IMPROVED METHODS FOR MEASURING ACHILLES TENDON MOMENT ARM
PROSPECTIVELY APPLIED TO TOTAL ANKLE ARTHROPLASTY PATIENTS

A Dissertation in
Kinesiology
by
Francesca Wade

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Submitted in Partial Fulfillment
of the Requirements
for the Degree of

Doctor of Philosophy

May 2020
The dissertation of Francesca Wade was reviewed and approved by the following:

Stephen J. Piazza  
Professor of Kinesiology, Mechanical Engineering, and Orthopaedics and Rehabilitation  
Dissertation Advisor  
Chair of Committee

Neil A. Sharkey  
Emeritus Professor of Kinesiology and Orthopaedics and Rehabilitation

Jonas Rubenson  
Associate Professor of Kinesiology

Gregory S. Lewis  
Assistant Professor of Orthopaedics and Rehabilitation

Jonathan Dingwell  
Professor of Kinesiology  
Graduate Program Director
ABSTRACT

The plantarflexion moment arm of the Achilles tendon (ATma) is critical to healthy locomotion. However, there is no consensus on the best method for measuring ATma. The first study of this dissertation outlines a novel method for quantifying ATma during dynamic tasks, using motion capture to define a functional axis of ankle rotation and ultrasound imaging to locate the Achilles tendon in vivo. The second study compares two methods for computing the axis of ankle rotation – using either a fixed functional axis or instantaneous helical axes. In plantarflexion and neutral, the fixed axis was the most appropriate choice, and provided estimates of ATma commensurate with previously reported values. The second half of this dissertation applies this novel method to total ankle arthroplasty (TAA). Study 3 found that the reduction in pain observed following TAA was associated with increased walking velocity and performance on locomotor tasks. However, improvements in pain did not fully explain improvements in locomotor function, which led to Study 4, where we investigated whether TAA altered ATma. The results indicated that approximately 37% of the change in ATma following TAA was attributable to a change in talar positioning. Further work is needed to determine any muscle adaptations to TAA, and to investigate whether a relationship exists between changes in ATma and changes in locomotor function following TAA.
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<td>2D</td>
<td>Two-dimensional</td>
</tr>
<tr>
<td>3D</td>
<td>Three-dimensional</td>
</tr>
<tr>
<td>6MWT</td>
<td>Six-Minute Walk Test</td>
</tr>
<tr>
<td>6MW</td>
<td>Six-Minute Walk</td>
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<tr>
<td>AES</td>
<td>Ankle Evolutive System</td>
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<tr>
<td>AOFAS</td>
<td>American Orthopaedic Foot and Ankle Society</td>
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<tr>
<td>ATma</td>
<td>Achilles Tendon Moment Arm</td>
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<tr>
<td>BMI</td>
<td>Body Mass Index</td>
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<tr>
<td>CoR</td>
<td>Center of Rotation</td>
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<tr>
<td>FA</td>
<td>Functional Axis</td>
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<tr>
<td>FDA</td>
<td>Food and Drug Administration</td>
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<td>GRF</td>
<td>Ground Reaction Force</td>
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<tr>
<td>IHA</td>
<td>Instantaneous Helical Axis</td>
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<tr>
<td>LCS</td>
<td>Low Contact Stress</td>
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<tr>
<td>MR</td>
<td>Magnetic Resonance</td>
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<td>MTJ</td>
<td>Myotendinous Junction</td>
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<tr>
<td>MVC</td>
<td>Maximal Voluntary Contraction</td>
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<tr>
<td>OA</td>
<td>Osteoarthritis</td>
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<tr>
<td>PROMs</td>
<td>Patient-Reported Outcome Measures</td>
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<td>ROM</td>
<td>Range of Motion</td>
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<tr>
<td>SF-36</td>
<td>RAND Short Form 36-Item Survey</td>
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<tr>
<td>SLBT</td>
<td>Single Leg Balance Test</td>
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<tr>
<td>STAR</td>
<td>Scandinavian Total Ankle Replacement</td>
</tr>
<tr>
<td>TA</td>
<td>Transmalleolar Axis</td>
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<td>TAA</td>
<td>Total Ankle Arthroplasty</td>
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<td>Transmalleolar Midpoint</td>
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ACKNOWLEDGEMENTS

This dissertation would not have been possible without my incredible support system. First and foremost, I have to acknowledge my Mum, whose constant emotional support, provision of English tea and chocolate, and exceptional proof-reading has meant I’ve managed to put together a coherent dissertation. Secondly, I must thank my Dad, for the support and reminders of how proud he is of me lifting my spirits when things got tough.

Alexa, Kristen, Nicole, Lauren – I don’t have the words to thank all the support you have given, except to say this wouldn’t have been completed without you. I know we’ll go on to do amazing things and I’m excited to see where we all end up. Emily – you’ve been there for all the highs and lows, even if you haven’t been physically here. Thanks for reminding me that there’s not 300 degrees in a circle!

To all the biomechanics graduate students I’ve worked with throughout this PhD, thanks for all the coffee, chats, and laughs.

To my committee, thanks for the constructive feedback and support you’ve given me. I’ve enjoyed the discussions we’ve had.

Steve, thank you for the guidance, the hours of edits and support you’ve given me throughout this process. I couldn’t have asked for a better advisor.
Dedication

This PhD is dedicated to all the graduate students who persist, despite the inequalities and opposition. You can do this.
Chapter 1

Introduction

1.1 Background

Healthy ankle function is critical to healthy locomotion. During the stance phase in young individuals, the hip and ankle joint moments are the primary contributors to support, and the ankle moment drives push-off (Winter, 1980). The role of the ankle is perhaps most critical during the late phase of stance, when push-off occurs (Farris and Sawicki, 2012; Kuo et al., 2005; Winter et al., 1990) and the muscles crossing the ankle generate power. The ankle generates approximately 50% of total positive power during walking and running (Farris and Sawicki, 2012), however this reduces with age and pathology. Deficiencies in ankle strength have been correlated to a reduced walking speed in the elderly (Bean et al., 2010; Marsh et al., 2006; Suzuki et al., 2001) and this may be due to an inability to generate sufficient ankle moment during push-off (Rasske and Franz, 2018; Winter et al., 1990). A compromised ability to generate joint moments has been observed in patients with lower-limb osteoarthritis (Astephen et al., 2008; Mündermann et al., 2005; Nüesch et al., 2012; Valderrabano et al., 2007b).

Recent work has suggested that the plantarflexion moment arm of the Achilles tendon (ATma) may be an important determinant of locomotor function. ATma is the distance from the ankle joint center of rotation to the line of action of the triceps surae force that is carried by the Achilles tendon. The magnitude of ATma has been posited to influence how much energy can be stored in the Achilles tendon (Raichlen et al., 2011; Scholz et al., 2008; Voigt et al., 1995) as well as where the triceps surae operate on their force-length and force-velocity curves (Ackland et al., 2012; Baxter et al., 2012; Baxter and Piazza, 2014; Lee and Piazza, 2009; Nagano and Komura, 2003). A joint with a
larger moment arm will cause the associated muscle to have an increased change in length and velocity for an equal amount of joint rotation compared with the same joint with a smaller moment arm. Evidence suggests that moment arms may change with aging due to reductions in muscle size (Csapo et al., 2014; Sugisaki et al., 2010) and migration of tendon insertion points (Kim et al., 2011).

There are several methods available for measuring ATma in vivo in humans. Direct measurement of the distance from the ankle joint center to the Achilles tendon can be obtained from magnetic resonance (MR) imaging (e.g., Rugg et al., 1990). The distance between the ankle center and the tendon can be inferred from the displacement of the myotendinous junction of the Achilles tendon with respect to the ankle joint rotation (tendon excursion, TE). TE may be tracked, using either ultrasound (e.g., Fath et al., 2010; Kawakami et al., 2001) or MR imaging (e.g., Wilson et al., 1999). There have also been several recent attempts to estimate ATma geometrically using ultrasound to image the tendon, with reflective markers attached to the probe and to bony landmarks (Manal et al., 2013; Rasske et al., 2017). The benefits of this method include that it requires no expensive MR scanner and that it can be used to estimate ATma during dynamic tasks.

![Figure 1-1](image.png)

**Figure 1-1.** Anterior-posterior and lateral radiographs of a Salto Talaris total ankle arthroplasty Haskell, 2012, pg 468
One surgery that has the potential to radically alter ankle function is total ankle arthroplasty (TAA). TAA is indicated when a patient has end-stage arthritis in the ankle joint that has not responded to conservative treatment. In a TAA, the arthritic bone and cartilage of the talus and distal tibia is resected and replaced with metal components that articulate with the aid of a polyethylene bearing. The popularity of TAA as an alternative to talocrural arthrodesis (ankle joint fusion) is growing in the U.S. (Pugely et al., 2014) as improved ankle prosthetic design has led to encouraging results (Vickerstaff et al., 2007). Much of the literature on TAA has focused on survivorship of the implants (Zaidi et al., 2013), including revision rates. Patient-reported outcomes following TAA show improvements in pain and self-reported function when compared with ankle fusion (Zaidi et al., 2013).

Functional improvements in range of motion (e.g., Kane et al., 2017; Valderrabano et al., 2003a) and gait (e.g., Brodsky et al., 2011) have been reported following TAA. The general consensus from the literature is that compared with arthrodesis and pre-surgery levels, TAA improves spatiotemporal gait parameters, gait symmetry, range of motion, and kinetic parameters such as ground reaction force profile and ankle moment, but that patients still show some deficits in comparison with controls. Specifically, these deficits were reported to persist in total plantarflexion range of motion (77% of healthy controls), maximal plantarflexion moment (89% of healthy controls), and maximal ankle joint power (63% of healthy controls; Valderrabano et al., 2007b). While rehabilitation following lower limb arthroplasty aims to restore gait symmetry to prevent unilateral overloading in the uninvolved limb (Rapp et al., 2015), the aforementioned deficits have been related to an asymmetrical walking pattern two-years following TAA (Queen et al., 2014a).
1.2 Objective, Specific Aims and Hypotheses

To date, there has been no investigation of the effect of TAA on ATma, and how this procedure might affect functional outcomes. ATma is dependent on the structure of the ankle joint, which determines the location of the ankle axis of rotation. It is conceivable that the axis of rotation at the ankle might be linked to the success of ankle arthroplasty, as stated by Deland and colleagues (2000):

“The complex dynamic nature of the ankle axis of rotation may be one reason for less than optimal results in total ankle replacement surgery”

The general purpose of this dissertation is to develop a novel method of quantifying ATma and to apply this method to TAA patients prospectively. The rationale is that TAA may change the location of the axis of ankle joint rotation, and therefore may change leverage behavior of the triceps surae.

The dissertation starts with the development of a method to compute three-dimensional estimates of Achilles tendon moment arm, first with a single fixed axis of ankle rotation (Study 1) and then with an axis of ankle rotation that moves with ankle rotation (Study 2). Prior to and following TAA, patients’ function was assessed using a Timed Up-and-Go (TUG) task, a Single Leg Balance test (SLBT), and a six minute walk (6MW), and related to any changes in self-reported pain. Patients’ function will be compared with that of age-matched controls to assess where any persisting deficits may arise (Study 3). Our novel method of ATma estimation was applied to patients before and after they undergo TAA to investigate any changes that may occur due to the surgery (Study 4).

The results of this work provides insight into the mechanical changes that might happen following TAA. Comparisons between TAA patients and healthy controls contribute to the literature on performance of the implants, whilst comparisons pre- and post-surgery within patients will elucidate the effect of the surgery on ATma.

This dissertation will address the following Specific Aims:
Aim 1: To investigate differences in ATma determined from an axis of rotation derived from ankle motion to those estimated using an axis dependent upon bony landmarks.

Hypotheses: ATma computed from the motion-derived axis will be larger than those derived from axes dependent on bony landmarks, and ATma quantified using the motion-derived axis will experience larger changes with ankle joint angle. Finally, ATma computed from the motion-derived axis will increase with loading more than ATma derived from anatomical landmarks.

Rationale: Comparing popular methods of determining ankle joint axis of rotation to observe their effect on ATma values is important as there is currently no consensus in the literature. Many studies assume the ankle joint center lies fixed at the midpoint of the malleoli; this is known to be incorrect from bone studies. The values obtained by others (e.g., Manal et al., 2013; Rasske et al., 2017) who utilized a combination of ultrasound and motion capture, with an anatomically defined center of rotation to compute ATma are low in comparison to the values reported from two-dimensional (2D) MR studies (e.g., Rugg et al., 1990). It is thought that a ‘functional axis’ of rotation, one that is found from motion of the foot with respect to the shank, might produce ATma values that are load- and angle-dependent, and that are similar to those reported from 2D MR studies, as a modified Reuleaux method also uses the motion of the joint to find a center of ankle rotation.

Aim 2: To determine how estimates of ATma using an axis of ankle rotation determined using instantaneous helical axis decomposition and allowed to move with ankle motion differ from ATma estimated using a single fixed axis of ankle rotation.

Hypothesis: ATma in the mid-range of motion will be similar when estimated from the two axis approaches, but that ATma will differ at the end range of motion. The instantaneous helical axis derived ATma will produce smaller values of ATma than FA.

Rationale: From in vivo studies of bone motion using roentgen stereophotogrammetry (Lundberg, 1989; Lundberg and Svensson, 1993) or bone pins (Arndt et al., 2004), it is known that the location of the axis of talocrural rotation is motion-dependent and its orientation as well as its anterior-
posterior location varies with joint angle. Three-dimensional (3D) imaging techniques such as MR provide a comprehensive way to locate the relative motions of the talus and tibia in addition to the soft-tissue structures such as the Achilles tendon. Studies using this method to compute ATma report an increase with plantarflexion, to a point, before leveling off or reducing at extreme plantarflexion (Clarke et al., 2015; Sheehan, 2012). A prior comparison of ATma estimated from a fixed point of rotation versus one that was permitted to move showed the two methods were comparable over the mid-range of motion, with the fixed point providing larger estimates of ATma than the point that was allowed to move (Rugg et al., 1990). MR imaging allows for the axis of rotation to move with joint angle, but uses a series of static images to emulate motion, restricting the types of movements that can be analyzed. The ultrasound-based geometric method allows for tendon imaging and joint tracking during dynamic tasks, and an instantaneous helical axis found from the motion might provide a more realistic approximation of talocrural rotation axis. We expect this method to produce ATma that are commensurate with those previously reported from 3D MR studies which use an instantaneous helical axis representation for ankle rotation.

**Aim 3: To determine pre- and post-TAA locomotor function in a prospective patient cohort compared to age-matched controls; and to determine the extent to which the observed changes are attributable to changes in pain levels.**

**Hypothesis:** Following TAA, 6MW average velocity will increase, SLBT time on the involved limb will increase and TUG time will decrease in comparison with pre-operative levels, but will not reach the level observed in healthy control participants. However, any improvements in locomotor function observed in the patient population will be related to reductions in pain.

**Rationale:** While TAA has been investigated thoroughly from an implant design and survivorship perspective, little focus has been given to the functional changes following the surgery. Gait analysis studies have been conducted but are confounded by changes in pain, and comparisons with control subjects have been limited. Previous work has shown notable improvements in gait
following TAA when compared with pre-operative performance (e.g., Piriou et al., 2008; Segal et al., 2018; Seo et al., 2017; Singer et al., 2013), but that some deficits still persist when compared with healthy controls (Flavin et al., 2013; Rouhani et al., 2012; Seo et al., 2017; Singer et al., 2013), including a reduced plantarflexion moment during push-off (Valderrabano et al., 2007b) and persisting asymmetry (Queen et al., 2014a). Queen et al. (2014b) evaluated performance on functional tasks (TUG and Four Square Step Test) and noted improvements following TAA. However, they did not compare to healthy controls to see if functional deficits were present as have been reported in gait. This study compared pre-operative patients and the same patients six months post-operatively to age-matched controls to investigate changes in functional performance due to TAA, and will also investigate the effect of pain levels on locomotor function in the patient population.

**Aim 4: To determine how, if at all, TAA alters ATma.**

*Hypothesis:* If changes exist following TAA, the pre-operative values of ATma will be correlated with the post-TAA value.

*Rationale:* This will be the first study to investigate ATma following TAA. One ubiquitous goal of total ankle implant design is to replicate physiologic anatomy as closely as possible (Cracchiolo and DeOrio, 2008). As such, the hypothesis is based on the assumption that TAA attempts to restore healthy structure and alignment to the ankle joint. This study compared ATma values in a cohort of patients pre-operatively to ATma obtained six months post-operatively. In addition to using our ultrasound-based motion capture method to quantify ATma, pre- and post-operative standing sagittal-plane radiographic images were analyzed to obtain an anterior-posterior distance from an approximate center of ankle joint rotation to the most posterior aspect of the calcaneus.
1.3 Organization of Dissertation

Chapter 2 will be a comprehensive literature review, outlining ankle arthritis, a history of total ankle arthroplasty and prosthetics, and a review of clinical and functional outcomes following TAA previously reported. Chapter 3 will describe the development of a method to determine ATma. Chapter 4 describes the evolution of this method to incorporate a moving instantaneous helical axis as a representation of ankle rotation. Chapter 5 outlines the functional changes, assessed over a range of tasks, before and after TAA surgery and compared with age-matched controls. It also evaluates the influence of self-reported pain levels on locomotor function. Chapter 6 investigates any changes in ATma between the pre- and post-surgery time points using the novel method outlined previously. Finally, Chapter 7 closes the dissertation with a summary of findings, a critique on the work conducted, and some suggestions for future work.
Chapter 2

Literature Review

2.1 Overview of Ankle Joint Anatomy

The ankle complex consists of both the talocrural and subtalar joints. The talocrural joint, also referred to as the ankle mortise or tibiotalar joint, is the articulation between the talar dome, tibial plafond, the medial malleolus, and the lateral malleolus (Close, 1956). The talus does not have a uniform shape, with the posterior three-fourths having a different curvature from that of the anterior fourth, forming a conical shape (Leardini et al., 1999). The subtalar joint is the articulation between...
the talus and the calcaneus and navicular bones, and predominantly allows for inversion (combination of internal rotation and supination) and eversion motions (Rockar, 1995).

The structure of the ankle permits a primary motion of plantar- and dorsi-flexion, while the protruding fibula reduces the range of eversion (external rotation and pronation) motion (Huson, 1987). The non-uniform talar shape lends itself to a shifting articulation contact area (Bonasia et al., 2010).

The ankle complex is surrounded by many ligaments that provide support and stability to the talar head. A thick fibrous medial ligament comprised of deep and superficial fibers forms a triangular support on the medial aspect of the ankle, while the lateral side is guided by the anterior and posterior talofibular and the calcaneofibular ligaments.

The ankle predominantly moves in the sagittal-plane (plantar- and dorsi-flexion), so in many studies, the joint is considered to function like a hinge. In fact, the seminal anatomy textbook ‘Gray’s Anatomy’ (Standring, 2016) refers to the talocrural joint as a hinge. A more appropriate approximation is that of a universal joint, or a pair of hinges. However, even this representation does not fully replicate the complex motion observed in the ankle joint, which includes sliding motions of the articulating bony surfaces observed in studies of bone kinematics (Arndt et al., 2004; Barnett and Napier, 1952; Isman and Inman, 1969; Lundberg, 1989; Lundberg and Svensson, 1993; Sammarco, 1977).

Motion of the ankle is produced by the surrounding muscles which act through their respective tendons. There are three main tendons acting at the ankle complex: two peroneal tendons and the Achilles tendon. The gastrocnemius and soleus muscles (triceps surae) act through the Achilles tendon, which inserts onto the posterior surface of the calcaneus. In addition to transferring triceps surae forces to produce plantarflexion, the Achilles tendon also plays a vital role in energy storage and release (Sawicki et al., 2009).
2.2 Methods to Obtain ATma

The ability to reliably and accurately compute ATma is necessary to understanding muscle and joint behaviors, and to create meaningful simulations of movement involving the ankle joint. Muscle force estimates are especially sensitive to errors in moment arm – a 20% error in moment arm can produce a 50% error in muscle force computation (Herzog, 1992). Geometric-based methods and tendon excursion methods are two prevalent approaches to estimate ATma. Some researchers have combined the strengths from each to create a portable approach for dynamic assessments (Franz et al., 2013; Lichtwark and Wilson, 2006; Manal et al., 2010).

2.2.1 Two Dimensional Methods: Geometric

People have been interested in finding the center of rotation (CoR) of articulating objects for hundreds of years, and often use a geometric approach. This classification of methods uses the motion of articulating objects about a point and applies mathematical rules of geometry to find that point, for example by fitting a sphere (e.g., Demarais et al., 2002) or finding the bisector of perpendicular lines. The Reuleaux method to find CoR was first presented in the 19th Century and has become a classic part of planar kinematics. Typically, the Reuleaux method is modified and applied to planar radiographic images of the ankle over a range of motion. For each joint angle, two points are located using bony landmarks and a line drawn between them for each segment. The intersection of the perpendicular bisectors for each point is the average location of the CoR for the motion. ATma is then calculated as the perpendicular distance from the CoR and the midline of the Achilles tendon.

Locating the CoR or axis of rotation is not as straightforward as it initially seems. Studies of bone kinematics, through the use of bone pins, roentgen stereophotogrammetry and radiographic
imaging, show moving articulation points and changes in orientation of the rotational axis of the joints that collectively produce the resultant ankle motion (Arndt et al., 2004; Barnett and Napier, 1952; Isman and Inman, 1969; Lundberg, 1989; Lundberg et al., 1989; Lundberg and Svensson, 1993; Sammarco et al., 1973; Sammarco, 1977). The talocrural axis is inclined inferiorly and laterally during dorsiflexion (Barnett and Napier, 1952; Isman and Inman, 1969; Lundberg, 1989; Lundberg et al., 1989) but moves throughout motion to a downwards medial orientation in plantarflexion, with some joint motion produced by interactions in joints of the foot (Lundberg, 1989). Throughout motion, most talocrural axes appear to pass through a point which represents the CoR. This central point lay lateral to the midpoint of a transmalleolar axis in eight healthy subjects studied over a range of static images (Lundberg et al., 1989). Their conclusion might lead to the use of a transmalleolar midpoint representation of CoR, but it only applies for the talocrural joint, not the resultant ankle motion.

Rugg and colleagues (1990) were one of the first groups to investigate the effect of a fixed or moving CoR using a modified Reuleaux method. They took a series of MR images with some unreported level of muscle activation in ten male ankles as they were rotated through 50°, with static images taken every 10°. They reported mean ATma that increased with plantarflexion, but were not observably influenced by changing between a moving or fixed CoR. Their study also reported reliability measures, finding a fixed CoR had an average deviation of 1.5 -2 mm across five separate calculations of CoR using the same MR images.

The modified Reuleaux method was applied to MR images of six male subjects to investigate the effect of muscle contraction on ATma (Maganaris et al., 1998). ATma increased with plantarflexion, and values reported with an isometric maximal voluntary contraction (MVC) were between 10 – 15 mm larger, dependent on joint angle.

Baxter and Piazza (2014) applied the modified Reuleaux method to MR images of the ankle complex in 20 males. They computed ATma as the shortest distance between CoR and tendon line
of action, and found that the capacity to produce torque at the ankle could be predicted by ATma and triceps surae muscle volume.

The CoR method is sensitive to measurement errors (Maganaris, 2004). The most egregious of these is the alignment of the radiographic plane to the plane in which motion occurs. If there is any deviation, ATma values are effected by a multiplier of the cosine of the angle between the two planes. Researchers take care to align the axis of the foot rotator with the lateral malleolus to minimize this error, but it is known that the true axis of rotation of the ankle joint does not remain fixed at this point (Isman and Inman, 1969; Lundberg, 1989; Lundberg et al., 1989; Sammarco, 1977).

2.2.2 Two Dimensional Methods: Tendon Excursion

The underlying principle is that moment arm determines the amount of tendon movement (excursion) over a joint rotation. Mathematically, the derivative of tendon excursion (TE) with respect to joint angle is equivalent to the moment arm. TE uses differentiation of the change in length of the Achilles tendon over the change in ankle angle to calculate ATma. The concept was first presented by Storace and Wolf (1979) and expanded by An et al. (1983), where they used the principal of virtual work to create this definition. In summary, TE relies on the assumption that tendon force remains constant during a motion, thus if the tendon stretches or relaxes during the movement, then TE cannot be attributable solely to joint rotation. Linear displacement of the tendon is most commonly determined using B-mode ultrasonography to track the motion of the myotendinous junction (MTJ) (Fath et al., 2010; Hashizume et al., 2016; Ito et al., 2000; Lee and Piazza, 2012).
Ultrasound-tracked tendon displacement has been used to determine ATma in twenty older adults (Lee and Piazza, 2012). The authors found a strong correlation between smaller ATma and slower elderly subjects’ preferred walking velocity, as determined by performance on a 6MW test. TE remains popular due to ease of use, and its lack of reliance on locating an appropriate CoR of the ankle joint, but the importance of constant tendon loading when using TE must not be overlooked. A varied load in the Achilles tendon during measurement, either through a change in the level of muscle contraction or the angle of the joint, led to varied estimates of ATma (Olszewski et al., 2015). The Achilles tendon has a slack length that changes based on ankle angle (Hug et al., 2013). Some researchers ‘precondition’ the tendon to minimize the effect of slack length on ATma measures (e.g., Fath et al., 2010; Franz et al., 2015), but many do not consider the change in tendon stiffness within their subjects. Comparison of moment arms in old and young tendons calculated using TE must be cautioned against due to the age-related change in tendon stiffness (Voigt et al., 1995).

The values obtained by geometric and TE methods have been compared. Fath and colleagues (2010) observed geometric values were significantly larger than TE ones, although the methods were correlated. The best agreement between CoR and TE occurred during plantarflexion (highest $R^2 = 0.94$ at an angular interval of $10^\circ$). Maganaris and colleagues (2000) compared the effect of maximal voluntary contraction (MVC) on moment arm and found that while TE ATma values changed with joint angle, increasing with plantarflexion, there was no difference between loading conditions. At rest, the moment arms reported with TE were not different to those computed using CoR (Maganaris et al., 2000).
2.2.3 Three Dimensional Methods: Geometric

The methods presented thus far have been two-dimensional but we do not exist in a planar world. Three-dimensional measures of ATma are necessary for the most accurate representations. While TE does not require an estimate of ankle center of rotation, many geometric based methods do, and 2D geometric methods have used a single point to represent a center of rotation of the ankle joint. Krähenbühl et al. (2017) suggest that the subtalar joint moves with a helical motion, and studies on bone kinematics of the ankle joint (Arndt et al., 2004; Lundberg, 1989; Lundberg et al., 1989; Lundberg and Svensson, 1993) imply that for 3D studies, a helical axis might be a close approximation of the true axis of rotation of the ankle.

Hashizume and colleagues found 2D ATma computed with the CoR approach were overestimated when compared to 3D values computed using a finite helical axis of rotation over a 20° angular displacement (Hashizume et al., 2012). The talocrural joint rotation axis was defined by the motion of the tibia relative to the talus. 3D ATma was determined as the shortest distance from the projection of the Achilles tendon force line of action onto the orthogonal plane of the joint axis to the finite helical axis. Using the same images and tendon line of action, CoR was found and used to compute 2D ATma. Their study showed the talocrural joint axis was not orthogonal to a sagittal MR imaging plane, contributing to the disparity between ATma values.

An alternative to a finite helical axis, which requires substantial rotation within a defined epoch of motion, is to compute an instantaneous helical axis. Sheehan (2012) obtained an instantaneous helical axis of calcaneal-tibial rotation from angular velocity in 3D dynamic MR data. The instantaneous helical axis relies on velocity data, and in most MR machines, this has to be determined from positions, introducing errors through differentiation. With cine-MR, angular velocity can be obtained directly to provide reported accuracy of < 0.5 mm error, and repeatability of < 1.8° and 1.5 mm (Sheehan, 2010). 3D ATma were larger in plantarflexion than dorsiflexion,
although there was no significant pattern of ATma increasing as plantarflexion increased, and 3D values were approximately 3mm less than 2D ATma (Sheehan, 2012).

While dynamic MR analyses are comprehensive, access to such technology is limited. It is clear that a method that is accessible, simple to use, inexpensive and portable is necessary for researchers interested in variables reliant on ATma. Lichtwark and Wilson (2005) were interested in the mechanical properties of the Achilles tendon in hopping motions. To obtain dynamic tendon information, they combined active markers attached to the ultrasound probe, and anatomical landmarks on the foot and shank with B-mode ultrasound imaging of the MTJ.

This method was further expanded to compute ATma (Manal et al., 2010). Ultrasound images were sampled in line with the line of action of the Achilles tendon, at the level of the ankle joint center (assumed to lie at the midpoint of the malleoli). Retroreflective markers were placed on the malleoli, to define CoR, and the ultrasound probe in order to determine the transformation between 2D and 3D location of the tendon. The method was initially tested on a lamb shank, with high repeatability (0.5 mm standard deviation over five trials), and a small difference (3%) in ATma obtained directly from the lamb. When applied to human ankles, the authors reported a minimal change in ATma with plantarflexion with no discernable change with MVC (Manal et al., 2013).

A similar ultrasound-based geometric method was applied to investigate the change in ATma during walking (Rasske et al., 2017). A dynamic assessment of ATma over the gait cycle was conducted on ten young healthy adults walking on a split belt treadmill. At the same joint angle, ATma differed slightly between stance and swing phases of gait, with peak ATma occurring just after toe-off. In addition to the load-dependence, the authors also observed a small change in ATma with plantarflexion angle.
2.3 Ankle Arthritis

Arthritis is an all-encompassing term referring to any pain or disease in skeletal joints, and may be classified as degenerative, inflammatory (rheumatoid), infectious, or metabolic arthritis (Lawrence et al., 1998). Degenerative arthritis, known as osteoarthritis (OA), is the most common form. OA is broadly defined as a debilitating condition where the articulating cartilage wears away, with increasing subchondral bone loss, eventually leading to bone-to-bone contact causing pain and further joint damage (Huch et al., 1997). Lower limb OA results in pain, reduced quality of life (Glazebrook et al., 2008), reduced mobility and joint stiffness, swelling in the joint (Huch et al., 1997), and altered walking mechanics (Schmitt et al., 2015).

Lower limb OA is growing in prevalence. Up to 25% of U.S. adults reported a diagnosis of arthritis between 1999 and 2014, and in the same population, the prevalence of OA has doubled from 6.6% to 14.3% (Park et al., 2017). Individuals who are older and obese have an increased risk of OA (Lementowski and Zelicof, 2008), and as the U.S. has an aging population with elevated average BMI (Crowninshield et al., 2006), the incidence of osteoarthritis will continue to rise. A report in 2008 established that approximately 27 million adults in the US population are diagnosed with OA (Lawrence et al., 2008), while the authors estimated that by the year 2020, 59.4 million U.S. adults will be diagnosed. While OA of the ankle is less prevalent than OA of other lower limb joints (Cushnaghan and Dieppe, 1991; Huch et al., 1997), it still impacts approximately 1% of the global adult population (Huch et al., 1997; Peyron, 1984; Valderrabano et al., 2009) and is associated with the development of kidney disease, heart failure, and development of OA in other joints (Saltzman et al., 2006).

Lower limb OA can develop from one of three categories – post-traumatic, primary and secondary. Post-traumatic OA accounts for 12% of overall symptomatic OA (Brown et al., 2006) but at the ankle, this increases to approximately 70% (Barg et al., 2013; Saltzman et al., 2005; Valderrabano
et al., 2009), and is attributable to 3.06 billion US dollars annually (Brown et al., 2006). For further information on primary and secondary OA, I refer the reader to Barg et al. (2013) and Taruc-Uy and Lynch (2013).

Patients presenting with post-traumatic OA are typically younger than those who present with primary OA, with an increased risk of developing OA over the age of 30 (Saltzman et al., 2005; Valderrabano et al., 2009). The risk of developing post-traumatic OA at the ankle also increases with fracture severity (Horisberger et al., 2009), malleolar ankle fracture (Lübbeke et al., 2012; Valderrabano et al., 2009), presence of a fracture-dislocation (Beris et al., 1997; de Vries et al., 2005), a Weber C fracture (Horisberger et al., 2009; Lübbeke et al., 2012), increased age (Horisberger et al., 2009; Lübbeke et al., 2012; Valderrabano et al., 2009), overweight or obese weight (Lübbeke et al., 2012; Valderrabano et al., 2009), and female sex (Horisberger et al., 2009; Lübbeke et al., 2012).

Radiographic imaging combined with patient-reported outcome measures (PROMs), like the American Orthopaedic Foot and Ankle Society Hind-Foot score (AOFAS-HF) and Short Form-36 (SF-36), pain assessment, and physical examinations of the joint can indicate how far the disease has progressed. In radiographic images, the physician can assess the presence of bone end sclerosis, presence of cysts in subchondral bone, osteophyte formation, and joint space narrowing, together comprising the Kellgren and Lawrence scale (Kellgren and Lawrence, 1957).

Assessment of mobility at the ankle joint is conducted during the physical examination, first by manually manipulating the joint to assess range of motion (RoM) and orientations which produce pain (Young et al., 2005), and by gait analysis. Patients with asymmetric ankle OA have lower hind-foot RoM, reduced peak ground reaction force (GRF), reduced ankle plantarflexion moment, and reduced ankle joint power when walking (Nüesch et al., 2013, 2012). The slower walking speed might be explained by muscular atrophy on the affected side, which is associated with advanced stage OA of the ankle (Wiewiorksi et al., 2012).
Classification of OA progression is not standardized, with many scales reported in the literature (Cheng et al., 2001; Giannini et al., 2007; Takakura et al., 1995; Tanaka, 2006). These scales are generally comprised of four or five stages, where a lower number represents normal alignment, early osteophyte formation and no reduction in joint space, and a higher number means obvious malalignment, almost no joint space and complete bone contact. Once classified, the appropriate treatment is discussed.

2.4 Treatments for Arthritis

The management of ankle OA takes many forms, but starts with a comprehensive, non-operative treatment. For patients who have contraindications for surgery, conservative treatments may be their only option. For further information on conservative treatments, the reader is referred to the review by Schmid and Krause (2013).

In patients whose symptoms fail to respond to conservative measures, surgical options are considered. Briefly, surgical treatment of ankle OA can be categorized as either joint-sacrificing or joint-saving (Bloch et al., 2015). Joint-preserving procedures for moderate OA include debridement, chondroplasty or osteophyte resection, or distraction arthroplasty and allografts for end-stage OA in younger patients (Bloch et al., 2015; Giannini et al., 2007). Joint-sacrificing surgeries include arthrodesis and arthroplasty (Weatherall et al., 2013). These procedures are not mutually-exclusive. Often, alignment surgery, which is thought of as joint-saving, is performed prior to a joint-sacrificing arthrodesis or arthroplasty (Cooke and Goldberg, 2014; DiDomenico and Anania, 2011; Haskell, 2012; Nihal et al., 2008).

Tibiotalar arthrodesis, or fusion, has been the surgically preferred treatment option for older patients with end-stage OA for many years, and is still considered the gold standard by some (Kennedy et al., 2006; Pugely et al., 2014; SooHoo, 2007). The ankle is debrided, and articular
cartilage removed. Once the joint is prepared, the talus and tibia are compressed together ensuring direct contact between the bony surfaces, and fixed in place by implants such as cannulated screws (Kennedy et al., 2006; Nihal et al., 2008). The surgery has been shown to improve short-term patient-reported pain, quality of life and clinical assessments of function (Coester et al., 2001; Fuchs et al., 2003; Fuentes-Sanz et al., 2012; Houdek et al., 2014).

The surgery is associated with a high complication rate (Crosby et al., 1996), especially when there is pre-existing disease in the knee or midtarsal joints. There have been reports of loss of subtalar motion (Coester et al., 2001; Thomas et al., 2006), further degeneration due to a change in joint loading (Wang et al., 2015), and degeneration at neighboring joints (Fuchs et al., 2003). In fact, an increased incidence of arthritis in adjacent joints is associated with arthrodesis (Fuchs et al., 2003; Sheridan et al., 2006; Thomas and Daniels, 2003) and reoperation rates are high (approximately 11%; SooHoo, 2007).

In addition to associated complications with ankle fusion, the surgery does not improve all function. Sagittal RoM is severely restricted following an arthrodesis (Van Der Plaat et al., 2015; Zhang et al., 2016). The consequence of reduced mobility in the joint is most clearly observed during tasks of daily locomotion, such as gait or stair-climbing (Fuentes-Sanz et al., 2012; Jastifer et al., 2015; Singer et al., 2013; Thomas et al., 2006).

2.5 Total Ankle Arthroplasty as a Treatment for Ankle Arthritis

There is much debate as to whether end-stage OA of the ankle is best managed with arthrodesis or arthroplasty (Jordan et al., 2014; Krause and Schmid, 2012) but TAA is increasing in popularity. Between 1991 and 2010, 7251 TAA were performed on Medicare beneficiaries, with an increase in the number of hospitals performing TAA from 3.1% of all U.S. hospitals to 12.6% (Pugely et al., 2014).
During a TAA, the distal end of the tibia is resected, a tibial window is created, the bone is cut away with trial implants inserted prior to the final prosthesis being implanted (Cooke and Goldberg, 2014; Harris, 2012; Haskell, 2012). Alignment, in all three planes, of the implant to the distal and proximal anatomy is crucial to consider (Barg et al., 2010; Barg et al., 2015; Cooke and Goldberg, 2014; Haskell, 2012).

### 2.5.1 Cost Effectiveness

Few studies have established the cost effectiveness of TAA in comparison with talocrural arthrodesis. SooHoo & Kominski (2004) reported total expected lifetime treatment costs of $15,568 for TAA and $6,990 for arthrodesis, but once adjusted for quality of life-adjusted years there was an incremental cost-effectiveness ratio for ankle arthroplasty of $18,419 per year. A more recent cost analysis (Courville et al., 2011) found similar costs, with a beneficial ratio of $11,800 per quality-adjusted life year gained. An interesting outcome was their sensitivity analysis, showing few variables result in TAA not being a cost-effective procedure. Their study also highlighted that TAA was more cost effective for younger patients ($9,600/quality-adjusted life year in 30 year olds vs $1,800/quality-adjusted life year in 80 year olds). Both studies (Courville et al., 2011; SooHoo and Kominski, 2004) considered direct medical costs only, and did not account for societal costs. A TAA is normally associated with a 2.3 day hospital stay (Pugely et al., 2014), in addition to long-term physical therapy (Hintermann et al., 2004; Valderrabano et al., 2007b). These costs and the associated functional deficits persisting following surgery have not been accounted for in any cost analyses.
2.5.2 Indications and Contraindications

The American Orthopaedic Foot and Ankle Society (AOFAS) position statement (AOFAS, 2014) cautions that the choice of TAA or arthrodesis is dependent on patient-specific criteria. Surgeons should consider patient age, the arthritis etiology, and any morphological deformity or instability, specifically whether such abnormalities can be surgically corrected (Clare and Sanders, 2002). Patients who have contraindications for an arthrodesis, such as those with previous hind-foot fusion and significant arthritic change in neighboring joints, may be good candidates for TAA, and are not immediately excluded from the surgery (Cooke and Goldberg, 2014; Haskell, 2012; Steck and Anderson, 2009).

Ankle OA generally affects older individuals and this is reflected in the mean age of patients undergoing TAA. Patients range between 17 and 95 years old, with an average age of 60, which is lower than that at the hip and knee (68 and 69 years respectively), and an average BMI of 28 kgm$^{-2}$ (range: 19 – 44 kgm$^{-2}$) (Zaidi et al., 2013). These patient demographics are influenced by relative contraindications for TAA.

Patients younger than 50 years were shown to have early failure rates and low clinical scores following TAA (Spirt et al., 2004). Given that younger patients tend to report increased physical activity, thus increasing the demands on the implant, surgeons have minimized risk of failure and determined young age to be a contraindication for TAA (AOFAS, 2014; Hintermann, 2005). This viewpoint has been challenged by Rodrigues-Pinto et al. (2013) who found no significant differences in implant survivorship between younger (< 50 years) and older (≥ 50 years) patients, with a trend towards fewer complications in the younger individuals. Kofoed and Lundberg-Jensen (1999) found similar results in their fifteen year follow-up of 100 arthroplasty patients. Demetracopoulos et al. (2015) found similar outcomes in old and young patients up to five years.
following TAA, with younger patients showing greater improvement in AOFAS-HF and SF-36 scores. Further research is necessary before the accepted contraindication of young age is revised. TAA is considered over arthrodesis if multiple joint disease is present (Cooke and Goldberg, 2014).

In accordance with the AOFAS position statement, the indications for TAA surgery are less clear and should be considered on an individual patient basis. Instead, contraindications are easier to define. Active infections, Charcot neuroarthropathy, avascular necrosis of the talar dome, peripheral neuropathy or vascular disease, chronic ankle instability, severe irreducible malalignment of the lower limb, and non-compliance are all absolute contraindications for TAA (DiDomenico and Anania, 2011; Guyer and Richardson, 2008; van den Heuvel et al., 2010). Misalignment of the tibia and talus can negatively influence survivorship of the implant due to abnormal loading (Guyer and Richardson, 2008; Pappas and Buechel, 2013) but can often be surgically corrected at the same time, provided the deformity is less than a 15° tilt in the coronal plane (Hennessy et al., 2008).

Other contraindications can be corrected surgically, these include mild misalignment of the lower limb, soft tissue imbalance and talar deformity (Kim et al., 2009; Steck and Anderson, 2009; Zhou and Tang, 2016). Many consider a hind-foot deformity greater than 20° varus or valgus to be an absolute contraindication (Queen et al., 2013; Trajkovski et al., 2013; Valderrabano et al., 2005; Wood and Deakin, 2003), although more recent research shows this arbitrary misalignment threshold might be patient- and surgeon-dependent (Dodd and Daniels, 2017; Shock et al., 2011).

Bony misalignment deformities can lead to an unacceptably high failure rate of an ankle implant. Two hundred Scandinavian Total Ankle Replacement (STAR) patients were followed for a mean of 46-months, with 163 surviving implants after seven years (Wood and Deakin, 2003). Most implant failures were due to malleolus fracture within surgery, or loading of the edge of the mobile bearing - both associated with misalignment. Patients who present with incongruent joints pre-arthroplasty were ten times more likely to develop edge-loading that progressively worsens over
time following the surgery, even when alignment deformities were corrected (Haskell and Mann, 2004). Modeling studies help to explain these observations. Espinosa et al. (2010) investigated the effect of misaligning the components of the ankle with validated finite element models. They found misalignment greater than 5° of the implant components led to contact stresses in excess of the yield stress of the mobile bearing.

Just as misaligned implant components increase contact stresses within the prosthesis, so too can increased weight. Obese patients are at greater risk of developing OA (Lementowski and Zelicof, 2008) due to increased joint forces, and there is concern that increased mechanical stress at a joint can lead to implant failure. In hip and knee arthroplasties, accelerated implant failures in obese patients have been observed (Bryan et al., 2013). At the ankle, obese patients were also more likely to have increased complication rates following TAA due to perioperative factors (Werner et al., 2015). Researchers have noticed similar short-term outcomes on TAA success between obese and non-obese patients (Barg et al., 2011; Bouchard et al., 2015; Gross et al., 2016), although in a long-term follow-up of patients who underwent TAA, obese patients had an increased risk of implant failure (Schipper et al., 2016). Further long-term analyses are necessary before the effect of obesity on TAA survivorship is understood.

2.5.3 Revision of TAA

The survival rate of TAA implants depends on the implant used, the surgical technique and expertise of the surgeon involved, and co-existing pathologies (Gougoulias et al., 2010). As such, the revision rates following the surgery vary. A systematic review of 1105 TAAs in 2010 identified 108 implant failures (9.8%), but these were salvaged in 62% of cases (Gougoulias et al., 2010). Another systematic review (Labek et al., 2011) determined the revision rate for different joint replacement surgeries from worldwide joint registries. They reported a mean rate of 3.29 revisions
per 100 observed component years for TAA. Simply, a third of patients will have to undergo revision surgery ten years following TAA. This is substantially greater than the revision rates reported for both total hip replacement (1.29 revisions per 100 observed component years) and total knee replacements (1.26 revisions per 100 observed component years). The most common causes of implant failure and need for revision were talar component failure due to aseptic loosening, aseptic osteolysis, and subsidence (Roukis and Elliott, 2015). Part of the explanation for increased revision rates at the ankle can be attributed to the implant used in the surgery, and the prostheses have undergone many changes to the design over the years.

2.6 Total Ankle Arthroplasty: Implant Design

2.6.1 First Generation Designs

TAA implant designs must consider the small joint size in combination with the high resultant moment (Sadeghi et al., 2001) and compressive forces (Tillmann et al., 1985) experienced at the joint. Further complicating design criteria is the complex motion at the ankle, consisting of both sliding and rolling motions.

The first description of an ankle prosthesis was proposed by Themistocles Glück in the 1800s (Brand et al., 2011) and TAA, in its most basic form, was first attempted in 1913 (Eloesser, 1913). The first obvious clinical trial was reported by Lord and Marotte (1973). An inverted hip replacement was inserted in 25 patients, with the acetabular cup in the calcaneus and the femoral stem of the implant implanted into the tibia. The surgery failed in 12 patients, and the first recommendation that arthrosis of the ankle should be treated by arthrodesis over TAA was given. Many continued to see the potential in arthroplasty, provided the implant design could be refined. As such, numerous designs, all two components with cement fixation to bone, were proposed in
the following decade (Anderson et al., 2003; Buchholz et al., 1973; Buechel et al., 1988; Buechel et al., 2003; Carlsson et al., 1994; Freeman et al., 1978; Kirkup, 1985; Kofoed, 1995; Newton III, 1982; Pappas et al., 1976; Schill et al., 1998; Stauffer and Segal, 1981; Waugh et al., 1976; Wynn and Wilde, 1992). The prostheses were primarily composed of a concave tibial component, constructed from polyethylene, and a convex metal talar component, and could be classified as either constrained or unconstrained designs that were secured with cemented fixation (Yu and Sheskier, 2014). Constrained implants limited free movement in the joint, however, any critical movement necessary to avoid injury would load the implant with unbearable stress, causing complications (Michelson et al., 2000). Unconstrained implants were held under control of the stabilizing ligaments surrounding the joint, permitting unhindered ankle motions. These designs increased joint contact forces due to their incongruent nature (Michelson et al., 2000).

Few designs attempted to replicate the natural anatomy of the joint's articulating surfaces (Freeman et al., 1978; Kempson et al., 1975; Kirkup, 1985; Matejczyk et al., 1979; Pappas et al., 1976). The exception was the Irvine Ankle total ankle arthroplasty, where 32 individual tali were measured to approximate talus morphology in the prosthesis (Waugh et al., 1976). The Irvine design was not perfect. It increased stress on the surrounding soft tissue due to separation caused by component rotation and led to failures of the implant (Evanski and Waugh, 1977).

Clinical trial results in the first generation were discouraging. For example, the Irvine total ankle arthroplasty was implanted 28 times, and reported two failures, although the follow-up was only for nine months (Evanski and Waugh, 1977). The CONAXIAL implant identified 90% loosening at ten years (Wynn and Wilde, 1992), the Mayo implant had a success rate of 19% over an average of a nine-year follow-up (Kitaoka and Patzer, 1996), and the Richard-Smith implant reported loosening of components in 29% of implants after seven years (Kirkup, 1985). Failure arose from component subsidence, loosening of screws, and complications such as infections, and poor wound healing (Gill, 2004). The incongruent designs loaded the prostheses in unpredictable ways when
implanted *in vivo*, and led to increased stress on the implants themselves (Pappas *et al*., 1976) or the surrounding tissues, leading to further damage as observed in the Irvine (Evanski and Waugh, 1977).

Initial implants were over- or under-constrained, implanted with biologically destructive fixation (Kofoed, 1995; Pappas *et al*., 1976), had a high risk of failure, and elevated reported pain levels; unsurprisingly, there were recommendations for discontinuation of their use (Carlsson *et al*., 2001; Demottaz *et al*., 1979; Kitaoka *et al*., 1994; Kitaoka and Patzer, 1996; Newton III, 1982; Stauffer and Segal, 1981; Wynn and Wilde, 1992). In an editorial published in the Journal of Bone and Joint Surgery, Hamblen (1985) concluded that the ankle joint could not be replaced using current techniques of the time, but with advances in technology including prosthetic design this view could be reversed. For a decade following, arthrodesis was again touted as the treatment of choice for ankle arthritis, until a resurgence of interest in the possibility of arthroplasty emerged in the 1990s (Gougoulias *et al*., 2009; Vickerstaff *et al*., 2007).

### 2.6.2 Second Generation Designs

Second-generation implant designs were reduced in quantity, but improved in quality. These designs were more attuned to ankle anatomy, mechanical alignment, and joint stability, and were implanted with a cementless fixation (Valderrabano *et al*., 2012). Many of these designs were adapted from a two-component to three-component implant (which, in addition to the talar and tibial components, utilizes an unconstrained mobile bearing) over time but due to FDA concerns about mobile bearings, the implants approved for use in the U.S. until 2009 were two-component designs (Park and Mroczek, 2011). The two prostheses from the second generation that garnered the most attention in the U.S. were the Agility (DePuy, Warsaw, IN) and the Buechel-Pappas Total Ankle (Endotec, Orange, NJ).
The first-generation low-contact stress (LCS) implant was a modified version of a cylindrical design first proposed by Pappas and colleagues (1976). The LCS implant was further refined into the Buechel-Pappas implant (Buechel and Pappas, 1992). The implant was fully congruent, with rotationally unconstrained meniscal bearings. Over a ten year follow-up of 50 implants, 88% of ankle surgeries were ranked as excellent or good, with pain scores significantly improved (Buechel et al., 2003). In the same study, no tibial component loosening or bearing subluxation was reported, although complications arose from wound healing, infections or malleolar fractures. Talar component subsidence was reported in one individual, and another experienced severe bearing wear (Buechel et al., 2003). Others found promising results with the Buechel-Pappas uncemented device (Ali et al., 2007; Doets, 2006), and reported survivorship of the implant was excellent, with 92% survival rate at 12-year follow-up (Buechel et al., 2004). Despite these findings, the implant is not in circulation in the U.S. (Yu and Sheskier, 2014).

The Agility total ankle prosthesis was comprised of a cobalt-chromium talar component, a titanium tibial component and a modular polyethylene insert that locked to the tibial component (Alvine, 2002). A review of all (n = 132) Agility arthroplasties conducted by a single surgeon between 1984 and 1994 concluded encouraging results following the surgery (Knecht et al., 2004). Revision surgery was performed in 11% of cases, but 90% of patients reported reduced pain and satisfaction following the surgery. Concern was voiced regarding radiographic markers of lucency and lysis, 76% of ankles reviewed showed some peri-implant radiolucency (Knecht et al., 2004). This concern prompted design changes in the Agility implant. Three features were improved in an attempt to minimize complications: a broad-based talar component was redesigned, the polyethylene insert became front loaded, and the ability of surgeons to use a variety of component sizes to best suit the patient’s needs (Cerrato and Myerson, 2008). Until 2007, the Agility implant was the only system cleared by the FDA (510(k)) and was therefore the most widely used by surgeons for over a decade (Roukis et al., 2015). Clinical results were initially encouraging but
were associated with high complication rates (Claridge and Sagherian, 2009). Despite alterations in implant design, the Agility prostheses had little compatibility across functional components, complicating revision options. As other TAA implants gained approval, the Agility systems fell in popularity (Roukis et al., 2015).

The Eclipse (Integra LifeSciences, NJ) is a two-component implant that received a Food and Drug Administration (FDA) approval of use via a 510(k) summary of safety and effectiveness in 2006 (Mulcahy and Chew, 2015). It uses a box stem fixation for the tibial component and a fin fixation for the talar component. The components are constructed from cobalt-chromium, with a polyethylene articulating surface fixed to the tibial component. This implant can be inserted from either a medial or lateral approach (Yu and Sheskier, 2014). Despite relatively recent FDA approval, few surgeons use the Eclipse design in favor of other implants, although it is indicated for use after failed prior ankle surgery (Roukis et al., 2015).

### 2.6.3 Third Generation Designs

Third-generation implants generally expand from the two-component designs to a three-component set-up with a mobile polyethylene bearing component. Two of these designs adopt similarities to the Buechel-Pappas – the Ankle Evolutive System (AES, Biomet) and the Mobility (DePuy, UK) implant. The AES implant has been in use in Scandinavia (Fevang et al., 2007; Henricson et al., 2007) and while short-term results seemed promising, high rates of aseptic loosening (Besse et al., 2009; Henricson et al., 2010; Kokkonen et al., 2011; Rodriguez et al., 2010) eventually caused the implant to be removed from market. The Mobility implant was used in 100 arthroplasties with a survival rate at four-year follow-up of 94% (Wood et al., 2010), and has promising short-term results (Rippstein et al., 2011), although comparison with the Buechel-Pappas design shows worse radiolucency results (Naal et al., 2009). The Mobility has been used in a series of investigations
into kinematic changes following arthroplasty (Valderrabano et al., 2003a, 2003b, 2003c), addressed in a later section of this review.

Another third-generation design took a slightly different approach, with a well-documented development process. The BOX implant (Finsbury Orthopedics, UK) is the result of years of research by Leardini et al. (1999, 2002, 2004). Initial clinical results showed low complication rates, good clinical scores and restoration of ankle mobility (Giannini et al., 2010).

For an in-depth review of implant design, the reader is referred to Cracchiolo and DeOrio (2008), Giannini et al. (2000), Gougoulias et al. (2009), and Vickerstaff et al. (2007).

2.7 Total Ankle Arthroplasty: Prostheses Implanted at Hershey Medical Center

2.7.1 Wright Medical INFINITY™ Total Ankle System

The INFINITY™ Total Ankle System was given FDA clearance in 2013 (Mulcahy and Chew, 2015). This fixed bearing prosthesis uses PROPHECY guidance system, which relies on preoperative images of the ankle to create patient-specific guides for resection (Reb and Berlet, 2017). These preoperative scans predict appropriate tibial component size in 92% of INFINITY™ arthroplasties, but only 46% of appropriate talar component size (Hsu et al., 2015). The INFINITY™ total ankle system has a low-profile tibial component to which an ultra-high molecular weight polyethylene insert is fixed, which articulates with a polished metal talar dome. The tibial component has three pegs angled to provide a strong fixation internally to the tibia, while the talar component has the same geometry to that of the INBONE (Wright Medical Technology, TN), allowing for interchangeable components based on a patient’s talus and surgeon preference (Anderson et al., 2015). Short-term survivorship of the INFINITY™ in one study was 95.3%, with significant improvements in Foot and Ankle Outcome Scores reported across pain, symptoms,
ability to participate in activities of daily life and quality of life (Saito et al., 2018). Two- to four-years following an INFINITY™ TAA, PROMS improved from preoperative scores and 65 of 67 implants survived, with two revisions for aseptic loosening (Penner et al., 2019).

2.7.2 Zimmer Trabecular Metal™ Total Ankle

The Zimmer Trabecular Metal™ total ankle prosthesis (Zimmer, Inc., Warsaw, IN) was cleared for use in the US market in 2012 by the FDA. It is a semi-constrained, fixed-bearing implant that was designed to minimize the amount of bone resection, maximize contact area and to mimic the natural anatomy of the joint (Brigido and DiDomenico, 2015). The tibial component is a trabecular metal made of titanium alloy, allowing for bone ingrowth. This theoretically improves long-term fixation of the prosthesis (Wilson et al., 2010), by minimizing changes in bone mineral density surrounding the implant (Meneghini et al., 2010). Published data on function following a Zimmer Trabecular Metal™ TAA is unavailable at the time of writing. However, a three-dimensional finite element model of the implant was developed from computed tomography images to investigate contact stresses and pressures (Martinelli et al., 2017). The finite element analysis showed that the average pressure at the articulating interfaces was mainly lower than the critical threshold for polyethylene integrity. Their findings were commensurate with finite element analyses of available implants with further research, implying that the implant might have similar survivorship outcomes to other available implants (Martinelli et al., 2017).

2.7.3 Tornier Salto Talaris® Total Ankle Prosthesis

The Salto Talaris® implant is a modified version of the Salto implant, which has been in use in Europe since 1997 (Gaudot et al., 2014). The Talaris received clearance for use in the U.S. by the
FDA in 2006 (Mulcahy and Chew, 2015). The Salto Talaris® implant is comprised of unconstrained tibial and talar components with a fixed polyethylene insert on the metal tibial component. The talar component is anatomically designed, with a smaller medial radius, made of cobalt chromium with a titanium spray coating (Coetzee and Deorio, 2010). The Salto (the implant available outside of the U.S.) has a hydroxyapatite coating, allowing for bone growth implantation.

In the U.S., the Salto Talaris® is intended for cemented use only (Tornier SAS, 2013). The tibial component is inserted to the distal tibia using a pedestal stem, and the polyethylene component of the Salto Talaris® locks to the tibia implant. The tibial component is inserted in the position producing the greatest conformity and least rotation between the components (Coetzee and Deorio, 2010).

A comparison of the fixed-bearing Salto Talaris® implant with the mobile-bearing Salto prosthesis was conducted by a group from France (Gaudot et al., 2014). Based off positioning assessments, clinical and radiographic mobility, and AOFAS scores the authors concluded that in the short-term there was no difference in clinical performance between the two implants.

Eighteen Salto Talaris® arthroplasty patients were included in an investigation on the correlation of PROMs and RoM following arthroplasty (Dekker et al., 2017). The patients receiving a Salto Talaris® implant had a total ankle RoM of 38 ± 12° and all PROMs improved following the surgery at a mean of 44-months.

A larger cohort (67 patients) was followed for a minimum of two years following a Salto Talaris® TAA (Schweitzer et al., 2013). Approximately three years following the surgery, the authors reported a survival of 96%. All patients reported improved pain scores, AOFAS-HF and functional scores (Schweitzer et al., 2013). The functional tests included the sit-to-stand test, the Timed Up-and-Go test (TUG) and average walking speed assessments. Sit-to-stand test and TUG performance improved significantly at both one- and two-years following the arthroplasty.
compared with preoperative values. Average preoperative walking speed was reported as a mean 0.86 ± 0.28 m/s. Two years following surgery this had improved to 1.21 ± 0.17 m/s.

2.8 Functional Changes Following Total Ankle Arthroplasty

The following studies include TAA implants outside those mentioned above due to the availability of the literature. Table 2-1 categorizes whether these implants are fixed- or mobile- bearing, with the year of FDA clearance or approval.

**Table 2-1.** Total Ankle Implant Model, Type, and FDA Status

<table>
<thead>
<tr>
<th>Implant</th>
<th>Bearing Type</th>
<th>FDA Date</th>
</tr>
</thead>
<tbody>
<tr>
<td>INBONE Total Ankle System</td>
<td>Fixed</td>
<td>510(k) clearance, 2010</td>
</tr>
<tr>
<td>(Wright Medical, TN, USA)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>HINTEGRA Total Ankle System</td>
<td>Mobile</td>
<td>510(k) clearance, 2015</td>
</tr>
<tr>
<td>(Newdeal SA, Lyon, France)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Scandinavian Total Ankle Replacement</td>
<td>Mobile</td>
<td>Approval, 2009</td>
</tr>
<tr>
<td>(STAR, Stryker, NJ, USA)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>BOX implant</td>
<td>Mobile</td>
<td>N/A</td>
</tr>
<tr>
<td>(Finsbury Orthopedics, UK)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mobility Total Ankle System</td>
<td>Mobile</td>
<td>N/A</td>
</tr>
<tr>
<td>(DePuy International, UK)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Agility Total Ankle System</td>
<td>Fixed</td>
<td>510(k) clearance, 1992 (first design)</td>
</tr>
<tr>
<td>(DePuy Orthopedics, Warsaw, IN)</td>
<td></td>
<td>510(k) clearance, 2002 (revised design)</td>
</tr>
</tbody>
</table>

2.8.1 Kinematics and Kinetics

Researchers have been conscious of the need to understand kinematics of the joint following TAA for decades (Conti *et al*., 2006; Demottaz *et al*., 1979; Michelson *et al*., 2000). The first kinematic analyses of currently available implants in the U.S. market occurred in 2003. Valderrabano and
colleagues published a series of three papers exploring changes in range of motion (Valderrabano et al., 2003b), movement transfer (Valderrabano et al., 2003a) and talar movement (Valderrabano et al., 2003c) in six fresh-frozen cadaver leg specimens. These tests were conducted in a non-operated ankle, a fused ankle, and an ankle which had either an Agility, HINTEGRA or STAR arthroplasty. Surgical intervention reduced RoM, although the effect was attenuated in the TAA specimens (Valderrabano et al., 2003b).

The second study in the series found the arthroplasty implants reduced the level of movement transfer between eversion and tibial rotation (Valderrabano et al., 2003a). Talar rotation and shift relative to the tibia during plantarflexion/dorsiflexion was investigated in the third study of the series, where HINTEGRA and STAR prostheses were not significantly different to the normal ankle (Valderrabano et al., 2003c). The outcomes of the three studies suggested that unconstrained prostheses with a mobile bearing more closely mimicked the natural kinematics of the ankle joint.

The effect of STAR implant positioning on RoM was investigated in six cadaver ankles (Tochigi, 2005). The talar component was positioned in neutral, 3mm, and 6 mm displacement both anteriorly and posteriorly. Maximal RoM was obtained with the talar component implanted in neutral (51.5±7.6°), although a slightly posterior insertion did not have a negative influence if joint height was preserved (Tochigi, 2005).

In patients, RoM following TAA has been reported from 22° (Salto prosthesis, Ajis et al., 2013) to 34° (Salto Talaris® prosthesis, Pedowitz et al., 2016). This RoM is similar to that reported in healthy individuals during gait and larger than that found in patients with osteoarthritis prior to surgical intervention (Nüesch et al., 2012). Valderrabano et al. (2007b) reported approximately 30° RoM one year following HINTEGRA TAA, half that observed in healthy controls. Dorsiflexion torque reached 89% of the healthy controls a year following the arthroplasty, while plantarflexion torque reached 80% of the controls, with no reported change in calf circumference (Valderrabano et al., 2007a).
RoM and torque production were positively influenced following arthroplasty, but did not fully reach levels of healthy controls (Ajis et al., 2013; Flavin et al., 2013; Rouhani et al., 2012; Seo et al., 2017; Singer et al., 2013; Valderrabano et al., 2007b). Some research has investigated the kinematics of prosthetic components during functional tasks as a way to explain some of the persisting deficits (Conti et al., 2006; Leszko et al., 2008; Renate List et al., 2012). Small anteroposterior translation (< 3.5 mm) of the contact position between the two components was found (Conti et al., 2006; Leszko et al., 2008), with greater variation of the contact point between the tibia and talus in the uninvolved limb. For inversion/eversion and internal/external rotation, greater variability was observed in the implanted ankles (Conti et al., 2006). No difference in prosthetic kinematics was observed between two groups of Mobility arthroplasty patients, those who presented with pain and those who were pain-free (List et al., 2012).

Component alignment was explored during the stance phase of gait (Yamaguchi et al., 2012). Component tilt and incongruence of the articulating surfaces were observed in four ankles, and the authors postulated that the changes in contact pressure and rotational torques might lead to component failure. Finite element analyses of the BOX prosthesis during simulation of the stance phase found maximal contact pressure of 10.3 MPa on the tibial component and 16.1 MPa on the talar component (Reggiani et al., 2006). Maximal force was observed during the latter part of gait in four ligaments (tibial calcaneal, posterior talofibular, deep anterior and deep posterior tibiotalar). These loads are comparable to static analyses of other prostheses (Elliot et al., 2014; Martinelli et al., 2017; McIff and Horton, 2004). These are high pressures in a joint - in comparison, contact pressure in the knee are reported at 3 – 6 MPa (Fukubayashi and Kurosawa, 1980), thus if prosthetic components are misaligned, it can lead to failure. For further detail, I refer the reader to a review of TAA designs from the perspective of stresses in the ankle joint (Kakkar and Siddique, 2011).
### 2.8.2 Gait Analysis

Jastifer and colleagues (2015) assessed 61 STAR arthroplasty patients in situations they will encounter in daily life. Walking performance on inclines, uneven surfaces, and stairs was assessed using a visual analog scale to rate difficulty of each task. Prior to TAA, patients found walking upstairs to be the most challenging task (6.6, where 0 was ‘no difficulty’ and 10 was ‘impossible’), followed by walking on an uneven surface (5.4) and walking downstairs (5.2). Walking downhill (4.4) was harder than walking uphill (3.7), while a level flat surface was the easiest (2.8). A year following TAA, the hardest tasks were walking on an uneven surface and walking downstairs (both rated as 1.8), while flat and uphill surfaces were the easiest (0.5). These tasks were all performed by patients while wearing their own footwear.

Gait parameters have been shown to be influenced by the use of shoes (Clarke et al., 1983; Jørgensen, 1990; McNair and Marshall, 1994; Nigg and Segesser, 1992; Stacoff et al., 2000), and there is no consensus as to whether patients should have their gait analyzed while shod or barefoot. Queen and Nunley (2010) found spatiotemporal parameters including stride length, step length, double support time and walking velocity were increased while wearing shoes. A pedobarographic investigation found maximal forces were increased in both the hind-foot and mid-foot when wearing running shoes, although contact time was no different (Arno and Roman, 2016). Queen and Nunley (2010) suggested for ease of comparison of gait parameters in clinical studies, patients should be analyzed barefoot in order to eliminate the influence of different footwear which may not be consistent between testing sessions or centers.

The effect of TAA on muscle activity has been investigated in patients who received the BOX prosthesis (Caravaggi et al., 2015; Cenni et al., 2013; Ingrosso et al., 2009). Gait analysis was conducted on 79 patients before, six- and 12- months following TAA, with concurrent electromyographic signals recorded from the biceps femoris, tibialis anterior, and gastrocnemius...
(Ingrosso et al., 2009). The authors suggested that the biceps femoris recovered functionality a year following TAA, and that co-contraction during mid-stance was observed pre-operatively and six months post-operatively in the tibialis anterior and gastrocnemius. Cenni et al. (2013) expanded on this observation, noticing the phasic activity between the tibialis anterior and the gastrocnemius was similar to controls while climbing stairs, but still exhibited some abnormality during stair descent. These persisting deficits may be due to insufficient follow-up; at five years post-TAA, electromyographic activity in the aforementioned muscles was not different from a control population (Caravaggi et al., 2015).

Comparison studies of fixed-bearing TAA and arthrodesis have been conducted. Both TAA and ankle fusion patients exhibit improvements compared to preoperative function and pain (Chopra et al., 2014; Flavin et al., 2013; Hahn et al., 2012; Piriou et al., 2008; Rouhani et al., 2012; Segal et al., 2018; Seo et al., 2017; Singer et al., 2013). TAA patients showed greater improvements across the studied gait parameters, although neither group’s function was restored to that of healthy controls (Flavin et al., 2013; Rouhani et al., 2012; Seo et al., 2017; Singer et al., 2013). Compared with controls, the stance phase of the gait cycle was longer, while the push-off duration was shorter in both operative groups, although TAA patients recovered gait symmetry while the arthrodesis patients maintained asymmetries (Chopra et al., 2014). This was highlighted by plantar pressure data; the arthrodesis group showed significant differences in over half the sub-regions between the two limbs, with minimal changes in TAA patients. These findings are in agreement with the symmetry of stance phase reported a year post-ankle arthroplasty by Piriou and colleagues (2008) and Flavin et al. (2013). TAA patients walked faster than arthrodesis patients, and were observed to have increased dorsiflexion RoM (Flavin et al., 2013; Singer et al., 2013).

Segal et al. (2018) showed that TAA patients increased cadence, with step length increased only in their uninvolved limb. This is in contrast to the findings of Piriou et al. (2008), who noticed no change in cadence between TAA and arthrodesis patients or healthy controls.
Ankle power, extension moment and ankle moment at heel rise were lowest in the arthrodesis patients, slightly elevated in the TAA patients, and greatest in the control group (Singer et al., 2013). Segal et al. (2018) observed no difference in peak joint moments or powers in the sagittal-plane between interventions, although the heel strike transient was significantly larger in arthrodesis patients, and occurred more frequently. This might be explained by the reduced ankle RoM placing the foot in a less favorable position at ground contact, although this was not explored by these authors.

Rouhani et al. (2012) divided the ankle and foot complex into the tibial-calcaneus, calcaneus-metatarsal and tibial-metatarsal joints, each of which had a reduced RoM in the osteoarthritic condition compared with controls. An arthrodesis only improved RoM of the calcaneal-metatarsal joint in the transverse plane, while there were persisting impairments reported in the other joints. The authors reported no significant differences in RoM in the TAA patients when compared with controls.

When gait was analyzed using a multi-segmental foot model, hind-foot sagittal RoM was significantly larger in TAA patients compared with arthrodesis patients (Seo et al., 2017). The increase in RoM was due to hind-foot dorsiflexion, although forefoot RoM also was significantly larger in TAA patients. One year following surgery, arthrodesis patients walked with greater hip RoM while TAA patients had increased ankle RoM (Hahn et al., 2012). This change in ankle RoM might explain why peak plantarflexion moment was greater in fused ankle patients and reduced in TAA patients compared with their pre-operative values.

TAA appears to outperform arthrodesis across many studies, so the focus of the remaining review is on fixed-bearing TAA gait analysis. Wang and Brown (2016) analyzed gait mechanics on five patients pre- and three months post- Salto Talaris® TAA. They reported an average improvement in walking speed of 0.19 m/s following the surgery, which is of smaller magnitude than that found by Schweitzer et al. (2013). However, Wang and Brown (2016) also reported an improvement in
RoM following TAA, an increase in peak plantarflexor moment, joint power absorption, joint power production, stride length, and a reduction in time spent in double support. This gait analysis study was in a very small cohort, and further studies are necessary to identify changes in gait following a Salto Talaris® arthroplasty. Another study, in which gait analysis was conducted three months following TAA, reported a worsening in kinematic and kinetic variables compared with preoperative values, before the improvements are noted at six- and 12-month follow-up (Valderrabano et al., 2007b).

Choi et al. (2013) compared preoperative temporospatial, kinematic and kinetic gait parameters to those at two-year follow-up following a Salto Talaris TAA. They noted increased walking velocity that was attributable to increases in both cadence (in agreement with Segal et al., 2018) and step length, in addition to reduced double support time. Ankle RoM significantly increased, primarily due to an increase in dorsiflexion range, agreeing with others (Flavin et al., 2013; Singer et al., 2013). Peak ankle power was also significantly increased post-operatively, although the authors reported no change in peak ankle moment.

Performance on functional tests, joint mechanics and GRF during walking were collected pre- and two-years post-operatively in 51 INBONE and the Salto Talaris TAA patients (Queen et al., 2012). In contrast to some (Choi et al., 2013; Flavin et al., 2013; Singer et al., 2013), the observed dorsiflexion RoM did not change across time points but all other variables of interest, including performance on the Four Square Step Test and TUG, improved from pre-operative levels. Pain was significantly improved, as measured by the AOFAS-HF.

Two years later, the same research group studied 78 patients who received either the INBONE or the Salto Talaris® implant (Queen et al., 2014c). They reported faster walking speed and improved gait symmetry, although this was insufficient improvement to regain a symmetrical walking pattern. This was best demonstrated by the dorsiflexion angle at heel-strike, which increased in the non-surgical limb but not on the involved side.
A total of 140 TAA patients underwent physical performance measures which were correlated with PROMs (McConnell and Queen, 2017). PROMs significantly improved between preoperative scores at both one and two- years post-operatively, and performance measures improved across time. The authors reported moderate correlations between PROMs and performance on the physical tasks, implying each provides unique information about recovery following TAA.

The effect of pre-operative alignment on vertical GRF, and lower limb mechanics during level walking was investigated in 93 patients (Grier et al., 2016). Ankle RoM, knee abduction at heel strike, hip adduction at heel strike and peak hip adduction increased significantly following TAA, regardless of alignment group. The authors concluded that pre-operative alignment did not appear to cause substantial differences in gait mechanics following the surgery, although they noted this was a two year follow-up and that prosthetic wear and long-term effects of dynamic loading may not be visible at such a time point.

2.8.3 Balance

Static balance following a HINTEGRA TAA was shown to be no different to control participants, but dynamic posture was poorer, with less reliance on the ankle to maintain balance (Lee et al., 2010). Patients who received either a hip, knee, or ankle arthroplasty were asked to perform a Single Leg Balance test (SLBT) approximately a year following their joint replacement (Butler et al., 2015). Only 9% of TAA patients (total 94 ankles) were able to maintain SLBT for ten seconds, showing a detrimental performance in comparison to the total hip (63% of 75 patients passed) and total knee (69% of 65 patients passed) patients. The pain relief associated with TAA two years following surgery was shown to improve static balance, without additional therapeutic intervention (Powell et al., 2017).
Chapter 3

Estimates of Achilles Tendon Moment Arm Differ When Axis of Ankle Rotation is Derived from Ankle Motion


3.1 Introduction

The plantarflexor moment arm of the Achilles tendon affects plantarflexor function through its combined influence on both leverage and muscle fiber length and shortening velocity, both of which affect muscle force generation. The plantarflexion moment arm of the Achilles tendon (ATma) determines the mechanical advantage of the triceps surae relative to the ground reaction force during the latter part of stance phase in activities requiring transmission of calf muscle forces (Carrier *et al.*, 1994). In addition, ATma determines the extent of muscle fiber shortening for a given ankle joint rotation (Lieber and Ward, 2011), and thus affects plantarflexor muscle operating points on the muscle force-length and force-velocity curves. Because ATma determines plantarflexor muscle moment and force generation, it is a critical determinant of the mechanics of push-off during gait (Rasske *et al.*, 2017; Takahashi *et al.*, 2016).

ATma may be found experimentally as a geometric distance on a two-dimensional (2D) magnetic resonance (MR) image of the ankle joint (Baxter and Piazza, 2014; Maganaris *et al.*, 1998; Rugg *et al.*, 1990). In the 2D geometric method, the center of rotation (CoR) between the tibia and talus is found from images of the ankle made with the ankle in different positions, and the ATma is taken to be the shortest distance from the CoR to the midline of the Achilles tendon. ATma have also
been found from three-dimensional (3D) MR images of the ankle complex, either using dynamic cine-MR (Sheehan, 2012) or static images (Clarke et al., 2015).

Ultrasound imaging of the Achilles tendon has also been employed to make estimates of ATma. Muscle moment arm may be found from measurements of tendon excursion and joint angle (Storace and Wolf, 1979), and many investigators have used this tendon excursion method to determine ATma (e.g., Fath et al., 2010; Ito et al., 2000; Spoor and van Leeuwen, 1992). The tendon excursion method does not require the location of a CoR, but does assume that all tendon excursion is attributable to joint rotation rather than tendon stretch and relaxation that are difficult to quantify in vivo (Olszewski et al., 2015). For this reason, the tendon excursion method as currently implemented may not be appropriate to quantify ATma during functional activities. Manal et al. (2010) described a hybrid method combining dynamic ultrasound imaging of the Achilles tendon with 3D tracking of the markers placed on the foot, the tibia, and the ultrasound probe. The Achilles tendon was located from ultrasound images while the motion data were used to identify and track the CoR, which was taken to be the midpoint between the medial and lateral malleoli. The ATma was then estimated throughout the motion by finding the shortest 3D distance between the CoR and the tendon midline. The combination of ultrasound imaging and motion tracking employed in this way allows for dynamic assessment of ATma during weight-bearing tasks such as walking (Rasske et al., 2017).

Results from studies using the hybrid method have produced ATma estimates that differ with those from MR-based studies, potentially due to errors in the location of the CoR. In both 2D and 3D MR studies, ATma has been found to be 8 – 15 mm larger during muscle contraction (Hashizume et al., 2014; Maganaris et al., 2000). Differences in ATma between conditions of maximum voluntary contraction and rest (Manal et al., 2013) and between stance and swing phase during walking (Rasske et al., 2017), however, have been found to be substantially smaller when hybrid methods have been used. MR-based estimates of ATma have also consistently shown ATma to increase with
plantarflexion angle (Hashizume et al., 2012; Maganaris, 2004; Sheehan, 2012), a trend that has not been evident when hybrid methods that assume a transmalleolar midpoint CoR have been used (Franz et al., 2019; Manal et al., 2013; Rasske et al., 2017). Previous efforts to assess ATma geometrically using ultrasound have specified the CoR as the transmalleolar midpoint (Manal et al., 2013; Rasske et al., 2017). While studies of relative bone motions have found that the CoR is well approximated by a transmalleolar midpoint (Siston et al., 2005), others have revealed differences between the landmark-based transmalleolar axis and the axis of tibiotalar rotation (Lundberg et al., 1989; Lundberg and Svensson, 1993; Sammarco, 1977). A rotation axis derived from marker-based tracking of the relative motion of the foot and shank might better represent the true axis of tibiotalar rotation and thus permit better estimates of ATma. The purpose of this study was to investigate differences in ATma determined using the ultrasound-motion analysis approach subject to three different methods for approximating the axis or center of ankle rotation. Specifically, we sought to compare ATma results from experiments in which a ‘functional axis’ is computed from measured ankle motion to results from the same trials in which the axis or center of rotation is determined using markers placed on the malleoli. The three representations of the ankle axis or center of rotation considered were: (1) functional axis (FA) identified using finite helical axis decomposition of the relative motions of the foot and shank; (2) transmalleolar axis of rotation (TA) through markers placed over the medial and lateral malleoli; and (3) the midpoint between the malleolar markers (TM). We hypothesized that: (i) ATma from FA will be larger than those estimated using TA or TM; (ii) a larger change in ATma will be seen with plantarflexion angle using FA than with TM or TA; and (iii) ATma computed with FA will increase with loading to a greater extent than will ATma computed with TM or TA.
3.2 Methods

Fifteen healthy young adults (8 F, 7 M; age: 26 ± 2 y; height: 1.70 ± 0.07 m; mass: 71 ± 12 kg) were recruited to participate in the study. The protocol was approved by the Institutional Review Board of The Pennsylvania State University, with informed consent obtained prior to data collection for each participant.

Calibration

In order to locate the tendon within the laboratory frame of reference, a static transformation between a reference frame defined by markers attached to the ultrasound probe frame and a frame aligned with the planar ultrasound image was found using a calibration procedure prior to data collection for each subject. Briefly, three cylindrical rod phantoms were imaged underwater using a 60 mm linear ultrasound probe (Telemed HL9.0/60/128Z-2; Lithuania). Retroreflective markers rigidly attached to either end of each rod and four markers attached to the ultrasound probe determined the locations of the rods and the probe in the laboratory reference frame. The centers of the intersections of each rod with the imaging plane were identified and used to create a homogeneous transformation between the probe and image frames. Further details of this calibration procedure are provided in the supplemental materials.

Motion trials

Following probe calibration, 12 mm-diameter retroreflective markers were attached with double-sided tape on the right lower limb over both malleoli and both femoral epicondyles. Two four-marker clusters on molded plastic plates were affixed to the anterior shank and the dorsum of the foot using double-sided tape and elastic wrap (3M Coban; St. Paul, MN). The Achilles tendon was imaged with the middle of the ultrasound probe placed vertically at the level of the malleoli and with the probe oriented to maximize the appearance of the tendon in the image (Figure 3-1).
Figure 3-1. Schematic illustration of ultrasound probe placement over the Achilles tendon, with a sample ultrasound image showing the Achilles tendon within the white box.

Figure 3-2. Illustrations of the reflective markers attached to the skin of a subject along with the ultrasound probe affixed over the Achilles tendon using a custom probe-holder. Clusters of reflective markers were secured to the dorsum of the foot and the anterior shank, and individual markers were placed over the lateral and medial malleoli (LM and MM) and lateral and medial femoral epicondyles (LE and ME). An anterior view is shown on the left and a posterolateral view on the right.
The probe was secured in place with a neoprene brace and foam support to minimize the motion of the probe relative to the leg (Figure 3-2). Participants performed standing toe raises in time to a metronome at 0.5 Hz. Ultrasound images, with 30 mm depth and at a frequency of 7 MHz with a dynamic range of 62 dB, were collected over three toe-raise cycles at a rate of approximately 60 frames per second, while marker coordinates were sampled at 100 Hz using eight Eagle cameras (Motion Analysis Corp.; Santa Rosa, CA). The above task was repeated three times, after which participants were asked to perform unloaded cyclical plantar- and dorsi-flexion motions at the same frequency while standing on the left leg with the right foot raised from the floor and extended in front; this motion was also repeated three times. These tasks were chosen to produce ranges of motion that were similar between the loaded and unloaded tasks and also similar to that of walking, without many of the attendant motions of the rest of the body that would occur during walking itself.

The ultrasound beamformer (Telemed LogicScan 128; Lithuania) produced a 5 V square-wave analog signal during image sampling. This signal was recorded using a data acquisition board (National Instruments PCIe-6259; Austin, TX) and Cortex software (Motion Analysis Corp.; Santa Rosa, CA) on the motion analysis system computer. The rising edge of this signal (plus an experimentally-determined time-delay) was used in subsequent analysis to synchronize the ultrasound images to the motion capture data.

\textit{ATma calculation}

Ultrasound images of the Achilles tendon and marker locations were analyzed using a custom-written MATLAB (Mathworks Inc., Natick, MA) script. The superficial and deep tendon borders were located in each grayscale ultrasound image by first applying the histeq() histogram equalization transformation in MATLAB with two bins to obtain a black-and-white image. Borders parallel to the image margins between white and black regions were then found automatically by marching in the deep and superficial directions from an initial user-identified starting point in the
center of the tendon. The tendon force line of action was taken to be the tendon midline that was 
midway between these two borders. The transmalleolar axis (TA) was defined as the line passing 
through the medial and lateral malleoli markers, while the coordinates of the transmalleolar 
midpoint (TM) were found as the spatial mean of the two malleoli.

The foot and shank coordinate systems were established following ISB recommendations (Wu et 
al., 2002). All marker coordinates were filtered using a bidirectional 4th order low-pass Butterworth 
filter with a cut-off frequency of 10 Hz. The coordinates of the foot and shank cluster markers were 
used to compute homogeneous transformations between the foot and shank segment coordinate 
systems using a least-squares fitting technique (Challis, 1995).

Finite helical axes were computed, at each time frame, using the shank-to-foot transformations 
from 25 frames (0.25 s) before and 25 frames after that time frame (Spoor and Veldpaus, 1980). 
As axes computed from skin-mounted markers are sensitive to noise when the magnitude of 
rotation is small, any helical axis for which the absolute value of the rotation about the axis was 
less than 0.2 rad was discarded (Woltring et al., 1985). A single axis representing the confluence 
(i.e., the “mean”) of all the remaining helical axes was found using a previously described 
procedure (Lewis et al., 2006) and this axis was termed the “functional” axis (FA).

For the methods that made use of an axis of rotation (FA and TA) rather than a point center of 
rotation, ATma was computed by first finding the moment of a unit force \( \vec{f} \) along the tendon line 
of action about an arbitrary point P on the axis of rotation:

\[
\vec{M}_p = \vec{PQ} \times \vec{f} \tag{1}
\]

where \( \vec{PQ} \) is the vector pointing from P to an arbitrary point Q on the tendon line of action.

Taking P’ to be a second point on the axis of rotation and \( \vec{u} \) to be a unit vector along PP’, the 
moment of the unit tendon force about the axis was computed as the projection of \( \vec{M}_p \) onto \( \vec{u} \):

\[
M_{P,P'} = \vec{u} \cdot (\vec{PQ} \times \vec{f}) \tag{2}
\]
Finally, ATma was found by shifting the terms in this scalar triple product and taking the absolute value to account for uncertainty in the direction of $\vec{u}$ along the axis of rotation:

$$ ATma = \left| \overrightarrow{PQ} \cdot (\overrightarrow{u} \times \overrightarrow{f}) \right| $$  

(3)

For the TM method, the shortest distance from the transmalleolar midpoint to the line of action of the Achilles tendon was computed and taken to be the ATma. Ankle joint angles were computed using ZXY Cardan decomposition (with $z$: medial-lateral; $x$: anterior-posterior; $y$: superior-inferior) of the rotation matrices between the shank and foot coordinate systems at each time frame. Second-order polynomials were fit to plots of ATma versus plantarflexion-dorsiflexion angle and interpolations of these polynomials were used to estimate moment arms at 5° increments from +10° (dorsiflexion) to -20° (plantarflexion).

**Statistical analysis:** A three-way repeated-measures ANOVA determined differences in ATma between methods (TA, TM, FA), loading conditions (loaded and unloaded) and ankle angle (+10°, +5°, 0°, -5°, -10°, -15°, and -20°) in SPSS (v23, IBM, USA) with the level of statistical significance set at $\alpha = 0.05$. When ANOVA indicated significance, simple effects post hoc analyses were conducted using a Bonferroni correction for multiple comparisons. Pairwise Pearson’s correlations were computed between ATma assessed in neutral ankle position in both loading conditions to identify associations between ATma computed using the three methods.

### 3.3 Results

Mean ATma found using FA were significantly larger (all $p < 0.001$) than those from TA (mean differences of 14.0 mm and 9.5 mm for loaded and unloaded conditions, respectively) and TM (mean differences of 11.0 mm and 6.3 mm), while TM moment arms were larger than TA for loaded (mean difference of 3.0 mm, $p = 0.011$) and unloaded (mean difference of 3.2 mm, $p = 0.006$), Figure 3-3). These patterns of differences were found across all joint angles examined.
(p = 0.024). The difference was most evident at 20° plantarflexion, for which FA values were 29% larger than TA values, and 24% larger than TM values, on average (Figure 3-3).

![Figure 3-3](image)

**Figure 3-3.** Achilles tendon moment arms calculated with a ‘functional’ ankle rotation axis (FA, filled circles), a transmalleolar rotation axis (TA, open circles), or a transmalleolar midpoint (TM, open squares) in loaded (A) and unloaded (B) conditions, plotted versus ankle angle. Plantarflexion angles are negative and dorsiflexion angles are positive. Error bars indicate standard deviation (n = 15).

When ATma was determined using the FA, a dependence on loading condition was observed. ATma from FA at maximum plantarflexion (20°) were on average 7.8 mm larger when loaded than when unloaded (p < 0.001). This significant difference remained but the effect was attenuated at smaller plantar flexion angles (all p < 0.07). As the ankle moved into dorsiflexion, however, no significant differences between loaded and unloaded ATma were observed for the FA method (all p > 0.05).

Loaded TA moment arms were marginally but not significantly larger (difference of 0.3 mm) than unloaded (p = 0.523). Within-angle comparisons revealed that loaded TA moment arms were not significantly different at any angle (all p > 0.278). There was a significant difference between TM moment arms measured at the two loading conditions at only the most plantarflexed position;
loaded TM moment arms were 1.27 mm larger than unloaded ($p = 0.046$), but at no other angles were the differences significant.

ATma from FA in the loaded condition were found to increase in magnitude from $10^\circ$ dorsiflexion to $20^\circ$ plantarflexion (Table 3-1). When unloaded, ATma increased from neutral position to $20^\circ$ plantarflexion. While some small but significant differences across joint angles were found for the TA and TM methods, the relationship between ATma and joint angle was less pronounced for TA and TM than for FA.

Table 3-1. Average differences between ATma values estimated at $20^\circ$ plantarflexion, neutral, and $10^\circ$ dorsiflexion for each loading condition found with each of the three methods (FA, TA, and TM). $P$-values are for Bonferroni-corrected post hoc mean comparisons between angle, with significant differences ($p < 0.05$) highlighted in bold type.

<table>
<thead>
<tr>
<th>Angle Comparison</th>
<th>FA Loaded</th>
<th>FA Unloaded</th>
<th>TA Loaded</th>
<th>TA Unloaded</th>
<th>TM Loaded</th>
<th>TM Unloaded</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean Difference (mm)</td>
<td>P-value</td>
<td>Mean Difference (mm)</td>
<td>P-value</td>
<td>Mean Difference (mm)</td>
<td>P-value</td>
</tr>
<tr>
<td>$20^\circ$ PF - $10^\circ$ DF</td>
<td>12.752</td>
<td>&lt; 0.001</td>
<td>5.099</td>
<td>0.008</td>
<td>0.981</td>
<td>1.000</td>
</tr>
<tr>
<td>$0^\circ$ - $10^\circ$ DF</td>
<td>5.575</td>
<td>&lt; 0.001</td>
<td>1.675</td>
<td>0.207</td>
<td>0.368</td>
<td>1.000</td>
</tr>
</tbody>
</table>

Pairwise Pearson’s correlations between FA, TA and TM in an unloaded neutral position revealed significant relationships in each case (Figure 3-4). Using standards set forth by Cohen (1988), TA and TM were strongly correlated ($r^2 = 0.853$, $p < 0.001$) while strong correlations were also found between FA and TM ($r^2 = 0.689$, $p < 0.001$), and FA and TA ($r^2 = 0.474$, $p = 0.005$). These correlations became slightly weaker for ATma measured in neutral position for the loaded condition (see Figure A3 for these plots).
Figure 3-4. Pairwise Pearson’s correlations of Achilles tendon moment arm in neutral (0°) position and in the unloaded condition. Correlations were made between moment arms measured using different methods: (A) FA and TA ($r^2 = 0.474$, $p = 0.005$), (B) FA and TM ($r^2 = 0.689$, $p < 0.001$), and (C) TA and TM ($r^2 = 0.853$, $p < 0.001$). Lines of agreement ($y=x$) are displayed with gray dashes.

3.4 Discussion

Our estimates of ATma measured using an ultrasound-motion tracking hybrid method differed substantially when computed with an FA as compared those found using TA or TM representations of ankle joint axis or center of rotation (Figure 3-3). Specifically, ATma from FA were larger, varied with plantarflexion angle more (such that ATma were larger in plantarflexion) and increasing with loading to a greater extent. Estimates of ATma that approximated the ankle using
anatomical landmarks (*i.e.*, TA or TM) did not vary appreciably with ankle joint angle or loading condition. Because the estimated position of the Achilles tendon was constant across the FA, TA, and TM methods, differences in ATma directly reflect differences in ankle axis position determined by those three methods.

To better understand why ATma computed with FA changed with joint angle and those computed with TA did not, we represented ATma as the product of two factors: (1) the shortest distance $D$ (distance along the mutual perpendicular) between the tendon axis and each joint axis; and (2) a factor of $\sin \beta$, where $\beta$ is the angle between the tendon axis and the joint axis:

$$ATma = D \sin \beta \quad (4)$$

Values of $D$ and $\sin \beta$ for a representative subject are shown in Figure 3-5. The shortest distance between the tendon and the FA varies considerably for both loaded and unloaded conditions, increasing with plantarflexion (Figure 3-5A). However, the shortest distance between the tendon and TA varies considerably less, and shows a small increase with dorsiflexion. The angular differences between the tendon axis and either joint axis are small; the tendon is nearly perpendicular to both FA and TA ($\sin \beta$ nearly unity) throughout the range of motion (Figure 3-5B). Variation in $D$ over the range of motion for FA is due to the obliquity of the FA axis; this is evident in a three-dimensional plot of the FA, TA, and tendon axes in dorsiflexion and plantarflexion for this representative subject that we have included in the supplementary material (Appendix A, Figure A4).
Our findings suggest that ATma estimated using FA provide a closer approximation of ATma estimated using MR-based geometric methods (Figure 3-6). ATma that increased with plantarflexion were observed when using FA, a pattern similar to that previously reported in 2D MR studies (Fath et al., 2010; Hashizume et al., 2012; Maganaris et al., 1998; Rugg et al., 1990). The same pattern was not observed when using TA or TM, commensurate with previously published results using a similar method (Manal et al., 2013; Rasske et al., 2017). The results from previous studies presented in Figure 3-6 along with the results of the present study suggest that ATma measured using techniques in which the center or axis of rotation is located based on ankle motion are larger than ATma measured using a center or axis of rotation defined by the malleoli. It is important to note, however, that comparisons of ATma magnitude across investigations must be interpreted with caution because ATma differences may be attributable in part to differences among participant groups such as body size.

**Figure 3-5.** (A) The shortest distance D between the functional axis (FA, thick lines) or transmalleolar axis (TA, thin lines) and the Achilles tendon for both loaded (solid) and unloaded (dashed) conditions for individual trials of a representative subject. (B) The sine of the angle (β) between these same axes. Variation in ATma with joint angle seen for FA but not TA was primarily attributable to variation in D with joint angle.
Figure 3-6. Loaded Achilles tendon moment arms found using a functional axis (FA, red), transmalleolar axis (TA, green), and a transmalleolar midpoint (TM, blue) plotted versus ankle angle. As in Fig. 3, plantarflexion angles are negative and dorsiflexion angles are positive. Moment arm values from previous studies, computed using ultrasound/motion capture (1 and 2), 2D MR (3-6), and 3D MR (7 and 8), are plotted in gray.

Loading condition influenced ATma computed with FA (Figure 3-3), in agreement with findings from MR studies. We found a 10% increase in FA moment arm for our loaded condition as compared to unloaded across all joint angles. Maganaris et al. (1998) reported a 22 – 27 % increase in ATma from MR between rest and maximal voluntary contraction (MVC) of the triceps surae, which they attributed to a change in tendon orientation due to muscle bulging with contraction. Hashizume et al. (2014) reported an approximately 15% increase in ATma with 30% of maximum effort. The toe-raise activity we used for our loaded condition likely did not produce maximal contractions, suggesting that our findings are in line with these previous studies. As in previous studies using the TM method, we measured only a small change (0.2 %) in ATma between loaded
and unloaded conditions. Manal et al. (2013) noted a 3.5% difference between MVC and passive conditions when using the TM method.

The functional axis (FA) and the transmalleolar axis and midpoint (TA and TM) were in roughly the same location for each participant, so it was perhaps unsurprising that the ATma estimated in neutral ankle position derived using these three methods were strongly correlated with one another (Figure 3-4). Only a little more than half the variance in FA moment arm was explained by TM or TA, however, suggesting that ATma found using FA were not simply scaled-up versions of their TM and TA counterparts. The bone kinematic measurements of Lundberg et al. (1989) showed TM to be a good approximation of the intersection of the tibiotalar axes of rotation derived throughout the range of ankle motion. It may be that variation in rotation axis orientation not captured when TM is used affects ATma estimates in important ways. Similarly, the findings of Lundberg et al. (1989) show that the ankle plantarflexion axis of rotation is similar to TA in the transverse plane, but when projected onto the sagittal-plane, substantial differences in axis inclination with TA were observed. The findings of Lundberg et al. (1989) are in agreement with other studies of bone kinematics (Isman and Inman, 1969; Lundberg, 1989; Sammarco et al., 1973).

Certain limitations affected this study, including the assumption that the Achilles tendon force line of action is located midway between the superficial and deep tendon borders. We computed FA from clusters of markers mounted on the tibia and the dorsum of the foot. As such, it is likely that the tibia-to-foot transformations included motions other than ankle motions, including talocalcaneal motion, tarsal and tarsometatarsal joint motion, and skin movement relative to the bones. The confounding effects of some of these extraneous motions could have been avoided if we had placed our foot cluster on the heel, but we found it difficult to do so without interfering with the ultrasound probe cluster. Foot deformation with foot loading may be substantial (Leardini et al., 2007; Xiong et al., 2009), and it is possible that the observed differences in ATma between loading conditions for FA are an artefact of such foot deformation rather than differences in rotation
axes. It is possible that inconsistency in the placement of the probe relative to the tendon could have contributed to variability in our moment arm estimates. When we tested the same subjects on different days, however, the differences in moment arm were typically 1 – 2 mm, a level of repeatability that we deemed acceptable relative to the differences in moment arm noted between the two methods and across our subjects.

The FA we used was a single fixed axis but studies of bone kinematics show that the axis of tibiotalar rotation moves throughout the range of ankle motion (Lundberg, 1989; Lundberg et al., 1989). We chose a fixed axis in the present study so that we could make a more direct comparison to the fixed transmalleolar axis (TA). Future work will build on the present study by considering the effects of using a moving instantaneous helical axis on estimates of ATma, as has been done in previous 3D MR studies (Sheehan, 2012, 2010).

 Representing ankle joint axis of rotation with a functional axis derived from relative motions of the foot and shank produced ATma that increased in magnitude with plantarflexion and loading, and was larger than those obtained from anatomical landmarks. ATma estimated using anatomical landmark representations of the ankle (TA and TM) failed to show substantial variation with either joint angle or loading condition.
Chapter 4

Achilles Tendon Moment Arms Are Similar When Computed Using A Single Fixed Axis Versus A Moving Instantaneous Helical Axis

This paper is under review for publication at the Journal of Biomechanics

4.1 Introduction

Muscle moment arms are essential determinants of muscle mechanical advantage that have been associated with locomotor function. At the ankle joint, the plantarflexion moment arm of the Achilles tendon (ATma) has been correlated with preferred gait velocity in older adults (Lee and Piazza, 2012), plantarflexor strength in young adults (Baxter and Piazza, 2014), and running economy (Scholz et al., 2008). Measurements of the ground reaction force and its point of application have been combined with measures of ATma to make subject-specific estimates of Achilles tendon force and stress (Lichtwark and Wilson, 2005). In computational simulations of movement, predictions of muscle function have been shown to be sensitive to the choice of muscle moment arm (Ackland et al., 2012), and metabolic cost and work rates during hopping tasks have been found to be sensitive to ATma (Voigt et al., 1995). The results of these studies suggest that accurate and reliable methods for experimentally determining subject-specific ATma are needed for understanding plantarflexor muscle function and the contributions of the triceps surae to locomotion.

Several methods have been proposed for estimating ATma in vivo, including approximation of ATma as the horizontal distance from the lateral malleolus to the heel in photographs (e.g., Scholz et al., 2008), measurement of the distance from the ankle joint center to the tendon on two-
dimensional magnetic resonance images (e.g., Maganaris et al., 2000; Rugg et al., 1990), and measures of ATma based on three-dimensional magnetic resonance imaging (e.g., Sheehan, 2012). ATma may also be estimated from the Achilles tendon excursion (assessed using ultrasound or magnetic resonance imaging) that occurs with respect to plantarflexion angle (e.g., Fukunaga et al., 1996), but these tendon excursion approaches rely on a tenuous assumption that tendon tension is maintained during in vivo movement. ATma has also been estimated using a hybrid method that combines ultrasound imaging of the Achilles tendon and video-based motion analysis (Manal et al., 2010; Manal et al., 2013). In this hybrid method, the tendon is located using an ultrasound probe that is tracked using the same motion analysis system that is used to locate the center of ankle rotation. Because it does not require magnetic resonance imaging, the hybrid method is relatively easy to implement and permits measurement of ATma during locomotor activities (e.g., Rasske and Franz, 2018).

We previously implemented a hybrid method combining ultrasound and motion analysis (Wade et al., 2019) to estimate ATma from: (1) a center of rotation (at the midpoint of the malleoli) similar to that used in previous implementations of this method (Manal et al., 2013; Rasske et al., 2017; Rasske and Franz, 2018); (2) a transmalleolar axis; and (3) an axis of ankle rotation derived from finite helical decomposition of measured ankle motion. This latter axis was a single axis that was fixed with respect to the shank and estimated from plantarflexion-dorsiflexion motions. Estimates of ATma made using this latter motion-derived axis were found to be similar to those computed from two-dimensional magnetic resonance imaging studies in which a center of rotation was also found from ankle motion. When compared to ATma values from three-dimensional magnetic resonance imaging (Hashizume et al., 2012; Sheehan, 2012), the moment arms we computed using the motion-derived axis were similar across the mid-range of motion, but differed in dorsiflexion and extreme plantarflexion. These differences are perhaps unsurprising given that the axis of ankle rotation has been demonstrated to vary with ankle motion (Lundberg, 1989), while the axis we used
was a single fixed axis estimated from a set of finite helical axes. The effect of assuming a fixed axis of rotation on ATma estimation was considered in one previous study (Rugg et al., 1990) employing two-dimensional magnetic resonance imaging; it was found that a moving center of rotation produced marginally (3%) smaller estimates of ATma than did a fixed center of rotation across the plantarflexion-dorsiflexion range of motion. The effect of assuming a fixed axis of ankle rotation in a three-dimensional analysis, however, has not been investigated to date.

The aim of this study was to determine whether hybrid-method estimates of ATma, made using an ankle axis allowed to move with ankle motion and determined using instantaneous helical axis (IHA) decomposition, differ from ATma estimated using a single fixed axis of ankle rotation (FA).

To these ends, we compared ATma estimated using the two techniques for joint axis estimation applied to motions measured in the same set of healthy subjects. We hypothesized that the two axis estimation techniques would produce similar ATma in the mid-range of ankle motion, but that ATma would differ at the ends of the range of motion (when the rotation axis deviates most from its orientation in the mid-range of motion), with IHA producing smaller ATma values than FA (based on the previous finding of Rugg et al., 1990).

4.2 Methods

Healthy participants (n = 20, 10 F, 10 M, age: 26.17 ± 2.79 y; height: 1.71 ± 0.01 m; mass: 71.48 ± 13.75 kg) performed a series of weight-bearing (toe-rise) and non-weight-bearing (seated with leg extended and elevated) cyclical plantar- and dorsiflexion motions at a rate of 0.5 Hz in time to a metronome. The experimental protocol for data collection has been outlined previously (Wade et al., 2019), but we summarize the methods briefly here. Reflective markers were placed on anatomical landmarks of the right leg, with additional rigid clusters of four markers each on the shank, the foot, and the ultrasound probe (HL9.0/60/128Z-2; Telemed, Lithuania). All procedures
were approved by the Institutional Review Board of The Pennsylvania State University, and all participants provided informed consent prior to testing.

Ultrasound images of the Achilles tendon were acquired by placing the ultrasound probe parallel to the tendon, at the level of the malleoli and securing it using a Velcro brace. These images were sampled at approximately 60 frames per second and synchronized to the marker coordinate data, which were collected at 100 Hz using seven Eagle cameras (Motion Analysis Corp.; Santa Rosa, CA). Synchronization was achieved using a 5 V square-wave signal that was emitted by the ultrasound beamformer (Telemed LogicScan 128; Lithuania) at the initiation of image sampling. This signal was recorded using a data acquisition board (National Instruments PCIe-6259; Austin, Texas) and Cortex software (Motion Analysis Corp.; Santa Rosa, CA) on the motion analysis computer. The rising edge of this signal (including an experimentally determined time-delay) was used to synchronize the ultrasound images to the motion capture data in the subsequent analyses.

Data were processed using routines custom-written in MATLAB (2018b, Mathworks Inc., Natick, MA). The Achilles tendon line of action was identified on the ultrasound images as the midline between the superficial and deep tendon borders and transformed into the laboratory reference frame. A bidirectional 4th order low-pass Butterworth filter with a cut-off frequency of 5 Hz was applied to all marker coordinates. Anatomical coordinate systems attached to the shank and foot were defined with reference to the recommendations set forth by the International Society of Biomechanics (Wu et al., 2002). Finite helical axes were computed from pairs of positions separated by rotations of at least 0.25 radians (Spoor and Veldpaus, 1980; Wade et al., 2019). The FA was defined as the single axis with the least deviation (in a least-squares sense) from all qualified finite helical axes (Lewis et al., 2006).

Instantaneous helical axis decomposition was applied throughout the motion as follows. A 4 x 4 homogeneous transformation matrix formed between the shank and the foot anatomical reference frames ($T_{SF}$) was determined using a least-squares fitting technique applied to the coordinates of
the foot and shank cluster markers (Challis, 1995). This transformation matrix was decomposed to obtain Euler/Cardan angles and a translation vector $\vec{t}$ at each time frame. Generalized cross-validated splines (Woltring et al., 1987) were fit to each of the three Euler/Cardan angles and each of the components of $\vec{t}$ with respect to time. The spline-fit angles were reconstituted into an orthonormal rotation matrix, $R$, representing the rotation between the shank and foot. At each time frame $i$ the time derivative of $R$ was approximated by:

$$\dot{R}_i = \frac{R_{i+1} - R_{i-1}}{2\Delta t}$$

(1)

where $\Delta t$ is the time step between successive frames of motion data. The skew-symmetric angular velocity matrix ($\omega$) was computed at each time frame according to (Craig, 1989):

$$\omega_i = \dot{R}_i \times R_i^T$$

(2)

The components of the angular velocity vector ($\vec{\omega}_i$) in the shank frame were then extracted from $\omega_i$. A unit vector directed along the IHA (defined in the shank reference frame) was found by normalizing $\vec{\omega}_i$:

$$\vec{u}_i = \frac{\vec{\omega}_i}{|\vec{\omega}_i|}$$

(3)

A point ($p_i$) on the IHA (in the shank reference frame) was defined in terms of $\vec{V}_i$, the rate change of $\vec{t}$ between frames $i + 1$ and $i - 1$, found using numerical differentiation (Sommer, 1992).

$$p_i = \vec{t}_i + \frac{\vec{\omega}_i \times \vec{V}_i}{|\vec{\omega}_i|^2}$$

(4)

The point on the IHA ($p_i$) and the unit vector along the IHA ($\vec{u}_i$) were transformed from the shank reference frame into the laboratory frame using the homogeneous transformation matrix ($T_{65}^i$) incorporating the spline-fit reconstructed $R_i$ and $\vec{t}_i$. 

As differentiated velocities obtained from the skin-mounted markers are sensitive to noise (De Lange et al., 1990), such as when rotations are small when movements are slow, IHA axes were deemed valid only when $|\vec{\omega}| > 0.8 \text{ rad s}^{-1}$. Therefore, the usable RoM in which ATma could be computed from IHA is less than that for FA-derived ATma.

At each time frame corresponding to those used to compute the IHA, the moment of a unit force ($\vec{f}$) along the line of action of the Achilles tendon about an arbitrary point A on the axis of rotation (i.e., $p$ in Eq. 4) was calculated:

$$M_A = \overline{AB} \times \vec{f}$$

(6)

where $\overline{AB}$ is the vector pointing from A to an arbitrary point B on the Achilles tendon line of action. If $A'$ is taken as a second arbitrary point on the axis of rotation, the moment of $\vec{f}$ about the axis was computed as the magnitude of the projection of $M_A$ onto $\vec{u}$:

$$M_{AA'} = \vec{u} \cdot (\overline{AB} \times \vec{f})$$

(7)

Finally, ATma was found by rearranging the terms in the scalar triple product and then taking the absolute value to account for uncertainty in the direction of $\vec{u}$ (i.e., $\vec{u}$ versus $-\vec{u}$):

$$ATma = |\overline{AB} \cdot (\vec{u} \times \vec{f})|$$

(8)

A repeated-measures three-way ANOVA was run in SPSS (v23, IBM, USA) to explain differences in ATma computed with an FA or IHA across five ankle angles (ranging from $5^\circ$ dorsiflexion to $15^\circ$ plantarflexion in $5^\circ$ increments) across loading conditions (weight-bearing and non-weight-bearing). Where significance was indicated, post-hoc simple effects analysis was conducted with a Bonferroni correction for multiple comparisons. Pairwise Pearson’s correlations were run to identify the relationship between ATma computed from a fixed and moving axis of rotation at each of the angles of interest. The level of statistical significance was set at $\alpha = 0.05$. 

\[
T_{LS}^{I} = \begin{bmatrix}
0 & R_i & \vec{z}_i \\
0 & 0 & 1
\end{bmatrix}
\]
4.3 Results

Table 4-1. Correlations between Achilles tendon moment arms computed with instantaneous helical axes of ankle rotation and Achilles tendon moment arms computed from a fixed axis of ankle rotation at five discrete angles of ankle rotation and two loading conditions. Positive angles represent dorsiflexion and negative angles represent plantarflexion.

<table>
<thead>
<tr>
<th>Angle</th>
<th>Loaded r</th>
<th>Loaded p-value</th>
<th>Unloaded r</th>
<th>Unloaded p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>+5°</td>
<td>0.310</td>
<td>0.184</td>
<td>0.647</td>
<td>0.002</td>
</tr>
<tr>
<td>0°</td>
<td>0.659</td>
<td>0.002</td>
<td>0.844</td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>-5°</td>
<td>0.696</td>
<td>0.001</td>
<td>0.898</td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>-10°</td>
<td>0.619</td>
<td>0.004</td>
<td>0.908</td>
<td>&lt; 0.001</td>
</tr>
<tr>
<td>-15°</td>
<td>0.478</td>
<td>0.033</td>
<td>0.904</td>
<td>&lt; 0.001</td>
</tr>
</tbody>
</table>

ATma estimated using the two methods of finding ankle axis of rotation (FA or IHA) were correlated (Table 4-1). The correlation between ATma from FA and IHA tended to be stronger for the non-weight-bearing condition at 0° (Figure 4-1), and in the center of the range of motion (0°, 5°, 10°).

Figure 4-1. Scatter plots of Achilles tendon moment arms computed with a fixed axis of rotation and with an instantaneous helical axis of ankle rotation at a neutral ankle angle for a) weight-bearing and b) non-weight-bearing conditions. Unity (y = x) is shown on each plot using a dashed line.
There were significant main effects of method \((p = 0.001)\), joint angle \((p < 0.001)\), and loading condition \((p = 0.001)\). There was also a significant three-way interaction effect between method x joint angle x loading condition \((p = 0.034)\), as well as significant two-way interactions between method x angle \((p = 0.009)\), and angle x load \((p < 0.001)\).

Mean ATma computed from IHA were, on average, 2.29 mm smaller than those computed from FA \((F_{1,19} = 4.294; \ p = 0.052; \ \eta^2 = 0.678)\) but the magnitude of this difference was angle- and load-dependent \((F_{3,17} = 6.297; \ p = 0.005; \ \eta^2 = 0.526)\). Post-hoc analysis of mean ATma within the loaded condition (Figure 4-2A) revealed that there were no significant differences between methods at any joint angle, except the most dorsiflexed position tested, where they differed by 9.35 ± 3.23 mm \((p = 0.001)\). However, during the non-weight-bearing condition (Figure 4-2B), there were significant differences between ATma from IHA and FA at all angles investigated \((all \ p \leq 0.015)\), with the exception of the most dorsiflexed position. At 5° dorsiflexion, IHA produced smaller estimates of ATma than FA \(\text{mean difference } 3.33 \text{ mm}\) but this difference did not reach the level of statistical significance \((p = 0.072)\). These differences were approximately constant across joint angles, with IHA producing estimates of ATma that were on average 3 mm smaller than those from FA.
Figure 4-2. Achilles tendon moment arm (ATma) computed with a fixed axis of rotation (FA, ●) or a moving axis (IHA, ▲) against ankle plantarflexion angle for weight-bearing (A) and non-weight-bearing (B) conditions. Positive angles represent dorsiflexion and negative angles represent plantarflexion. Note that IHA-derived ATma are reported over a reduced range of motion than ATma derived from FA, due to the IHA being ill-defined at low-velocity, such as that observed at the end ranges of motion.

4.4 Discussion

This study is the first to determine, using axes of ankle rotation, how estimates of ATma differ when computed with a moving, instantaneous helical axis compared with a fixed axis. We found that during weight-bearing, such differences in ATma were present only in dorsiflexion. Our hypothesis that ATma estimated using the IHA would be smaller than those estimated from FA was partially accepted because the IHA produced a 21% smaller estimate of ATma at 5° dorsiflexion than did the FA, and the IHA yielded a 6% smaller estimate of ATma in neutral position. In plantarflexion, however, the IHA yielded ATma that were on average just 3% larger than those from the FA. Rugg et al. (1990) also found small differences in moment arm when
moving and fixed points of rotation were used in a two-dimensional analysis, but these differences were in the opposite direction, with 3% smaller moment arms for a center of rotation that moved with ankle motion. In our study, the magnitude of differences in ATma between the IHA and the FA was 3.4 mm, comparable to the mean difference 3.1 mm difference reported between ATma computed from a moving axis and a fixed transmalleolar axis from dynamic magnetic resonance (Sheehan, 2012). Sheehan (2012) found that moment arms increased from dorsiflexion to plantarflexion, a finding similar to our own, with ATma leveling off with further plantarflexion.

To understand better how the movement of the axis is related to the differences in ATma that we observed, we conducted further analysis of the moving axis orientation across a range of joint angles in a representative subject. This analysis revealed that in plantarflexion (red lines, Figure 4-3) the IHA was higher on the medial side and oriented closer to the FA than in dorsiflexion, when the axis was more horizontal or slightly elevated on the lateral side. The distance from the FA to the posterior calcaneus where the Achilles tendon inserts is greater than the distance from the IHAs found in dorsiflexion to the same point (Figure 4-3, frontal view), thus providing a visual explanation for the differences in the two computed ATma in dorsiflexion. These IHA axis orientations during loading are similar to those published from studies of bone motion (Arndt et al., 2004; Lundberg et al., 1989).
Figure 4.3. Ankle axes of rotation during weight-bearing, with color changing from dorsiflexion (blue line) to plantarflexion (red line), with the functional axis (solid black line) for a representative subject. The foot is shown in a plantarflexed position to facilitate viewing of the axes. Note that in the frontal plane view, the functional axis (black line) is farther from the posterior calcaneus than the dorsiflexed instantaneous helical axes (blue line).

In a previous study in which ATma were computed from IHA (Sheehan, 2012), similar results to those of the present study were found, in that the magnitude of ATma reduced in dorsiflexion. However, Sheehan (2012) reported ATma in plantarflexion were relatively constant across the range of motion studied. There were differences in methodologies between the present study and Sheehan that should be acknowledged. Sheehan directly obtained angular velocity of bones from cine-magnetic resonance imaging, and used an angular velocity threshold of $|\omega| > 0.3$ rad s$^{-1}$. This threshold is lower than the $0.8$ rad s$^{-1}$ threshold, a difference that was necessitated due to the additional noise and uncertainty involved in obtaining velocity from the positions of skin-mounted markers. Our use of skin-mounted markers is an inherent limitation as we could not track the
motions of the bones directly. We attempted to reduce the effects of noise with both low-pass filtering and spline fitting of the Euler angles and translations.

For the non-weight-bearing condition, in plantarflexion and neutral ankle positions, ATma computed from IHA were significantly smaller ($p < 0.05$) than those computed from FA, by a mean difference of approximately 3 mm. In dorsiflexion, this pattern persisted, although the mean difference did not reach the level of significance ($p = 0.072$). To explore further why the mean difference in the non-weight-bearing condition is smaller than in the weight-bearing condition, we plotted the IHA orientations again in dorsiflexion (blue line, Figure 4-4) and plantarflexion (red line, Figure 4-4) compared with the FA (black line, Figure 4-4). The difference in axis orientations between plantarflexion and dorsiflexion was found to be smaller in the non-weight-bearing condition than in the weight-bearing conditions (Figure 4-3). Three-dimensional magnetic resonance imaging has been used to compute ATma with an FA and passive rotation (Clarke et al., 2015), and produced comparable estimates of ATma to those we produced with FA. This implies that weight-bearing, not just muscle activation, has an influence on the motion of the axis of rotation. The effect of weight-bearing on the moving axis location may be explained by weight-bearing leading to forced conformity in the ankle joint, as has been observed at the knee (Hsieh and Walker, 1976). Another potential limitation of our study is that we estimated the Achilles tendon line of action as a straight line through the midpoint of the superficial and deep borders of the tendon image. However, an investigation into changes in Achilles tendon curvature across active and passive ranges of motion showed that a simple straight-line model of the tendon is an appropriate assumption (Obst et al., 2014).
Figure 4-4. Ankle axes of rotation during non-weight-bearing, with color changing from dorsiflexion (blue line) to plantarflexion (red line), with the functional axis (solid black line) for a representative subject. The foot is shown in a plantarflexed position to facilitate viewing of the axes.

Whilst the IHA we measured deviated from a single fixed axis throughout the range of motion, the ATma calculated using FA and IHA were commensurate across much of the studied range of motion. The greatest difference between the ATma estimated using FA and IHA occurred in dorsiflexion. During walking, the greatest plantarflexor moments and power generation occur when the ankle is plantarflexed in late stance. As such, the differences between the methods of computing axes of ankle rotation may have limited impact when applied to studies of human locomotion. Our findings suggest that both FA and IHA produce similar estimates of ATma. Indeed, use of FA to compute ATma may be preferable because ATma estimation is restricted to a limited range of motion for IHA due to the IHA being ill-defined at points of low velocity seen at the ends of the range of motion studied.
Chapter 5

Changes in Locomotor Function and Pain Following Total Ankle Arthroplasty

This paper is in preparation for submission at the Journal of Orthopaedic Research

5.1 Introduction

Total ankle arthroplasty (TAA) is a surgical treatment for end-stage osteoarthritis at the ankle that is indicated after failure of conservative treatment options. Historically, the surgical treatment for ankle osteoarthritis was tibiotalar arthrodesis, in which the tibia is fused to the talus (Nihal et al., 2008). Fusion of bones that severely restricts range of motion is not ideal, but was preferred to TAA in the past due to insufficient implant designs. Recent implants, however, show improved survivorship and outcomes (O’Connor et al., 2018) and, as a result, TAA is growing in prevalence, with an increase from 524 surgeries in 2005 to 1514 in 2012 (Law et al., 2018). Another study using the Nationwide Inpatient Sample database also reported an increase in TAA as a treatment for ankle arthritis from 9% in 2006 to 26% in 2010 (Raikin et al., 2014).

TAA comprises of resection of arthritic bone from the tibia and talus, followed by implantation of a metal tibial component, a metal talar component, and a polyethylene bearing that is either fixed to the tibial component or partly free to move between the two metal components. Mobile bearing designs are popular in Europe (Gougoulas et al., 2009) and have been the primary focus of TAA research, given one of the designs (Scandinavian Total Ankle Replacement, Stryker, NJ) underwent a comprehensive controlled Food and Drug Administration trial. Mobile bearing designs are associated with increased risk of bearing subluxation (Stengel et al., 2005; Zaidi et al., 2013).
Fixed-bearing implants are more prevalent in the U.S., and are considered to be more reliable and offer improved survivorship, with better clinical outcomes (Gaudot et al., 2014) and lower risk of revision (Roukis and Elliott, 2015).

Outcomes following TAA are largely positive, with high levels of satisfaction (e.g., Hendy et al., 2018; McConnell and Queen, 2017; Nunley et al., 2012; Pedowitz et al., 2016; Queen et al., 2014b; Saltzman et al., 2010; Tenenbaum et al., 2017). Several authors have noted patients report alleviation of pain and improvement in quality of life, up to 15 years following TAA (Brunner et al., 2013; Cody et al., 2018; Daniels et al., 2015; Saito et al., 2018), with increased activity levels and a return to sport for many (Bonnin et al., 2009). In stark contrast to other lower limb replacements, approximately 22% of TAAs require revision surgery within the first five years (Labek et al., 2011), mainly due to infections or loosening of components (Cody et al., 2019), and a systematic review identified residual pain in up to 60% of cases (Gougoulias et al., 2010). One study found no relationship between implant kinematics during gait and pain following TAA (List et al., 2012), but they did not consider overall locomotor function. As yet, no study has considered the relationship between the alleviation of pain and improvement of function following TAA.

The relationship between pain and walking velocity has been investigated to a limited extent in other lower limb joints. For example, patients with low back pain walk slower, with shorter steps, than healthy controls (Keefe and Hill, 1985). Patients with patellofemoral pain also were shown to walk slower than pain-free controls, with reduced peak loading rate of the involved limb in both self-selected and fast walking (Powers et al., 1999). Those with knee pain walk 11% - 24% (Manetta et al., 2002; Powers et al., 1999) slower than those who are pain free. When pain-relieving injections were used in those with osteoarthritic knee pain, level walking velocity increased by 5%, but the pain relief was insufficient to improve stair stepping (Shrader et al., 2004). These studies
suggest an association between pain and walking speed, and that reducing pain does increase walking velocity, but it is unclear how pain and walking speed are related following TAA.

Walking velocity is substantially reduced in those with ankle osteoarthritis when compared with age-matched adults (Khazzam et al., 2006; Nüesch et al., 2012; Stauffer et al., 1977; Valderrabano et al., 2007b), but seems to be restored to healthy levels 12-months following TAA (Valderrabano et al., 2007b). Restoring walking velocity should be a priority of TAA, as it is an indicator of health in older adults (the primary population who undergo TAA), with a 12% increased risk of mortality for every 0.1 m/s reduction in gait velocity (Studenški et al., 2011). Whilst spatiotemporal parameters (including cadence and stride length in addition to velocity) return to levels of healthy controls one year following TAA, deficits may persist in total plantarflexion range of motion and ankle kinetics (Valderrabano et al., 2007b). However, most gait analysis following TAA considers patients walking only a short distance within a gait laboratory (Brodsky et al., 2011; Demottaz et al., 1979; Doets et al., 2007; Flavin et al., 2013; Hahn et al., 2012; Queen et al., 2017; Roselló Añón et al., 2014; Singer et al., 2013; Valderrabano et al., 2007b) and thus may be more conducive for measuring maximal gait speed as opposed to typical gait speed (Graham et al., 2008). Gait velocity over a longer distance may be more affected by pain, and so a longer duration task, such as the Six-Minute Walk (6MW), might be more appropriate for an accurate representation of typical gait speed.

Locomotor function assessed using simple tasks in a clinical setting has advantages over more complex gait analysis protocols in that these tasks are faster to implement and do not require any specialist equipment, but provide useful information on the rehabilitation process and function following TAA. However, such assessments have been sparsely reported in the literature. Queen et al. (2012) used the Four Square Step Test as a measure of step stability and the Timed Up-and-Go (TUG) as a measure of walking function in patients before and after TAA. Another assessment of
function in those undergoing lower limb joint replacement (Wylde et al., 2012) used the TUG, a sit-to-stand task and a Single Leg Balance task (SLBT), which is representative of standing balance and can be a good predictor of fall risk or balance impairments (Springer et al., 2007).

There have, to date, been no studies of the relationship between performance between common clinical assessments of locomotor function and measures of pain in patients before and after TAA. The purposes of this study were (1) to examine pre- to post-operative changes in function in TAA patients along with comparisons to healthy controls, and (2) to determine the extent to which the observed changes are attributable to reduction in pain. We hypothesized that at six months following total ankle arthroplasty, function would improve compared to pre-operative levels (i.e., 6MW velocity would increase, SLBT time on the involved limb would increase, TUG time would reduce, pain levels would decrease) but would not reach the levels observed in healthy controls. We further hypothesized that observed improvements in function would be related to reductions in pain.

5.2 Methods

Participants and study design

Fourteen TAA patients (age: 64.04 ± 9.26 y, height: 1.71 ± 0.07 m, body mass: 98.55 ± 20.07 kg) and 15 age-matched control subjects (age: 63.14 ± 7.95 y, height: 1.68 ± 0.10 m, body mass: 75.57 ± 13.66 kg), provided informed written consent prior to participation in the study. All procedures involving testing of human subjects were approved by the Penn State Hershey Medical Center Institutional Review Board. The subjects in the control group had healthy ankles and lower limbs, and underwent all study tasks at a single time-point, in our Penn State University Park campus laboratory. The patients scheduled to receive a fixed-bearing total ankle arthroplasty, who met all
inclusion criteria and were willing to participate in the study performed all tasks pre-operatively and six months post-operatively in a motion laboratory on the Penn State Hershey Medical Center Campus. Patients were excluded if they were ineligible for total ankle arthroplasty, underwent a dual-limb arthroplasty, were younger than 45, or were unable to provide informed written consent. A list of concomitant procedures for all 14 patients can be found in Table 5-1.

Incomplete datasets were obtained for three patients. One patient withdrew from the study after the initial visit, as they elected not to undergo the surgical procedure, and their data were excluded from the study. One patient was unable to complete the 6MW test pre-operatively and could not perform any of the functional assessments post-operatively due to hip problems, and was therefore excluded from the study. One patient had significant pain six months post-operatively, with a distal medial tibial stress response with varus heel, and was immobilized in a boot. Due to these complications, he was removed from the study. Therefore, the results presented are from 11 patients.

Patients received either a Salto Talaris® (n = 5; Integra LifeSciences Corporation, Plainsboro, NJ), a Zimmer Trabecular Metal™ Total Ankle (n = 4; Zimmer Inc., Warsaw, IN), or an INFINITY™ Total Ankle System (n = 2; Wright Medical Group, Memphis, TN), with each surgery performed by one of three experienced foot and ankle specialist surgeons. Choice of implant was at the discretion of the attending surgeon. An anterior approach was used with the Salto Talaris® and INFINITY™ total ankle implants, while a lateral approach was used with the Zimmer Trabecular Metal™ Total Ankle implant.
Table 5-1. Further information regarding participants, including implant received, details of any concomitant surgeries or post-operative procedures, and demographics.

<table>
<thead>
<tr>
<th>ID</th>
<th>Implant Received</th>
<th>Concomitant Surgery</th>
<th>Post-Operative Procedures</th>
<th>Age</th>
<th>BMI</th>
<th>Sex</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibular osteotomy and fibular realignment Deltoïd reconstruction with semitendinosis allograft Medial displacing calcaneal osteotomy First tarsometatarsal plantarflexion fusion</td>
<td>None</td>
<td>67.39</td>
<td>38.63</td>
<td>F</td>
</tr>
<tr>
<td>2</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibular osteotomy with valgus correction and repair Broström anterior talofibular ligament revision reconstruction</td>
<td>None</td>
<td>64.57</td>
<td>34.83</td>
<td>M</td>
</tr>
<tr>
<td>3</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>78.84</td>
<td>28.88</td>
<td>M</td>
</tr>
<tr>
<td>4</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>72.74</td>
<td>33.74</td>
<td>M</td>
</tr>
<tr>
<td>5</td>
<td>Zimmer Trabecular Metal Total Ankle</td>
<td>Revision total hip arthroplasty - removed from study</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>Salto™ Talaris</td>
<td>None</td>
<td>Steroid injection</td>
<td>58.54</td>
<td>38.65</td>
<td>F</td>
</tr>
<tr>
<td>7</td>
<td>INFINITY™ Total Ankle</td>
<td>Percutaneous Achilles tendon lengthening Calcaneal closing wedge Dwyer osteotomy to correct the heel varus Talar cyst curettage and grafting with autograft</td>
<td>None</td>
<td>46.99</td>
<td>24.34</td>
<td>M</td>
</tr>
<tr>
<td>8</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Lateral fibular osteotomy Broström anterior talofibular ligament revision reconstruction Percutaneous Achilles tendon lengthening</td>
<td>Gutter debridement, saucerization of distal medial tibia and talus</td>
<td>66.89</td>
<td>28.67</td>
<td>F</td>
</tr>
<tr>
<td>9</td>
<td>Salto Talaris®</td>
<td>Calcaneal osteotomy with a Zimmer 6.5 screw Percutaneous Achilles tendon lengthening</td>
<td>Steroid injection</td>
<td>55.52</td>
<td>44.73</td>
<td>F</td>
</tr>
<tr>
<td>10</td>
<td>INFINITY™ Total Ankle</td>
<td>None</td>
<td></td>
<td>66.49</td>
<td>34.15</td>
<td>M</td>
</tr>
<tr>
<td>11</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>71.17</td>
<td>35.87</td>
<td>M</td>
</tr>
<tr>
<td>12</td>
<td></td>
<td>Elected not to undergo total ankle arthroplasty surgery</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibula correction of valgus deformity and malunion of varus repair First tarsometatarsal plantar flexion fusion Percutaneous Achilles tendon lengthening</td>
<td>None</td>
<td>58.35</td>
<td>25.71</td>
<td>M</td>
</tr>
<tr>
<td>14</td>
<td>Salto Talaris®</td>
<td>Removed from study at 6-month follow-up due to complications</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
Patient reported outcome measures

Each participant completed the RAND 36-Item Health Survey 1.0 (SF-36), from which we extracted the physical component score and the bodily pain score (scoring as outlined in Hays et al., 1993). For each sub-score, a higher score indicates better function/less pain. Participants verbally answered the American Orthopaedic Foot and Ankle Society – Hind-foot Scale (AOFAS-HF), in which 50 points are assigned to function, 40 to pain, and 10 points to hind-foot alignment with a higher score indicating better function (Kitaoka et al., 1994). Pain scores from the AOFAS-HF are delineated into four categories (‘None’; ‘Mild, occasional’; ‘Moderate, daily’; ‘Severe, almost always present’), therefore only the global pain score from the SF-36 was used in our analysis. Both the SF-36 and the AOFAS-HF have been used extensively to evaluate outcomes for foot and ankle procedures (Lau et al., 2005; Safavi et al., 2019).

Functional assessments

Figure 5-1. Patients were asked to perform the following functional tasks: A) Single Leg Balance Time (SLBT) on the involved and contralateral limbs, B) Timed Up-and-Go, and C) a six minute walk.
Each participant in both the patient and control groups performed three tasks to assess their locomotor function (Figure 5-1). First, they were instructed to perform a SLBT on both the involved (i) and contralateral (c) limbs while barefoot. Control subjects were instructed to stand on the leg they felt most comfortable balancing on to determine their ‘contralateral’ limb, and the less comfortable leg was deemed their ‘involved’ limb. This task required each participant to balance unilaterally, with eyes open for as long as possible until either (1) a ceiling time of one minute was reached, (2) the raised foot touched the ground or the other limb, or (3) the foot on the floor moved in an effort to maintain or regain balance. The SLBT has been used previously with lower extremity joint arthroplasty patients (Butler et al., 2015), and has found to be a reliable and valid measure of balance (Flansbjer et al., 2012; Roberts, 2016; Springer et al., 2007).

The TUG task was performed with participants wearing their own shoes. In this test, participants rose from a chair (seat height: 46 cm), walked forward three meters, turned, and returned to the seated starting position as quickly as possible without running (Podsiadlo and Richardson, 1991). This measure has been found to have excellent reliability, and to be predictive of safe ambulation (Podsiadlo and Richardson, 1991).

The final functional task was a 6MW, which was conducted in a straight and level hallway. Participants were read standardized instructions, to ‘walk as fast as possible, without running or jogging’ (Crapo et al., 2002), before proceeding to walk for six minutes, pivoting around cones placed at the start and 30 m marks. The 6MW has been assessed for reliability and validity of physical endurance in older adults (Rikli and Jones, 1998), and is a common functional mobility test for older adults (Tiedemann et al., 2008).

**Statistical approach**

Two-tailed paired t-tests were conducted to establish changes in each functional measure and patient reported outcome measures between the two time-points in the patient data. Welch’s two
sample independent t-tests were run on the changes in functional measures and patient reported outcome measures between the pre-operative data and control data, and the post-operative data and the control data. Two-tailed Pearson’s product moment correlations investigated the relationship between the pre-operative and post-operative function. To investigate how much of the change in function could be attributed to a change in pain, linear regression analysis was performed, incorporating covariates of change in pain ($\Delta$pain) and BMI. To determine repeatability, we performed the protocol on five young adult participants (age: 25.88 ± 2.49 y; height: 1.76 ± 0.05 m; weight: 77.14 ± 16.59 kg) over two separate days. A paired-samples t-test was run to see if there were significant differences in 6MW velocity, TUG performance, and SLBT time. All analyses were conducted in R (R Core Team, 2018; www.r-project.org), with significance level $\alpha = 0.05$.

5.3 Results

There were no significant differences in young adult performance over the two testing days (all $p > 0.615$). For 6MW average velocity, there was a mean difference of 0.028 m/s ($p = 0.620$), for normalized average velocity, the mean difference was 0.016 stature/s ($p = 0.615$), and TUG time differed by an average of 0.076 s ($p = 0.839$).

Self-reported outcome measures indicated that patients perceived improvement following surgery, but did not reach the levels of physical function perceived by our control subjects (Table 5-2). Patients perceived the largest improvements in the pain (mean difference = 20 points out of a possible 40) and function (mean difference = 15.19 points out of a possible 50) sub-scores of the AOFAS-HF (both $p < 0.05$). The related sub-scores of the SF-36, role of physical functioning (mean difference post-pre = 31.82) and pain (mean difference post-pre = 34.32), also saw the largest average improvements (both $p < 0.05$).
Table 5-2. Sub-scores (mean ± SD) for the two patient-reported outcome measures before (“Pre”) and six months after (“Post”) TAA, along with control participants scores.

<table>
<thead>
<tr>
<th>Measure</th>
<th>Pre  (mean ± SD)</th>
<th>Post (mean ± SD)</th>
<th>Controls (mean ± SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AOFAS</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total (/100)</td>
<td>36.45 ± 13.50 *</td>
<td>71.55 ± 15.45*</td>
<td>99.47 ± 1.46†</td>
</tr>
<tr>
<td>Pain (/40)</td>
<td>9.09 ± 10.44*</td>
<td>29.09 ± 7.01*</td>
<td>40.00 ± 0.00†</td>
</tr>
<tr>
<td>Function (/50)</td>
<td>20.45 ± 5.79*</td>
<td>35.64 ± 8.05*</td>
<td>49.47 ± 1.46†</td>
</tr>
<tr>
<td>Alignment (/10)</td>
<td>6.82 ± 3.37</td>
<td>6.82 ± 3.37</td>
<td>10.00 ± 0.00†</td>
</tr>
<tr>
<td>SF-36 (RAND)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Physical functioning (/100)</td>
<td>51.82 ± 23.27</td>
<td>62.73 ± 29.27</td>
<td>90.00 ± 16.15†</td>
</tr>
<tr>
<td>Role functioning/physical (/100)</td>
<td>34.09 ± 43.69*</td>
<td>65.91 ± 42.24*</td>
<td>96.67 ± 12.91†</td>
</tr>
<tr>
<td>Role functioning/emotional (/100)</td>
<td>81.82 ± 40.45</td>
<td>84.85 ± 34.52</td>
<td>95.56 ± 17.21</td>
</tr>
<tr>
<td>Energy/fatigue (/100)</td>
<td>57.73 ± 25.22</td>
<td>67.73 ± 25.04</td>
<td>75.33 ± 11.72</td>
</tr>
<tr>
<td>Emotional well-being (/100)</td>
<td>81.82 ± 40.45*</td>
<td>91.27 ± 9.43*</td>
<td>83.47 ± 7.69†</td>
</tr>
<tr>
<td>Social functioning (/100)</td>
<td>72.73 ± 27.28</td>
<td>82.95 ± 26.38</td>
<td>97.50 ± 7.01</td>
</tr>
<tr>
<td>Pain (/100)</td>
<td>34.77 ± 20.81*</td>
<td>69.09 ± 23.62*</td>
<td>92.33 ± 8.58†</td>
</tr>
<tr>
<td>General Health (/100)</td>
<td>72.73 ± 23.06</td>
<td>75.91 ± 17.58</td>
<td>79.33 ± 10.67</td>
</tr>
</tbody>
</table>

*indicates a significant difference post – pre, α= 0.05
†indicates a significant difference controls - post, α= 0.05

Post-operative locomotor function was better than that at the pre-operative timepoint, but when post-operative assessments were compared to results for age-matched control subjects, it was apparent that some deficits persisted (Figure 5-2). At six months following TAA, insignificant increases were observed in 6MW average velocity (mean difference 0.11 m/s, p = 0.072, Cohen’s d = 0.36), normalized 6MW average velocity (mean difference 0.07 statures/s, p = 0.079, Cohen’s d = 0.36), and an insignificant mean reduction of 1.04 s was observed in TUG performance (p = 0.27, Cohen’s d = 0.31). However, SLBT time significantly increased on the involved limb after TAA (mean difference: 6.17 s, p = 0.048, Cohen’s d = 0.52). When performance on these four tasks were compared between patients at six months following surgery and the age-matched control subjects, there were large and statistically significant deficits (all p ≤ 0.001, Cohen’s d ≥ 1.12).
Figure 5-2. Box and whisker plots depicting performance in functional tasks (a) average walking velocity from the six minute walk (6MW velocity), (b) average walking velocity normalized to height (6MW velocity), (c) Single Leg Balance Time (SLBT) on the involved (_i) and contralateral (_c) limbs, (d) Timed Up-and-Go (TUG) time for pre-operative (‘Pre’), post-operative (‘Post’), and control (‘Cont’) data. (a) and (b) also depict average distance (or normalized distance) covered in the six minute walk task.

Pain reduced strongly and significantly following the surgery ($p < 0.001$, Cohen’s $d = 1.09$), with the score improving an average of 34.3 points, although there was still a significant difference compared with controls (mean difference = 24.87, $p = 0.006$, Cohen’s $d = 0.99$). The mean AOFAS-HF score significantly ($p < 0.001$, Cohen’s $d = 1.71$) increased from 36.5 (range: 20 – 57) to 69.1 (range: 51 – 100).
Figure 5-3. Scatter plots depicting performance in functional tasks (a) average walking velocity from the six minute walk task (6MW velocity; $R^2 = 0.426; p = 0.030$), (b) average walking velocity normalized to height (6MW velocity; $R^2 = 0.389; p = 0.430$), (c) Single Leg Balance time on the involved limb (SLBT; $R^2 = 0.483; p = 0.018$), and (d) Timed Up-and-Go time (TUG; $R^2 = 0.051; p = 0.504$) with pre-operative data on the $x$-axis and post-operative data on the $y$-axis. Regression (black line) and unity line ($y = x$; dashed line) also depicted.

Correlations of function (Figure 5-3) between the pre-operative and post-operative time point indicate whether the observed change is related to the pre-operative level of function. For 6MW velocity ($r = 0.653, p = 0.030$), normalized 6MW velocity ($r = 0.624, p = 0.040$), and SLBT time ($r = 0.695, p = 0.018$) pre-operative levels had significant correlation with post-operative function. Pre- and post-operative TUG performance had no correlation ($r = 0.226, p = 0.504$), whilst pain levels had a moderate correlation ($r = 0.646, p = 0.032$).
Table 5-3. Multiple regression output for models investigating the influence of Δpain, BMI and their interaction on change in function.

<table>
<thead>
<tr>
<th>Function</th>
<th>Intercept</th>
<th>Δpain</th>
<th>BMI</th>
<th>Δpain * BMI</th>
<th>Adj. R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>6MW Velocity</td>
<td>-3.415 ***</td>
<td>0.074 ***</td>
<td>0.103 ***</td>
<td>-0.002 ***</td>
<td>0.787 **</td>
</tr>
<tr>
<td>Normalized 6MW</td>
<td>-2.098 **</td>
<td>0.046 **</td>
<td>0.064 **</td>
<td>-0.001 **</td>
<td>0.750 **</td>
</tr>
<tr>
<td>Velocity</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SLBT Time</td>
<td>-8.942</td>
<td>1.033</td>
<td>0.279</td>
<td>-0.026</td>
<td>0.259</td>
</tr>
<tr>
<td>TUG</td>
<td>47.393 *</td>
<td>-0.937 *</td>
<td>-1.414 *</td>
<td>0.027 *</td>
<td>0.482</td>
</tr>
</tbody>
</table>

Significance indicated by: ‘***’ p ≤ 0.001 ‘**’ p ≤ 0.01 ‘*’ p ≤ 0.05

The linear regression model did not predict change in any assessed function from Δpain alone (all p ≥ 0.344). However, when BMI and the interaction of BMI with Δpain was included, moderate to good fits for change in function were obtained for the resulting models (Table 5-3). For example, a 1 point Δpain (lower pain is indicated by a score closer to 100) results in a 0.074 m/s increase in 6MW average velocity, and an increase in BMI (Table 5-3). We found that variation in improvements in pain (when accounting for BMI) accounted for the majority of variation in improvement in average walking velocity, and up to half of the variation in improvement in TUG (Table 5-3).

5.4 Discussion

Our study is the first to relate changes in locomotor function to changes in pain following TAA. Our first goal was to quantify the changes in locomotor function from a pre-operative time point to six-month post-TAA, and compare patient function to a group of age-matched controls. As we predicted, locomotor function at six months post-operation was improved, with self-report scores of function similar to those previously reported in the literature (Dekker et al., 2017; Lachman et al., 2019; McConnell and Queen, 2017; Queen et al., 2014c), at both pre-operative and six months post-operative time points. However, with respect to both self-reported function and outcomes of
functional assessments, there were still deficits when compared to healthy age-matched controls. This is not unexpected as rehabilitation following TAA takes time. Persisting pain, no increase in walking distance or activity levels, and gait abnormalities are still present two-years following TAA and beyond (Claridge and Sagherian, 2009; Valderrabano et al., 2007b). It is reasonable to expect that at six months follow-up, patients are still rehabilitating and should not be expected to have regained full function. However, pain does seem to have the biggest reduction immediately following surgery (e.g., Ingrosso et al., 2009), and so we were curious as to how much this reduction in pain is associated with the improvements in function we observed.

Our hypothesis was that any observed changes in locomotor function would be related to reductions in pain following surgery. However, we found that pain improvement assessed using the pain score from the SF-36, did not explain variation in observed improvements in function. Increased pain levels are associated with decreased mobility in osteoarthritis patients (Zhang and Jordan, 2010), and pain, obesity and mobility appear to be interconnected in osteoarthritis patients (Christensen et al., 2007; Messier et al., 2004). When we included the interaction between pain and BMI in our multivariate regression models of changes in function, we found that almost 80% of the changes in 6MW average velocity could be explained, approximately 50% of TUG time change could be explained, and 26% of the change in SLBT time on the involved limb was captured by pain and BMI.

The interaction of pain and BMI had differing impacts across the functional tasks we studied. Pain and BMI appeared to have a greater influence on those tasks that rely on gait speed (6MW and TUG). Along with previous investigators, we observed an increase in mean gait speed following TAA (Brodsky et al., 2011; Chopra et al., 2014; Demottaz et al., 1979; Detrembleur and Leemrijse, 2009; Doets et al., 2007; Dyrby et al., 2004; Flavin et al., 2013; Ingrosso et al., 2009; Kane et al., 2017; Queen et al., 2014b; Queen et al., 2017; Roselló Añón et al., 2014; Singer et al., 2013;
Valderrabano et al., 2007b). An association between BMI and 6MW velocity in older adults has been reported (Beriault et al., 2009; Camarri et al., 2006; Lee and Piazza, 2012), with a larger BMI associated with slower 6MW velocity. TUG does not depend only on gait velocity, as the task also assess ability to transfer out of a chair and requires good dynamic balance at the pivot point; therefore TUG is a good predictor of functional mobility (Podsiadlo and Richardson, 1991). A reduction in TUG time of 0.9 – 1.4 s has been identified as being clinically important (Mesquita et al., 2016), and therefore the 1.04 s mean reduction from pre- to six months post-operative we observed suggests that patients are improving their dynamic balance and explosiveness in rising from a chair, in addition to gait speed. Given TUG is a short term test of function, it may be easier for patients to “push through” pain than in a 6MW, thus explaining the reduced effect of change of pain and BMI on TUG compared with 6MW velocity. In functional assessments following total knee arthroplasty, 6MW velocity was predictive of longer duration walking ability (e.g., 30 minutes), and the TUG performance was a strong predictor of physical function, and this is also a reasonable assumption for our results (Ko et al., 2013).

We ran a simple regression model to see if BMI alone might explain variation in change in locomotor function. In this model, 6MW average velocity ($R^2 = 0.043$, $p = 0.541$), normalized 6MW average velocity ($R^2 = 0.043$, $p = 0.543$), and TUG performance ($R^2 = 0.168$, $p = 0.210$) were all poorly described just by BMI. Therefore, it appears as though it is specifically the interaction of Δ pain and BMI that has a strong association with walking velocity. However, BMI by itself did seem to best explain the variation in SLBT time on the involved limb due to TAA ($R^2 = 0.385$, $p = 0.042$), compared with the Δ pain, or the interaction of pain change and BMI. SLBT is a challenging task for patients with ankle osteoarthritis (Sun et al., 2011), and pain management, such as lidocaine injections, has been shown to not influence single leg stance time (Riemann et al., 2004), suggesting that pain may not have a large role in balance. Although there are no specific
studies of BMI on SLBT time, other balance tasks are influenced by BMI (Balogun et al., 1994; Greve et al., 2007).

The interaction between Δpain and BMI explained 25.9% - 78.7% of the variation in locomotor function following TAA, therefore other factors must be considered to contribute to improved function following TAA. While age is related to locomotor function, with older adults walking slower and with higher step frequencies than younger adults (Prince et al., 1997), adding age to our model of change in function, pain, and BMI did not improve the fit. Interestingly, our study is not the only one to find that age has little effect on function following lower limb arthroplasty (Jones et al., 2001). TAA alters many mechanical factors, and the macro-effects of the surgery on gait have been demonstrated through studies of gait kinetics (e.g., Brodsky et al., 2011; Valderrabano et al., 2007b), and we observed a deficit in function at six months post-operatively compared with control subjects. These deficits may also be influenced by the change of ankle joint structure caused by replacing biological tissues with uniform implants. The persisting deficits we observed may also be explained by the fact that our patients are still in the rehabilitation phase, and it is unclear if the increased improvements in function following TAA up to two years post-surgery (Valderrabano et al., 2007b) are due to long-term physical therapy rehabilitation causing tissue-level adaptations. Answering these questions is beyond the scope of the current research, but are important directions for future investigations.

The findings we report are based on a small sample of patients undergoing TAA, with a short-term follow-up of six months, where patients are clearly still in the rehabilitation phase and are not expected to have regained full function. Nevertheless, investigating the effect of pain from TAA on post-operative locomotor function is novel and important, indicating other factors beyond alleviating pain must be considered when restoring function. We assessed pain using the global pain score from the SF-36 rather than an ankle specific pain score, and did not collect data on
methods of pain management. However, similar methods have been used in prior TAA literature (e.g., Benich et al., 2017). We collected patient perspectives on their ankle pain and function using the AOFAS-HF, which has recently had concerns raised over its validity (Madeley et al., 2012; Pinsker and Daniels, 2011). However, the score is prevalent in TAA research (Naal et al., 2010) and we included its scores as a minor part of this study for ease with comparison of outcomes to other TAA literature. The sub-scores (pain, function, and alignment) are correlated with the foot function index (Ibrahim et al., 2007), and these are presented in Appendix B for completeness.

Other limitations of our study include the use of three different fixed-bearing total ankle implants. As minimal differences in functional outcomes were found between fixed- and mobile-bearing implants (Queen et al., 2017; Valderrabano et al., 2012), there is likely an insignificant effect of the different type of fixed-bearing implants used in the present study on functional outcomes. There may be an influence of surgical approach, and surgeon experience, but we have insufficient and unequal patient numbers receiving each implant to comment on this. A breakdown of findings by implant type is presented in Appendix B. Finally, we did not control for concomitant or follow-up procedures. Research has indicated that there were equivalent outcomes for patients undergoing TAA who received a concomitant triceps surae lengthening and those who did not (Queen et al., 2014b).

This study indicates that BMI and changes in pain predict changes in 6MW velocity and TUG performance following TAA. Our results agree with our first hypothesis that locomotor function at six months post-operation is improved when compared with the pre-operative function, but has not yet reached the levels of healthy controls. Our second hypothesis, that changes in function would be predicted by changes in pain following TAA was only confirmed when accounting for the interaction with BMI, and for 6MW velocity (both raw, and normalized) and TUG time. BMI was the best predictor for changes in SLBT time. Future studies should investigate the changes in
mechanics, such as alterations in joint structure, and physiology, such as triceps surae muscle and tendon adaptations, on improvements in function following total ankle arthroplasty to optimize locomotor function following TAA.
Chapter 6

Achilles Tendon Moment Arm Changes With Fixed-Bearing Total Ankle Arthroplasty

This paper is in preparation for submission at the Journal of Bone and Joint Surgery

6.1 Introduction

Total ankle arthroplasty (TAA) is a surgical intervention for end-stage arthritis that is performed in growing numbers (Pugely et al., 2014). TAA involves replacing diseased bone surfaces and cartilage with metal and plastic implants, and it has been shown to be effective in reducing or eliminating joint pain (Dekker et al., 2017; Gougoulias et al., 2010; List et al., 2012; Neufeld and Lee, 2000; Schweitzer et al., 2013). TAA implants have been designed with goals to require minimal bone resection, retain natural ligament support, and restore physiological joint function (Gross et al., 2018).

Whilst TAA has been found to improve gait kinematics (e.g., Brodsky et al., 2011; Queen et al., 2012; Valderrabano et al., 2007b) deficits in kinetic gait variables have also been observed. For example, Valderrabano and colleagues (2007b) noted that one year following TAA, total plantar flexion range of motion, maximal plantarflexion moment, maximal adduction moment, and maximal ankle joint power were reduced in patients when compared with control participants. Deficits in ankle joint moments and powers may be attributed to a reduction in the force-generating capacity of ankle joint muscles, but it is also important to consider the alterations to joint structure that may arise following TAA. One key parameter indicative of joint structure with the potential to be influenced by TAA is the plantarflexion moment arm of the Achilles tendon (ATma). ATma
relates the forces produced by the triceps surae to the amount of moment the ankle joint can produce, and is defined mathematically as the component of moment produced by a unit muscle force about the axis of rotation.

ATma is a critical factor in determining the mechanical advantage of the triceps surae as they actively plantarflex the ankle during locomotor activities. For example, recent reports found an association between ATma and ankle plantarflexor moment generation (Baxter and Piazza, 2014; Rasske and Franz, 2018), which is important during the push-off phase of gait (Winter, 1980). A strong, positive correlation between the walking speed of slower-walking elderly adults and ATma was reported by Lee and Piazza (2012). To date, however, no study has investigated how TAA might alter ATma, and what, if any, implications this might have for elderly gait.

In contrast to TAA and ATma, the effects of knee arthroplasty on knee extensor moment arm has received substantial attention (e.g., Browne et al., 2005; D’Lima et al., 2001; Gerus et al., 2013; Hamilton et al., 2013; Pal et al., 2007). For example, D’Lima et al. (2001) tested cadaver knees in a closed kinetic chain knee simulator, under one of three conditions – unoperated; with a control total knee replacement; and with an implant designed to increase the knee extensor moment arm of the quadriceps. The implant that increased the moment arm reduced muscle force, suggesting that this design would facilitate tasks of daily living in patients. This finding was corroborated and expanded upon when Browne et al. (2005) reported that a total knee replacement design that increased quadriceps moment arm reduced patellofemoral compressive forces. To our knowledge, there exist no similar studies of the effects of TAA on ankle joint mechanics, even if only to quantify the effects of TAA on ATma.

The purpose of this study was to perform the first investigation of the pre- to post-operative changes in ATma that may occur with TAA. We used ultrasound imaging combined with three-dimensional motion analysis to measure ATma in patients prior to their TAA and again six months post-
We hypothesized that pre-operative ATma would be strongly correlated with the post-TAA value, as one goal of this procedure is to reconstruct the patient’s physiologic anatomy.

6.2 Methods

Study design
A repeated-measures prospective cohort study was completed in which 14 patients were recruited, each of whom was scheduled to receive a fixed-bearing total ankle arthroplasty performed by one of three experienced foot and ankle specialists. Participants were deemed eligible for this study if their surgeon identified them as a candidate for TAA and were older than 45 years with no prior ankle arthrodesis.

Patients received one of three fixed-bearing implants, either the INFINITY™ Total Ankle System (Wright Medical Group, Memphis, TN), the Salto Talaris® (Integra LifeSciences Corporation, Plainsboro, NJ) or the Zimmer Trabecular Metal™ Total Ankle (Zimmer, Inc., Warsaw, IN). Standard operating procedures were used for all implant designs. All study procedures received approval from the Penn State Hershey Medical Center Institutional Review Board and all participants provided informed written consent prior to participation in the study.

After the initial pre-operative visit, one subject withdrew from the study following a decision not to undergo TAA. Incomplete datasets were also obtained for three other subjects who were excluded from analysis for this reason. One of these subjects received significant post-TAA surgery and did not attend the post-operative visit; one was unable to perform the required tasks post-operatively due to hip problems that arose after surgery; and we could not obtain ATma data in a third subject due to a failure with synchronization between the motion capture and the ultrasound.

Further details on the individual participants are presented in Table 6-1. The ATma data reported are thus the results of ten patients (age at surgery: $62.86 \pm 9.72$ y; height: $1.72 \pm 0.08$ m; body mass:
for whom complete datasets were obtained at both the pre-operative and post-operative visits.

An additional five young, healthy adults (age: 25.88 ± 2.49 y; height: 1.76 ± 0.05 m; body mass: 77.14 ± 16.59 kg) participated in repeatability testing. These subjects underwent the ATma protocol outlined below in two separate sessions separated by at least two days. This participant pool also provided informed written consent prior to participation.

Measurement of Achilles tendon moment arm

We used our previously published methods to determine ATma in the subjects (Wade et al., 2019), and the approach is briefly summarized here. A 60 mm linear ultrasound probe (Telemed HL9.0/60/128Z-2; Lithuania), was fitted with four reflective markers that were rigidly attached to a custom-made probe case. The probe was held in place over the Achilles tendon at the level of the malleoli, parallel to the tendon using a neoprene brace and foam support. Reflective markers were also placed on the malleoli, the femoral epicondyles, and clusters of four markers mounted on a rigid plastic plate were secured to skin over the dorsum of the foot and the anterior shank with double-sided tape (3M; St. Paul, MN). Each participant then performed three sets of three seated cyclical plantar- and dorsi-flexion motions in time to a metronome at 0.5 Hz, with the knee fully extended and elevated (Figure 6-1). Each participant was asked to maintain the largest comfortable range of motion during these tasks, given their pain and mobility level at the time of data collection.
Table 6-1. Further information regarding participants, including implant received, details of any concomitant surgeries or post-operative procedures, and demographics.

<table>
<thead>
<tr>
<th>ID</th>
<th>Implant Received</th>
<th>Concomitant Surgery</th>
<th>Post-Operative Procedures</th>
<th>Age</th>
<th>BMI</th>
<th>Sex</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibular osteotomy and fibular realignment, Deltoïd reconstruction with semitendinosis allograft, Medial displacing calcaneal osteotomy, First tarsometatarsal plantarflexion fusion</td>
<td>None</td>
<td>67.39</td>
<td>38.63</td>
<td>F</td>
</tr>
<tr>
<td>2</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibular osteotomy with valgus correction and repair, Broström anterior talofibular ligament revision reconstruction</td>
<td>None</td>
<td>64.57</td>
<td>34.83</td>
<td>M</td>
</tr>
<tr>
<td>3</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>78.84</td>
<td>28.88</td>
<td>M</td>
</tr>
<tr>
<td>4</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>72.74</td>
<td>33.74</td>
<td>M</td>
</tr>
<tr>
<td>5</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Revision total hip arthroplasty - removed from study</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>Salto Talaris®</td>
<td>None</td>
<td>Steroid injection</td>
<td>58.54</td>
<td>38.65</td>
<td>F</td>
</tr>
<tr>
<td>7</td>
<td>INFINITY™ Total Ankle</td>
<td>Percutaneous Achilles tendon lengthening, Calcaneal closing wedge Dwyer osteotomy to correct the heel varus, Talar cyst curettage and grafting with autograft</td>
<td>None</td>
<td>46.99</td>
<td>24.34</td>
<td>M</td>
</tr>
<tr>
<td>8</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Lateral fibular osteotomy, Broström anterior talofibular ligament revision reconstruction, Percutaneous Achilles tendon lengthening</td>
<td>Gutter debridement, saucierization of distal medial tibia and talus</td>
<td>66.89</td>
<td>28.67</td>
<td>F</td>
</tr>
<tr>
<td>9</td>
<td>Salto Talaris®</td>
<td>Calcaneal osteotomy with a Zimmer 6.5 screw, Percutaneous Achilles tendon lengthening</td>
<td>Steroid injection</td>
<td>55.52</td>
<td>44.73</td>
<td>F</td>
</tr>
<tr>
<td>10</td>
<td>INFINITY™ Total Ankle</td>
<td>Removed from study due to synchronization problems</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>11</td>
<td>Salto Talaris®</td>
<td>None</td>
<td></td>
<td>71.17</td>
<td>35.87</td>
<td>M</td>
</tr>
<tr>
<td>12</td>
<td>Elected not to undergo total ankle arthroplasty surgery</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>Zimmer Trabecular Metal™ Total Ankle</td>
<td>Fibula correction of valgus deformity and malunion of varus repair, First tarsometatarsal plantar flexion fusion, Percutaneous Achilles tendon lengthening</td>
<td>None</td>
<td>58.35</td>
<td>25.71</td>
<td>M</td>
</tr>
<tr>
<td>14</td>
<td>Salto Talaris®</td>
<td>Removed from study at 6-month follow-up due to complications</td>
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Figure 6-1. Experimental approach to compute Achilles tendon moment arm from seated, non-weight-bearing dorsiflexion and plantarflexion motions. Reflective markers were placed on the femoral epicondyles, malleoli, and four rigidly attached markers were placed on the anterior shank and dorsum of the foot. Four markers were also rigidly attached to a linear ultrasound probe (blue case) which concurrently imaged the Achilles tendon.

Three-dimensional marker coordinates were collected using five Eagle cameras (Motion Analysis Corp., Santa Rosa, CA) sampling at 100 Hz, and the ultrasound images were simultaneously sampled at approximately 60 frames per second. The two sets of data were synchronized using the rising edge of a 5 V square wave signal emitted from the ultrasound beamformer when images were recorded. The rising edge of the square wave signal triggered a solenoid to which a reflective marker was attached. The motion of this marker was used to synchronize the motion data with the ultrasound data (a schematic of the circuit that controlled the solenoid is presented in Appendix C). The axis of ankle rotation was located using a custom-written MATLAB script (Mathworks Inc., Natick, MA) that performed finite helical axis decomposition of the motion of the foot relative to the shank (Wade et al., 2019). The ATma was found by finding the component of a moment produced by a unit muscle force (acting along the tendon line of action) along the axis of ankle rotation for each tendon position obtained from the ultrasound images. ATma computed from this method in young healthy subjects compared favorably to values in the literature computed from magnetic resonance imaging (Wade et al., 2019). For each trial the value of ATma at a neutral
ankle position was interpolated by fitting either a first, second, or third order polynomial (whichever was the best fit, as decided by $R^2$ fit, residual plot, and sum of the square error value) to the moment arm versus ankle angle plot. ATma values for each subject at a single visit were obtained by taking an average of moment arms across trials (procedure outlined in Appendix D).

**X-ray measures**

Pre-operative (mean time before TAA: 7.10 months; range: 1.43 – 15.27 months) and post-operative (mean time following TAA: 6.86 months; range: 2.70 – 11.23 months) sagittal-plane weight-bearing radiographic images were analyzed to determine the position of the natural talus (pre-operatively) or the talar component (post-operatively) relative to the attachment of the Achilles tendon. Radiographs were obtained at the participant’s regularly scheduled clinical visit. Both pre- and post-operative images were analyzed using routines custom-written in MATLAB. For each image, the most posterior aspect of the calcaneal tuberosity (assumed to approximate the insertion of the Achilles tendon) and the superior aspect of the facet for cuboid were digitized (Figure 6-2).

The horizontal distance between these points was used as an estimate of calcaneal length for normalization. Eight points on the sagittal-plane margins of the talar dome or talar component were digitized and a circle was fit in a least-squares sense to those points. The center of this circle was taken to represent the ankle joint center. The anterior-posterior distance, henceforth $d_R$, between the joint center and the posterior aspect of the calcaneal tuberosity was computed and normalized by calcaneal length as this distance did not change due to TAA. To test the reliability of this measure, the pre-TAA and post-TAA radiographs from a randomly selected participant were digitized five times and the coefficient of variation was found to be less than 1% for both the pre- and post-operative images.
Figure 6-2. An example (A) pre-operative and (B) post-operative x-ray image. The posterior calcaneus and identifying landmark (red dots) were used to compute calcaneal length. A circle was fit to self-selected points around the talar dome (magenta dots) and the center was used as a proxy of the superior aspect of the talar dome to compute $d_R$, as a proportion of calcaneal length.

**Statistical analysis**

Linear regression analysis was performed in R (R Core Team, 2018; www.r-project.org) to quantify any relationship between changes in ATma and $d_R$. Assumptions of homoscedasticity and normality were assessed by inspection of residual plots and no obvious deviations were observed. A one-sample t-test was used to quantify whether the absolute differences in ATma and $d_R$ between pre-operative testing and post-operative testing were different from zero, and a paired t-test was run to quantify any intersession differences in ATma repeatability testing. A Pearson’s product moment correlation was also used to investigate the similarity between pre-operative and post-operative ATma, $d_R$, and inter-session testing in the young group. The level of statistical significance was set at $\alpha=0.05$.

**6.3 Results**

In the group of young healthy subjects we studied to assess reliability, there were no significant intersession differences in ATma (mean difference: 4.81 mm or 9.97%; $t_4=2.176; p=0.095$; 95%
Confidence Interval [-1.33, 10.94]), and the values of ATma between sessions were strongly and significantly correlated \((r^2 = 0.906; t_3 = 5.383; p = 0.013; 95\% \text{ Confidence Interval } [0.44, 1.00])\).

Based on these repeatability metrics, we used a threshold of 10\% to identify meaningful differences between pre- and post-operative measures of ATma.

![Figure 6-3](image)

**Figure 6-3.** Post-total ankle arthroplasty Achilles tendon moment arm (ATma) plotted against pre-operative ATma for all ten included participants. Regression line (black, \(R^2 = 0.461, p = 0.031\)) and unity line \((y = x, \text{ dashed line})\) are also plotted. Numbers correspond to participant ID for lookup in Table 6-1.

The absolute value of the change in ATma following TAA was significantly different from zero (mean absolute difference = 9.626 mm, \(t_9 = 4.501, p = 0.001\); 95\% Confidence Interval [4.788, 14.464]). This is best illustrated by those participants who exhibited large differences between pre- and post-operative ATma. For example, participants 1, 3, and 8 had the largest deviation from the unity line (Figure 6-3), experienced changes of -54.22\%, +64.14\% and +123.98\% (pre-to-post) respectively. Whilst a moderate correlation between pre- and post-operative ATma was found \((r^2 = 0.461, p = 0.031; 95\% \text{ Confidence Interval } [0.086, 0.920])\), only 46.1\% of the variance in post-operative ATma was explained by pre-operative ATma (Figure 6-3).
The absolute value of the difference in normalized horizontal distance, \( d_R \), measured from the radiographs also significantly changed pre- to post-TAA (mean absolute difference: 4.583 \% of calcaneal length; \( t_9 = 4.880; p = 0.001; 95\% \text{ Confidence Interval } [2.246 \text{ 6.707}] \)), and \( d_R \) pre- to post- were not correlated (\( r^2 = 0.192; p = 0.2057; 95\% \text{ Confidence Interval } [-0.265 \text{ 0.837}] \); Figure 6-4). The normalized horizontal distance, \( d_R \), represented the sagittal-plane location of the Achilles tendon relative to the joint center, and thus might be useful as a means of understanding how ATma was changed by TAA.

We tested for this association and found there to be a linear, positive relationship between the change in ATma and the change in \( d_R \) (\( R^2 = 0.497, p = 0.023 \)), indicating approximately 50\% of the change in ATma was due to an alteration in sagittal-plane positioning of the talar component (Figure 6-5).
Figure 6-5. Change in distance from posterior calcaneus to the ankle joint center, $d_R$, as a percentage of calcaneal length plotted against change in ATma for all participants. Change is computed as post-operative value – pre-operative value. Numbers correspond to participant ID for lookup in Table 6-1.

6.4 Discussion

Our primary aim in this research was to investigate, for the first time, whether TAA alters ATma, and our results supported our original hypothesis, that pre-operative ATma could predict post-operative ATma at the six month follow-up visit. However, our original hypothesis is supported only when the mean, and not absolute, differences are considered, as there were sizeable differences evident within individuals (e.g., participants 1, 3, and 8; Figure 6-3).

The absolute change in ATma following TAA was substantial, with a mean absolute difference of 9.63 mm difference in ATma due to the surgery. In seven of the ten patients, changes of greater than 10% were greater in magnitude than the variation we observed in the repeatability testing.

The average ATma we found in this study was 38.7 ± 13.7 mm pre-operatively, and 42.2 ± 15.1 mm post-operatively. These ATma values were approximately 10 mm smaller than those we previously reported for an identical method (Wade et al., 2019). Our previous study was of young
adults, however, and ATma has previously been shown to decrease with age (Rasske and Franz, 2018).

We found a significant absolute change in mean anterior-posterior talar dome location following TAA from weight-bearing sagittal-plane radiographic images, and there were large changes on an individual basis (Figure 6-4), similar to those found for ATma. Previous radiographic investigations of TAA implants focused on identifying subluxation, subsidence, radiographic lucency and other factors related to loosening and other poor outcomes (Datir et al., 2013; Kim et al., 2016; Saito et al., 2018). Evaluation of surgical technique tends to be focused on tibiotalar alignment and conformity of the talar component with the tibial component, as these measures are indicative of successful survivorship, reducing the risk of osteolysis (Kim et al., 2016). Therefore, practicing surgeons may not be considering the anterior-posterior location of the talar component with respect to the Achilles tendon, and the implications for post-operative function this distance might represent.

Our results show individual participants responded variably to the influence of TAA on ATma, and that approximately 50% of the change in ATma could be explained by a change in talar dome positioning. In other words, the positioning of the talar component with respect to the calcaneus did influence post-operative ATma along with other factors. When pre-operative talar dome position is not replicated following TAA, there may be good reasons, such as preserving bone stock and maintaining alignment with the tibial component (e.g., Reb and Berlet, 2017). Thirty-seven percent of the change in ATma was explained by the change in $d_R$, suggesting that ATma could be manipulated by implant design, similar to how knee extensor moment arm is given consideration in the design of total knee arthroplasty components (e.g., Mahoney et al., 2002). For example, a design with an anteriorly-skewed talar dome would increase ATma without sacrificing alignment with the tibial component. We can only speculate on the functional consequences of such a change in design but they might include increased walking velocity (Lee and Piazza, 2012) and greater
plantarflexor strength (Baxter and Piazza, 2014). If an individual obtains a greater ATma following TAA, an equivalent plantarflexor moment can be generated with less Achilles tendon force. A reduction in Achilles tendon force is beneficial, as it allows for reduced effort and reduced compressive force within the ankle joint, reducing the risk of total ankle implant component loosening (Sopher et al., 2017). We are currently preparing a study of the relationship between the change in function (as assessed in Chapter 5) and the change in ATma (as assessed in Chapter 6); preliminary results are presented in Appendix E.

A reduction in muscle size has been posited to move the line of action of the Achilles tendon (Sugisaki et al., 2010), thus if ATma is sensitive to muscle size, it is possible that our subjects who have larger ATma are more active and have larger muscle mass. A study found muscle function was improved two-years after TAA, with persisting deficits in muscle size and electromyographic frequency (Valderrabano et al., 2007a). In a longer follow-up period, change in muscle size due to increased post-operative activity levels may be more of a concern than at the six month time point. An investigation into the influence of implant positioning on muscle activity could further advance our understanding of the consequences of TAA on the mechanical advantage at the ankle.

Caution must be used when interpreting the results of this study as it is subject to several important limitations. We obtained reliability measures with younger, non-TAA participants, and thus our reliability findings may not apply to older TA patients. Despite less variability in weight-bearing ATma measures, we computed ATma using non-weight-bearing cyclical plantarflexion and dorsiflexion motions as the weight-bearing range of motion was limited in patients pre-operatively. Future studies could employ partial weight-bearing (e.g., supporting weight on a walker) for more precise estimates of ATma. The radiographic measures of talar dome location were made on two-dimensional sagittal-plane images. The motion of the implant may not be well represented by a hinge joint, and there is some three-dimensional play, as such, the center of the fitted circle is an approximation of talar dome location. Lee et al. (2013) stated that the average time for radiographic
complications to arise was 74 weeks post-TAA, we were not assessing clinical outcomes and therefore feel justified in our measures at the six month time point. As this was an initial investigation into ATma and TAA, we included patients who received concurrent surgeries, such as Achilles tendon lengthening. As such, we cannot be confident that changes in ATma are solely due to the TAA. Patients received one of three implant designs, and we do not have sufficient numbers to extract the effect of implant design on ATma. Research suggests that there is no obvious difference in clinical outcome between fixed and mobile-bearing designs (Gaudot et al., 2014; Queen et al., 2017; Valderrabano et al., 2012), and the three implants used in this study were fixed-bearing. However, we cannot state the degree to which differences in design account for differences in ATma outcome.

This is the first study to investigate the effect of TAA on ATma. We found that there was a significant mean difference in ATma following TAA, and these changes were partially explained by a change in location of the talar dome. Further investigations are necessary to determine the extent to which TAA changes ATma, and the functional consequences with respect to ankle joint mechanics.
Chapter 7

Discussion and Conclusion

7.1 Summary

The purpose of this dissertation was to develop a versatile method of measuring the plantarflexion moment arm of the Achilles tendon (ATma) and apply this method to understand joint mechanical changes following total ankle arthroplasty (TAA), specifically to determine whether ATma is altered following TAA. This was achieved, in Chapter 3 and Chapter 4, by creating and testing a novel method of measuring the plantarflexion moment arm of the Achilles tendon in vivo, using motion tracking and ultrasound imaging. Chapter 5 quantified the changes in locomotor function following TAA, relating the improvements in function to the alleviation of pain, whilst Chapter 6 implemented this novel method for ATma measurement in patients undergoing a TAA to determine that TAA altered ATma in our patient population.

Several methods for quantifying ATma in vivo have been proposed, but the field has not settled on a single “gold standard” methodology. Each of the existing approaches suffer from some limitations, and the research question and environment dictate which method might be most appropriate for a given study. Obtaining ATma from three-dimensional (3D) magnetic resonance (MR) imaging (e.g., Sheehan, 2012) is the most comprehensive technique, allowing 3D location of the tendon and 3D assessment of relative bone motions to identify joint axes during dynamic tasks. However, this 3D MR approach is expensive, time consuming, not available to all researchers, and is restricted in the types of motion able to be analyzed. A promising approach, termed the hybrid method (Manal et al., 2010; Manal et al., 2013), sought to address some of these limitations, with
the primary focus of creating a method for computing ATma that could be used in dynamic tasks. The hybrid method implements ultrasound imaging of the Achilles tendon and motion analysis, tracking both the ultrasound probe and locating bony landmarks that define the ankle joint axis (Manal et al., 2010). This method has been implemented successfully in analysis of walking (Rasske and Franz, 2018), but the approach identified the center of rotation of the ankle at the midpoint of the malleoli, which is an approximation of true relative bone motions (Barnett and Napier, 1952; Isman and Inman, 1969; Lundberg et al., 1989; Sammarco, 1977).

In Chapter 3, we present a study which compared ATma measured with a similar hybrid method, with ankle rotation defined about 1) a point center of rotation at the midpoint of the malleoli, fixed in the shank reference frame (e.g., Manal et al., 2010); 2) a transmalleolar axis – fixed in the shank reference frame; and 3) an axis we termed a ‘functional axis’ (FA), fixed in the shank frame but defined from helical axis decomposition of the foot motions with respect to the shank. We found significant differences between the ATma computed from the anatomically-based axes and the motion-derived axis, which led to a comparison of the various ATma values we obtained to those in from MR literature. The FA produced ATma very similar to those reported from 2D MR studies, and somewhat similar to those reported in 3D MR studies. Use of the malleolar midpoint produced ATma values similar to those previously reported using the same method (Manal et al., 2013; Rasske et al., 2017), but these ATma differed from those obtained using MR.

Studies of bone motions have shown that the location and orientation of the axis of ankle rotation is dependent on joint angle and on loading that varies with task (Arndt et al., 2004; Barnett and Napier, 1952; Isman and Inman, 1969; Lundberg et al., 1989; Sammarco, 1977). A limitation of the study presented in Chapter 3, and in previous estimates of ATma, is that they implement a single fixed axis of rotation, rather than attempt to replicate axis movement with respect to bone reference frames that is known to occur physiologically. There has been one study that computed the axis of ankle rotation as an instantaneous helical axis (IHA) that was allowed to move (Sheehan, 2012),
and quantified ATma using 3D MR imaging. However, no study to date has compared the effect of computing ATma with moving instantaneous helical axes or with a single fixed axis. Chapter 4 presents the results of such a study, that sought to determine whether the use of an IHA offered advantages in ATma determination over FA that compensated for certain tradeoffs in terms of range of motion. In the weight-bearing condition, the IHA and FA approaches produced ATma that were comparable over much of the range of motion, diverging only in dorsiflexion. Additionally, the values of ATma obtained using IHA were comparable to those reported by Sheehan (2012) who used 3D MR imaging. This study provides support for use of the fixed FA approach when plantarflexion is of interest.

Once we had established an acceptable method of computing ATma using ultrasound imaging of the Achilles tendon and motion capture, we aimed to use this novel method to quantify any changes in ATma that might arise due to TAA. Prior research on TAA has focused on surgical outcomes, such as implant survivorship, pain reduction, and patient satisfaction (Brunner et al., 2013; Cody et al., 2019; Daniels et al., 2015; Hendy et al., 2018; McConnell and Queen, 2017; Nunley et al., 2012; Pedowitz et al., 2016; Queen et al., 2014c; Saito et al., 2018; Saltzman et al., 2010; Tenenbaum et al., 2017). Some previous research has conducted gait analysis on patients pre- and post-TAA, specifically noting improvements spatiotemporal parameters (Brodsky et al., 2011; Demottaz et al., 1979; Doets et al., 2007; Flavin et al., 2013; Hahn et al., 2012; Queen et al., 2017; Roselló Añón et al., 2014; Singer et al., 2013; Valderrabano et al., 2007b), and a handful of studies have looked at kinetics, noting some deficits in the involved limb (Nüesch et al., 2015; Queen et al., 2014a; Queen et al., 2014c; Valderrabano et al., 2007b). No one study, however, has related these improvements in function to improvements in pain, and comparisons of functional performance between patients and control subjects has been limited. These questions needed to be investigated before we were able to add complexity in our understanding of TAA by investigating changes in leverage.
For the third study (Chapter 5), patients, approximately one month before and six months following TAA, performed a battery of functional tasks and their results were compared to the performance of age-matched controls. The age-matched control subjects performed better on all tasks, with deficits persisting in the patient group at six months following TAA. In addition to comparing the performance of TAA patients to controls, we also investigated the influence of pain on improvements in function post-TAA. Using regression models, change in walking velocity (both raw, and normalized to height) was best predicted by the interaction between change in pain and BMI. In other words, walking velocity was most dependent on pain in the patients with larger BMI. This was also the case for change in TUG performance. However, BMI alone was the best predictor of SLBT time. This study provided additional support that BMI is related to performance and functional outcomes, as has been demonstrated following total hip arthroplasty (e.g., Slaven, 2012) and total knee arthroplasty (e.g., Giesinger et al., 2018).

The final study of my dissertation (Chapter 6) assessed ATma in the same group of patients at both pre-operative and six month post-operative time points, using the novel method outlined in Chapter 3. This was the first investigation of its kind looking at changes in leverage at the ankle following TAA. Previous investigations of joint replacement and leverage have been conducted for the muscles of the knee and hip joints, but similar studies have not been performed for TAA. For example, it has been shown that increasing the extensor moment arm at the knee through implant design has a positive impact on quadriceps efficiency (Browne et al., 2005; D’Lima et al., 2001; Hamilton et al., 2013). In this study, ATma was found to alter by a mean absolute difference of almost 10 mm between pre- and post-TAA, and ATma varied substantially on an individual basis following TAA. Approximately 50% of the change in ATma was attributable to the anteroposterior position of the talar component relative to the preoperative position of the anatomical talar dome. Further investigations are needed to investigate the muscular and tendon adaptations to TAA, and
to test for any relationship between ATma and functional performance (preliminary data are shown in Appendix E).

7.2 Implications

This dissertation presents a novel method for measurement of ATma, and applies this method to enhance our understanding of how TAA affects ankle joint mechanics. The method of measuring ATma from a functional axis provides an alternative approach to quantifying ATma in vivo during dynamic tasks, building from previous work that only used a midpoint (e.g., Manal et al., 2013; Rasske and Franz, 2018b). Prior research has utilized several different approaches for measuring ATma in vivo, ranging from the simplest form of measuring the distance from the posterior calcaneus to the malleoli from photographs (e.g., Scholz et al., 2008) to the most advanced form, using cine-MR imaging and 3D image rendering to compute ATma from the location of the IHA and Achilles tendon (e.g., Sheehan, 2012).

This dissertation also investigated the effect of using fixed versus moving axes of rotation in assessment of ATma, the results of which showed that the fixed functional axis of ankle rotation was sufficient for computing ATma in neutral and plantarflexion ranges of motion. A previous study came to a similar conclusion when comparing a moving to a fixed center of rotation as computed from 2D MR (Rugg et al., 1990). It is hoped that this alternative approach offers a closer approximation of the anatomical axis of ankle rotation, and that researchers will consider a functional axis over a center of rotation in the future.

One goal of the work presented in this dissertation was to encourage exploration of how TAA alters ankle mechanics and what this might mean functionally. The work in TAA patients highlights the relationship between improvements in pain (and its interaction with BMI) and improvements in function following TAA, which had the strongest relationship for six minute walking velocity.
However, not all of the functional improvements were related to reductions in pain. BMI has an established inverse relationship with walking velocity in older adults (e.g., Beriault et al., 2009; Camarri et al., 2006; Lee and Piazza, 2012), but relationships to pain are less well documented. TAA is well established in improving pain (Saito et al., 2018; Schweitzer et al., 2013; Zaidi et al., 2013), and an individual’s locomotor function improves following the surgery (Brodsky et al., 2011; Chopra et al., 2014; Demottaz et al., 1979; Detrembleur and Leemrijse, 2009; Doets et al., 2007; Dyrby et al., 2004; Flavin et al., 2013; Ingrosso et al., 2009; Kane et al., 2017; Queen et al., 2014c; Roselló Añón et al., 2014; Singer et al., 2013; Valderrabano et al., 2007b). If function improves independently of pain, as we observed, to what else is this improvement attributable? One potential answer is that some of the variation in the function of post-operative TAA patients is attributable to altered mechanics.

The data collected on TAA patients were further analyzed to determine whether TAA did lead to altered mechanics, as indicated by changes in ATma. TAA does appear to alter ATma substantially for some patients, although on average, there was no significant change in ATma following the surgery. The change in ATma observed following TAA was partially explained by a pre-to-post change in talar dome location. This is the first study of its kind to investigate whether TAA caused changes in plantarflexion moment arm of the Achilles tendon. Further work should aim to quantify any structural changes that might lead to mechanical adaptations, including any muscular changes that arise following TAA before there is sufficient evidence to conclude whether TAA influences ankle mechanics.

7.3 Limitations

As with any research, this dissertation has some limitations, and they should be acknowledged and addressed. The method used to compute ATma throughout this dissertation uses a cluster of
markers located on the dorsum of the foot, rather than on the posterior calcaneus, due to methodological considerations. As such, foot deformations under load, such as those occurring at the tarsometatarsal and subtalar joints could have contributed to the ankle joint axes computed from relative bone motions. This may go some distance to explaining the variability in ATma under unloaded conditions when foot joints are relatively unlocked. Related to this, ATma was computed in the patient population solely using the more variable unloaded plantarflexion and dorsiflexion cycles, due to a restricted available RoM when weight-bearing pre-operatively.

Our inclusion criteria were broad for this initial foray into TAA research, so we enrolled subjects who received concomitant surgeries, and those who had other lower-limb concerns. As such, it is hard to tell if the changes reported are solely due to TAA and future work should implement stricter inclusion criteria. We also utilized three different implant designs, and the surgeries were performed by one of three surgeons, and implant design varies considerably. However, the three implants used – the INFINITY™, the Zimmer Metal™ Total Ankle, and the Salto Talaris® – all are fixed-bearing designs. In a study comparing fixed- to mobile-bearing ankle prostheses, no significant differences were identified in functional outcomes (Queen et al., 2017; Valderrabano et al., 2012), therefore we feel justified in our inclusion of patients receiving one of three implants.

The measures of talar dome location we made for our TAA participants were from planar radiographic images. However, there is no constraint for the bone/implant motion to be hinge-like, and the true motions are three-dimensional, not fully planar. As such, the distance obtained and reported in Chapter 6 may not be representative of the articular point of rotation about which the FA might be thought to act.

Finally, patients were monitored six months following their TAA due to time constraints of completing this dissertation. Valderrabano et al. (2007b) presented data that implied patients may not recover full function until two years following a TAA. As such, at the six month time point,
patients are still rehabilitating, which is why we chose to focus on improvements in pain and function.

7.4 Future Work

The work presented herein on total TAA is an initial investigation and more research is needed to understand the interplay between altered ankle joint structure and locomotor function following TAA. For one, an investigation into the relationship between changes in ATma to the observed improvements in function (preliminary data in Appendix E). Understanding whether there is a relationship between structure and function is the logical next step.

This dissertation did not consider muscle-level adaptations following TAA. Investigations into muscle adaptations with ankle arthritis and the subsequent changes in the muscle with surgery would be interesting to understand whether ATma changes change muscle structures. We also only looked at ATma as a potential mechanical explanation for changes following TAA – there are many others including muscle adaptations, adaptations within the tendon itself, changes in the shape of the articulating surface, and bilateral considerations. However, our work presented within is an initial investigation, and forms the foundation for many further exploratory questions to be studied.
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Appendix A

Ultrasound Probe Calibration Procedure

In order to locate the Achilles tendon in the laboratory frame of reference, a time-varying transformation between a reference frame attached to the ultrasound probe and the laboratory frame was needed. Markers rigidly attached to the ultrasound probe (and therefore fixed in a probe frame of reference P) were used to track the probe throughout the data collection. A calibration procedure was followed to determine the locations of these probe markers in probe frame P. Once the locations of the probe markers in P are known, their locations in the ground reference frame G may be used to derive the homogeneous coordinate transformation $T_{PG}$ between G and P and thus locate the Achilles tendon in the laboratory-fixed frame G.

Calibration was accomplished using three cylindrical rods (PLA filament plastic, 2.85 mm diameter; 3Doodler, WobbleWorks Inc., NY, USA) used as ultrasound phantoms that defined a technical reference frame T simultaneously within the ground reference frame G and the probe reference frame P (Figure A1). Each rod was fitted with 12 mm retroreflective markers at both ends such that the line connecting the marker centers was also the central cylinder axis. The phantoms were threaded through holes in a plastic container, forming an inverted triangle when viewed end-on (Figure A2). This container was filled with water, submerging the rods in order to enhance ultrasound images of the rods. Ultrasound images were obtained using a 60 mm linear ultrasound probe (HL9.0/60/128Z-2; Telemed, Lithuania), sampling at a depth of 30 mm and with a sampling rate of approximately 60 fps. A single trial 1 s in duration was collected, during which the probe was held with its imaging plane perpendicular (as judged by eye) to the phantom rods. Locations of the markers fixed to the probe and to the ends of the rods were tracked using eight Eagle cameras (Motion Analysis Corp., USA) operating at 100 Hz. The ultrasound beamformer (LogicScan 128,
Telemed, Lithuania) emitted a 5 V square-wave analog pulse when image sampling commenced, and the rising edge of this pulse was detected by the same data acquisition board (PCIe-6259, National Instruments, USA) used by the motion analysis system. The 5 V pulse enabled synchronization of the marker location and ultrasound image data.

Ultrasound image data and marker coordinates were processed using custom-written routines in MATLAB (MathWorks Inc., USA). A single time frame corresponding to the 10th ultrasound images was selected for analysis and the nearest time frame from the motion analysis data was identified and was taken to correspond to the ultrasound image. The conversion factor between ultrasound image pixels and millimeters was found by digitizing points at a depth of 0 mm and 30 mm on the ultrasound image. On the same ultrasound image each phantom appeared as a circular outline, and the centers of these circles were digitized to determine the coordinates of points R_{1P}, R_{2P}, and R_{3P} in the probe reference frame P. The locations of the points R_{1G}, R_{2G}, and R_{3G} were also found in the laboratory-fixed ground reference frame G by finding the intersections of each phantom cylinder axis (as defined by the markers on the rods) with the imaging plane, which was defined in a probe-fixed frame by three virtual markers established using a pointer fitted with markers.

The technical reference frame T in probe frame P was constructed as follows: The origin of T was located at R_{1P} and basis vectors that defined the reference frame were computed as follows: A unit vector x_P was found that pointed from R_{1P} to R_{3P}, and a quasi-y vector pointed from R_{1P} to R_{2P}. The z_P unit vector was found by taking the cross product of x_P and quasi-y vectors, then normalizing the result. Finally, the y_P unit vector was determined by taking the cross product of the z_P and x_P unit vectors. The 4x4 homogeneous transformation between the probe and technical coordinate systems was thus constructed as:

\[
T_{pt} = \begin{bmatrix}
1 & 0 & 0 & 0 \\
R_{1P} & x_P & y_P & z_P
\end{bmatrix}
\] (A1)
A similar transformation between the ground reference frame and the technical reference frame $T$ was defined using the points $R_{1G}$, $R_{2G}$, and $R_{3G}$ as follows:

$$T_{GT} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ R_{1G} & x_G & y_G & z_G \end{bmatrix}$$  \hspace{1cm} (A2)

The transformation between the ground and probe reference frames ($T_{PG}$) was defined as

$$T_{PG} = T_{PT}(T_{GT})^{-1}$$  \hspace{1cm} (A3)

This transformation was then used to find the locations of the four markers rigidly attached to the ultrasound probe in the probe reference frame by transforming their ground frame coordinates to the probe frame. This transformation was performed for every time frame of the calibration trial, and then an average location for each of the probe-fixed markers was found. During the motion trials, the locations of these probe-fixed markers were used to establish a least-squares fit transformation (Challis, 1995) between $G$ and $P$ that permitted us to locate the tendon image in the ground frame $G$. 
**Figure A1.** Global (G), image (A), and probe (P) coordinate systems necessary for calibration.

**Figure A2.** (A) Calibration set-up and (B) side-view schematic of markers on the ends of each of the three phantoms.
**Figure A3.** Pearson’s correlations between ATma assessed using the three different methods (FA, TA, TM) in neutral position (0°) during the loaded condition.
Figure A4. Plots showing the joint axes TA (red) and FA (blue) along with the Achilles tendon axes in dorsiflexion (black solid line) and plantarflexion (black dashed line) for a single representative subject’s loaded trial. These axes are plotted in the laboratory frame, with displacement transformations applied in order to make the dorsiflexion and plantarflexion joint axes coincident. Two views are shown: At left, the view is down the TA axis (causing TA to appear as a circle); at right, the view is down the FA axis. Looking down the TA axis (left), we see that the tendon axes in dorsiflexion and plantarflexion are relatively close to TA and that this distance does not change substantially between dorsiflexion and plantarflexion. Looking down the FA axis (right), however, we see that the tendon axis is farther away from FA and that the shortest distance between FA and the tendon axis increases in plantarflexion.
Appendix B

Breakdown of Functional Performance by Implant Type

Table B1. Age, BMI, RAND-36 Item Pain Scores, Six Minute Walk (SMW) Distance, Normalized walking speed, Timed Up-and-Go (TUG) time, and Single Leg Balance Time for the involved limb (SLBTi), by implant type, with pre-operative and 6-months post-operative values (mean ± SD) shown.

<table>
<thead>
<tr>
<th>Implant</th>
<th>Zimmer</th>
<th>INFINITY™</th>
<th>Salto Talaris®</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (y)</td>
<td>66.42 ± 5.96</td>
<td>56.74 ± 13.79</td>
<td>63.99 ± 9.80</td>
<td>63.14 ± 7.95</td>
</tr>
<tr>
<td>BMI (kg/m²)</td>
<td>32.09 ± 5.08</td>
<td>29.24 ± 6.94</td>
<td>33.57 ± 6.76</td>
<td>26.64 ± 3.49</td>
</tr>
<tr>
<td>Pain: Pre</td>
<td>43.00 ± 13.85</td>
<td>46.25 ± 33.59</td>
<td>28.13 ± 19.26</td>
<td>92.33 ± 8.58</td>
</tr>
<tr>
<td>Pain: Post</td>
<td>50.63 ± 30.10</td>
<td>88.75 ± 15.91</td>
<td>67.08 ± 15.92</td>
<td>92.33 ± 8.58</td>
</tr>
<tr>
<td>SMW Distance (m)</td>
<td>290.57 ± 152.23</td>
<td>440.92 ± 19.28</td>
<td>391.32 ± 82.97</td>
<td>557.37 ± 90.24</td>
</tr>
<tr>
<td>Pre</td>
<td>391.11 ± 104.91</td>
<td>491.63 ± 50.06</td>
<td>398.43 ± 52.54</td>
<td>557.37 ± 90.24</td>
</tr>
<tr>
<td>TUG (s)</td>
<td>14.36 ± 5.71</td>
<td>7.74 ± 1.89</td>
<td>9.54 ± 1.50</td>
<td>7.11 ± 1.27</td>
</tr>
<tr>
<td>Pre</td>
<td>19.2 ± 0.89</td>
<td>9.19 ± 6.88</td>
<td>6.04 ± 10.67</td>
<td>32/42 ± 23.68</td>
</tr>
<tr>
<td>Post</td>
<td>24.07 ± 8.04</td>
<td>23.55 ± 8.18</td>
<td>2.55 ± 1.81</td>
<td>32/42 ± 23.68</td>
</tr>
</tbody>
</table>
Appendix C

Schematic of Synchronization Between Motion Capture and Ultrasound Image

Figure C1. Schematic of circuit used to activate solenoid for synchronization between ultrasound imaging and motion capture. If voltage is applied to the gate (due to analog signal) of the MOSFET transistor, current flows between drain and source, activating the solenoid, causing the marker to be pulled into the ‘on’ position. If there is no voltage to the gate, the circuit is open.
Appendix D

Procedure for Determining Outlier Trials Using the Motion-derived Functional Axis

For each participant, at each time-point and working within the limits of the ankle joint (as defined by the location of the malleoli markers), the angle and minimum distance between the FA from each trial was computed. Axes were deemed outliers if the angle and minimum distance values were outside of the range of mean ± 1 standard deviation of all computed axes. For each pairing (trial 1 & trial 2, trial 2 & trial 3, trial 1 & trial 3), the angle and the minimum distance between the pair of FA was computed as follows.

For example, the angle was computed as:

$$\theta_{1,2} = \cos^{-1}(\vec{u}_1 \cdot \vec{u}_2) \tag{D1}$$

where $\vec{u}_i$ is the unit vector of the FA for trial $i$.

A point on each axis was defined by the equation:

$$pt_1 = p_{FA_1} + s_1 \vec{u}_{FA_1} \tag{D2}$$
$$pt_2 = p_{FA_2} + s_2 \vec{u}_{FA_2} \tag{D3}$$

where $s_i$ was a fraction of the quasi-mediolateral distance between the medial and lateral malleoli, and $p_{FA_1}$ and $p_{FA_2}$ were points on each FA obtained from the finite helical decomposition. The
Euclidean distance between the points was computed, and the minimum distance was taken as the smallest value of $dist$.

$$dist(u_{FA_1}, u_{FA_2}) = ||pt_2 - pt_1||$$  \hspace{1cm} (D4)

The angle and minimum distances of all trials and participants were then used to compute an average and standard deviation that were used to define an acceptable usable range. Axes that lay outside of the range defined by mean angle/distance ± 1SD were deemed outliers and eliminated from further analysis. To determine which trials were to be included, we followed the algorithm outlined in Table D1.

**Table D1.** Comparison of functional axes across three individual trials to determine outliers

<table>
<thead>
<tr>
<th>FA1 &amp; FA2</th>
<th>FA2 &amp; FA3</th>
<th>FA1 &amp; FA3</th>
<th>Included Trials</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>In</td>
<td>In</td>
<td>In</td>
<td>1,2,3</td>
<td>All axes similar to each other</td>
</tr>
<tr>
<td>Out</td>
<td>Out</td>
<td>Out</td>
<td>1,2,3</td>
<td>No similar axes; use all</td>
</tr>
<tr>
<td>In</td>
<td>Out</td>
<td>Out</td>
<td>1,2</td>
<td>FA1 &amp; FA2 similar; exclude FA3</td>
</tr>
<tr>
<td>Out</td>
<td>In</td>
<td>Out</td>
<td>2,3</td>
<td>FA2 &amp; FA3 similar; exclude FA1</td>
</tr>
<tr>
<td>Out</td>
<td>Out</td>
<td>In</td>
<td>1,3</td>
<td>FA1 &amp; FA3 similar; exclude FA2</td>
</tr>
<tr>
<td>In</td>
<td>Out</td>
<td>In</td>
<td>1</td>
<td>FA1 is middle axis; exclude FA2 &amp; FA3</td>
</tr>
<tr>
<td>In</td>
<td>In</td>
<td>Out</td>
<td>2</td>
<td>FA2 is middle axis; exclude FA1 &amp; FA3</td>
</tr>
<tr>
<td>Out</td>
<td>In</td>
<td>In</td>
<td>3</td>
<td>FA3 is middle axis; exclude FA1 &amp; FA2</td>
</tr>
</tbody>
</table>
Appendix E

Relationships Between Change in ATma and Change in Locomotor Function

These results show the relationship between change in ATma pre- to post-TAA and four measures of locomotor function, combining the work from Chapter 5 and Chapter 6. Our original hypothesis was that those participants who experienced an increase in ATma would have improvement in function, as demonstrated by increased six minute walk (6MW) velocity, increased Single Leg Balance (SLBT) time, and reduced Timed Up-and-Go (TUG) time. Simple linear regression analysis was performed in R (R Core Team, 2018) to investigate whether there was any relationship between the observed changes in ATma (Chapter 6) and the changes in function (Chapter 5). Significant relationships were present between the changes in ATma and 6MW velocity ($R^2 = 0.549$, $p = 0.014$, Figure E1), normalized 6MW velocity ($R^2 = 0.543$, $p = 0.0151$, Figure E2), and TUG ($R^2 = 0.544$, $p = 0.0148$, Figure E3). Changes in SLBT time on the involved limb were poorly predicted by changes in ATma ($R^2 = 0.174$, $p = 0.231$, Figure E4).

Figure E1. A scatter plot of change in 6MW velocity (m/s) against change in ATma. The regression equation is $\Delta 6MW\, velocity = 0.161 - 0.012(\Delta ATma)$.
Figure E2. A scatter plot of change in normalized 6MW velocity (statures/s) against change in ATma. The regression equation is $\Delta \text{Normalized 6MW velocity} = 0.099 - 0.008(\Delta \text{ATma})$

Figure E3. A scatter plot of change in Timed Up-and-Go (TUG) time (s) against change in ATma. The regression equation is $\Delta \text{TUG} = -1.832 + 0.197(\Delta \text{ATma})$
Figure E4. A scatter plot of change in Single Leg Balance time (SLBT) on the involved limb(s) against change in ATma. The regression equation is $\Delta SLBT_{\text{involved}} = 4.211 + 0.330 (\Delta ATma)$
VITA
FRANCESCA E. WADE

EDUCATION
Ph.D., Kinesiology (Biomechanics), Pennsylvania State University, USA 2020
M.Sc., Sports Biomechanics (Distinction), Loughborough University, UK 2016
B.Sc. (Hons), Sport and Exercise Science (2:1), University of Bath, UK 2014

PUBLICATIONS
Peer-Reviewed Publications


In Review
Wade F, Hickox L, and Piazza S. Achilles tendon moment arms are similar when computed using a single fixed axis versus a moving instantaneous helical axis. Journal of Biomechanics.

In Preparation


TEACHING EXPERIENCE
Head Teaching Assistant, KINES384 – Biomechanics Fall 2019, 2018, 2017; Spring 2020, 2019, 2018

Teaching Assistant, KINES384 – Biomechanics Fall 2017, 2016; Spring 2017

FUNDING, HONORS & AWARDS
$8,000, Mini Project Grant, College of Medicine, 2018
Harold F. Martin Graduate Assistant Outstanding Teaching Award, Pennsylvania State University, 2018-19
Finalist, Doctoral Competition, American Society of Biomechanics, 2018
Qualysis Best Podium Award, 2nd Place, American Society of Biomechanics East, 2018
Recipient of College of Health & Human Development Professional Development Endowment, 2017-18
Penn State Team Semi-Finalist, Novel Shoe Design for Limited Dexterity, Nike Ease Competition, 2017