COPPOCK, JAMES AVERY, M.S. Validation of the Thigh Stabilization System for a Novel Multi-Planar Instrumented Knee Arthrometer. (2019)
Directed by Dr. Randy J Schmitz. 92 pp.

Knee joint laxity characterizes the structural proprieties of the connective tissues and supporting structures within the knee joint. In the past, knee joint laxity has been measured subjectively by clinicians who assess joint integrity through manual manipulation of the joint. More recently however, instrumented knee arthrometers have provided clinicians and researchers alike with objective measures of knee joint laxity. To this end, arthrometry has become an important tool for use in to characterizing knee joint laxity and how it differs across broad populations.

Despite the many technological advances in instrumented knee arthrometry over the past three decades, there are still significant issues with the reliability, and generalizability of these measurements. These issues inhibit our understanding of how knee joint integrity changes in response to joint insult and hormonal fluctuations. Therefore, a novel instrumented arthrometer must be developed to specifically address these deficiencies. To this end, this thesis examines and discusses the gaps in current instrumented arthrometry. Furthermore, it proposes a solution to address a key measurement issue associated with thigh segment stabilization and attempts to validate this solution via a stabilization study utilizing cadaveric specimens.

The evidence presented herein suggests that, while the *a priori* benchmarks for this study were not completely met, the stabilization system was clearly able to provide sufficient stability such that an arthrometric assessment of the joint could be repeatedly administered. Moreover, with minor changes to the current stabilization system it may be entirely possible to obtain truly generalizable and highly repeatable arthrometric evaluations.

VALIDATION OF THE THIGH STABILIZATION SYSTEM FOR A NOVEL MULTI-PLANAR INSTRUMENTED KNEE ARTHROMETER

by

James Avery Coppock

A Thesis Submitted to
the Faculty of The Graduate School at
The University of North Carolina at Greensboro
in Partial Fulfillment
of the Requirements for the Degree
Master of Science

Greensboro
2019

Approved by

_____
Committee Chair

Mura gcuirfidh tú san earrach ní bhainfidh tú san fhómhar.

APPROVAL PAGE

This thesis written by James Avery Coppock has been approved by the following committee of the Faculty of The Graduate School at The University of North Carolina at Greensboro.

Committee Chair _____
Randy J Schmitz

Committee Members _____
Sandra J Shultz

_____
Jeffrey B Taylor

_____
Date of Acceptance by Committee

_____
Date of Final Oral Examination

ACKNOWLEDGMENTS

Thank you to Randy and Sandy who both believed in this project and helped see it through some difficult times. Furthermore, I would like to thank Jeff, whose clinical insights were instrumental in evoking a more clinically useful device. Financial Support for this project was provided through the North Carolina Biotechnology Center's Biotechnology Innovation Grant (SJ Shultz, PI).

TABLE OF CONTENTS

LIST OF TABLES

Page

LIST OF FIGURES

Page

CHAPTER I

INTRODUCTION

Ligamentous knee laxity is a property of the connective tissues within and around the knee joint. Knee joint laxity describes how the structures deform under fixed force application [12]. Laxity in the tibiofemoral knee joint is characterized by relative displacement of the tibia relative to the femur. This displacement happens within three planes of motion and is comprised of six degrees of freedom: three translations and three rotations.

The majority of clinical and scientific focus has surrounded the characterization of anterior – posterior (AP) knee laxity, which heavily focuses on assessing functionality of the cruciate ligaments. One of these driving factors is due to the discovery that AP (sagittal) plane laxity has been shown to be a prospective risk factor associated with anterior cruciate ligament (ACL) injury [13]. Furthermore, traumatic knee injuries, such as an ACL rupture, are a known risk factor for the early development of osteoarthritis (OA) [14–17]. For those who have experienced a traumatic knee injury there is an estimated 40% chance that the injured knee will develop osteoarthritis (OA) within the joint [14–18]. OA poses significant health and quality of life (QOL) risks as osteoarthritis and can impose significant disability and pain on those who are symptomatic [14–17].

Recent evidence suggests that, in addition to sagittal plane laxity, frontal and transverse plane laxity characterizations are also important predictors for knee instability and subsequent injury [19–34].

In the case of non-contact ACL injuries, Koga et al. found that those injured invariably involved rapid transverse and frontal plane rotations [25–27]. Additionally, While ACL injuries represent only a portion of knee injuries and broader injury mechanisms, the evidence that supports the involvement of a tri-planar injury mechanism demonstrates its potential importance in various other mechanopathologies [23,30,31,35]. Furthermore, evidence suggests that laxity profiles are not uniformly stiff or lax within a subject [19,20,23–25,30,30–32]. Shultz et al. have demonstrated that knee joint laxity is probably based upon multiple physical characteristics [23]. Further, it has also been demonstrated that laxity profiles are dynamic and cyclical over time, exhibiting both changes on multiple timescales [20, 22, 29–31, 36–40]. To this end, Shultz et al. showed that one's measured laxity changes cyclically across the menstrual cycle and during acute exercise intervals [20, 23, 29].

From this perspective, it is apparent that a lone measure for knee joint laxity is insufficient to characterize one's joint stability. In the clinical setting, knee laxity is often graded subjectively using a battery of manually administered tests. Manual laxity tests allow the clinician to gauge the structural integrity of the ligaments and other joint structures through a series of manual manipulations to the shank. Each individual test is graded one an ordinal scale, generally, from 1-3 where 1 represents normal laxity and 3 represents a suspected injury. A meta-analysis of manual laxity assessment techniques found that the Lachman's test was the most valid test for the assessment of sagittal plane laxity, specific to diagnosing ACL ruptures [41].

Despite showing evidence of laxity tests being reasonably reliable when conducted by the same examiner these tests are nevertheless criticized for their subjectivity and lack of generalizability [42].

To address this issue, instrumented knee arthrometers have aimed to objectify and generalize measurements of knee joint laxity so that clinicians and researchers alike may develop a better understanding of knee laxity from a cross-sectional rather than personal perspective.

Of the commercially available devices the KT – 1000 (MEDmetric, Corp., San Diego, CA) is widely considered the gold-standard in instrumented arthrometry. The KT – 1000 and its successor, the KT – 2000 were designed to emulate the Lachman's test while providing an objective measurement of anterior/posterior translation of the tibia relative to the displacement sensor in contact with the patient's patella. Comparative studies confirm the KT – 1000/2000s validity and reliability against other well-established devices [43–48]. However, the company producing the device has since ceased its manufacturing operations. More recently the GNRB (Genourob, Laval, France) and the Robotic Knee Testing system (RKT, Arthrometrix LLC, Atlanta, Georgia), both robotic arthrometers, have taken aim at filling the void left behind by the KT devices. Initial studies concerning the GNRB suggested that the GNRB performed more reliably compared to the KT devices [49]. However, subsequent studies have challenged these claims, citing significant sources of operator and measurement- based error [50,51]. Similarly, the RKT has sought to eliminate reliability concerns in instrumented arthrometry with its tri-axial system [52–54]. Initial reports from Branch and colleagues showed excellent transverse plane repeatability and interrater reliability (ICC = 0.87-0.99) [52, 54].

However, a recent study noted that, while the RKT did show promising results for frontal and transverse plane rotations for specific injuries, the measurements provided negligible diagnostic information compared to the 'healthy' contralateral limb [53].

While instrumented arthrometers have no doubt furthered our understanding of laxity and its relations to knee injury risk and OA development, they are not without their limitations. The purpose of these devices is to establish a measure that is indicative of true bony movement associated with an applied force, however current device designs often have not been validated against true bony movement. For non-robotic devices, such as the KT – 1000, operator introduced variability is a primary concern for measurement reproduction [55,56]. However, operator-dependent reliability concerns are not only limited to manually manipulated devices [50,51]. Furthermore, while it has been demonstrated that tibial rotation plays a significant role in measured tibial motion, many devices do not have a well-controlled method for ensuring repeatable tibial rotation during testing [6,10].

Another primary issue in instrumented arthrometry is soft tissue artefact. Soft tissue artefact, by definition, is a broad term given to the measurement error associated with calculating true bony movement, particularly in kinematics research when soft tissues are present. This phenomenon arises due to the unpredictable volumetric deformation of non-osseous tissues (primarily muscular, adipose and dermal) during movement tasks. Soft tissue artefact is particularly evident during sagittal plane arthrometry where the calf is the primary loading point for measurement of anterior displacement [50,51].

Although soft tissue artefact is a recognized issue in arthrometry, there is very little research quantifying its effect on arthrometric measurements. Alqatahni et al., however, found evidence supporting the notion that body-mass-index (BMI) affects measurement readings on the GNRB arthrometer [51]. This concern was echoed by Mouarbes et al. who were critical of the GNRBs methodology for measuring anterior/posterior displacement [50].

While soft tissue artefact has been a primary concern at the point of loading or measurement application, less attention has been paid to its presence at the thigh. To my knowledge, only one such study exists that examines the effect of error introduced by the thigh. In this study Draganich et al. demonstrated that arthrometric measurement variance was significantly reduced in the GENUCOM system when using an enhanced thigh restraint method during multiplanar laxity assessment [57]. In fact, the enhanced restraint thigh restraint method reduced day-to-day measurement variability nearly two-fold when compared to the manufacturer's recommended procedure [57].

This finding is critically important when considering the current state of instrumented arthrometry. While attention has been paid to eliminating operator error through the development of robotic testing protocols, no currently available arthrometer explicitly addresses the issue of thigh movement during arthrometric assessment, which is an evident source of significant measurement error.

Formally, this highlights the need for a device that addresses this issue. Therefore, the purpose of this thesis is to develop an evidence-based thigh stabilization system for use in combination with a novel tri- axial instrumented arthrometer.

Effective thigh stabilization system should improve the overall validity, generalizability and reliability of measurements obtained using the novel knee arthrometer by ensuring that thigh segment motion is minimized during multi-planar arthrometric evaluation. Moreover, improved thigh stabilization system has the potential to reduce both systematic and random measurement errors in instrumented arthrometry first by reliably fixating the thigh such that it minimizes movement during arthrometric evaluation in all three planes of motion.

I.1. Specific Aims: Validation the Thigh Stabilization System

H1: The thigh stabilization system will significantly limit thigh movement in the sagittal plane by preventing the thigh from producing any significant (0.5±0.5mm) amount of translation in the sagittal plane during laxity assessment.

H2: The thigh stabilization system will significantly limit thigh movement in the frontal plane by preventing the thigh from producing any significant (1.0 ± 0.5^{o}) amount of rotation in the frontal plane during laxity assessment.

H3: The thigh stabilization system will significantly limit thigh movement in the transverse plane by preventing the thigh from producing any significant (2.0 ± 1.0^{o}) amount of rotation in the transverse plane during laxity assessment.

In order to test these hypotheses, optical 3D motion capture markers will monitor motion of device relative to optical markers embedded in the femur of the limbs being tested (n =3) during a series of arthrometric tests (n = 10 per limb) all testing planes. Due to the previously specified need to track true bony movement of the femur during testing, this study will investigate the validity of the thigh stabilization system using a cadaver limb model. In order to establish agreement between analyses, the Bland-Altman Limits of Agreement methods will be employed as a test for all hypotheses [58].

CHAPTER II

REVIEW OF THE LITERATURE

II.1. Anatomy

To more fully understand how stabilization of the thigh and proximal portion of the knee joint is a critical factor in the objective measurement of knee joint laxity, it is first necessary to understand its basic anatomy. The knee joint is a modified hinge type synovial joint generally considered to be comprised of two lesser joints. These lesser joints, the tibiofemoral and patellofemoral joints are joined together by three bones, five primary ligaments, and acted upon by two major muscle groups.

II.1.1. Bony Anatomy

The three constituent bones of the knee joint are the femur, tibia and patella. The femur, the largest bone in the body, has two rounded bony prominences at its distal end characterized as the medial and lateral condyles [59–63]. These condyles are the femur's articular surface for the tibiofemoral joint. Proximal and lateral to the femoral condyles lie the femoral epicondyles. The epicondyles serve as primary proximal attachment points for the collateral ligaments [59]. The femoral trochlea (groove) lies between these two prominences. Posterior to the femoral trochlea is the intercondylar fossa, which bifurcates the medial and lateral condyles. The intercondylar fossa is the primary proximal attachment points for the anterior and posterior cruciate ligaments (ACL/PCL).

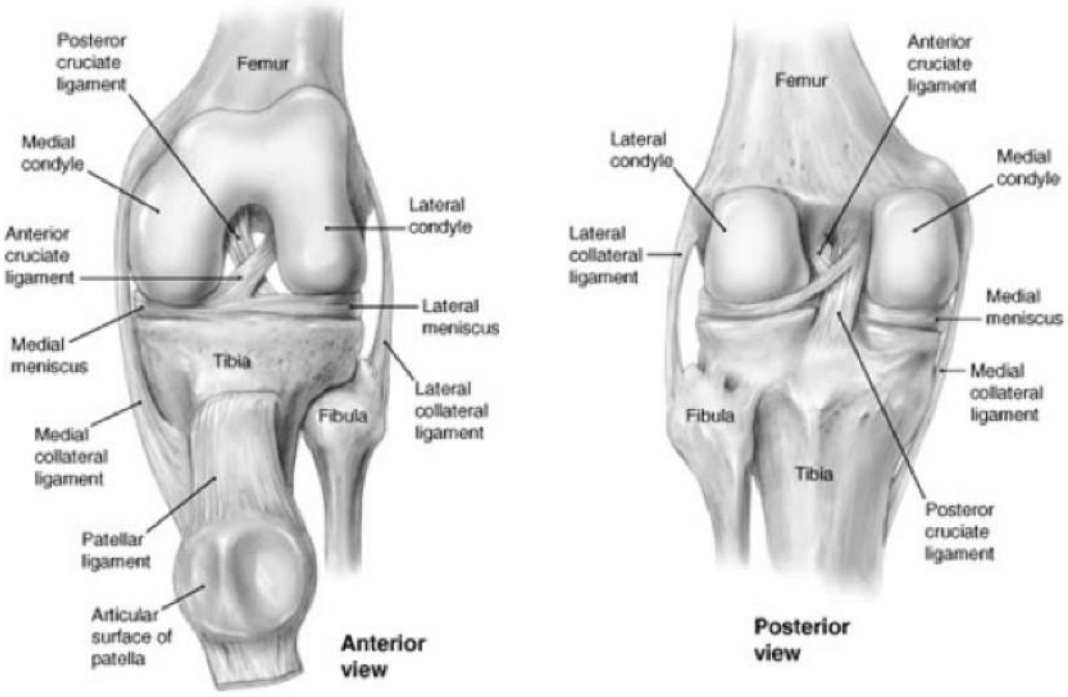

Figure II.1. Gross Anatomy of the Knee Joint [1]

The tibia lies distal to the femur. On its proximal end the tibia has two corresponding articular surfaces also known as the medial and lateral tibial condyles. Together these condyles make up what is known as the tibial plateau, which interfaces and articulates with the femoral condyles. The intercondylar eminence divides the two condyles and gives rise to the insertions of the ACL and PCL. Slightly inferior to and on the anterior surface of the tibia lies a bony prominence known as the tibial tuberosity. The tibial tuberosity is the primary insertion of the patellar ligament.

The smallest bone of the knee joint, the patella, lies anterior to the articular surfaces of the femur and tibia within the patellofemoral groove. The patella is the largest sesamoid bone in the body and plays an important functional role in the control of the knee joint. Its position on the anterior surface of the femur and tibia

allows it to augment extensor muscle efficiency by increasing their lever arm about the joint [64]. The inferior apex of the patella is connected to the tibial tuberosity via the patella tendon, while the superior base of the patella serves as the insertion for the quadriceps tendon. Its posterior surface has both medial and lateral facets, which articulate with the medial and lateral condyles of the femur.

II.1.2. Joint Anatomy

The larger of the two constituent joints, the tibiofemoral joint, joins the thigh at the distal end of the femur to the shank at the proximal end of the tibia [61,63, 65]. The femur's medial and lateral condyles are convexly curved such that they interface and rest within the concave architecture of the corresponding tibial plateaus [61,63,65]. The articulating surfaces of tibiofemoral joint are each lined with distinct cartilaginous tissues [61,63,65]. The femoral condyles are lined in hyaline (articular) cartilage, an avascular, aneural and alymphatic structure. Opposite to the condylar surfaces, lie the menisci, which are superiorly attached to the medial and lateral tibial condyles [62,63,65–67]. The menisci are comprised of fibrocartilage and contain both vascular/avascular as well as neural/aneural regions. In comparison to hyaline cartilage, fibrocartilage has a much denser collagen matrix, making it much more well-suited to sustain heavy, repeated loading. To this end, the articular cartilage present in synovial joints such as in the human knee greatly diminishes viscous friction and aids in force dispersion across the joint, preventing damage under high loads [61].

The patellofemoral joint is situated anteriorly to tibiofemoral joint which sits the patella bone within the femoral trochlea [62,63,65–68]. Similar to the tibiofemoral joint the patella is cushioned by the articular cartilage of the distal femur [62,63,65–68]. This cartilage allows the patella to glide within the trochlear groove, creating a smooth fulcrum around which the greater knee joint rotates in the sagittal plane [62,63,65–68].

II.1.3. Active Stabilizers of the Knee

Active stabilization is a result of contractile forces acting on the joint. In the knee, active stabilization of the joint is controlled primarily by the quadriceps and hamstring muscle groups [59,61,63,67,69].

The quadriceps femoris group has its insertion on the patella and tibial tuberosity via the quadriceps tendon and patellar ligament, respectively. The four muscles included in this group are the rectus femoris, the vastus lateralis, the vastus intermedius and the vastus medialis. Contraction of these four constituent muscles through their common insertion on the patella causes superior displacement of the sesamoid within the femoral trochlea and subsequent extension of the knee joint [59, 61,63,67–69].

Conversely, the hamstring group, causes flexion of the knee joint when contraction occurs. The hamstring group is composed of the biceps femoris, the semimembranosus and the semitendinosus. The biceps femoris has its insertion on the lateral condyle of the tibia and the head of the fibula, while the semimembranosus and semitendinosus have their insertion on the tibia along the posterior medial condyle [70]. In the frontal and transverse planes, the primary active stabilizers about the knee joint are the hamstring group, vastus lateralis, adductor magnus, gracilis, sartorius, tensor fascia latae and the popliteus [61–63].

From the hamstring group, the semitendinosus and semimembranosus serve to create medial rotation of the joint while the biceps femoris acts to create lateral rotation of the joint.

Similarly, the gracilis and adductor magnus are primarily responsible for adduction of the femur, while the tensor fascia latae and the vastus lateralis serve to externally rotate and abduct the femur. In a non-clinical population control from these muscles alleviates strain on the passive stabilizers and prevents excessive motion of the joint during movement tasks [70, 71].

II.1.4. Ligament Anatomy

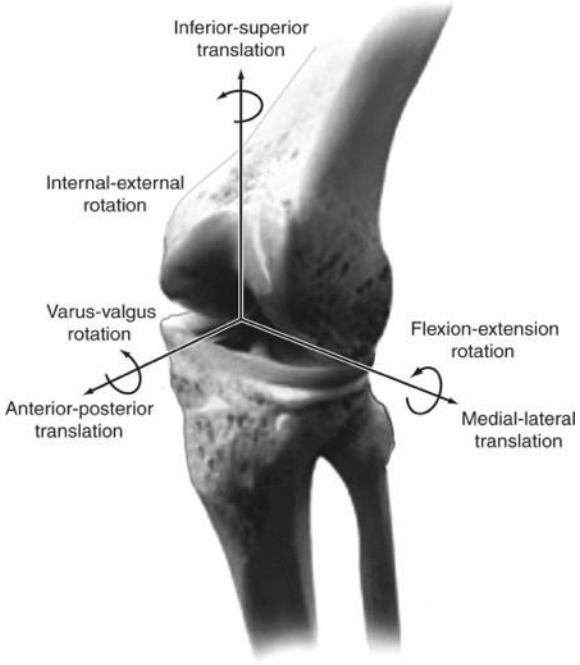

Figure II.2. Six Degrees-of-Freedom Within the Knee Joint [2]

The knee joint is capable of moving with six degrees of freedom and is comprised of three translations and three rotations (Figure II.2). These motions are guided, passively, by the ligaments of the knee as well as its aforementioned joint structure.

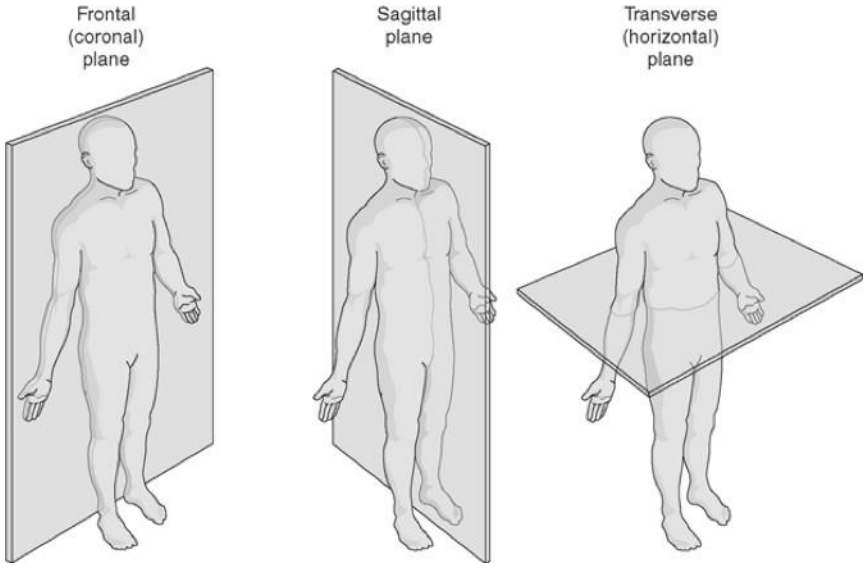

Figure II.3. Anatomic Planes of Motion [3]

From a descriptive perspective, we can describe these motions as happening within three distinct planes of motion (Figure II.3) with knee motion being described as Anterior – Posterior in the sagittal plane; Varus – Valgus in the frontal plane, and Internal – External in the transverse plane. Ligaments are a collection of dense, fibrous and relatively inelastic collagenous tissue that join bone to bone.

Their primary purpose, in the knee, is to provide stability to the joint and to prevent excessive movement of the femur, tibia and patella in any one of the three planes of motion. They do this by acting in tension to resist loads applied to the joint. In contrast to tendons, ligaments do not directly transmit contractile forces and are, as such, considered to be passive stabilizers of the joint.

The passive stabilization of the knee joint is generally controlled by the five primary ligaments of the knee [61,63]. Each of these five ligaments demonstrates specific functional characteristics based on its attachments and fiber directionality, from which it derives its strength. Despite their differential attachments, each ligament does not accept any load singularly. Rather, all applied loads are shared between at least two passive structures.

II.1.4.1 Collateral Ligaments

The collateral ligaments of the knee are a pair of two relatively flat ligaments that arise from the femur and run to the tibia and fibula. Their name is indicative of their directionalities, which is side-by-side or parallel with the long axis of the leg. In general, they resist frontal plane motion of the tibia relative to the femur.

II.1.4.2 Lateral Collateral Ligament (LCL)

The LCL is a broad, tubular ligament located on the lateral aspect of the joint capsule. The LCL has its proximal attachment on the lateral epicondyle of the femur and its distal attachment on the head of the fibula [60, 61, 63, 65, 67, 69, 72]. Biomechanically, the LCLs primary function is to resist varus stress, though it also accepts secondary loads during external rotation and posterior translation of the tibia [61,65,72]. During extension, the LCL accepts approximately 55% of varus rotational forces at 5 of flexion and 69% of varus forces at 25 of flexion [73]. Maximum varus deflection occurred at 30 of flexion [73].

II.1.4.3 Medial Collateral Ligament (MCL)

The MCL is often described as being comprised of two smaller ligaments, the superficial (sMCL) and deep (dMCL) medial collateral ligaments [59,61–63,65,67,69,72].

Like the LCL, the sMCL has its femoral attachment on the medial epicondyle of the femur, however, it has two distal attachments [59, 61–63, 65, 67, 69, 72]. The more proximal attachment merges with the semimembranosus tendon of the hamstring group while the more distal attachment inserts itself on the posteromedial crest of the tibia [59, 61–63, 65, 67, 69, 72]. The dMCL can be further divided into two subsidiary ligaments: the meniscofemoral and meniscotibial ligaments [59, 61–63, 65, 67, 69, 72]. The meniscofemoral ligament has its femoral attachment slightly distal to the sMCLs on the medial epicondyle, while its tibial attachment inserts itself into the medial meniscus [59, 61–63, 65, 67, 69, 72]. The meniscotibial ligament, while part of the MCL, does not actually cross the joint line of the knee. Instead, it has its proximal attachment on the medial meniscus with a more distal attachment to the articular cartilage of the medial tibial plateau [59, 61–63, 65, 67, 69, 72]. As a whole, the MCL's primary biomechanical purpose is to resist valgus rotation as well as external rotational forces. Grood et al. determined that around 5Âř of flexion the MCL provided approximately 57% of restraint to valgus rotation and up to 80% of the same forces at 25 of flexion [73].

II.1.4.4 Patellar Ligament

The patellar ligament, which as noted previously, extends from the apex of the patella and inserts on the tibial tuberosity. The patellar ligament is collinear with the insertion of the quadriceps tendon and as such appear seamless along the anterior surface of the knee joint.

II.1.4.5 Cruciate Ligaments

The cruciate ligaments lie between the femoral and tibial condyles within the joint capsule. They are named due to their crossing of each other within the joint capsule.

II.1.4.6 Posterior Cruciate Ligament (PCL)

The PCL arises from the lateral edge of the medial femoral condyle and inserts on the posterior aspect of the tibial plateau [67]. Biomechanical studies have determined that the primary responsibility of the PCL is to accept loads causing posterior translation of the tibia relative to the femur, though it also acts to stabilize the joint when varus-valgus and internal/external rotational loads are applied [67]. The PCL is also tensioned during hyperextension of the joint [67].

II.1.4.7 Anterior Cruciate Ligament (ACL)

The ACL originates from the posteromedial corner of the lateral femoral condyle and runs inferiorly, medially and anteriorly to its insertion on the anterior horn of the medial meniscus which lies within the intercondylar eminence of the tibia [61,63,65,74]. Similar to the MCL, the ACL is comprised of two separately functional bundles: the anteromedial (AM) and the posterolateral (PL) [61, 63, 65, 74].

The two bundles have been shown to accept functionally different anteroposterior and rotational loads [12]. Specifically, the AM bundle accepts greater loads at increasing angles of flexion compared to the PM bundle.

Functionally, the ACL's primary biomechanical purpose is to resist anterior translation of the tibia relative to the femur [12].

Estimates of this resistance predict that the ACL accepts greater that 80% of anteriorly directed forces on the tibia [12,61].

The two bundles have been shown to accept functionally different anteroposterior and rotational loads [12]. Specifically, the AM bundle accepts greater loads at increasing angles of flexion compared to the PM bundle.

Functionally, the ACL's primary biomechanical purpose is to resist anterior translation of the tibia relative to the femur [12]. Estimates of this resistance predict that the ACL accepts greater that 80% of anteriorly directed forces on the tibia [12,61].

The ACL also resists significant varus – valgus and internal/external rotational torques as indicated by the natural internal rotation of the tibia during unimpeded anterior translation of the tibia and frequent injury mechanism [12]. The ACL and PCL are also play an important role, structurally, in guiding the knee throughout its range of motion as their intersection serves as its axis of rotation [12, 15, 61, 63, 67, 73].

II.1.5. Mechanoreceptors and Joint Stability

In addition to their role in passive structural stabilization of the knee, the ligaments play an integral role in the neuromuscular control of the joint. Connections between the passive and active structures of the knee joint are facilitated by the proprioceptors within the ligaments. Mechanoreceptors such as Golgi tendon-like organs, Pacinian corpuscles, Ruffini corpuscles and free nerve endings, send kinetic and kinematic feedback to higher centers in the central nervous system which produce responses allowing the joint to compensate and react to changing stimuli.

In a pilot investigation conducted by Freeman and Wyke (1967), it was determined that ligaments were integral to normal neuromuscular function in felines. During their study they demonstrated that a lack of articular proprioceptive function presented itself following the resection of feline posterior or medial articular nerves [75]. This resection had significant implications to the voluntary movement patterns of the affected limb(s). In showing that articular feedback has demonstrable effects on gait biomechanics without disrupting the passive joint structures, it has been inferred that articular receptors play an integral role in the neuromuscular control of the lower limb [75].

Further investigations have given rise to evidence that the passive structures of the knee play an important role in dynamic control of the human knee joint [76–79]. Histologically, Kennedy et al. found that the passive structures of the knee were most notably afferently innervated by the posterior articular nerve, a branch of the tibial nerve and the terminal endings of the femoral nerve [80]. They found this innervation was comprised of a variety of nerve endings, including Golgi-tendon-like organs as well as Ruffini receptors [80]. Functionally, Tsuda et al demonstrated in vivo that the ACL has verifiable intrinsic control over the hamstring muscle group [76]. This was accomplished by direct stimulation of the ACL in a locally anesthetized human limb using a live-wire [76]. The electrical impulse administered to the ACL elicited a reciprocal contraction of the muscle group. In effect, this response stiffens the joint against impending anterior translation and guards the ACL from absorbing the maximal force of the translation.

When these structures are damaged, such as in the case of a ruptured ACL, the body no longer has the ability to sense excessive stretching of the joint through the effected pathways [76, 81–83]. Furthermore, even in the absence of a complete or partial rupture a decrease in proprioceptive capability has been shown to worsen with increasing knee joint laxity [76, 81, 82]. In both conditions, the diminished sensory capacity leads to the development of instability in the knee joint [15, 79]. A recent 2016 study reaffirmed this notion. Marreiros et al. concluded that subjects presenting knee instability in the form of excessive laxity were significantly worse performing than their gender-matched control subjects on a proprioception task designed to assess their ability to sense knee position [84]. Furthermore, the presence of increased joint laxity lead to an increased likelihood of developing osteoarthritis within the effected joint [84]. For this reason, the study of knee joint laxity and its longitudinal effects on joint health and injury risk have gained much traction in the academic communities over the past 50 years.

II.2. Laxity

Knee joint laxity is a measure of the mechano-elastic properties of the soft structural tissues of the knee joint, namely the ligaments. Mechanically, knee joint laxity can be characterized by the knee's total deformation at a fixed applied load. [42]. Though subjective, these manual laxity tests provide clinicians with a hands-on diagnostic tool that when performed masterfully, can be useful and reasonably accurate [74, 85].

Perhaps the most widely recognized manual laxity assessment is the Lachman's test. The Lachman's test is performed to assess anterior – posterior laxity in a patient which assesses ACL function.

To administer a Lachman's test the examiner supports the distal femur of the patient, who is lying on an examination table, with one hand and with the other hand on the proximal tibia administers anteriorly directed pulls. By manipulating the shank, the Lachman's test allows the examiner to, in their own way, assess total tibial translation under a manual load. Furthermore, the Lachman's allows the performer to assess what is clinically referred to as 'End Feel' [74,85]. End Feel is described as the feeling of a structural support resisting the manual pull at the end range of displacement [74, 85].

Similar to the Lachman's, the Anterior Drawer test also aims to assess the integrity of the ACL via an anteriorly directed pull. The test is administered first by placing the patient's hip in 45° of flexion and their knee in 90° of flexion [42, 74,85]. Once the patient is properly situated, the examiner fixates the lower leg by anchoring the limb with their body weight and then locates the joint line by palpation of the joint. The examiner then directs an abrupt series of pull anteriorly and assess tibial translation resulting from the pulls. A test is considered positive when more than 6mm of tibial translation occurs [42,74,85].

In both tests, it is necessary for the clinician to establish a patient-specific baseline laxity value by first assessing and comparing the injured side to the non-involved side (i.e. bilateral comparison) [42, 74, 85]. This enables the clinician to make a subjective assessment of each knee's ligamentous integrity relative to their own continuum [42, 74, 85].

The pivot-shift test is a dynamic test to examine anterolateral rotary instability of the knee. It is a more complex manual test to administer as it involves significant coordination efforts from the examiner [42,74,85].

To administer a pivot-shift test the examiner must control the patient's leg from full extension to approximately 40 degrees of flexion while also administering valgus and internal rotational forces at the joint line and ankle, respectively [42, 74, 85].

The test is considered positive when the examiner is able to create dynamic subluxation (partial-dislocation) of the tibia in the sagittal plane. A positive pivot-shift sign is highly specific for examining ACL injury, but may also relate to damage of other structures limiting anterolateral rotary instability such as the LCL, posterolateral capsule, arcuate complex and the Iliotibial Band [42, 74, 85]. Given its complexities the pivot-shift test is difficult to administer and often uncomfortable for the patient. Discomfort may cause guarding, which diminishes the test's ability to elicit a true dynamic subluxation sign when it does exist.

Examination of Varus – Valgus laxity is accomplished via the Varus – Valgus stress tests. These tests evaluate the degree of varus and valgus deflection, or joint opening, experienced when a load is applied [42, 74, 85]. To administer the tests the patient lies supine with their leg in either full extension, or in 20-30° of flexion. Once situated the examiner applies a varus or valgus load to the shank, while supporting the knee joint, to create rotation about the joint line. The examiner then evaluates the amount of opening that occurs about the joint line to make determinations about the structures effected by this test, which include but are not limited to, the ACL, PCL, MCL, LCL, and other accessory structures or muscles [42, 74, 85].

II.2.1. Clinical Effectiveness

In a meta-analysis examining the accuracy of manual laxity assessments in the detection of ACL injury the Lachman's test was found to be the most valid diagnostic assessment tool, having an 85% sensitivity (true-positive rate) and 94% (true-negative rate) specificity [42]. Sensitivity and Specificity are defined below in the equations below.

$$Sensitivity = \frac{\#\ of\ True\ Positives}{(\#\ of\ True\ Negatives + \#\ of\ False\ Negatives)} \qquad (II.1)$$

$$Specificity = \frac{\#\ of\ True\ Negatives}{(\#\ of\ False\ Positives + \#\ of\ True\ Negatives)} \qquad (II.2)$$

The anterior drawer also demonstrated strong specificity (92%) and sensitivity (91%) in chronic conditions (as defined by the clinicians in each study) but was not an effective test in acute conditions [42]. The pivot-shift test seems to be highly specific (98%) but not highly sensitive (24%) when the patient is not anesthetized, indicating limited clinical usefulness [42]. Furthermore, the varus – valgus stress test does not appear to be highly reliable in diagnosis of specific injury [42, 86]. Harilainen et al. found the varus – valgus stress test to have a sensitivity of 86% but did not report specificity [86].

Overall, manual tests for sagittal plane laxity assessment appear to be useful in general diagnostic scenarios. Other manual tests that assess transverse or frontal plane laxity, however, appear to have limited usefulness even in strictly for diagnostic purposes. Coupled with their subjectivity, manual laxity tests provide very little scientific insight into why, mechanically, someone might be presenting with an injury.

II.2.2. Instrumented Laxity Assessment

Despite being reasonably quick and useful to administer, manual laxity assessments have been under scrutiny since the mid 1970s due to their subjectivity. As evidence has accrued suggesting that laxity is an important physiological measure for diagnosing general joint instability and a predictor of traumatic knee injury and subsequent or independent development of debilitating joint diseases such as osteoarthritis, a push has been made to objectify these measurements [15,87–89]. To this end, instrumented knee arthrometers were developed to bring an air of objectivity to a process that has lagged behind the cutting edge of medical technologies.

II.2.2.1 KT – 1000/2000

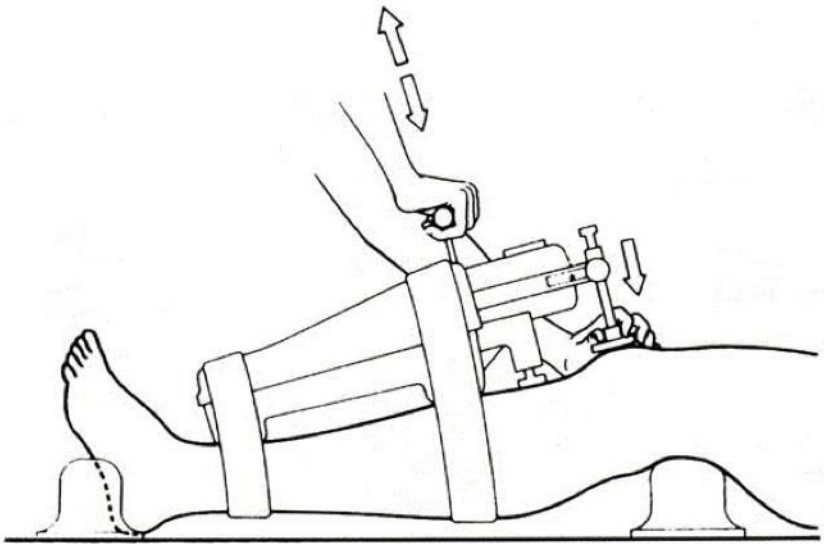

Figure II.4. The KT – 1000 Instrumented Knee Arthrometer [93]

The KT – 1000 (Figure II.4) Knee Ligament Arthrometer (MEDmetric Corp, San Diego, CA) is perhaps the most widely studied and frequently used instrumented knee arthrometer. Furthermore, it is often considered the 'Gold-Standard' against which all other instrumented arthrometers are compared [93]. The KT – 1000 and its successor, the KT – 2000, which includes a computerized graphical user interface for modulus evaluation, are designed to mimic the Lachman's test for anterior tibial translation. Figure II.4 shows how the device interfaces with the patient's leg. Once fastened to the patient, using the two straps seen in Figure II.4 the examiner uses the handle on the anterior face of the device to pull the tibia and body of the device anteriorly. The device, itself, obtains an objective measurement of displacement, read in 0.5-millimeter increments (mm), by tracking the relative movement of the device in relation to the platform used to seat the patella during testing.

To maintain consistency of measurement, the device emits audible beeps at three different frequencies to indicate when the examiner has reached a certain force-level [87, 90]. Typical force levels for examiner pulls are 89N, 134N, 200N and manual maximum pull [87, 90].

As in other manual laxity tests, it is commonplace for clinicians to examine both limbs to assess a patient-specific AP laxity profile. This is especially important when examining a suspected injury. For diagnostic purposes, most studies agree that a 3mm absolute side-to-side difference in displacement measurement between limbs is sufficient to indicate an ACL injury [47, 87, 90–92]. This differential value was first established in a large repeated measures study by Daniel et al. in 1985 who found that 100% of patients exhibiting an absolute differential laxity measurement of 3mm had an ACL disruption [91]. Despite this 3mm differential threshold Daniel et al. established as absolute, they also found that 95% of patients whose left-right difference were in the 2.0-2.5mm range also had an ACL injury [91]. These results remain true at both the lowest, 89N, and highest, manual maximum, force levels [91].

Though effective at obtaining an objectified measurement, the KT devices have drawn criticism surrounding the repeatability, generalizability and validity of its measurements. Sherman et al. reported that the KT – 1000 was able to correctly identify torn ACLs in 90-95% of cases, indicating strong validity of the device [92]. Similar results have been found in other studies, noting that validity of the KT devices invariably increase with increased applied force [6, 46–48, 87, 92, 93]. The KT draws its largest criticisms from its repeatability, and therefore the generalizability of its results to a greater population [6,56,94]. Its reliability is particularly low for interrater, between examiners, reliability [6, 56, 94].

Studied examining the reliability of the KT devices find that interrater reliability ranges from 0.65 - 0.92, while intrarater reliability ranges from 0.67 - 0.99 [43, 94, 95]. Furthermore, the KTs reliability compared to a clinician's manual assessment is also held in question [44]. Wiertsiema et al. found that the KT – 1000 was less reliable than the Lachman's test having an interrater reliability of 0.14 vs. 0.77 and an intrarater reliability of 0.47 vs. 1.0, respectively [44].

II.2.2.2 Rolimeter

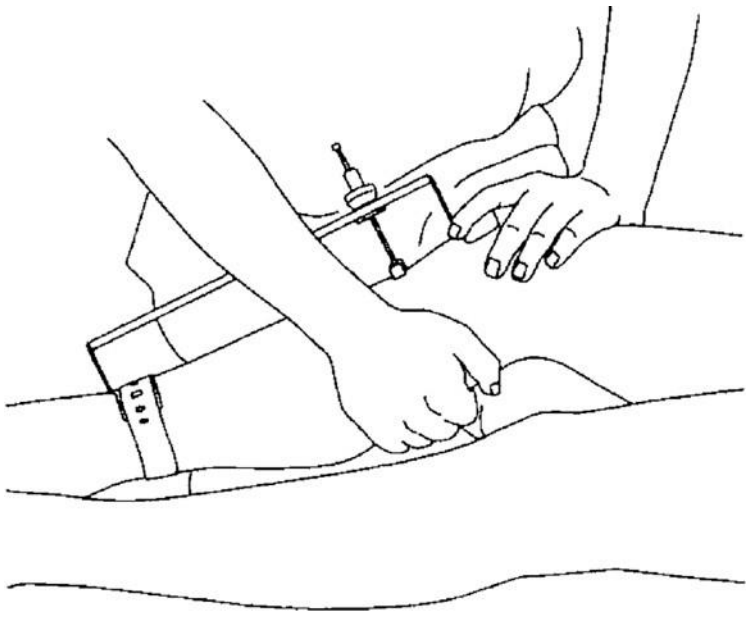

Figure II.5. The Rolimeter Instrumented Knee Arthrometer [5]

The Rolimeter is a basic instrumented arthrometer that allows the examiner to perform a Lachman's test and obtain an objective reading. The device measures anterior displacement of the tibial tubercle relative to the patella during manual manipulation of the shank via a linear displacement slide that measures in 2mm increments [6, 96].

As seen in Figure II.5, a measurement may be taken once the displacement slide is properly positioned over the tibial tubercle. Logistically, the Rolimeter measures sagittal plane laxity in a similar way to the KT, which also uses the patella and tibial tubercle as reference points for measurement.

Several studies have examined the Rolimeter's reliability and found it to be reliable between testers and across multiple time points [6,96,97]. However, Papandreou et al. found the Rolimeter to only have a pooled interrater reliability of ICC = 0.69, indicating moderate reliability [98]. Furthermore, Muellner et. al demonstrated that the device's reliability suffers when used by an inexperienced examiner [99]. It would also seem that the increased reliability of this device may be due to its relatively low measurement resolution of 2mm [97]. Based on the data presented by Ericsson et al. it seems likely that the examiner interpolates the device's reading to the nearest 1mm increment if the test does not fall exactly on one of the 2mm increments [97].

The Rolimeter is a newer device than the KT and thus has less evidence supporting or refuting its usefulness, however, given its similarities to both manual laxity assessments and the KT device it can be assumed to have many of the same drawbacks. Most notably, the Rolimeter lacks the ability to maintain consistent tibial transverse plane orientation and has no defined measurement protocol. The latter issue has more to do with fixation of the patella pad which relies on the examiner to fixate it against the patient.

II.2.2.3 Stryker ® Knee Laxity Tester

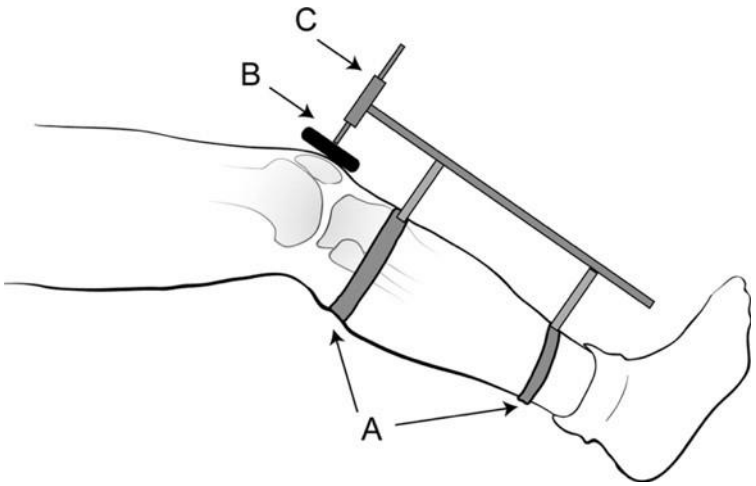

Figure II.6. The Stryker Knee Laxity Testing Device [6]

The Stryker Knee Laxity Tester (Stryker Corporation, Kalamazoo, MI), like the Rolimeter, is a manually manipulated laxity device. The device as seen is FigureII.6 is fixated to the shank using two hook-and-loop straps and measurements are obtained by pushing or pulling the spring-loaded piston which rests on the patella. Measurement readings were given in 1mm increments from the piston. Highgenboten et al. found that the Stryker device to be less reliable (Inter-test ICC = 0.74) during anterior testing than both the KT – 1000 (ICC = 0.87) and the Genucom system (ICC = 0.96) [48]. Similarly, a comparative study of both healthy and ACL-deficient knees determined the Stryker device was comparably accurate to the KT – 1000 [87]. However, it should be noted that Highenboten et al. maintains that their reliability metrics were only obtained by a well-trained examiner [48].

In contrast to the findings of Anderson et al., a meta-analysis examining the merits of the Stryker device concluded that it was inferior to the KT – 1000 on four key metrics: sensitivity, specificity, accuracy, and positive predictive value [41]. They did conclude, however, that the Stryker device was performed comparably in their positive likelihood ratio, which describes the ratio between sensitivity and specificity [41]. Ultimately, research did not support the adoption of the Stryker device over the other devices of the time, namely the KT – 1000. This presumably contributed to the eventual discontinuation of the device and continued rise in popularity of the KT devices.

II.2.2.4 Genucom Knee Analysis System

The Genucom Knee Analysis System (FARO medical Technologies Inc, Montreal, Canada) is a defunct multiplanar laxity device first described by Oliver and Coughlin in 1987. The computer-aided Genucom device was able to assess both absolute displacement and stiffness in the sagittal, frontal and transverse planes. [48, 100, 101].

In the pilot study conducted by Oliver and Coughlin, the device demonstrated its ability to measure various planar movements (Anterior – Posterior and Varus – Valgus) [101]. Prior to testing a six degree-of-freedom soft-tissue artefact compensation test was performed on the distal femur to enable assessment of true bony movement. According to Patent # US4649934A the soft tissue compensation test is conducted manually with a probe that measures tissue displacement under loading [102]. The measured displacement is verified through repeated testing and then cancelled out during laxity testing [102]. The force and displacement measurements calculated by the device were recorded on a 6-degree-of-freedom electrogoniometer [101].

When compared to limited clinical evaluations the Genucom performed reliably [100, 101].

However, subsequent investigations attempting to establish more rigorous reliability metrics did not agree on the device's overall usefulness [41,48,87,100]. Highenboten et al. found the device to be both accurate and reasonably reliable, while in contrast, Wroble et al. found the device to exhibit significant variability between day and patient [48,100]. The authors attributed much of this variability to the complex set-up required to obtain the precise measures of stress and true bony movement [48, 100]. Furthermore, the devices validity was called into question due to an abnormally high false positive indication during anterior-posterior testing [48, 87, 103]. Ultimately, despite promising initial test results the Genucom failed to produce consistent results, leading to its demise in the research community.

II.2.2.5 UCLA Instrumented Clinical Testing Apparatus

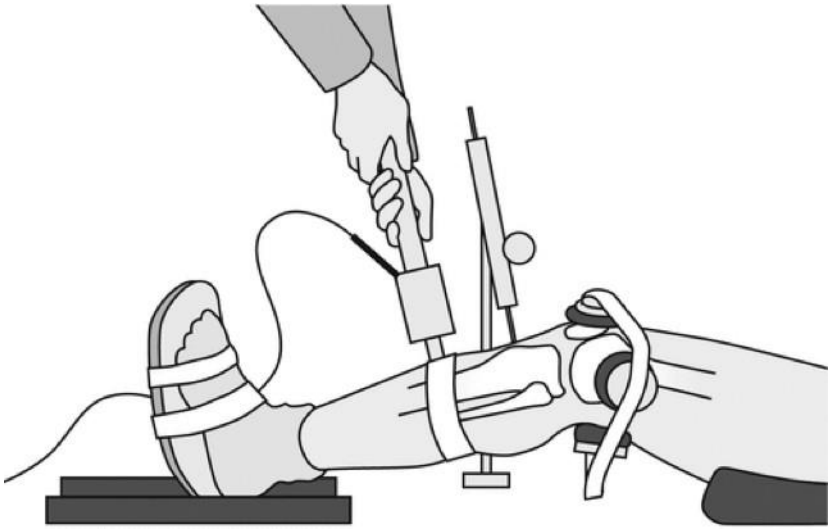

Figure II.7. The UCLA Clinical Testing Apparatus [7]

The UCLA Clinical Testing Apparatus was a computerized arthrometer which arose from a prototype device first described by Markolf et al. in a 1976 publication examining the devices ability to quantify knee stability [104,105]. The device itself was first described in a 1978 study on human limbs [105]. Figure II.7 depicts the general device set-up. To obtain a measurement the examiner manipulates the patient's limb using the force transducer. Displacement is measured separately with the linear displacement gauge which is positioned over the patient's tibial tuberosity. The device's modular set-up allowed for evaluation of displacement and stiffness characteristics in both the sagittal and frontal planes. Test data was recorded on a local computer.

In a pilot study utilizing the UCLA Clinical Testing Apparatus, Markolf et al. compared the structural stability of healthy cadaver knees to the same cadaver knees with artificially injured internal structures [104].

This study was able to characterize how laxity and stiffness of the knee joint was altered in various angles of flexion and in multiple planes [104]. The follow-up study, using the non-cadaver prototype found that the UCLA device was valid and able to detect ACL ruptures in up to 95% of presenting cases [105]. In addition, a 1987 paper highlighted the devices ability to discern ACL-deficient patients from those that had intact ACLs. It also made note that many patients with reconstructed ACLs have laxity profiles similar those of an ACL-deficient patient after certain time points [106]. While results supporting the device were promising, it was never recreated commercially, thus evidence supporting its utility are limited [92, 104–106].

II.2.2.6 TELOS

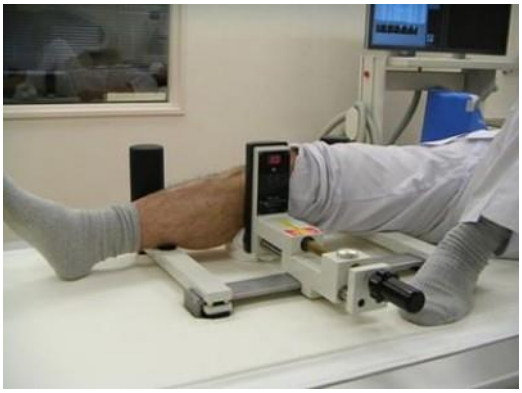

Figure II.8. TELOS Instrumented Knee Arthrometer [8]. To load the joint the black rotary handle extends the padded arm. Applied force is measured via sensors located within the extensible arm, while displacement is measured radiographically.

The TELOS (Telos GmbH, Laubscher, Hölstein, Switzerland) device is a simple three-point bending device that allows for static loading and radiographic imaging to assess sagittal and frontal plane laxity. The device utilizes a linearly extensible pad with an embedded force sensor to load the knee joint (Figure II.8).

Studies examining the TELOS' validity and reliability have been met with mixed results. Garavaglia et al. found the TELOS to have a high interobserver reliability ICC = 0.95 - 99 as well as a sensitivity of 93.3% - 97.6% and a specificity of 77.8 - 86.7% based upon the subject's positioning [107]. Schulz et al. found similar results, where in a study examining device reliability when assessing posterior knee joint laxity (n = 787) the TELOS boasted an interrater reliability of ICC = 0.91 and an intrarater reliability of ICC = 0.95 [108]. They also found that the TELOS produced similar results between experienced and inexperienced examiners [108]. Hewett et al. determined that the TELOS was superior in assessing posterior, sagittal plane laxity [109].

However, other studies have found the TELOS' usefulness to be limited [45, 110–112]. In a few studies, the TELOS has reported sensitivity and specificity values as low as 59% and 75%, respectively [45,111–114]. Furthermore, the device's relatively low measurement resolution of 0.5mm creates diagnostic issues when using the well-accepted 3mm-differential cut-off for diagnosing ligament ruptures. Ultimately, the TELOS device has not proven to be a more accurate or reliable arthrometer than its competitors. Given the device's requisite radiation exposure, it is therefore, difficult to recommend its repeated use in non-diagnostic scenarios.

II.2.2.7 Vermont Knee Laxity Device (VKLD)

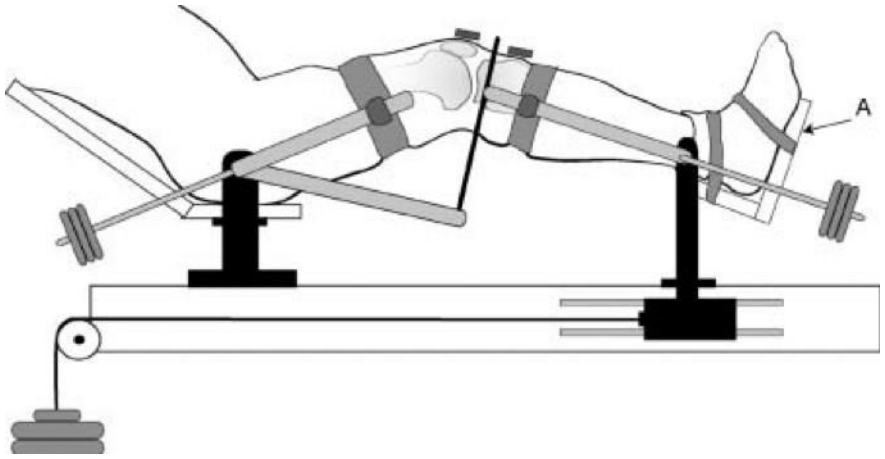

Figure II.9. The VKLD Instrumented Knee Arthrometer [92]. The VKLD enables multiplanar arthrometric evaluation in both weightbearing and non-weightbearing conditions. Force application is done manually through as system of levers and slides. Force magnitude and displacement measurements are monitored by computerized software via a system of sensors.

The VKLD was heralded as a comprehensive laxity assessment device developed and first described by researchers at the University of Vermont. The VKLD was developed to assess how knee stability is affected during non-weight-bearing, weight-bearing and transitional non-weight-bearing to weight-bearing states. Figure II.9 depicts the devices basic setup. The VKLD characterized laxity profiles in these three separate states through a system of linear slides, hinges and pulleys which allowed the limb to be loaded using weights proportional to the subjects' body mass [93, 115, 116]. Uh et al. first described the devices ability to characterize A-P laxity during non-weight-bearing and weight-bearing conditions [92].

In the non-weight-bearing condition they found that the device's measurements were not significantly different than the measurements of the KT – 1000, although the VKLD did exhibit greater standard deviation between subjects at both force levels (89/200N) [93]. In the weight-bearing condition the device measured nearly 70% less AP laxity [115]. A follow-up study examining the effects of transitional non-weight-bearing to weight-bearing on anterior translation of the tibia, found that the ACL deficient limb exhibited translations nearly four-times greater than the intact limb [115].

A 2007 study repurposed the device to examine laxity for both varus-valgus and internal – external rotations [116]. Shultz et al. determined that the VKLD is able to provide clinically relevant and reliable measurements for both planes of motion in both the non-weight-bearing and weight-bearing conditions [19, 116]. Given the device's reliability, several follow-up studies using the VKLD have examined the device's ability to characterize laxity and stiffness profiles in all three planes [21, 23]. These studies have highlighted that laxity profiles are different between genders and across the menstrual cycle in females [20–23, 38].

The VKLD, though a prolific research tool, is an impractical clinical tool due to its immense size, complicated setup and operational procedures. That being said, the VKLD has been instrumental in establishing a clinical need for multiplanar laxity assessment. The device has not produced any studies since 2012, however, it remains an important benchmark tool in instrumented arthrometry.

II.2.2.8 GNRB

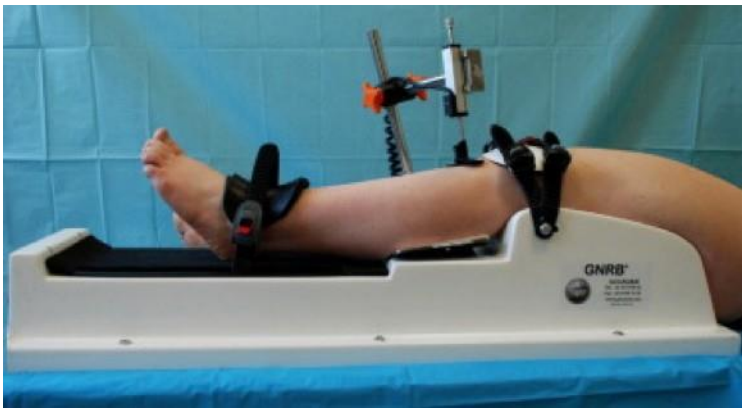

Figure II.10. The GNRB ® Arthrometer [9]

The GNRB (Genourob Inc., Laval, France) is a recently developed robotic 'laximetry' device. In its basic form, the device is capable of mimicking a Lachman's test by applying a robotically controlled 'thrust' to the base of the calf. Anterior displacement of the tibia relative to the femur is measured via an articulating linear displacement transducer placed on the tibial tubercle with 0.1 mm precision [9]. A separately sold attachment allows for posterior displacement measurements to be applied by fixating the shank to the robotically controlled platform and allowing it to be pulled posteriorly [9]. Figure II.10 shows the GNRB as set up for measurement of anterior displacement. Additionally, the GNRB monitors electrical activity of the knee flexors and patella 'pressurization' during testing. Muscle activity is monitored via surface electromyography and halts testing if significant muscular activity is detected during testing. Similarly, patella fixation pressure is measured via sensors embedded within the patella strap mechanism.

With the decline of MEDmetric and discontinued support of the KT devices, the GNRB has gained increasing popularity amongst clinicians and researchers alike since its introduction in 2007 [9]. The initial publication conducted by Robert et al. consisted of three separate studies. The first examined healthy subjects; the second examined subjects with complete ACL ruptures and the third examined those with confirmed partial ruptures. The first experiment, conducted on n = 20 subjects (17 male, 3 female) determined that the device demonstrated a significantly greater interrater reliability compared to the KT – 1000 between two different operators with varied experience [9]. In the second study, examining subjects with complete ACL rupture, they determined that at the traditional 3mm differential diagnostic value the sensitivity and specificity of the GNRB was 70% and 99% at 134 N of force, respectively [9]. This resulted in a correct diagnosis in 88% of subjects [9]. Similarly, the third study found that at a 1.5mm differential cutoff the GNRB could diagnose partial ACL ruptures with a sensitivity and specificity of 80% and 87%, respectively, resulting in a correct diagnosis in 81% of cases [9].

Subsequent studies have echoed these findings,concluding that the GNRB is a superior arthrometer to the KT – 1000, which has long been considered the gold standard in instrumented arthrometry. To this end, Collette et al. found that the GNRB was more reliable than the KT – 1000 based on Mean ±Standard Deviation repeated measures over a 10-day testing protocol [49]. Bouguennec et al. and Ryu et al. concluded that the reproducibility and diagnostic value of the GNRB was superior to that of the TELOS, respectively [110,111]. Furthermore, Jenny et al. demonstrated that the GNRBs measurement of bony movement was reasonably accurate when compared to a fluoroscopic navigation tool [117].

These results are, however, contested. At least one study advocating for usage of the GNRB cites conflicts of interest in reporting due to the authors being the primary inventors of the device [9]. Additional tests comparing the GNRB to the KT – 1000, TELOS and Rolimeter have asserted that the GNRB demonstrates poor reliability and validity [50,51,110,118]. In these studies, the authors have found the GNRB's reproducibility to be as low as ICC = 0.210 for intrarater and ICC = 0.220 for interrater reliabilities [50,51,118]. Furthermore, recent evidence suggests that the devices patella fixation system, and tibial displacement sensor, imparts significant measurement variability [50,51]. Given the mixed support regarding the GNRB it is difficult to say with confidence that, as has been claimed in early studies, that the device is superior.

II.2.2.9 ROTAM/ROTAB

Figure II.11. The ROTAM (left) the ROTAB (right)[10,11]. The devices are used to assess transverse plane (internal/external) rotational laxity and simultaneous medial rotation of the tibia during assessment of anterior laxity, respectively.

The ROTAM (Genourob Inc, Laval, France) (Figure II.11) is a relatively new device, so little is known or published about this device However, according to Samson et al. the device functions similarly to the GNRB but includes a fixation boot on the shank and foot through which computer monitored internal and external rotational torques are applied [98]. The authors claim that the device applies between 3-10Nm of torques and is capable of measuring rotations to 0.1 of precision through the use of a gyroscope [98]. Aside from the pilot study, Ruiz et. Al used the ROTAM to examine the effects of the ACL and other anterolateral structures on the examination of internal rotation [99]. Using the ROTAM they found that transection of these structures, in a cadaveric limb, significantly increased internal rotation under loading [99].

The ROTAB (Genourob Inc, Laval, France) (Figure II.11) is also a newer device, with many similarities to the GNRB. The ROTAB, however, simultaneously measures anterior translation of the tibia and passive rotation of the limb under loading conditions. A study conducted by Senioris et al. aimed to assess the ROTAB's reliability in obtaining these simultaneous measurements. They concluded that the device was reliable ICC = 0.97) when examining (n= 14) fresh cadaver limbs. [96].

II.2.2.10 Robotic Knee Testing System (RKT)

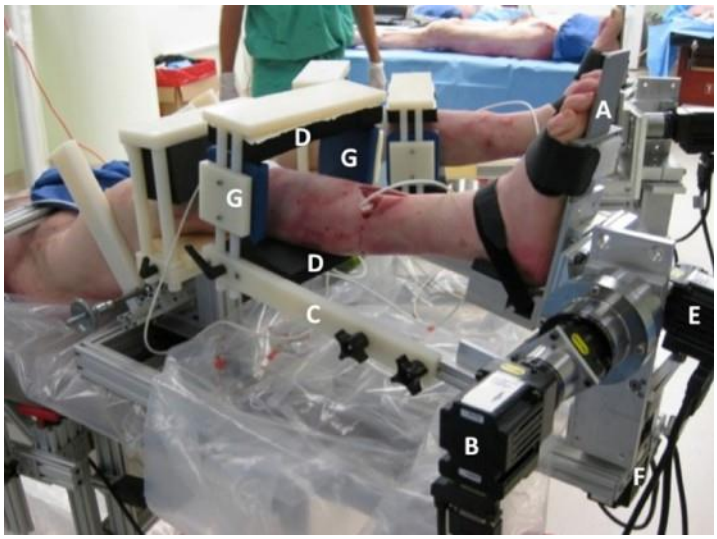

Figure II.12. The Robotic Knee Testing Apparatus (RKT). The RKT is depicted with a cadaveric limb fastened into footplates (A). Sagittal plane manipulation is controlled via a motor (B) which applies force through the application arm (C). Pads (D)transfer force anteriorly to the tibia and beneath the calf. Internal - external tibial rotation is generated by a second motor (E). A third motor (F) creates varus-valgus rotational movement by manipulating a second set of pads (G) [53].

The RKT (Figure II.12) is a multiplanar, non-commercially available, instrumented arthrometer developed by TP Branch and colleagues in association with Branch's company (Arthrometrix LLC, Atlanta, Georgia). In a study first published in 2015 Branch et al. examined this device's ability to quantify and classify injured vs non-injured knees as compared to clinician's assessment across a battery of manually administered tests [53]. To assess joint integrity the device performed three uni-axial laxity assessments, one in each plane [53].

Tests results from the RKT yielded separable measurements for injured versus non-inured knees in the frontal and transverse planes, but not in the sagittal [53]. In the present study no meaningful statistical analyses were done, presumably due to the relatively small sample size (n =4 knees). Other studies have, however, examined the reliability of the RKT [52–54]. In one study, Branch et al. found that the intrarater reliability for the RKT was very strong (ICC = 0.87 – 0.99) for internal – external rotational laxity measures [54].

Despite showing positive reliability results in initial testing, Branch and colleagues have shown little evidence of the device's efficacy in the other relevant planes of motion. In at least one study they specifically mention that the results in the transverse and frontal planes only provided limited information based on the injury that was simulated, while the sagittal plane did not provide meaningful evidence in any scenario [53]. Thus, without additional studies there is insufficient information supporting their device's usefulness outside of the sagittal plane [53].

II.3. Issues in Instrumented Knee Arthrometry

As clinicians and researchers aim to draw generalizable insights from the results obtained through instrumented arthrometry, it becomes apparent that instrumented arthrometry is far from a perfected science. Instrumented arthrometers are plagued by a variety of issues that detract from each device's overall reliability. Most notably, the devices are susceptible to significant operator error, fail to control for excessive motion, and neglect to account for error introduced by soft tissue artefact [6, 44, 88].

II.3.1. Operator Error

Numerous studies have sought to examine aspects that effect operator-dependent arthrometer reliability [43–47, 49, 56, 95–99, 116, 118–120]. To this end, there are key operator-dependent reliability factors highlighted throughout the literature pertaining exclusively to applications related to instrumented arthrometry: experience, test force, handedness, gender and positioning issues [43–47, 95–99, 116, 119, 120].

Forster and Warren-Smith (1989) sought to examine whether the KT – 1000 was a reliable measurement device [120]. Though they did not directly seek to answer whether or not the device was affected by user experience, they did find that inexperienced users produced approximately 64.5% of measures with excessive measurement variability between tests (> Δ 2.00mm between measures) [120]. Ballantyne et al. hypothesized experience would affect device reliability. They found that inexperienced users (ICC = 0.12) are significantly more likely to produce less reliable measurements on the same person than an experienced user (ICC = 0.78) [119]. Berry et al. also demonstrated an experiential effect on reliability. In their study the difference in interrater reliability between novice and experienced users was significant, where novice users demonstrated an ICC = 0.65 and ICC = 0.79 for experienced users [121]. Device reliability between tests and examiners is also influenced by the applied force during testing. Generally speaking, it has been found that larger, more gradually forces applied to the joint yield more accurate diagnostic tests [11, 41, 91]. This trend has been studied most often with the KT devices and in the sagittal plane, where the maximum manual test force has been established as the most reliable diagnostic procedure [41, 42, 95]. While more accurate than lower applied loads, the manual maximum test force neglects key controls that detract from device interrater reliability and generalizability [93].

While not all practitioners utilize this loading schema, it is important that a strict loading protocol be developed to standardize measurement practices.

Gender has also been shown to influence examiner reliability scores. Ballantyne et al. also sought to examine the effects of gender on device reliability. To this end they found that gender had a significant effect on reliability (ICC_{male} = 0.84, ICC_{female} = 0.68), though not as pronounced as the effect of experience [119].

While the literature supporting gender differences in laxity assessment reliability is limited, it is important to note that instrumented arthrometers such as the KT – 1000 and Rolimeter have been shown to be most reliable and most useful for diagnostic purposes, when being performed at maximum manual force levels [41, 87, 119]. Given the average inherent discrepancy between male and female strength, particularly in the sagittal plane, it is apparent that these measurements may not be entirely comparable [55].

Similarly, evidence suggests that, the handedness of the operator affect manually applied laxity measurements [56, 122]. Sernert et al. highlighted that the measured laxity values differed between legs based on the dominant hand of the operator [56]. Though reliability was not reported in this study, the laxity values obtained by each examiner was significantly higher when measuring the knee that corresponded to their handedness. A post-hoc analysis reflected these results in a separate study by Sernert et al., where only a single right-hand-dominant physician was used to obtain anterior laxity measures [122].

Operator-dependent errors are clearly a source of significant error in instrumented arthrometry. Therefore, reduction in outside error is paramount to creating a reliable and more generalizable system for obtaining meaningful laxity measurements. If these factors can be reasonably controlled, accounting for device specific error will become a much easier task, enabling for more precise and reliable measurements.

II.3.2. Positional Control

Reliability of these devices are also negatively impacted by lack of sufficient positional control. A primary point in the literature concerns the degree to which the tibia is rotated during testing [6, 88, 92, 94, 97, 119, 123, 124]. Guskiewicz et al. demonstrated that the degree of tibial rotation has a significant impact on anterior tibial displacement during testing with the KT – 1000 [124]. Specifically, they found that, in an externally rotated position, anterior laxity measured nearly twice that of the same internally rotated limb [124]. Indeed, this trend is supported throughout the literature [9, 23, 94, 124–127]. This phenomenon has been traced anatomically, in part, to the disengagement of the iliotibial tract when the tibia is externally rotated, placing more load-bearing responsibility onto the ACL which results in greater anterior translation under loading [124]. Without a well-prescribed and/or controlled set-up procedure it is difficult to generalize or even compare measurements obtained on the same device. Furthermore, it is more interesting and insightful to gather measurements when tibial rotation is known.

II.3.3. Soft Tissue Artefact

Finally, we examine the an often-overlooked issue in instrumented arthrometry termed soft tissue artefact. Soft tissue artefact is a term to describe measurement error associated with the unwanted movement of soft tissue, such as muscle and dermis, and adipose tissues when trying to measure bony movement [128]. As discussed above previously, the goal in instrumented arthrometry is to provide a precise and accurate measurement of true bony movement under specific loading schemes, so examining the effect of soft tissue artefact on laxity measurement is a key issue that must be addressed.

It was noted above that the KT attaches itself to the shank with a series of two straps: one placed proximally around the gastrocnemius and the other more distally around the calcaneus tendon. During measurement, these straps cause noticeable compression of the gastrocnemius. This was evidenced in a study conducted by Shino et al. which found that a significant portion of measured anterior displacement (μ = 5.3mm) was attributed to soft tissue compression when using an apparatus comparable to the KT devices [128].

Despite little empirical evidence of error associated with soft tissue artefact in instrumented arthrometry, it is a well-recognized yet poorly characterized source of error in other biomechanical fields [50, 128–130]. Furthermore, in many cases the effects of soft tissue artefact are significant [129,130]. Therefore, while it is difficult to say, with certainty, how much error soft tissue artefact imparts on an instrumented arthrometer, this source of error cannot be ignored during device development.

II.4. Summary

It is clear from the vast amount of research surrounding instrumented arthrometric assessment of knee joint laxity that there are many advantages to being able to objectively quantify knee joint integrity, particularly as it relates to assessing injury risk and overall joint health [6-8,14-16]. However, while it is well established that sagittal plane knee joint laxity is an important measure of joint integrity, recent evidence suggests that a single planar measurement does not sufficiently characterize the multidimensionality of the joint and its compositional structures [19-54].

Furthermore, though many novel instrumented arthrometers have attempted to address these issues, they are not without their faults [43–47, 95–99, 116, 119, 120]. In this regard, while many studies have sought to examine issues related to operator error, much less attention has been paid to issues such as positional control and soft tissue artefact [57,94,124-127,129]. That being said, recent devices have attempted to address control of tibial rotation during arthrometric evaluation [6,10,19-24,34,53,54]. This cannot be said, however, for the issue of soft tissue artefact as it relates to instrumented arthrometry.

In fact, to the knowledge of the author, only four studies to date have examined soft tissue artefact and its relation to instrumented arthrometric evaluation [50,51,57,129]. In all but one of these studies, evaluation of soft tissue artefact has been constrained to the shank [50,51,129]. While it is important to consider and reduce measurement error in all facets of instrumented arthrometry, the fact is that measurement error introduced by the thigh has largely been ignored.

Considering the measurement of knee joint laxity is fundamentally intended to be a measure of displacements and rotations of the tibia relative to the femur, it is imperative that the stability of the femur throughout testing is ensured. Doing so will enhance arthrometer precision by minimizing this largely unstudied source of error during testing. Accordingly, the study to follow will aim to minimize, quantify and validate the efficacy of a novel thigh stabilization system during tri-planar instrumented arthrometric evaluation.

CHAPTER III

METHODS

The Methods (e.g. participants, equipment, testing methods) described herein are a part of a larger, ongoing, grant project. The purpose of this grant was to develop and validate a clinically accessible novel tri-planar knee arthrometer that included an optimized thigh stabilization component. Specifically, the first specific aim of the grant was to build and validate a thigh stabilization system such that motion of the thigh during arthrometric evaluation is minimized to remain within acceptable, predefined, limits. Only those data processing and analysis techniques which are pertinent to this thesis will be described.

III.1. Participants

Three human, fresh-frozen cadaveric lower limbs were utilized to validate the thigh stabilization system. The limbs were chosen to maximize variability in physical characteristics while ensuring that joint integrity and musculature wasting -issues are not a concern. A detailed description of the physical characteristics of each limb can be found in Table III.1. Similarly, a description of the specimens cause of death may be found in Table III.2.

Table III.1. Physical Description of Cadavers

Cadaveric Limb Information					
Subject	**Sex**	**Age**	**Mass (kg)**	**BMI**	**Limb**
2	Female	58	72.7	26	Right
3	Male	39	90.9	23	Left
4	Female	36	80.5	30	Left

Table III.2. Cause of Death

Cause of Death	
Subject	**Description**
2	Asphyxia
3	Central Pontine Myelinolysis
4	Asphyxia

III.2. Equipment

III.2.1. Thigh Stabilization System (TSS)

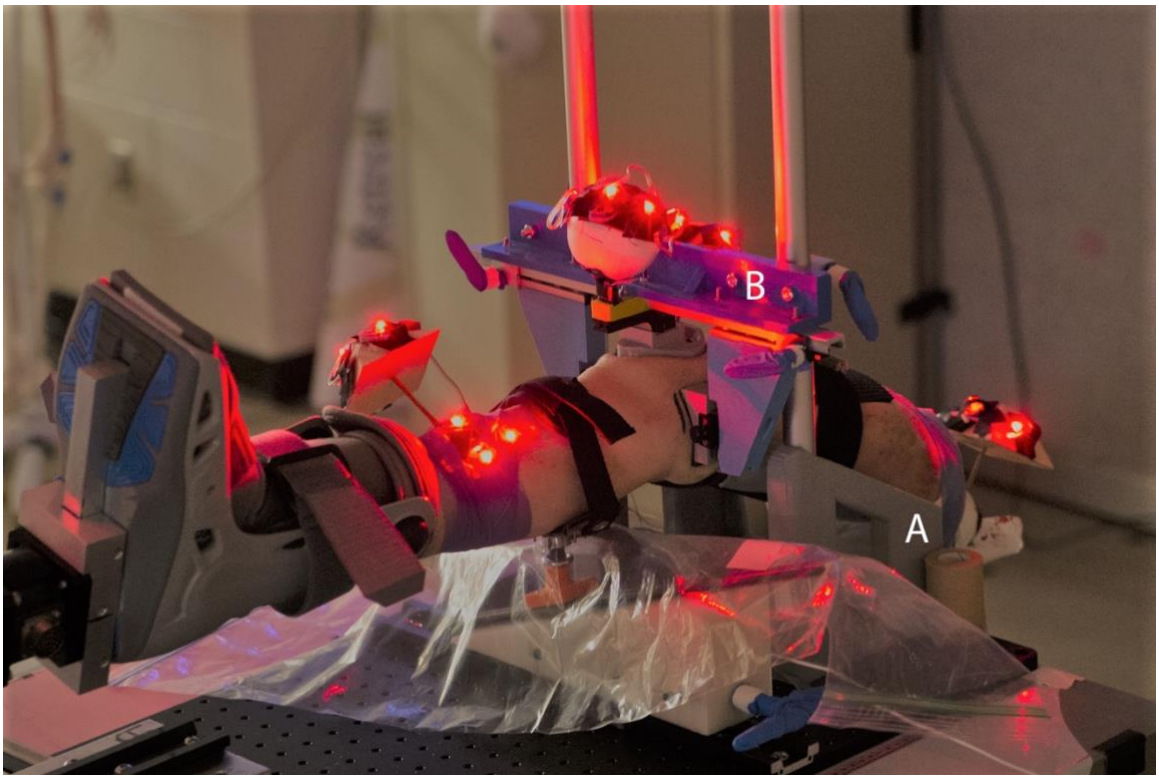

Figure III.1. The Thigh Stabilization System (TSS). The TSS Consists of the (A) Thigh Cradle and (B) U-bar system.

The Thigh Stabilization System (U-bar and Thigh Cradle) is a novel immobilization system designed with two main purposes. Its first purpose is to position and comfortably support the thigh such that the u-bar and measurement systems may be properly interfaced with the leg. Subsequently, the second purpose of the Thigh Stabilization System is to immobilize the thigh to prevent movement of the femur during laxity assessment.

The structure of the Thigh Cradle serves to elevate the subject's thigh such that, the knee joint will be placed in approximately 25-30 degrees of flexion when interfaced with the measurement systems. Fixation, in conjunction with a compressive thigh strap, is accomplished via the U-bar.

The U-bar consists of two, subject-interfacing, tunable and affixable components: the contoured condyle pads, and the patellar pad. The condyle pads are contoured to mimic the curvature of the distal femoral condyle of the knee and, when in use, apply horizontal compressive forces along the medial and lateral condylar region of the distal thigh. The patella pad applies vertical compressive force to the anterior surface of the patella. With the knee joint in approximately 30 degrees of flexion, the compression applied to the patella immobilizes it in the femoral intercondylar groove. Compression is applied by hand by affixing the U-bar's rail clamps to the rails inserted into the distal end of the thigh cradle. Together, the patellar, bilateral condyle pads and distal thigh strap will stabilize the femur in the thigh cradle, preventing movement of the femur during testing in all three anatomic planes of motion.

III.2.2. Shank Stabilization System (SSS)

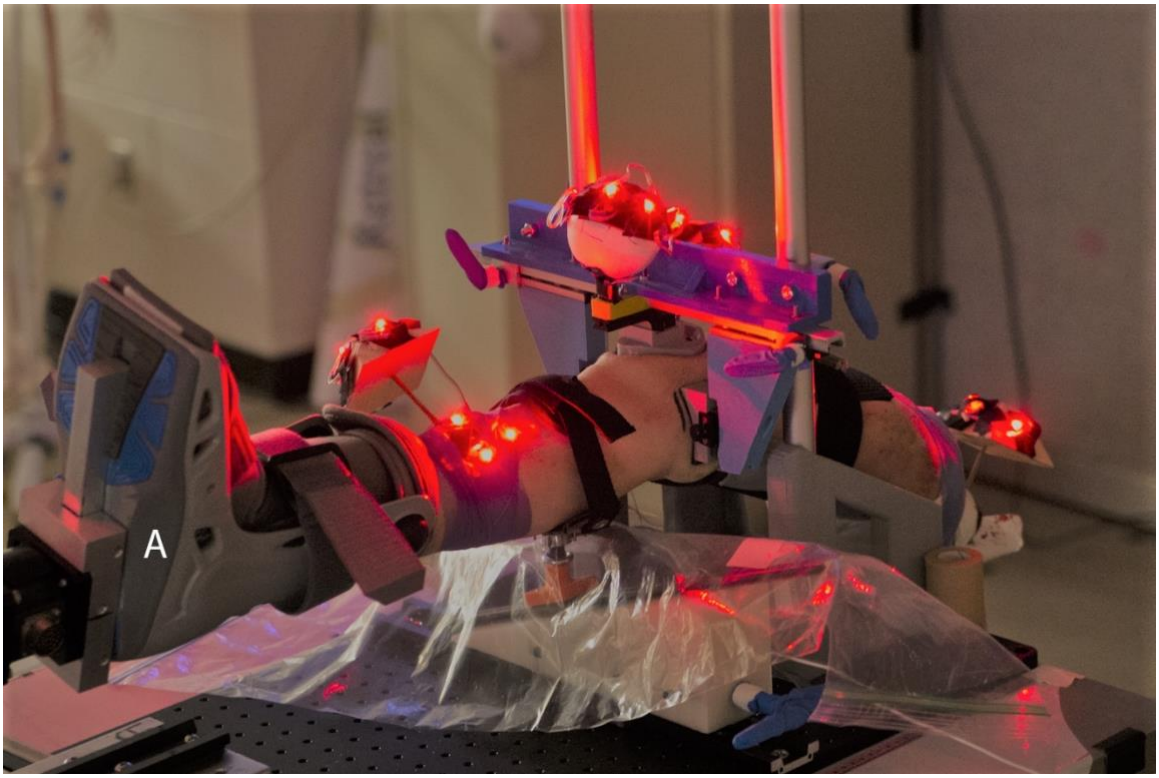

Figure III.2. The Shank Stabilization System (SSS). The Heel Cradle (A) Interfaces Directly with the VV and IE Loading Systems.

The shank stabilization system consists only of a heel cradle (A), which is designed to immobilize the medial and lateral malleoli of the ankle joint while keeping the joint at approximately 90 degrees of flexion. The heel cradle, fashioned from a walking boot, uses two hook-and-loop straps and an inflatable air-bladder to securely fasten the subject's distal shank and foot to the Varus – Valgus and Internal – External loading systems.

III.2.3. Anterior – Posterior (AP) Loading System

The Anterior – Posterior Loading System manipulates the AP cradle by translating horizontal into vertical motion via a translational CAM mechanism. A single, low-friction, linear guide coupled to a wedge CAM creates both positive and negative vertical translations of the AP cradle as the wedge CAM slides perpendicularly to the subject's shank. Load monitoring for the AP Loading System is recorded on an in-line tension-compression load cell attached to the AP cradle.

III.2.4. Internal – External (IE) Rotational Loading System

The Internal – External Rotational Loading System lies in series with the heel cradle and consists of a lockable rotary shaft which is affixed to the heel cradle via a six-degrees-of-freedom load cell and mounting plate. During Internal – External arthrometric evaluation the operator must first unlock the rotary shaft. Once the rotary shaft is unlocked the operator applies torque directly to the heel cradle via the rotary shaft Torque applied to the shaft is recorded on the load cell.

III.2.5. Varus – Valgus (VV) Loading System

The Varus – Valgus Loading System consists of two perpendicular, low-friction, non-motorized linear tracks which allow for radial deflection of the shank as force is applied to the heel cradle. The operator manipulates the VV measurement system via a s-beam tension-compression load cell, which is attached to the malleolar region of the heel cradle.

III.2.6. Measurement System(s)

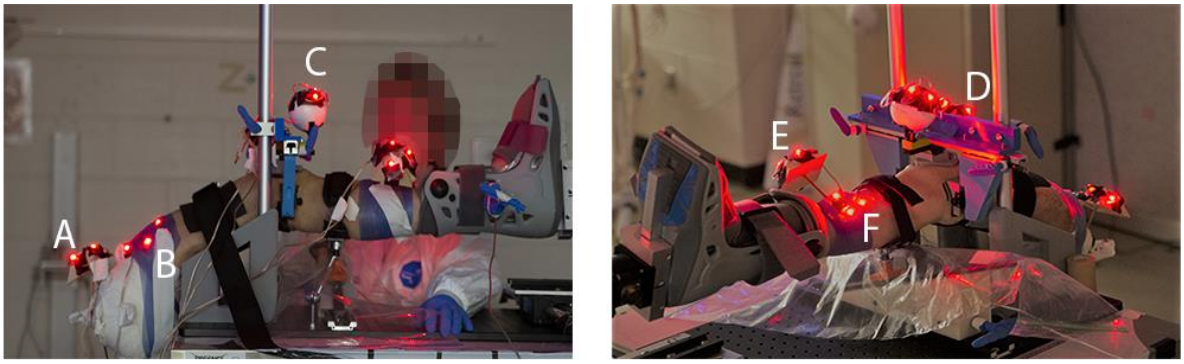

Figure III.3. Optical Marker Placement. The Optical Marker System Allows for Measurement of Limb and Device Displacements/Rotations

Prior to testing the MotionMonitor (Innovative Sports Training, Inc., Chicago, IL) system environment will have been initialized according to the manufacturers initialization instructions. Optical sensors paired to the MotionMonitor system will be used to assess gross movement between the limb and the device. As Seen in Figure III.7 optical sensors will be placed on the femoral head (A), thigh surface (B), thigh cradle (C), patella (D), tibial plateau (E), and the shank surface (F). Kinematic data will be recorded via the Motion Monitor system at 240Hz.

III.3. Procedures

Data collection occurred over a 48-hour period in which each limb (n=3) underwent manual arthrometric evaluation in each of the three anatomic planes of motion (Frontal, Sagittal and Transverse).

III.3.1. Device Set-up

Prior to situation within the device the operator marked the location of the subject's medial and lateral joint line. These marks guided the placement of the subject within the thigh stabilization system in order to avoid obstructing the joint line. Once the condyles had been marked the subject's limb was placed within the device, allowing their thigh and shank to be supported by their respective stabilization systems.

The subject's limb was first situated and fastened into the Shank Stabilization System, ensuring that the sole foot was flat against the insole of the Heel Cradle. Once the hook-and-loop straps were fastened and the air bladder filled, the operator ensured that shank was parallel with the device's base.

If adjustments to the angle needed to be made this was be done using the angular adjustment clamps of the measurement linkage system. Confirming that the heel cradle is unlocked and able to freely move, the limb was then aligned with the condylar clamps.

To begin situating the limb in within the Thigh Cradle the operator began by aligning the subject's lateral femoral condyle with the distal edge of the corresponding condyle pad. Once the condyle pads had been aligned with the subject's condyles, the patella pad was positioned over the subject's patella. At this point the U-bar was then interfaced and fixated to the subject for testing. This was done first by compressing the U-bar and affixing its clamps to the rails. Following fixation of the U-bar, the condylar clamps were affixed to the subject by manually compressing and locking the condylar clamps first on the lateral side and then on the medial side. Finally, once the U-bar was fixated in the proper position, the thigh strap was placed around the thigh through the cut outs in the cradle.

Because the limb was disarticulated from the hip, the proximal head of the femur was loosely affixed to the testing surface. Stabilization of the femoral head was achieved by fastening two nylon cords around the head of the femur. These cords were then tensioned to bolts on the testing surface. This method of fixation was thought to provide a moderate amount of downward and medio-lateral fixation force to the femur without completely immobilizing it.

With the limb stabilized, the optical markers for testing were then affixed to the limb. Bony markers were affixed to the limb using screws embedded into the bone, while surface markers were affixed to the limb using PreWrap ® tape.

Manual testing was completed by a single, experienced, clinician.

III.3.2. . Anterior – Posterior Testing

Before the first trial of 3 repetitions begins, the operator first exerted three conditioning loading cycles, a standard practice for manual and instrumented laxity evaluation. Following the last conditioning cycle, three manual cycles at 130N Anterior and 90N Posterior were administered.

III.3.3. Varus – Valgus Testing

Prior to frontal plane testing, the AP cradle was unfastened and removed such that the shank remains unimpeded. Similar to the testing technique described for the sagittal plane, the operator first applied three conditioning torques prior to testing. Following the conditioning cycles, ten tests of 10Nm varus – valgus loading will be applied to each limb.

III.3.4. Internal – External Testing

Likewise, following the conditioning trials ten test of 5Nm torques were applied in the transverse plane through the long axis of the tibia. These loading schemas reflect those described in previous studies [23, 52–54, 93, 131].

III.3.5. Data Analysis

Kinematic data was collected during the arthrometric evaluation cycles using the MotionMonitor System. Displacement and rotational output data from each marker were processed using the software provided through MotionMonitor. Loading data was recorded with the previously described device-embedded load cells and processed using the software provided via MotionMonitor. Prior to export, the kinematic and loading data was processed using a fourth-order, 6Hz low pass zero-lag Butterworth filter. Any dropped kinematic data was reconstituted using a cubic spline interpolation function; however, every attempt was made to redo trials in which a significant amount of data was lost.

Outlier detection and removal was performed on subject-subject basis. First, a Z-score analysis was performed on each data column (25 per test) to detect the location of data points that were greater than 3.0 standard deviations away from the mean. In the case where a point was detected, all data points from that corresponding index (row) were removed from all following analyses.

As previously stated, the purpose of this thesis was to validate the usefulness and efficacy of a novel thigh stabilization system in evoking more generalizable arthrometric measurements by effectively immobilizing the thigh during arthrometric evaluation(s). Therefore, to evaluate this hypothesis, only sensors on the Thigh Stabilization System (U-bar) and femur will be analyzed.

To this end, the absolute error and random error was established by comparing the observed rotations and translations of the ground truth, control, sensor (thigh cradle) to the experimental sensor (femoral head) in each plane.

Each trial, which consists of three full (e.g. Anterior – Posterior) loading cycles compared discrete displacements and rotations at specific loading points. In the sagittal plane, data to be considered for analysis was pulled at 130N anterior and 90N posterior, while in the transverse and frontal planes data was pulled at ±5Nm for Internal – External Rotation (Transverse) and ±10Nm for Varus – Valgus (Frontal) Rotation. Specifically, During AP arthrometric testing, only displacement data from the anterior – posterior axis was considered for analysis. During varus – valgus and internal – external laxity testing, LOA analyses were formed by analyzing frontal and transverse plane rotational data. In the case where multiple data points lie close to the threshold loading level, only the first to cross the loading threshold was considered. The 95% limits of agreement (LOAs) were constructed by creating confidence intervals (± 1.96 multiplied by the standard deviation (SD) of the individual difference scores) around the mean differences of segments.

Bland-Altman plots were created both individually for each subject and collectively to show agreement between: anterior – posterior displacement of the femur- relative to the device during Anterior – Posterior testing; frontal and transverse plane rotation of the femoral-head relative to the device for Varus – Valgus testing; and frontal and transverse plane rotation of the femur relative to the device for Internal – External testing; Furthermore, plots to qualitatively assess the differences between detected device and femoral-head sensor motions were constructed for all of the aforementioned scenarios.

Successful completion of the clinical thigh stabilization prototype is indicated by constraining motion of the bony femur relative to the device within 0.5±0.5mm for AP displacements, 1.0±0.5^{o}for frontal plane rotations, and 2.0±1.0^{o} for transverse plane rotations.

All processing and analyses were completed using MATLAB (The MathWorks, Inc., Natick, Massachusetts, United States) and the Python computing language (Python Software Foundation. Python Language Reference, version 3.7. Available at http://www.python.org).

CHAPTER IV

RESULTS

IV.1. Anterior – Posterior Testing

Figure IV.1 (top) depict transverse plane (Anterior – Posterior) displacements of the bony femur and device sensors as captured at peak anterior (130N) loading. Similarly, Figure IV.1 (bottom) depicts transverse plane displacements of the bony femur and device sensors for all maximum posterior (90N) loads.

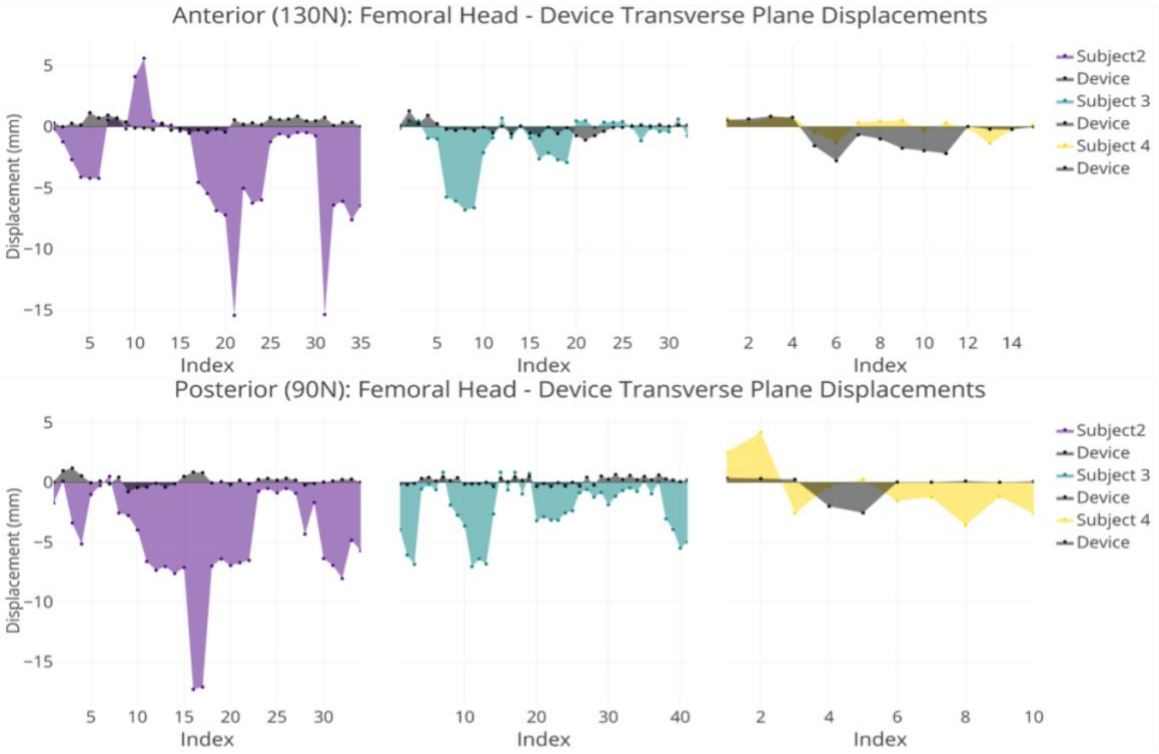

Figure IV.1. Anterior (130N) and Posterior (-90N) Displacement Comparisons: Femoral Head-Device.

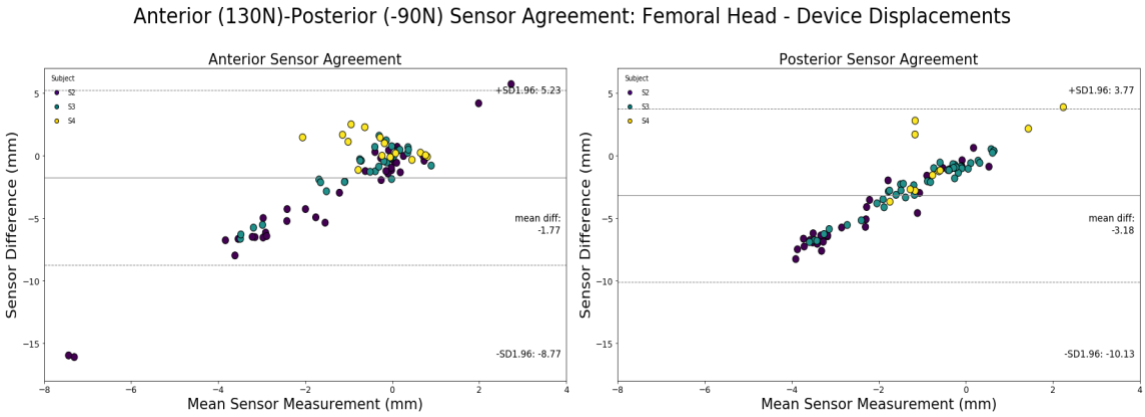

Figure IV.2. Anterior (130N) and Posterior (-90N) Sensor Agreement: Femoral Head-Device Displacement.

Bland-Altman comparisons (Figure IV.2) show that, for anterior displacements, the average systematic difference between the displacement as measured from the sensor embedded in the femoral head compared to the device embedded sensor was -1.77± 7.00, 95% LOA [-8.77,5.23] mm. In the posterior direction, Bland-Altman, comparisons showed an agreement of -3.18± 6.95, 95% LOA [-10.13,3.77] mm.

On a subject-level (Figure IV.3) it was found that Subject 2 demonstrated a mean sensor agreement of -3.36± 8.75, 95% LOA [-12.11,5.39] mm in the anterior direction and a -5.00± 8.25, 95% LOA [-13.25, 3.24] mm in the posterior direction. For Subject 3 we found a mean sensor agreement of -1.18± 4.16, 95% LOA [-5.34,2.99] mm in the anterior direction and a -2.38± 4.10, 95% LOA [-6.48,1.73] mm in the posterior direction. Meanwhile, for Subject 4 we found a mean sensor agreement of 0.68± 1.98, 95% LOA [-1.30,2.66] mm in the anterior direction and a -0.27± 4.99, 95% LOA [-5.22,4.68] mm in the posterior direction.

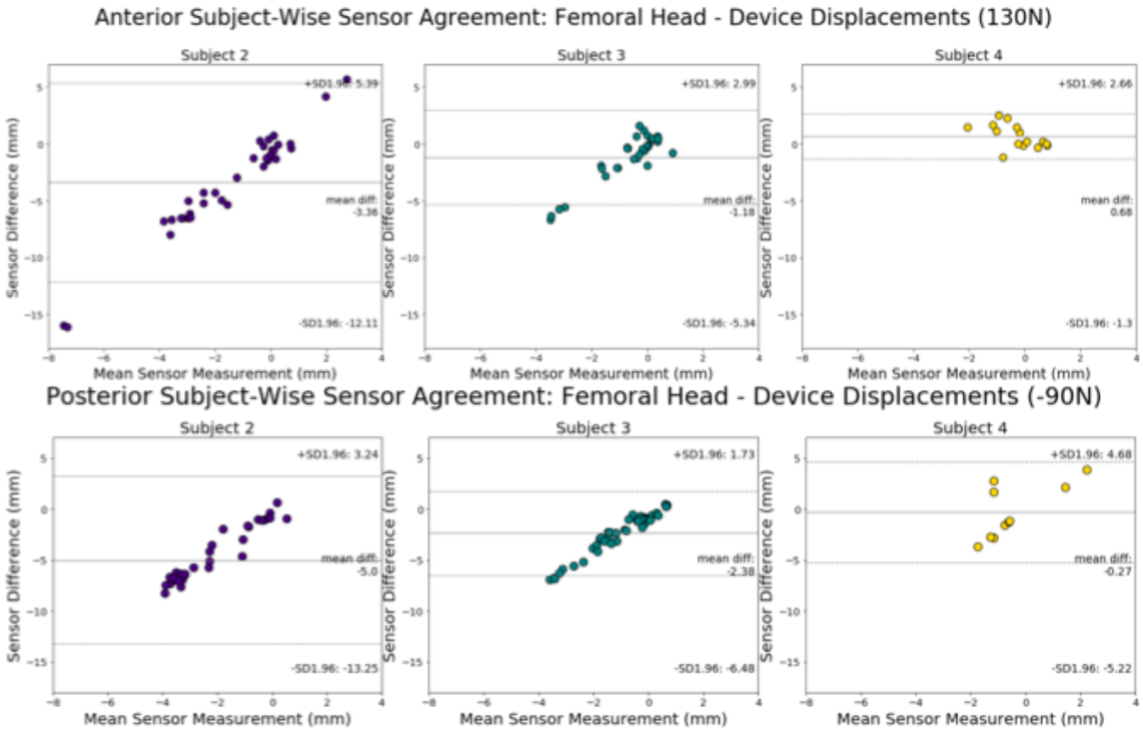

Figure IV.3. Anterior (130N) and Posterior (-90N) Subject-Wise Sensor Agreement: Femoral Head-Device Displacement.

IV.2. Varus – Valgus Rotational Testing

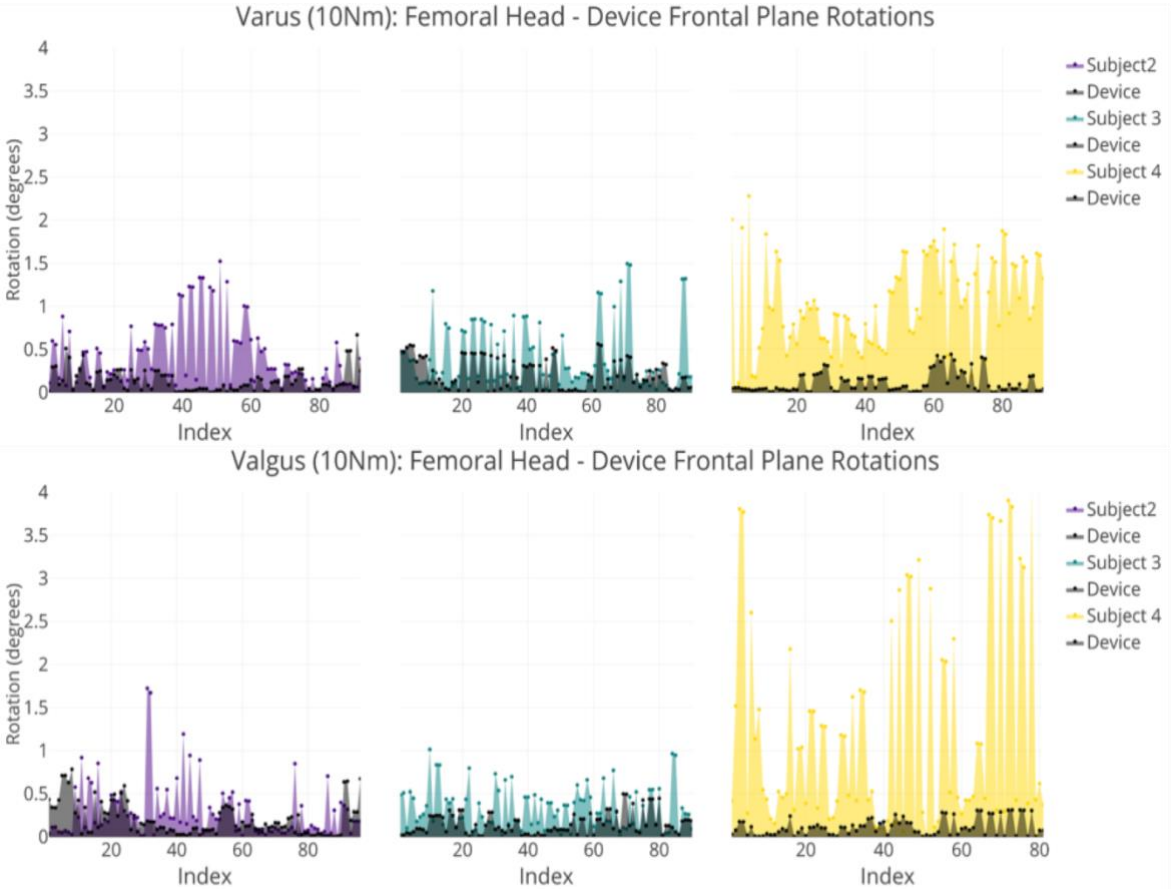

Figure IV.4. Varus – Valgus (10Nm) Frontal Plane Rotational Comparisons: Femoral Head-Device.

Figure IV.4 (top) shows frontal plane rotations for all varus (10Nm) rotations of the bony femur sensor, compared to the device. Similarly, in Figure IV.4 (bottom) valgus rotations of the bony femur sensor as compared to the device within the frontal plane. Figure IV.5, like Figure IV.4, shows femoral head rotations compared to device-related rotations in transverse planes for varus and valgus (10Nm) moments.

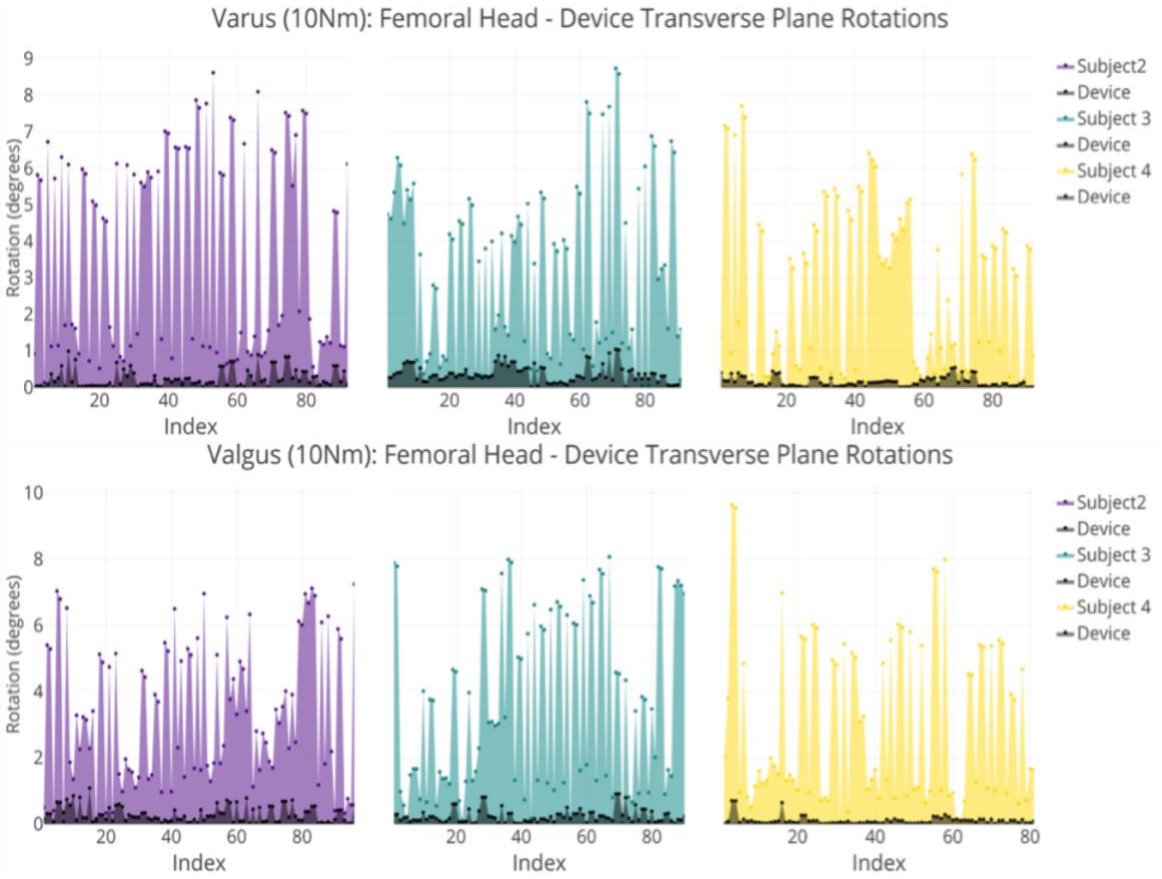

Figure IV.5. Varus – Valgus (10Nm) Transverse Plane Rotational Comparisons: Femoral Head-Device.

Bland-Altman comparisons (Figure IV.6 and Figure IV.7) show that, for varus rotations, the average systematic rotational difference as measured between femoral head and the device embedded sensors were 1.37± 1.07, 95% LOA [-0.60,1.54] degrees, in the frontal plane. In the valgus direction, Bland-Altman, comparisons show an agreement of 0.44± 1.65, 95% LOA [-1.21,2.09] degrees for frontal plane rotations. Transverse plane rotations show a mean difference of 3.04 ± 4.74, 95% LOA [-1.71,7.78] degrees in the varus direction, while in the valgus direction a mean difference of 3.13 ± 4.65, 95% LOA [-1.52,7.78] degrees was found.

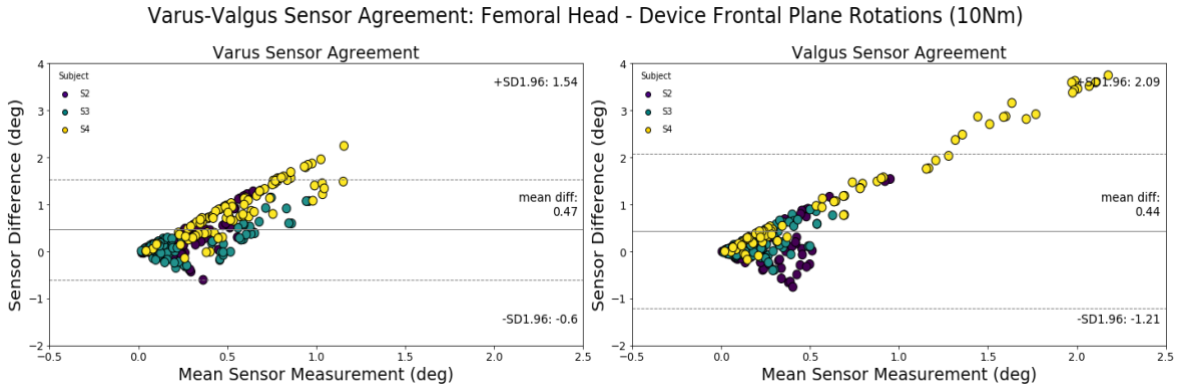

Figure IV.6. Varus – Valgus (10Nm) Sensor Agreement: Frontal Plane Femoral Head-Device Rotations.

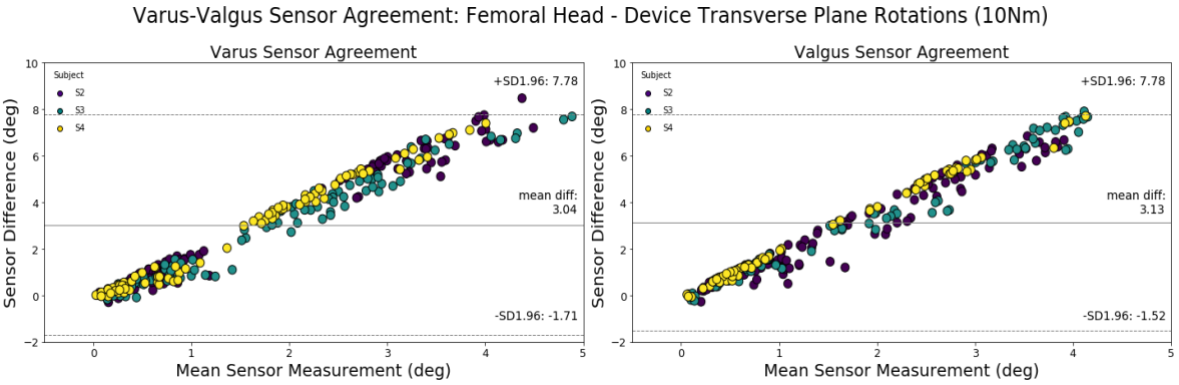

Figure IV.7. Varus – Valgus (10Nm) Sensor Agreement: Transverse Plane Femoral Head-Device Rotations.

Looking subject-wise about for frontal plane rotations (Figure IV.8 it was found that Subject 2 demonstrated a mean sensor agreement of 0.29± 0.82, 95% LOA [-0.53,1.11] degrees in the varus direction and a 0.09±0.78, 95% LOA [-0.70,.87] degrees in the valgus direction. Subject 3 produced a mean sensor agreement of 0.18± 0.65, 95% LOA [-0.47,0.83] degrees in the varus direction and a 0.19± 0.54, 95% LOA [-0.36,0.73] degrees in the valgus direction.

Similarly, For Subject 4 Bland-Altman analysis determined there was a mean sensor agreement of 0.92± 1.06, 95% LOA [-0.14,1.98] degrees in the varus direction and a 1.13± 2.28, 95% LOA [-1.15,3.41] degrees in the valgus direction.

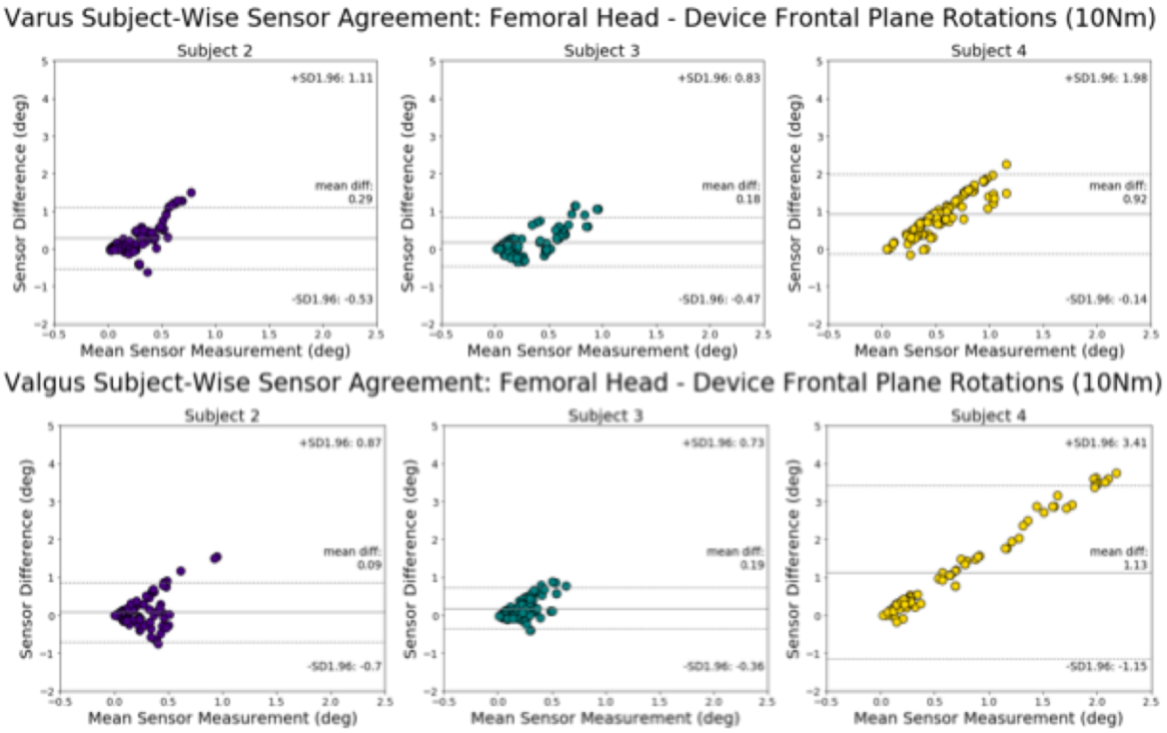

Figure IV.8. Varus – Valgus (10Nm) Subject-Wise Sensor Agreement: Frontal Plane Femoral Head-Device Rotations.

For transverse plane rotations (Figure IV.9) it was found that Subject 2 demonstrated a mean sensor agreement of 3.53± 5.21, 95%LOA [-1.67,8.74] degrees in the varus direction and a 3.08± 4.08, 95% LOA [-1.00,7.16] degrees in the valgus direction. Subject 3 produced a mean sensor agreement of 2.91± 4.38 95% LOA [-1.47,7.29] degrees in the varus direction and a 3.4± 5.09, 95% LOA [-1.69,8.49] degrees in the valgus direction.

Similarly, For Subject 4 we found a mean sensor agreement of 2.66± 4.43, 95% LOA [-1.78,7.09] degrees in the varus direction and a 2.89± 4.71, 95% LOA [-1.81,7.60] degrees in the valgus direction.

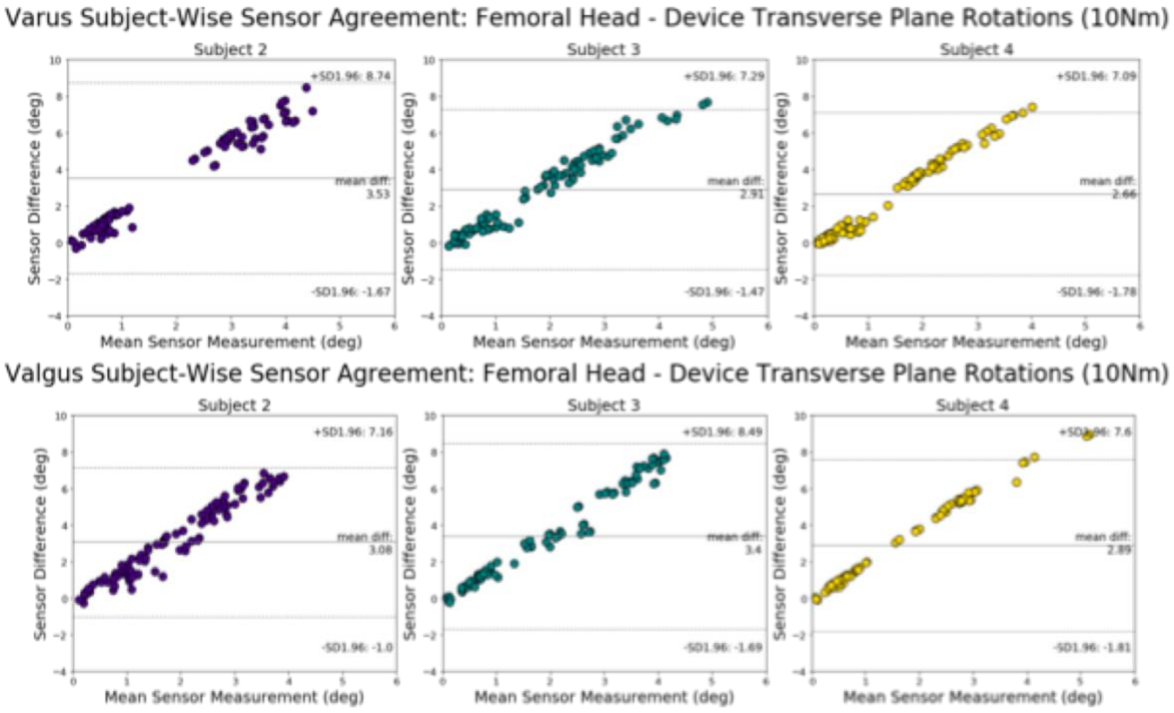

Figure IV.9. Varus – Valgus (10Nm) Subject-Wise Sensor Agreement: Transverse Plane Femoral Head-Device Rotations.

IV.3. Internal – External Rotational Testing

Figure IV.10 (top) shows frontal plane rotations for all internal (5Nm) rotations of the bony femur sensor, compared to the device. Similarly, Figure IV.10 (bottom) shows external rotations of the bony femur sensor as compared to the device within the transverse plane. Figure IV.11, like Figure IV.10, shows femoral head rotations compared to device-related rotations about in the transverse planes for internal and external (5Nm) moments.

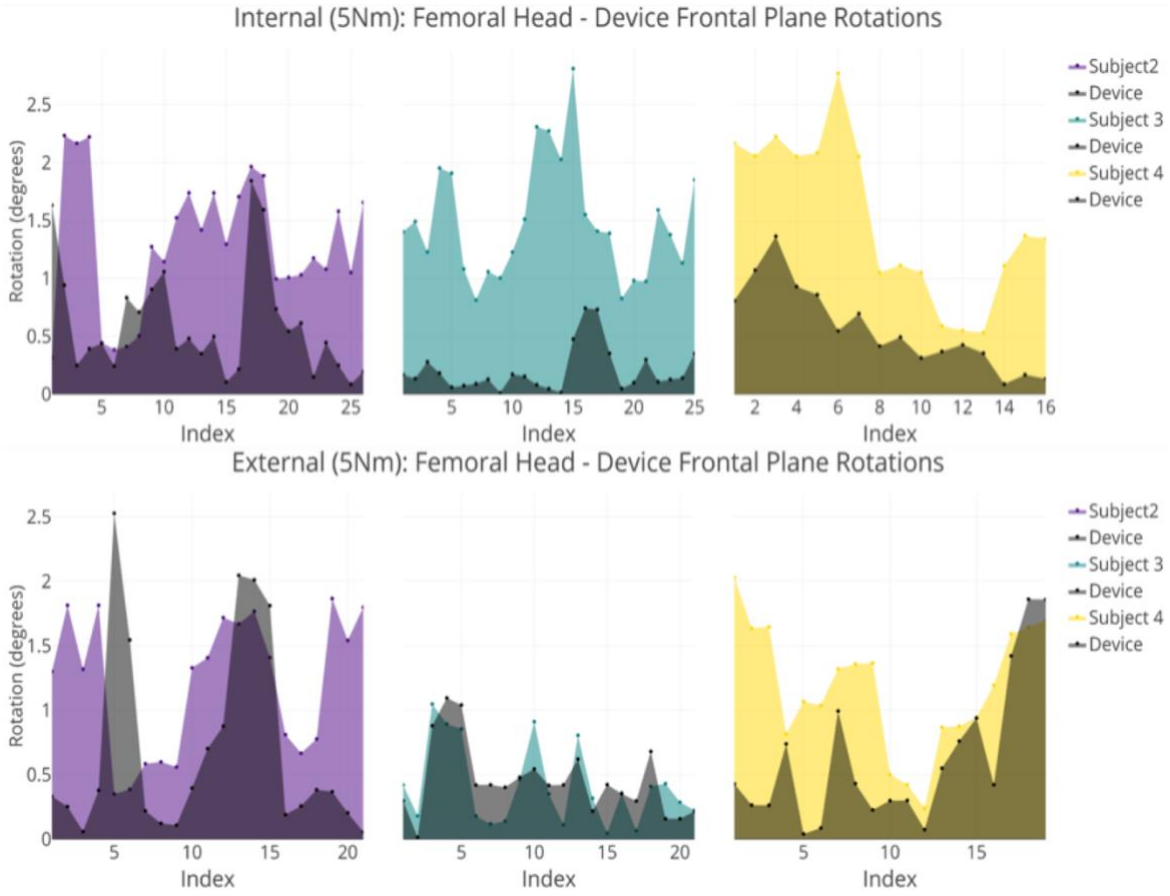

Figure IV.10. Internal – External(5Nm) Frontal Plane Rotational Comparisons: Femoral Head-Device.

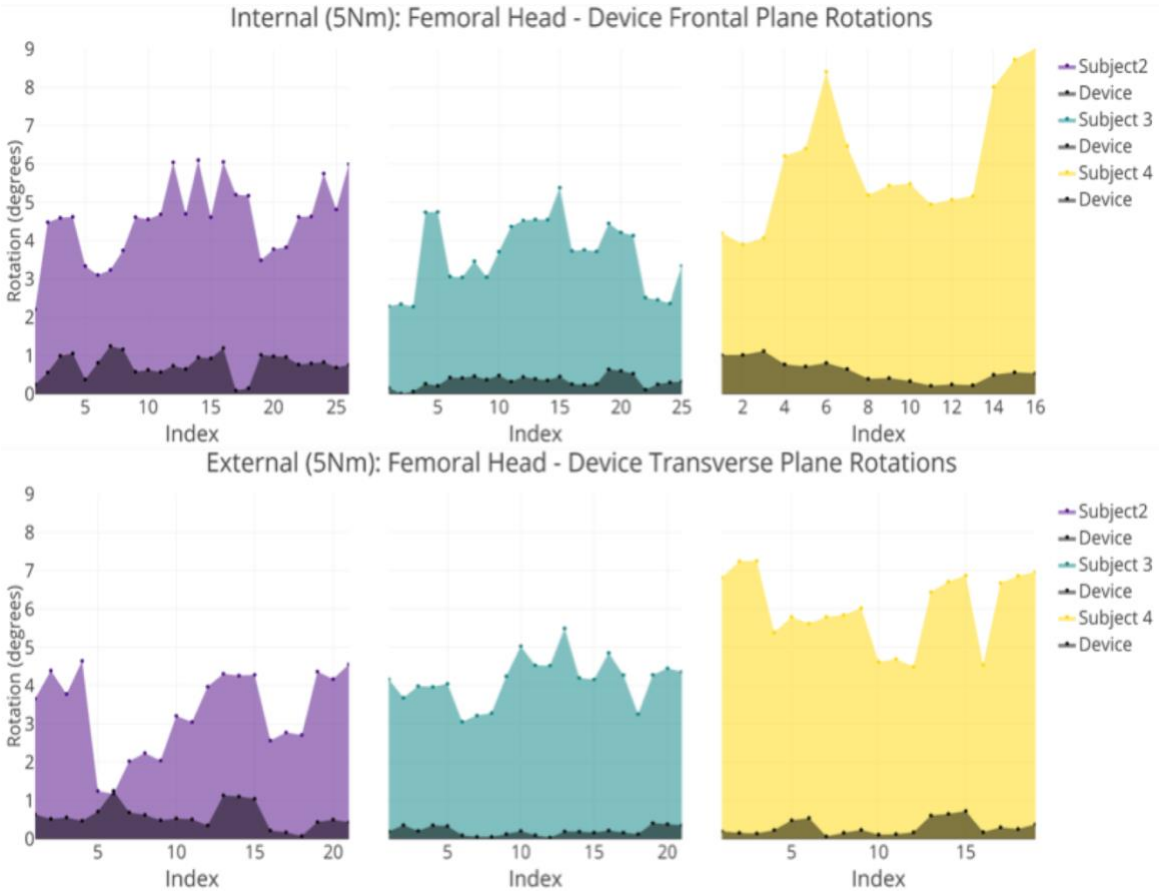

Figure IV.11. Internal – External (5Nm) Transverse Plane Rotational Comparisons: Femoral Head-Device.

Bland-Altman comparisons (Figure IV.12 and Figure IV.13) show that, for internal rotations, the average systematic rotational difference as measured between femoral head and the device embedded sensors are 0.97± 1.3, 95% LOA [-0.32,2.27] degrees, about for frontal plane rotations. In the external direction, Bland-Altman, comparisons show an agreement of 0.33± 1.37, 95% LOA [-1.04,1.70] degrees about in the frontal plane. Transverse plane rotations show a mean difference of 4.00 ± 2.80, 95% LOA [1.20,6.80] degrees in the internal direction, while in the external direction a mean difference of 4.07 ± 2.97, 95% LOA [1.11,7.04] degrees was found.

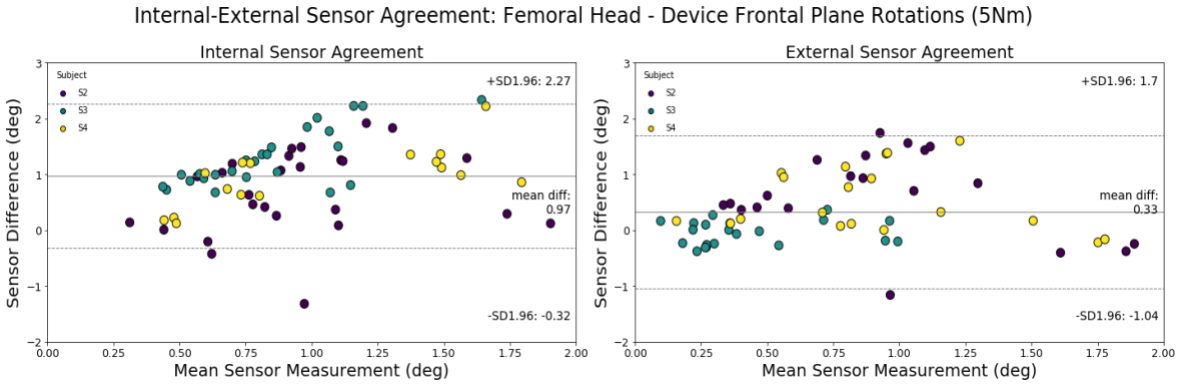

Figure IV.12. Internal – External (5Nm) Sensor Agreement: Frontal Plane Femoral Head-Device Rotations.

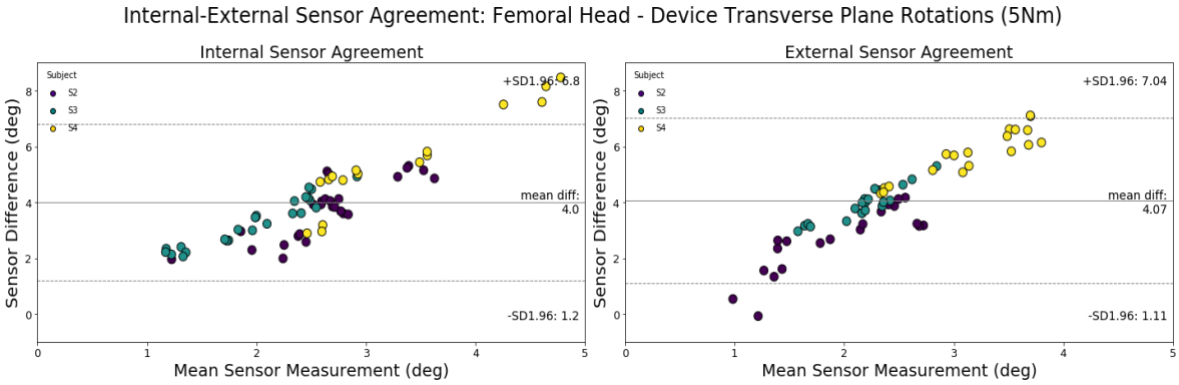

Figure IV.13. Internal – External (5Nm) Sensor Agreement: Transverse Plane Femoral Head-Device Rotations.

Looking subject-wise in the frontal plane (Figure IV.14) it was found that Subject 2 demonstrated a mean sensor agreement of 0.69± 1.46, 95% LOA [-0.76,2.15] degrees in the internal direction and a 0.51±1.84, 95% LOA [-1.33,2.35] degrees in the external direction. Subject 3 produced a mean sensor agreement of 1.29± 0.98, 95% LOA [0.30,2.27] degrees in the internal direction and a -0.04± 0.42, 95% LOA [-0.46,0.37] degrees in the external direction.

Similarly, For Subject 4 we found a mean sensor agreement of 0.94± 1.02, 95% LOA [-0.07,1.96] degrees in the internal direction and a 0.54± 1.10, 95% LOA [-0.55,1.64] degrees in the external direction.

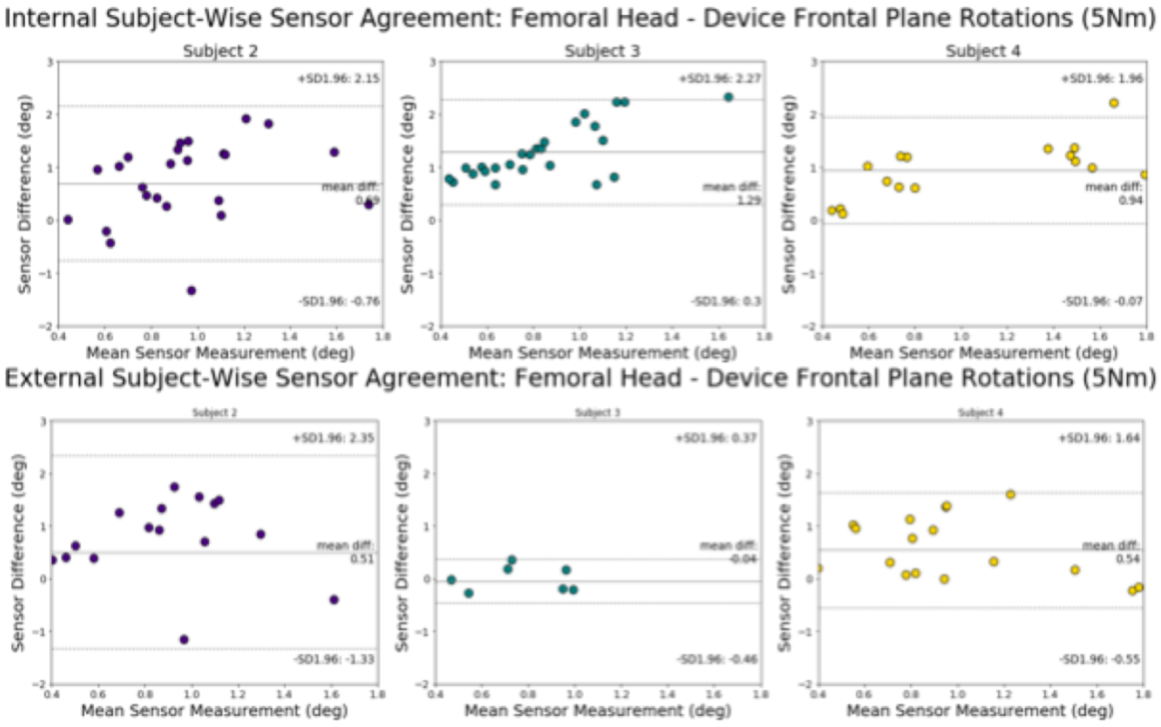

Figure IV.14. Internal – External (5Nm) Subject-Wise Sensor Agreement: Frontal Plane Femoral Head-Device Rotations.

For transverse plane rotations (Figure IV.15) it was found that Subject 2 demonstrated a mean sensor agreement of 3.78± 1.99, 95% LOA [1.78,5.77] degrees in the internal direction and a 2.71± 2.22, 95% LOA [0.50,4.93] degrees in the external direction. Subject 3 produced a mean sensor agreement of 3.30± 1.63 95% LOA [1.67,4.93] degrees in the internal direction and a 3.94± 1.13, 95% LOA [2.81,5.07] degrees in the external direction.

Similarly, For Subject 4 we see a mean sensor agreement of 5.45± 3.29, 95% LOA [2.15,8.74] degrees in the internal direction and a 5.73± 1.71, 95% LOA [4.03,7.44] degrees in the external direction.

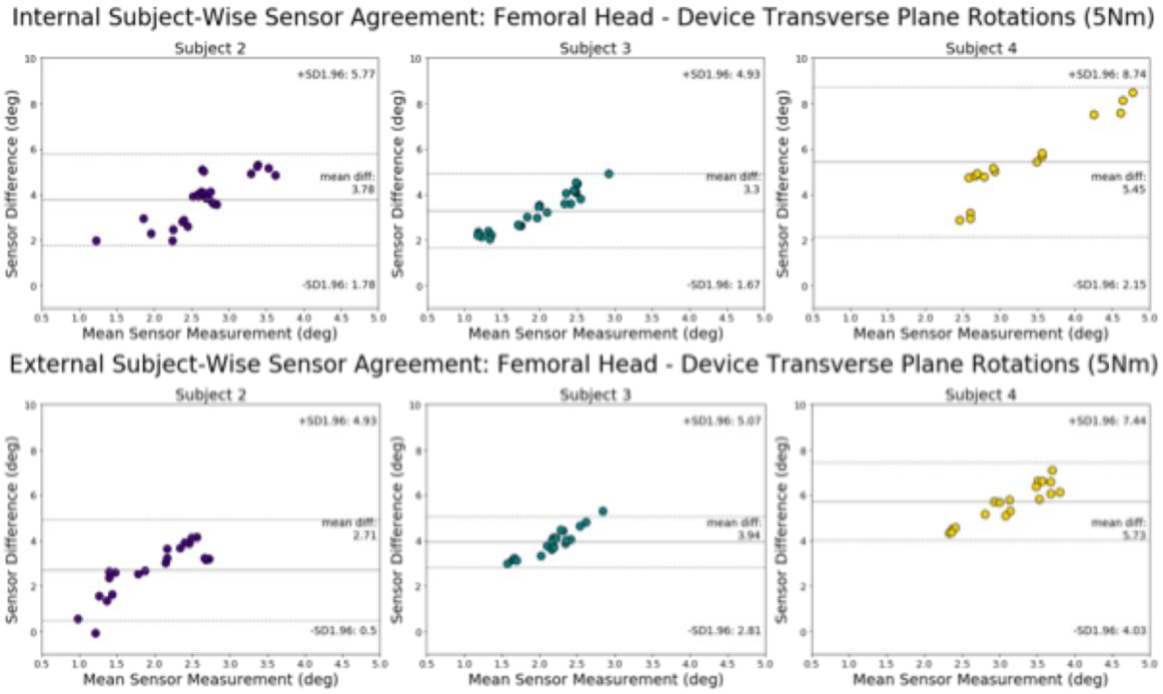

Figure IV.15. Internal – External (5Nm) Subject-Wise Sensor Agreement: Transverse Plane Femoral Head-Device Rotations.

CHAPTER V

DISCUSSION

This study presents a novel method for stabilizing the femur as a basis for minimizing measurement error during multiplanar arthrometric (laxity) evaluation. As summarized in Table IV.1, the primary results of this study, as presented in the previous section, suggest that rotation of the femur was most significantly constrained about for frontal plane rotations during both Varus – Valgus and Internal – External Rotation testing.

The device, in its current form, did not reject the null hypotheses in any given plane. However, this study has elucidated key limitations about the current device that must be addressed as well as model limitations that may have adversely affected the results of this study. Ultimately, the insights about the study discussed herein will inform our device and study design moving forward. The discussion to follow will begin by addressing the agreement of femoral to device motion within each plane and will then explore the implications of these findings as bounded by the limitations and future directions of the project.

V.1. Anterior – Posterior Testing

Testing in the Anterior – Posterior plane yielded results that fell outside of the previously defined *a*-priori limits. Overall, it was determined that femoral motion during anterior testing differed relative to the device by -1.77±7.00mm, while for posterior testing, the difference was found to be -3.18±6.95mm.

With reference to the device, anterior – posterior translations of the femur during AP arthrometric evaluation were constrained anteriorly by the U-bar fixation system and posteriorly by the Thigh Cradle. When considering the LOAs on a subject- by subject level, it is apparent that Subject 2 and to a lesser extent, Subject 3 had an inordinate effect on the error calculations with clustered difference scores ranging reading nearly 8-20mm posterior differences. The magnitude of these differences can be appreciated in Figure IV.1, where the device and bony sensor readings can be visually paired.

Examining the subject's as a whole, there appears to be two distinct clusters both anteriorly and posteriorly. Within these two clusters, the points of greatest disagreement occur mostly within Subject 2. While it is difficult to envision how a 8-20mm disagreement could occur when bounded by the thigh cradle and u-bar, it is plain to see that the majority of these disagreements occur posteriorly. It is possible that this phenomenon is occurring due to soft tissue artefact of the thigh, presumably related to the compression of the posterior musculature and connective tissues. In the cadaver model, the lack of muscular tone and muscle wasting would seemingly exaggerate this source of error when compared to human limbs. In the case of Subject 2, their age and relative lack of muscle tone may have contributed to this disparity. Moreover, Subject 3 had considerable amounts of muscle wasting, possibly due to their pre-existing condition (Table III.2).

Furthermore, the posterior displacements seen during AP testing may be in response to the disarticulation of the hip. Without the support of the joint and its surrounding tissues, the U-bar may essentially act as a fulcrum. Given the rigid nature of the U-bar, it is reasonable to assume that only anteriorly directed levering could occur at the femoral head.

Considering displacements measured by the sensor embedded in the femoral head are modeled about the distal end of the femur, it is easy to see how an 8-20mm posterior displacement of the femur could occur. However, while it is relatively clear that these issues with the cadaver models are not a concern for true human testing, it is difficult to estimate their effects on the sensor agreement.

AP testing was also affected by a few key limitations, primarily related to the components of the device. While the u-bar assembly is undoubtedly a novel aspect of this arthrometer, it is still a prototype in development. In order to speed up this development, the choice was made to utilize some readily available components on the u-bar, for this initial round of testing. Specifically, the affixable components on the U-bar utilized friction over more desirable custom mechanical solutions. Because these pieces utilized metal-metal clamping friction, it was inevitable that the components would experience fatigue. To this end, the frequency and force required to repeatedly clamp the U-bar and condylar pads resulted in fatigue to both the clinician and the components of the device. In fact, one condylar-pad braking component failed during testing, necessitating a backup component to be installed. As a result, there were limited instances, particularly for Subject 2, where these components slipped during testing, potentially resulting in the errant sensor disagreements during AP testing.

V.2. Varus – Valgus Rotational Testing

Results obtained from the Varus – Valgus portion of testing primarily fell outside of the previously defined *a*-priori limits, however in some cases systematic errors fell within the pre-defined *a*-priori limits. Overall, it was determined that femoral motion during varus rotational testing differed relative to the device by -0.47±1.07 degrees in the frontal plane and 3.04±4.74 degrees in the transverse plane.

Similarly, femoral differences for valgus rotations were shown to be 0.44±1.60 degrees for frontal plane rotations and 3.13±4.65 degrees for transverse plane rotations. For Varus – Valgus arthrometric evaluation, the results described for frontal plane stabilization are highly encouraging primarily because it demonstrates that the thigh stabilization system was able to constrain femoral rotation in the frontal plane near the pre-defined *a*-priori limits. In this case, frontal plane stabilization is the main metric of success as it directly correlates to the test being administered.

Subject-subject results show remarkably similar error distributions. In all cases, for frontal plane rotations, the difference between the bony and device sensors fails to cross 2.5^o disagreement for varus and 4.5^o for valgus loadings. Given our *a priori* clinically relevant LOAs, the systematic error for frontal plane rotations falls well within the $\pm 1^o$ of allowable error. However, the same cannot be said for the extent of the random error in the system. Despite coming close to these marks, there is still approximately 0.6^o -1.1^o of excessive error for Varus and Valgus measurements, respectively.

Transverse plane results indicate that there is a marked increase in femoral rotations as compared to the frontal plane rotations. This was a surprising finding given the load was applied in the frontal plane. Several reasons may exist for this transverse plane rotation during VV arthrometric testing. It is possible that the positioning of the U-bar system may have created an unbalanced fulcrum point that led to increased limb rotations in the transverse plane. Because the condylar clamps are parallel and collinear with one another it is possible that the compressive forces of the pads were not equally distributed about the subject's medial and lateral femoral condyles.

This imbalance could feasibly predispose the limb to rotate or dislodge from its pre-testing position about the femur's long axis in the transverse plane as is moves to a position of greater stability. Furthermore, because the femoral head was disarticulated from the hip and fixation of the femur provided little transverse plane stabilization, the femur was able to move most freely about it's long axis. Given these considerations and despite the results lying outside the *a priori* limits by approximately$1.1^o \pm 3.75^o$, they are still encouraging.

It is difficult to predict to what extent these issues had on the performance of the thigh stabilization system and the arthrometer in general, but it is the belief of the author and clinicians that the limb model used during testing does not adequately represent the kinematics present in the human acetabulofemoral joint.

V.3. Internal – External Rotational Testing

Results obtained from Internal – External rotational testing showed similar results as compared to varus – valgus testing. Results fell primarily outside of the previously defined *a*-priori limits, however in some cases systematic and random errors fell within or close to the pre-defined *a*-priori limits, respectively. Overall, it was determined that femoral motion during internal rotational testing differed relative to the device by 0.97±1.30 degrees in the frontal plane and 4.00±2.80 degrees in the transverse plane. Similarly, femoral differences for external rotations were shown to be 0.33±1.37 degrees for frontal plane rotations and 4.07±2.97 degrees for transverse plane rotations.

While frontal plane rotation of the femur can be considered an accessory motion during Internal – External rotational arthrometric testing the results described previously are, nevertheless, a promising sign of an early prototype. Frontal plane results suggest that only minor disagreements occur between the femoral head and device embedded sensors.

Accordingly, there is no more than 2.5◦ disagreement for internal and 2.0 ◦ for external loadings. Systematically, frontal plane rotations fall well the a priori defined ±2◦ clinically allowable error limit. This cannot be said for the extent of the random error between measurements, which falls approximately 0.3◦-0.4◦ outside of the allowable range.

Transverse plane results, which represents the primary plane of stabilization for IE testing, do not support the notion that the thigh was adequately stabilized during testing. From the clinical perspective the results lie outside the a priori limits by approximately 2.10◦ ±2.90◦, for both internal and external rotations.

Considering the limitations previously mentioned, this did not come as a surprise during data analysis. Interestingly, however, the largest sensor disagreements for both internal and external testing occurred during testing on Subject 4. This result, at first glance, is surprising considering the that Subject 2 was the only individual with noted condylar slippage.

That being said, Subject 4 was by far the largest limb tested (Table III.1). With this consideration, it would seem logical that the device would struggle at limiting transverse plane rotations for individuals with greater BMIs. Though logical, this insight reveals limitations to the current design.

V.4. Limitations

As with all studies, the present study was not without its limitations. While it was necessary to obtain true, ground-truth, measurements of thigh and shank movement during testing the usage of cadaveric specimens as subjects in this study was arguably its greatest limitation. There is no doubt that the inability to correctly model the fixation of the femoral-head within the acetabulofemoral detracts from the overall stability of the limb within the device.

Additionally, certain aspects of the devices current design limited it performance. As previously discussed, the friction clamping design employed on the U-bar resulted in minor failures due to fatigue to the metal clamping components. These design choices have always been a temporary fix and should be eliminated in future redesigns.

V.5. Future Directions

In its current form, the device's thigh cradle only supports one limb, in human pilot testing this leaves the contralateral limb off to the side and out of the way of the device an. As it operates, the contralateral limb has no pre-defined position, often leading the subject to lay their limb flat on the table causing the subject's pelvis to become misaligned.

To address this issue, and the greater issue of femoral stability within the device, the thigh cradle is set to become one designed to accommodate two limbs at once. In doing so, it is the opinion of the author and other collaborators on the project that, leveling the hip such that both limbs are in comparable amounts of flexion will have a stabilizing effect on the limb being tested.

Moreover, the device will undergo a series of component revisions and additions to enhance its overall ease-of-use and fixation security. For instance, as the condylar pads currently stand, they are mounted to roller tracks.

While this configuration allowed for smooth, quick adjustment of the pads all on a relatively low-profile rail, they also had significant limitations. Compared to geared or even profile-rail solutions the roller tracks necessitated greater component clearances, which created some amount of instability or 'slop' in the clamping system. Ultimately, slop is an entirely undesirable quality in a stabilization system and must be addressed moving forward.

To address this, pieces such as the frictional clamping components of the device, which had evident drawbacks, will be replaced with mechanical, geared solutions. These parts will enhance the overall ease-of-use of the device while simultaneously minimizing 'slop' and preventing slippage of the fixation components.

Additional cadaver testing will be completed to further validate the efficacy of the Thigh Stabilization System and device as a whole to obtain valid, multiplanar, arthrometric evaluations.

BIBLIOGRAPHY

[1] agdiffusion.com. Basic knee joint anatomy. [Online; accessed November 13, 2018].

[2] Willyan Hasenkamp. Flexible micro/nanofabricated systems for medical devices and implants.

[3] Human Anatomy Planes [Digital image]. (n.d.). Retrieved October 14, 2018, fromhttps://www.anatomynote.com/human-anatomy/gross-view-of-human-body/human-anatomy-planes/

[4] C. W. Schauberger, B. L. Rooney, L. Goldsmith, D. Shenton, P. D. Silva, and A. Schaper. Peripheral joint laxity increases in pregnancy but does not correlate with serum relaxin levels. *Am J Obstet Gynecol*, 174(2):667–71, 1996.

[5] G. C. Keene, P. Jeer, and R. Gul. Intraoperative arthrometry in anterior cruciate ligament reconstruction. *Acta Orthop Belg*, 71(5):565–70, 2005.

[6] L. Pugh, R. Mascarenhas, S. Arneja, P. Y. Chin, and J. M. Leith. Current concepts in instrumented knee-laxity testing. *Am J Sports Med*, 37(1):199–210, 2009.

[7] Najeeb Khan, Eric Dockter, Donald Fithian, Ronald Navarro, and William Luetzow. *Development of Arthrometry*, pages 115–129. Springer International Publishing, Cham, 2017.

[8] T. Tsukeoka and Y. Tsuneizumi. Varus and valgus stress tests after total knee arthroplasty with and without anesthesia. *Arch Orthop Trauma Surg*, 136(3):407–11, 2016.

[9] H. Robert, S. Nouveau, S. Gageot, and B. Gagniere. A new knee arthrometer, the gnrb: experience in acl complete and partial tears. *Orthop Traumatol Surg Res*, 95(3):171–6, 2009.

[10] A. Senioris, T. Rousseau, M. L'Hermette, S. Gouzy, F. Duparc, and F. Dujardin. Validity of rotational laxity coupled with anterior translation of the knee: A cadaveric study comparing radiostereometry and the rotab(r). *Knee*, 24(2):289– 294, 2017.

[11] Romain Seil, Henri Robert, Daniel Theisen, and Caroline Mouton. *Instrumented Static Laxity Evaluation*, pages 413–428. Springer Berlin Heidelberg, Berlin, Heidelberg,2017.

[12] J. Dargel, M. Gotter, K. Mader, D. Pennig, J. Koebke, and R. Schmidt-Wiethoff. Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction. *Strategies Trauma Limb Reconstr*, 2(1):1–12, 2007.

[13] J. M. Uhorchak, C. R. Scoville, G. N. Williams, R. A. Arciero, P. St Pierre, and D. C. Taylor. Risk factors associated with noncontact injury of the anterior cruciate ligament: a prospective four-year evaluation of 859 west point cadets. *Am J Sports Med*, 31(6):831–42, 2003.

[14] R. Bitton. The economic burden of osteoarthritis. *Am J Manag Care*, 15(8 Suppl):S230–5, 2009.

[15] D. Blalock, A. Miller, M. Tilley, and J. Wang. Joint instability and osteoarthritis. *Clin Med Insights Arthritis Musculoskelet Disord*, 8:15–23, 2015.

[16] D. T. Felson, R. C. Lawrence, P. A. Dieppe, R. Hirsch, C. G. Helmick, J. M. Jordan, R. S. Kington, N. E. Lane, M. C. Nevitt, Y. Zhang, M. Sowers, T. McAlin- don, T. D. Spector, A. R. Poole, S. Z. Yanovski, G. Ateshian, L. Sharma, J. A. Buckwalter, K. D. Brandt, and J. F. Fries. Osteoarthritis: new insights. part 1: the disease and its risk factors. *Ann Intern Med*, 133(8):635–46, 2000.

[17] Y. Zhang and J. M. Jordan. Epidemiology of osteoarthritis. *Clin Geriatr Med*, 26(3):355–69, 2010.

[18] L. Hendren and P. Beeson. A review of the differences between normal and osteoarthritis articular cartilage in human knee and ankle joints. *Foot (Edinb)*, 19(3):171–6, 2009.

[19] S. J. Shultz, Y. Shimokochi, A. D. Nguyen, R. J. Schmitz, B. D. Beynnon, and D. H. Perrin. Measurement of varus-valgus and internal-external rotational knee laxities in vivo–part ii: relationship with anterior-posterior and general joint laxity in males and females. *J Orthop Res*, 25(8):989–96, 2007.

[20] S. J. Shultz, R. J. Schmitz, Y. Kong, W. N. Dudley, B. D. Beynnon, A. D. Nguyen, H. Kim, and M. M. Montgomery. Cyclic variations in multiplanar knee laxity influence landing biomechanics. *Med Sci Sports Exerc*, 44(5):900–9, 2012.

[21] S. J. Shultz, M. L. Pye, M. M. Montgomery, and R. J. Schmitz. Associations between lower extremity muscle mass and multiplanar knee laxity and stiffness: a potential explanation for sex differences in frontal and transverse plane knee laxity. *Am J Sports Med*, 40(12):2836–44, 2012.

[22] S. J. Shultz, B. J. Levine, A. D. Nguyen, H. Kim, M. M. Montgomery, and D. H. Perrin. A comparison of cyclic variations in anterior knee laxity, genurecurvatum, and general joint laxity across the menstrual cycle. *J Orthop Res*, 28(11):1411–7, 2010.

[23] S. J. Shultz, W. N. Dudley, and Y. Kong. Identifying multiplanar knee laxity profiles and associated physical characteristics. *J Athl Train*, 47(2):159–69, 2012.

[24] J. B. Taylor, H. M. Wang, R. J. Schmitz, C. K. Rhea, S. E. Ross, and S. J. Shultz. Multiplanar knee laxity and perceived function during activities of daily living and sport. *J Athl Train*, 50(11):1199–206, 2015.

[25] B. P. Boden, F. T. Sheehan, J. S. Torg, and T. E. Hewett. Noncontact anterior cruciate ligament injuries: mechanisms and risk factors. *J Am Acad Orthop Surg*, 18(9):520–7, 2010.

[26] H. Koga, R. Bahr, G. Myklebust, L. Engebretsen, T. Grund, and T. Krosshaug. Estimating anterior tibial translation from model-based image-matching of a noncontact anterior cruciate ligament injury in professional football: a case report. *Clin J Sport Med*, 21(3):271–4, 2011.

[27] H. Koga, A. Nakamae, Y. Shima, J. Iwasa, G. Myklebust, L. Engebretsen, R. Bahr, and T. Krosshaug. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med*, 38(11):2218–25, 2010.

[28] S. G. McLean, X. Huang, A. Su, and A. J. Van Den Bogert. Sagittal plane biomechanics cannot injure the acl during sidestep cutting. *Clin Biomech (Bristol, Avon)*, 19(8):828–38, 2004.

[29] S. J. Shultz. Acl injury in the female athlete: a multifactorial problem that remains poorly understood. *J Athl Train*, 43(5):455, 2008.

[30] H. C. Smith, P. Vacek, R. J. Johnson, J. R. Slauterbeck, J. Hashemi, S. Shultz, and B. D. Beynnon. Risk factors for anterior cruciate ligament injury: a review of the literature-part 2: hormonal, genetic, cognitive function, previous injury, and extrinsic risk factors. *Sports Health*, 4(2):155–61, 2012.

[31] H. C. Smith, P. Vacek, R. J. Johnson, J. R. Slauterbeck, J. Hashemi, S. Shultz, and B. D. Beynnon. Risk factors for anterior cruciate ligament injury: a review of the literature - part 1: neuromuscular and anatomic risk. *Sports Health*, 4(1):69–78, 2012.

[32] H. J. Braun, R. Shultz, M. Malone, W. E. Leatherwood, A. Silder, and J. L. Dragoo. Differences in acl biomechanical risk factors between field hockey and lacrosse female athletes. *Knee Surg Sports Traumatol Arthrosc*, 23(4):1065–70, 2015.

[33] J. B. Taylor, K. R. Ford, R. J. Schmitz, S. E. Ross, T. A. Ackerman, and S. J. Shultz. Biomechanical differences in female basketball and soccer players during multi-directional jump landings. *J Strength Cond Res*, 2017.

[34] D. A. Padua, L. J. DiStefano, T. E. Hewett, W. E. Garrett, S. W. Marshall, G. M. Golden, S. J. Shultz, and S. M. Sigward. National athletic trainers' association position statement: Prevention of anterior cruciate ligament injury. *J Athl Train*, 53(1):5–19, 2018.

[35] L. Y. Griffin, J. Agel, M. J. Albohm, E. A. Arendt, R. W. Dick, W. E. Garrett, J. G. Garrick, T. E. Hewett, L. Huston, M. L. Ireland, R. J. Johnson, W. B. Kibler, S. Lephart, J. L. Lewis, T. N. Lindenfeld, B. R. Mandelbaum, P. Marchak, C. C. Teitz, and E. M. Wojtys. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *J Am Acad Orthop Surg*, 8(3):141–50, 2000.

[36] S. J. Shultz, R. J. Schmitz, A. D. Nguyen, B. Levine, H. Kim, M. M. Montgomery, Y. Shimokochi, B. D. Beynnon, and D. H. Perrin. Knee joint laxity and its cyclic variation influence tibiofemoral motion during weight acceptance. *Med Sci Sports Exerc*, 43(2):287–95, 2011.

[37] S. J. Shultz, L. Wideman, M. M. Montgomery, K. N. Beasley, and B. C. Nindl. Changes in serum collagen markers, igf-i, and knee joint laxity across the menstrual cycle. *J Orthop Res*, 30(9):1405–12, 2012.

[38] R. J. Schmitz and S. J. Shultz. Anterior knee stiffness changes in laxity "responders" versus "nonresponders" across the menstrual cycle. *J Athl Train*, 48(1):39–46, 2013.

[39] S. J. Shultz, R. J. Schmitz, J. R. Cone, T. J. Copple, M. M. Montgomery, M. L. Pye, and A. J. Tritsch. Multiplanar knee laxity increases during a 90-min intermittent exercise protocol. *Med Sci Sports Exerc*, 45(8):1553–61, 2013.

[40] S. J. Shultz, R. J. Schmitz, J. R. Cone, R. A. Henson, M. M. Montgomery, M. L. Pye, and A. J. Tritsch. Changes in fatigue, multiplanar knee laxity, and landing biomechanics during intermittent exercise. *J Athl Train*, 50(5):486–97, 2015.

[41] C. F. van Eck, M. Loopik, M. P. van den Bekerom, F. H. Fu, and G. M. Kerkhoffs. Methods to diagnose acute anterior cruciate ligament rupture: a meta-analysis of instrumented knee laxity tests. *Knee Surg Sports Traumatol Arthrosc*, 21(9):1989–97, 2013.

[42] A. Benjaminse, A. Gokeler, and C. P. van der Schans. Clinical diagnosis of an anterior cruciate ligament rupture: a meta-analysis. *J Orthop Sports Phys Ther*, 36(5):267–88, 2006.

[43] R. Vauhnik, M. P. Perme, M. G. Barcellona, D. Rugelj, M. C. Morrissey, and F. Sevsek. Robotic knee laxity testing: reliability and normative data. *Knee*, 20(4):250–5, 2013.

[44] S. H. Wiertsema, H. J. van Hooff, L. A. Migchelsen, and M. P. Steultjens. Reliability of the kt1000 arthrometer and the lachman test in patients with an acl rupture. *Knee*, 15(2):107–10, 2008.

[45] P. Boyer, P. Djian, P. Christel, X. Paoletti, and R. Degeorges. [reliability of the kt-1000 arthrometer (medmetric) for measuring anterior knee laxity: comparison with telos in 147 knees]. *Rev Chir Orthop Reparatrice Appar Mot*, 90(8):757–64, 2004.

[46] C. Jardin, C. Chantelot, H. Migaud, F. Gougeon, M. J. Debroucker, and A. Duquennoy. [reliability of the kt-1000 arthrometer in measuring anterior laxity of the knee: comparative analysis with telos of 48 reconstructions of the anterior cruciate ligament and intra- and interobserver reproducibility]. *Rev Chir Orthop Reparatrice Appar Mot*, 85(7):698–707, 1999.

[47] R. R. Wroble, L. A. Van Ginkel, E. S. Grood, F. R. Noyes, and B. L. Shaffer. Repeatability of the kt-1000 arthrometer in a normal population. *Am J Sports Med*, 18(4):396–9, 1990.

[48] C. L. Highgenboten, A. Jackson, and N. B. Meske. Genucom, kt-1000, and stryker knee laxity measuring device comparisons. device reproducibility and interdevice comparison in asymptomatic subjects. *Am J Sports Med*, 17(6):743–6, 1989.

[49] M. Collette, J. Courville, M. Forton, and B. Gagniere. Objective evaluation of anterior knee laxity; comparison of the kt-1000 and gnrb(r) arthrometers. *Knee Surg Sports Traumatol Arthrosc*, 20(11):2233–8, 2012.

[50] D. Mouarbes, E. Cavaignac, P. Chiron, E. Berard, and J. Murgier. Evaluation of reproducibility of robotic knee testing device (gnrb) on 60 healthy knees. *J Orthop*, 15(1):94–98, 2018.

[51] Y. Alqahtani, J. Murgier, P. Beaufils, P. Boisrenoult, C. Steltzlen, and N. Pujol. Anterior tibial laxity using the gnrb(r) device in healthy knees. *Knee*, 25(1):34–39, 2018.

[52] S. K. Stinton, R. Siebold, H. Freedberg, C. Jacobs, and T. P. Branch. The use of a robotic tibial rotation device and an electromagnetic tracking system to accurately reproduce the clinical dial test. *Knee Surg Sports Traumatol Arthrosc*, 24(3):815–22, 2016.

[53] T. P. Branch, S. K. Stinton, R. Siebold, H. I. Freedberg, C. A. Jacobs, and W. C. Hutton. Assessment of knee laxity using a robotic testing device: a comparison to the manual clinical knee examination. *Knee Surg Sports Traumatol Arthrosc*, 25(8):2460–2467, 2017.

[54] T. Branch, S. Stinton, M. Sternberg, W. Hutton, F. Lavoie, C. Guier, and P. Neyret. Robotic axial lower leg testing: repeatability and reproducibility. *Knee Surg Sports Traumatol Arthrosc*, 23(10):2892–9, 2015.

[55] S. Kumar, Y. Narayan, and C. Bacchus. Symmetric and asymmetric two-handed pull-push strength of young adults. *Hum Factors*, 37(4):854–65, 1995.

[56] N. Sernert, J. Helmers, C. Kartus, L. Ejerhed, and J. Kartus. Knee-laxity measurements examined by a left-hand- and a right-hand-dominant physiotherapist, in patients with anterior cruciate ligament injuries and healthy controls. *Knee Surg Sports Traumatol Arthrosc*, 15(10):1181–6, 2007.

[57] L. F. Draganich, M. R. Sathy, and B. Reider. The effect of thigh and goniometer restraints on the reproducibility of the genucom knee analysis system. *Am J Sports Med*, 22(5):627–31, 1994.

[58] F. M. Griffin, K. Math, G. R. Scuderi, J. N. Insall, and P. L. Poilvache. Anatomy of the epicondyles of the distal femur: Mri analysis of normal knees. *J Arthro- plasty*, 15(3):354–9, 2000.

[59] J. Martin Bland & Douglas G. Altman (2007) Agreement Between Methods of Measurement with Multiple Observations Per Individual, Journal of Biopharmaceutical Statistics, 17:4, 571-582, DOI: 10.1080/10543400701329422

[60] E. W. James, C. M. LaPrade, and R. F. LaPrade. Anatomy and biomechanics of the lateral side of the knee and surgical implications. *Sports Med Arthrosc Rev*, 23(1):2–9, 2015.

[61] F. Flandry and G. Hommel. Normal anatomy and biomechanics of the knee. *Sports Med Arthrosc Rev*, 19(2):82–92, 2011.

[62] R. F. LaPrade, A. H. Engebretsen, T. V. Ly, S. Johansen, F. A. Wentorf, and Engebretsen. The anatomy of the medial part of the knee. *J Bone Joint Surg Am*, 89(9):2000–10, 2007.

[63] T. A. Blackburn and E. Craig. Knee anatomy: a brief review. *Phys Ther*, 60(12):1556–60, 1980.

[64] A. J. Fox, F. Wanivenhaus, and S. A. Rodeo. The basic science of the patella: structure, composition, and function. *J Knee Surg*, 25(2):127–41, 2012.

[65] J. M. Postel. [knee joint. anatomy and biomechanics]. *Soins Chir*, (184):5–9, 1997.

[66] K. E. Kuettner and A. A. Cole. Cartilage degeneration in different human joints. *Osteoarthritis Cartilage*, 13(2):93–103, 2005.

[67] Jr. Bowman, K. F. and J. K. Sekiya. Anatomy and biomechanics of the posterior cruciate ligament, medial and lateral sides of the knee. *Sports Med Arthrosc Rev*, 18(4):222–9, 2010.

[68] D. S. Hungerford and M. Barry. Biomechanics of the patellofemoral joint. *Clin Orthop Relat Res*, (144):9–15, 1979.

[69] A. Steinbruck, S. Milz, M. Woiczinski, C. Schroder, S. Utzschneider, V. Jansson, and A. Fottner. [anatomy and biomechanics of the patellofemoral joint: physiological conditions and changes after total knee arthroplasty]. *Orthopade*, 40(10):848, 850–2, 854, 2011.

[70] R. A. Brand, R. D. Crowninshield, C. E. Wittstock, D. R. Pedersen, C. R. Clark, and F. M. van Krieken. A model of lower extremity muscular anatomy. *J Biomech Eng*, 104(4):304–10, 1982.

[71] R. R. Neptune and C. P. McGowan. Muscle contributions to frontal plane angular momentum during walking. *J Biomech*, 49(13):2975–2981, 2016.

[72] B. R. Meister, S. P. Michael, R. A. Moyer, J. D. Kelly, and C. D. Schneck. Anatomy and kinematics of the lateral collateral ligament of the knee. *Am J Sports Med*, 28(6):869–78, 2000.

[73] E. S. Grood, S. F. Stowers, and F. R. Noyes. Limits of movement in the human knee. effect of sectioning the posterior cruciate ligament and posterolateral structures. *J Bone Joint Surg Am*, 70(1):88–97, 1988.

[74] K. Jensen. Manual laxity tests for anterior cruciate ligament injuries. *J Orthop Sports Phys Ther*, 11(10):474–81, 1990.

[75] M. A. Freeman and B. Wyke. The innervation of the ankle joint. an anatomical and histological study in the cat. *Acta Anat (Basel)*, 68(3):321–33, 1967.

[76] E. Tsuda, Y. Okamura, H. Otsuka, T. Komatsu, and S. Tokuya. Direct evidence of the anterior cruciate ligament-hamstring reflex arc in humans. *Am J Sports Med*, 29(1):83–7, 2001.

[77] R. L. Cooper, N. F. Taylor, and J. A. Feller. A systematic review of the effect of proprioceptive and balance exercises on people with an injured or reconstructed anterior cruciate ligament. *Res Sports Med*, 13(2):163–78, 2005.

[78] J. Haus and Z. Halata. Innervation of the anterior cruciate ligament. *Int Orthop*, 14(3):293–6, 1990.

[79] H. Johansson, P. Sjolander, and P. Sojka. A sensory role for the cruciate ligaments. *Clin Orthop Relat Res*, (268):161–78, 1991.

[80] J. C. Kennedy, I. J. Alexander, and K. C. Hayes. Nerve supply of the human knee and its functional importance. *Am J Sports Med*, 10(6):329–35, 1982.

[81] G. Pap, A. Machner, W. Nebelung, and F. Awiszus. Detailed analysis of proprioception in normal and acl-deficient knees. *J Bone Joint Surg Br*, 81(5):764– 8, 1999.

[82] B. Reider, M. A. Arcand, L. H. Diehl, K. Mroczek, A. Abulencia, C. C. Stroud, Palm, J. Gilbertson, and P. Staszak. Proprioception of the knee before and after anterior cruciate ligament reconstruction. *Arthroscopy*, 19(1):2–12, 2003.

[83] B. L. Riemann and S. M. Lephart. The sensorimotor system, part ii: The role of proprioception in motor control and functional joint stability. *J Athl Train*, 37(1):80–4, 2002.

[84] Sara S. P. Marreiros, Jody L. Riskowski, and Martijn P. Steultjens. pro- prioception and laxity in patients with joint hypermobility syndrome with and without knee instability. Rheumatology, 55(suppl$_1$): i178 – –i179, 2016.

[85] R. Sandberg, B. Balkfors, A. Henricson, and N. Westlin. Stability tests in knee ligament injuries. *Arch Orthop Trauma Surg*, 106(1):5–7, 1986.

[86] A. Harilainen. Evaluation of knee instability in acute ligamentous injuries. *Ann Chir Gynaecol*, 76(5):269–73, 1987.

[87] A. F. Anderson, R. B. Snyder, C. F. Federspiel, and A. B. Lipscomb. Instrumented evaluation of knee laxity: a comparison of five arthrometers. *Am J Sports Med*, 20(2):135–40, 1992.

[88] J. C. Kupper, B. Loitz-Ramage, D. T. Corr, D. A. Hart, and J. L. Ronsky. Measuring knee joint laxity: a review of applicable models and the need for new approaches to minimize variability. *Clin Biomech (Bristol, Avon)*, 22(1):1–13, 2007.

[89] M. K. Lotz and V. B. Kraus. New developments in osteoarthritis. posttraumatic osteoarthritis: pathogenesis and pharmacological treatment options. *Arthritis Res Ther*, 12(3):211, 2010.

[90] G. P. Graham, S. Johnson, C. M. Dent, and J. A. Fairclough. Comparison of clinical tests and the kt1000 in the diagnosis of anterior cruciate ligament rupture. *Br J Sports Med*, 25(2):96–7, 1991.

[91] D. M. Daniel, M. L. Stone, R. Sachs, and L. Malcom. Instrumented measurement of anterior knee laxity in patients with acute anterior cruciate ligament disruption. *Am J Sports Med*, 13(6):401–7, 1985.

[92] O. H. Sherman, K. L. Markolf, and R. D. Ferkel. Measurements of anterior laxity in normal and anterior cruciate absent knees with two instrumented test devices. *Clin Orthop Relat Res*, (215):156–61, 1987.

[93] B. S. Uh, B. D. Beynnon, D. L. Churchill, L. D. Haugh, M. A. Risberg, and B. C. Fleming. A new device to measure knee laxity during weightbearing and non-weightbearing conditions. *J Orthop Res*, 19(6):1185–91, 2001.

[94] E. M. Rohman and J. A. Macalena. Anterior cruciate ligament assessment using arthrometry and stress imaging. *Curr Rev Musculoskelet Med*, 9(2):130–8, 2016.

[95] T. P. Branch, H. O. Mayr, J. E. Browne, J. C. Campbell, A. Stoehr, and C. A. Jacobs. Instrumented examination of anterior cruciate ligament injuries: minimizing flaws of the manual clinical examination. *Arthroscopy*, 26(7):997–1004, 2010.

[96] H. Balasch, M. Schiller, H. Friebel, and F. Hoffmann. Evaluation of anterior knee joint instability with the rolimeter. a test in comparison with manual assessment and measuring with the kt-1000 arthrometer. *Knee Surg Sports Traumatol Arthrosc*, 7(4):204–8, 1999.

[97] D. Ericsson, A. H. Ostenberg, E. Andersson, and M. Alricsson. Test-retest reliability of repeated knee laxity measurements in the acute phase following a knee trauma using a rolimeter. *J Exerc Rehabil*, 13(5):550–558, 2017.

[98] M. G. Papandreou, E. Antonogiannakis, C. Karabalis, and K. Karliaftis. Inter-rater reliability of rolimeter measurements between anterior cruciate ligament injured and normal contra lateral knees. *Knee Surg Sports Traumatol Arthrosc*, 13(7):592–7, 2005.

[99] T. Muellner, W. Bugge, S. Johansen, C. Holtan, and L. Engebretsen. Inter- and intratester comparison of the rolimeter knee tester: effect of tester's experience and the examination technique. *Knee Surg Sports Traumatol Arthrosc*, 9(5):302–6, 2001.

[100] R. R. Wroble, E. S. Grood, F. R. Noyes, and D. J. Schmitt. Reproducibility of genucom knee analysis system testing. *Am J Sports Med*, 18(4):387–95, 1990.

[101] J. H. Oliver and L. P. Coughlin. Objective knee evaluation using the genucom knee analysis system. clinical implications. *Am J Sports Med*, 15(6):571–8, 1987.

[102] Fraser GA. Mar. 17, 1987 1987. Canada patent 4,649,934.

[103] A. F. Anderson and A. B. Lipscomb. Preoperative instrumented testing of anterior and posterior knee laxity. *Am J Sports Med*, 17(3):387–92, 1989.

[104] K. L. Markolf, J. S. Mensch, and H. C. Amstutz. Stiffness and laxity of the knee– the contributions of the supporting structures. a quantitative in vitro study. *J Bone Joint Surg Am*, 58(5):583–94, 1976.

[105] K. L. Markolf, A. Graff-Radford, and H. C. Amstutz. In vivo knee stability. a quantitative assessment using an instrumented clinical testing apparatus. *J Bone Joint Surg Am*, 60(5):664–74, 1978.

[106] K. L. Markolf and H. C. Amstutz. The clinical relevance of instrumented testing for acl insufficiency. experience with the ucla clinical knee testing apparatus. *Clin Orthop Relat Res*, (223):198–207, 1987.

[107] G. Garavaglia, A. Lubbeke, V. Dubois-Ferriere, D. Suva, D. Fritschy, and J. Menetrey. Accuracy of stress radiography techniques in grading isolated and combined posterior knee injuries: a cadaveric study. *Am J Sports Med*, 35(12):2051–6, 2007.

[108] M. S. Schulz, K. Russe, G. Lampakis, and M. J. Strobel. Reliability of stress ra- diography for evaluation of posterior knee laxity. *Am J Sports Med*, 33(4):502–6, 2005.

[109] T. E. Hewett, F. R. Noyes, and M. D. Lee. Diagnosis of complete and partial posterior cruciate ligament ruptures. stress radiography compared with kt-1000 arthrometer and posterior drawer testing. *Am J Sports Med*, 25(5):648–55, 1997.

[110] N. Lefevre, Y. Bohu, J. F. Naouri, S. Klouche, and S. Herman. Validity of gnrb(r) arthrometer compared to telos in the assessment of partial anterior cruciate ligament tears. *Knee Surg Sports Traumatol Arthrosc*, 22(2):285–90, 2014.

[111] N. Bouguennec, G. A. Odri, N. Graveleau, and P. Colombet. Comparative reproducibil- ity of telos and gnrb(r) for instrumental measurement of anterior tibial translation in normal knees. *Orthop Traumatol Surg Res*, 101(3):301–5, 2015.

[112] J. Beldame, S. Bertiaux, X. Roussignol, B. Lefebvre, J. M. Adam, F. Mouilhade, and F. Dujardin. Laxity measurements using stress radiography to assess anterior cruciate ligament tears. *Orthop Traumatol Surg Res*, 97(1):34–43, 2011.

[113] S. M. Ryu, H. D. Na, and O. J. Shon. Diagnostic tools for acute anterior cruciate ligament injury: Gnrb, lachman test, and telos. *Knee Surg Relat Res*, 30(2):121–127, 2018.

[114] J. Beldame, S. Mouchel, S. Bertiaux, J. M. Adam, F. Mouilhade, X. Roussignol, and F. Dujardin. Anterior knee laxity measurement: comparison of passive stress radiographs telos((r)) and "lerat", and gnrb((r)) arthrometer. *Orthop Traumatol Surg Res*, 98(7):744–50, 2012.

[115] B. D. Beynnon, B. C. Fleming, R. Labovitch, and B. Parsons. Chronic anterior cruciate ligament deficiency is associated with increased anterior translation of the tibia during the transition from non-weightbearing to weightbearing. *J Orthop Res*, 20(2):332–7, 2002.

[116] S. J. Shultz, Y. Shimokochi, A. D. Nguyen, R. J. Schmitz, B. D. Beynnon, and D. H. Perrin. Measurement of varus-valgus and internal-external rotational knee laxities in vivo–part i: assessment of measurement reliability and bilateral asymmetry. *J Orthop Res*, 25(8):981–8, 2007.

[117] J. Y. Jenny, B. Puliero, G. Schockmel, S. Harnoist, and P. Clavert. Experimental validation of the gnrb((r)) for measuring anterior tibial translation. *Orthop Traumatol Surg Res*, 103(3):363–366, 2017.

[118] R. Vauhnik, M. C. Morrissey, M. P. Perme, F. Sevsek, and D. Rugelj. Inter-rater reliability of the gnrb(r) knee arthrometer. *Knee*, 21(2):541–3, 2014.

[119] B. T. Ballantyne, A. K. French, S. L. Heimsoth, A. F. Kachingwe, J. B. Lee, and G. L. Soderberg. Influence of examiner experience and gender on interrater reliability of kt-1000 arthrometer measurements. *Phys Ther*, 75(10):898–906, 1995.

[120] I. W. Forster, C. D. Warren-Smith, and M. Tew. Is the kt1000 knee ligament arthrometer reliable? *J Bone Joint Surg Br*, 71(5):843–7, 1989.

[121] J. Berry, K. Kramer, J. Binkley, G. A. Binkley, P. Stratford, S. Hunter, and K. Brown. Error estimates in novice and expert raters for the kt-1000 arthrometer. *J Orthop Sports Phys Ther*, 29(1):49–55, 1999.

[122] N. Sernert, Jr. Kartus, J. T., L. Ejerhed, and J. Karlsson. Right and left knee laxity measurements: a prospective study of patients with anterior cruciate ligament injuries and normal control subjects. *Arthroscopy*, 20(6):564–71, 2004.

[123] C. Mouton, D. Theisen, and R. Seil. Objective measurements of static anterior and rotational knee laxity. *Curr Rev Musculoskelet Med*, 9(2):139–47, 2016.

[124] K. M. Guskiewicz, D. H. Perrin, D. E. Martin, D. M. Kahler, B. M. Gansneder, and F. C. McCue. Effect of acl reconstruction and tibial rotation on anterior knee laxity. *J Athl Train*, 30(3):243–6, 1995.

[125] I. Fiebert, J. Gresley, S. Hoffman, and K. Kunkel. Comparative measurements of anterior tibial translation using the kt-1000 knee arthrometer with the leg in neutral, internal rotation, and external rotation. *J Orthop Sports Phys Ther*, 19(6):331–4, 1994.

[126] A. Lorenz, H. Rottgerkamp, E. Bobrowitsch, C. I. Leichtle, and U. G. Leichtle. Tibial rotation influences anterior knee stability–a robot-aided in-vitro study. *Clin Biomech (Bristol, Avon)*, 32:131–7, 2016.

[127] K. L. Markolf, W. L. Bargar, S. C. Shoemaker, and H. C. Amstutz. The role of joint load in knee stability. *J Bone Joint Surg Am*, 63(4):570–85, 1981.

[128] K. Shino, M. Inoue, S. Horibe, H. Nakamura, and K. Ono. Measurement of anterior instability of the knee. a new apparatus for clinical testing. *J Bone Joint Surg Br*, 69(4):608–13, 1987.

[129] Tsung-Yuan Tsai, Tung-Wu Lu, Mei-Ying Kuo, and Cheng-Chung Lin. Effects of soft tissue artifacts on the calculated kinematics and kinetics of the knee during stair-ascent. *Journal of biomechanics*, 44:1182–8, 02 2011.

[130] Alberto Leardini, Lorenzo Chiari, Ugo Della Croce, and Aurelio Cappozzo. Human movement analysis using stereophotogrammetry. part 3. soft tissue artifact assessment and compensation. *Gait Posture*, 21(2):212–225, Feb 2005.

[131] T. P. Branch, S. K. Stinton, W. C. Hutton, and P. Neyret. The combination of tibial anterior translation and axial rotation into a single biomechanical factor improves the prediction of patient satisfaction over each factor alone in patients with acl reconstructed knees. *Knee Surg Sports Traumatol Arthrosc*, 25(4):1038–1047, 2017.