Recovery of Balance and Lower Extremity Joint Contributions in Total Ankle Arthroplasty Patients

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## Abstract

Ankle arthritis is a debilitating condition that causes severe pain and decreased function in the affected limb on the order of end-stage hip arthrosis, end-stage kidney disease, and congestive heart failure. Total ankle replacement is a viable surgical option for treating end-stage ankle arthritis, but few have studied its effects on balance over time. Therefore, the purpose of this study was to test the accuracy of a single-marker method of tracking center of mass, evaluate center of pressure measurements in total ankle replacement patients, and analyze lower extremity joint contributions over a two-year recovery period. Subjects stood on two force platforms for ten seconds in different conditions, and relevant variables were calculated from the force platform and 3D motion capture data. Results showed that increasing recovery time restored partial symmetry between the surgical and non-surgical limbs in ground reaction force, ankle range of motion, and ankle and hip moment contribution in static balance tasks. Furthermore, the ankle and hip may have different roles in postural stability. The results of the studies suggest that total ankle replacement is an effective treatment for end-stage ankle arthritis in terms of restoring postural stability. While patients may not have returned to the level of healthy control subjects, they are more functional and more stable after a two-year recovery period. While further work is needed, the results are encouraging for the outlook of ankle arthritis patients who may need total ankle replacement surgery.

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## **General Audience Abstract**

Ankle arthritis is a debilitating condition that causes severe pain and decreased function in the affected limb on the order of end-stage hip arthrosis, end-stage kidney disease, and congestive heart failure. Total ankle replacement is a viable surgical option for treating end-stage ankle arthritis, but few have studied its effects on balance over time. Therefore, the purpose of this study was to test the accuracy a simplified method to track the center of gravity of the human body, evaluate center of pressure (the point where the force of body weight acts) measurements in total ankle replacement patients, and analyze lower extremity joint contributions to balance over a two-year recovery period. Subjects stood on two force measurement platforms for ten seconds in different conditions, and relevant variables were calculated from the force platform and 3D motion capture data. Results showed that increasing recovery time restored partial symmetry between the surgical and non-surgical limbs in weight-bearing force, center of pressure excursion, and ankle and hip contributions to stability. The results of the study suggest that total ankle replacement is an effective treatment for end-stage ankle arthritis in terms of restoring balance. While patients may not have returned to the level of healthy people, the results suggest they are more stable after a two-year recovery period. While further work is needed, the results are encouraging for the outlook of ankle arthritis patients who may need total ankle replacement surgery.

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#### **Chapter 1: Introduction**

#### Ankle Osteoarthritis and Total Ankle Replacement

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year [1], [2]. The effects on mental health due to endstage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease [3], [4]. Patients can suffer from severe pain, muscle atrophy, limited or no ability to walk unassisted, and difficulty performing daily activities [5], [6]. Unlike other joints, arthritis of the ankle is most commonly developed as the result of a traumatic injury rather than primary arthritis resulting from aging and natural use of the joint or secondary to conditions like obesity or rheumatoid arthritis. Furthermore, it is speculated that ankle arthritis incidence may rise in the future due to increasing participation in sports activities, leading to traumatic injuries, and increased life expectancy [7]-[10]. Post-traumatic ankle osteoarthritis, as it is known, accounts for anywhere from 70-80% of new ankle OA cases each year [2], [3], [11]. Among these cases, the most common causes are ankle fractures, specifically of the malleolus (39%) and the tibial plafond (14%) [11]. Ankle OA from this cause can develop in a patient of nearly any age, and can sometimes delay presentation for several years [12]. Ligament injuries (16%), tibial shaft and talar fractures (7%), and tibiotalar joint injuries are less common forms of post-traumatic injuries that can lead to ankle OA [11], [13]–[16].

There are two widely accepted treatments for end-stage ankle OA: arthrodesis and arthroplasty. In general, surgical options are only considered when the disease has reached its end stage, pain and mobility limitations are severe, and other treatments, such as braces or NSAIDs have proven insufficient [17]. Once it has been determined that surgery is the best course of action, the condition of the ankle is evaluated for bone quality, hindfoot alignment, vasculature, lifestyle demands of the patient, and other areas to determine if arthroplasty or arthrodesis is necessary, or if other joint-preserving surgery such as osteotomy, ligament or cartilage repair, or tendon transfer will be sufficient [17]–[19]. Depending on these factors and a decision made by the surgeon and patient, ankle arthrodesis may also be performed. Ankle arthrodesis (also called ankle fusion) is achieved by orienting the ankle joint in the preferred angle (usually slight

plantarflexion) and inserting one or several surgical screws or nails through the bones to be fused [20], [21]. The bones are then left to naturally fuse and eliminate motion in the affected joint. Several techniques are used, but the most common is isolated tibiotalar arthrodesis, which involves fusing the tibia and the talus. This technique is sometimes expanded to include fusion of the subtalar joint, which requires fusion of the calcaneus to the talus. The technique is known as combined tibiotalocalcaneal (CTTC) fusion, and can be accomplished in 1 main step using an intramedullary nail [20], although this approach is not always used and more advanced procedures have been introduced. While the CTTC technique limits motion even more severely than the isolated tibiotalar technique, it has been shown to reduce complications and further joint degeneration [21]. In cases where arthritis has spread to the hindfoot and cause major deformation, triple arthrodesis of the talocalcaneal, calcaneocuboid, and talonavicular joints may be performed [22]. After an average follow-up of 44 years, degenerative changes in adjacent joints were found in all of the 67 ankles evaluated, and this procedure decreases sagittal motion by up to 15 degrees and decreases coronal motion by 60% [22], [23]. Many other approaches exist and are appropriate for some patients and their specific needs and symptoms. Some consider arthrodesis to be the best treatment for many debilitating ankle conditions, including arthrosis and arthritis, despite major drawbacks such as loss of nearly all ankle motion, possible spread of arthritis to (68% in hindfoot following arthrodesis) and increased stress on adjacent joints, and decreased functional ability [23]-[26]. In some very specific cases, painful arthrodesis has be converted to an arthroplasty if nonunion of the bones occurs [25], [27]. However, this is not a common treatment, and most surgeons will elect to perform a fusion revision.

However, another treatment option is available for patients that may allow greater range of motion to be maintained. The total ankle replacement (also known as arthroplasty, or TAA) is a procedure intended to improve motion of the joint and relieve pain. The surgery generally begins with the opening of the ankle through a standard, anterior approach. The adjacent bones and ligaments are corrected if necessary to fix any pre-existing foot deformities. Each different model of prosthesis has different surgical procedures, but in general, the prosthesis is fixed to the talus and the tibia [26]. If done correctly and without complications, arthroplasties can restore most of the ankle range of motion [28], minimize gait changes, and achieve overall better long-term outcomes compared to arthrodesis [23]. However, some authors have reported higher

revision rates and pain scores in arthroplasty cases [29], [30]. In cases of arthritis of the ankle and subtalar (or other hindfoot) joints, combined arthroplasty and fusion has been performed with promising results [31], [32]. Preservation of most ankle motion was achieved with the replacement, while pain was relieved by the fusion, all while maintaining the early functional outcomes of isolated arthroplasty [31]. While joint arthroplasties at other lower extremities are quite common, the development of ankle replacements was impeded by the failures of the first generation of implants [33]. First generation implants often showed positive short-term results, but deteriorated as loosening occurred, improper positioning and implant size caused malleolar fractures, and small surface area of implants resulted in reoccurring pain for patients [33]. Development of more anatomical designs, addition of a third component, and improved surgical technique helped to bring about a second generation of ankle replacements that have grown in success in recent years [33]. Since the development of the new generation of ankle prostheses, researchers are conducting outcome studies and finding improved revision and failure rates [29], [34]–[36]. Furthermore, due to the complex nature of the surgery and the learning curve that exists with the procedure for many of the implants, it has been suggested that increased surgeon experience and familiarity with the procedure will decrease the need for revision surgeries [34]. As part of the second generation of ankle implants, a number of different options are available that employ different fixation types, numbers of components, bearing types, and more [23]. While only five implants have been approved for use in the US as of 2011, other types have been used with success in European studies [37], [38].

One class of implants are two-component, fixed-bearing total ankle systems. These generally consist of two stem-like components that extend into the tibia and talus. The bearing that connects the two components is fixed, meaning it is not capable of anteroposterior (AP) or mediolateral (ML) translation, and frontal plane ankle motion is limited to 4° [39]–[42]. Of the five implants approved for use in the United States, four are two component, fixed-bearing designs [37]. The Agility<sup>TM</sup> Total Ankle System (DePuy Synthes), designed by Dr. Frank Alvine in 1984, was the first FDA-approved total ankle implant [37]. It consist of a wide tibial component, which allows some additional rotation motion, and a talar component [43]. It is currently approved in the US only with the use of cement [37]. Follow-up studies have reported up to 93 percent satisfaction rates with the Agility<sup>TM</sup> ankle [44], [45]. The INBONE® Total

Ankle Replacement is a unique design similar to knee replacements, which has a tibal stem component that can be lengthened to extend further up the tibia [37], [43]. It is also a fixed-bearing design, and has been only approved for use with cement. It has only been in use since 2004, but a new iteration (INBONE II) has been created, and it is a popular choice for patients with low bone stock and heavier patients [46]. Wright Medical, who produces the INBONE®, released another implant in 2014 - the Infinity® system. The talar component is interchangeable with the INBONE II, and is more low profile than the INBONE [47]. The Salto-Talaris implant is based off a three-component, mobile-bearing design widely used in Europe. The US design is a fixed-bearing, two-component design, which uses a conical tibia component. It has been in use in the US since 2006 [37]. The Eclipse ankle (Kinetikos Medical) is one of the few that uses a medial or lateral approach as opposed to the standard anterior approach. It is not widely used in the US [37]. Only one implant used in the US is a three-component, mobile-bearing implant.

The Scandinavian Total Ankle Replacement (STAR) was created in 1978 by Hakon Kofoed, and was originally a two-component, fixed bearing design. After receiving FDA approval in 2009, it is the only three-component, mobile-bearing prosthesis used in the US and the only implant approved for use without cement [26], [37], [43]. A polyethelene component between the tibial and talar component allows AP translation of the bearing which is designed to allow more motion [37], [43]. Surgical outcomes of STAR implants have been reported by several researchers with promising results [34], [36]. Other mobile-bearing, three-component systems are used in Europe, and among the most popular are the HINTEGRA, STAR, and the European version of the Salto-Talaris. These are mobile-bearing implants that do not use cement, and are not used in the US [43]. However, researchers in Europe have reported positive results with those implants as well [41], [48]. Although European researchers and surgeons prefer mobile-bearing designs, studies in both the US and Europe have found few significant differences between fixed and mobile-bearing implants in terms of patient-reported outcomes, spatiotemporal variables, and joint moments [39], [49].

Since the rise in popularity of ankle arthroplasties, many studies have compared outcomes of the procedure with the previous "gold standard" for end-stage arthritis – arthrodesis. In a study considering data from 1997-2010, researchers found that the arthrodesis rate was about 65%

compared to 45% for arthroplasty [50]. On the other hand, in another study over the years 2001-2007, researchers found that 72% of the ankle patients seen in the participating clinics were given an ankle replacement [29]. However, the first study only considered incidence rates in Finland, where arthroplasties were nearly abandoned near the end of the study range [50], while the second study considered data in Canada. Commonly used measure to evaluate improvements to the ankle after surgery are patient-reported outcomes (PRO). Examples of PRO, are the Ankle Osteoarthritis Scale (AOS), American Orthopaedic Foot and Ankle Society (AOFAS) Ankle-Hindfoot Scale, and Short Form-36 (SF-36). These measures evaluate pain and function of the ankle according to ratings given by the patient, and have been validated and suggested by researchers for use in ankle replacement populations [6], [51], [52]. Studies involving PRO have shown improvements in pain relief for TAA compared to fusion [53], but also more show patients requiring additional surgeries or post-op complications [53]. Additionally, TAA was found to give patients a better chance of returning to recreational activities compared to arthrodesis [35], [54]. However, more complications tend to arise from arthroplasty, with 3-6 year implant survival rates reported anywhere from 70-98% and 8-12 year rates at 80-95% [35], although most tend to range from 70-80% [55], [56]. Revision surgeries are also thought to be more common in TAA [29]. Proponents of arthrodesis will point to this as a downfall of arthroplasty, but most studies report arthrodesis fusion rates as just over 90%, while the rates can be as low as 72% [24], [57]. Furthermore, a meta-analysis of 49 studies conducted by Haddad et. al. concluded that revision and amputation rates for arthroplasty were in fact lower than those of arthrodesis [55]. Common complications of arthrodesis include nonunion (11%), infection (2%), spread of arthritis to adjacent joints (57%), and malposition of the joint or technical error (5%)[58], [59]. Although it is uncommon, failed ankle fusions can be salvaged with arthroplasty in certain cases [25], [27] Total ankle complications mainly arise from improper positioning of the prosthesis, and complications vary based on where it was positioned. The main issues, apart from those that apply to any surgery, include malleolar fractures, nonunion of the ankle syndesmosis, loss of motion, aseptic loosening, loss of bone stock, polyethylene wear, and implant subsidence [60]. In the case of some of these problems, the replacement can be revised with a new prosthesis or arthrodesis [49]. However, amputation is also a possibility in some cases [35].

Abnormal gait is a common and significant symptom of ankle OA. Thus, gait analysis is a common research topic in ankle OA populations. Limited range of motion [28, p. 1], [61]–[63], loss of muscle [64], [65], and plantar loading asymmetry [66] are just a few factors that may play a role in altering gait. Horisberger et. al. measured the plantar pressure distribution in gait of endstage post-traumatic ankle OA patients, and found that the patients put higher loads and pressures and increased contact time and area on their unaffected limbs. They also noted that patients put less pressure on the hindfoot of the affected limb in comparison to the unaffected, transferring more weight to the forefoot and toes [66]. This suggests that patients favor the healthy foot in a unilateral ankle arthritis situation, leading to asymmetrical loading. Additionally, several studies have found decreased ankle range of motion in ankle OA patients compared to controls [19], [62], [63], [67], [68]. Researchers have studied the functional deficits in the lower leg muscles in ankle OA patients and the effects it caused on ankle moment. They found decreased plantarflexion and dorsiflexion torque in both the affected and, to a lesser extent, the unaffected legs of arthritic patients compared to healthy controls [64], [65], [69]. The frequency and intensity of muscle activation for the tibialis anterior, medial gastrocnemius, and peroneal muscles have also been found to have drastically decreased in OA patients compared to controls [65]. Nüesch et. al. explored asymmetric ankle OA and how the progression of the disease affected gait. They found that ankle OA has a significant effect on several gait parameters even before it has progressed past its early stage, which was defined as an arthritic ankle joint with less than 50% osteoarthritic surface. Walking speed, dorsiflexion range of motion, peak weight-acceptance ground reaction force (GRF), and peak plantarflexion power were all significantly decreased in the affected leg of OA patients compared to controls [19]. Patients with early-stage ankle OA who opted for realignment surgery, while reporting similar quality of life to control subjects, have significantly greater pain scores, lower walking speed, smaller ankle range of motion, and lower dorsiflexion moments in a mid-term outcome gait study [70]. As the disease progresses to the end-stage, reduced toe-off GRF, reduced plantarflexion and dorsiflexion moments and powers, and decreased triplanar range of motion were all observed in end-stage ankle OA patients compared to healthy controls [62], [71]. Likewise, in addition to limitations in plantarflexion and dorsiflexion, Schmitt et. al. found an increased reliance on hip extension in the latter stages of stance phase in ankle OA patients [68].

Thus, one of the main goals of arthrodesis or arthroplasty is to restore some of the changes in gait mechanics resulting from ankle OA.

Ankle fusion has been shown to improve some gait parameters compared to osteoarthritic ankles, but not to the level of healthy ankles. Research focusing on ankle fusion outcomes suggests that most pain from ankle OA was relieved, but in some cases, patients had developed arthritis in joints adjacent to those that had been fused, which was affecting gait. Furthermore, range of motion in the hindfoot and midfoot of fused ankles has been measured during walking using a multi-segment foot model and was found to be significantly reduced in the sagittal, frontal, and coronal planes compared to controls [58], [59]. Another study of patients with ankle fusion demonstrated improvements in knee and hip moments and work compared to the same patients before surgery. No significant changes in ankle angles, moments, or work were found [72]. In a series of studies on cadaveric ankles, researchers found that fused ankles attained significantly lower triplanar ranges of motion than controls, poorly replicated the movement transfer of controls between the foot and leg (as measured by the relation of output movement of the ankle relative to input manual movement of the leg), and displayed negligible amounts of talar shift (as measured by degrees of rotation) during dorsiflexion and plantarflexion. In all three areas, fused ankles were nowhere near the performance of the normal ankles, suggesting that the ability of a fused ankle to act normally during walking is compromised [28, p. 1], [73, p. 2], [74, p. 3]. Other gait studies have found that arthrodesis patients had faster gait and longer step lengths, but more asymmetry in stance time and percent stance at toe-off compared to ankle OA patients [75]. They also have exhibited decreased plantarflexion angle at toe-off [76]. Overall, ankle arthrodesis appears to significantly improve gait compared to the same pre-operative ankles. However, significant differences still exist between fused ankles and control ankles as shown in a number of studies [28], [58], [59], [72]–[74].

As ankle replacements have improved, more gait analysis studies have been conducted on total ankle patients. Some studies have compared gait mechanics between joint replacements, and found that not enough significant differences in gait parameters were present to say that one prosthesis is superior [39], [49]. However, the series of studies in cadaveric ankles from Valderrabano et. al. showed that prostheses come very close to replicating normal ankle range of

motion, foot to leg movement transfer, and talar rotation. All the prostheses outperformed the fusion ankles, but several types of ankle replacements were compared. The designs that most closely replicated the structure of an actual ankle joint (three-component, mobile-bearing prostheses) also replicated the motion and mechanics more closely [28, p. 1], [73, p. 2], [74, p. 3]. However, using cadaveric ankles changes the results because ankle motion is not influenced by pain or subject control and relates more to implant design than effects on gait. Using some of the same prostheses, another study reported a large gap in plantarflexion angle between controls and arthroplasties, which contradicts the cadaver study [59]. Several other studies have shown that post-operative ankle replacement patients produce GRF patterns, gait loading symmetry, ankle range of motion, and ankle moments closer to controls than arthrodesis patients [59], [62], [75], [77], [78]. These results suggest that the differences between arthroplasty and arthrodesis are minimal shortly after the patient has recovered from surgery. However, other studies have analyzed gait of arthroplasty patients up to 2 years post-op, and have noticed improvement in many gait variables, including step length, stride length, walking speed, ankle range of motion, and medial GRF [47], [59]. This suggests that while arthroplasty patients may not show great improvement over arthrodesis patients in terms of gait soon after surgery, they will continue to improve as the recovery continues. Studies examining total ankle replacement post-op gait changes have found improvements in spatiotemporal variables such as walking speed, stride length, and percent stance at toe-off as well as ankle angles, moments, and ground reaction forces. Researchers found that performance declined right after surgery, but showed full or partial rehabilitation by one or two years after surgery [62], [79], [80]. Results suggest that total ankle replacement improves gait compared to the same patients before surgery, and continues to improve over time.

Previous research has shown that gait and balance parameters are related, and that changes in one can indicate changes in the other. Several gait and posture studies have been conducted in patient populations with neurological disorder or injury [81]–[83], diabetes [84], and the elderly [85]–[88]. Generally, the results indicated that subjects who walked more carefully (longer double support time, shorter step length) displayed decreased balance performance in static and dynamic tasks [81], [85]. Muscle degeneration, destruction of mechanoreceptors, and proprioceptive deterioration are key factors in postural stability [89]–[93]. Traumatic ankle damage has been

linked to severe damage of mechanoreceptors and subsequent declines in ankle proprioception [94]. Articular mechanoreceptors are located in joint receptor fibers near ankle ligaments, and are often damaged with repeated ankle injury. One of the more common ankle OA etiologies, ligamentous trauma injuries, run a high risk of mechanoreceptor damage due to the low-strength of joint receptor fibers compared to ligaments [89]. Additional receptors are present in articular cartilage and muscles such as the gastrocnemius and tibialis anterior [89]. Thus, a history of ankle injuries causing damage to joint receptor fibers, coupled with the loss of receptors resulting from muscle and cartilage degradation, could leave ankle OA patients with a reduced sense of ankle position. Therefore, the "ankle strategy" for maintenance of postural equilibrium could be compromised in this patient population. The ankle strategy is likely used in slow, low frequency adjustments when the center of mass is not in immediate danger of moving beyond the limits of stability [89]. Ankle strategy could also be considered a long-term balance strategy as opposed to quick center of mass adjustments for critical balance situations. Other studies have suggested that the ankle strategy is used primarily in the anteroposterior (AP) direction, meaning that ankle patients could have additional problems with sway in that direction [95]. Therefore, patients with ankle replacements or ankle OA, especially from a post-traumatic etiology, would be at risk for declines in proprioceptive feedback that could lead to deficits in postural stability, particularly in the AP direction.

#### Balance and Postural Stability

There are several terms that are used to describe the human ability to balance; postural stability, postural steadiness, postural sway, postural dynamics, postural control, and others. The basic principle that is of concern is the human body's ability to maintain a static, upright posture in response to the body's natural tendency to sway [96], [97]. Posture is a particular orientation of the body's linkages, but postural control is a constant and dynamic process with the goal of maintaining a certain posture [98]. In recent years, many researchers have come to the similar conclusion that posture in "quiet stance" is actually comprised of small, corrective motions rather than the previous assumption that it is a truly static task [99], [100]. Within the field of postural control, there are two subsets: static and dynamic. Static balance would be described as simply maintaining one position for some amount of time. This is the more common variation of postural studies and is generally studied using bipedal or unipedal stance. However, dynamic

postural control can be useful for investigating some populations. It involves stability in moving tasks, such as leaning, sit-to-stand, four square step, and others. Dynamic balance can better replicate and analyze everyday tasks that may be affected in the population of interest. However, in certain pathologic populations, testing is often limited to low-risk dynamic tasks or only static tasks due to patients' inability to complete some actions. In either subset of postural stability, several different methods exist to quantify performance.

In the case of a somewhat abstract concept in postural stability, it is often difficult to quantify. Some have used computerized dynamic posturography, or other computerized tests that measure reaction times in response to perturbations and subjects' ability to control their movements [101]. Measures from posturography, such as equilibrium score and postural stability index, have been determined to be moderately reliable [102], [103]. But the most common measures involving postural stability of measures of the center of pressure (COP). The COP is defined as the point on the body segment in ground contact (the foot) through which the ground reaction force (GRF) can be assumed to act [104]. Deviations in the COP demonstrate the reactions of the body in response to perturbations in order to re-stabilize the system. Using ground reaction forces and moments, the COP can be calculated using the methods described by Hufschmidt et. al. [105]. The location of the COP by itself does not reveal much information about the ability of the subject to maintain balance control. If plotted parametrically, the x and y coordinates of the COP do yield a plot known as a stabilogram. A stabilogram can be useful in identifying periods of a balance test during which the subject was unstable, which would be indicated by substantial deviations of the COP from the origin of the test. Prieto et. al. described several other measures of postural steadiness calculated using the COP [106]. Of the dozens of measures described, one of the most commonly used is excursion. Excursion is the total length of the path drawn out by the COP over the length of the test, calculated by summing the distances between each successive set of points [106]. Excursion can also be measured in just the AP or ML directions, and is one of the most common measures of COP used in postural steadiness research. It can put a quantifiable value on how much a subject's COP is deviating from quiet stance. Other measures described by Prieto et. al., such as COP velocity, frequency, and resultant distance express information about the COP in both the time and frequency domain that can be used to assess balance of a subject and possibly identify causes of balance deficits.

An important part of postural control is the neuromuscular and sensory feedback aspect. Proprioception has been defined as "one's ability to integrate the sensory signals from various mechanoreceptors to thereby determine body position and movements in space" [90], [107], [108]. The visual, vestibular, and somatosensory systems are responsible for supplying this information, and the central nervous system (CNS) responds by correctly timing postural correction actions (sensory organization) and executing correct muscle responses (muscle coordination) [91]. The most important component of the afferent system is somatosensation, which concerns the orientation of body parts with respect to one another. Proprioception is a specialized component of touch that is a type of somatosensation, and is key in balance at the ankle [89]–[91], [107], [108]. Through proprioception, the brain is able to sense the static position of body segments and information about their movements. The afferent information comes from mechanoreceptors, which are present within joint articulations and muscles. Articular mechanoreceptors are located in joint receptor fibers near ankle ligaments, and can sense forces, deformations, position, and other information regarding the joint. Muscle receptors, such as golgi tendon organs and muscle spindles, provide statuses on tension, contraction velocity, and strain of muscles. Together, along with cutaneous receptors and the other senses, these receptors allow for quick motions to correct the swaying center of gravity [89]. Because it is generally the only body part touching the ground, the foot-ankle complex is very important to overall body proprioception [90]. Furthermore, proprioception has been shown to be reduced in subjects who have sustained repeated ankle injuries [109]. The increased risk for proprioceptive deficits, the fact that proprioception is considered the most important sensory system for the maintenance of postural stability in older adults [110], and research that has shown improvement in postural stability following a knee replacement [111] indicate that ankle arthroplasty may have a significant positive effect on ankle OA patients whose postural stability has been compromised.

Patients with osteoarthritis, who have undergone total ankle replacement, or others who have sustained traumatic or repeated ankle injuries may have a compromised ability to employ the ankle balance recovery strategy due to the loss of mechanoreceptors [112]. As previously mentioned, surgery to correct ankle osteoarthritis involves the resection of articular structures, which contain a large number of mechanoreceptors [89]. These patients often have a history of

traumatic ankle injuries, which can include ligament disruption [13]. Thus, multiple ligamentous injuries can result in the cumulative loss of mechanoreceptors and deafferentiation of the joint [89]. Although a previous study found no significant differences in the joint position sense of TAA patients compared to controls, the sample size was just 13 and patients had two years to recover from surgery [113]. Significant deficits could be present at earlier time points, and no studies have been done to gather similar data in ankle OA patients. Furthermore, ankle arthritis has been shown to cause lower leg muscle atrophy [69]. Due to the high number of receptors within muscles that aid in balance, atrophy of the lower limb muscles such as the gastrocnemius or tibialis anterior can severely compromise the ability to sense the status of muscles and position sense [89]. While not reaching the muscle activation or torque levels of the healthy leg after a year, TAA patients did show marked improvement over their pre-op muscle function [65]. These proprioceptive deficits may lead to decreased postural stability in ankle OA patients.

Some researchers have investigated the effects of ankle arthritis on postural control. Although the patient population is often younger than those of arthritis in other joints [7], [114]-[116], many elderly people are still affected by the disease. Some already have trouble keeping their balance due to slower reaction time and muscle latency [117]. Balance research on ankle OA patients is sparse, and most studies only examine a small part of the picture. Previous research in healthy controls has shown that weight-bearing asymmetry during quiet stance increases center of pressure (COP) sway velocity and excursions of the more unloaded foot [118]. Additionally, these trends were identified in healthy control subjects, meaning the increased COP excursion and velocity were likely due to asymmetrical loading as opposed to impaired sensorimotor function [119], [120]. Thus, the control issues can be primarily attributed to weight-bearing asymmetry, which suggests that unilateral ankle OA patients, who do not load their limbs evenly [66], are more likely to display increased COP excursion. Furthermore, a meta-analysis of a number of studies that focused on balance changes in patients with unstable ankles clearly showed the balance deficit in injured ankles in terms of COP measures, single-leg-stance times, and excursion tests [121]. Ankle instability and ankle OA have both been linked to repeated ankle sprains and injuries [13], [122]. A study examining the differences in unilateral and bilateral ankle OA patients in quiet standing tasks showed that clinically meaningful differences in COP measures may exist, particularly in the AP direction [120]. Additionally, Wikstrom and

Anderson found increased COP excursion of ankle OA patients in both the AP and ML direction, and Hubbard et. al. found increased resultant COP excursion, as well as increased COP velocity in the ML and resultant directions compared to controls, all in quiet standing tasks [61], [123]. The consensus on the effect of ankle OA on balance is that there is certainly some reduction in control, but more research must be conducted to understand everything that is occurring and the causes behind the actual declines in postural control. However, it is clear that ankle OA could affect postural stability, and therefore interferes with daily activities.

With compromised balance comes loss of balance, or in other words, fall risk. Decreased ankle proprioception has been linked to increased risk of falling and injury sustained from falling [124]. Furthermore, general foot injuries, such as foot lesions and structural deformities were shown to increase fall risk and decrease balance performance [125], [125], [126]. Even without specific pathologies, the results clearly demonstrated the effect of some general foot problem on balance [127]. The same authors further investigated the issue to identify certain characteristics of the ankle and ankle movement that could be linked to increased fall risk. Of those that were identified, ankle range of motion, plantar sensitivity, and plantarflexion strength are notable [126]. Further research was conducted to show that in addition to the previous identifiers of fall risk, fallers were more likely to exhibit valgus deformity and experience disabling foot pain [125]. Severe pain has been identified as a hallmark of ankle OA, and coronal plane malalignment is quite common in ankle OA patients and candidates for total ankle replacement [128], [129]. Furthermore, ankle weakness has been directly linked to increased fall frequency. Whipple et. al. showed that ankle moments and powers were significantly decreased in subjects identified as having more than one unexplained fall within the last year [130]. All of these factors point to decreased postural stability and increased fall risk [131]. Several studies have linked fall risk with osteoarthritis as well as the decreases in range of motion and muscle atrophy that is a hallmark of ankle OA [131]–[136]. In addition to the disruption of normal life that patients experience due to altered gait, ankle OA may pose a risk to patients' health due to the increased likelihood of balance impairments and falls.

The current literature in the area of TAA balance is thin. Along with two studies that have examined non COP-related measures in total ankle patients [101], [137], only one other study

addressed the topic. Garde and Kofoed conducted stabilometry analysis of total ankle patients in 1996, but only 8 subjects were tested and only unilateral balance tests were done [138]. The results, although limited by study size, showed significant improvement in clinical evaluation score, but no differences in stabilometric analysis from pre-to post-op. However, the Garde and Kofoed study is outdated, and vast improvements have been made in the design and function of ankle implants in the second generation of design. Furthermore, multiple studies have been conducted regarding COP-related measures of balance in patients of other lower extremity joint arthroplasties, finding some improvement in COP measures in surgical limbs compared to controls, but still significant deficits [139]–[144]. Further research of a similar fashion is needed on ankle arthroplasty patients. Furthermore, while research has been done on TAA gait across time [39], [59], [62], [79], [80], [145], the few studies that have investigated TAA balance have not considered multiple time points beyond the surgery.

The information already discussed demonstrates a high probability that the balance of TAA patients is compromised. While arthroplasty can correct some of the underlying issues that cause balance deficits in arthritic patients, some damage to joint receptors cannot be undone. Furthermore, information regarding how ankle replacement patients progress in their recovery can aid clinicians in designing rehabilitation strategies. Previous research has not explored the subject of balance in TAA patients thoroughly. Some studies, such as Butler et. al., have studied the balance of ankle replacement patients in comparison to other total joint arthroplasty patients, but only using a pass/fail test during timed single leg stance. During this study, the researchers were focusing on assessing balance in the clinical setting and therefore did not record COP measures during the single leg balance assessment [137]. Other studies have utilized dynamic posturography to conduct impairment assessments of TAA patients and stability responses to changing visual stimuli and support perturbations [101]. Other research investigates the weightbearing asymmetry in total ankle and ankle osteoarthritis patients. While not directly measuring balance performance, weight-bearing asymmetry has been suggested to affect balance and COP excursions [118]. Results from some studies support the conclusion that patients with an ankle prosthesis are more likely to put higher pressure on the non-surgical limb and shift the COP of the surgical limb posteriorly and laterally when performing static balance tasks [146], [147]. Furthermore, some studies have determined that COP excursions in static balance of healthy

controls are significantly higher in the limb that experienced less loading [118], [119]. Therefore, TAA patients may place higher loads on their non-surgical limb, leading to increased COP excursion in the surgical limb [66], [118]. While further research is necessary, this research shows that TAA patients likely have some balance deficits that are related to loading symmetry, residual sensorimotor deficits, and mechanical limitations. However, there are still many unanswered questions about the differences in the surgical and non-surgical limbs and differences between the ML and AP directions in terms of COP-related measures.

#### Purpose and Hypothesis

Therefore, the purpose of this research was to investigate the improvement of balance as quantified by measures of the COP during recovery (pre, 1 year, 2 years post-op) following TAA, to investigate lower extremity joint kinematics and kinetics to explain some of the differences in COP measures, and to compare different methods of tracking body center of mass to possibly provide a simplified alternative method for measuring balance performance. It was hypothesized that: balance in the surgical limb will improve at each subsequent time point, as evidenced by decreasing COP excursions, velocities and resultant distances and increasing ground reaction force symmetry; the hip will contribute higher moments at earlier time points to compensate for reduced ankle strength, but will equalize with ankle contributions as time progresses; and the sacral marker method will provide a reliable alternative to tracking whole-body center of mass location in order to investigate stability.

## **Chapter 2: Center of Mass Tracking**

#### Abstract

Maintaining control of the whole-body center of mass (COM) is paramount in maintaining steady, quiet stance. Lack of stability in COM position may indicate postural control deficits, which have been linked to increased fall risk. A simplified method characterized by tracking a single point on the lower back has been suggested as an alternative that has been successfully tested in slow walking conditions. The purpose of this study was to evaluate the effectiveness of a simplified method of tracking the whole-body COM in the assessment of static balance in ankle osteoarthritis patients. This study was a secondary analysis of previously collected data from 391 ankle osteoarthritis patients. Each patient was asked to perform three 10-second, quiet standing trials in two conditions. Coefficients of multiple determination (CMDs) of the mean time series of each task were calculated. Negative CMDs were set equal to 0. Mean COM positions were tested for normality using a Shapiro-Wilks test and found to be not normal. Therefore, the methods were compared for each task with Mann-Whitney tests of the mean anteroposterior (AP) and mediolateral (ML) COM positions (p < 0.05). Mann-Whitney tests revealed significant differences between the location of the two methods in the ML direction only (FT: p = 0.0027; SW: p < 0.001). The mean CMD values indicate that there is limited association between the two measurement methods in the AP and ML directions for the FT and SW tasks. Taking into account the mostly weak correlations and differences in mean positions, we conclude that the data does not support our hypothesis. The findings of this study suggest that the single-marker sacral method for tracking COM is not reliable in static balance tasks. Further research is needed to compare this method to the gold standard in patient populations.

#### Introduction

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year [1], [2]. The effects on mental health due to end-stage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease [3], [4]. Patients can suffer from severe pain, muscle atrophy, limited or no ability to walk unassisted, and difficulty performing daily activities [5], [6].

Some researchers have investigated the effects of ankle arthritis on postural control. Although the patient population is often younger than those of arthritis in other joints [7], [114]–[116], many elderly people are still affected by the disease. Some already have trouble keeping their balance due to slower reaction time and muscle latency. With compromised balance comes loss of balance, or in other words, fall risk. Decreased ankle proprioception, which could be present in ankle arthritis patients [89], [112], has been linked to increased risk of falling and injury sustained from falling [124]. Furthermore, general foot injuries, such as foot lesions and structural deformities were shown to increase fall risk and decrease balance performance [125], [125], [126]. Even without specific pathologies, the results clearly demonstrated the effect of some general foot problem on balance [127]. The same authors further investigated the issue to identify certain characteristics of the ankle and ankle movement that could be linked to increased fall risk. Of those that were identified, ankle range of motion, plantar sensitivity, and plantarflexion strength are notable [126]. Furthermore, research suggests that the center of mass (COM) plays an important role in relation to fall risk [148].

The COM and center of pressure (COP) are often confused or used interchangeably, but the process of maintaining stability relies on minimizing the distance between the two [89]. The center of mass is the point through which gravitational force is considered to be acting. Each body segment has its own center of mass, and the weighted sum of all segments amounts to the whole body center of mass [104]. In this paper, center of mass (COM) will refer to the whole-body center of mass. In addition, the center of gravity (COG) is defined as a two-dimensional location of the COM in the horizontal plane. Theoretically, there are several ways to calculate the COM. The most accurate way is to use anthropometric data and track the location of each

segmental COM and calculate a weighted sum. This generally requires a motion capture system and a full marker set in order to precisely track the location of each segment [149]. While this method is the most accurate, it is also the most time consuming, for the subject and the researcher. Another technique that is often used is known as the double-integration method. This method uses the shear force component in the direction of the COM coordinate to be calculated, and integrates twice to obtain the change in the location of that component of the COM [149]. However, this approach requires initial conditions, or it only results in the change in position from time 0 [150]. However, correcting the error in that data with moving averages of the COP has been suggested as a solution and a reliable way to calculate COM from force plate data alone [151]. Finally, some researchers simply use a single point on the sacrum to estimate the COM. Validations of the sacral method have been conducted, mainly for use in walking studies. In walking, both the sacral method and the integration method have shown promising results at lower speeds, with deviations rising as walking speed increases [152], [153]. In slipping, the sacral method has also shown strong correlation with the segmental method in one study [153], but showed significant differences when compared to a simplified segmental model in another [154]. However, to my knowledge, no previous studies have investigated the use of a singlemarker COM tracking method in a patient population performing static balance tasks. Ankle arthritis patients are a suitable population to test this method, because they often experience excessive body sway. While neither the integration method or a single marker method is as accurate as the segmental method, the sacral method should be validated against the integration method, which is an accepted technique for COM measurement [149]. Therefore, the purpose of this study was to determine the agreement of a single-marker method for tracking COM with force plate integration. The hypothesis was that the methods would reasonably agree as evidenced by low root-mean-square deviations and high coefficients of multiple determination.

#### Methods

This study was a secondary analysis of previously collected data from 391 ankle osteoarthritis patients. The patients' demographic information is displayed in Table 2-1. All subjects had end-stage ankle arthritis as diagnosed by an orthopedic surgeon and were scheduled for a total ankle replacement within two weeks of initial testing. In order to participate in the study, all subjects had to be capable of independent ambulation without the use of an assistive device and be able to

maintain bilateral, quiet, upright stance for 10 seconds. Potential subjects were excluded if they had experienced or been diagnosed with pain or degeneration of any other lower extremity joint ipsilaterally or contralaterally, had a previous ankle arthrodesis, had a history of lower extremity joint arthroplasty or spinal surgery, or any other neuromuscular deficiencies that affected their activities of daily living, or had a previous ankle arthrodesis. Prior to study initiation, all subjects signed informed consent that was approved by the institutional review board.

Variable	Mean	Std. Dev.	Min	Max
Age at Surgery (yrs)	63.10	9.73	25.07	83.38
Height (m)	1.71	0.10	1.50	2.05
Mass (kg)	86.91	18.02	49.80	145.0
BMI	29.14	5.40	18.04	52.07

Table 2-1: Study 1 Demographics

Each patient was asked to perform three 10-second, quiet standing trials in two conditions. Trials were conducted with patients standing barefoot on two force plates sampling at 1200 Hz (BP600-900, AMTI, Watertown, MA, USA) with feet shoulder-width apart and together. Kinematic data were collected using an 8-camera motion capture system sampling at 120 Hz (Cortex, Motion Analysis, Santa Rosa, CA, USA). A reflective marker was affixed to the subject on the L4-L5 vertebral joint as shown in Figure 2-1.



Figure 2-1: Single marker at the L4/L5 vertebral joint, or sacrum, used to track COM position

Force plate and marker position data was exported from Cortex and COM calculations were performed using a custom developed Matlab program (Mathworks, Natick, MA). Force plate data was filtered using a 4<sup>th</sup> order, low-pass, recursive, Butterworth filter with a cutoff frequency of 30 Hz, and marker data with an identical filter using a cutoff frequency of 7 Hz [68], [155], [156]. COM was calculated for both conditions using two methods. COM was first calculated by twice integrating force platform data as described in Equations 2-1 - 2-4 [151]:

$$a(t) = \frac{F_h}{m} \tag{2-1}$$

$$v(t) = \int [a(t) - a_0] dt$$
 (2-2)

$$s(t) = \int [v(t) - v_0] dt$$
 (2-3)

$$r(t) = s(t) - [MA_s(t) - MA_{COP}(t)]$$
(2-4)

where  $F_h$  is horizontal force, m is subject mass,  $a_0$  and  $v_0$  are the means of a(t) and v(t), MA is the moving average, and r(t) is the estimated COM in the AP or ML direction. Moving averages

were calculated using a 4 second window [151]. The COM was also calculated by tracking the position of a single marker placed on the L4/L5 vertebral joint. Both methods represent center of mass displacement relative to the mean and not absolute COM locations in the global coordinate system. Coefficients of multiple determination (CMDs) of the mean time series of each task were calculated according to methods described by Kadaba et. al [157]. Negative CMDs were set equal to 0. For each direction and condition, the RMS difference between the two methods was calculated. The mean RMS differences for each direction and condition were determined to be not normally distributed and were tested using a Wilcoxon signed-rank test for differences from 0. All statistical analysis was performed using STATA (StataCorp LLC, College Station, TX).

#### Results

Figure 2-2 shows the mean RMS difference for each direction and condition. Wilcoxon signed-rank test for each value showed that all 4 values were significantly greater than 0 (p < 0.001).



Figure 2-2: RMS Deviation of the center of mass displacement. Mean (+ 95% CI) RMS differences for COM displacement over each direction and condition (AP – anteroposterior, ML – mediolateral; FT – feet together, SW – shoulder width). \* indicates significant difference from 0 (p < 0.05).

Figure 2-3 shows the mean CMD values for each task and direction. As a CMD of 1 shows absolute agreement, the values for each direction and condition show moderate agreement.



Figure 2-3: CMD values for center of mass displacement. Mean (+95% CI) CMD values for the mean time series of each task and direction (AP – anteroposterior, ML – mediolateral; FT – feet together, SW – shoulder width)

Table 2-2 reports the mean CMD values for each task and direction. These values indicate that there is moderate association between the two measurement methods in the AP and ML directions for the FT and SW tasks.

 Table 2-2:Average CMD values for FT and SW task in AP and ML direction

Task	AP		ML	
	CMD		CMD	
	Mean	Std. Err.	Mean	Std. Err.
FT	0.739	0.012	0.725	0.012
SW	0.789	0.011	0.513	0.014

Figure 2-4 shows representative examples of COM paths using both methods, and the CMD value for the data. 2-4a shows a trial with excellent correlation, 2-4b shows a trial with moderate correlation, and 2-4c shows a trial with extremely poor correlation.



*Figure 4: Examples of trials with a) high CMD (0.9921) b) moderate CMD (0.7811) and c) poor CMD (0.0455)* 

#### Discussion

The purpose of this study was to determine if a simplified, single marker tracking method would accurately represent the position of the whole-body center of mass in end-stage ankle osteoarthritis patients. The results of the study did not support the hypothesis that the simplified method would reasonably replicate the integration method. These results indicate that the single-marker method for COM tracking is not suitable for assessing COM movement during these tasks. However, positive results were achieved in other studies using similar methods in healthy subjects performing more dynamic tasks [153], [154]. Other possibilities are that the force platform integration method proposed by Chan is not reliable, or that the tasks in this study have COM movement that cannot be differentiated from signal noise. However, since other studies have employed Chan's method with success [158], [159], the former is not a likely explanation. Due to the unknown integration constants (initial velocity and position), this method appears to be prone to large errors in some subjects.

Although the integration method was considered to be the standard of comparison in this study, it gave unrealistic values for many balance trials. The basis of the calculation is that the difference between the moving and average of the COP and the moving average of the twice-integrated terms should be more or less equal. Therefore, subtracting the difference of those terms from the integrated term should give an estimate of the COM [151]. However, double integration methods for finding COM are highly sensitive to boundary conditions, particularly in quiet standing [151]. Chan's method assumes that the initial acceleration and velocity are equal to the mean values of the trial. This could be the reason for some trials diverging; if the actual initial values are much different than the mean values, the calculation will be inaccurate. Therefore, if the integration was begun at a point with known conditions (a local maximum with velocity=0), the error propagation may not be as drastic. Thus, this alteration to Chan's method could reduce error and may be especially appropriate for quiet standing tasks.

This study had a number of limitations. Unfortunately, due to the secondary nature of the analysis, the data collection methods were not able to be determined at the start of this study. COM was not an anticipated interest when the data was being collected, so motion capture data

was not collected for the whole body. Future studies should compare the single marker method to the segmental method to ensure the impact of the results.

The findings of this study suggest that the single-marker sacral method for tracking COM is not completely reliable in static balance tasks. Due to the lack of whole-body motion capture data in this study, the proposed method was compared to a force-platform double integration method which, while considered an acceptable substitute for the gold-standard segmental method, may not provide reliable data in less dynamic tasks such as those described in this study. Additional research is needed to compare this method to the gold standard in patient populations and to modify the integration method used in this study to reduce error. Furthermore, the COM has been suggested as being a better indicator of balance performance, while COP may be better suited as an indicator of balance strategy [120]. Future work should calculate measures of the COM similar to those often calculated with the COP (e.g. excursion, velocity, frequency) and compare the two alongside performance-based balance tests. Additionally, a relation between the COM and COP equaling the difference in excursions normalized to the COP excursion would quantify the amount of "overshoot" in the COP. Winter described this overshoot as the product of the COP being a control variable while the COM is a response variable [160]. The COP is controlled by the body in order to keep the COM within the body's limits of stability, which produces the overshoot effect [161]. This overshoot variable will quantify the degree to which the COP is needed to make large excursions to correct perturbations in COM. Stable and unstable surface athletes have been shown to exhibit different COP profiles, and those differences could be present in the general population or certain patient populations, but cannot always be detected by traditional COP measures [162]. This variable would help to compare people or patients who may use different balance strategies to maintain upright stance and may provide a more definitive measure of postural control.

## **Chapter 3: Evaluation of Center of Pressure Measures**

#### Abstract

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year. The effects on mental health due to end-stage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease. One promising surgical solution for this disease is total ankle replacement, but its effects on balance are not well understood. Therefore, the purpose of this study was to analyze the balance performance, as measured by ground reaction force (GRF) and center of pressure (COP) measures over a period of two years after total ankle replacement surgery. A total of 408 subjects (177 left and 231 right ankles) diagnosed with end-stage ankle OA and scheduled for a total ankle replacement within two weeks of testing. The data was compared across the three time points using a linear mixed effects, maximum likelihood estimation model with time and limb as main effects and sex, age, and BMI as covariates. Significance was set at p < 0.05 for all tests, and all statistical analysis was performed using STATA (StataCorp LLC, College Station, TX). Results showed that surgical limb excursion decreased over time in the feet together condition (p < 0.001) and was not significantly different from the non-surgical limb after 2 years in 3 of 4 direction/condition combinations. Additionally, results showed that patients displayed higher COP frequencies in the mediolateral direction compared to the anteroposterior direction in both limbs and conditions (p < 0.001). Finally, results showed that ground reaction force decreased in the non-surgical limb, while increasing in the surgical limb over time (p < 0.001) to become not significantly different after 2 years and that the two limbs show different changes in resultant COP excursion with changing load symmetry. In conclusion, total ankle replacement does improve balance in the surgical limb compared to the pre-op condition and restores some symmetry between limbs. The difference in relationship between excursion and loading symmetry could be useful for identifying instability in other patient populations.

#### Introduction

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year [1], [2]. The effects on mental health due to end-stage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease [3], [4]. Patients can suffer from severe pain, muscle atrophy, limited or no ability to walk unassisted, and difficulty performing daily activities [5], [6]. Total ankle replacement is a viable surgical treatment for end-stage ankle arthritis, but its effects on balance are not well understood.

Some researchers have investigated the effects of ankle arthritis on postural control. Although the patient population is often younger than those of arthritis in other joints [7], [114]–[116], many elderly people are still affected by the disease. Some already have trouble keeping their balance due to slower reaction time and muscle latency [117]. With compromised balance comes loss of balance, or in other words, fall risk. Decreased ankle proprioception, which could be present in ankle arthritis patients [89], [112], has been linked to increased risk of falling and injury sustained from falling [124]. Furthermore, general foot injuries, such as foot lesions and structural deformities were shown to increase fall risk and decrease balance performance [125], [125], [126]. Center of pressure (COP) measures are one of the most common measurement tools used to assess balance [163]. Studies have shown that analysis of COP measures are able to detect differences in stability in older adults, and patients with neurological disorders [163]–[165]. In particular, COP velocity has been validated as measure of postural instability [164].

Other research has investigated the weight-bearing asymmetry in total ankle and ankle osteoarthritis patients. While not directly measuring balance performance, weight-bearing asymmetry has been suggested to affect balance and COP excursions [118]. Results from some studies support the conclusion that patients with an ankle prosthesis are more likely to put higher pressure on the non-surgical limb and shift the COP of the surgical limb posteriorly and laterally when performing static balance tasks [146], [147]. Furthermore, some studies have determined that COP excursions in static balance of healthy controls are significantly higher in the limb that experienced less loading [118], [119]. Therefore, TAA patients may place higher loads on their non-surgical limb, leading to increased COP excursion in the surgical limb [66], [118]. Other

studies have found increased COP excursions and velocities, some in the AP direction, some in ML, and some in both [61], [120], [123]. Furthermore, previous work has linked decreases in postural stability with fall risk, which could lead to further injury, especially in older adults [131]. While studies have been done to examine postural stability in ankle OA patients, few have followed up with those patients after their ankle replacement to determine improvements or declines in balance after surgery. Butler et. al. found that only 9% of ankle replacement patients passed a single-leg stance test one year after surgery, while hip and knee patients passed 63% and 69% of the time, respectively [137]. Lee et. al. showed that ankle replacement patients utilize more hip motion than controls, had greater difficulty controlling weight shift in the AP direction (as characterized by velocity), and showed asymmetrical loading in quiet stance [101]. However, no studies have investigated COP-related balance measures in TAA patients across time.

Therefore, the purpose of this study was to investigate the improvement of balance as quantified by measures of the COP during recovery (pre-op, 1 year, 2 years post-op) following TAA. It was hypothesized that balance in the surgical limb will improve at each subsequent time point, as evidenced by decreasing COP excursions, velocities and resultant distances and increasing ground reaction force symmetry.

#### **Methods**

A secondary analysis of data from 408 subjects (177 left and 231 right ankles) were analyzed for this study. The demographics for the patients in the study are listed below in Table 3-1. All subjects had end-stage ankle arthritis as diagnosed by an orthopedic surgeon and were scheduled for a total ankle replacement within two weeks of initial testing. Subjects returned approximately one and two years after their total ankle replacement surgery to repeat the same testing procedures that were completed prior to surgery. In order to participate in the study, all subjects had to be capable of independent ambulation without the use of an assistive device and be able to maintain bilateral, quiet, upright stance for 10 seconds. Potential subjects were excluded if they had experienced or been diagnosed with pain or degeneration of any other lower extremity joint ipsilaterally or contralaterally, had a previous ankle arthrodesis, had a history of lower extremity joint arthroplasty or spinal surgery, or any other neuromuscular deficiencies that affected their
activities of daily living, or had a previous ankle arthrodesis. Prior to testing, height, weight, foot length, and foot width were measured and recorded for each subject at each time point. The same measurements were taken before testing at the post-op visits. In addition, age at the time of surgery was determined from patient medical records and confirmed by patient self-report. Prior to study initiation, all subjects signed informed consent that was approved by the institutional review board.

Variable	Mean	Std. Dev.	Min	Max
Age at Surgery (yrs)	63.39	9.78	25.07	83.41
1-Year Test Age (yrs)	64.73	9.64	33.52	84.41
2-Year Test Age (yrs)	65.32	10.21	27.07	85.41
Height (m)	1.71	0.10	1.40	2.05
Mass (kg)	86.74	18.03	49.80	145.0
BMI	29.58	5.55	18.29	51.37

Table 3-1: Patient Demographics

390 subjects were analyzed at the pre-op time point, 251 at the 1 year post-op time point, and 161 at the 2-year post-op time point. Each patient was asked to perform three 10-second, quiet standing trials in two conditions. Trials were conducted with patients standing barefoot on two force plates sampling at 1200 Hz (BP600-900, AMTI, Watertown, MA, USA) with feet shoulder-width apart and together. The force data were exported from Cortex (Motion Analysis Corporation, Santa Rosa, CA), filtered using a 4<sup>th</sup> order, low-pass, recursive, Butterworth filter with a cutoff frequency of 30 Hz [155], [156] and analyzed using a custom developed MATLAB program (MATLAB, Mathworks, Natick, MA). This cutoff frequency is lower than those used in many studies, but quiet standing allows for a lower cutoff frequency due to the less dynamic nature of movement. The raw force plate data were used to calculate the coordinates of the center of pressure throughout each trial according to the methods described by Hufschmidt et. al. [105]. Anteroposterior (AP) and mediolateral (ML) center of pressure (COP) along with resultant (RES), AP, and ML COP excursions were calculated using methods described by Prieto et. al. [106]. COP excursion is a commonly used measure in postural stability studies and can measure the directional movement of the COP. The AP and ML excursion lengths were calculated according to Equation 3-1 [106]:

$$TOTEX_{AP} = \sum_{n=1}^{N-1} |AP[n+1] - AP[n]|$$
(3-1)

where N is the total number of data points collected, and AP is the location in the AP plane of the COP at time point n. The same equation is used for ML excursions. Similarly, the total excursion lengths were calculated according to Equation 3-2 [106]:

$$TOTEX_{RES} = \sum_{n=1}^{N-1} [(AP[n+1] - AP[n])^2 + (ML[n+1] - ML[n])^2]^{1/2}$$
(3-2)

The instantaneous COP velocity was calculated by taking the first derivative of the AP and ML COP position data. Mean and peak values were calculated from the velocity data. COP velocity has been shown to be a reliable measure in quiet stance and has been linked to changes in stability [164]. Mean resultant distance (RD) was calculated as the average COP location relative to the mean location. Mean RD was calculated for each subsequent point and averaged as described by Equation 3-3 [106]:

$$RD = \frac{1}{n} \sum_{n=1}^{N} [(AP[n] - \overline{AP})^2 + (ML[n] - \overline{ML})^2]^{1/2}$$
(3-3)

where  $\overline{AP/ML}$  is the mean location of the COP in the AP or ML direction. Resultant distance shows changes in the average COP location taking both AP and ML directions into account. The mean frequency was calculated as the frequency of a sinusoidal oscillation of value mean distance and total length of total excursion as described by Equation 3-4 [106]:

$$MFREQ_{AP} = \frac{TOTEX_{AP}}{4\sqrt{2}MDIST_{AP}T}$$
(3-4)

Where MDIST is the average distance of the AP/ML time series relative to the mean location. Mean frequency can be used to identify how subjects are balancing by analyzing oscillations of COP movement. Furthermore, the mean vertical ground reaction force (GRF) normalized to body weight on each limb was computed.

The variables of interest were surgical and non-surgical mean vertical ground reaction force and mean resultant distance as well COP excursion, mean COP velocity, peak COP velocity, and

mean frequency in the AP and ML direction for the feet together and shoulder width conditions. The data was compared across the three time points using a linear mixed effects, maximum likelihood estimation model with time and limb as main effects and sex, age at testing, and BMI as covariates. This model was chosen over a repeated measures ANOVA due to the significant number of subjects who did not return for each testing session or were not able to complete all tasks at each session. The maximum likelihood estimation solution to the mixed effects model is able to account for missing data at different time points and still include those subjects in the analysis who completed at least one testing session. Histograms for each variable were examined and the distributions were determined to be sufficiently normal for the analysis. Significance was set at p < 0.05 for all tests, and all statistical analysis was performed using STATA (StataCorp LLC, College Station, TX).

#### Results

Figure 3-1 shows bidirectional COP excursions for each condition by limb over the two-year testing period. In the AP direction of the FT condition, the surgical limb COP excursion decreased with time (p < 0.001), while the non-surgical did not (p=0.751). COP excursion was significantly lower at both post-op time points compared to pre-op (post-1: p < 0.001, post-2: p=0.016) and the excursion on the surgical side was greater than the non-surgical side at the pre-op time point (p=0.014). In the ML direction, the COP excursion beneath the surgical limb deceased with time while the non-surgical increased (p < 0.001). Additionally, surgical side excursion was greater than the non-surgical side at the pre-op (p < 0.001) and 1-year time points (p=0.008).

In the AP direction of the SW condition, the non-surgical side excursion was greater than the surgical side (p=0.014) and the COP excursion was significantly lower at the 2-year time point compared to the pre-op time point (p=0.015). There were no significant differences between limbs at any time point. In the ML direction, the surgical side COP excursion was greater than on the non-surgical side (p < 0.001). Surgical side COP excursion was greater than on the non-surgical side at each time point (pre: p < 0.001, p1: p=0.009, p2: p=0.037).

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Figure 3-1:Center of pressure excursion. AP and ML Center of Pressure excursions for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions. \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

Figure 3-2 shows bidirectional mean resultant distance for each condition by limb over the twoyear testing period. In the FT condition, the non-surgical limb showed greater resultant distance (p < 0.001) and resultant distance was greater on the non-surgical side when compared with the surgical side at each time point (pre: p < 0.001, p1: p < 0.001, p2: p=0.019). In the SW condition, there was a significant time-limb interaction (p=0.006). Non-surgical was significantly greater at each time point (pre: p < 0.001, p1: p < 0.001, p2: p=0.001).



*Figure 3-2: Mean COP Resultant Distance. Mean RD for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions.* \*-*significant difference between limbs* 

Figure 3-3 shows bidirectional COP mean velocity for each condition by limb over the two-year testing period. In the AP direction of the FT condition, no significant effects were found. In the ML direction, a significant decrease in velocity was found with time (p=0.045) and velocity at the 2-year point was lower than the 1-year (p=0.016).

In the AP direction of the SW condition, velocity decreased with time (p=0.031) and the surgical limb showed significantly greater velocity compared to non-surgical (p=0.003). There was a decrease between pre-op and 1-year (p=0.010) and between 1-year and 2-year (p=0.034), and the surgical side velocity was greater at 2-year (p=0.031). No significant differences were found in the ML direction.



*Figure 3-3: Mean Center of pressure velocity. AP and ML mean COP Instantaneous Velocity for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions. \*-significant difference between limbs; +-significant difference compared to pre-op; #-significant difference compared to 1 year post-op* 

Figure 3-4 shows bidirectional COP peak velocity for each condition by limb over the two-year testing period. The only significant difference was the greater non-surgical velocity at the 2-year time point in the AP direction for the FT condition (p=0.045).



*Figure 3-4: Peak Center of pressure velocity. AP and ML peak COP Instantaneous Velocity for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions. \*-significant difference between limbs* 

Figure 3-5 shows bidirectional COP mean frequency for each condition by limb over the twoyear testing period. In the AP direction of the FT condition, a significant time-limb interaction was found (p < 0.001). The surgical limb showed higher frequencies (p < 0.001) and decreased frequency with time (p < 0.001). The surgical limb COP velocity was greater than on the nonsurgical side at the pre-op (p < 0.001) and 1 year (p=0.048) time points. In the ML direction, the surgical limb COP frequency was greater (p < 0.001) than the non-surgical limb. In addition, the non-surgical COP frequency increased with time (p=0.002). Frequency was greater at 1-year (p < 0.001) and 2-year points (p=0.042) compared to pre-op time point and the surgical side COP frequency was greater at all time points (pre-op: p < 0.001, post-1: p < 0.001, post-2: p=0.001) when compared to the non-surgical side.

In the AP direction of the SW condition, a significant time-limb interaction was found (p < 0.001). Surgical limb COP frequency significantly decrease with time (p < 0.001), and the surgical limb frequency was greater (p < 0.001) than the non-surgical limb. There was a significant decrease in frequency between pre-op and 2-year time points (p=0.008) and the

surgical limb was greater at all time points (pre-op: p < 0.001, post-1: p < 0.001, post-2: p=0.012). In the ML direction, non-surgical frequency increased with time (p=0.001), however, the surgical limb results was greater (p < 0.001) than the non-surgical. An increase in frequency was found between pre-op and 1-year (p < 0.001) and a decrease between 1 and 2 years (p=0.037) as well as greater surgical side COP frequency at the pre-op time point (p < 0.001).



Figure 3-5: Mean Center of pressure frequency. AP and ML mean COP Frequency for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions. \*-significant difference between limbs; +-significant difference compared to pre-op; #-significant difference compared to 1 year post-op; %-significant time effect in surgical limb

Figure 3-6 shows comparisons of the mean COP frequency in each limb and condition by direction over all time points. All comparisons between directions were significant (p < 0.001).



Figure 3-6: Comparison of mean COP frequency in AP and ML directions. \*-indicates significant difference between directions (p < 0.001) in a Wilcoxon signed-rank test

Figure 3-7 shows mean vertical ground reaction force for each condition by limb over the twoyear testing period. In the FT condition, the non-surgical limb force was significantly greater (p < 0.001), non-surgical force decreased with time while surgical increased over time (p < 0.001). The non-surgical limb showed greater force pre-op and 1-year post-op (p < 0.001). In the SW condition, the non-surgical limb force was also greater (p < 0.001) and the non-surgical limb force decreased over time (p < 0.001). Non-surgical limb force was greater at all-each time point (pre: p < 0.001, post-1: p < 0.001, post-2: p=0.014).



Figure 3-7: Mean vertical ground reaction force. Mean vertical ground reaction force for surgical and non-surgical limbs in Feet Together and Shoulder Width conditions. \*-significant difference between limbs; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

Figure 3-8 shows the resultant COP plotted against vertical GRF for each condition in males and females at each time point. A value of 0.5 on the x axis would represent equal load being placed on each limb. The regression showed greater slopes in the surgical limb compared to the non-surgical limb at the pre-op time point for both conditions (p < 0.001). At 1 and 2-year post-op time points, significant differences did not exist between limbs (FT: p1=0.612, p2=0.785; SW: p1=0.216, p2=0.270).



Figure 3-8: Resultant center of pressure excursion vs. vertical ground reaction force. Resultant COP/vertical GRF regression and scatter for FT and SW condition at each time point. \* indicates significance difference between limbs (p < 0.001)

#### Discussion

The purpose of this study was to examine changes in COP measures and vertical ground reaction force over the two-year period of recovery from total ankle replacement. The excursion results support the hypothesis, showing COP excursion decreases in the surgical limb with time and an increase in ML excursion with time. By the 2-year time point, only ML excursion in the SW condition was different between limbs. This suggests that recovery time improves stability in the surgical limb and restores some symmetry between limbs by 2 years. The resultant distance was greater in the non-surgical limb in both condition at all-time points. This suggests that the COP of the non-surgical limb is, on average, further away from the base of support than the surgical limb. This could mean that the non-surgical limb is more responsible for correcting the COP to keep the COM within the base of support. The significant velocity results were few and did not provide much insight into the behavior over time. The mean frequency results showed that in

each case, the frequency of the surgical limb was greater than that of the non-surgical. This corresponds to more quick changes in the location of the COP in the surgical limb, and more long-term adjustments made by the non-surgical limb. However, the surgical side COP frequency decreased in the AP direction with time while the non-surgical side increased in the ML direction. Furthermore, the ML frequencies were greater than AP frequencies. This agrees with previous results that have suggested that the hip is primarily used for rapid motions to recover balance when the COM is the near the limits of stability [166]. The results from the ground reaction force data agree with the hypothesis that symmetry would be restored as time increased. They also agree with previous results that show that end-stage ankle OA patients (time 0) place increased load on their healthy limb [66]. The decreasing non-surgical limb force coupled with the increasing surgical force again suggests that symmetry is being partially restored over time. By 2 years after surgery, there was no longer a difference between the limbs in the FT condition, which is the more challenging task. The decreases in resultant COP with increased GRF within limbs agree with previous results that found higher excursions in the less loaded limb [118]. It is also interesting to note that the effect was greater in the surgical limb than in the non-surgical only in the pre-op condition. This suggests that changes in loading symmetry affect the stability of the surgical limb more than the non-surgical limb. Furthermore, this difference in effect of loading asymmetry on excursion no longer exists after surgery. This suggests that a difference in loading asymmetry effect on COP excursion may be indicative of instability of the limb. This could provide a valuable tool for analysis in patients with possible unilateral injury or neurological defect.

This study had some limitations which should be taken into account. Each patient only completed three trials in each condition. This was due to the number of other tasks the patients were asked to do at each testing session in addition to quiet standing. However, an increased number of trials for each condition would decrease the effects of outlying trials in which the patient was uncharacteristically unstable. Furthermore, each trial was only 10 seconds in length. This was due to the inability of some patients to maintain quiet stance unassisted for much longer than that time. Other patients were excluded from the study because they were not even able to maintain quiet stance for 10 seconds. Furthermore, the balance tests were conducted as part of the Short Physical Performance Battery (SPPB), which dictates 10 second trials. However, 10

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seconds may not be long enough to detect all differences in COP frequency analysis. It would have been helpful to have longer trials to analyze, but a longer time was likely not feasible in this population.

In summary, the results of this study suggest that there are significant recoveries in balance as TAA patients recover from surgery that tend to restore symmetry. The non-surgical limb may be more responsible for maintaining balance at first, but the contribution balances with the surgical limb over time. The causes of these changes are unclear – the ankle replacement could be removing mechanical hindrances that prevented patients from maintaining stability, or the patients' balance could be improving simply because they are no longer in as much pain. The differences in the effects of loading asymmetry between the surgical and non-surgical COP excursion at the pre-op time point could provide a valuable analysis tool to diagnose stability problems in patients who may have unilateral pathologies or neurological disorders that affect balance. Specifically, this difference could indicate that the improvements in balance seen after recovery may be due to a decreased reliance on the non-surgical limb, resulting in the equality of the excursion-GRF slopes. Future work could expand upon the number of trials collected in these patients and more closely examine the changes in the effect of loading asymmetry on COP excursion in different populations.

# **Chapter 4: Joint Dynamics**

## Abstract

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year. The effects on mental health due to end-stage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease. One promising surgical solution for this disease is total ankle replacement, but its effects on balance are not well understood. It is known that a large number of TAA patients have sustained multiple ankle injuries, and this often leads to damage of ankle mechanoreceptors. Those who have sustained multiple ankle injuries may have a decreased mechanoreceptor count, decreased ankle proprioception, and decreased ability to use the ankle for postural control. Although multiple studies have been conducted that consider lowerextremity joint kinetics in the gait of TAA patients, there have been no studies on similar measures of lower-extremity joint moments in balance. Information on the contributions of each joint, particularly the ankle and hip, could reveal compensation mechanics used by TAA patients to maintain static balance in spite of their compromised ankles. A total of 408 subjects (177 left and 231 right ankles) were analyzed for this study. Sagittal and frontal lower-extremity joint angles and moments, ankle joint moment arms, and joint moment contribution percentages were calculated. The data was compared across the three time points using a linear mixed effects, maximum likelihood estimation model with time and limb as main effects and sex, age at testing, and BMI as covariates. Significance was set at p < 0.05 for all tests, and all statistical analysis was performed using STATA (StataCorp LLC, College Station, TX). Results showed that ankle range of motion decreased in the surgical limb over time (p=0.026) and that plantarflexion moment in both limbs showed no significant difference by the two-year time point (p=0.067). Furthermore, ankle contributions were shown to be significantly greater than hip contributions in the sagittal plane with the opposite occurring in the frontal plane (p < 0.001). These results suggest that patients are regaining stability and symmetry compared to their pre-op condition and that the ankle and hip may have specifically defined roles in postural stability that may be useful for creating targeted training and rehabilitation strategies.

## Introduction

Ankle arthritis is a debilitating condition that affects about 1% of the world population, with around 50,000 new cases reported each year [1], [2]. The effects on mental health due to end-stage ankle arthritis have been compared to those associated with other chronic diseases such as congestive heart failure, end-stage hip arthritis, and kidney disease [3], [4]. Patients can suffer from severe pain, muscle atrophy, limited or no ability to walk unassisted, and difficulty performing daily activities [5], [6]. Total ankle replacement is a viable surgical treatment for end-stage ankle arthritis, but its effects on balance is not well understood.

It is known that a large number of TAA patients have sustained multiple ankle injuries, and this often leads to damage of ankle mechanoreceptors [94]. Those who have sustained multiple ankle injuries may have a decreased mechanoreceptor count, decreased ankle proprioception, and decreased ability to use the ankle for postural control [91], [108], [166]. The body employs three main strategies to keep the center of mass (COM) within the limits of stability (LOS): ankle strategy, hip strategy, and step strategy. Step strategy is only used when the COM has strayed beyond the LOS and could be considered an "emergency recovery" of postural stability. The ankle strategy involves contraction of the gastrocnemius or the tibialis anterior about the ankle joint. It is often used in slower, low frequency COM movements to maintain equilibrium [89]. Some research has suggested that an ankle-dominant strategy is used primarily in AP sway control [146]. The hip strategy is used more for large, rapid motions to restore balance when the COM is near the LOS. It employs rotation about the hip in concert with antiphase ankle rotation to quickly change the location of the COM and restore postural equilibrium [166]. Others have also indicated that the hip strategy may be used more often in ML balance [95], [146].

Although multiple studies have been conducted that consider lower-extremity joint kinetics in the gait of TAA patients [62], [79], [145], [167], there have been no studies on similar measures of lower-extremity joint moments in balance. Information on the contributions of each joint, particularly the ankle and hip, could reveal compensation mechanics used by TAA patients to maintain static balance in spite of their compromised ankles. Winter theorized that the overall support moment (or summative moment) remains constant between subject who may have varying styles of gait [168]. Other studies have since used this concept to examine compensation

mechanisms in patients with ACL reconstruction as well as older adults [169]–[171]. Lee et. al. showed that TAA patients use more hip motion to maintain balance [101], but it is unknown if that effect persists as the patients recover from surgery and regain function of the ankle or if that increased motion is translated into increased hip moment.

Another measure, ankle joint moment arm, measures the distance away from the ankle joint where the force is acting. Few studies have examined this measure, and those that did were interested in the distance between the gastrocnemius and the center of the ankle joint [172], [173]. The moment arm as defined by the distance between the ankle joint and the point of application of the force is more related to the COP and how far from the body the patient allows it to move. Lower moment arms may correspond to a closer COP and a greater stability.

Therefore, the purpose of this study was to measure lower extremity joint kinematics and kinetics in TAA patients over the recovery period to determine changes in the way the patients are maintaining balance. It was hypothesized that patients would show greater hip moments and range of motion before surgery, but would transition to a more even strategy with the ankle as recovery progressed. Additionally, it was hypothesized that ankle joint moment arm would decrease as recovery progressed as patients recovered balance and keep their COP closer to their body.

#### Methods

A secondary analysis of data from 408 subjects (177 left and 231 right ankles) were analyzed for this study. The patient demographics are shown in Table 4-1. All subjects had end-stage ankle arthritis as diagnosed by an orthopaedic surgeon and were scheduled for a total ankle replacement within two weeks of initial testing. Subjects returned approximately one and two years after their total ankle replacement surgery to repeat the same testing procedures that were completed prior to surgery. In order to participate in the study, all subjects had to be capable of independent ambulation without the use of an assistive device and be able to maintain bilateral, quiet, upright stance for 10 seconds. Potential subjects were excluded if they had experienced or been diagnosed with pain or degeneration of any other lower extremity joint ipsilaterally or contralaterally, had a previous ankle arthrodesis, had a history of lower extremity joint

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arthroplasty or spinal surgery, or any other neuromuscular deficiencies that affected their activities of daily living, or had a previous ankle arthrodesis. Prior to testing, height, weight, foot length, and foot width were measured and recorded for each subject at each time point. The same measurements were taken before testing at the post-op visits, and the implant type was obtained from the patients' medical records. In addition, age at the time of surgery was determined from patient medical records and confirmed by patient self-report. Prior to study initiation, all subjects signed informed consent that was approved by the institutional review board.

Variable	Mean	Std. Dev.	Min	Max
Age at Surgery (yrs)	63.39	9.76	25.07	83.41
1-Year Test Age (yrs)	64.73	9.64	33.52	84.41
2-Year Test Age (yrs)	65.32	10.21	27.07	85.41
Height (m)	1.71	0.10	1.40	2.05
Mass (kg)	86.74	18.03	49.80	145.0
BMI	29.6	5.54	18.29	51.37

Table 4-1: Patient Demographics

390 subjects were analyzed at the pre-op time point, 251 at the 1-year post-op time point, and 161 at the 2-year post-op time point. Each patient was asked to perform three 10-second, quiet standing trials in two conditions. Trials were conducted with patients standing barefoot on two force platforms sampling at 1200 Hz (AMTI, Watertown, MA, USA) with feet shoulder-width apart and together. Reflective markers were placed on the patient on landmarks specified in Figure 4-1. Kinematic data was collected using an 8-camera system collecting at 120 Hz (Motion Analysis Corporation, Santa Rosa, CA). Force plate and kinematic data were exported to Visual 3D v6 (C-Motion, Bethesda, MD). Marker data was filtered using a 4<sup>th</sup> order, low-pass, recursive Butterworth filter with a cutoff frequency of 7 Hz, and force plate data was filtered using an identical filter with a cutoff frequency of 30 Hz [68], [155], [156]. Joint angles were calculated as Cardan angles between adjacent segments with a rotation order of flexion-extension, abduction-adduction, and internal external rotation. Joint moments were calculated using Visual 3D using inverse dynamics and were reported as body mass-normalized internal moments (Nm/kg) [68].



Figure 4-1: Modified Helen-Hayes marker set used for kinematics. Landmarks: Scapula (not pictured), R/L Anterior Superior Illiac Spine, R/L Posterior Superior Illiac Spine, L4/L5 (sacrum), R/L Greater Trochanter, R/L Thigh, R/L Lateral Knee, R/L Shank, R/L Lateral Malleolus, R/L Superior Heel, R/L Inferior Heel, R/L Lateral Heel, R/L Toe. Additional markers used for joint center location: R/L Illiac Crest, R/L Medial Knee, R/L Medial Ankle, R/L 1<sup>st</sup> and 5<sup>th</sup> Metatarsal Heads

Sagittal, frontal, and transverse plane angles and moments were calculated for the ankle, knee, and hip. In addition to those variables, ankle joint moment arm was calculated by dividing the sagittal and frontal plane ankle moment by the vertical and AP ground reaction force, respectively. The moment arms were rectified and normalized to foot length, which was calculated as the AP distance between the inferior heel and toe marker. Additionally, summative moment was calculated as the sum of the absolute value of each joint moment at each time point. While this is contrary to the method described by Winter [168], his method was for the stance phase of gait, while this study focused on quiet standing. Thus, the directions of the moments from the knee moment as Winter suggested would have sometimes resulted in cancellation of the moments. Thus, calculation of the individual joint contribution percentages would have resulted in values that did not add to 100%. Finally, percent ankle and hip contributions were calculated as the absolute value of the ankle/hip moments divided by the summative moment.

The variables of interest were the range of motion, mean, and peak angles and moments for the hip and ankle, ankle joint moment arm, summative moment, and ankle and hip moment contribution percent for the sagittal and frontal planes. The data was compared across the three time points using a linear mixed effects, maximum likelihood estimation model with time and limb as main effects and sex, age at testing, and BMI as covariates. This model was chosen over a repeated measures ANOVA due to the significant number of subjects who did not return for each testing session or were not able to complete all tasks at each session. The maximum likelihood estimation solution to the mixed effects model is able to account for missing data at different time points and still include those subjects in the analysis who completed at least one testing session. Histograms for each variable were examined and the distributions were determined to be sufficiently normal for this analysis. Significance was set at p < 0.05 for all tests, and all statistical analysis was performed using STATA (StataCorp LLC, College Station, TX).

#### Results

# Joint Angles and Range of Motion

Figure 4-2 shows the mean ankle angles of each plane for each condition by limb over the twoyear testing period. No significant interactions, main effects or differences were found.



*Figure 4-2: Mean ankle angle. Biplanar mean ankle angles for surgical and non-surgical limbs in FT and SW condition.* 

Figure 4-3 shows the mean hip angles of each plane for each condition by limb over the two-year testing period. In the sagittal plane of the FT condition, a significant increase in hip flexion with time was found in the surgical (p < 0.001) and non-surgical limbs (p=0.006). There were significant increases in hip flexion angle between pre-op and 1-year (p < 0.001) and between pre-op and 2-year (p=0.006). In the frontal plane, the surgical limb hip adduction angle increased with time (p < 0.001). There was a significant increase in hip adduction angle between pre-op and 2-year time points (p=0.030) and the non-surgical hip adduction was greater at the pre-op time point (p < 0.001).

In the sagittal plane of the SW condition, a significant increase in hip flexion angle was found with time in the surgical limb (p=0.005) and the non-surgical limb (p=0.004). Significant increases in hip flexion angle were found between pre-op and 1-year (p < 0.001) and between pre-op and 2-year (p=0.019). In the frontal plane, the non-surgical limb hip adduction angle was greater than the surgical limb (p < 0.001) and the surgical hip adduction angle increased with

time (p < 0.001). Significantly greater hip adduction angles in the non-surgical limb were found at pre-op (p < 0.001) and 1-year (p=0.041) time points.



*Figure 4-3: Mean hip angle. Biplanar mean hip angles for surgical and non-surgical limbs in FT and SW condition.* \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

Figure 4-4 shows the mean ankle range of motion (ROM) in each plane for each condition by limb over the two-year testing period. In the sagittal plane of the FT condition, time was found cause a significant decrease in ankle ROM in the surgical limb (p=0.026) and there was a significant decrease in ankle ROM between pre-op and 2-year (p=0.013) time points. In the frontal plane, the non-surgical ankle ROM was significantly increased (p < 0.001) overall, and was greater at each time point (pre-op: p < 0.001, p1: p < 0.001, p2: p=0.029).

In the sagittal plane of the SW condition, the non-surgical limb ankle ROM was significantly greater than on the surgical side at the 1-year time point (p=0.020). In the frontal plane, the non-surgical ankle ROM was significantly greater (p=0.018) overall, and the ankle ROM in the non-surgical limb was significantly greater at the 1-year time point (p=0.017).



Figure 4-4: Mean ankle range of motion. Biplanar mean ankle range of motion for surgical and non-surgical limbs in FT and SW condition. \*-significant difference between limbs; %-significant time effect in surgical limb

Figure 4-5 shows the mean hip range of motion in each plane for each condition by limb over the two-year testing period. In the sagittal plane for the FT condition, hip ROM decreased with time (p=0.018) and hip ROM was significantly lower at 1-year compared to pre-op (p=0.005). In the frontal plane, hip ROM decreased with time in the surgical and non-surgical limbs (p < 0.001), and was significantly lower at 1-year and 2-year time points compared to pre-op (p < 0.001).

In the sagittal plane of the SW condition, sagittal plane hip ROM decreased in the non-surgical limb with time (p=0.044) and the 2-year point showed significantly lower hip ROM than the pre-op (p=0.032) and 1-year (p=0.019) time points. In the frontal plane, the hip ROM decreased with time in the surgical (p=0.002) and non-surgical (p=0.004) limbs and the 2-year frontal plane hip ROM was significantly lower when compared with the pre-op (p < 0.001) and 1-year (p < 0.001) time points.



Figure 4-5: Mean hip range of motion. Biplanar mean hip range of motion for surgical and nonsurgical limbs in FT and SW condition. +-significant difference compared to pre-op; #-significant difference compared to 1 year post-op; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

## Mean and Peak Joint Moments

Figure 4-6 shows the mean ankle moment in each plane for each condition by limb over the twoyear testing period. In the sagittal plane of the FT condition, the surgical limb showed significantly lower plantarflexion moment (p < 0.001) when compared with the non-surgical side. However, surgical plantarflexion moment did increase across time, (p=0.041) while the non-surgical plantarflexion moment decreased (p=0.001). The surgical limb showed lower plantarflexion moment at the pre-op (p < 0.001 and 1-year (p=0.023) time points. In the frontal plane, there were no significant effects.

In the sagittal plane of the SW condition, the surgical limb showed significantly lower plantarflexion moment (p < 0.001), and the non-surgical moment decreased with time (p=0.041). The surgical limb was lower than the non-surgical at the pre-op (p < 0.001) and 1-year (p=0.003) time points. In the frontal plane, there were no significant effects.



Figure 4-6: Mean ankle moment. Biplanar mean ankle moments for surgical and non-surgical limbs in FT and SW condition. \*-significant difference between limbs; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

Figure 4-7 shows the mean hip moment in each plane for each condition by limb over the twoyear testing period. In the sagittal plane for the FT condition, there was a significant increase in hip extension with time (p=0.020) and the surgical limb showed lower hip extension moment (p < 0.001). Hip extension moment at the 2-year time point was significantly greater than pre-op (p=0.007) and the surgical hip extension moment was less than non-surgical at each time point (pre-op: p < 0.001, p1: p=0.006, p2: p=0.029). In the frontal plane, surgical limb hip abduction moment increased with time (p < 0.001) and the surgical limb showed lower hip abduction than non-surgical (p < 0.001). Hip abduction moment at 1 and 2-year time points were greater than pre-op (p < 0.001) and 2-year was greater than 1-year (p=0.031). Surgical hip abduction moment was lesser than non-surgical at pre-op (p < 0.001) and 1-year time points (p=0.001).

In the sagittal plane for the SW condition, non-surgical hip extension moment increased with time (p=0.002). 1-year (p=0.002) and 2-year (p < 0.001) hip extension moments were greater compared to pre-op. In the frontal plane, surgical hip abduction moment was greater than non-surgical hip abduction moment (p < 0.001), and surgical hip abduction moment increased with



time (p=0.010). The surgical limb showed lower hip abduction moment at the pre-op and 1-year time points (p < 0.001).

Figure 4-7: Mean hip moment. Biplanar mean hip moment for surgical and non-surgical limbs in FT and SW condition. \*-significant difference between limbs; +-significant difference compared to pre-op; #-significant difference compared to 1 year post-op

Figure 4-8 shows the peak ankle moments in each plane for each condition by limb over the twoyear testing period. Figure 4-8a shows peak dorsiflexion and eversion moments of the ankle. In the sagittal plane of the FT condition, the non-surgical limb ankle dorsiflexion moment decreased with time (p=0.003). The 2-year time point showed significantly greater ankle dorsiflexion moment than at 1 year (p=0.036) and surgical ankle dorsiflexion moment was greater than in the non-surgical limb at pre-op (p < 0.001). In the frontal plane, there were no significant main effects.

In the sagittal plane of the SW condition, the surgical limb showed significantly greater ankle dorsiflexion moment than non-surgical limb (p=0.008) and also had a greater ankle dorsiflexion moment at the pre-op (p < 0.001) and 1-year time points (p=0.043). In the frontal plane, the non-surgical limb showed greater ankle dorsiflexion moment at the pre-op time point (p=0.039).

Figure 4-8b shows peak ankle plantarflexion and inversion moments. In the sagittal plane of the FT condition, surgical ankle plantarflexion moment increased (p=0.020) while non-surgical ankle plantarflexion moment decreased (p=0.001). Ankle plantarflexion moment was lesser in the surgical limb than the non-surgical at pre-op and 1-year time points (p < 0.001). In the frontal plane, there were no significant main effects.

In the sagittal plane of the SW condition, surgical ankle plantarflexion moment was lesser than in the non-surgical limb (p < 0.001) and non-surgical ankle plantarflexion moment decreased with time (p=0.032). Surgical ankle plantarflexion moment was lesser at pre-op and 1-year time points (p < 0.001) compared to non-surgical. There were no significant main effects in the frontal plane.



Figure 4-8: Peak ankle moment. Biplanar peak ankle moment for surgical/non-surgical limbs in FT/SW condition a) peak positive (dorsiflexion, eversion) moment b) peak negative (plantarflexion, inversion) moment. \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

Figure 4-9 shows the peak hip moments in each plane for each condition by limb over the twoyear testing period. Figure 4-9a shows peak hip flexion and adduction moments. For the sagittal plane of the FT condition, hip flexion moment decreased with time (p=0.007) and the surgical limb showed greater hip flexion moment compared to non-surgical (p < 0.001). Hip flexion moments at 1-year (p=0.037) and 2-year (p=0.003) time points were significantly less than at pre-op and the hip flexion moment in the surgical limb was significantly greater than nonsurgical at each time point (pre-op: p < 0.001, p1: p=0.009, p2: p=0.038). In the frontal plane, there was a significant time-limb interaction in hip adduction moment (p < 0.001). Hip adduction moment decreased at each successive time point (p1: p < 0.001, p2: p=0.037) compared to preop and hip adduction moment in the surgical limb was greater at pre-op (p < 0.001) and 1-year (p=0.002) compared to the non-surgical limb.

In the sagittal plane of the SW condition, hip flexion moment decreased with time in the surgical (p=0.047) and non-surgical limbs (p=0.001) and was significantly different from pre-op at 1 and 2-year time points (p < 0.001). In the frontal plane, the surgical limb showed greater hip adduction moment (p < 0.001) and decreased with time (p=0.014). Surgical limb showed greater hip adduction moment at pre-op and 1-year time points (p < 0.001).

Figure 4-9b shows peak hip extension and abduction moments. For the sagittal plane of the FT condition, hip extension moment increased with time (p=0.027) and was greater in the non-surgical limb (p < 0.001). The hip extension moment at the 2-year point was greater than pre-op (p=0.008) and non-surgical hip extension moment was greater than in the surgical limb at all time points (pre-op: p < 0.001, p1: p=0.003, p2: p=0.030). In the frontal plane, surgical hip abduction moment increased with time (p < 0.001) and was greater in the non-surgical limb than in the surgical limb. Hip abduction moment at 1-year and 2-year time points were greater than at pre-op (p < 0.001) and 2-year hip was greater than 1-year (p=0.031). Surgical hip abduction moment was greater than non-surgical at pre-op and 1-year time points (p < 0.001).

For the sagittal plane in the SW condition, non-surgical hip extension moment increased with time (p=0.006). Non-surgical hip extension moment increased at 1 and 2-year time points (p1: p=0.011, p2: p=0.002) compared to pre-op. In the frontal plane, hip abduction moment was

greater in the non-surgical limb (p < 0.001) and the surgical limb hip abduction moment increased with time (p=0.012). Non-surgical hip abduction moment was significantly greater than surgical at pre-op and 1-year points (p < 0.001).



Figure 4-9: Peak hip moment. Biplanar peak hip moment for surgical/non-surgical limbs in FT/SW condition peak a) positive (flexion, adduction) moment b) negative (extension, abduction) moment. \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb; ^-significant time effect in non-surgical limb

#### Ankle Joint Moment Arms

Figure 4-10 shows the mean normalized ankle joint moment arm in each plane for each condition by limb over the two-year testing period. In the sagittal plane of the FT condition, the surgical limb showed a greater moment arm compared to the non-surgical limb (p < 0.001) and the surgical moment arm decreased with time (p=0.005). The surgical limb showed a significantly greater moment arm only at the pre-op time point (p < 0.001). In the frontal plane, the nonsurgical limb showed a greater moment arm compared to the surgical limb (p=0.011), and was significantly greater at the pre-op time point (p=0.033).

In the sagittal plane for the SW condition, there were no significant main effects. In the frontal plane, non-surgical limb showed a greater moment arm (p < 0.001) overall and at all three time points (pre: p=0.002, p1/p2: p < 0.001) compared to the surgical limb. Additionally, the moment arm was greater overall at the 2-year point compared to pre-op (p=0.044).



Figure 4-10: Mean ankle joint moment arm. Mean ankle joint moment arms (normalized to foot length) of the sagittal and frontal plane moments for each condition and limb. Note: Multiple data points were excluded from the Frontal/Transverse plane due to impossible results. \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb;

## Summative Moments and Joint Contributions

Figure 4-11 shows the mean summative moments in each plane for each condition by limb over the two-year testing period. In the sagittal plane of the FT condition, the non-surgical limb had a greater moment compared to the surgical limb (p < 0.001). The non-surgical limb had significantly greater moment only at the pre-op time point (p < 0.001). In the frontal plane, summative moment increased with time, was greater in the non-surgical limb, and the summative moment in the surgical limb increased with time (p < 0.001). Moment was greater at 1-year and 2-year time points (p < 0.001) compared to pre-op, and moment at 2 years was greater than at 1 year (p=0.049). The non-surgical limb was greater at pre-op and 1-year points (p < 0.001).

In the sagittal plane of the SW condition, the non-surgical summative moment was greater overall (p < 0.001) and was significantly greater at the pre-op (p < 0.001) and 1-year (p=0.024) time points. In the frontal plane, moment in the non-surgical limb was greater than the surgical limb (p < 0.001) overall and at each time point (pre-op: p < 0.001, p1: p < 0.001, p2: p=0.004).



Figure 4-11: Mean summative moment. Mean summative moments in the sagittal, frontal, and transverse planes for each condition and limb. \*-significant difference between limbs; +- significant difference compared to pre-op; #-significant difference compared to 1 year post-op; %-significant time effect in surgical limb

Figure 4-12 shows the mean joint contribution percentage in each plane for each condition by limb over the two-year testing period. Figure 4-12a shows the ankle contribution. For the sagittal plane in the FT condition, ankle % decreases with time (p=0.015) and ankle % in the surgical (p=0.011) and non-surgical limb (p=0.014) increased with time. Ankle % was lower at the 1-year point compared to pre-op (p=0.006) and greater in the surgical limb compared to non-surgical at the 2-year time point (p=0.011). In the frontal plane, ankle % decreased with time in the surgical limb (p=0.002) and was greater in the surgical limb compared to the non-surgical (p=0.041). Ankle % was lower at 1-year (p=0.036) and 2-year time points (p=0.010) compared to pre-op and was greater in the surgical limb at pre-op (p < 0.001).

In the sagittal plane of the SW condition, the ankle % in the non-surgical limb was greater than the surgical at the pre-op time point (p=0.008). The non-surgical limb ankle % also significantly decreased with time (p=0.014). In the frontal plane, the ankle % in the non-surgical limb was greater than the surgical ankle % at pre-op (p=0.027) and the surgical limb percentage decreased with time (p=0.048).

4-12b shows the hip contribution percentage. For the sagittal plane in the FT condition, hip % decreased with time (p < 0.001). Hip % was lower at 1-year and 2-year time points compared to pre-op (p < 0.001) and the non-surgical hip % was greater at the pre-op time point (p=0.034). In the frontal plane, hip % increased with time (p=0.010), the non-surgical limb was greater (p < 0.001), and the surgical hip % increased with time (p < 0.001). 1-year (p=0.025) and 2-year (p=0.007) hip % were greater than at pre-op and the non-surgical limb hip % was greater at pre-op (p < 0.001) compared to the surgical limb.

In the sagittal plane of the SW condition, hip % decreased with time (p < 0.001) overall and at 1year and 2-year time points compared to pre-op (p < 0.001). In the frontal plane, the non-surgical limb had a greater hip % compared to the surgical limb (p=0.006) and the surgical limb hip % increased with time (p=0.012). The non-surgical limb hip % was significantly greater at the preop time point only compared to the surgical limb (p < 0.001).



Figure 4-12: Mean joint contribution percentage. Mean joint contribution percentage in the sagittal and frontal planes for each condition and limb for the a) ankle and b) hip. \*-significant difference between limbs; +-significant difference compared to pre-op; %-significant time effect in surgical limb

Figure 4-13 compares ankle and hip contribution percentage over all time points for each direction and condition. Each comparison between joints showed a significant difference in contribution percentage. In the AP (sagittal plane) comparisons for each leg, the ankle showed a significantly greater percentage, while the hip was significantly greater for ML (frontal plane) comparisons (all p < 0.001).



Figure 4-13: Joint contribution percentage by direction. Mean ankle and hip contribution percentages for AP (sagittal) and ML (frontal) direction in S and NS limbs for a) feet together and b) shoulder width condition. \* indicates significant difference between ankle/hip (p < 0.001)

## Discussion

The purpose of this study was to examine changes in lower extremity joint angles and moments over the recovery period. Sagittal ankle range of motion in the surgical limb decreased in the FT condition. This is counterintuitive to results in gait studies that show increases in range of motion after surgery [59], [62], [75], [77], [78]. However, gait uses a larger range of motion than quiet standing. In quiet standing, an increase in range of motion may in fact indicate more instability. This theory is supported by the results of this study, which show that ankle range of motion decreases in the surgical limb at each time point. This suggests less ankle movement and more stability as time increased. The convergence of the peak PF moment between limbs in the sagittal plane suggests that the ankle is regaining strength and ability to contribute to stability in the sagittal plane. This agrees with previous studies that have shown that the ankle contributes more to AP balance [95], [146]. A similar result showed the two limbs regaining symmetry in peak hip ABD moment as patients recovered. This supports the theory that the ankle is more involved in AP balance while the hip is more linked to ML control. This hypothesis is also supported by the joint contribution results. The ankle contribution in the sagittal plane increased with each time point in the surgical limb, and the hip contribution in the frontal plane of the surgical limb increased with time. Regardless of time point or limb, the ankle contributed more in the sagittal plane and the hip more in the frontal plane. The data suggests that the recovery period helps the surgical ankle regain symmetry with the non-surgical limb in sagittal moment contribution and regain overall symmetry of the limbs, as shown by the frontal plane hip contributions. The contribution results along with the partial recovered symmetry in ankle and hip moments lend support to the hypothesis that the ankle is responsible for AP control, while the hip is responsible for ML [95], [146]. However, the hypothesis of this study that the hip would contribute more at early time points was not supported. Instead, the results suggest that the two joints are simply used for different purposed in balance and may not compensate for one another. These findings could be helpful in identification of balance deficits in patients with particular pathologies, since ankle injuries may manifest themselves more in AP balance deficits and vice versa for hip injuries. Furthermore, these findings could prove useful for balance training strategies to target joints that should be strengthened for improvement in specific balance tasks.

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There were some limitations in this study. As with the COP analysis, the number of trials and the length of time limit the impact of the results. Additionally, much of the ankle joint moment arm data had to be excluded due to impossible values. This was likely due to very small forces in the AP and ML directions at some points. Since the moment arm was calculated by dividing by force, very small force values would cause singularities in the moment arm values. If calculations were not accurate at some points, they may not have been accurate in some of the points that were included in the data. Thus, the moment arm data should not be considered relevant.

In summary, the joint data shows that there may be weight to the theory that the ankle is more involved in AP postural control, while the hip is more involved in the ML direction. Decreases in ankle range of motion may suggest that as patients recover, their ankles decrease in the amount of deviation they experience. This is likely due to the nature of quiet stance tasks, which do not test the range of motion in the same way as gait tasks. Symmetry between limbs at the 2-year time point in metrics such as ankle plantarflexion moment and hip abduction moment may suggest that symmetry is being partially restored with recovery time. However, much of the data did not show clear results, and could benefit from control data for comparison. Future work in this area should focus on obtaining joint kinematics and kinetics for quiet standing in a healthy population and further investigating the theory that the ankle and hip joints are specialized to control certain directions.

## **Chapter 5: Conclusions**

The results of these studies indicate that recovery time in TAA patients has an impact in recovery of postural stability. In addition, the proposed single-marker method for center of mass tracking proved to be unreliable for this population and set of conditions. Further validation against the proven segmental method may be beneficial, and the force plate integration method that was chosen may need to be optimized for analysis in quiet standing tasks. Center of pressure analysis showed that increased loading asymmetry causes changes in excursion, specifically that it affects the surgical limb more than the non-surgical. After surgery, the difference in effect of loading asymmetry in the two limbs became essentially equal. Results also showed that patients regain some symmetry between their limbs after 2 years of recovery. However, whether this improvement is due to changes in mechanics from the surgery or simply due to pain relief is unclear. Joint contribution and center of pressure frequency analysis showed that the ankle and hip may have different roles in postural stability. Future work in the area should focus on a more in-depth look at the COM-COP relationship and validating a single marker method against the segmental method, further exploration of the diagnostic power of the COP excursion-GRF relationship, and further inspection of the specific roles of the ankle and hip in postural control.

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