. (1998a), who found differences between 22 - 27%. This discrepancy can potentially be explained by the different knee angle used by Maganaris et al. (1998a) ( $90^\circ$ ) compared to the present investigation ( $180^\circ$ ). A knee angle of  $180^\circ$  was adopted in an attempt to minimise the influence of tendon slack, which has the potential to introduce error into the  $MA_{AT}$  estimation when using the TE method (An et al., 1984). The increase in  $MA_{AT}$  during MVC compared to rest can be explained by a shift of the Achilles tendon away from the joint centre, due to a shortening of the plantarflexor muscles (Maganaris et al., 1998a). This shift is possibly smaller in magnitude when the knee is fully extended compared to a more flexed position. Support for this speculation comes from Riemann, DeMont, Ryu, and Lephart (2001) who demonstrated that muscle stiffness of the GM is greater during full knee extension compared to more flexed positions. A direct consequence of the greater stiffness could be a reduction of AT movement during MVC and therefore a reduced increase in  $MA_{AT}$  during MVC when compared to rest. Our results, in combination with those of Maganaris et al. (1998a), let us speculate that there is an interaction between knee angle,  $MA_{AT}$  and plantarflexor contraction level. Future research should be conducted to specifically test this hypothesis.

The present findings have specific implications for musculoskeletal modelling. Our descriptive results can be used as guidance for modellers to account for the dependence of  $MA_{AT}$  on ankle angle and contraction level. However, it is important to consider the within-group variability observed in our participants. The somewhat large standard deviations reported here indicate that the interaction between ankle angle, muscular contraction level and  $MA_{AT}$  can differ between individuals. This variability should be taken into consideration by performing appropriate sensitivity analyses.

# Chapter 5

# 5 The impact of Achilles tendon moment arm methods and anatomical modifiers on Achilles tendon forces during an inverse dynamic simulation of submaximal cycling

## 5.1 Introduction

Quantifying muscular forces is important in many contexts. Individual muscle forces can be used to monitor and increase sporting performance or support the process of rehabilitation in order to prevent injuries of the muscle-tendon complex (Orishimo et al., 2008). While direct measurements of muscular forces are highly invasive (Gregor, Komi, & Järvinen, 1987; Gregor et al., 1991; Finni et al., 1998; Komi et al., 1987), non-invasive modelling techniques such as inverse dynamics can be used to estimate muscle torques (e.g., gait analysis: Zajac et al. (2002); cycling: Hull and Jorge (1985); Neptune, Kautz, and Hull (1997)). The net muscle torque is the combined rotational force of a muscle group about a particular joint. The sum of muscle forces produced by individual muscles within the muscle group can be obtained by dividing the net muscle torque by the corresponding muscle-tendon moment arms. If muscles and tendons act in a similar way, the muscle force will be equal to the tendon force as the tendons connect the muscles with the corresponding joint.

When using a multi-segmental musculoskeletal model, moment arms are usually based on the musculoskeletal geometry of the model as moment arms are computed as the ratio of changes in muscle-tendon length to joint rotation (An et al., 1984; Delp et al., 1990; Hoy, Zajac, & Gordon, 1990; Zajac, 1989). In this context, the musculoskeletal geometry of the model (i.e. the origin and insertion sites of the muscle-tendon complex to the bony structures) is taken from cadaveric measurements from the literature (Brand et al., 1982; Wickiewicz et al., 1983). However, it has been shown that experimentally determined moment arms increase the accuracy of individual musculoskeletal models as an under- or overestimation of moment arm has a direct impact of net muscle forces (Anderson & Pandy, 2001; Zajac, Neptune, & Kautz, 2003). Ericson et al. (1985) as well as Orishimo et al. (2008), for example, quantified the Achilles tendon force by dividing

plantarflexor torque about the ankle joint by a constant Achilles moment arm ( $MA_{AT}$ ) whereas Raasch et al. (1997) determined the moment arm based on the musculoskeletal geometry taking ankle angle changes into account. To show the importance of accurate moment arm estimations, invasive measurements of the Achilles tendon force using a buckle-type tendon force transducer revealed an overestimation of Achilles tendon force when compared to studies using an inverse dynamic approach (Ericson et al., 1985; Gregor et al., 1987, 1991). As both studies report similar values for maximal plantarflexor torque, an explanation could be the length of the  $MA_{AT}$ .

For muscular forces to be estimated accurately and to make meaningful comparisons between studies, it is important to consider the methods that were used in order to determine  $MA_{AT}$ . Several methods exist to determine  $MA_{AT}$ . These include cadaveric measures (Visser et al., 1990), the geometric approach of the centre of rotation method (COR: Reuleaux (1875)) using magnetic resonance imaging (MRI) (Maganaris et al., 1998a; Rugg et al., 1990), the tendon excursion method (TE) *in vitro* (Spoor et al., 1990) and *in vivo* using ultrasonography or MRI (Maganaris, 2003a; Maganaris et al., 2000).

From such studies it can be learned that  $MA_{AT}$  changes as a function of ankle angle and contraction state (Maganaris et al., 1998a; Spoor & van Leeuwen, 1992). It has been shown that  $MA_{AT}$  increases up to 25% as a function of ankle angle from dorsiflexion to plantarflexion ( $-15^\circ$  to  $30^\circ$ ) (Maganaris et al., 1998a). Maganaris et al. (1998a) also demonstrated that  $MA_{AT}$  increases by 22% - 27% between rest and a maximal voluntary contraction (MVC), which is caused by the thickening of the triceps surae (Maganaris & Paul, 1999). In contrast, in chapter 4, it was demonstrated that  $MA_{AT}$  increases only by 1% - 8% between rest and MVC which is also supported by Baxter et al. (2011). It can be speculated, that the likely reason for the discrepancy of these results could be the difference in knee angle as the only methodological difference was the knee angle ( $90^\circ$  Maganaris et al. (1998a);  $180^\circ$  in chapter 4). Based on these results, it is reasonable to believe that the relationship between  $MA_{AT}$  and contraction state is likely to be influenced by knee angle (Maganaris, 2003a). Finally, it has been shown that the magnitude of  $MA_{AT}$  depends on the method being employed. In chapter 3, it was demonstrated that  $MA_{AT}$ s are 30% greater when the COR method compared to the TE method is used.

It becomes clear that the interactions between ankle angle, knee angle, contraction state and method used need to be taken into consideration when using biomechanical models that estimate Achilles tendon force from muscle torques and  $MA_{AT}$ .

The default settings of the available biomechanical analysis software often do not take all these interactions into consideration. For instance, the default setting of OpenSim and SIMM (an open source and commercial software, respectively, that allows to build and analyse computer models of the musculoskeletal system and dynamic simulations of movement) and AnyBody (a commercially available software similar to OpenSim which

anatomical representations were taken from CT scans) for example uses a  $MA_{AT}$ , which is derived from the tendon excursion method (An, Chao, Cooney III, & Linscheid, 1979; Delp et al., 1990; Wu, An, Cutlip, Andrew, & Dong, 2009). It does take changes in  $MA_{AT}$  as a function of ankle angle into account, but changes in  $MA_{AT}$  as a function of contraction state are not considered. Whilst it is likely that different methods of calculating  $MA_{AT}$ , and different ways of accounting for the moment arm dependence on joint angles and contraction state effect the  $MA_{AT}$  length, the practical impact of this is not known. The overall aim of this study was to determine the impact of the interactive effects of method, ankle angle, knee angle and contraction state on estimates of  $MA_{AT}$  and consequently on Achilles tendon forces during submaximal cycling. To illustrate this effect, an inverse dynamics approach (Hull & Jorge, 1985; Korff, Hunter, & Martin, 2009) was employed and divided plantarflexor torque by  $MA_{AT}$  to obtain Achilles tendon forces. The specific aims of this study were to quantify 1) how Achilles tendon force depends on the method of obtaining  $MA_{AT}$  (assumption of a constant  $MA_{AT}$  for the TE and the COR methods), 2) how Achilles tendon force would change if the  $MA_{AT}$  dependence on ankle angle was taken into consideration 3) how Achilles tendon force would change if both ankle angle and contraction state were taken into consideration and 4) how Achilles tendon force would change if ankle angle, contraction state and knee angle were taken into consideration. To test the interactive effects of method, ankle angle, knee angle and contraction state, five modifications of  $MA_{AT}$  to take these variables into account were used. To test especially the influence of contraction state on  $MA_{AT}$  and on the resulting Achilles tendon force, the submaximal cycling was performed at two different submaximal resistances.

## 5.2 Methods

### 5.2.1 Participants

Twelve healthy participants (6 men, 6 women) volunteered to participate in this study. Mean ( $\pm$  SD) age, stature and mass were  $27.8 \pm 3.7$  years,  $172.6 \pm 8.1$  cm and  $69.2 \pm 8.5$  kg, respectively. All participants were physically active non competitive cyclists with no recent lower limb musculoskeletal injuries. Institutional ethical approval for this study was granted by the Brunel University research ethics committee, and written informed consent was provided by all participants. All participants visited the laboratory on two occasions. On the first day, they visited the laboratory for familiarisation purposes, where they were introduced to the testing devices as well as the testing protocols. On the second day participants performed a cycling task (protocol 1), a maximum plantarflexor task to determine maximal ankle torque (protocol 2), and a passive ankle rotation to determine  $MA_{AT}$  (protocol 3). All tests were separated by a minimum of 30 minutes.

### **5.2.2 Experimental protocol: Kinematic and kinetic analysis of cycling (protocol 1)**

Before the testing, anthropometric measures (stature, seated height) were taken for each participant. These measurements were used to adjust seat height and crank length of the ergometer to 108.5% and 20% of leg length (body height minus seated height; results were approximated to the nearest 2.5 mm), respectively (Martin, Farrar, Wagner, & Spirduso, 2000; Martin & Spirduso, 2001). To set the crank length, we used a crank that was adjustable from 165 - 185 mm with 2.5 mm increments (varicrank, RacerMate, USA). Before testing, participants performed a warm up protocol that consisted of 5 min cycling at a self selected cadence at 50 W. For the testing, participants were asked to cycle on a magnetically braked cycle ergometer (Velotron, Racermate, USA) for 1.5 minutes at 90 revolutions per minute (rpm). In order to standardise the power output across participants, the instantaneous peak power output based on lean thigh volume in accordance with a method presented by (Martin et al., 2000) was estimated. In order to show the impact of  $MA_{AT}$  methods on Achilles tendon forces at two resistances, participants were then required to cycle at both 10% and 20% of this estimated peak value. The order of the trials was randomised for each participant with three minutes rest between each trial. To ensure a constant pedalling rate throughout the trial, participants received visual feedback on a monitor. During the last 30 seconds of each trial, kinematic and kinetic data were collected. Kinematic data were sampled at 100 Hz using a 4 - camera motion analysis system (Motion Analysis, Santa Rosa, CA, USA). Two reflective markers were placed at the level of the pedal spindle and at the posterior end of the pedal, on the 5<sup>th</sup> metatarsophalangeal joint, the lateral malleolus, the lateral femoral epicondyle, the greater trochanter (GRT), and the anterior superior iliac spine (ASIS) of the right leg. Pedal force data of the right foot were collected at 3000 Hz using a custom built force pedal with two piezoelectric force transducers (Kistler, model 9251AQ01, Winterthur, Switzerland). The electric charge signal was converted into a voltage for each transducer signal by charge amplifiers (Kistler model 5038A, Winterthur, Switzerland) and transformed into a digital signal by a 12-bit analogue-to-digital card (model PCI-6071E, National Instruments, Austin, TX, USA). Both kinematic and kinetic signals were synchronised using Cortex (Motion Analysis, Santa Rosa, CA, USA) and low-pass filtered using a 2<sup>nd</sup>-order, zero phase-lag, Butterworth filter with a cut-off frequency of 10 and 20 Hz, respectively.

### **5.2.3 Calculation of ankle torque during the cycling tasks**

The net muscle torque of the right ankle was derived using an inverse dynamics algorithm (Hull & Jorge, 1985). For this purpose, the foot, shank and thigh were assumed to be rigid segments, and the ankle, knee and hip joints were assumed to be frictionless hinge joints. The foot was assumed to be stationary with respect to the pedal.

In order to calculate the vertical and radial force components in the inertial reference frame, the angles between the position of the two pedal markers and the crank with respect to the global reference frame were used. The centre of the hip was estimated by using the GRT and ASIS markers (Neptune & Hull, 1995). Segmental lengths for the thigh and the shank were determined using the Euclidean distance between the GRT and lateral femoral epicondyle, as well as the lateral femoral epicondyle and the lateral malleolus, respectively.

Relative joint angles in the sagittal plane were determined from the positional data of the estimated joint centres and the segment lengths. Linear and angular velocities and accelerations of the lower limb were determined using Woltring's generalised cross-validation natural B-spline filter (Woltring, 1986). In combination with the pedal forces and the joint positions, the corresponding joint torques for hip, knee and ankle were obtained (Hull & Jorge, 1985). Segmental mass proportions, radii of gyration and centre of mass were estimated based on anthropometric measures as published in the literature (Jensen, 1989). Ankle torque profiles were determined over 15 consecutive crank cycles (see Figure A.4 for a free-body diagram).

#### **5.2.4 Experimental protocol: Maximal isometric plantarflexion torque (protocol 2)**

To determine the maximal isometric plantarflexor torque ( $T_{max}$ ) about the ankle joint, participants were seated on an isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, Inc., NY) with their right knee straight ( $180^\circ$ ) and a relative hip angle of  $85^\circ$ . To prevent movement of the foot and the lower limb during the rotation, the foot was fixed to the dynamometer's footplate with two inelastic straps. In order to measure the torque about the ankle joint, the centre of the ankle joint (lateral malleolus) was aligned with the shaft centre of the isokinetic dynamometer.  $T_{max}$  was determined with the foot plate at  $90^\circ$  (i.e., the sole of the foot was perpendicular to the tibia). Participants were instructed to perform three maximal isometric plantarflexion contractions (MVC). They were instructed to rotate their foot as hard and fast as possible and to hold the contraction for at least four seconds. Verbal encouragement was provided during the maximal contraction. As the measured torque about the ankle joint is the net torque of all the forces produced by agonist and antagonist muscles, the torque produced by the plantarflexor muscles can be underestimated during maximal isometric plantarflexions (Maganaris, Baltzopoulos, & Sargeant, 1998b). To take antagonistic activation of the tibialis anterior (TA) into account, the EMG-torque relationship was established (Brown & McGill, 2008). For this purpose, in addition to the three plantarflexor MVCs, the participants also performed a ramped dorsiflexion contraction. To establish the EMG-torque relationship for the TA, the EMG signals of the gastrocnemius medialis (GM) as a representative of the plantarflexor muscles and TA as well as the torque about the ankle

joint were recorded at 1000 Hz. Electromyographic activity was measured using bipolar self-adhesive surface electrodes with a centre to centre distance of 20 mm. EMG electrodes were placed at the midpoint of the TA and GM muscles parallel to the tibia along the midsagittal plane. Before electrode placement, the skin was shaved and cleaned with isopropyl alcohol pads to reduce impedance. The EMG signals were amplified at a gain of 1000 (Telemyo 2400R, Noraxon USA Inc., Arizona, USA) and converted into a digital signal using a 12-bit analogue-to-digital card (model PCI-6071E, National Instruments, Austin, TX, USA). Then the digital signals were band-pass filtered (20 - 500 Hz), full-wave rectified and filtered ( $2^{nd}$ -order low-pass Butterworth, zero-lag, 6.5 Hz cut-off). Torque data from the isokinetic dynamometer were low-pass filtered ( $4^{th}$ -order Butterworth, zero-lag, 12.47 Hz cut-off).

### 5.2.5 Derivation of dependent variables

To determine the influence of co-contraction on the maximal plantarflexor torque during plantarflexion MVCs, the rectified, filtered TA EMG signal was plotted against the measured torque during the ramped dorsiflexion contraction (up to 50% of the maximal torque value). The data were fitted with a  $3^{rd}$ -order polynomial and the resulting regression equation was used to determine the antagonist torque contribution during the plantarflexion MVCs (Maganaris et al., 1998b). To obtain the corrected isolated plantarflexor torque ( $T_{max\_corr}$ ), the antagonistic contribution was added to the net  $T_{max}$  torque. As the force-length relationship of the GM and therefore the resulting maximal isometric torque is dependent on ankle and knee angle (Maganaris, 2003a), a function based on these published results and adjusted the  $T_{max\_corr}$  measurement to account for ankle and knee angle changes ( $T_{max\_corr\_knee}$ ) was created. These relationships were later used for correction for antagonistic activation of ankle torque ( $T_{max\_corr\_ank}$ ) obtained by the inverse dynamics during cycling. To account for ankle angle the equation was:

$$T_{max\_corr\_ank} = T_{max\_corr} \cdot (-0.002 \cdot \alpha_{ankle}^2 + 0.47) \quad (5.1)$$

To account for the knee angle, we used:

$$T_{max\_corr\_knee} = T_{max\_corr} \cdot (0.0033 \cdot \alpha_{knee} + 0.42) \quad (5.2)$$

### 5.2.6 Experimental protocol: Achilles tendon moment arm (protocol 3)

In order to estimate the  $MA_{AT}$  with respect to the ankle joint, the tendon excursion method was applied (An et al., 1983). Seating position of the participants was the same as for the MVC trials. Tendon displacement was determined using a 40-mm linear ultrasound probe (10-MHz B-mode; Esoate Megason GPX, Italy) during three consecutive passive

rotations over a range of 45° of the ankle joint at 10° s<sup>-1</sup> (-15° dorsi to 30° plantar). During the passive rotations, the muscle-tendon junction (MTJ) of the gastrocnemius medialis (GM) was recorded at 25 Hz and later manually digitised. The ultrasound signal was synchronised with the position data from the isokinetic dynamometer, which was sampled at 1000 Hz, and converted into a digital signal using a 12-bit analogue-to-digital card (model PCI-6071E, National Instruments, Austin, TX, USA). Both signals were low-pass filtered using a 4<sup>th</sup>-order digital Butterworth, zero-lag, filter with a 3.75 Hz and 2.63 Hz cut-off frequency, respectively, as determined by the residual analysis (Winter, 2005). During the passive ankle rotations, participants were instructed to relax their muscles in order to avoid muscular forces effecting tendon excursion and therefore the moment arm calculation. Real time electromyography setup from the plantarflexion MVC trials of the TA and the GM muscles was used to visually monitor muscle activation during the data collection (Telemyo 2400R, Noraxon USA Inc., Arizona, USA).

### 5.2.7 Derivation of Achilles tendon moment arm dependent variables

The tendon displacement data were plotted against joint angular displacement and approximated by fitting a 2<sup>nd</sup>-order polynomial (mean ( $\pm$  SD) coefficient of determination:  $.997 \pm .002$ ). To calculate MA<sub>AT</sub>, the polynomial was analytically differentiated at the ankle angles of interest (i.e. -15°, 0°, 15° and 30°). To quantify the inter-test reliability, MA<sub>AT</sub> measurements were analysed three times for each participant (the mean ( $\pm$  SD) coefficient of variation across all participants was 3.18% ( $\pm$  2.30%)).

### 5.2.8 Five methods of deriving Achilles tendon moment arm

To calculate the Achilles tendon force, net muscle torque about the ankle joint obtained using inverse dynamics during cycling was divided by the corresponding MA<sub>AT</sub> for each participant. As MA<sub>AT</sub> changes as a function of method, ankle angle, knee angle and contraction state (chapter 4 Maganaris et al., 1998a; Spoor et al., 1990), five different modifications of the MA<sub>AT</sub> using the tendon excursion method at rest to account for these changes were used. These five modifications were based on results from previous studies. First, a constant MA<sub>AT</sub> at an ankle angle of 0 (MA<sub>TE</sub>) was taken as it best represents the average ankle angle during cycling (Stienen, Schouten, Schuurmans, & van Der Helm, 2007). Second, as MA<sub>AT</sub> is dependent on the method being employed (see chapter 3)(e.g. TE and COR), a 49% increase of MA<sub>TE</sub> to obtain COR<sub>rest</sub> MA<sub>AT</sub> (MA<sub>COR</sub>) as described in chapter 4 was performed. The method comparison was only performed at the constant MA<sub>AT</sub>. The reason for this was that any method dependent difference in Achilles tendon force would also be present for all other MA<sub>AT</sub> adjustments (ankle angle, knee angle and contraction state). Third, to account for changes in MA<sub>AT</sub> due to changes in ankle angle during cycling, a linear relationship for each participant

based on  $MA_{AT}$  measurements at  $-15^\circ$ ,  $0^\circ$ ,  $15^\circ$  and  $30^\circ$  ( $MA_{TE\_ank}$ ) was used. The fourth modification took both changes in ankle angle and contraction state into consideration ( $MA_{TE\_ankMVC}$ ). As for the contraction state, the  $MA_{TE\_ank}$  was increased by 4.5% based on findings of chapter 4. The fifth condition took changes in ankle and knee angle, as well as contraction state into consideration ( $MA_{TE\_ankkneeMVC}$ ). As an average increase in  $MA_{TE}$  of 4.5% at a knee angle of  $180^\circ$  was found (see chapter 4) and Maganaris et al. (1998a) reported an average increase in  $MA_{TE}$  of 25% at a knee angle of  $180^\circ$ , a linear relationship for the knee angle dependence during contraction state was applied (see Appendix A.1.1). The last  $MA_{AT}$  modification was the most differentiated method, as it takes the largest number of anatomical modifiers into consideration. The Achilles tendon force derived from the resulting moment arms were used as the baseline measure against the tendon forces derived from the remaining conditions.

In order to quantify the effects of the  $MA_{AT}$  modifications on the Achilles tendon force profiles during the cycling task across the two resistances, each of the moment arm manipulations to  $MA_{TE\_ankkneeMVC}$  by calculating the relative absolute deviation across 15 revolutions were compared (RAD, see equation 5.3) (Korff & Jensen, 2007) ( $n = 66$  for each revolution as participants cycled at 90 rpm at a sampling rate of 100 frames per second). The RAD (equation 5.3) provides a percentage mean difference between the force profiles across one revolution and therefore helps to interpret the effect of  $MA_{AT}$  modifications on the Achilles tendon force profiles and the maximal force output in particular.

$$RAD = 100 \cdot \frac{\frac{1}{n} (\sum_{i=1}^n |MA_{TE\_ankkneeMVCi} - MA_{modificationi}|)}{range(MA_{TE\_ankkneeMVC})} \% \quad (5.3)$$

### 5.2.9 Statistical treatment

To compare the RAD by quantifying the difference in Achilles tendon force profiles across the two resistances, descriptive statistics (percentage differences according to equation 5.3) were used. To test the difference of adjusting the magnitude of  $MA_{AT}$  on the resulting maximal Achilles tendon force across the two resistances, a repeated measures two-way ANOVA ( $MA_{AT}$  modification  $\times$  resistance) was used. If the interaction effect was significant, follow-up repeated measures one-way ANOVAs were performed for each resistance. Post hoc paired  $t$ -test (Bonferroni corrected) were then used to determine the difference. Statistical significant was accepted at  $p < .05$ .

## 5.3 Results

Participant characteristics for the cycling were  $91.2 \pm 4.8$  cm for seat height,  $16.8 \pm 0.9$  cm for crank length and  $108.2 \pm 15.8$  W at 10% and  $216.3 \pm 31.7$  W at 20% of their predicted maximal power, respectively. Descriptive results for cycling can be found in

Table 5.1. Descriptive kinematic results during submaximal cycling at two resistances (mean  $\pm$  SD).

resistance (%)	rpm	$T_{max}$ ankle (N · m)	ankle angle		knee angle	
			min (°)	max (°)	min (°)	max (°)
10	90.2	38.6	82.7	102.7	71.8	136.9
	( $\pm$ 0.6)	( $\pm$ 8.6)	( $\pm$ 5.6)	( $\pm$ 6.4)	( $\pm$ 5.2)	( $\pm$ 5.9)
20	90.2	56.9	81.5	105.9	71.1	136.4
	( $\pm$ 0.9)	( $\pm$ 15.7)	( $\pm$ 10.2)	( $\pm$ 7.7)	( $\pm$ 4.3)	( $\pm$ 6.0)

Note: The resistance values refer to 10% and 20% of the predicted peak power output during cycling; rpm = revolutions per minute;  $T_{max}$  ankle = the maximal torque during cycling about the ankle joint as predicted by the inverse dynamics.

Table 5.1. Across all participants,  $MA_{AT}$  estimates obtained using the TE method at ankle angles of  $-15^\circ$ ,  $0^\circ$ ,  $15^\circ$  and  $30^\circ$  were  $28.1 \pm 4.9$ ,  $33.2 \pm 2.8$ ,  $38.2 \pm 3.0$  and  $42.9 \pm 5.4$  mm, respectively.

The RAD ranged from 4.7% to 9.72% with the highest error for  $F_{TEankkneeMVC} - F_{COR}$  and the lowest for  $F_{TEankkneeMVC} - F_{TEankMVC}$  (see Table 5.2).

The repeated measures two-way ANOVA revealed significant main effects for  $MA_{AT}$ -modification ( $F(4, 44) = 64.470$ ,  $p < .01$ ) and resistance ( $F(1, 11) = 46.39$ ,  $p < .01$ ). In addition, there was also an interaction effect for  $MA_{AT}$ -modification  $\times$  resistance ( $F(4, 44) = 17.960$ ,  $p < .01$ ). To locate the differences for a resistance of 10% of predicted peak power, the repeated measures one-way ANOVA revealed a significant main effect for  $MA_{AT}$ -modification ( $F(4, 44) = 59.790$ ,  $p < .01$ ). Bonferroni corrected paired  $t$ -tests revealed significant differences for all comparisons ( $p < .05$ ) with the exception of  $F_{TE} - F_{TEank}$  and  $F_{TE} - F_{TEankMVC}$  ( $p > .05$ ). To locate the differences for a resistance of 20% of predicted peak power, the repeated measures one-way ANOVA revealed a significant main effect for  $MA_{AT}$ -modification ( $F(4, 44) = 55.579$ ,  $p < .01$ ). Bonferroni corrected paired  $t$ -test revealed significant differences for all comparisons ( $p < .05$ ) with the exceptions of  $F_{TE} - F_{TEank}$  and  $F_{TE} - F_{TEankMVC}$  ( $p > .05$ ).

## 5.4 Discussion

The overall aim of this experiment was to determine the impact of the interactive effects of method, ankle angle, knee angle and contraction state on  $MA_{AT}$  and consequently on the estimates of Achilles tendon force during submaximal cycling.

To determine the impact of  $MA_{AT}$  on the Achilles tendon force, five different modifications of  $MA_{AT}$  which were used to divide plantarflexor torque obtained from the inverse dynamics algorithm. To test the influence of  $MA_{AT}$  obtained using the TE and the COR

Table 5.2. The impact of five Achilles tendon moment arm modifications on the maximal resulting Achilles tendon force and the Achilles tendon force profiles (RAD) during submaximal cycling.

resistance (%)	$F_{TE}$ (N)	$F_{COR}$ (N)	$F_{TEank}$ (N)	$F_{TEankMVC}$ (N)	$F_{TEankkneeMVC}$ (N)
10	1165.1 ( $\pm 239.8$ )	782.0*** ( $\pm 160.9$ )	1129.3 ( $\pm 223.5$ )	1113.7 ( $\pm 219.5$ )	982.5** ( $\pm 212.4$ )
20	1714.7 ( $\pm 426.1$ )	1150.8*** ( $\pm 286.9$ )	1717.4 ( $\pm 436.2$ )	1686.8 ( $\pm 435.8$ )	1477.9** ( $\pm 449.2$ )
<i>Relative absolute difference (RAD)*</i>					
10	8.6%	9.7%	5.1%	4.7%	—
20	7.8%	8.6%	6.4%	5.8%	—

*Note:* All Achilles tendon forces are means ( $\pm$  SD) across 15 crank revolutions at 90 rpm. The resistance was set to 10% and 20% of the predicted peak power; the following five Achilles tendon moment arm ( $MA_{AT}$ ) modifications were used to determine Achilles tendon force from plantarflexor torque:  $F_{TE}$  = a constant  $MA_{AT}$  using the tendon excursion (TE) method;  $F_{COR}$  = a constant  $MA_{AT}$  using the centre of rotation (COR) method;  $F_{TEank}$  =  $MA_{AT}$  that changes as a function of ankle angle (TE method);  $F_{TEankMVC}$  =  $MA_{AT}$  that changes as a function of ankle angle and contraction state (TE method);  $F_{TEankkneeMVC}$  =  $MA_{AT}$  that changes as a function of ankle angle, knee angle and contraction state (TE method); \* = the relative absolute difference (RAD) of the Achilles tendon force profiles for each modification was compared to the force profile of  $F_{TEankkneeMVC}$ ; \*\* & \*\*\* = Achilles tendon force is significantly smaller compared to all forces being calculated using the TE method ( $p < .01$ ).

methods (modifications 1 and 2), a constant  $MA_{AT}$  for each of the methods was applied. When using musculoskeletal models, joint angle dependent moment arm changes are often taken into consideration (Buchanan et al., 2004) but not contraction-state-dependent changes (Delp et al., 1990). In order to partition out the impact of both approaches on Achilles tendon force,  $MA_{AT}$  that only changes as a function of ankle angle (modification 3) in comparison to  $MA_{AT}$  that changes as a function of contraction state and ankle angle (modification 4) was used. Maganaris et al. (1998a) reported an increase in  $MA_{AT}$  from rest to MVC from 22% - 27% at a knee angle of 90°. In this study, an average increase of 4.5% at a knee angle of 180° was found (see chapter 4). The bigger increase in  $MA_{AT}$  can potentially be explained by a larger shift of the Achilles tendon from the joint centre (Maganaris et al., 1998a). Therefore we included a  $MA_{AT}$  that changed as a function of ankle angle, knee angle and contraction state (modification 5). As the last modification is the most differentiated approach which takes the largest number of anatomical modifiers into consideration, we used the Achilles tendon force profile of this  $MA_{AT}$  modification to compare it the remaining profiles.

Our findings show that different ways of deriving  $MA_{AT}$  affect the estimation of Achilles tendon forces during cycling considerably, both in terms of peak Achilles tendon forces and derivation across the whole crank cycle. More specifically, our results show that when

applying  $MA_{AT}$  obtained using the TE and the COR methods, the Achilles tendon force was significantly smaller for the COR  $MA_{AT}$  ( $F_{COR}$ ) due to a bigger  $MA_{AT}$ . Although only a constant  $MA_{AT}$  to show the impact of the MA method on the Achilles tendon force was used, the significant difference between forces would still be the same if  $MA_{AT}$  changed as a function of ankle angle and contraction state. The reason for this is that  $MA_{AT}$  changed for both methods in a similar fashion across ankle angles (see chapter 4). These results highlight the importance for researchers to carefully consider which method to use for their musculoskeletal research as it has a significant influence on the force output of the model.

Surprisingly, no significant differences in Achilles tendon forces were found when plantarflexor torque was divided by a constant  $MA_{AT}$  determined at  $0^\circ$  ( $F_{TE}$ ) compared to  $MA_{AT}$  which changes a function of ankle angles ( $F_{TEank}$ ) on the one hand and as a function of ankle angles and contraction state on the other hand ( $F_{TEankMVC}$ ). These results can be explained by the fact that the constant  $MA_{AT}$  was determined at an ankle angle of  $0^\circ$ . The average ankle angle during the cycling trials ranged between  $-13^\circ$  to  $17^\circ$  ( $0^\circ$  foot perpendicular to tibia segment). This range of ankle angles is in line with the literature (Chen, Kautz, & Zajac, 2001). As the maximal plantarflexor torque is produced during the downstroke phase at a crank angle of about  $90^\circ$  (Hull & Jorge, 1985; Neptune, 1999), the ankle angle is close to be  $90^\circ$  (Chen et al., 2001). Therefore, the maximal plantarflexor torque is produced by a similar ankle angles and consequently a difference between a constant  $MA_{AT}$  and a  $MA_{AT}$  which changes as a function of ankle angle cannot be found. Furthermore, an explanation why a difference when adjusting  $MA_{AT}$  to account for changes in ankle angle and contraction state - although a main effect for our two resistances (10% and 20% of predicted peak power) was found- was not found, could be due to the submaximal nature of experimental design. The results indicate, that the maximal plantarflexor torque during the highest resistance was 34% of  $T_{max\_corr}$ . This refers to a maximal increase in  $MA_{AT}$  of 1.5%, respectively (100%  $T_{max\_corr}$  refers to an increase of 4.5% as described in chapter 4). However, studies looking at higher resistances or maximal cycling should be aware of the influence of contraction state on the resulting Achilles tendon force.

Finally, when taking ankle angle, knee angle and contraction state into consideration for the  $MA_{AT}$  estimation, the resulting Achilles tendon force ( $F_{TEankkneeMVC}$ ) was significantly smaller compared to all other modifications with the exception of  $F_{COR}$ . A possible explanation for this result is that the muscle stiffness of the GM is greater during full knee extension when compared to knee flexion (Riemann et al., 2001). A greater stiffness results in a smaller shift of the Achilles tendon during the muscular contraction (MVC) and consequently leads to a smaller  $MA_{AT}$  during a more extended knee position. This difference can be seen when isolating the effect of knee angle by comparing  $F_{TEankMVC}$  and

$F_{TEankkneeMVC}$ . As the greatest plantarflexion torque is produced during the downstroke, the knee angle is in a more flexed position. Therefore, the corresponding  $MA_{AT}$  is bigger due to the greater shift of the Achilles tendon during the contraction. Within the context of cycling, it becomes apparent that  $MA_{AT}$  should take changes in knee angle, ankle angle and contraction state into account. This is particularly important for musculoskeletal models, as they only take ankle angle and knee angle dependence into account but not changes in contraction state. Our results show, that if contraction state is not taken into consideration during submaximal cycling, Achilles tendon forces are overestimated by approximately 12% and 14% at 10% and 20% of predicted peak power, respectively ( $F_{TEank}$  vs.  $F_{TEankkneeMVC}$ ).

In addition, with an increase in resistance from 10% to 20%, the relative absolute difference (RAD) between  $F_{TEank}$  as well as  $F_{TEankMVC}$  in comparison to  $F_{TEankkneeMVC}$  became greater. This shows that the influence of knee angle on Achilles tendon force estimates becomes more prominent. However, this conclusion has to be interpreted with caution. In this experiment it was assumed that the  $MA_{AT}$ s at MVC change linearly from 180° to 90° knee angle (i.e. 4.5% to 25%). While previous research reported that plantarflexion torque decreases as a function of knee flexion (Maganaris, 2003a), future research should focus on how  $MA_{AT}$  is influenced by changes in knee and angle angles during MVC between 90° and 180°.

In addition, our results show the importance of carefully considering how to model  $MA_{AT}$  as a deviation not only exists between the maximal Achilles tendon forces but also between their force profiles across crank cycles (RAD up to 9.7% from the  $F_{TEankkneeMVC}$  profile). Across both resistances, the RAD is higher when using a constant moment arm ( $MA_{TE}$  and  $MA_{COR}$ ) as compared to the modifications that take the angle and contraction state dependence into account. Depending on the specific content, researchers have to decide whether they can accept such a difference across the force profile. However, although our maximal Achilles tendon forces are in line with previous studies (Ericson et al., 1985), direct invasive measurements during cycling report that the Achilles tendon force at 88 W and 265 W was 480 N and 661 N (both at 90 rpm) (Gregor et al., 1987). These results are considerably lower than the present results obtained when dividing the plantarflexor torque by the  $MA_{AT}$  using the TE method. Interestingly, when taking all anatomical modifiers for the TE method into the calculation ( $F_{TEankkneeMVC}$ ), the resulting force is the closest to the direct measurements for the TE method. However, when adjusting the  $MA_{COR}$  to take all anatomical modifiers into the calculation ( $F_{CORankkneeMVC}$ ), 659 N (108 W) and 992 N (216 W) were obtained. Although these forces are still higher than the ones reported by Gregor et al. (1987), using the COR method for  $MA_{AT}$  ( $F_{CORankkneeMVC}$ ) provides the closest Achilles tendon force in comparison to the direct invasive measurements. This finding is in line with Gregor et al. (1991) who used the COR method to determine  $MA_{AT}$

for cycling. As the length of  $MA_{AT}$  using the COR method is significantly larger than using the TE method (see chapter 4), the findings of this experiment let us speculate about the validity of the TE  $MA_{AT}$  results. However, as the Gregor et al. (1987) only used one participant, the results have to be interpreted with caution. Therefore, future studies should focus on comparing estimates of Achilles tendon force using musculoskeletal modeling techniques with direct Achilles tendon force measurements.

In summary, the significant impact of different  $MA_{AT}$ -modifications on the resulting Achilles tendon force during submaximal cycling was shown. Although a significant main effect for our  $MA_{AT}$ -modifications was found, the ankle angle and contraction state dependence of  $MA_{AT}$  was found to be nonsignificant when compared to a constant  $MA_{AT}$ . However, when taken the knee angle dependence into the  $MA_{AT}$  calculation, the resulting Achilles tendon force was closest to results from direct Achilles tendon force measurements (Gregor et al., 1987). In this context, when using  $MA_{AT}$  obtained using the COR method, Achilles tendon forces were significantly smaller but closer to the direct measurements as compared to results obtained using the TE method. Our results from the inverse dynamics analysis show that when calculating muscular forces from joint torques during dynamic movements, it is important to use  $MA_{AT}$  which accounts for changes in ankle angle and contraction state. In addition, our results let us speculate that changes in knee and ankle angle during muscular contraction play an important role for the  $MA_{AT}$  estimation and should be taken into consideration when using musculoskeletal models. Collectively, these present results will enable researchers to make a more informed decision on how to adjust  $MA_{AT}$  for dynamic musculoskeletal models.

# Chapter 6

## 6 General discussion

Muscles and tendons cross joints at a distance allowing for muscle forces to cause joint torques and ultimately segmental rotations. The determination of muscle-tendon moment arms is important because it is a determinant for an analysis of muscle function to better understand the individual's strength, movement economy and sprinting performance with respect to mechanical effectiveness. In addition, within a developmental context, it is important in order to explore age-related changes in human movement. From a methodological perspective, moment arms are important because they allow us to calculate muscular forces from torques. To estimate muscle forces from joint torques, it is vital that methodological and anatomical modifiers are taken into consideration.

The overall purpose of this thesis was to gain insights into the interactions between methodological and anatomical modifiers of Achilles tendon moment arm ( $MA_{AT}$ ) and their effect on the derivation of muscular forces from torque. In order to answer the research questions, three experiments were performed. In experiment one, two common methods of obtaining  $MA_{AT}$  were compared. Following these results, experiment two explored the interaction effect of method, ankle angle and contraction on  $MA_{AT}$  estimates. In experiment three, the interactive effects of methodological and anatomical modifiers of  $MA_{AT}$  on the estimation of Achilles tendon forces during cycling were determined.

### 6.1 Summary of findings

Results from experiment one showed that  $MA_{AT}$  length was significantly smaller when using the tendon excursion (TE) method and ultrasonography as compared to using the centre of rotation (COR) method and magnet resonance imaging. However, besides the significant difference in  $MA_{AT}$  length, a high correlation between both methods across different ankle angles was found. Furthermore, the experiment provided evidence that the selection of fitting techniques in order to describe  $MA_{AT}$  as the ratio of tendon to angular displacement influences the reliability of  $MA_{AT}$  estimates. In addition,  $MA_{AT}$ s were more reliable for the TE method when the ankle joint was rotated in a dorsiflexed direction. Findings of experiment two extended the results of experiment one by showing that  $MA_{AT}$  obtained using the TE method at rest and the COR method at rest as well as MVC are well correlated and change in a similar trend as a function of ankle angles.

To test the practical significance of these findings, it was demonstrated that method-

ological and anatomical modifiers play a significant role when deriving muscle-tendon forces from torques during submaximal cycling. The significantly greater  $MA_{AT}$  length as obtained by the COR method, resulted in a significant smaller Achilles tendon force in comparison to all remaining moment arm modifications. In addition, results also highlight the importance of taking the ankle angle and contraction state dependence of  $MA_{AT}$  into account when using biomechanical modelling techniques. Furthermore, it was shown that when  $MA_{AT}$ -dependence on knee angle was taken into account, the resulting Achilles tendon force was decreased in comparison to only taking the ankle angle and contraction state dependence of  $MA_{AT}$  into account.

## 6.2 Effect of moment arm method on Achilles tendon moment arm estimates

There is increasing evidence that the TE method using ultrasonography produces smaller  $MA_{AT}$  compared to MR imaging or using the COR method (Karamanidis et al., 2011; Lee & Piazza, 2009; Maganaris, 2003a; Spoor et al., 1990). Results from experiment one support this assumption and showed that  $MA_{AT}$  estimates obtained using the COR method are 25% - 30% larger than those obtained using the TE method at rest (Maganaris et al., 1998a). Interestingly, however, the  $MA_{AT}$  estimates for both methods were well correlated. In addition, as shown in experiment two,  $MA_{AT}$  estimates also changed in a similar way across the range of angles. Although this had not been shown before, previously published  $MA_{AT}$  data are suggestive of these relationships (Maganaris et al., 2000). The absolute difference in  $MA_{AT}$  length between the methods needs to be interpreted within the context of the assumptions inherent to each of these methods. Hashizume et al. (2011) recently showed that when comparing 2D and 3D  $MA_{AT}$  using the COR method, the 2D  $MA_{AT}$  was overestimated by approximately 25% when directly compared to 3D estimates. Their findings showed that this difference was caused by a deviation of the medial or lateral misalignment of the 2D sagittal scanning plane. Although the TE method during these present experiments and previous publications was only applied during 2D, the 3D results as presented by Hashizume et al. (2011) are similar to  $MA_{AT}$  length estimates from these studies (Lee & Piazza, 2009; Karamanidis et al., 2011). Within this context,  $MA_{AT}$  estimates from Maganaris (2003a) using the TE method appear to overestimate  $MA_{AT}$  when compared to findings of experiment two (chapter 4) or previous publications (Karamanidis et al., 2011; Lee & Piazza, 2009). This difference could be explained by a methodological difference, as Maganaris (2003a) used individual scans at  $\pm 5^\circ$  of the angle of interest to determine the change in tendon length. In contrast, the remaining studies quantified the change in tendon length continuously over a range of motion (i.e. Karamanidis et al. (2011):  $20^\circ$ ; Lee and Piazza (2009):  $30^\circ$ ). Quantifying the ratio of length changes in

tendon length to changes in ankle angle over larger intervals in experiment one resulted in more reliable  $MA_{AT}$  estimates than by using smaller intervals (such as  $\pm 5^\circ$ ). However, the influence of using still images at specific angles in contrast to a continuous ankle joint rotation is still unknown and needs further research.

Results of experiment three demonstrated the effect of the method-dependent difference in  $MA_{AT}$  length on the Achilles tendon force during submaximal cycling. Interestingly, Achilles tendon forces determined using  $MA_{AT}$  from the TE method were similar to those of Ericson et al. (1985). However, when using  $MA_{AT}$  from the COR method, Achilles tendon forces were closer to the results from direct invasive *in vivo* measurements (Gregor et al., 1987, 1991). These results suggest, that the  $MA_{AT}$  length obtained using the COR method might be more valid compared to using  $MA_{AT}$  estimates obtained using the TE method.

### **6.3 Effect of anatomical modifiers on Achilles tendon moment arm and Achilles tendon force**

Previous studies have shown that  $MA_{AT}$  changes as a function of ankle angle (Klein, 1996; Maganaris et al., 2000) and contraction state (Baxter et al., 2011; Maganaris et al., 1998a) based on changes in the anatomy of the ankle joint as well as in muscle thickness (see sections 2.4.1 and 2.4.2) (Maganaris et al., 1998c; Maganaris & Paul, 1999). Present results of experiment two extend these findings by showing that  $MA_{AT}$  obtained using the TE and the COR methods at rest change in a similar way as a function of ankle angle. As the TE method is based on the principle of virtual work (An et al., 1984), no energy should be lost during a muscle contraction as it violates the principle. Given that muscles and tendons store, release and dissipate elastic energy during muscle contractions, the principle of virtual work is violated, and this violation is likely to be more significant when large muscle forces are produced as during a maximal contraction. Within this context, it was shown for the TE method that  $MA_{AT}$ s obtained during a dorsiflexion ankle rotation were more reliable than those during a plantarflexion rotation. As we found a significant higher passive torque during the dorsiflexion rotations of the ankle joint (chapter 3), the violation of the virtual work principle is likely to be more significant and therefore the  $MA_{AT}$  length is likely to be more erroneous. However,  $MA_{AT}$  obtained using the TE method at rest and the COR method during muscle contraction were well correlated and changed in a similar way as a function of ankle angle (Maganaris et al., 2000). Interestingly, however, the relative increase in  $MA_{AT}$  between rest and MVC for the COR method was only between 1% to 8% across the range of motion whereas Maganaris et al. (1998a) reported an increase of 22% to 27%. The potential difference can be explained by the different knee angles during the measurement as described in chapter 4. Results of experiment

three demonstrated that the presumed knee-angle dependence for the  $MA_{AT}$  estimation during contraction state had a significant effect on the resulting Achilles tendon force. However, the presumed knee-angle dependence has not been specifically quantified yet. Interestingly, the angle- and contraction-state-dependence of  $MA_{AT}$  was not significantly different when using a constant  $MA_{AT}$ . As the cycling task was only performed under submaximal resistance (10% and 20% of the predicted peak power), the influence of the contraction-state-dependence of  $MA_{AT}$  should be explored during higher resistances.

## 6.4 Significance of findings and suggestions for future research

By using a direct comparison approach, the results of this thesis advance previous findings by giving novel insights into the influence of moment arm method and anatomical modifiers on  $MA_{AT}$  estimates and their effect on Achilles tendon forces during dynamic movements. These findings extend previous findings in the area of  $MA_{AT}$  estimations using the TE and the COR methods at rest (Hashizume et al., 2011; Karamanidis et al., 2011; Lee & Piazza, 2009; Maganaris et al., 2000; Maganaris, 2003a; Sheehan, 2012) and during muscular contraction (Baxter et al., 2011; Maganaris et al., 1998a, 1999). Previous studies applying the TE method using ultrasonography and MR imaging reported contradicting  $MA_{AT}$  length estimates and therefore suggested an influence of moment arm method on  $MA_{AT}$  estimates (Karamanidis et al., 2011; Lee & Piazza, 2009; Maganaris et al., 2000; Maganaris, 2003a). Results of this thesis provide explicit evidence that the moment arm method significantly influences  $MA_{AT}$  estimates. As demonstrated in chapter 5, this method-dependence has a direct influence on the estimation of Achilles tendon forces when using biomechanical modelling techniques such as inverse dynamics. In addition to the method dependence, results of these present studies highlight the importance of taking the the movement direction og the ankle joint, the ankle angle and contraction state dependence of  $MA_{AT}$  estimations into account. More specifically, the presumed influence of knee angle on  $MA_{AT}$  estimates during contraction state was shown to have a significant effect on the resulting Achilles tendon force. However, the influence of knee angle changes on  $MA_{AT}$  is not completely understood and needs further research.

The results of this thesis highlight the importance of carefully considering the influence of moment arm method on the results of biomechanical simulations when comparing with previous studies. Current biomechanical models using the TE method may overestimate the simulated forces based on the reported method difference and because they do not change the moment arm as a function of contraction state (Maganaris et al., 1998a). For an analysis of muscle function, the method-dependence has to be also taken into consideration. In order to make muscular forces results comparable across groups, the same moment arm method should be used. This also implies to studies looking at movement economy. The

influence of moment arm method for the gear ratio of sprinters, for example, can be seen by two recent studies (Baxter et al., 2011; Lee & Piazza, 2009). While Lee and Piazza (2009) used the TE method to estimate the  $MA_{AT}$  of sprinters and non-sprinters, Baxter et al. (2011) applied the COR method for the same purpose. Lee and Piazza (2009) reported  $MA_{AT}$  estimates which were 29% smaller for non-sprinters and 40% smaller for sprinters compared to Baxter et al. (2011). Although both studies found a significant difference in  $MA_{AT}$  between both groups, based on the results from Lee and Piazza (2009) the gear ratio of the sprinters and non-sprinters would be higher for a given foot length. Within a developmental context, the high correlation found between the TE and COR methods allow researchers to examine growth-related changes independent of the method used.

The  $MA_{AT}$  of all our participants changed significantly as a function of moment arm methods. Although the absolute length of  $MA_{AT}$ s for each method is in line with previous publications for both methods (Karamanidis et al., 2011; Lee & Piazza, 2009; Maganaris et al., 1998a; Maganaris, 2003a), the different sample sizes as well as the heterogeneity of the participant groups make it difficult to compare the results. Future studies may wish replicate the direct method comparison by using a more homogeneous participant group. Within this context, as both methods produce different but reliable  $MA_{AT}$  results, the question of the validity of  $MA_{AT}$  length for both methods remains unknown

Although  $MA_{AT}$  changes as function of contraction state and ankle angle, there is some evidence from these present studies that the  $MA_{AT}$  is also influenced by knee angle changes during contraction state. The results from experiment three (chapter 5) show for the first time the significant influence of knee angle on  $MA_{AT}$  estimates and consequently the Achilles tendon force during cycling. In experiment three, it was assumed that the increase in  $MA_{AT}$  to the contraction state changed linearly between knee angles of 90° and 180°. Future studies may focus on how  $MA_{AT}$  is influenced during contraction state throughout the full range of motion of the knee joint.

## 6.5 Hypotheses

### Study 1: Research hypothesis

$H_0$ : *In vivo*  $MA_{AT}$  obtained using the tendon excursion method will be significantly smaller than the  $MA_{AT}$  obtained using the centre of rotation method.

$H_0$ : Larger intervals or higher polynomial fittings for the ratio of tendon and angular displacement data will result in more reliable measurements of  $MA_{AT}$ .

**ACCEPTED**

$H_0$ : Passive dorsiflexion in comparison to plantarflexion rotations of the ankle will result in a higher reliability of  $MA_{AT}$  measures.

**ACCEPTED**

### Study 2: Research hypothesis

$H_0$ : *In vivo*  $MA_{AT}$  obtained using the tendon excursion method at rest will be highly correlated to those obtained from the centre of rotation method during a maximal voluntary contraction.

**ACCEPTED**

$H_0$ : There will be no significant interaction effect on  $MA_{AT}$  between moment arm method, joint angle and contraction state.

**ACCEPTED**

### Study 3: Research hypothesis

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  obtained using the tendon excursion and centre of rotation methods.

**ACCEPTED**

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  which changes as a function of ankle angle during submaximal cycling.

**REJECTED**

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  which changes as a function of ankle angle and contraction state during submaximal cycling.

**REJECTED**

$H_0$ : There will be significant differences in Achilles tendon force when using  $MA_{AT}$  that changes as a function of ankle angle, knee angle and contraction state.

**ACCEPTED**

## 6.6 Limitations

The methods being used in three experimental studies of this thesis have been carefully selected and were to the best of the knowledge of the PhD candidate at the time. This also included the collection and analysis as well as the statistical processing of data. However, besides the carefully designed experiments, a number of limitations have to be acknowledged.

The first limitation refers to the 2D design of the TE and the COR methods to estimate  $MA_{AT}$ . In section 2.3 the influence of 2D versus 3D ankle movement was described in detail. Recently, two studies quantified the impact of using the 2D COR method in contrast to a 3D approach on  $MA_{AT}$  using MR imaging (Hashizume et al., 2011; Sheehan, 2012). The author found a significant overestimation of 2D  $MA_{AT}$  of about 25% in comparison to 3D  $MA_{AT}$  (Hashizume et al., 2011). If these findings were applied to the COR  $MA_{AT}$  results of the present studies, a dependence of  $MA_{AT}$  method would not

be detectable. In addition, when measuring tendon displacement by tracking the MTU, researchers have used skin marker to determine the relative movement of the probe with respect to the MTJ during contraction state (Magnusson et al., 2001; Muramatsu et al., 2001). However, as changes in tendon length were only determined during rest, Maganaris (2005) demonstrated that the skin marker relative to the probe only deviated by 0.2 mm, while calcaneal and MTJ displacement was equal. Based on these findings, the use of skin marker was neglected. However, small deviations in tendon displacement may have occurred.

A second limitation refers to relatively small sample size in experiment two ( $n = 6$ ). The sample size number was derived based on the sample sizes used in previous studies (Maganaris et al., 1998a, 2000). In order to perform an a priori power analysis, reference data have to exist to draw a comparison. Since Maganaris et al. (2000) suggested that there is no “gold standard” to estimate  $MA_{AT}$ , therefore an a priori power analysis was not possible. However, results of experiments 1 and 2 indicate moderate to strong agreements of  $MA_{AT}$  values obtained using the TE method at rest and the COR method at rest as well as at MVC. When looking at the results reported by Maganaris et al. (2000) for the COR and TE methods using MR imaging, the results show a similar trend as  $MA_{AT}$  for TE method at rest increased from 4.3 to 5.6 cm and 5.4 to 7 cm for the COR method at MVC. In both cases there is an increase from dorsi- to plantarflexion of 25% and 30%, respectively. Thus, besides the small sample size, it is believed that the external validity of the present results and the correlation between the methods will not change if a higher sample size is used.

A third limitation occurs by neglecting the contribution of muscular co-activation to the net muscle torque (Maganaris et al., 1998b). As the net muscular torque is divided by the  $MA_{AT}$ , it is assumed that net muscular torque is only produced by the plantarflexor muscles. However, co-activation of the dorsiflexor muscle is also present during a plantarflexion movement. Thus, the contribution of the plantarflexor muscles is underestimated.

## 6.7 Summary

The present experiments of this thesis extend the existing knowledge on how to estimate  $MA_{AT}$  at rest and during contraction state and which anatomical modifiers have to be taken into consideration. In experiment one it was shown that the length of  $MA_{AT}$  is dependent on the method being employed. When using the TE method,  $MA_{AT}$  was significantly smaller compared to  $MA_{AT}$  obtained using the COR method. However, besides the method-dependent differences, results were well correlated. As the TE method cannot be applied during contraction state,  $MA_{AT}$ s in experiment two were compared for the TE method at rest and the COR method as rest and MVC across the range of motion.

Similar to experiment one, the results show that  $MA_{AT}$  obtained using the TE method at rest are well correlated with those obtained using the COR method at rest and MVC. In addition, the results of both methods changed in a similar way across the range of motion of the ankle joint. Future research should further explore the method-dependence and correlation for additional moment arm estimates (i.e. TA moment arm). Finally, the results of experiment three showed the effect of moment arm method on the Achilles tendon force during submaximal cycling. It was found that besides the  $MA_{AT}$  dependence of method, ankle angle and contraction state, changes in knee angle during the contraction state should also be considered. Within this context, future studies should aim to explore how the presumed knee angle-dependence influences  $MA_{AT}$  estimates during contraction throughout its range of motion. As most dynamic movements involve changes in knee angle, this will improve Achilles tendon force estimates for those movements. Additionally, the effect of contraction state and knee angle should be also be explored during maximal cycling or alternative maximal movement tasks. Because of the submaximal nature of experiment three, changes in  $MA_{AT}$  due to changes in the contraction state were not significant.

## 6.8 Conclusions

The results of the three experimental studies of this thesis show that  $MA_{AT}$  obtained using the TE method are significantly smaller than those obtained using the COR method. In spite of the differences,  $MA_{AT}$ s are well correlated and change in a similar way as a function of ankle angle and contraction state. In addition, besides the significantly larger Achilles tendon force due to a smaller  $MA_{AT}$  during an inverse dynamics simulation of submaximal cycling, changes in knee angle during the contraction should be taken into account. Together, these results highlight the importance of carefully considering the  $MA_{AT}$  dependence of method, ankle angle, knee angle and contraction state for future research using biomechanical models.

# References

## References

- Aagaard, P., Simonsen, E. B., Andersen, J. L., Magnusson, P., & Dyhre-Poulsen, P. (2002). Increased rate of force development and neural drive of human skeletal muscle following resistance training. *Journal of Applied Physiology*, *93*(4), 1318–1326. DOI: 10.1152/japplphysiol.00283.2002
- Ackland, D. C., Pak, P., Richardson, M., & Pandy, M. G. (2008). Moment arms of the muscles crossing the anatomical shoulder. *Journal of Anatomy*, *213*(4), 383–390. DOI: 10.1111/j.1469-7580.2008.00965.x
- An, K., Chao, E., Cooney III, W., & Linscheid, R. (1979). Normative model of human hand for biomechanical analysis. *Journal of Biomechanics*, *12*(10), 775–788. DOI: 10.1016/0021-9290(79)90163-5
- An, K., Takahashi, K., Harrigan, T. P., & Chao, E. Y. (1984). Determination of muscle orientations and moment arms. *Journal of Biomechanical Engineering*, *106*(3), 280. DOI: 10.1115/1.3138494
- An, K., Ueba, Y., Chao, E., Cooney, W., & Linscheid, R. (1983). Tendon excursion and moment arm of index finger muscles. *Journal of Biomechanics*, *16*(6), 419–425. DOI: 10.1016/0021-9290(83)90074-X
- Andersen, L. L., & Aagaard, P. (2006). Influence of maximal muscle strength and intrinsic muscle contractile properties on contractile rate of force development. *European Journal of Applied Physiology*, *96*(1), 46–52. DOI: 10.1007/s00421-005-0070-z
- Anderson, F. C., & Pandy, M. G. (2001). Static and dynamic optimization solutions for gait are practically equivalent. *Journal of Biomechanics*, *34*(2), 153–161. DOI: 10.1016/S0021-9290(00)00155-X
- Arampatzis, A., Morey-Klapsing, G., Karamanidis, K., DeMonte, G., Stafilidis, S., & Brüggemann, G.-P. (2005). Differences between measured and resultant joint moments during isometric contractions at the ankle joint. *Journal of Biomechanics*, *38*(4), 885–892. DOI: 10.1016/j.jbiomech.2004.04.027
- Baltzopoulos, V. (1995). A videofluoroscopy method for optical distortion correction and measurement of knee-joint kinematics. *Clinical Biomechanics*, *10*(2), 85–92. DOI: 10.1016/0268-0033(95)92044-M
- Baxter, J. R., Novack, T. A., Van Werkhoven, H., Pennell, D. R., & Piazza, S. J. (2011). Ankle joint mechanics and foot proportions differ between human sprinters and

- non-sprinters. *Proceedings. Biological sciences / The Royal Society*. DOI: 10.1098/rspb.2011.2358
- Blazevich, A. J., Coleman, D. R., Horne, S., & Cannavan, D. (2009). Anatomical predictors of maximum isometric and concentric knee extensor moment. *European Journal of Applied Physiology*, *105*(6), 869–878. DOI: 10.1007/s00421-008-0972-7
- Bobbert, M. F., Huijing, P. A., & van Ingen Schenau, G. J. (1986). A model of the human triceps surae muscle-tendon complex applied to jumping. *Journal of Biomechanics*, *19*(11), 887–898. DOI: 10.1016/0021-9290(86)90184-3
- Brand, R. A., Crowninshield, R. D., Wittstock, C. E., Pedersen, D. R., Clark, C. R., & van Krieken, F. M. (1982). A model of lower extremity muscular anatomy. *Journal of Biomechanical Engineering*, *104*(4), 304–311. DOI: 10.1115/1.3138363
- Brown, S. H. M., & McGill, S. M. (2008). Co-activation alters the linear versus non-linear impression of the EMG-torque relationship of trunk muscles. *Journal of Biomechanics*, *41*(3), 491–497. DOI: 10.1016/j.jbiomech.2007.10.015
- Buchanan, T. S., Lloyd, D. G., Manal, K., & Besier, T. F. (2004). Neuromusculoskeletal modeling: estimation of muscle forces and joint moments and movements from measurements of neural command. *Journal of Applied Biomechanics*, *20*(4), 367–95.
- Chen, G., Kautz, S., & Zajac, F. (2001). Simulation analysis of muscle activity changes with altered body orientations during pedaling. *Journal of Biomechanics*, *34*(6), 749–756. DOI: 10.1016/S0021-9290(01)00014-8
- Delp, S. L., Loan, J. P., Hoy, M. G., Zajac, F. E., Topp, E. L., & Rosen, J. M. (1990). An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Transactions on Biomedical Engineering*, *37*(8), 757–767. DOI: 10.1109/10.102791
- Donatelli, R. A. (1996). *The biomechanics of the foot and ankle* (2nd ed.). Philadelphia, PA: D. A. Davis.
- Erdemir, A., McLean, S., Herzog, W., & van den Bogert, A. J. (2007). Model-based estimation of muscle forces exerted during movements. *Clinical Biomechanics*, *22*(2), 131–154. DOI: 10.1016/j.clinbiomech.2006.09.005
- Ericson, M., Ekholm, J., Svensson, O., & Nisell, R. (1985). The forces of ankle joint structures during ergometer cycling. *Foot & Ankle*, *6*(3), 135–142.
- Fath, F., Blazevich, A. J., Waugh, C. M., Miller, S. C., & Korff, T. (2010). Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans. *Journal of Applied Physiology*, *109*, 1644–1652. DOI: 10.1152/jappphysiol.00656.2010
- Finni, T., Komi, P. V., & Lukkariniemi, J. (1998). Achilles tendon loading during walking: application of a novel optic fiber technique. *European Journal of Applied Physiology*, *77*(3), 289–291. DOI: 10.1007/s004210050335

- Fregly, B. J., & Zajac, F. E. (1996). A state-space analysis of mechanical energy generation, absorption, and transfer during pedaling. *Journal of Biomechanics*, *29*(1), 81–90. DOI: 10.1016/0021-9290(95)00011-9
- Fukashiro, S., Hay, D. C., & Nagano, A. (2006). Biomechanical behavior of muscle-tendon complex during dynamic human movements. *Journal of Applied Biomechanics*, *22*(2), 131–147.
- Fukashiro, S., Komi, P., Järvinen, M., & Miyashita, M. (1993). Comparison between the directly measured achilles tendon force and the tendon force calculated from the ankle joint moment during vertical jumps. *Clinical Biomechanics*, *8*(1), 25–30. DOI: 10.1016/S0268-0033(05)80006-3
- Fukashiro, S., Rob, M., Ichinose, Y., Kawakami, Y., & Fukunaga, T. (1995). Ultrasonography gives directly but noninvasively elastic characteristic of human tendon in vivo. *European Journal of Applied Physiology and Occupational Physiology*, *71*(6), 555–557. DOI: 10.1007/BF00238560
- Fukunaga, T., Ito, M., Ichinose, Y., Kuno, S., Kawakami, Y., & Fukashiro, S. (1996). Tendinous movement of a human muscle during voluntary contractions determined by real-time ultrasonography. *Journal of Applied Physiology*, *81*(3), 1430–1433.
- Fukunaga, T., Roy, R. R., Shellock, F. G., Hodgson, J. A., & Edgerton, V. R. (1996). Specific tension of human plantar flexors and dorsiflexors. *Journal of Applied Physiology*, *80*(1), 158–165.
- Gregor, R., Komi, P., Browning, R., & Järvinen, M. (1991). A comparison of the triceps surae and residual muscle moments at the ankle during cycling. *Journal of Biomechanics*, *24*(5), 287–297. DOI: 10.1016/0021-9290(91)90347-P
- Gregor, R., Komi, P., & Järvinen, M. (1987). Achilles Tendon Forces During Cycling. *International Journal of Sports Medicine*, *8*(S 1), S9–S14. DOI: 10.1055/s-2008-1025698
- Grieve, D., Pheasant, S., & Cavanagh, P. (1978). Prediction of gastrocnemius length from knee and ankle joint posture. In E. Asmussen & K. Jorgensen (Eds.), *Biomechanics-VI-A, International Series on Biomechanics* (pp. 405–412). Baltimore: University Park Press.
- Hashizume, S., Iwanuma, S., Akagi, R., Kanehisa, H., Kawakami, Y., & Yanai, T. (2011). In vivo determination of the Achilles tendon moment arm in three-dimensions. *Journal of Biomechanics*, 1–5. DOI: 10.1016/j.jbiomech.2011.10.018
- Hawkins, D., Lum, C., Gaydos, D., & Dunning, R. (2009). Dynamic creep and preconditioning of the Achilles tendon in-vivo. *Journal of Biomechanics*, *42*(16), 2813–2817. DOI: 10.1016/j.jbiomech.2009.08.023
- Herzog, W., Read, L., & Keurs, H. ter. (1991). Experimental determination of force-length relations of intact human gastrocnemius muscles. *Clinical Biomechanics*, *6*(4), 230–

238. DOI: 10.1016/0268-0033(91)90051-Q
- Hicks, J. H. (1953). The mechanics of the foot. I. The joints. *Journal of Anatomy*, 87(4), 345–357.
- Hintermann, B., Nigg, B. M., & Sommer, C. (1994). Foot movement and tendon excursion: an in vitro study. *Foot & Ankle International*, 15(7), 386–395.
- Hoy, M. G., Zajac, F. E., & Gordon, M. E. (1990). A musculoskeletal model of the human lower extremity: The effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle. *Journal of Biomechanics*, 23(2), 157–169. DOI: 10.1016/0021-9290(90)90349-8
- Hull, M., & Jorge, M. (1985). A method for biomechanical analysis of bicycle pedalling. *Journal of Biomechanics*, 18(9), 631–644. DOI: 10.1016/0021-9290(85)90019-3
- Huxley, A. F. (1957). Muscle structure and theories of contraction. *Progress in Biophysics and Biophysical Chemistry*, 7, 255–318.
- Isman, R. E., & Inman, V. T. (1969). Anthropometric studies of the human foot and ankle. *Bulletin of Prosthetics Research*, 11, 97–129.
- Ito, M., Akima, H., & Fukunaga, T. (2000). In vivo moment arm determination using B-mode ultrasonography. *Journal of Biomechanics*, 33(2), 215–218. DOI: 10.1016/S0021-9290(99)00154-2
- Jensen, R. K. (1989). Changes in segment inertia proportions between 4 and 20 years. *Journal of Biomechanics*, 22(6-7), 529–536. DOI: 10.1016/0021-9290(89)90004-3
- Karamanidis, K., Albracht, K., Braunstein, B., Catala, M. M., Goldmann, J.-P., & Brüggemann, G.-P. (2011). Lower leg musculoskeletal geometry and sprint performance. *Gait & Posture*, 34(1), 138–141. DOI: 10.1016/j.gaitpost.2011.03.009
- Karamanidis, K., Stafilidis, S., DeMonte, G., Morey-Klapsing, G., Brüggemann, G.-P., & Arampatzis, A. (2005). Inevitable joint angular rotation affects muscle architecture during isometric contraction. *Journal of Electromyography and Kinesiology*, 15(6), 608–616. DOI: 10.1016/j.jelekin.2005.02.001
- Kay, A., & Blazevich, A. (2008). Reductions in active plantarflexor moment are significantly correlated with static stretch duration. *European Journal of Sport Science*, 8(1), 41–46. DOI: 10.1080/17461390701855505
- Kellis, E., & Baltzopoulos, V. (1999). In vivo determination of the patella tendon and hamstrings moment arms in adult males using videofluoroscopy during submaximal knee extension and flexion. *Clinical Biomechanics*, 14(2), 118–124.
- Klein, P. (1996). Moment arm length variations of selected muscles acting on talocrural and subtalar joints during movement: An In vitro study. *Journal of Biomechanics*, 29(1), 21–30. DOI: 10.1016/0021-9290(95)00025-9
- Komi, P., Salonen, M., Järvinen, M., & Kokko, O. (1987). In Vivo Registration of Achilles Tendon Forces in Man. *International Journal of Sports Medicine*, 08(S 1), S3–S8.

DOI: 10.1055/s-2008-1025697

- Korff, T., Hunter, E. L., & Martin, J. C. (2009). Muscular and non-muscular contributions to maximum power cycling in children and adults: implications for developmental motor control. *The Journal of Experimental Biology*, *212*(Pt 5), 599–603. DOI: 10.1242/jeb.020180
- Korff, T., & Jensen, J. L. (2007). Age-related differences in adaptation during childhood: the influences of muscular power production and segmental energy flow caused by muscles. *Experimental Brain Research*, *177*(3), 291–303. DOI: 10.1007/s00221-006-0684-3
- Lear dini, A., OConnor, J., Catani, F., & Giannini, S. (1999). Kinematics of the human ankle complex in passive flexion; a single degree of freedom system. *Journal of Biomechanics*, *32*(2), 111–118. DOI: 10.1016/S0021-9290(98)00157-2
- Lee, S. S. M., Lewis, G. S., & Piazza, S. J. (2008). An algorithm for automated analysis of ultrasound images to measure tendon excursion in vivo. *Journal of Applied Biomechanics*, *24*(1), 75–82.
- Lee, S. S. M., & Piazza, S. J. (2009). Built for speed: musculoskeletal structure and sprinting ability. *The Journal of Experimental Biology*, *212*(Pt 22), 3700–3707. DOI: 10.1242/jeb.031096
- Lundberg, A., & Svensson, O. (1993). The axes of rotation of the talocalcaneal and talonavicular joints. *The Foot*, *3*(2), 65–70. DOI: 10.1016/0958-2592(93)90064-A
- Lundberg, A., Svensson, O. K., Németh, G., & Selvik, G. (1989). The axis of rotation of the ankle joint. *The Journal of Bone and Joint Surgery*, *71*(1), 94–99.
- Maganaris, C. N. (2000). In vivo measurement-based estimations of the moment arm in the human tibialis anterior muscle-tendon unit. *Journal of Biomechanics*, *33*(3), 375–379. DOI: 10.1016/S0021-9290(99)00188-8
- Maganaris, C. N. (2003a). Force-length characteristics of the in vivo human gastrocnemius muscle. *Clinical Anatomy*, *16*(3), 215–223. DOI: 10.1002/ca.10064
- Maganaris, C. N. (2003b). Tendon conditioning: artefact or property? *Proceedings. Biological sciences / The Royal Society*, *270* Suppl, S39–42. DOI: 10.1098/rsbl.2003.0004
- Maganaris, C. N. (2004). Imaging-based estimates of moment arm length in intact human muscle-tendons. *European Journal of Applied Physiology*, *91*(2-3), 130–139. DOI: 10.1007/s00421-003-1033-x
- Maganaris, C. N. (2005). Validity of procedures involved in ultrasound-based measurement of human plantarflexor tendon elongation on contraction. *Journal of Biomechanics*, *38*(1), 9–13. DOI: 10.1016/j.jbiomech.2004.03.024
- Maganaris, C. N., Baltzopoulos, V., Ball, D., & Sargeant, A. J. (2001). In vivo specific tension of human skeletal muscle. *Journal of Applied Physiology*, *90*(3), 865–872.

- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1998a). Changes in Achilles tendon moment arm from rest to maximum isometric plantarflexion: in vivo observations in man. *The Journal of Physiology*, *510*(3), 977–985. DOI: 10.1111/j.1469-7793.1998.977bj.x
- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1998b). Differences in human antagonistic ankle dorsiflexor coactivation between legs; can they explain the moment deficit in the weaker plantarflexor leg? *Experimental Physiology*, *83*(6), 843–855.
- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1998c). In vivo measurements of the triceps surae complex architecture in man: implications for muscle function. *The Journal of Physiology*, *512*(2), 603–614. DOI: 10.1111/j.1469-7793.1998.603be.x
- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1999). Changes in the tibialis anterior tendon moment arm from rest to maximum isometric dorsiflexion: in vivo observations in man. *Clinical Biomechanics*, *14*(9), 661–666. DOI: 10.1016/S0268-0033(99)00018-2
- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (2000). In vivo measurement-based estimations of the human Achilles tendon moment arm. *European Journal of Applied Physiology*, *83*(4-5), 363–369. DOI: 10.1007/s004210000247
- Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (2006). Human calf muscle responses during repeated isometric plantarflexions. *Journal of Biomechanics*, *39*(7), 1249–1255. DOI: 10.1016/j.jbiomech.2005.03.011
- Maganaris, C. N., & Paul, J. (2000). Load-elongation characteristics of in vivo human tendon and aponeurosis. *Journal of Experimental Biology*, *203*(4), 751–756.
- Maganaris, C. N., & Paul, J. P. (1999). In vivo human tendon mechanical properties. *The Journal of Physiology*, *521*(1), 307–313. DOI: 10.1111/j.1469-7793.1999.00307.x
- Magnusson, S. P., Aagaard, P., Rosager, S., Dyhre-Poulsen, P., & Kjaer, M. (2001). Load-displacement properties of the human triceps surae aponeurosis in vivo. *The Journal of Physiology*, *531*(1), 277–288. DOI: 10.1111/j.1469-7793.2001.0277j.x
- Manal, K., Cowder, J. D., & Buchanan, T. S. (2010). A hybrid method for computing Achilles tendon moment arm using ultrasound and motion analysis. *Journal of Applied Biomechanics*, *26*(2), 224–228.
- Manal, K., Gravare-Silbernagel, K., & Buchanan, T. S. (2011). A real-time EMG-driven musculoskeletal model of the ankle. *Multibody System Dynamics*. DOI: 10.1007/s11044-011-9285-4
- Martin, J. C., Farrar, R. P., Wagner, B. M., & Spirduso, W. W. (2000). Maximal Power Across the Lifespan. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *55*(6), M311–M316. DOI: 10.1093/gerona/55.6.M311

- Martin, J. C., & Spirduso, W. (2001). Determinants of maximal cycling power: crank length, pedaling rate and pedal speed. *European Journal of Applied Physiology*, 84(5), 413–418. DOI: 10.1007/s004210100400
- Menegaldo, L. L., & Oliveira, L. F. de. (2009). Effect of muscle model parameter scaling for isometric plantar flexion torque prediction. *Journal of Biomechanics*, 42(15), 2597–2601. DOI: 10.1016/j.jbiomech.2009.06.043
- Miller, J. A., & Melissa Gross, M. (1998). Locomotor advantages of Neandertal skeletal morphology at the knee and ankle. *Journal of Biomechanics*, 31(4), 355–361. DOI: 10.1016/S0021-9290(98)00031-1
- Morse, C. I., Thom, J. M., Mian, O. S., Birch, K. M., & Narici, M. V. (2007). Gastrocnemius specific force is increased in elderly males following a 12-month physical training programme. *European Journal of Applied Physiology*, 100(5), 563–570. DOI: 10.1007/s00421-006-0246-1
- Morse, C. I., Tolfrey, K., Thom, J. M., Vassilopoulos, V., Maganaris, C. N., & Narici, M. V. (2008). Gastrocnemius muscle specific force in boys and men. *Journal of Applied Physiology*, 104(2), 469–474. DOI: 10.1152/jappphysiol.00697.2007
- Muramatsu, T., Muraoka, T., Takeshita, D., Kawakami, Y., Hirano, Y., & Fukunaga, T. (2001). Mechanical properties of tendon and aponeurosis of human gastrocnemius muscle in vivo. *Journal of Applied Physiology*, 90(5), 1671–1678.
- Murray, W. M., Delp, S. L., & Buchanan, T. S. (1995). Variation of muscle moment arms with elbow and forearm position. *Journal of Biomechanics*, 28(5), 513–525. DOI: 10.1016/0021-9290(94)00114-J
- Nagano, A., Komura, T., Himeno, R., & Fukashiro, S. (2003). Optimal Digital Filter Cutoff Frequency of Jumping Kinematics Evaluated Through Computer Simulation. *International Journal of Sport and Health Science*, 1(2), 196–201. DOI: 10.5432/ijshs.1.196
- Narici, M. V., Binzoni, T., Hiltbrand, E., Fasel, J., Terrier, F., & Cerretelli, P. (1996). In vivo human gastrocnemius architecture with changing joint angle at rest and during graded isometric contraction. *The Journal of Physiology*, 496(Pt 1), 287–297.
- Neptune, R. R. (1999). Optimization Algorithm Performance in Determining Optimal Controls in Human Movement Analyses. *Journal of Biomechanical Engineering*, 121(2), 249–252. DOI: 10.1115/1.2835111
- Neptune, R. R., & Hull, M. (1995). Accuracy assessment of methods for determining hip movement in seated cycling. *Journal of Biomechanics*, 28(4), 423–437. DOI: 10.1016/0021-9290(94)00080-N
- Neptune, R. R., & Hull, M. L. (1998). Evaluation of Performance Criteria for Simulation of Submaximal Steady-State Cycling Using a Forward Dynamic Model. *Journal of Biomechanical Engineering*, 120(3), 334–341. DOI: 10.1115/1.2797999

- Neptune, R. R., Kautz, S., & Hull, M. (1997). The effect of pedaling rate on coordination in cycling. *Journal of Biomechanics*, *30*(10), 1051–1058. DOI: 10.1016/S0021-9290(97)00071-7
- O'Brien, T. D., Reeves, N. D., Baltzopoulos, V., Jones, D. A., & Maganaris, C. N. (2009). Moment arms of the knee extensor mechanism in children and adults. *Journal of Anatomy*, *215*(2), 198–205. DOI: 10.1111/j.1469-7580.2009.01088.x
- O'Brien, T. D., Reeves, N. D., Baltzopoulos, V., Jones, D. A., & Maganaris, C. N. (2010). Mechanical properties of the patellar tendon in adults and children. *Journal of Biomechanics*, *43*(6), 1190–1195. DOI: 10.1016/j.jbiomech.2009.11.028
- Orishimo, K. F., Burstein, G., Mullaney, M. J., Kremenec, I. J., Nesse, M., McHugh, M. P., et al. (2008). Effect of knee flexion angle on Achilles tendon force and ankle joint plantarflexion moment during passive dorsiflexion. *The Journal of Foot and Ankle Surgery*, *47*(1), 34–39. DOI: 10.1053/j.jfas.2007.10.008
- Pandy, M. G. (1999). Moment arm of a muscle force. *Exercise and Sport Sciences Reviews*, *27*, 79–118.
- Pandy, M. G., & Zajac, F. E. (1991). Optimal muscular coordination strategies for jumping. *Journal of Biomechanics*, *24*(1), 1–10. DOI: 10.1016/0021-9290(91)90321-D
- Panjabi, M. M. (1979). Centers and angles of rotation of body joints: A study of errors and optimization. *Journal of Biomechanics*, *12*(12), 911–920. DOI: 10.1016/0021-9290(79)90059-9
- Panjabi, M. M., Goel, V. K., & Walter, S. (1982). Errors in kinematic parameters of a planar joint: Guidelines for optimal experimental design. *Journal of Biomechanics*, *15*(7), 537–544. DOI: 10.1016/0021-9290(82)90007-0
- Piazza, S. J., & Delp, S. L. (1996). The influence of muscles on knee flexion during the swing phase of gait. *Journal of Biomechanics*, *29*(6), 723–733. DOI: 10.1016/0021-9290(95)00144-1
- Raasch, C. C., Zajac, F. E., Ma, B., & Levine, W. S. (1997). Muscle coordination of maximum-speed pedaling. *Journal of Biomechanics*, *30*(6), 595–602. DOI: 10.1016/S0021-9290(96)00188-1
- Raichlen, D. A., Armstrong, H., & Lieberman, D. E. (2011). Calcaneus length determines running economy: implications for endurance running performance in modern humans and Neandertals. *Journal of Human Evolution*, *60*(3), 299–308. DOI: 10.1016/j.jhevol.2010.11.002
- Redl, C., Gfoehler, M., & Pandy, M. G. (2007). Sensitivity of muscle force estimates to variations in muscle-tendon properties. *Human Movement Science*, *26*(2), 306–319. DOI: 10.1016/j.humov.2007.01.008
- Reuleaux, F. (1875). *Theoretische Kinematik: Grundzüge einer Theorie des Maschinewe-*

- sens.* Braunschweig: Vieweg.
- Riemann, B., DeMont, R., Ryu, K., & Lephart, S. (2001). The effects of sex, joint angle, and the gastrocnemius muscle on passive ankle joint complex stiffness. *Journal of Athletic Training*, *36*(4), 369–377.
- Rosager, S., Aagaard, P., Dyhre-Poulsen, P., Neergaard, K., Kjaer, M., & Magnusson, S. P. (2002). Load-displacement properties of the human triceps surae aponeurosis and tendon in runners and non-runners. *Scandinavian Journal of Medicine and Science in Sports*, *12*(2), 90–98. DOI: 10.1034/j.1600-0838.2002.120205.x
- Rugg, S., Gregor, R., Mandelbaum, B., & Chiu, L. (1990). In vivo moment arm calculations at the ankle using magnetic resonance imaging (MRI). *Journal of Biomechanics*, *23*(5), 495–501. DOI: 10.1016/0021-9290(90)90305-M
- Sammarco, G. J., Burstein, A. H., & Frankel, V. H. (1973). Biomechanics of the ankle: a kinematic study. *The Orthopedic Clinics of North America*, *4*(1), 75–96.
- Schatzmann, L., Brunner, P., & Stäubli, H. (1998). Effect of cyclic preconditioning on the tensile properties of human quadriceps tendons and patellar ligaments. *Knee Surgery, Sports Traumatology, Arthroscopy*, *6*, S56–S61. DOI: 10.1007/s001670050224
- Scholz, M. N., Bobbert, M. F., van Soest, A. J., Clark, J. R., & van Heerden, J. (2008). Running biomechanics: shorter heels, better economy. *The Journal of Experimental Biology*, *211*(Pt 20), 3266–3271. DOI: 10.1242/jeb.018812
- Sheehan, F. T. (2007). The 3D patellar tendon moment arm: quantified in vivo during volitional activity. *Journal of Biomechanics*, *40*(9), 1968–1974. DOI: 10.1016/j.jbiomech.2006.09.029
- Sheehan, F. T. (2012). The 3D in vivo Achilles' tendon moment arm, quantified during active muscle control and compared across sexes. *Journal of Biomechanics*, *45*(2), 225–230. DOI: 10.1016/j.jbiomech.2011.11.001
- Smidt, G. L. (1973). Biomechanical analysis of knee flexion and extension. *Journal of Biomechanics*, *6*(1), 79–92. DOI: 10.1016/0021-9290(73)90040-7
- Spiegelman, J. J., & Woo, S. L.-Y. (1987). A rigid-body method for finding centers of rotation and angular displacements of planar joint motion. *Journal of Biomechanics*, *20*(7), 715–721. DOI: 10.1016/0021-9290(87)90037-6
- Spoor, C., & van Leeuwen, J. (1992). Knee muscle moment arms from MRI and from tendon travel. *Journal of Biomechanics*, *25*(2), 201–206. DOI: 10.1016/0021-9290(92)90276-7
- Spoor, C., van Leeuwen, J., Meskers, C., Titulaer, A., & Huson, A. (1990). Estimation of instantaneous moment arms of lower-leg muscles. *Journal of Biomechanics*, *23*(12), 1247–1259. DOI: 10.1016/0021-9290(90)90382-D
- Stienen, A. H. A., Schouten, A. C., Schuurmans, J., & van Der Helm, F. C. T. (2007). Analysis of reflex modulation with a biologically realistic neural network. *Journal of*

- Computational Neuroscience*, 23(3), 333–348. DOI: 10.1007/s10827-007-0037-7
- Storage, A., & Wolf, B. (1979). Functional analysis of the role of the finger tendons. *Journal of Biomechanics*, 12(8), 575–578. DOI: 10.1016/0021-9290(79)90076-9
- Thelen, D. G., & Anderson, F. C. (2006). Using computed muscle control to generate forward dynamic simulations of human walking from experimental data. *Journal of Biomechanics*, 39(6), 1107–1115. DOI: 10.1016/j.jbiomech.2005.02.010
- Tsaopoulos, D. E., Baltzopoulos, V., Richards, P. J., & Maganaris, C. N. (2007). In vivo changes in the human patellar tendon moment arm length with different modes and intensities of muscle contraction. *Journal of Biomechanics*, 40(15), 3325–3332. DOI: 10.1016/j.jbiomech.2007.05.005
- Tsaopoulos, D. E., Baltzopoulos, V., Richards, P. J., & Maganaris, C. N. (2009). A comparison of different two-dimensional approaches for the determination of the patellar tendon moment arm length. *European Journal of Applied Physiology*, 105(5), 809–814. DOI: 10.1007/s00421-008-0968-3
- van den Bogert, A. J., Smith, G. D., & Nigg, B. M. (1994). In vivo determination of the anatomical axes of the ankle joint complex: An optimization approach. *Journal of Biomechanics*, 27(12), 1477–1488. DOI: 10.1016/0021-9290(94)90197-X
- Visser, J. J., Hoogkamer, J. E., Bobbert, M. F., & Huijing, P. A. (1990). Length and moment arm of human leg muscles as a function of knee and hip-joint angles. *European Journal of Applied Physiology and Occupational Physiology*, 61(5-6), 453–460. DOI: 10.1007/BF00236067
- Voigt, M., Bojsen-Møller, F., Simonsen, E., & Dyhre-Poulsen, P. (1995). The influence of tendon young's modulus, dimensions and instantaneous moment arms on the efficiency of human movement. *Journal of Biomechanics*, 28(3), 281–291. DOI: 10.1016/0021-9290(94)00071-B
- Waugh, C. M., Blazeovich, A. J., Fath, F., & Korff, T. (2011). Can Achilles tendon moment arm be predicted from anthropometric measures in pre-pubescent children? *Journal of Biomechanics*, 44(10), 1839–1844. DOI: 10.1016/j.jbiomech.2011.03.023
- Waugh, C. M., Blazeovich, A. J., Fath, F., & Korff, T. (2012). Age-related changes in mechanical properties of the Achilles tendon. *Journal of Anatomy*, 220(2), 144–155. DOI: 10.1111/j.1469-7580.2011.01461.x
- Wickiewicz, T. L., Roy, R. R., Powell, P. L., & Edgerton, V. R. (1983). Muscle architecture of the human lower limb. *Clinical Orthopaedics and Related Research*, 179, 275–283.
- Wilson, N. A., & Sheehan, F. T. (2009). Dynamic in vivo 3-dimensional moment arms of the individual quadriceps components. *Journal of Biomechanics*, 42(12), 1891–1897. DOI: 10.1016/j.jbiomech.2009.05.011
- Winter, D. (2005). *Biomechanics and motor control of human movement* (3rd ed.). Hoboken, NJ: John Wiley & Sons.

- Winter, D. (2009). *Biomechanics and motor control of human movement* (4th ed.). Hoboken, NJ: John Wiley & Sons.
- Woltring, H. J. (1986). A FORTRAN package for generalized, cross-validatorspline smoothing and differentiation. *Advances in Engineering Software*, 8(2), 104–113.
- Wood, L., Dixon, S., Grant, C., & Armstrong, N. (2006). Elbow flexor strength, muscle size, and moment arms in prepubertal boys and girls. *Pediatric Exercise Science*, 18(4), 457–469.
- Wu, J. Z., An, K.-N., Cutlip, R. G., Andrew, M. E., & Dong, R. G. (2009). Modeling of the muscle/tendon excursions and moment arms in the thumb using the commercial software anybody. *Journal of Biomechanics*, 42(3), 383–388. DOI: 10.1016/j.jbiomech.2008.11.008
- Xiao, M., & Higginson, J. (2010). Sensitivity of estimated muscle force in forward simulation of normal walking. *Journal of Applied Biomechanics*, 26(2), 142–149.
- Yamaguchi, G., Sawa, A., Moran, D., Fessler, M., & Winters, J. (1990). A survey of human musculotendon actuator parameters. In J. Winters & S. L.-Y. Woo (Eds.), *Multiple muscle systems: Biomechanics and movement organization*. (pp. 717–773). New York: Springer-Verlag.
- Zajac, F. E. (1989). Muscle and tendon: Properties, models, scaling, and application to biomechanics and motor control. *Critical Reviews in Biomedical Engineering*, 17, 359–411.
- Zajac, F. E., Neptune, R. R., & Kautz, S. A. (2002). Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. *Gait & Posture*, 16(3), 215–232.
- Zajac, F. E., Neptune, R. R., & Kautz, S. A. (2003). Biomechanics and muscle coordination of human walking: part II: lessons from dynamical simulations and clinical implications. *Gait & Posture*, 17(1), 1–17.

# Appendices

# A Appendix

## A.1 Additional information

### A.1.1 Influence of knee angle on Achilles tendon moment arm during contraction state

The following linear relationship was used to account for changes in Achilles tendon moment arm ( $MA_{AT}$ ) due to changes in knee angle during contraction state (maximal voluntary contraction). In chapter 4 of this thesis it was reported that  $MA_{AT}$  increased by 4.5% on average at an knee angle of  $180^\circ$  from rest to MVC. In contrast, Maganaris et al. (1998a) reported an increase of 25% at an knee angle of  $90^\circ$  (Figure A.1). Based on these results, the estimation of  $MA_{AT}$  during contraction state which changed as a function of knee angle was based on the following linear relationship (Note: The linear equation has to be multiplied with the corresponding  $MA_{AT}$  at rest; the result has to be divided by 100 as the increase in  $MA_{AT}$  is expressed as a percentage):

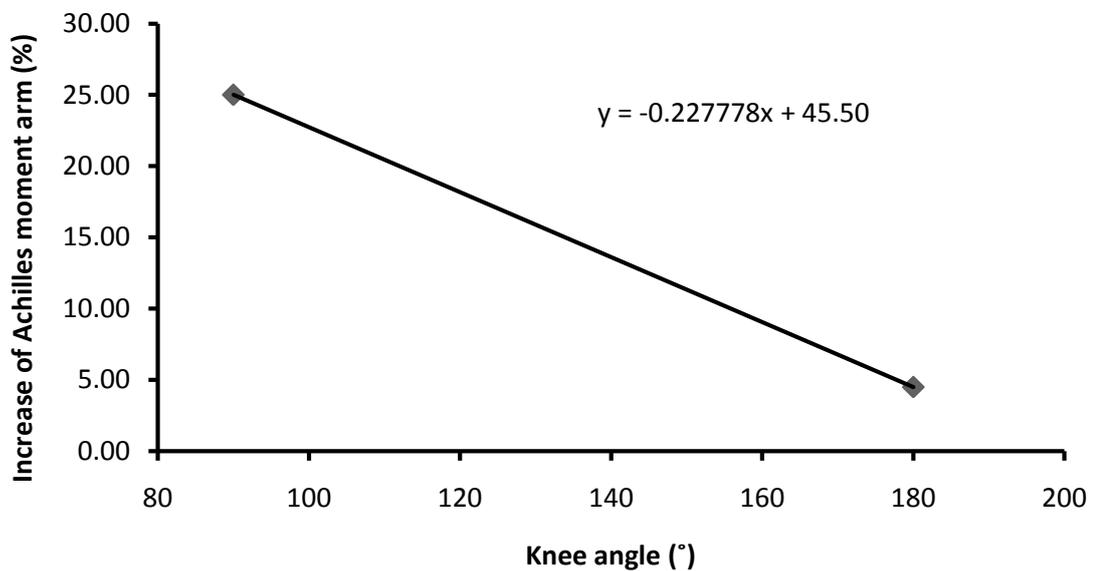


Figure A.1. Graph showing the relationship between knee angle and the percentage increase in Achilles tendon moment arm from rest to a maximal voluntary contraction.

### A.1.2 Residual analysis

The cutoff frequencies for each of the digital filters used in this thesis were determined using the residual analysis as described by Winter (2009). In short, the residual between (see equation A.1) the filtered and unfiltered signal was plotted as a function of the filter cutoff frequency.

$$R(f_c) = \sqrt{\frac{1}{N} \sum_{i=1}^N (X_i - \hat{X}_i)^2} \quad (\text{A.1})$$

where

$f_c$  = is the cutoff frequency of the filter,

$X_i$  = are the raw data at  $i$ th sample,

$\hat{X}_i$  = is the filtered signal at the  $i$ th sample.

The residuals of a custom-built Matlab program were determined until half the sampling frequency ( $0.5 f_s$ ). When the data only consisted of random noise, the residual were represented by a straight line decreasing from an intercept at 0 Hz to an intercept at the abscissa at  $0.5 f_s$ . However, when the data consisted of random noise as well as a true signal, the residual rose above the straight random noise line as the cutoff frequency was reduced (Winter, 2009). As the choice of the cutoff frequency was a compromise between the signal distortion and the level of noise which was acceptable, the cutoff frequency at which the true signal started to rise ( $d$ ) was individually selected. When using the Matlab program, this cutoff frequency was visually selected from the residual plot. Based on this choice, the cutoff frequency for the filter was determined by the program by creating the straight line representing the random noise from  $d$  to  $0.5 f_s$  ( $e$ ). A horizontal line from the intercept of the ordinate ( $a$ ) with the signal curve was drawn. The resulting intercept ( $d$ ) provided the suggested cutoff of the filter (see Figures A.2 and A.3).

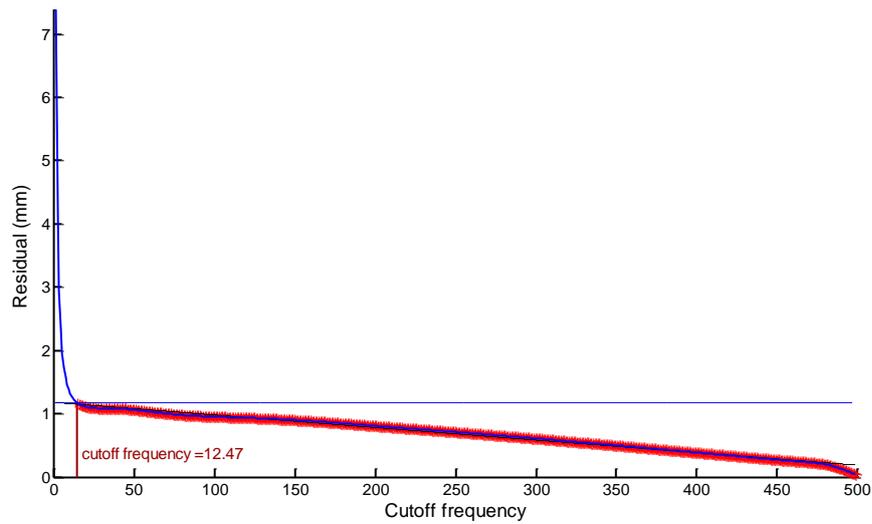


Figure A.2. Plot of the residual between a filtered and an unfiltered torque signal which changes as a function of the 4<sup>th</sup>-order Butterworth filter cutoff frequencies (until 500 Hz)

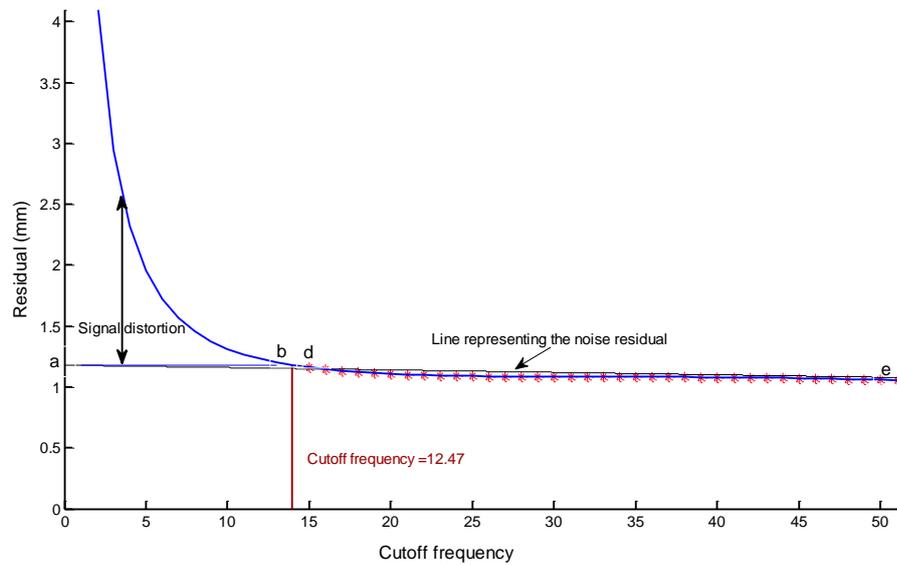


Figure A.3. Illustration of the estimation of the cutoff frequency as shown in Figure A.2 (zoomed in). Line  $d$  to  $e$  represents the residual of noise;  $a$  represents the intersect of line  $de$  with 0 Hz;  $b$  represents the intersect of a horizontal line from  $a$  to the true signal which rises from line  $de$ . The x-coordinate of point  $b$  represents the final cutoff frequency of the filter.

### A.1.3 Free-body diagram of the foot segment during cycling

The following figure (see FigureA.4) is an example of the cycling free-body diagram that was used during the inverse dynamics approach in chapter 5.

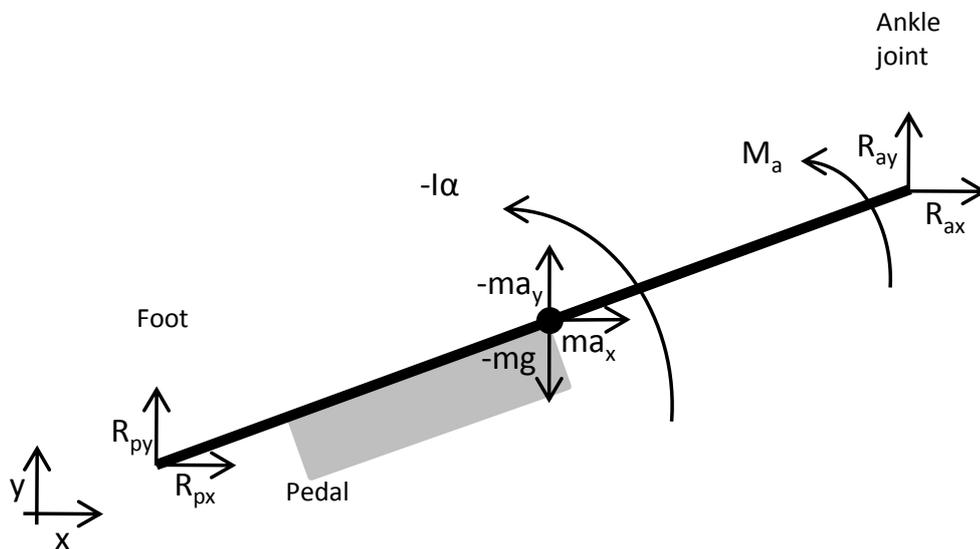
$ma_x, ma_y, mg$  = acceleration of segment of centre of mass

$I\alpha$  = angular acceleration of segment in plane of movement

$R_{px}, R_{py}$  = pedal reaction forces at the distal end of the foot segment

$R_{ax}, R_{ay}$  = ankle joint reaction forces at the proximal end of the foot segment

$M_a$  = net muscle torque acting at the ankle joint



*Figure A.4.* Free-body diagram of the foot segment constraint to the pedal showing reaction and gravitational forces, ankle joint net torque and all linear and angular accelerations. The right foot was assumed to be a rigid segment as well as being stationary with respect to the pedal.

## A.2 Ethical approval

Head of School of Sport & Education  
Professor Susan Capel

Stuart Miller  
Research Student  
School of Sport and Education  
Brunel University

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1<sup>st</sup> May 2008

Dear Stuart

**RE71- 07 – Validation of Reuleaux's method using motion analysis**

I am writing to confirm the Research Ethics Committee of the School of Sport and Education received your application connected to the above project. Your application has been independently reviewed and I am pleased to confirm your application complies with the research ethics guidelines issued by the University.

On behalf of the Research Ethics Committee, I wish you every success with your study.

Yours sincerely



Dr Simon Bradford  
Chair of Research Ethics Committee

Head of School of Sport & Education  
Professor Susan Capel

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12<sup>th</sup> August 2009

Dear Stuart

**RE71-07 – Validation of Reuleaux's method using motion analysis**

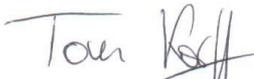
I am writing to confirm the Research Ethics Committee of the School of Sport and Education received your application to amend the above mentioned research study. Your application has been independently reviewed to ensure it complies with the University Research Ethics requirements and guidelines.

The Chair, acting under delegated authority, is satisfied with the decision reached by the independent reviewers and is pleased to confirm there is no objection on ethical grounds to the proposed amendment to your study.

Any further changes to the protocol contained within your application and any unforeseen ethical issues which arise during the conduct of your study must be notified to the Research Ethics Committee for further consideration.

On behalf of the Research Ethics Committee for the School of Sport and Education, I wish you every success with your amended study.

Yours sincerely



Signed on behalf of Dr Simon Bradford  
**Chair of Research Ethics Committee**  
School Of Sport and Education

Head of School of Sport & Education  
Professor Susan Capel

**Brunel**  
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15<sup>th</sup> June 2008

Dear Florian

**RE70-07 – Age-related Differences in Tendon Stiffness and the Effect on Force Production**

I am writing to confirm the Research Ethics Committee of the School of Sport and Education received your application connected to the above project. Your application has been independently reviewed and I am pleased to confirm your application complies with the research ethics guidelines issued by the University.

On behalf of the Research Ethics Committee, I wish you every success with your study.

Yours sincerely



Dr Simon Bradford  
**Chair of Research Ethics Committee**

Head of School of Sport & Education  
Professor Susan Capel

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Florian Fath  
PhD (Sport Science) Student  
School of Sport and Education  
Brunel University

4<sup>th</sup> August 2010

Dear Florian

**RE48-09 – The effect of changes in muscle-tendon stiffness on the neural activation during cycling**

I am writing to confirm the Research Ethics Committee of the School of Sport and Education received your application connected to the above mentioned research study. Your application has been independently reviewed to ensure it complies with the University/School Research Ethics requirements and guidelines.

The Chair, acting under delegated authority, is satisfied with the decision reached by the independent reviewers and is pleased to confirm there is no objection on ethical grounds to the proposed study.

Any changes to the protocol contained within your application and any unforeseen ethical issues which arise during the conduct of your study must be notified to the Research Ethics Committee for further consideration.

On behalf of the Research Ethics Committee for the School of Sport and Education, I wish you every success with your study.

Yours sincerely



Dr Gary Armstrong  
**Chair of Research Ethics Committee**  
School Of Sport and Education

## A.3 Participants health questionnaire

### Brunel's Standard Health Assessment & Consent Form PRE-PARTICIPATION HEALTH CHECK QUESTIONNAIRE

Health and safety is of paramount importance. For this reason we need to be aware of your current health status before you begin any testing procedures. The questions below are designed to identify whether you are able to participate now or should obtain medical advice before undertaking this investigation. Whilst every care will be given to the best of the investigators ability, an individual must know his/her limitations.

Subject name: ..... DOB: .....

Doctors Surgery Address: .....

Emergency Contact Name & Number: .....

Please answer the following questions:

	YES	NO
1. Has your doctor ever diagnosed a heart condition or recommend only medically supervised exercise?	<input type="checkbox"/>	<input type="checkbox"/>
2. Do you suffer from chest pains, heart palpitations or tightness of the chest?	<input type="checkbox"/>	<input type="checkbox"/>
3. Do you have known high blood pressure? If yes, please give details (i.e. medication)	<input type="checkbox"/>	<input type="checkbox"/>
4. Do you have low blood pressure or often feel faint or have dizzy spells?	<input type="checkbox"/>	<input type="checkbox"/>
5. Do you have known hypercholesteremia?	<input type="checkbox"/>	<input type="checkbox"/>
6. Have you ever had any bone or joint problems, which could be aggravated by physical activity?	<input type="checkbox"/>	<input type="checkbox"/>
7. Do you suffer from diabetes? If yes, are you insulin dependent?	<input type="checkbox"/>	<input type="checkbox"/>
8. Do you suffer from any lung/chest problem, i.e. Asthma, bronchitis, emphysema?	<input type="checkbox"/>	<input type="checkbox"/>
9. Do you suffer from epilepsy? If yes, when was the last incident?	<input type="checkbox"/>	<input type="checkbox"/>
10. Are you taking any medication?	<input type="checkbox"/>	<input type="checkbox"/>
11. Have you had any injuries in the past? E.g. back problems, muscle strains etc...	<input type="checkbox"/>	<input type="checkbox"/>
12. Are you currently enrolled in any other studies?	<input type="checkbox"/>	<input type="checkbox"/>
13. I have recently participated in a blood donation program	<input type="checkbox"/>	<input type="checkbox"/>
14. Are you a smoker?	<input type="checkbox"/>	<input type="checkbox"/>
15. Do you exercise on a regular basis (at least 60 min a week)?	<input type="checkbox"/>	<input type="checkbox"/>
16. Describe your exercise routines (mode, frequency, intensity/speed, race times):	<input type="checkbox"/>	<input type="checkbox"/>

If you feel at all unwell because of a temporary illness such as a cold or fever please inform the investigator. Please note if your health status changes so that you would subsequently answer YES to any of the above questions, please notify the investigator immediately.

**I have read and fully understand this questionnaire. I confirm that to the best of my knowledge, the answers are correct and accurate. I know of no reasons why I should not participate in physical activity and this investigation and I understand I will be taking part at my own risk.**

Participant's name & signature: \_\_\_\_\_ Date: \_\_\_\_\_

Investigator's name & signature: \_\_\_\_\_ Date: \_\_\_\_\_

## A.4 Consent form

### CONSENT FORM

The participant should complete the whole of this sheet himself		
	Please tick the appropriate box	
	YES	NO
Have you read the Research Participant Information Sheet?	<input type="checkbox"/>	<input type="checkbox"/>
Have you had an opportunity to ask questions and discuss this study?	<input type="checkbox"/>	<input type="checkbox"/>
Have you received satisfactory answers to all your questions?	<input type="checkbox"/>	<input type="checkbox"/>
Who have you spoken to?		
Do you understand that you will not be referred to by name in any report concerning the study?	<input type="checkbox"/>	<input type="checkbox"/>
Do you understand that you are free to withdraw from the study: at any time	<input type="checkbox"/>	<input type="checkbox"/>
without having to give a reason for withdrawing?	<input type="checkbox"/>	<input type="checkbox"/>
(where relevant) without affecting your future employment as a member of staff of the University or your progression or assessment as a student of the University.	<input type="checkbox"/>	<input type="checkbox"/>
Do you agree to take part in this study?	<input type="checkbox"/>	<input type="checkbox"/>
<b>Signature of Research Participant:</b>		
Date:		
Name in capitals:		
<u>Witness statement</u>		
I am satisfied that the above-named has given informed consent.		
<b>Witnessed by:</b>		
Date:		
Name in capitals:		

## A.5 Conference abstract

Poster presentation at the 15th ECSS Congress 23-26 June 2010 Antalya/Turkey

### **Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MRI**

Florian Fath<sup>1</sup>, Anthony Blazeovich<sup>2</sup>, Charlie Waugh<sup>1</sup>, Stuart Miller<sup>1</sup>, & Thomas Korff<sup>1</sup>

<sup>1</sup> School of Sport and Education, Brunel University, UK

<sup>2</sup> School of Exercise, Edith Cowan University, WA

#### **Introduction**

The quantification of muscle forces during human movement is important in many contexts. In vivo muscle forces can be estimated by dividing the muscular torque about a joint by the moment arm of the corresponding muscle tendon unit. Two different methods have been proven useful to estimate the moment arm of the Achilles tendon with respect to the ankle joint. When employing the centre of rotation (COR) method, the COR is geometrically constructed using MRI images of the foot. The perpendicular distance to the Achilles tendon is then measured directly. When employing the tendon excursion (TE) method, the moment arm is calculated by the first derivative of the tendon excursion (which can be obtained from ultrasound images) with respect to the angular displacement of the corresponding joint. The purposes of this study were a) to compare the moment arms obtained from COR and TE methods and b) to determine the reliability of each method.

#### **Methods**

Achilles tendon moment arms of 9 participants were obtained from both COR and TE methods. Using MR images of the foot at different angles, the COR at the neutral foot position was geometrically constructed. The perpendicular line to the Achilles tendon was measured using imaging software (COR method). For the TE method, the ankle was passively rotated (10°/s) 3 times through its range of motion. Tendon displacement at the muscle tendon junction of the gastrocnemius medialis was measured using ultrasound imaging. Moment arms were obtained by differentiating the tendon displacement with respect to the corresponding angular excursion using different differentiation techniques. For each participant and each method, three separate measurements were taken.

#### **Results**

Moment arms obtained from the TE method correlated well with those obtained from the COR method, but they were dependent on the mathematical differentiation technique. The highest correlation between the two methods was found when the tendon displacement was differentiated over an angular interval of 20° ( $r = .99$ ,  $R^2 = .98$ ). The coefficients of variation across the three separate measurements were 2.34% and 3.93% for TE and COR methods, respectively.

**Discussion**

Our results show good agreement between the TE and COR methods of determining Achilles tendon moment arm. Furthermore, both methods are very reliable. These findings demonstrate that both methods are robust against their individual limitations. They thereby allow researchers to choose the method that is most appropriate and accessible within their individual contexts.

## A.6 Peer-reviewed publication

*J Appl Physiol* 109: 1644–1652, 2010.  
First published September 16, 2010; doi:10.1152/jappphysiol.00656.2010.

### Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans

Florian Fath,<sup>1</sup> Anthony J. Blazeovich,<sup>2</sup> Charlie M. Waugh,<sup>1</sup> Stuart C. Miller,<sup>1,3</sup> and Thomas Korff<sup>1</sup>

<sup>1</sup>Centre for Sports Medicine and Human Performance, Brunel University, <sup>2</sup>London Sport Institute, Middlesex University, London, United Kingdom; and <sup>3</sup>School of Exercise, Biomedical, and Health Sciences, Edith Cowan University, Perth, Western Australia

Submitted 14 June 2010; accepted in final form 14 September 2010

**Fath F, Blazeovich AJ, Waugh CM, Miller SC, Korff T.** Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans. *J Appl Physiol* 109: 1644–1652, 2010. First published September 16, 2010; doi:10.1152/jappphysiol.00656.2010.—Accurate and reliable estimation of muscle moment arms is a prerequisite for the development of musculoskeletal models. Numerous techniques are available to estimate the Achilles tendon moment arm in vivo. The purposes of this study were 1) to compare in vivo Achilles tendon moment arms obtained using the center of rotation (COR) and tendon excursion (TE) methods and 2) to assess the reliability of each method. For the COR method, magnetic resonance (MR) images from nine participants were obtained at ankle angles of  $-15^\circ$ ,  $0^\circ$ , and  $+15^\circ$  and analyzed using Reuleaux' method. For the TE method, the movement of the gastrocnemius medialis-Achilles tendon junction was recorded using ultrasonography as the ankle was passively rotated through its range of motion. The Achilles tendon moment arm was obtained by differentiation of tendon displacement with respect to ankle angular excursion using seven different differentiation techniques. Moment arms obtained using the COR method were significantly greater than those obtained using the TE method ( $P < 0.01$ ), but results from both methods were well correlated. The coefficient of determination between moment arms derived from the COR and TE methods was highest when tendon displacement was linearly differentiated over a  $\pm 10^\circ$  interval ( $R^2 = 0.94$ ). The between-measurement coefficient of variation was 3.9% for the COR method and 4.5–9.7% for the TE method, depending on the differentiation technique. The high reliabilities and strong relationship between methods demonstrate that both methods are robust against their limitations. The large absolute between-method differences (~25–30%) in moment arms have significant implications for their use in musculoskeletal models.

center of rotation; tendon excursion; musculoskeletal model

QUANTIFICATION OF HUMAN MUSCLE FORCES in vivo is difficult, because direct measurements are highly invasive (3, 4). Therefore, muscle forces are typically estimated indirectly using biomechanical modeling techniques. One method used to estimate muscle forces is calculation of the ratio of the muscular moment about a joint and the moment arm of the muscle or tendon of interest. A number of techniques can be used to obtain muscle moment arms, including cadaver dissection (6, 7, 11, 27), magnetic resonance (MR) imaging (16, 26), and ultrasound imaging (9, 12–14).

Two of these methods in particular have become popular within the scientific community. The first technique uses sagittal plane two-dimensional MR images to estimate the center

of rotation (COR) of a joint (25, 26). The perpendicular distance between the center of the joint and the line of action of the muscle or tendon of interest is then measured directly (16, 18, 26). The second technique is the tendon excursion (TE) method. The principle of virtual work (1, 28) is used to compute the moment arm as the ratio of the linear displacement of the tendon to the angular excursion of the corresponding joint (9, 17, 27). Thus it does not require knowledge of the location of the COR.

The COR and TE methods have advantages and disadvantages. One limitation of the COR method is that the multiple steps of manual MR image processing, required to determine the COR, can introduce errors in the COR calculation and, therefore, in the moment arm estimation (16). Another disadvantage of the COR method is the limited accessibility and relatively high costs involved in using MR scanners. A major advantage of MR imaging, however, is the high visibility of the underlying anatomic structures about the joint. In particular, the line of force can be easily identified. The major advantage of the TE method is that it does not require knowledge of the COR or the line of action of muscle or tendon force. In addition, ultrasonography, which is often easier to access and more time- and cost-efficient than MR scanning, can be used for the TE method. The main limitation of the TE method is the assumption that no internal forces, including friction, act in the joint of interest during a passive rotation (principle of virtual work) (1, 28). Thus it is assumed that active and passive forces within the joint are negligible and that all muscles spanning the joint of interest are inactive.

Lee and Piazza (13) used the TE method to determine in vivo values of Achilles tendon moment arms. The moment arms reported by Lee and Piazza are smaller than those reported by Maganaris et al. (16), who used the COR method. While these differences can have multiple explanations (including the use of participants with different anthropometric characteristics), these findings raise the question whether moment arms obtained using these two methods are comparable. However, to our knowledge, a direct comparison between these two methods has not been made. Therefore, differences in the moment arms reported in the literature cannot incontrovertibly be attributed to methodological differences. The first purpose of the present study was to compare moment arm measures of the Achilles tendon using the COR method (MR imaging) and the TE method (ultrasound imaging).

When scientific measurement techniques are evaluated, the reliability of the dependent measure is another important consideration. Coefficients of variation (CVs) of moment arms obtained using the COR method have been reported to be 7.9% (16, 17). The mean between-day difference in moment arms

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