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The purpose of this investigation was to establish day to day reliability of varus/valgus and internal/external rotational stiffness measures and then compare stiffnesses between males and females. Twenty healthy college students underwent varus/valgus (non-weightbearing) and internal/external (non-weight and weightbearing) applied torques to 10, and 5 Nm, respectively. Ten subjects returned a second day to establish reliability measures. Stiffness constants were calculated for each displacement created by a .5 Nm incrementally applied torque. Results revealed mean female stiffness was significantly less than males for valgus, varus, and weightbearing external rotational stiffness. Interactions demonstrated that female knees were less stiff during initial loading. Female knee joint stiffness increased to equal male stiffness during internal rotation, external rotation, and weightbearing internal rotation. These results suggest that with respect to males, females are in different joint positions as loads are applied, potentially causing a need for alternate strategies to control joint orientation.

VARUS/VALGUS AND INTERNAL/EXTERNAL ROTATIONAL KNEE JOINT STIFFNESS IN MALES AND FEMALES

by

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APPROVAL PAGE

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

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CHAPTER I

INTRODUCTION

Stiffness is a physical quantity often studied in biomechanics that is used to describe the deformability of an object or body under a given load. Stiffness is considered to have implications for both performance and for injury (Butler, Crowell, & McClay-Davis, 2003). Biomechanical stiffness can be examined at varying degrees of detail and complexity. Leg-spring stiffness, joint stiffness, and the stiffness of individual structures (such as ligaments) can and have all been reported (Butler et al., 2003; Latash & Zatsiorsky, 1993). .

Human knee joint stiffness can be represented as a torsional spring (Farley, Houdijk, Van Strien, & Louie, 1998) and thusly described through mechanical means. Although specific knee joint stiffness has been included in sagittal plane studies of overall leg spring stiffness (Arampatzis, Bruggemann, & Metzler, 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002), stiffness and laxity in varus/valgus and internal/external rotation is less often reported. Among these are studies assessing varus/valgus (Crowninshield, Pope, & Johnson, 1976; Gollehon, Torzilli, & Warren, 1987; Grood, Stowers, & Noyes, 1988; Markolf, Bargar, Shoemaker, & Amstutz, 1981; Markolf, Mensch, & Amstutz, 1976; Nielsen, Rasmussen, Ovesen, & Andersen, 1984) and internal/external rotational stiffness (Crowninshield et al., 1976; Gollehon et al., 1987; Grood et al., 1988; Hsieh & Walker, 1976; Markolf et al., 1981; Markolf et al., 1976; Nielsen, Rasmussen et al., 1984; Shoemaker & Markolf, 1985; Suggs, Wang, & Li, 2003; Wang & Walker, 1974) in vitro. Studies that examine angular displacement and stiffness in both varus/valgus and internal/external rotation *in vivo* are fewer (Markolf, Kochan, & Amstutz, 1984; Mills & Hull, 1991a, 1991b).

In vivo stiffness characteristics in varus/valgus and internal/external rotation may be a contributing factor to the sex bias seen in acute anterior cruciate ligament (ACL) injury (Hsu 2006) as ACL injury has been associated with both valgus and external rotational knee joint mechanics (Fung & Zhang, 2003; Olsen, Myklebust, Engebretsen, & Bahr, 2004).

There appears to be a paucity of studies that compare varus/valgus and internal/external rotational stiffness between males and females. Hsu et al (2006) demonstrated in cadaver knees that females have decreased stiffness compared to males in combined rotary loads of 10 Nm Valgus and ±5 Nm internal tibial torque. However, it remains unclear if there are sex differences in varus/valgus or internal/external rotational stiffness in vivo in the healthy, intact knee joint.

Additionally, traditionally reported curves of angular displacement under load have presented two stiffness “phases”. The breakpoint-style two-phase model (Hsu, Fisk, Yamamoto, Debski, & Woo, 2006; Markolf et al., 1976; Shoemaker & Markolf, 1982) uses an instantaneous stiffness at zero displacement and its intersection with a “terminal stiffness” calculated as the slope at some maximal moment. Due to different load ranges and tangent point choices in analysis of these curves, the use of a two-phase stiffness model, though widely accepted, may be limited in its generalizability. Inconsistencies and lack of methodological detail exist in the choice of that maximal terminal tangent point.

Finally, there is a possibility that examining stiffness only as the inverse of laxity, which is displacement under a given load, may conceal information. By more thoroughly understanding the changing stiffness in an overall laxity measure, it can be better determined where, during the measured displacement, that the knee joint is more stiff, and, conversely, where the knee is less resistive to applied loads.

.	Thus, the purposes of this study are first, to determine and characterize the day to day measurement consistency of healthy in vivo knee joint stiffness as a physical quantity in varus/valgus rotation and internal/external rotation when loaded by external torques and second, to compare stiffness in males and females.

Objectives

Objective 1

Determine the between day measurement consistency of incrementally derived stiffness across the varus/valgus and internal/external rotational loading cycles.

Hypothesis 1: Incrementally derived stiffness will demonstrate day to day measurement consistency.

Objective 2

Form characteristic varus/valgus and internal/external rotation torque by angular displacement curves for sample for 40 healthy normally-functioning knees (20 male, 20 female) by averaging the displacements by known loads over entire loading path. Describe these curves for whole sample, males, females, and weightbearing vs. non-weightbearing conditions (internal/external rotation only).

Objective 3

Derive incremental stiffness constants for each 0.5 Nm of applied torque, and determine the point(s) at which incremental stiffness changes along the torque by displacement curves while also determining if sex differences exist.

Hypothesis 2: Females will exhibit lower stiffness in varus/valgus and internal/external rotation at some points in the curve than males.

Limitations / Assumptions

Limitations to this study include the following:

1. Subjects are healthy with no history of knee ligament injury or surgery, and no history of injury or chronic pain in either lower extremity for the past 6 months. Results may only be generalized to a similar population.
2. Though every effort is made to secure the femur and prevent its rotation from occluding tibial displacement results, this is an *in vivo* study, and soft tissue deformation during displacements is inevitable to some degree.
3. Measured displacements are dependent upon the ability of electromagnetic sensors to accurately depict the anatomy of focus.
4. All measurements are made in a laboratory setting and not in athletic or other functional task. The results may not be generalizable to the knee as functioning in sport or other tasks.

Delimitations

1. The Varus/valgus moment was exerted at a known distance from the knee joint center on the distal tibia using a hand-held transducer for force measurement, while internal/external rotational moments were exerted using the VKLD foot cradle and were directly measured by the foot cradle's transducer.
2. The stiffness reported is a quasi-stiffness (Latash & Zatsiorsky, 1993) in that it is unknown if the measurements are taken at equilibria, and the viscous and inertial impacts upon force are not directly addressed.

Operational Definitions

Stiffness: The deformability of an object by external load. The spring stiffness constant *k* (unit = N/m linearly and Nm/degree angularly) is derived by dividing change in force (between equilibria) by change in position (between equilibria) in the case of linear displacements and change in moment by change in angle at equilibria for angular displacements. In true spring stiffness, potential energy is stored as the spring is displaced. This energy is equal to the work required to cause the displacement.

Incremental Stiffness: Defined as the stiffness constant *k* derived over the displacement caused by each successive .5 Nm applied torque (calculated as $\Delta M/\Delta\theta$ over that displacement). For example, the stiffness for the displacement caused from 1.5-2.0 Nm applied torque, then from 2.0-2.5 Nm applied torque, and so forth.

Valgus Rotational Stiffness: The stiffness of the knee joint when loaded in the valgus direction.

Varus Rotational Stiffness: The stiffness of the knee joint when loaded in the varus direction.

Internal Rotational Stiffness: The stiffness of the knee joint when loaded in the internal rotation direction. For the purposes of the present study, this could be in a non-weightbearing or a weightbearing condition.

External Rotational Stiffness: The stiffness of the knee joint when loaded in the external rotation direction. Again, for this study, such a load could be in the weightbearing or non-weightbearing condition.

VKLD: The Vermont Knee Laxity Device. A device designed for the measurement of knee laxity, and adapted for use in measuring knee joint stiffness for the present study. Details are presented in Uh, et al (2001) and in the methods section of this study.

Quasi-stiffness: The instantaneous quotient of force by position (linear) or moment by angle (angular). This expression assumes the effects of viscosity and inertia to be inherent in the force on/by the spring, and does not require the measurements to be made at equilibrium. Graphically, it is the slope of the load by displacement curve at a given point and is expressed mathematically as:

$k(t) = \frac{dF}{dx}$ linearly and $k(t) = \frac{dM}{d\theta}$ angularly. (Unit = N/m linearly and Nm/degree angularly).

Torque: (Moment) A force couple that results in angular acceleration of a body. It is the cross product of force and the distance from the axis of rotation to the point the force is applied (lever or moment arm). Unit = Newton-meter

Angular displacement: The change in angle from an initial position to the final position after movement. Unit = degrees.

(Vertical) Leg spring stiffness: A modeling of the body as a classic mass-spring system (McMahon & Cheng, 1990; McMahon, Valiant, & Frederick, 1987). The leg is modeled as the spring and stiffness is derived as the force applied vertically to the legs divided by the vertical displacement of the body's center of mass.

Viscosity: Fluid friction. This is a resistance (force) to attempted displacement and is proportional in magnitude and opposite in direction to velocity. In an oscillating system, viscosity would act to reduce amplitude of motion with each cycle – this is known as viscous damping.

Inertia: Mass. This is the property of any body to resist acceleration by an applied force. Rotational inertia describes a body's resistance to angular acceleration by an applied moment.

Work: The dot product of an applied force and the displacement over which the force acts. In the angular sense, it is the product of moment and angular displacement over which the moment acts. For a torsional spring, the work done on the spring is equal to the energy dissipated by, plus the energy stored in the spring. Unit = Newton-meters, or Joules.

CHAPTER II

REVIEW OF THE LITERATURE

Characterizing knee joint stiffness is not a simply defined task due to the myriad of methods in which stiffnesses of anatomical structures are described. To fully characterize stiffness it must be understood how stiffness has been defined and characterized mathematically in previous biomechanical studies. The focus of this literature review is the definition, the biomechanical background, and the mathematical interpretation of knee joint stiffness.

Stiffness Defined

Classical mechanics background states stiffness quantifies the deformability of an object in terms of the quotient of force applied to displacement (in accordance with Hooke's law). The simplest example is that of an ideal spring in one dimension. To measure its stiffness, a force is applied axially to the spring, and the displacement along the same path is measured. The ratio of force applied to linear displacement is a proportionality constant and is referred to as the spring constant, or stiffness, *k*. Mathematically this appears:

$$F = -k \cdot x \qquad \text{(eq. 1)}$$

Where "x" denotes linear displacement of the spring at the point of force application, and *k* is the spring constant (stiffness).

If a force is applied when the spring is at equilibrium, and a new equilibrium is established, the spring constant can be found using:

$$k = \frac{-\Delta F}{\Delta x} \qquad \text{(eq. 2)}$$

The negative sign on the right side of both equations suggests that the force applied and the displacement of the spring are opposite in direction.

Equations 1 and 2 imply that the spring is massless and that stiffness, measured with force and displacement, is independent of time. So, the equation(s) will only hold true when the measurements are taken at static equilibria. Also ignored in these equations are any inertial elements of the spring system. In reality, springs are not massless, and are also typically part of systems that involve inertial components as well as viscous (velocity-opposing) damping components. Further, reported stiffness measurements are often made dynamically, surpassing the ability of these equations to adequately capture stiffness. It is therefore imperative that biomechanical studies reporting stiffness as a physical quantity explain the way stiffness is derived.

Often, in biomechanics, the instantaneous expression of stiffness is used to address the concern of time dependency or a dynamic system:

$$k(t) = \frac{dF}{dx} \qquad \text{(eq. 3)}$$

Where **dF/d*x*** represents the differential of force with respect to displacement in one dimension, and ***k*(*t*)** is the stiffness as a function of time. This allows for dynamically changing stiffness, but ignores inertia and viscosity. To understand the preference for this expression, the full expression of spring-exerted force must be considered.

Equation 3 is a simplification. A more complete picture of stiffness, including inertial components, viscous damping elements, and inherent stiffness appears:

$$F(t) = m(t)\,\frac{d^2x}{dt^2} + b(t)\,\frac{dx}{dt} + k(t)\,x(t) \qquad \text{(eq. 4)}$$

This is force by a spring, where m(*t*) is inertia (mass), b(*t*) is viscosity (velocity dependent and always opposed to velocity in direction), k(*t*) is stiffness, x is length, and *t* is time.

To derive instantaneous spring stiffness, eq. 4 is differentiated by *t* and both sides divided by d*x*/d*t* to render dF/d*x* as in eq. 3. However, dF/d*x* derived this way is not simply *k*(*t*), but rather *k*(*t*) plus other inertial load-dependent and velocity-dependent terms. Reporting ***k*(*t*)** as the derivative of force with respect to displacement as in equation 3, then, conceals some omissions.

Velocity-dependent viscosity and inertial considerations that would be reflected in the acceleration term at the fore of the right side of equation 4 are ignored. It is important to know that stiffness reported as in eq.3 is a simplification (viscosity and inertial effects are not addressed in the force component), and is actually termed quasi-stiffness. A more thorough treatment of this can be found in Latash and Zatsiorsky (1993), and Barger and Olsson (1995). To include every contributor to stiffness in a mathematical expression becomes impractical, and may include so may assumptions so as to be invalid (Butler et al., 2003).

The term quasi-stiffness is used in that viscous damping and inertia are ignored (requirements of eq. 4) and the measurements are not made at static equilibrium (requirements of eq. 2) yet we consider the constant *k* derived therefrom to be reflective of a Hooke's law-abiding spring. This simplified approach has been utilized in many lower extremity stiffness studies (Farley et al., 1998; Markolf, Graff-Radford, & Amstutz, 1978; Markolf et al., 1984; Markolf et al., 1976; McMahon & Cheng, 1990; Zhang, Nuber, Butler, Bowen, & Rymer, 1998; Zhang & Wang, 2001). Latash & Zatsiorsky (1993) have pointed out that many of these studies reporting stiffness have actually calculated a quasi-stiffness.

Torsional Spring Stiffness

Only linear motion and stiffness in one dimension have been examined thus far. Linear stiffness is sometimes used to depict leg spring stiffness (Arampatzis et al., 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002; McMahon & Cheng, 1990; McMahon et al., 1987) The leg spring can be considered a kinetic chain of the hip, knee, and ankle joints. To measure individual joint contributions to leg spring stiffness, quasi-stiffness can be adapted to angular measure.

A joint is not a deformable object per se (Latash & Zatsiorsky, 1993). Rather than an object, it is a complex articulation of objects moving angularly with substructures moving both linearly and/or angularly. As such, a joint can be simplified as an axis that two segments share in their counter rotations with the prime mover being a moment or torque. Joint velocity and displacement are described angularly with moments, angular displacements, and angular velocities and accelerations replacing forces, linear displacements, and linear velocities and accelerations respectively. So, linear equations 1- 3 become the respective angular analogs (equations 5- 7):

$$\mathbf{M = -}\boldsymbol{k\theta} \qquad \text{(eq. 5)}$$

$$k = \frac{-\Delta M}{\Delta \theta} \qquad \text{(eq. 6)}$$

$$k(t) = \frac{dM}{d\theta} \qquad \text{(eq. 7)}$$

Where M = moment, θ = angular displacement, *k* = stiffness, and *t* = time. In a dynamic system, equation 7 is best suited for describing (quasi) stiffness as it allows for time dependency and freedom from attaining static equilibrium.

As in the linear case, angular stiffness as measured in isolated joints is reported with some simplification. Describing a joint as two segments sharing an axis in one dimension may be useful for modeling. In reality, a joint is made up of many substructures, each contributing to the overall stiffness of the joint. Further, a joint is rarely confined to move solely in one dimension. Lastly, joint stiffness has inertial and viscous components analogous to the linear case. So, equation 7 is a simplified quasi-stiffness in the same regard as equation 3.

Mechanical Stiffness of the Knee Joint

Knee joint stiffness can be examined in any plane or multiple ones. When considered as a contributor to leg spring stiffness, it is most often offered as quasi-stiffness in the sagittal plane (Arampatzis et al., 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002; McMahon & Cheng, 1990; McMahon et al., 1987). Quasi-stiffness is reported as stiffness most often in the sagittal plane even when not part of overall leg spring stiffness and other planes are considered (McFaull & Lamontagne, 1998; Zhang et al., 1998). The following section will attempt to report the tasks through which stiffness is attained and the functional outcome of the stiffness measurement.

Hopping tasks (Farley et al., 1998), running mechanics (McMahon et al., 1987), and effects of stiffness on performance (Gunther & Blickman, 2002) have previously studied dynamically changing stiffness. Few studies have been centered on the aforementioned viscous damping elements of the knee joint and such studies are focused on representing the pure mechanical properties of the knee (McFaull & Lamontagne, 1998; Oatis, 1993; Zhang et al., 1998; Zhang & Wang, 2001). When attempting to apply the practical implications of these mechanical properties, quasi-stiffness would actually be a representation of effective stiffness as an overall property of the knee. This overall behavior is important as it has often been associated with both performance and injury risk (Butler et al., 2003)

This overall joint (quasi) stiffness is a summation of individual structure contributions (ligaments, tendons, and menisci) to the joint behavior in much the same way that leg spring stiffness is a product of individual joint contributions to that overall stiffness. These individual structures have their own inherent stiffnesses (Crowninshield et al., 1976). Further, the joint's stiffness is affected by neuromuscular activity (Latash & Zatsiorsky, 1993).

To quantify the contributions of individual structures, healthy, intact knees can be compared to knees that lack structures such as ACL or have suffered injury to specific structures. Researchers are able to take this approach in vivo (Beynnon, Fleming, Labovitch, & Parsons, 2002; Markolf et al., 1984; Shoemaker & Markolf, 1982). Another approach is to use cadaveric knees and find the contributions of individual structures to overall stiffness by systematic transection (Crowninshield et al., 1976; Gollehon et al., 1987; Hsieh & Walker, 1976; Markolf et al., 1981; Markolf et al., 1976; Nielsen, Rasmussen et al., 1984; Shoemaker & Markolf, 1985). In this way, the theoretical function of each ligament, etc. can be tested. For example, from these studies it has been learned that the ACL contributes heavily to internal rotational stiffness, resistance to hyperextension, and anterior shear force (Hsieh & Walker, 1976). It also assumes more support for valgus and external rotational stiffness once the MCL is damaged and the knee is more flexed (Hsieh & Walker, 1976). The MCL resists both internal and external rotational excursions, but its largest contribution to overall knee joint stiffness is under valgus loads (Markolf et al., 1976; Nielsen, Rasmussen et al., 1984). These types of studies shed light on mechanisms of injury of joint substructures, and also encourage studies that may combine types of load such as Mills' (1991a), which examined the displacement of the knee joint under combined external tibial rotation (on the femur) and valgus displacement in order to study mechanisms and risks of ski injuries.

Methods to Quantify Knee Joint Stiffness

With the task of characterizing knee joint stiffness mathematically, understanding the methods and obtained data of previous work is most useful as background to the application of load to the joint and the measurement of ensuing displacements. Ultimately, the relationship between load and displacement is what underlies (quasi) stiffness. The section will attempt to demonstrate how previous work has attempted to quantify knee joint stiffness.

A common approach is to exert external loads on the joint and measure the resulting displacements (Markolf et al., 1981; Markolf et al., 1978; Markolf et al., 1984; Markolf et al., 1976; Olmstead, Wevers, Bryant, & Gouw, 1986; Uh et al., 2001; Zhang et al., 1998; Zhang & Wang, 2001). This type of stiffness can be derived from any data which includes kinetic and kinematic data for a joint in a given plane. This is often done by direct measurement of load and displacement at the joint (Crowninshield et al., 1976; Louie & Mote Jr., 1987; Markolf et al., 1981; Markolf et al., 1978; Markolf et al., 1984; Markolf et al., 1976; Mills & Hull, 1991b; Olmstead et al., 1986) or separated from overall leg spring stiffness as a joint contribution derived by inverse dynamics (Arampatzis et al., 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002).

It is important to note that this direct measurement of load and displacement is not the only approach to deriving spring stiffness. Other approaches include isolating individual joint work (and then stiffness) as in the leg spring stiffness model (Arampatzis et al., 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002; McMahon & Cheng, 1990; McMahon et al., 1987). Also, reliance upon the relationship between stiffness and frequency of vibration of the joint system has been used to derive stiffness from observed frequencies of motion for both the leg spring (Cavagna, Franzetti, Heglund, & Willems, 1988; Farley et al., 1998; McMahon et al., 1987) and the individual joint (McFaull & Lamontagne, 1998; Oatis, 1993; Zhang et al., 1998; Zhang & Wang, 2001).

A summary of studies that use these varied approaches to characterize knee joint stiffness was published in 2003 (Butler et al.). Considering the various ways to calculate and report stiffness, a commonality among these approaches emerges. Previous work has tended to: 1) simplify the knee joint as two segments hinged as a joint in one dimension (for each displacement considered), and 2) choose a level of detail in the characterization of knee joint stiffness that ranges from overall vertical leg spring stiffness down to the contributions of individual joint structures. Only a limited number of studies considered the fullest of detail expressed in equation 4 (Oatis, 1993; Zhang & Wang, 2001). The vast majority of researchers have seemingly expressed quasi-stiffness in a simplified, but useful, way. The level of detail appears to depend upon the research question and the ability of the researcher to measure the contributors to joint stiffness.

As mentioned, it is evident that in the majority of these biomechanical studies the stiffness reported is actually quasi-stiffness. If a researcher seeks to report stiffness as an overall quality of the knee joint, with the stiffness as a sum of all the implied factors influencing it by structure, neural behavior, and muscular contribution ; then quasi-stiffness appears to be an acceptable approach. Further, If the purpose of a study involves considering the knee's behavior as a hookean torsional spring, quasi-stiffness would be a justified method.

Knee Joint Stiffness in Secondary Planes

As previously described, the knee joint is much more complicated than the one-dimensional hinge model often utilized with the majority of studies on overall leg-spring stiffness considering primarily the sagittal plane (Arampatzis et al., 1999; Dutto & Braun, 2004; Farley et al., 1998; Gunther & Blickman, 2002; McMahon & Cheng, 1990; McMahon et al., 1987). Given the nature of the running and hopping tasks often employed in such studies it is logical to consider that plane first, and often solely. However, varus/valgus and rotary angular stiffnesses exist and can be studied.

At the knee, several authors have isolated varus/valgus stiffness (Crowninshield et al., 1976; Markolf et al., 1981; Mills & Hull, 1991a; Nielsen, Rasmussen et al., 1984; Olmstead et al., 1986; Pope, Johnson, Brown, & Tighe, 1979; Zhang & Wang, 2001) or its inverse, varus/valgus laxity (Gollehon et al., 1987; Grood et al., 1988). Displacement values from these studies range from 1.9°-19.5° total varus/valgus motion (Markolf et al., 1976) (Mills & Hull, 1991a). Comparing reported laxities becomes problematic, though, due to measurements taken at widely varying moment loads and under varied muscular tension, loading, and flexion angle conditions. Stiffness values in these studies also range widely from 2.94 Nm/deg (Bryant & Cooke, 1988) up to 16.5 Nm/deg (Markolf et al., 1976) of varus/valgus stiffness. Again, there are a large variety of loading, tension, and flexion conditions.

In the transverse plane, attempts to quantify internal and external rotational stiffness have also been made (Crowninshield et al., 1976; Hsu et al., 2006; Louie & Mote Jr., 1987; Markolf et al., 1981; Markolf et al., 1984; Markolf et al., 1976; Mills & Hull, 1991a; Zhang & Wang, 2001). The same gamut of results is seen, presumably from vastly assorted loading, flexion, and tension conditions – these influences are discussed below. Displacements range from 6.3° up to 61.7° of combined internal/external rotation (Louie & Mote Jr., 1987) (Wang & Walker, 1974). Stiffness ranges from 0.13 Nm/deg (Markolf et al., 1981) up to 2.54 Nm/deg (Louie & Mote Jr., 1987) in internal/external rotation. These wide ranges of stiffnesses and displacements suggest that the joint response to loading is dependent upon a wide variety of factors. The following section will attempt to address some of these factors.

Influences on Knee Joint Stiffness

In these secondary planes of motion, knee flexion angle is no longer part of the measured stiffness, but must be considered in the derivation of stiffness. Differences in varus/valgus and internal/external rotational stiffness are seen at different angles of flexion (Crowninshield et al., 1976; Markolf et al., 1981; Markolf et al., 1984; Markolf et al., 1976; Mills & Hull, 1991a; Nielsen, Rasmussen et al., 1984). A general trend revealed by these studies is one of increased laxity and decreased stiffness in varus/valgus and internal/external rotation as the knee is flexed. This research supports that the knee is least lax and most stiff in secondary planes when the knee is fully extended.

Multiple attempts have been made to account for neuromuscular effects on stiffness in several studies by controlling for muscle tension and relaxation (Louie & Mote Jr., 1987; Markolf et al., 1978; McFaull & Lamontagne, 1998; Olmstead et al., 1986; Pope et al., 1979; Shoemaker & Markolf, 1982; Zhang et al., 1998; Zhang & Wang, 2001). This is often done by instruction to participants and the monitoring of muscle activation where possible by EMG. Muscular activation increases spring stiffness of the joint in all planes, and therefore affects overall joint spring stiffness (Markolf et al., 1978; Olmstead et al., 1986; Zhang et al., 1998; Zhang & Wang, 2001).

Another important consideration is the effect of weightbearing on joint stiffness. Several researchers have included this effect and found the joint to be "stiffer" when bearing weight, from an anterior/posterior laxity standpoint, as well as in varus/valgus and internal/external rotation (Beynnon et al., 2002; Markolf et al., 1981; Uh et al., 2001). This increased stiffness with weightbearing is probably associated with the bony anatomy of the knee. Markolf (1976), for example, proposed that a stiffness decrease is seen at epicondylar liftoff in varus/valgus displacement.

Laxity and stiffness

Even where laxity is the variable of interest, it is inextricably tied to stiffness. A continuous moment by displacement curve really represents all infinitesimal differentials of laxity. Though laxity is not the focus of the present investigation, it can be considered to be related to stiffness (Markolf et al., 1981; Markolf et al., 1978; Markolf et al., 1984; Markolf et al., 1976). As noted earlier, it is not strictly the inverse of stiffness, however, as laxity is typically measured once the entire load is exerted (Uh et al., 2001), and stiffness as measured in this study is more incremental along the loading path, allowing the possible detection of changes in stiffness during loading.

For some decades, researchers have used varied approaches to determining knee joint stiffness. In their course, the effects of musculature, weightbearing, and the contributions of individual structures within the joint have been addressed in varied ways. A tabular overview of stiffness reporting follows in table 1.

Table 1: Existing stiffness literature:

Year	Author	n	Int/ext? Var/val?	Cadaveric? In vivo?	Passive? Weightbearing?	Moment(s) used to load	Results	Notes
1974	Wang & Walker	27	Int/ext	Cadaveric	Passive – sim. weightbearing	0-250 kgf-cm (=0-24.5 Nm)	33.43 tot I-E degrees @ 50 kgf-cm; 48.27 @ 125 kgf-cm; 61.74 @ 250 kgf-cm	Just laxity reported up to 5 kgf-cm, secondary measured beyond. 25 degrees knee flexion.
1976	Crowninshield, et al	6	Both	Cadaveric	Passive, NWB	unknown	Stiffness reported as ratio (0 - 1) to stiffness at full extension.	Comparison of model to cadaver samples and comparison of intact to serially transected knees. 0-90 degrees knee flexion.
1976	Markolf, et al.	35	both	cadaveric	Passive, NWB	0-29 Nm in var/val; 0-8 Nm in int/ext.	11.0 -14.0/16.5 Nm/deg @ 0 deg, 0.7 - 12.4/9.6 Nm/deg @ 135 deg for Var/val; 0.7 - 2.3/2.5 Nm/deg @ 0 deg, 0.1 - 2.5/2.3 Nm/deg for int/ext.	Breakpoints created by intersection of terminal stiffness slopes and neutral stiffness at origin. 0-135 degrees knee flexion.
1976	Hsieh & Walker	8	Int/ext	cadaveric	Passive – sim. weightbearing	0 - 50 kg-cm (4.90 Nm)	21.6/11.0 deg @ 0 deg flexion unloaded/loaded in int/ext. 26.0/12.1 deg @ 30 deg flexion unloaded/loaded in int/ext.	Laxity reported as opposed to true stiffness. Examines role of joint (axial) load in stability. 0, 30 deg knee flexion.
1978	Markolf, et al.	28m 21 f	Var/val	In vivo	Passive and tensive, NWB	0- 20 Nm	7.0 - 8.3 Nm/deg at full extension in var/val	Compares in vivo results to previous cadaveric study. Effect of muscle tension considered. Left/right differences found in individuals - but no trend for one or other side as group. 0 deg knee flexion.
1981	Markolf, et al.	4 m 4 f	both	cadaveric	Passive – axial load of 925 N	0 - 10Nm in IE, 0 - 20 Nm in V/V	.31/.34 Nm/deg unloaded/loaded in IE at 0 deg; .13/.18 Nm/deg unloaded/loaded in IE at 20 deg. Only qualitatively given in var/val.	Markolf's breakpoint analysis of moment by displacement curves. Examines role of joint load in stability/stiffness. 0, 20 deg knee flexion.
1982	Shoemaker & Markolf	20 m	Int/ext	In vivo	Tensive, NWB	0-10 Nm	33.0/41.0 deg tot IE rot @ 20 deg knee flexion (hips 10/90 deg flexed); 47.0/47.0 deg tot IE rot @ 90 deg knee flexion (hips 10/90 deg flexed)	Laxity reported. Examines effect of musculature. Also examines rotation of foot during the same excursions of tibia. 20, 90 degrees knee flexion.
1984	Nielsen, et al.	20	both	cadaveric	Passive, NWB	0-3 Nm	Unspecific for total laxity - reported only as differences in laxity created by selectively	Laxity reported. Examines stability changes with transaction of ind. Structures. 30, 70 deg knee flexion.

1984	Markolf, et al.	35	both	In vivo	Passive, NWB	0 - 20 Nm var/val; 0 - 10 Nm Int/ext.	Too extensive for this table - laxities at 20 Nm and 10 Nm reported for v-v and I-E respectively. Stiffness as instantaneous value at those points	Stiffness not reported as in other Markolf studies - in this case it is a comparison at the max. used torque of laxity and instantaneous stiffness of healthy and ACL - def. knees. 0, 20 deg knee flexion.
1985	Shoemaker & Markolf	7	Int/ext	Cadaveric	Passive, WB	0-10 Nm	23.8 deg, 20.0 deg @ 0 deg flexion for unloaded, loaded conditions. 37.8 deg, 35.0 deg @ 20 deg flexion for same conditions.	Laxity reported. Comparison between intact knees and knees with serially transected ligaments. 0, 20 deg knee flexion.
1986	Olmstead, et al.	5	Var/val	In vivo	Passive and tensive, NWB	0-20 Nm	Reports percent changes in initial and terminal stiffness with muscle tension.	No hard numbers – just percent changes with muscle activation. 0 deg knee flexion.
1987	Gollehon, et al.	17	both	cadaveric	Passive, NWB	int/ext: 0-6 Nm. Var/val: 0-15 Nm	5.5-10.5 deg val from 0-90 deg flexion. About 6-8 deg var from 0-90 deg flexion. About 15-24 deg int from 0-45 deg	Laxity reported. Examines effect of different flexion angle. 0-90 degrees knee flexion.
1987	Louie & Mote	4 m	Int/ext	In vivo	Passive, tensive, NWB	about 14 - 42 Nm - highly variable.	42.6 total degrees to 6.3 total degrees in relaxed to tensed conditions.	Laxity reported. Highly variable loading. Examines effect of muscle activation on laxity. 10, 90 deg knee flexion.
1988	Bryant & Cooke	24 m 17 f	Var/val	In vivo	Passive, NWB	0-20 Nm	3.5 Nm/deg val and 2.94 Nm/deg var (max values)	Attempts to standardize stiffness and laxity measures for the knee joint, as well as make some comparisons between males and females. 1-2 deg knee flexion.
1988	Grood, et al.	15	both	cadaveric	Passive, NWB	0-5 Nm int/ext, 0-20 Nm var/val.	about 10-30 deg int, 5-12 deg ext from 0-90 deg flexion. About 3-10 deg var, 3-6 deg val from 0-90 deg flexion.	Laxity reported. Very generally reported laxity in each direction - specifically reports changes in laxity with serial transaction and change in flexion angle. 0-90 deg knee flexion.
1991	Mills & Hull	6 m	both	In vivo	Passive, NWB	0-60 Nm for var/val, 0-20 Nm for int/ext.	13.2 - 19.5 deg total var/val at 15 - 60 deg flexion. 43.2 - 46.7 deg total int/ext at 15-	Laxity reported. Uses terms "terminal rotation" and "flexibility" to denote total

1991	Mills & Hull	3	both	In vivo	Passive, NWB	0-20Nm int/ext, 0-60 Nm var/val	8.5-14.0 deg total var/val depending on constraint of foot, about 40.9 - 50.8 tot deg int/ext depending on constraint.	Laxity reported. Primarily a test of apparatus and of effect of foot constraint on results. Only a small pilot-size sample. 15, 60 deg knee flexion.
2001	Zhang & Wang	8 m	Var/val	In vivo	Passive, tensive, NWB	0 - 20 Nm	285.3 Nm/rad - 452.2 Nm/rad in relaxed and tensed states. (4.9 Nm/deg - 7.9 Nm/deg)	Examines dynamic stiffness change with muscle tension specifically to oppose intended motion. 0 deg knee flexion.
2006	Hsu, et al.	44 m 38 f	Int/ext	cadaveric	Passive, NWB	0-5 Nm int/ext concurrent with 10 Nm var/val	0.79 Nm/deg, 1.06 Nm/deg for fem, male @ 15 deg flexion; .85 Nm/deg, 1.03 Nm/deg for fem, male @ 30 deg flexion	Comparative of females/males in stiffness and laxity. Curve analysis built upon Markolf model. 15, 30 deg knee flexion.

Summary

The focus of this literature review is on the definition, the biomechanical background, and the mathematical interpretation of knee joint stiffness. The definition of knee joint stiffness is apparently simple, but often carries assumptions that must be understood. There are multiple approaches to measuring knee joint spring stiffness. Which approach is best depends entirely on the research question at hand. Similarly determined is whether to report true stiffness or quasi-stiffness.

As many researchers examine laxity in lieu of stiffness, it is important to understand that the two concepts are inextricably linked, both being relationships of load and displacement. Again, the preference of one to the other is a matter of research question and purpose.

The literature provides a foundation for understanding that the outcome stiffness measured is truly a combined effect of substructures with their accompanying inertial and viscous complications. The literature underscores the large influence of neuromuscular function and axial loading on the knee joint displacements under a load, giving justification to minimizing the effects of musculature in finding stiffness as an inherent mechanical property of the knee joint. Some points not addressed in the literature include the addressing of gender differences for stiffness in varus/valgus and internal/external rotation, and the day to day consistency of stiffness measurement.

CHAPTER III

METHODS

Subjects

Twenty subjects (10 men and 10 women, age = 25.1 ± 3.4, height = 173.1 ± 8.0 cm, mass = 73.6 ± 12.0 kg) were recruited. Inclusion criteria included no history of knee ligament injury or surgery, no history of injury or chronic pain in both lower extremities for the past 6 months, and otherwise healthy. Subjects read and signed a consent form that had been approved by the Institutional Research Board at UNCG prior to the actual data collection. Data were collected for both limbs. In order to establish the reliability of the data, the first 10 subjects were asked to participate in a second identical data collection session within 24 to 48 hours. All procedures and instrumentations are identical to those found in Shultz, et al (2006, in review).

Instrumentation

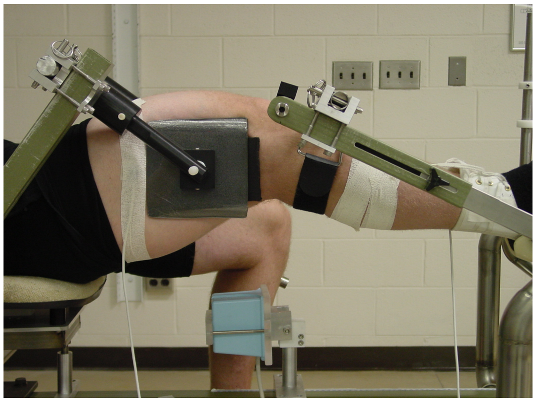
The Vermont Knee Laxity Device (VKLD) was used to measure the amount of varus/valgus and internal/external rotational stiffness. This device was developed at the University of Vermont and allows kinematic measurement in the coronal, sagittal, and transverse planes while simulating non-weight bearing and weight bearing conditions in a supine position. Details of the VKLD have been described in previous studies (Uh et al., 2001).

Each subject was positioned in the VKLD in a supine position with the foot secured in the foot plate, and the hip in 10° of flexion. In this position, the greater trochanter and the lateral malleolus were aligned to the rotational axis of the VKLD counter weight lever arm. Each swingarm for the thigh and shank were attached to the corresponding segment, and counterweights were used to support the tare weight of each segment.

When simulating the weight bearing condition, 40% of body weight, which was connected to the foot cradle by a pulley system, pulled the foot cradle superiorly. A shoulder device was adjusted to fit snugly to the subject to prevent superior movement during the weight bearing condition.

In order to minimize the femur movement in varus/valgus and internal/external rotational tests when applying torques to the knee, the thigh was clamped as tight as possible without causing any pain. The shank was loosely strapped to the proximal tibia which is connected to the swingarm for the shank and moved freely laterally and medially.

A 3D electromagnetic tracking system (Ascension MiniBird Hardware, Ascension Technology, Burlington, VT, Motion Monitor Software, Innovative Sports Training, Chicago, IL) was used to obtain 3D position and orientation of each segment. Two motion sensors were attached to the lateral aspect of the thigh just proximal to the clamping plate (along the iliotibial band) and the tibial shaft just distal to the shank strap (see fig. 1). Extensive pilot work showed that these placements resulted in minimum skin movement during application of torque to the knee joint, and it is believed that these sensor placements best represent the movements of the femur and tibia in this study.



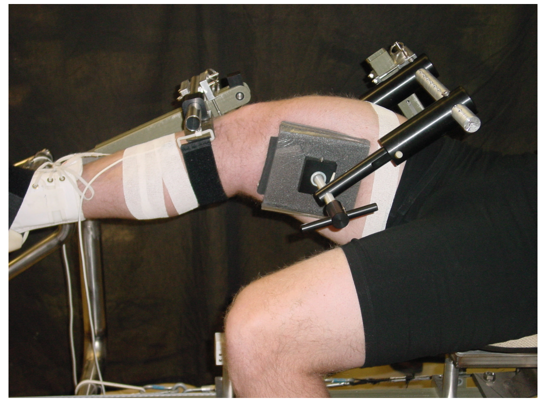


Fig. 1a: Lateral view of femoral clamp

Fig 1b: Medial view of clamp

For valgus-varus torque application to the knee joint, a hand held force transducer (Model SM-50, Interface, Scottsdale, AZ) was used to push the distal tibia while simultaneously recording the applied force. For subject comfort, a concave molded orthoplast attachment was attached to the force transducer to increase the contact area between the tibia and the force transducer (see fig. 2). The signal from the force transducer was amplified through a strain gauge transducer (Model 9820, Interface Advanced Force Measurement, AZ) and delivered to the personal computer.

Rotational torque to the knee joint was applied through a T-handle connected to a six degree-of-freedom (6DOF) loadforce transducer (Model MC3A, Advanced Medical Technology, Inc; Watertown, MA), firmly fixed to the foot plate. The foot was secured in the foot plate such that if an examiner rotated the T handle, the tibia was rotated with an equal torque. When aligning the 6DOF forceload transducer, care was taken to ensure the tibia and the 6DOF load transducer were aligned so that the torque about the vertical axis of the load transducer and the torque about the longitudinal axis of the tibia were the same. Signals from the 6DOF load transducer was delivered to and stored in the personal computer for further analysis.

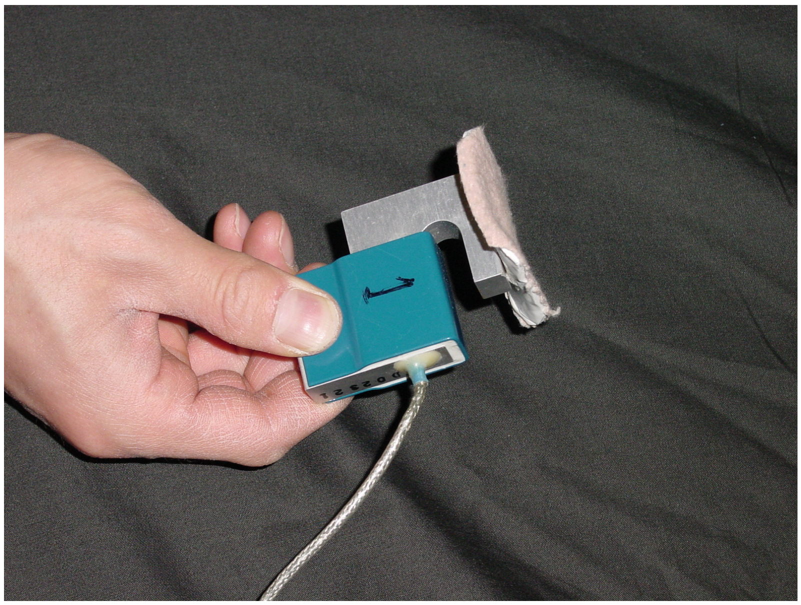


Fig. 2: Handheld force transducer

Procedure

The limb of interest was positioned in the VKLD with the foot strapped to the foot plate, and the anatomical flexion axes of the ankle and hip joints aligned with the mechanical axes of rotation of the VKLD counterweight system. The subject's foot was fitted with an ankle brace and secured in the foot plate with any excess space filled with additional padding. Once properly positioned in the VKLD with the foot cradle in the locked position, the thigh and leg counterweights were applied. The thigh clamp was observed for movement during this process to ensure the thigh was sufficiently secured as the counterweight was applied.

The joint coordinate system for the lower extremity was constructed using the following procedures. The center of rotation of the knee and ankle joint centers were estimated using the centroid method that calculated the midpoint between the medial and lateral epicondyles of the femur and the medial and lateral malleoli, respectively. The hip joint center was also estimated using the centroid method, calculating the midpoint of a line defined anteriorly by a point placed medially from the ASIS-greater trochanter midpoint at a distance equal to half the ASIS-greater trochanter line distance, and defined posteriorly by a point placed posteriorly from the ASIS-greater trochanter midpoint at a distance equal to the ASIS-greater trochanter line distance.

To construct the segment coordinate system for each segment, the following landmarks were digitized: greater trochanter and lateral and medial femoral epicondyles for the femur, and the most medial and lateral parts of the tibial plateau and, the medial malleolus for the shank, and most lateral part of the tibial plateau for the shank were digitized in order to construct the segment coordinate system for each segment. The y axes of the segment coordinate systems for the thigh and shank were parallel to the lines between the greater trochanter and lateral femoral epicondyle and between the most medial part of the tibial plateau and medial malleolus respectively. Z axes for thigh and femur were parallel to the lines between the lateral and medial femoral epicondyles and between the most medial and lateral part of the tibial plateau, respectively. X axes for each segment were perpendicular to both y and z axes of the segment coordinate system for each joint. Pilot data supported the use of the digitization method to compute the hip joint center, and provided valid position data for knee flexion angle as verified by a standard, hand held goniometer. Following digitization of joint centers, the ankle and knee were flexed to 90° and 20° respectively, and the subjects were instructed to relax their leg muscles.

Varus/Valgus Loading

Varus/Valgus stiffness was measured in the non-weight bearing condition, and then Internal/external rotational stiffness was measured in the non-weight bearing and weight bearing condition. The order of leg to be measured (right or left side) across the subjects was counterbalanced. To obtain not just displacement data from the neutral position but also the total displacement in both directions, one cycle of torque application consisted of both directions (i.e.

varus –valgus rotation or internal-external rotation). Therefore, the order of the direction for torque applications was also counterbalanced across the subjects.

Each subject was positioned in their own neutral position for the varus/valgus testing: Axial rotation (of the footplate) was first locked with the second metatarsal visually aligned perpendicular to the horizontal plane. The foot plate and foot cradle were then unlocked to allow free movement in both the coronal and sagittal planes, respectively. Subjects were then asked to straighten their knee and relax. Finally the knee flexion angle was adjusted to 20° and the foot cradle was locked in this position. Knee flexion angle was confirmed (within $\pm 5^{o}$) with both a hand held goniometer and real time knee flexion angle data obtained from the motion sensors.

Ten Nm of external varus and valgus torque were applied to the knee joint by applying force to the medial and lateral aspect of distal tibia. The same examiner always applied force manually to the tibia using a hand held force transducer (see fig. 3). While applying the force to the tibia, great caution was taken so that the direction of the force vector was always directed perpendicular to the long axis of the tibia. The amount of force applied to the tibia to create 10 Nm of torque at the knee joint was determined based on a lever arm distance of 0.34 m from the axis of the knee joint.

Tibiofemoral positional data were recorded to determine the initial varus-valgus angle and confirmed (within $\pm 5^{o}$) with clinical measurement. Subjects were instructed to relax all muscles during the Varus/valgus loading. Prior to data collection, the hand held force transducer was calibrated and a series of three alternating V-V loads were applied to the knee to familiarize subjects with the procedure and to ensure that the subject could tolerate the 10 Nm of V-V loading. Following the familiarization process, data were collected on three separate trials, consisting of a single cycle of V-V loading. While the examiner applied the force to the distal tibia, another examiner always monitored the amount of force applied using a real-time oscilloscope and indicated the point where the force created 10Nm of torque at the knee joint, at which time, loading ceased. The limb was returned to the neutral position between each trial by following the same procedure previously described.

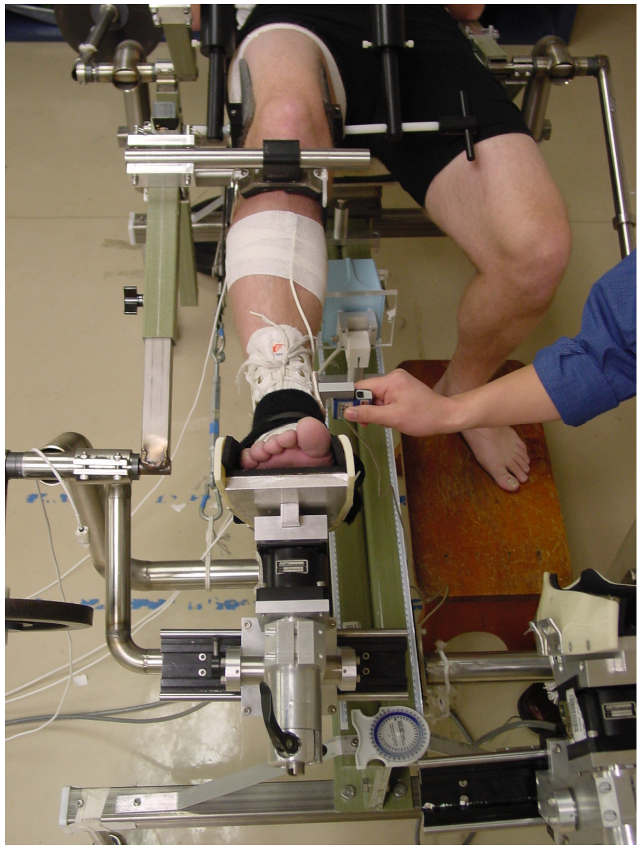


Fig. 3: Varus/valgus loading with handheld transducer

Internal/External Rotational Loading

Neutral position of the limb was achieved following a similar procedure previously described. Subjects were asked to straighten their knee and return to a relaxed position. For this test, the 6DOF foot plate was also unlocked, allowing axial rotation of the tibia in the transverse plane. Once subjects were adjusted to 20º of knee flexion, coronal and sagittal plane movements were locked. A series of three alternating internal-external (I-E) torque, about the longitudinal axis of the tibia, was applied to the knee to familiarize subjects with the procedure and to ensure that subjects could tolerate the 5 Nm of I-E torque. The amount of torque was monitored with the use of a real-time oscilloscope in the same manner described above, and the I-E torque was released as soon as the amount of torque reached 5 Nm. Following the familiarization process, data were collected for three separate trials, consisting of a single cycle of I-E loading. The limb was returned to the neutral position between each trial by following the same procedure previously described (see fig. 4).

Internal/External Rotational Loading during Weightbearing

For measurement of rotational stiffness during weight bearing conditions, the same procedure was repeated as above with an additional compressive force equal to 40% of the subject's bodyweight. To position the subject's knee in a neutral position, the following procedures were performed: 1) the foot cradle and the 6DOF foot plate were unlocked to allow movement of the shank in all coronal, transverse, and sagittal plane motions, 2) subjects were asked to actively straighten their knee and relax, 3) 40% of the body weight was gradually released by the examiner holding the foot cradle, and subjects were asked to push the foot plate and accept the weight, 4) subjects were instructed to flex or extend the knee to achieve a knee angle of 20º of flexion using a real-time goniometer, and 5) the medial-lateral movement of the foot plate was locked in this position. The subjects were instructed to maintain the position while applying the rotational torque. A compressive force equal to 40% of bodyweight was chosen to simulate the loading condition experienced during double leg stance (assuming 50% of bodyweight applied to each leg, and 10% of bodyweight distributed below the knee).

The compressive force was applied by unlocking the foot cradle allowing movement in the sagittal plane. While the subjects maintained 20 º of knee flexion angle, the same procedures were made for both the familiarizing trial and actual data collection trials. As in the previous measurements, neutral position of the knee was achieved every time before each trial. Subjects were instructed to maintain the same knee position (20^{o} knee flexion) upon joint loading while 3 separate trials of I-E torques were applied to the knee.

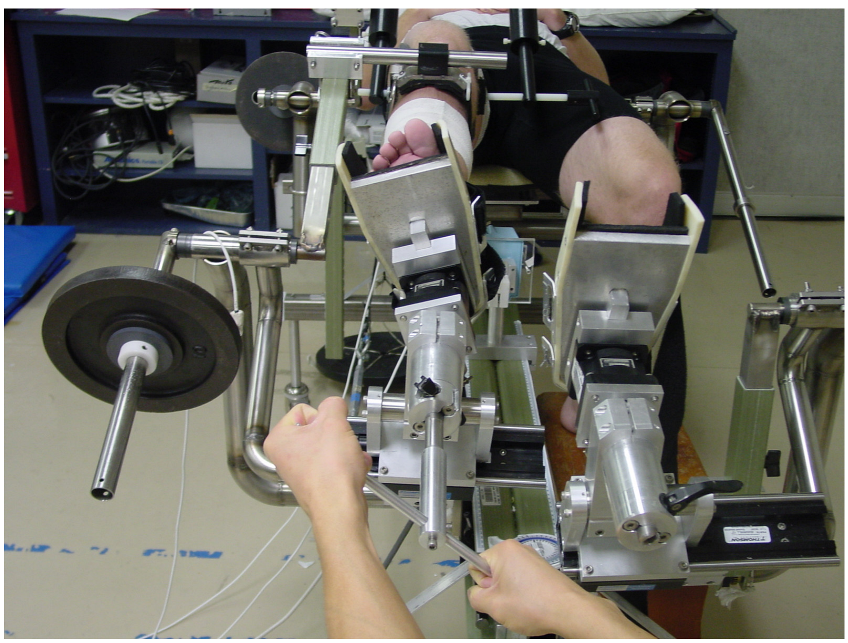


Fig. 4: Internal/external rotational torque applied by t-bar

Data Acquisition

Position data were collected at 100 Hz using the electromagnetic tracking system. The signals from the motion sensors and both hand and 6DOF force transducers were filtered using a low-pass filter at 10 Hz and 20Hz using a 4th order zero lag Butterworth filter, respectively.

The y axes of the segment coordinate system for the thigh and shank were directed superiorly along the longitudinal axis of the thigh and shank, and were parallel to the lines between the greater trochanter and lateral femoral epicondyle and between the most medial part of the tibial plateau and medial malleolus, respectively. Z axes for thigh and shank were directed laterally for the right leg and medially for the left leg, and were parallel to the lines between the lateral and medial femoral epicondyles and between the most medial and lateral part of the tibial plateau, respectively. X axes for each segment were perpendicular to both y and z axes of the segment coordinate system for each joint and directed anteriorly. To obtain the knee joint angle, A segmental reference system quantified the three dimensional kinematics of the knee during the transition from non-weight bearing to weight bearing. For each segment the +Z axis was directed laterally, the +Y axis was directed superiorly, and the +X axis was directed anteriorly. Euler's equations were used to describe joint motion about the knee with a rotational sequence of Z Y' X" (Kadaba et al., 1989).

Data Reduction

To form the moment by displacement curve as described by Shoemaker & Markolf (1982), displacements at each 0.1 Nm were plotted. The displacements for each of three trials in each condition were averaged at each 0.1 Nm increment to give one curve. This yields a representative torque by displacement curve for each subject, as well as group average curves for males, females, and all subjects.

Each curve was then broken into sections of 0.5 Nm load increase. For each section, an incremental stiffness was derived in a traditional hookean way:

$$k = \frac{-\Delta M}{\Delta \theta} \qquad \text{(eq. 6)}$$

This resulted in 20 incremental stiffnesses for the valgus and varus rotational loadings and 10 incremental stiffnesses for the internal and external rotational loadings. If an individual subject's incremental stiffnesses were negative, the data were eliminated from statistical analyses. These considerations resulted in the elimination of 1 male and 1 female knee for valgus loading (N=38 total knees), 3 male and 2 female knees for varus loading (N=35 total knees), 2 male and 2 female knees for internal rotational loading – 3 male, 2 female knees in weightbearing - (N=36 total knees, 35 total knees respectively), and 2 male and 2 female knees for external rotational loading – both weightbearing and non-weightbearing (N=36 total knees).

Additionally, average stiffnesses at the start of displacement for males and females were seen to be negative and/or unrealistically large (10 – 30 X the averages at all other increments). This problem was addressed by considering the loading cycle from a point immediately following the unrealistic stiffnesses. For Varus and Valgus loading, data were considered from 1.5 - 10.0 Nm load. For Internal and External rotational loading, data were considered from 0.5 - 5.0 Nm load.

Data Analyses

Objective 1

To ascertain the between day measurement consistency of incrementally derived stiffness for the first 10 subjects, separate repeated measures ANOVA for each direction and condition (valgus, varus, internal rotation (WB and NWB), and external rotation (WB and NWB),) were used to calculate intraclass correlation coefficients ($ICC_{2,k}$) and standard errors of measurement (SEM).

Objective 2

Graphs were created for varus, valgus, internal rotational (weightbearing and non-weightbearing), and external rotational (weightbearing and non-weightbearing) loading for the entire sample of normal knees. They were averaged in the manner described above and graphed with an envelope of standard deviation around them.

This was repeated for male and female knees in order to create a representative graph for males and females that also included the envelope of standard deviation. This allowed qualitative comparison between male and female knee torque by displacement (Hsu et al., 2006).

Objective 3:

Six separate repeated measures ANOVAs were performed for the sets of valgus, varus, internal, and external incremental rotation (WB & NWB) stiffness constants, with sex as the between factor, and the successively measured stiffnesses as the within factor. The alpha level was set a priori at $P<0.05$ and Tukey's Post hoc testing was used to identify differences.

CHAPTER IV

RESULTS

Objective 1

Valgus

ICC values for day to day valgus stiffness measurement ranged from .39 to .92 with an average ICC of .72 (table 2). 12 of the 17 constants were .68 or higher. For details of trial, between, and error variances for all loading conditions, see appendix G.

Table 2: ICC 2,k and SEM for valgus stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
1.50	2.99	±	1.69	2.63	±	1.79	0.82	0.75
2.00	2.49	±	1.60	2.10	±	1.31	0.80	0.71
2.50	2.27	±	1.45	1.91	±	1.09	0.65	0.86
3.00	2.12	±	1.16	1.64	±	0.76	0.59	0.74
3.50	2.18	±	1.47	1.54	±	0.58	0.40	1.14
4.00	1.81	±	1.05	1.84	±	1.02	0.87	0.38
4.50	1.76	±	0.73	1.60	±	0.67	0.55	0.49
5.00	1.67	±	0.50	1.67	±	0.67	0.74	0.34
5.50	1.82	±	0.88	1.75	±	0.58	0.71	0.48
6.00	1.80	±	0.80	1.64	±	0.52	0.68	0.45
6.50	1.61	±	0.66	1.74	±	0.61	0.79	0.29
7.00	1.49	±	0.57	1.90	±	0.78	0.74	0.39
7.50	1.92	±	0.96	2.02	±	0.91	0.78	0.45
8.00	1.97	±	1.07	2.18	±	1.36	0.92	0.38
8.50	2.02	±	1.11	2.39	±	1.28	0.78	0.60
9.00	1.93	±	1.01	2.67	±	1.51	0.70	0.82
9.50	2.59	±	2.38	2.82	±	1.62	0.66	1.39

Varus

ICC values in day to day varus stiffness measurement ranged from -.48 to .87 with an average value of .45 (table 3). Values from .56 to .87 occurred up to 5 Nm of applied torque. The reliability of these initial stiffness ranges is important as many of the later described sex differences of objective 3 exist in the early loading ranges. For further detail, including day 1 and 2 means with standard deviations and the variances used in calculations, see appendix G.

Table 3: ICC 2,k and SEM for varus stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
1.50	2.50	±	1.47	2.67	±	1.61	0.82	0.68
2.00	2.35	±	2.04	2.53	±	1.17	0.56	1.34
2.50	2.14	±	0.89	1.62	±	0.62	0.74	0.46
3.00	2.03	±	0.84	1.74	±	0.94	0.87	0.34
3.50	2.11	±	1.02	1.76	±	0.50	0.72	0.54
4.00	1.98	±	0.76	1.66	±	0.54	0.63	0.46
4.50	1.94	±	0.65	1.69	±	0.69	0.67	0.39
5.00	2.19	±	0.97	1.55	±	0.54	0.72	0.52
5.50	2.23	±	0.88	1.86	±	0.78	0.64	0.53
6.00	2.06	±	0.73	1.69	±	0.58	0.00	0.73
6.50	2.09	±	0.84	1.75	±	0.70	-0.48	1.02
7.00	2.30	±	0.86	2.00	±	0.88	-0.06	0.90
7.50	1.96	±	0.57	1.89	±	0.87	0.06	0.84
8.00	2.20	±	0.88	1.76	±	0.30	-0.14	0.94
8.50	2.39	±	0.80	2.12	±	0.77	0.83	0.33
9.00	2.26	±	0.87	1.81	±	0.66	0.83	0.36
9.50	2.88	±	1.38	1.81	±	0.47	0.33	1.13

Internal Rotation

ICC values for internal rotation ranged from -.65 to .91 (average = .43). Initial stiffness ICC values range from .72 to .91 (table 4) through 2.0 Nm of applied torque. This range of reliability corresponds to where male female differences are later reported and discussed. For further detail, including day 1 and 2 means with standard deviations and the variances used in calculations, see appendix G.

Table 4: ICC 2,k and SEM for IR stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
0.50	0.35	±	0.23	0.38	±	0.18	0.88	0.69
1.00	0.34	±	0.16	0.41	±	0.19	0.85	0.65
1.50	0.32	±	0.11	0.40	±	0.16	0.72	0.56
2.00	0.40	±	0.14	0.43	±	0.17	0.91	0.68
2.50	0.44	±	0.13	0.58	±	0.20	0.15	0.09
3.00	0.52	±	0.13	0.68	±	0.52	0.41	0.24
3.50	0.67	±	0.43	0.51	±	0.09	-0.65	-0.45
4.00	0.64	±	0.11	0.68	±	0.29	0.42	0.24
4.50	0.63	±	0.10	0.62	±	0.18	0.18	0.11

External Rotation

ICC values for external rotation were higher than internal rotation overall (average = .64), but were lower at later levels of incrementally measured stiffness (table 5). Reliability in initial stiffnesses remained at or above .74 through 2.5 Nm of applied torque. For further detail, including day 1 and 2 means with standard deviations and the variances used in calculations, see appendix G.

Table 5: ICC 2,k and SEM for ER stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
0.50	0.45	±	0.35	0.34	±	0.22	0.86	0.69
1.00	0.37	±	0.22	0.26	±	0.12	0.77	0.68
1.50	0.33	±	0.15	0.30	±	0.10	0.90	0.83
2.00	0.36	±	0.18	0.31	±	0.10	0.74	0.67
2.50	0.37	±	0.14	0.34	±	0.12	0.77	0.71
3.00	0.44	±	0.12	0.35	±	0.09	0.42	0.39
3.50	0.45	±	0.12	0.38	±	0.11	0.61	0.57
4.00	0.49	±	0.14	0.45	±	0.12	0.39	0.36
4.50	0.51	±	0.10	0.52	±	0.16	0.26	0.25

Internal Rotation in Weightbearing

ICC values for weightbearing internal rotation ranged from -.14 to .92 (average=.40), and were highest up to 2.0 Nm of applied torque (table 6). This range corresponds to where male female differences are later reported and discussed. For further detail, including day 1 and 2 means with standard deviations and the variances used in calculations, see appendix G.

Table 6: ICC 2,k and SEM for IRWB stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
0.50	1.56	±	1.15	1.24	±	0.96	0.92	0.33
1.00	1.31	±	0.72	1.22	±	0.60	0.86	0.27
1.50	1.39	±	0.52	1.53	±	0.81	0.60	0.51
2.00	1.03	±	0.35	1.47	±	0.63	0.72	0.33
2.50	1.13	±	0.34	1.55	±	0.86	-0.07	0.89
3.00	1.19	±	0.52	1.98	±	1.39	0.49	0.99
3.50	1.54	±	0.46	2.69	±	3.26	0.27	2.78
4.00	1.39	±	0.36	2.03	±	1.26	-0.07	1.30
4.50	1.56	±	0.52	2.35	±	1.02	-0.14	1.09

External Rotation in weightbearing

Overall, ICC values for external weightbearing rotation averaged .77, with eight of nine values exceeding .70 (table 7). Weightbearing external rotation and valgus rotation had the highest overall reliability from day to day. Again, for further detail, including day 1 and 2 means with standard deviations and the variances used in calculations, see appendix G.

Table 7: ICC 2,k and SEM for ERWB stiffness measures day to day.

Torque (Nm)	Day 1 Mean (Nm/deg)		SD (Nm/deg)	Day 2 Mean (Nm/deg)		SD (Nm/deg)	ICC 2,K	SEM (Nm/deg)
0.50	1.93	±	1.78	1.17	±	0.98	0.81	0.77
1.00	1.49	±	1.34	1.02	±	0.67	0.76	0.65
1.50	1.50	±	1.25	0.90	±	0.54	0.76	0.61
2.00	1.07	±	0.71	0.86	±	0.45	0.88	0.24
2.50	1.13	±	0.94	0.89	±	0.44	0.70	0.51
3.00	0.96	±	0.45	0.84	±	0.37	0.90	0.14
3.50	1.21	±	0.80	0.90	±	0.28	0.55	0.54
4.00	1.05	±	0.44	0.87	±	0.28	0.76	0.22
4.50	0.95	±	0.28	0.98	±	0.41	0.81	0.18

Objective 2

Displacement curves are presented below with envelopes of standard deviation. A brief period of steep moment by displacement slope (which would indicate a high stiffness) exists in several of the graphs. Specifically, All the valgus, varus, internal rotation, external rotation (though it is less-pronounced), and male internal weight-bearing and external weight-bearing feature this brief period of higher stiffness.

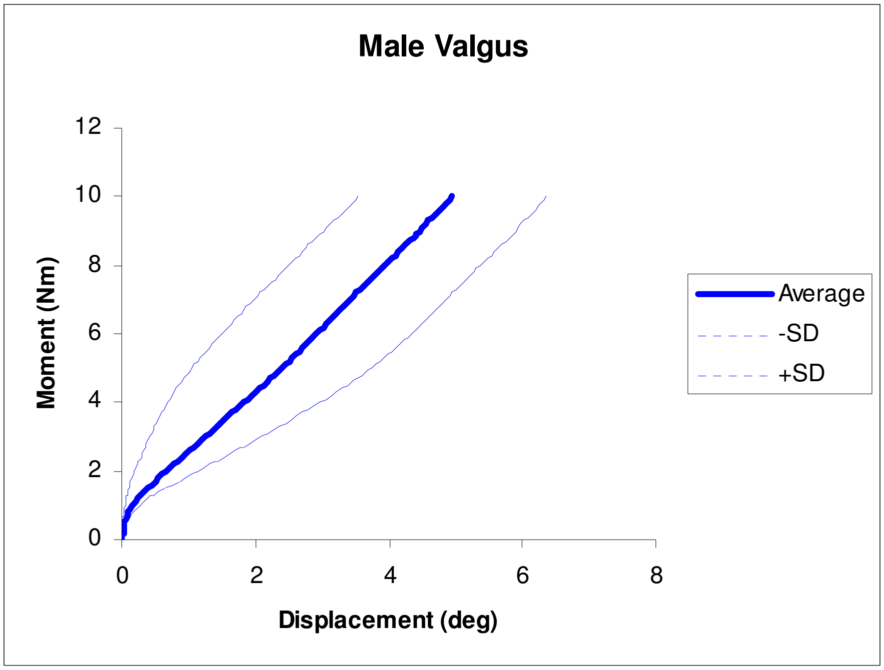


Fig. 5: Male valgus torque by displacement curve

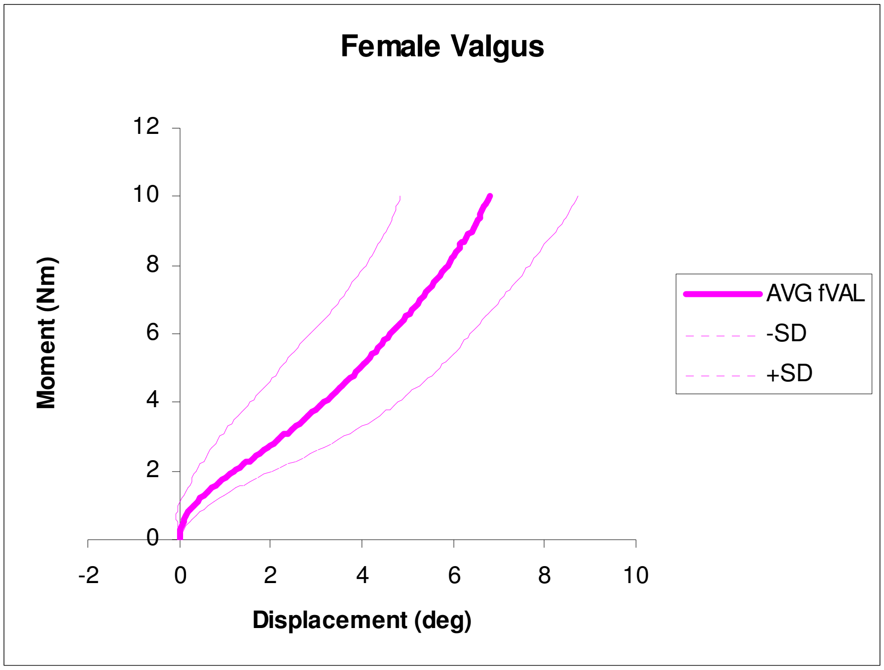


Fig 6: Female valgus torque by displacement curve

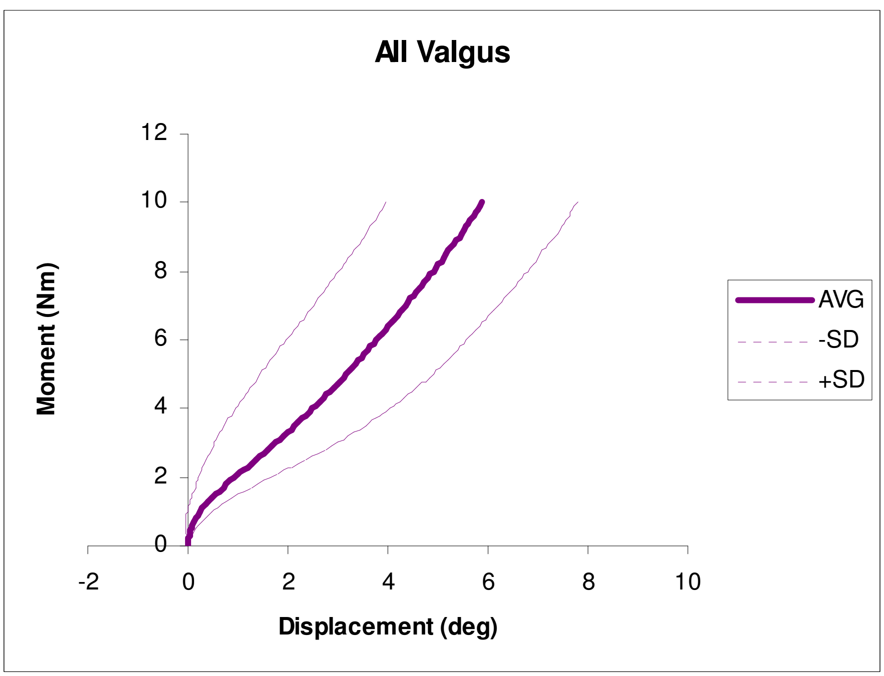


Fig 7: Valgus torque by displacement curve for all

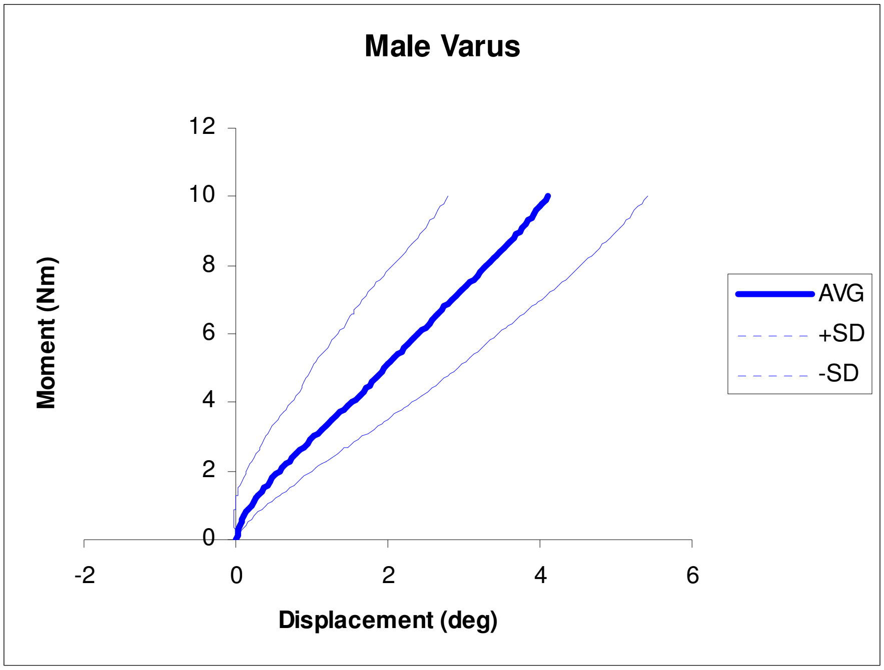


Fig 8: Male varus torque by displacement curve

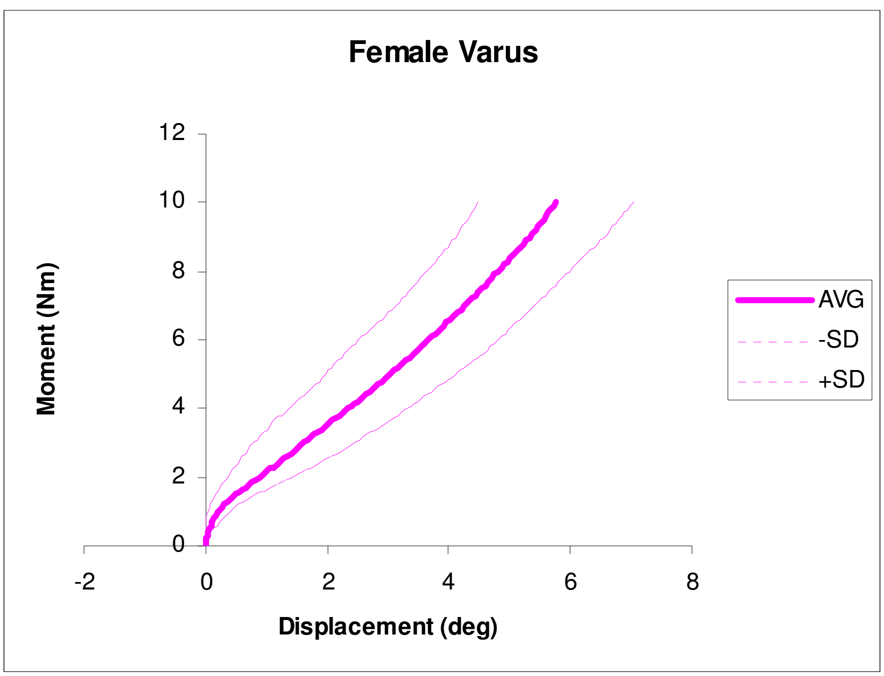


Fig 9: Female varus torque by displacement curve

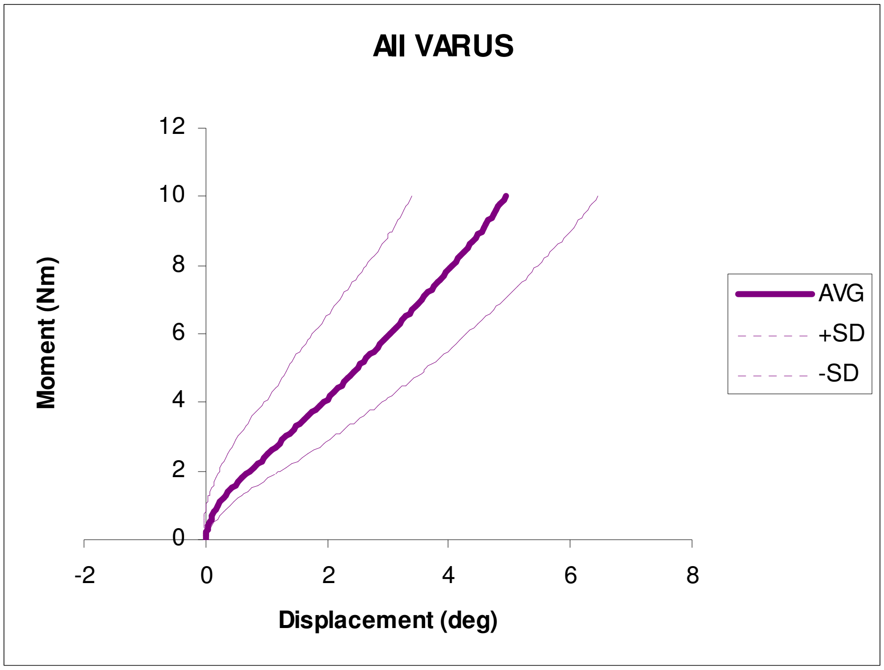


Fig 10: Varus torque by displacement curve for all

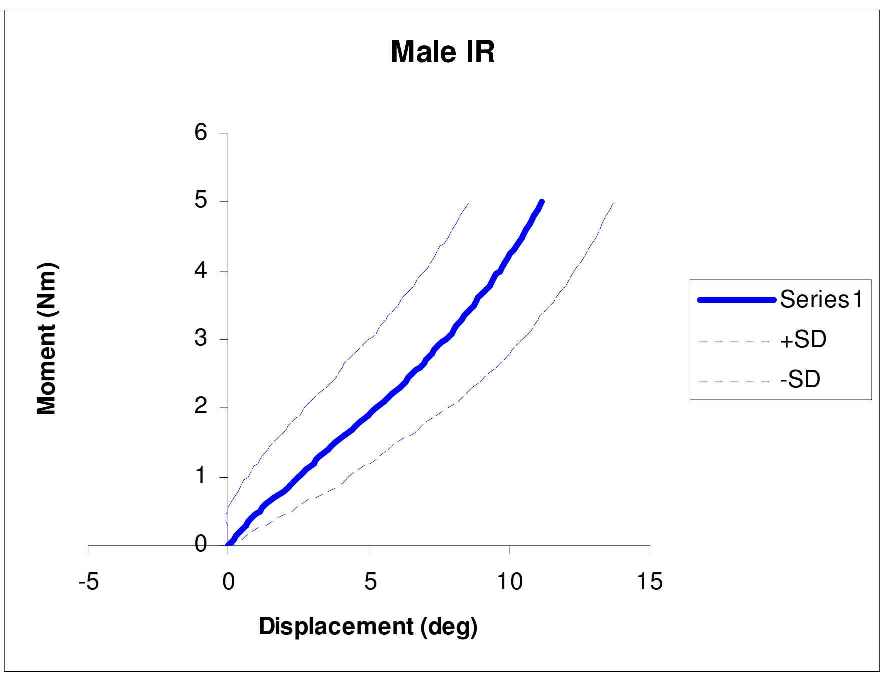


Fig 11: Male internal rotational torque by displacement curve

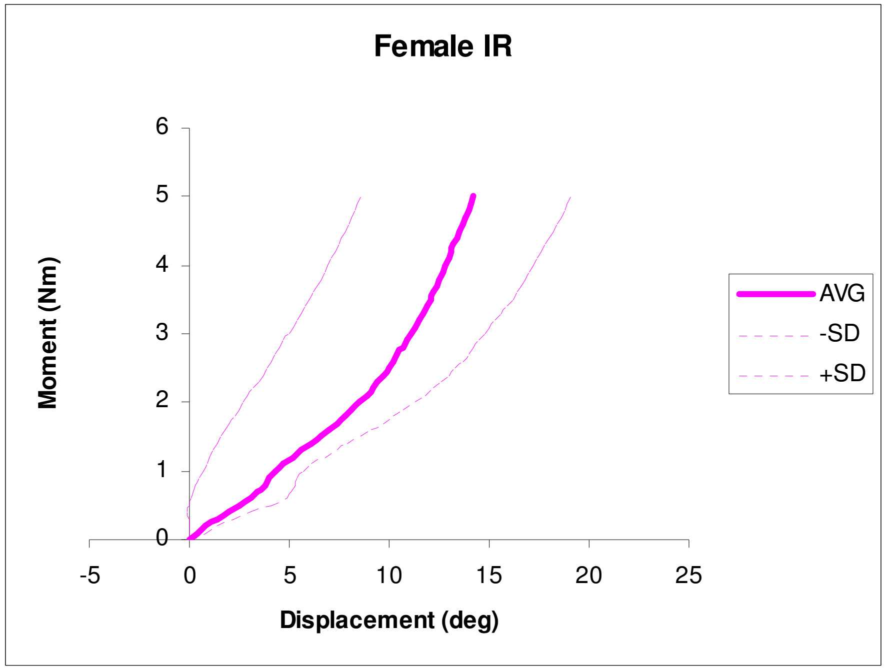


Fig 12: Female internal rotational torque by displacement curve

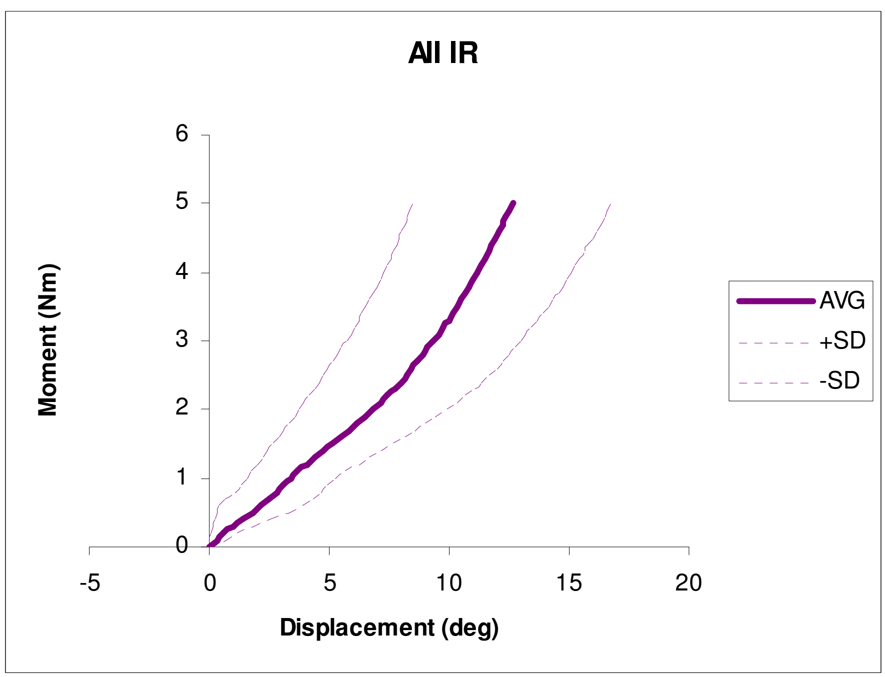


Fig 13: Internal rotational torque by displacement curve for all

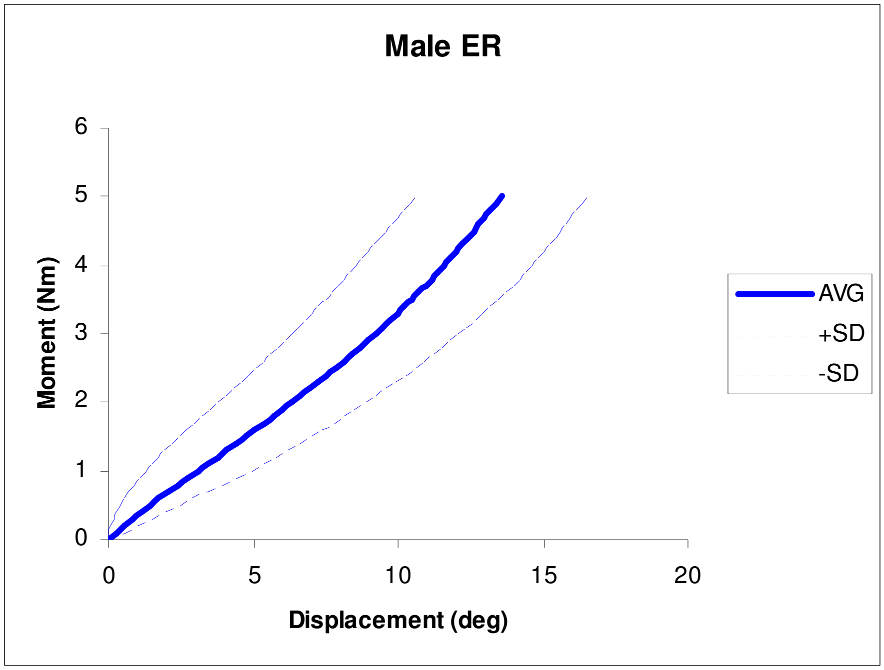


Fig 14: Male external rotational torque by displacement curve

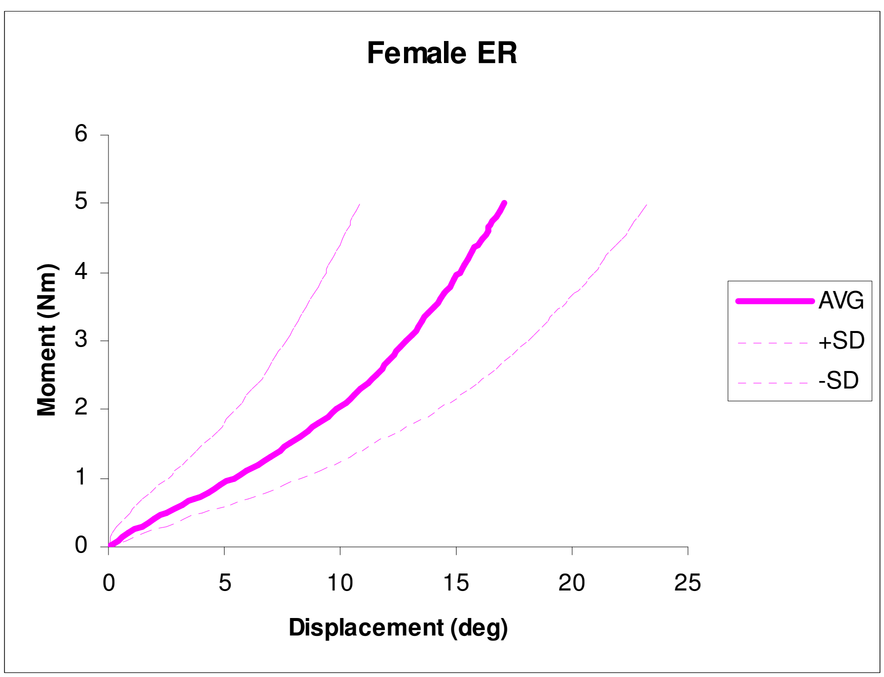


Fig 15: Female external rotational torque by displacement curve

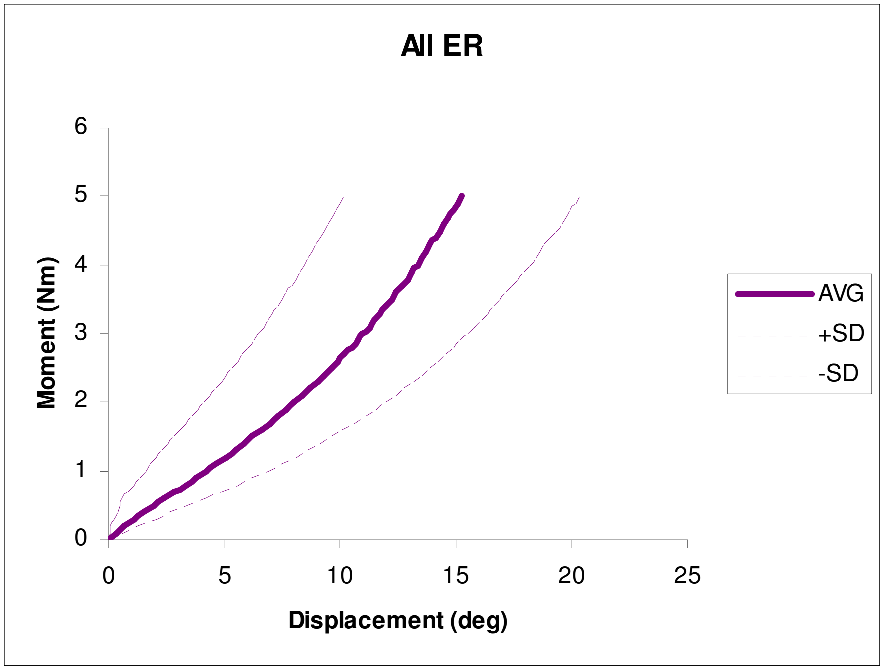


Fig 16: External rotational torque by displacement curve for all

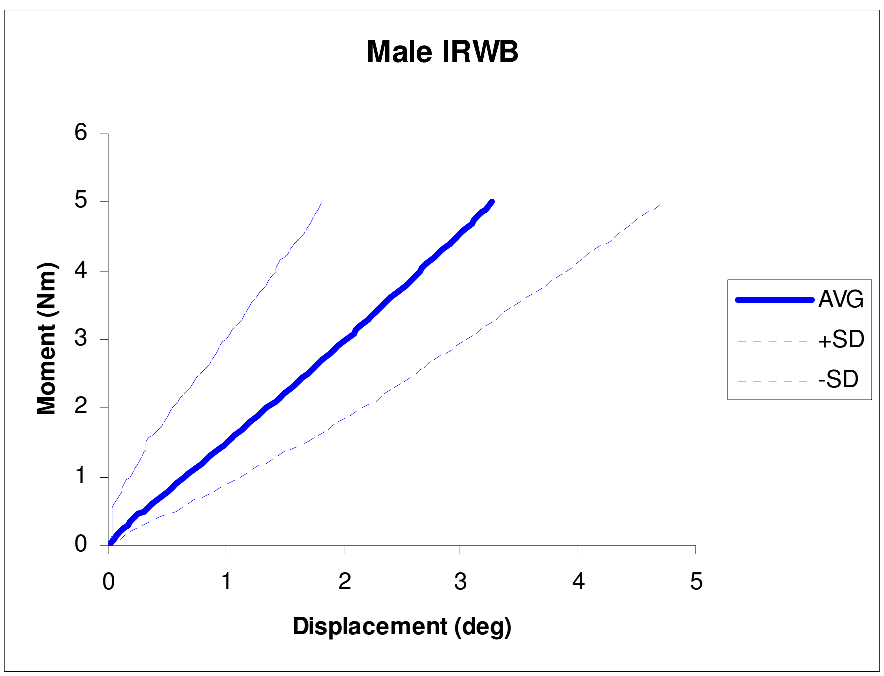


Fig 17: Male internal rotational torque by displacement curve (WB)

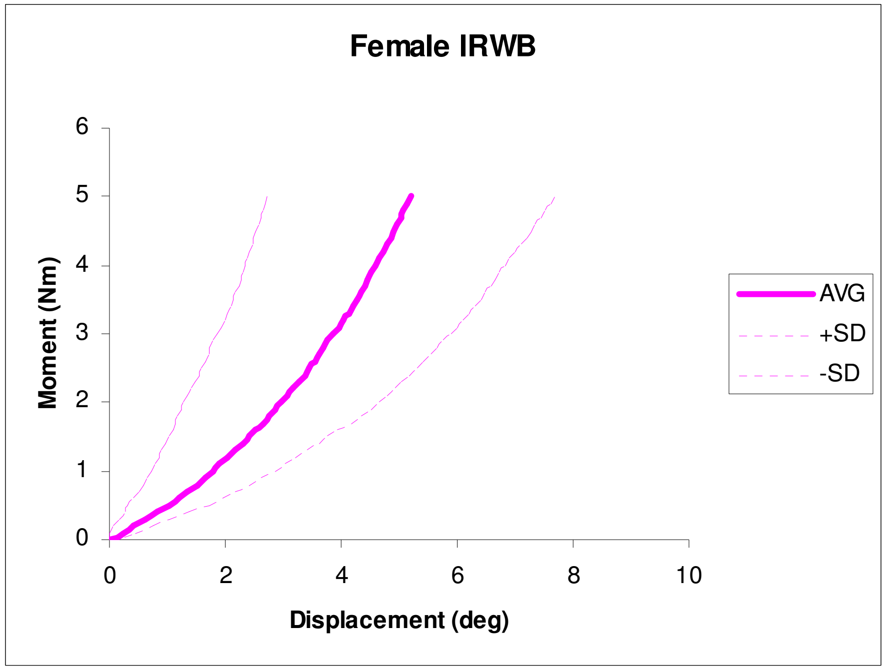


Fig 18: Female internal rotational torque by displacement curve (WB)

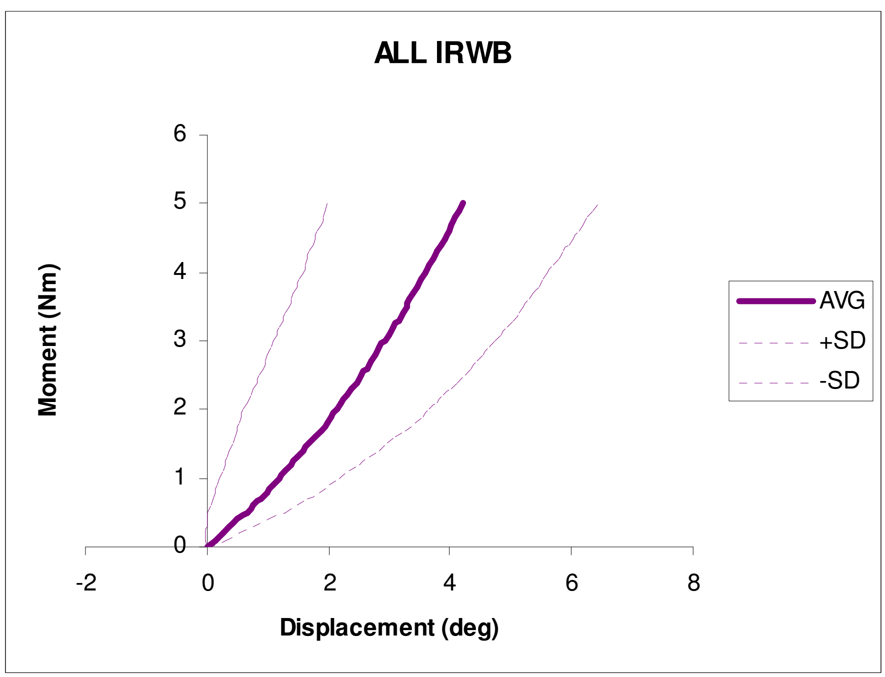


Fig 19: Internal rotational torque by displacement curve for all (WB)

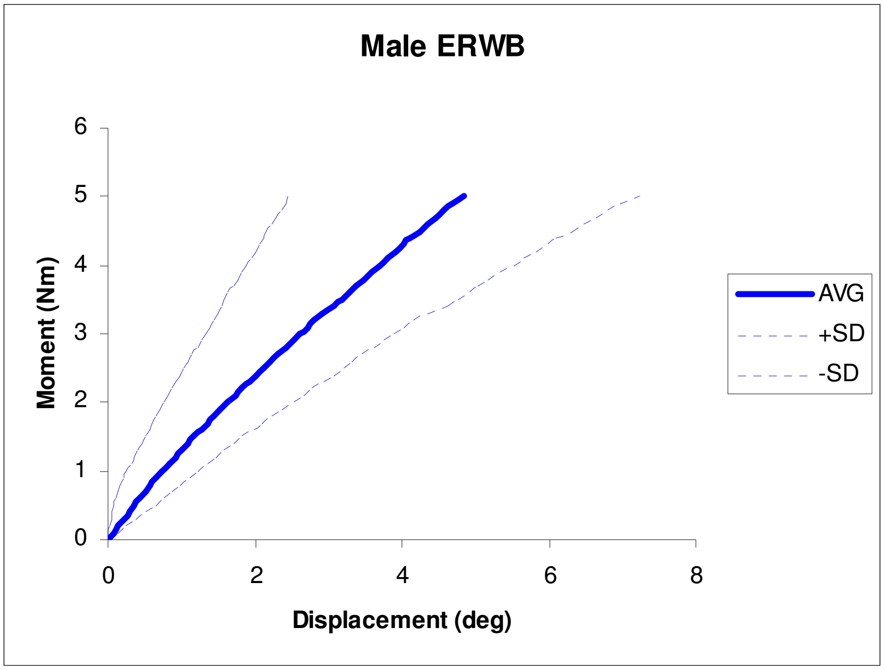


Fig 20: Male external rotational torque by displacement curve (WB)

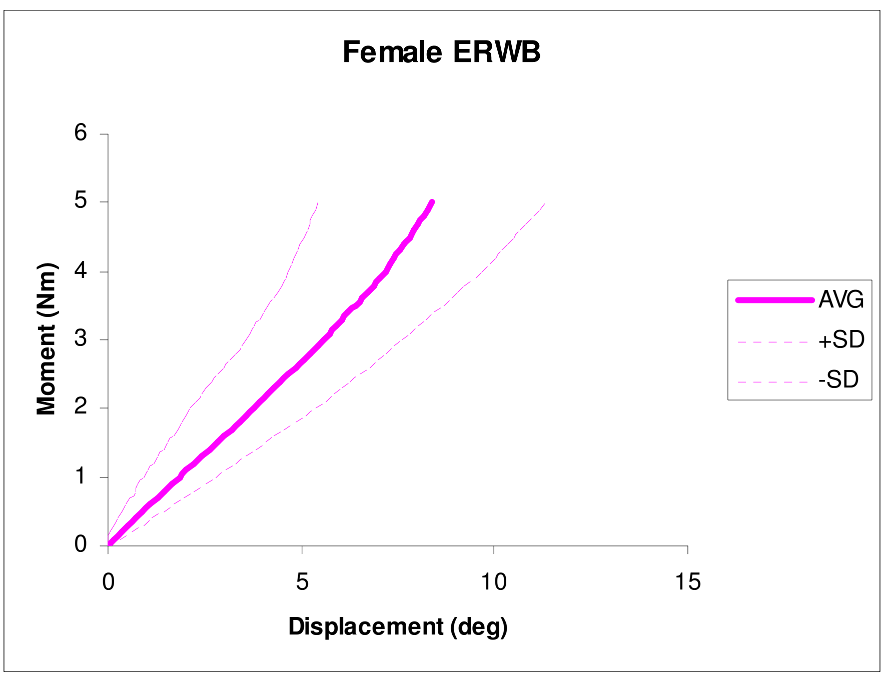


Fig 21: Female external rotational torque by displacement curve (WB)

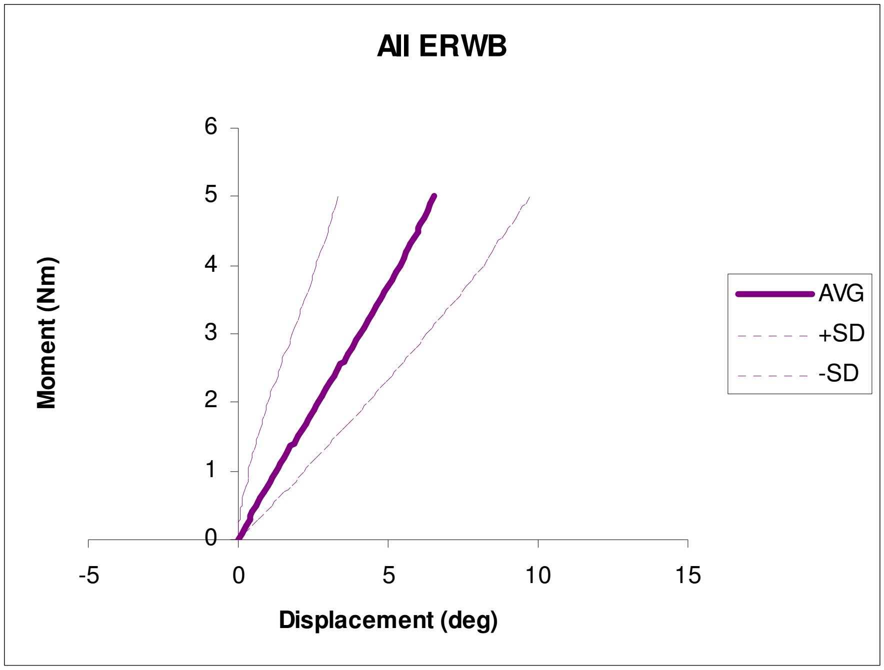


Fig 22: External rotational torque by displacement curve for all (WB)

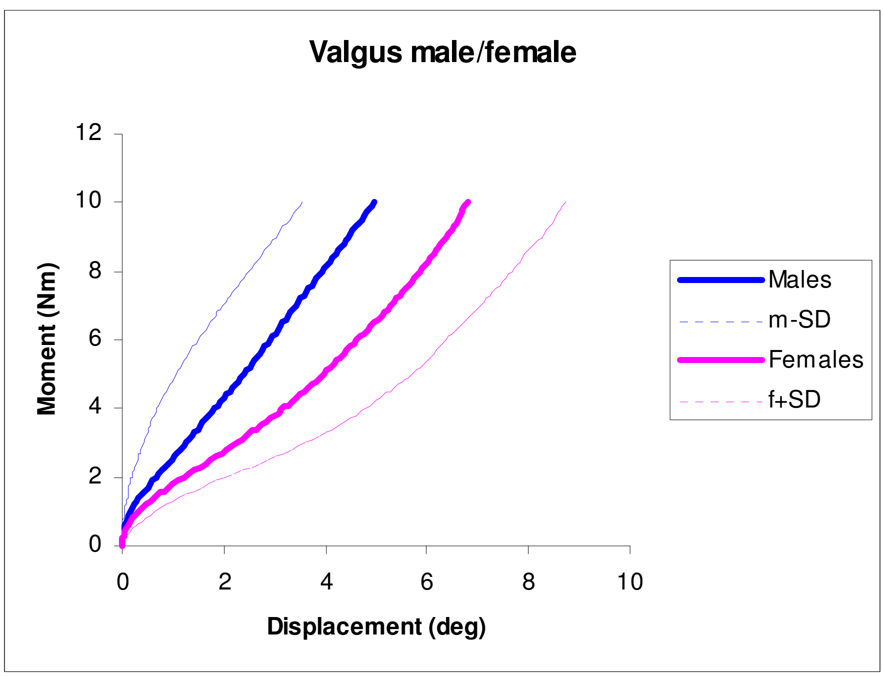


Fig 23: Comparison of male and female valgus

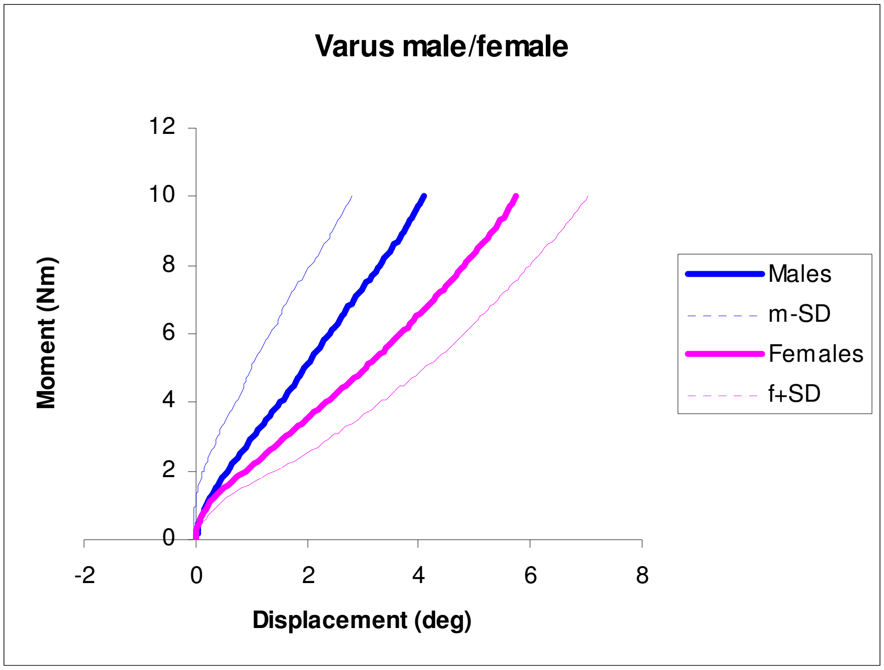


Fig 24: Comparison of male and female varus

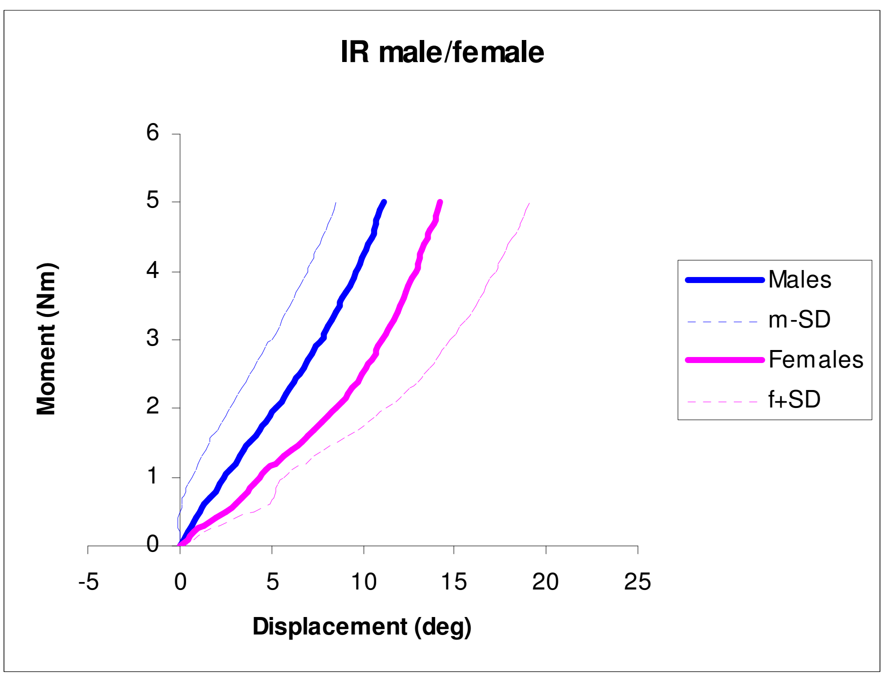


Fig 25: Comparison of male and female internal rotation

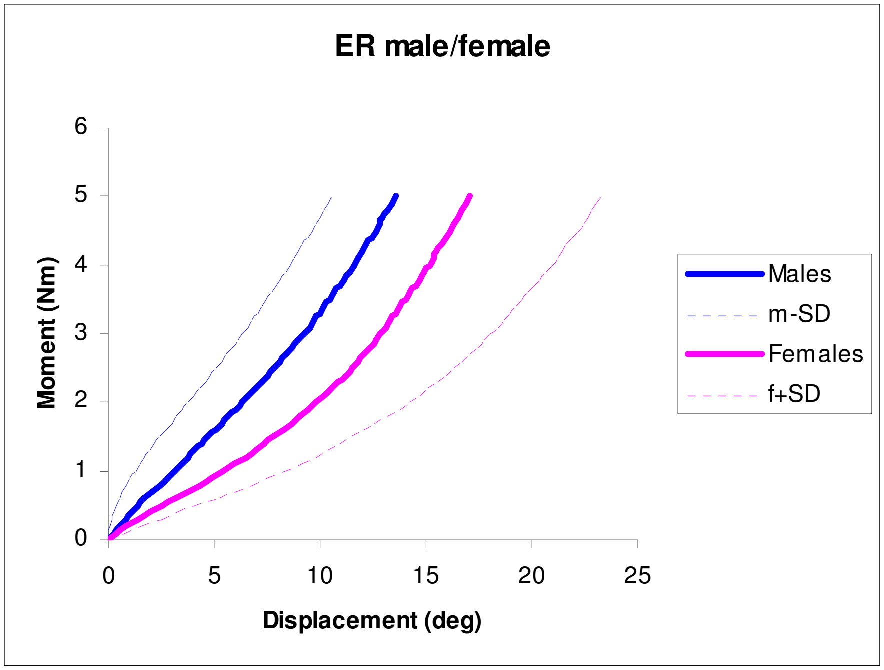


Fig 26: Comparison of male and female external rotation

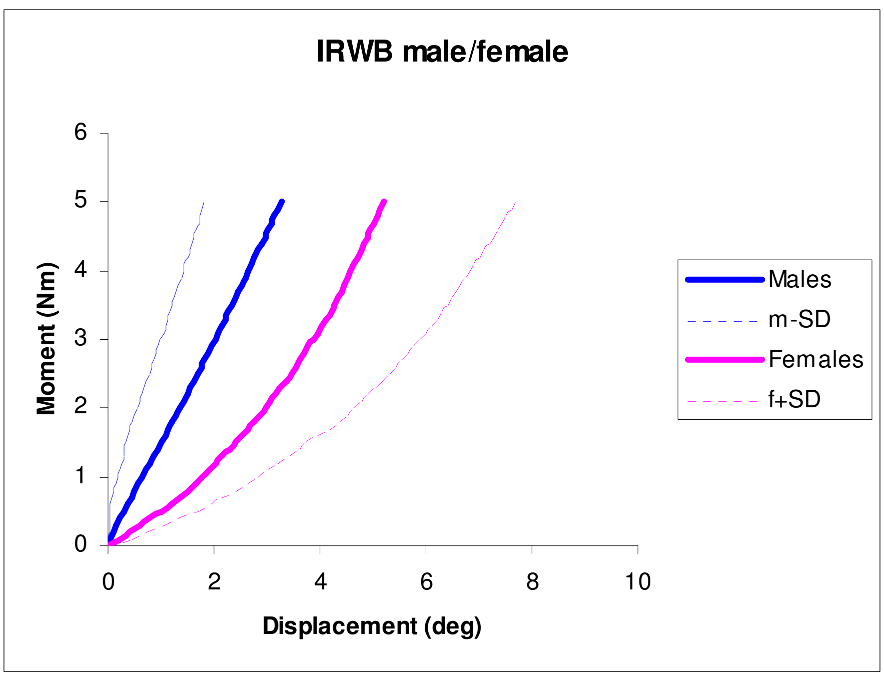


Fig 27: Comparison of male and female internal rotation (WB)

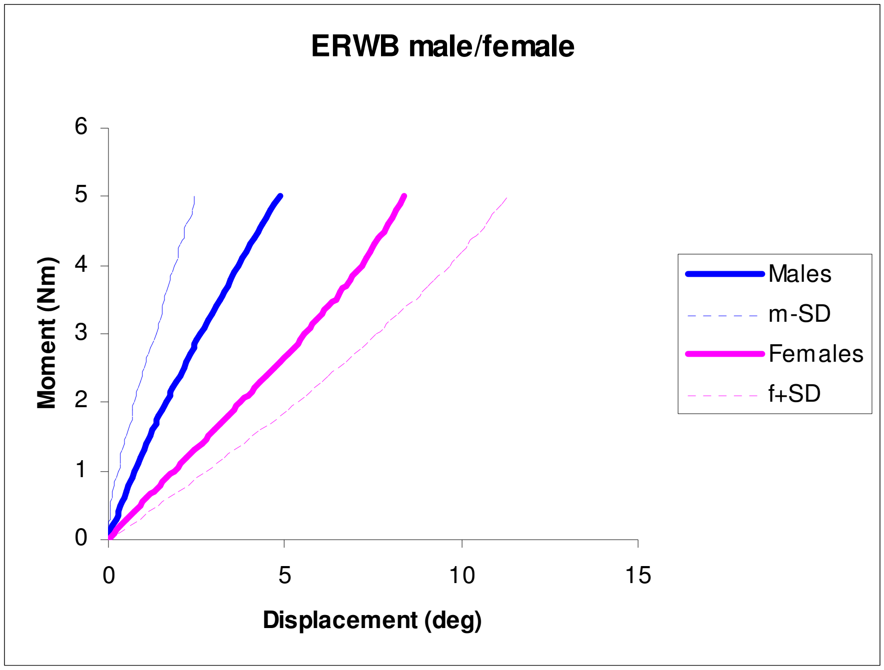


Fig 28: Comparison of male and female external rotation (WB)

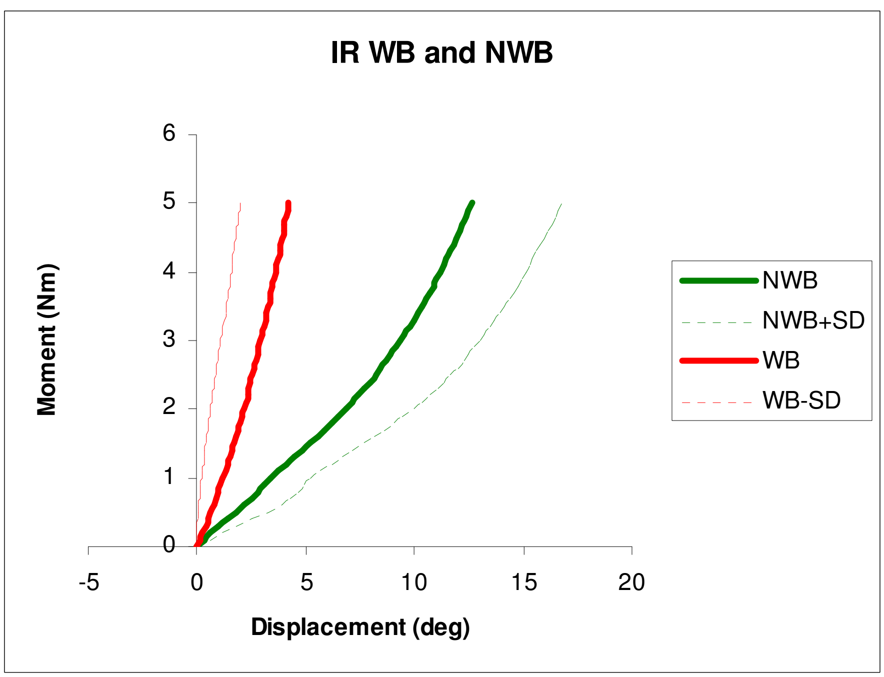


Fig 29: Comparison of internal rotation (WB and NWB)

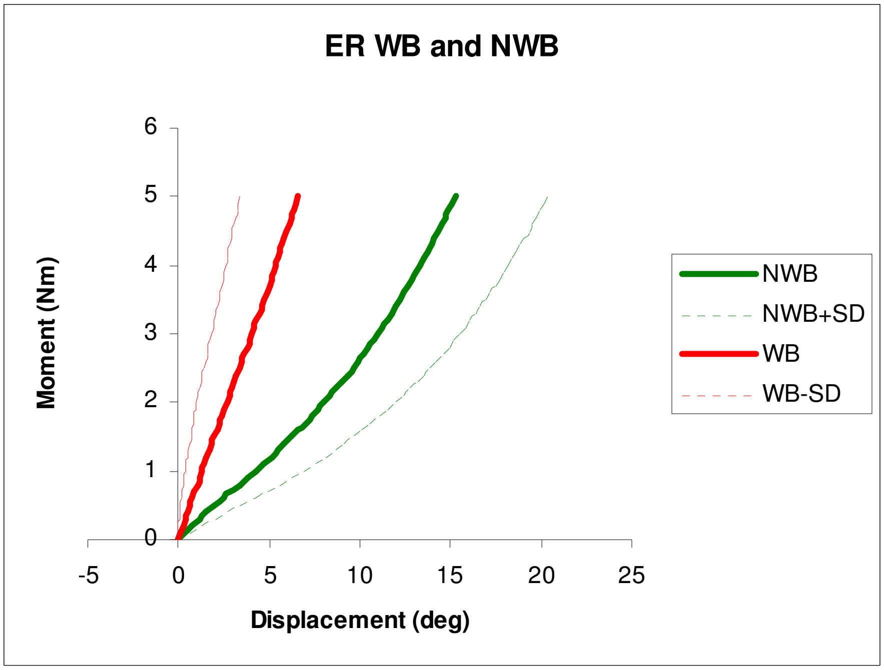


Fig 30: Comparison of external rotation (WB and NWB)

Objective 3

Valgus

The repeated measures ANOVA revealed a significant main effect ($F_{1,36}$=5.56, p=.024) for sex (see appendix A for SPSS output) with males having a greater mean valgus stiffness than females (2.32 ± .27 Nm/deg and 1.88 ± .27 Nm/deg respectively). A significant interaction of sex and incremental stiffness was also found ($F_{16,576}$=3.84, p<.01). Post hoc testing revealed that females had lower stiffness values early in the displacement (from 1.5 to 3.5 Nm of applied torque) (Figure 31). Additionally, while males experienced no increases in incremental stiffness as applied torque increased, females had increased incremental stiffness from 8.5 to 10 Nm of applied torque (Figure 31). Table 8 contains the data means and standard deviations for the male-female valgus stiffness comparison.

Table 8: Means and SD for female and male valgus

Applied Torque	Sex	Mean		SD	
1.5-2.0 Nm	female	1.40	±	0.71	Nm/deg
	male	3.01	±	1.88	Nm/deg
	Total	2.20	±	1.62	Nm/deg
2.0-2.5 Nm	female	1.27	±	0.74	Nm/deg
	male	2.39	±	1.45	Nm/deg
	Total	1.83	±	1.27	Nm/deg
2.5-3.0 Nm	female	1.19	±	0.55	Nm/deg
	male	2.39	±	1.31	Nm/deg
	Total	1.79	±	1.17	Nm/deg
3.0-3.5 Nm	female	1.22	±	0.44	Nm/deg
	male	2.35	±	1.11	Nm/deg
	Total	1.78	±	1.01	Nm/deg
3.5-4.0 Nm	female	1.22	±	0.36	Nm/deg
	male	2.26	±	1.18	Nm/deg
	Total	1.74	±	1.01	Nm/deg
4.0-4.5 Nm	female	1.28	±	0.37	Nm/deg
	male	2.10	±	1.00	Nm/deg
	Total	1.69	±	0.85	Nm/deg
4.5-5.0 Nm	female	1.42	±	0.50	Nm/deg
	male	2.08	±	0.88	Nm/deg
	Total	1.75	±	0.78	Nm/deg
5.0-5.5 Nm	female	1.60	±	0.56	Nm/deg
	male	2.05	±	0.79	Nm/deg
	Total	1.83	±	0.71	Nm/deg
5.5-6.0 Nm	female	1.77	±	0.74	Nm/deg
	male	2.11	±	0.65	Nm/deg
	Total	1.94	±	0.71	Nm/deg
6.0-6.5 Nm	female	1.72	±	0.81	Nm/deg
	male	2.05	±	0.56	Nm/deg
	Total	1.88	±	0.71	Nm/deg
6.5-7.0 Nm	female	2.12	±	1.33	Nm/deg
	male	2.27	±	0.98	Nm/deg
	Total	2.20	±	1.15	Nm/deg
7.0-7.5 Nm	female	2.07	±	0.91	Nm/deg
	male	2.36	±	1.19	Nm/deg
	Total	2.22	±	1.05	Nm/deg
7.5-8.0 Nm	female	2.31	±	1.15	Nm/deg
	male	2.13	±	0.68	Nm/deg
	Total	2.22	±	0.94	Nm/deg
8.0-8.5 Nm	female	2.28	±	0.95	Nm/deg
	male	2.11	±	0.87	Nm/deg
	Total	2.20	±	0.91	Nm/deg

8.5-9.0 Nm	female	2.46	±	1.05	Nm/deg
	male	2.31	±	1.12	Nm/deg
	Total	2.39	±	1.08	Nm/deg
9.0-9.5 Nm	female	3.13	±	1.92	Nm/deg
	male	2.65	±	1.83	Nm/deg
	Total	2.89	±	1.87	Nm/deg
9.5-10.0 Nm	female	3.41	±	2.15	Nm/deg
	male	2.83	±	2.20	Nm/deg
	Total	3.12	±	2.16	Nm/deg

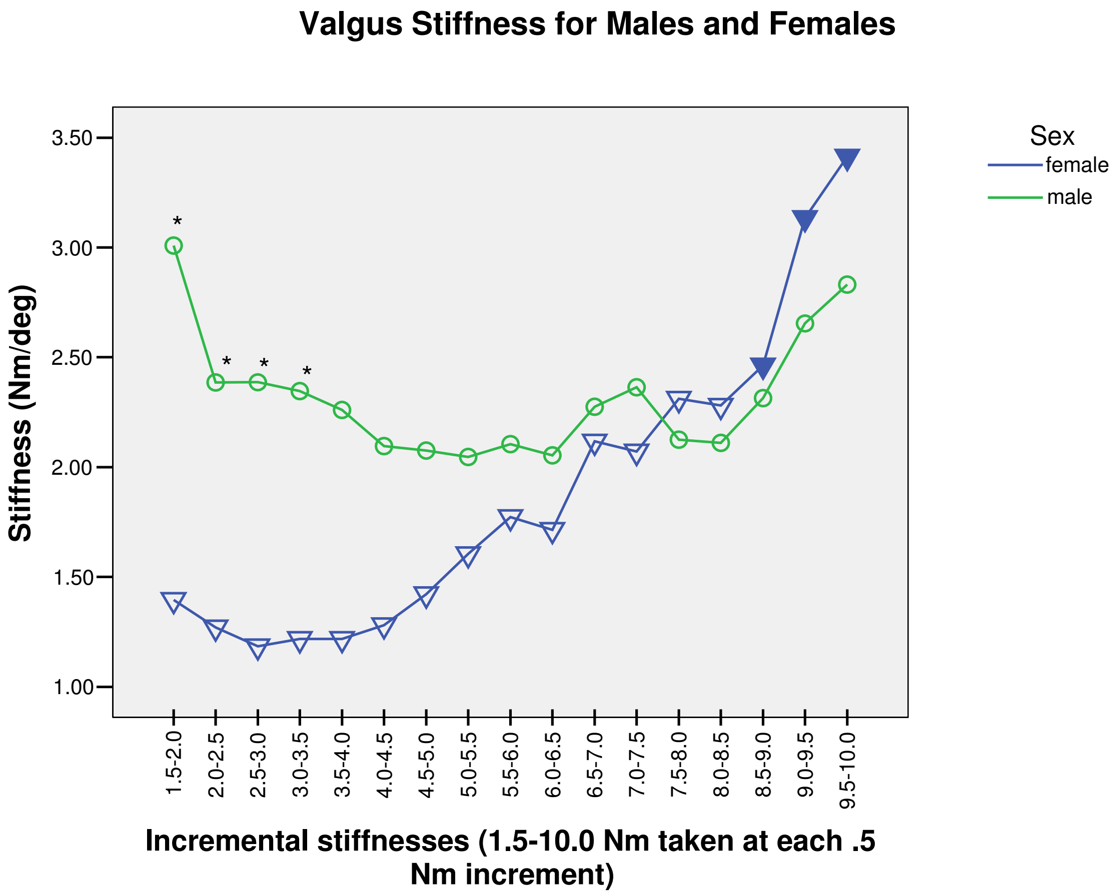


Figure 31: Profile plot for male and female valgus. * indicates a significant difference between males and females (occurring from 1.5-3.5 Nm applied torque). Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness in female displacement (occurring from 8.5-10.0 Nm of applied torque).

Varus

The repeated measures ANOVA revealed a significant main effect ($F_{1,33}$=9.37, p=.004) for sex (see appendix B for SPSS output) with males having a greater mean varus stiffness than females (2.68 ± .39 Nm/deg and 1.85 ± .37 Nm/deg respectively). A significant interaction of sex and incremental stiffness was found for the varus case as well ($F_{16,528}$=2.74, p<.01). Post hoc testing revealed that females had one lower stiffness value early in the displacement (at 1.5 Nm of applied torque) (Figure 32). Males experienced stiffness changes early in the loading cycle (between 1.5 and 2.0 Nm applied torque) and again at the loading cycle's end (between 9.5 and 10.0 Nm applied torque). Females again had increased incremental stiffness from 8.5 to 10 Nm of applied torque (Figure 32). Table 9 contains the data means and standard deviations for the male-female varus stiffness comparison.

Table 9: Means and SD for female and male varus

Applied Torque	Sex	Mean		SD	
1.5-2.0 Nm	female	1.54	±	1.14	Nm/deg
	male	3.37	±	2.22	Nm/deg
	Total	2.43	±	1.96	Nm/deg
2.0-2.5 Nm	female	1.38	±	0.51	Nm/deg
	male	2.80	±	1.68	Nm/deg
	Total	2.07	±	1.40	Nm/deg
2.5-3.0 Nm	female	1.36	±	0.46	Nm/deg
	male	2.45	±	1.14	Nm/deg
	Total	1.89	±	1.01	Nm/deg
3.0-3.5 Nm	female	1.44	±	0.44	Nm/deg
	male	2.35	±	1.00	Nm/deg
	Total	1.88	±	0.88	Nm/deg
3.5-4.0 Nm	female	1.44	±	0.41	Nm/deg
	male	2.42	±	1.33	Nm/deg
	Total	1.91	±	1.08	Nm/deg
4.0-4.5 Nm	female	1.46	±	0.35	Nm/deg
	male	2.56	±	1.28	Nm/deg
	Total	1.99	±	1.07	Nm/deg
4.5-5.0 Nm	female	1.62	±	0.56	Nm/deg
	male	2.45	±	1.12	Nm/deg
	Total	2.02	±	0.96	Nm/deg
5.0-5.5 Nm	female	1.58	±	0.39	Nm/deg
	male	2.60	±	1.25	Nm/deg
	Total	2.07	±	1.04	Nm/deg
5.5-6.0 Nm	female	1.71	±	0.38	Nm/deg
	male	2.82	±	1.31	Nm/deg
	Total	2.25	±	1.10	Nm/deg
6.0-6.5 Nm	female	1.82	±	0.44	Nm/deg
	male	2.61	±	1.26	Nm/deg
	Total	2.20	±	1.00	Nm/deg
6.5-7.0 Nm	female	1.83	±	0.49	Nm/deg
	male	2.51	±	1.34	Nm/deg
	Total	2.16	±	1.04	Nm/deg
7.0-7.5 Nm	female	2.13	±	0.95	Nm/deg
	male	2.34	±	0.92	Nm/deg
	Total	2.23	±	0.93	Nm/deg
7.5-8.0 Nm	female	2.10	±	0.54	Nm/deg
	male	2.40	±	0.84	Nm/deg
	Total	2.25	±	0.71	Nm/deg
8.0-8.5 Nm	female	2.03	±	0.64	Nm/deg
	male	2.65	±	1.47	Nm/deg
	Total	2.33	±	1.15	Nm/deg

8.5-9.0 Nm	female	2.29	±	0.90	Nm/deg
	male	2.87	±	1.67	Nm/deg
	Total	2.57	±	1.34	Nm/deg
9.0-9.5 Nm	female	2.83	±	1.59	Nm/deg
	male	2.90	±	1.19	Nm/deg
	Total	2.87	±	1.39	Nm/deg
9.5-10.0 Nm	female	2.88	±	1.35	Nm/deg
	male	3.43	±	1.71	Nm/deg
	Total	3.15	±	1.54	Nm/deg

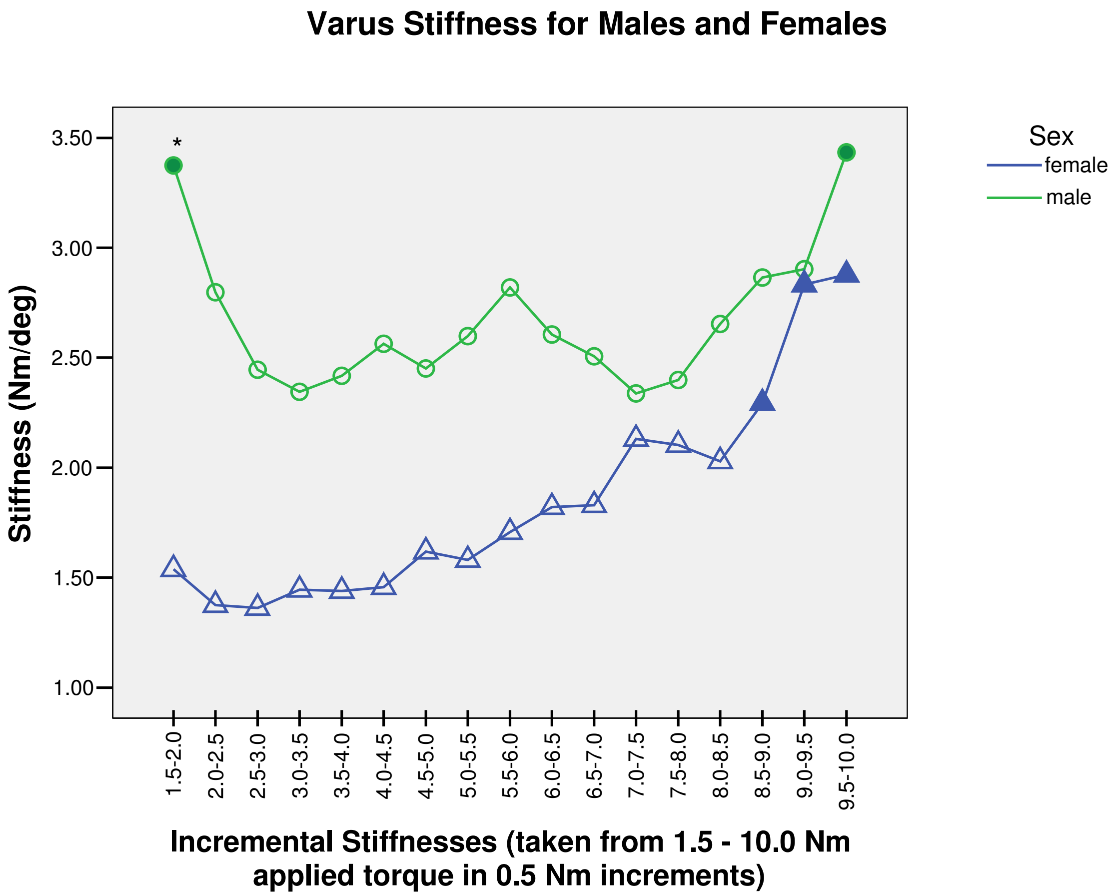


Figure 32: Profile plot for male and female varus. * indicates a significant difference between males and females (occurring from 1.5 to 2.0 Nm applied torque). Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness (2-4 Nm) in female displacement (occurring from 8.5-10.0 Nm of applied torque). Solid circles indicate differing stiffness than open circles in male displacement (occurring from 1.5-2.0 Nm and again from 9.5-10.0 Nm of applied torque).

Internal Rotation

The repeated measures ANOVA for internal rotation revealed no significant main effect for sex ($F_{1,34}$=.001, p=.98) (see appendix C for SPSS output). A significant interaction of sex and incremental stiffness was found ($F_{8,272}$=4.78, p<.01), with males having significantly greater mean stiffness early in the loading cycle (at between .5-1.0 Nm) (figure 33). Post hoc testing revealed that females had two increases in stiffness during loading (between 2.0 Nm and 2.5 Nm of applied torque and again between 3.5 and 4.0 Nm) (Figure 33). Males experienced a stiffness change later in the loading cycle (between 4.5 and 5.0 Nm applied torque). Table 10 contains the data means and standard deviations for the male-female internal rotational stiffness comparison.

Table 10: Means and SD for female and male internal rotation

Applied Torque	Sex	Mean		SD	
0.5-1.0 Nm	female	0.04	±	0.85	Nm/deg
	male	0.46	±	0.25	Nm/deg
	Total	0.25	±	0.65	Nm/deg
1.0-1.5 Nm	female	0.27	±	0.16	Nm/deg
	male	0.43	±	0.18	Nm/deg
	Total	0.35	±	0.18	Nm/deg
1.5-2.0 Nm	female	0.31	±	0.18	Nm/deg
	male	0.39	±	0.16	Nm/deg
	Total	0.35	±	0.17	Nm/deg
2.0-2.5 Nm	female	0.45	±	0.36	Nm/deg
	male	0.42	±	0.12	Nm/deg
	Total	0.44	±	0.26	Nm/deg
2.5-3.0 Nm	female	0.52	±	0.27	Nm/deg
	male	0.44	±	0.09	Nm/deg
	Total	0.48	±	0.20	Nm/deg
3.0-3.5 Nm	female	0.60	±	0.29	Nm/deg
	male	0.53	±	0.11	Nm/deg
	Total	0.56	±	0.22	Nm/deg
3.5-4.0 Nm	female	0.76	±	0.39	Nm/deg
	male	0.59	±	0.16	Nm/deg
	Total	0.67	±	0.31	Nm/deg
4.0-4.5 Nm	female	0.86	±	0.48	Nm/deg
	male	0.67	±	0.18	Nm/deg
	Total	0.76	±	0.37	Nm/deg
4.5-5.0 Nm	female	0.94	±	0.63	Nm/deg
	male	0.79	±	0.22	Nm/deg
	Total	0.86	±	0.47	Nm/deg

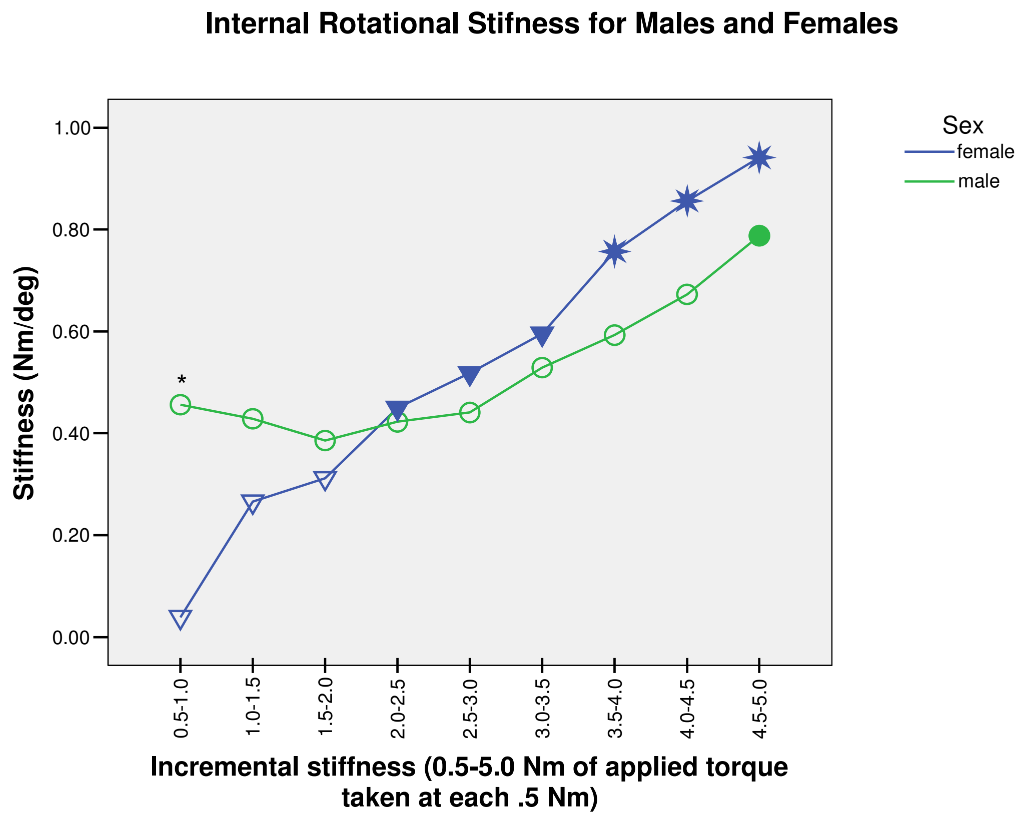


Figure 33: Profile plot for male and female internal rotation. * indicates male/female difference at .5-1.0 Nm applied torque. Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness in female displacement (occurring at between 2.0-2.5 Nm of applied torque). Solid stars indicate further differing stiffness values (from initial solid triangle) for females (occurring at between 3.5-4.0 Nm applied torque). Solid circles indicate differing stiffness than open circles in male displacement (occurring at between 4.5-5.0 Nm of applied torque).

External Rotation

The repeated measures ANOVA for external rotation revealed no significant main effect for sex ($F_{1,34}$=.60, p=.44) (see appendix D for SPSS output). A significant interaction of sex and incremental stiffness again was found ($F_{8,272}$=7.17, p<.01). Post hoc testing revealed differences in stiffness during loading for females: once at between 2.5-3.0 Nm and then again at between 4.0 and 4.5 Nm of applied torque. Again, males showed increase during loading at between 4.0 and 4.5 Nm of applied torque (Figure 34). Males and females showed stiffness mean difference at between 0.5-1.0 Nm applied torque. Table 11 contains the data means and standard deviations for the male-female external rotational stiffness comparison.

Table 11: Means and SD for female and male external rotation

Applied Torque	Sex	Mean		SD	
0.5-1.0 Nm	female	0.22	±	0.12	Nm/deg
	male	0.43	±	0.31	Nm/deg
	Total	0.33	±	0.26	Nm/deg
1.0-1.5 Nm	female	0.24	±	0.11	Nm/deg
	male	0.37	±	0.18	Nm/deg
	Total	0.31	±	0.16	Nm/deg
1.5-2.0 Nm	female	0.28	±	0.10	Nm/deg
	male	0.35	±	0.13	Nm/deg
	Total	0.31	±	0.12	Nm/deg
2.0-2.5 Nm	female	0.33	±	0.11	Nm/deg
	male	0.36	±	0.12	Nm/deg
	Total	0.34	±	0.12	Nm/deg
2.5-3.0 Nm	female	0.40	±	0.12	Nm/deg
	male	0.39	±	0.11	Nm/deg
	Total	0.40	±	0.11	Nm/deg
3.0-3.5 Nm	female	0.46	±	0.13	Nm/deg
	male	0.42	±	0.10	Nm/deg
	Total	0.44	±	0.12	Nm/deg
3.5-4.0 Nm	female	0.48	±	0.15	Nm/deg
	male	0.46	±	0.13	Nm/deg
	Total	0.47	±	0.14	Nm/deg
4.0-4.5 Nm	female	0.55	±	0.18	Nm/deg
	male	0.50	±	0.12	Nm/deg
	Total	0.53	±	0.15	Nm/deg
4.5-5.0 Nm	female	0.59	±	0.16	Nm/deg
	male	0.54	±	0.15	Nm/deg
	Total	0.57	±	0.15	Nm/deg

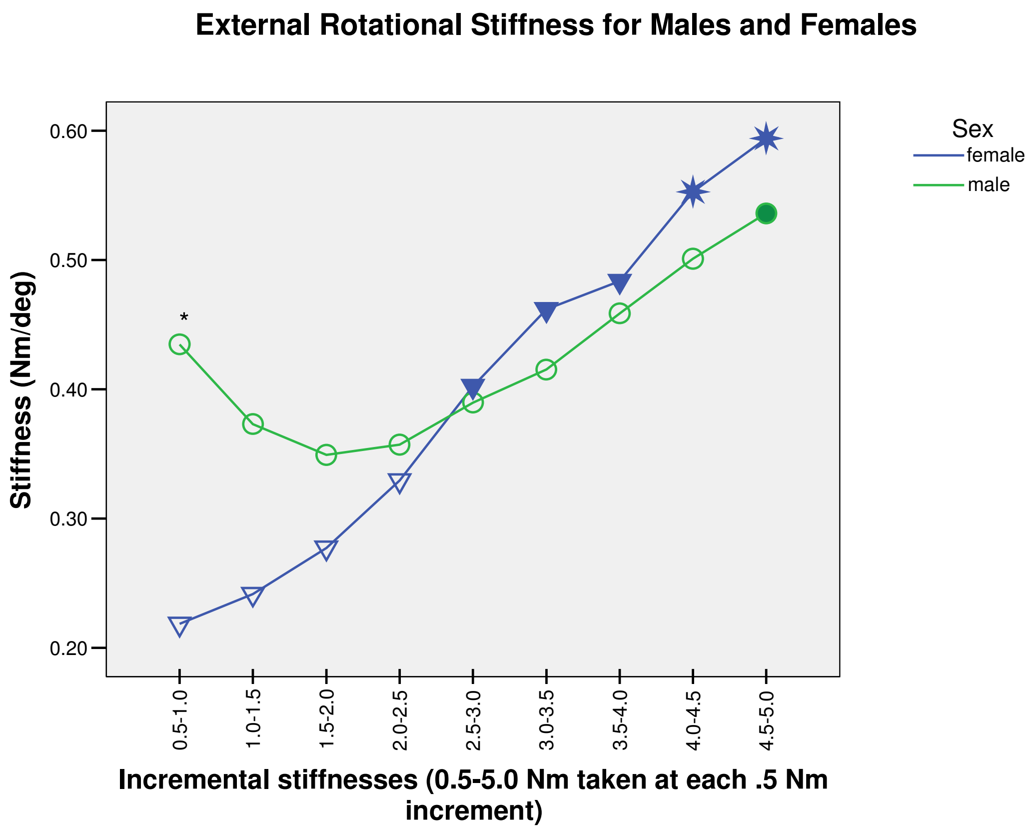


Figure 34: Profile plot for male and female external rotation. * indicates male/female difference at .5-1.0 Nm applied torque. Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness in female displacement (occurring at between 2.5-3.0 Nm of applied torque). Solid stars indicate further differing stiffness values for females (occurring at between 4.0-4.5 Nm applied torque). Solid circles indicate differing stiffness than initial stiffness in male displacement (occurring at between 4.5-5.0 Nm of applied torque).

Internal Rotation in Weightbearing

The repeated measures ANOVA for internal rotation in weightbearing revealed no significant main effect for sex ($F_{1,33}$=.001, p=.98) (see appendix E for SPSS output). There was, however, an interaction of sex and incrementally measured stiffness ($F_{8,264}$=6.28, p<.01). Post hoc testing revealed differences in stiffness during loading for females: once at between 3.0-3.5 Nm and then again at between 4.0 and 4.5 Nm of applied torque. Males did not have differing stiffness during loading (Figure 35). Table 12 contains the data means and standard deviations for the male-female external rotational stiffness comparison.

Table 12: Means and SD for female and male internal rotation (WB)

Applied Torque	Sex	Mean		SD	
0.5-1.0 Nm	female	0.85	±	0.54	Nm/deg
	male	1.70	±	0.96	Nm/deg
	Total	1.26	±	0.88	Nm/deg
1.0-1.5 Nm	female	0.96	±	0.61	Nm/deg
	male	1.47	±	0.61	Nm/deg
	Total	1.21	±	0.65	Nm/deg
1.5-2.0 Nm	female	1.23	±	0.83	Nm/deg
	male	1.46	±	0.50	Nm/deg
	Total	1.34	±	0.69	Nm/deg
2.0-2.5 Nm	female	1.32	±	0.81	Nm/deg
	male	1.50	±	0.53	Nm/deg
	Total	1.41	±	0.68	Nm/deg
2.5-3.0 Nm	female	1.43	±	0.72	Nm/deg
	male	1.52	±	0.50	Nm/deg
	Total	1.47	±	0.61	Nm/deg
3.0-3.5 Nm	female	1.56	±	0.80	Nm/deg
	male	1.51	±	0.35	Nm/deg
	Total	1.54	±	0.62	Nm/deg
3.5-4.0 Nm	female	2.05	±	1.02	Nm/deg
	male	1.50	±	0.49	Nm/deg
	Total	1.78	±	0.85	Nm/deg
4.0-4.5 Nm	female	2.34	±	1.81	Nm/deg
	male	1.63	±	0.70	Nm/deg
	Total	2.00	±	1.41	Nm/deg
4.5-5.0 Nm	female	2.27	±	1.61	Nm/deg
	male	1.77	±	0.75	Nm/deg
	Total	2.03	±	1.28	Nm/deg

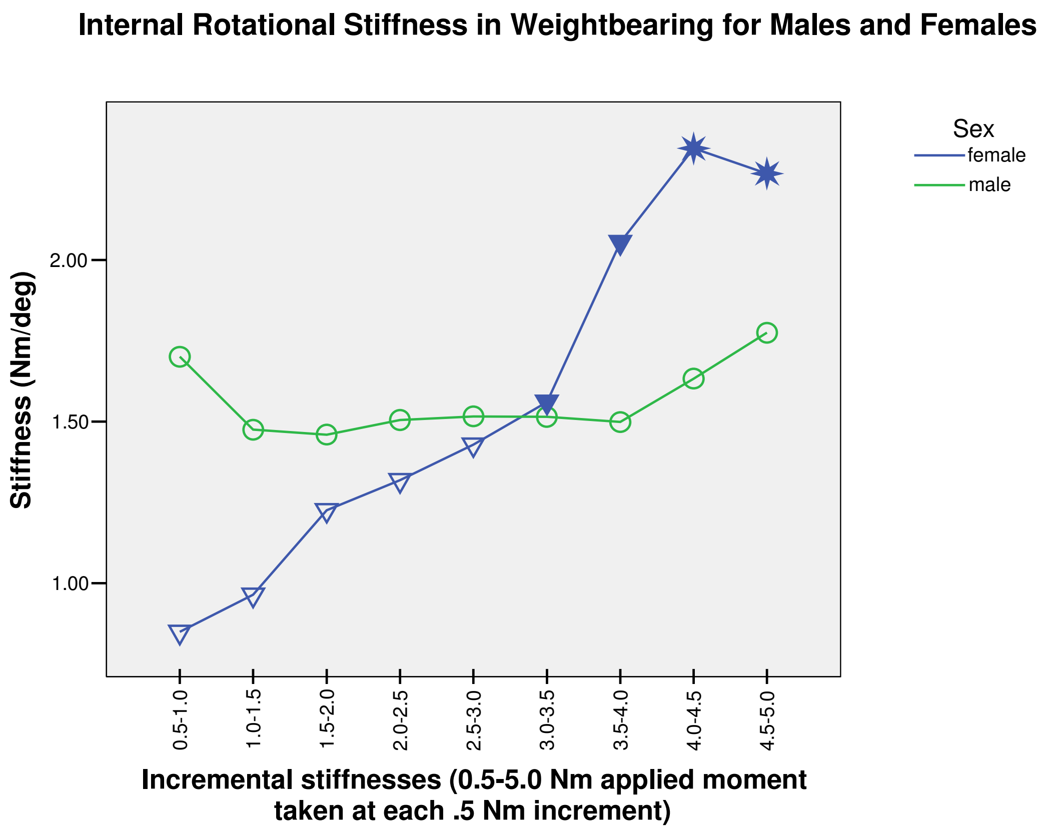


Figure 35: Profile plot for male and female internal rotation (WB). Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness in female displacement (occurring at between 3.0-3.5 Nm of applied torque). Solid stars indicate further differing stiffness values for females (occurring at between 4.0-4.5 Nm applied torque).

External Rotation in Weightbearing

The repeated measures ANOVA revealed a significant main effect ($F_{1,34}$=7.65, p=.009) for sex (see appendix F for SPSS output) with males having a greater mean stiffness than females (1.28 ± .24 Nm/deg and .80 ± .24 Nm/deg respectively). A significant interaction of sex and incremental stiffness was also found ($F_{8,272}$=6.41, p<.01). Post hoc testing revealed that females had lower stiffness values than males early in the displacement (from 0.5 to 2.0 Nm of applied torque) (Figure 36). Females increased incremental stiffness at between 4.5-5.0 Nm of applied torque (Figure 36). Males decreased in stiffness at 4.5-5.0 Nm of applied torque. Table 13 contains the data means and standard deviations for the male-female external rotation in weightbearing stiffness comparison.

Table 13: Means and SD for female and male external rotation (WB)

Applied Torque	Sex	Mean		SD	
0.5-1.0 Nm	female	0.88	±	1.12	Nm/deg
	male	1.77	±	1.16	Nm/deg
	Total	1.33	±	1.21	Nm/deg
1.0-1.5 Nm	female	0.60	±	0.29	Nm/deg
	male	1.47	±	0.84	Nm/deg
	Total	1.03	±	0.76	Nm/deg
1.5-2.0 Nm	female	0.64	±	0.25	Nm/deg
	male	1.40	±	0.85	Nm/deg
	Total	1.02	±	0.73	Nm/deg
2.0-2.5 Nm	female	0.66	±	0.36	Nm/deg
	male	1.19	±	0.51	Nm/deg
	Total	0.93	±	0.51	Nm/deg
2.5-3.0 Nm	female	0.67	±	0.34	Nm/deg
	male	1.22	±	0.68	Nm/deg
	Total	0.95	±	0.60	Nm/deg
3.0-3.5 Nm	female	0.77	±	0.44	Nm/deg
	male	1.13	±	0.53	Nm/deg
	Total	0.95	±	0.51	Nm/deg
3.5-4.0 Nm	female	0.83	±	0.44	Nm/deg
	male	1.24	±	0.74	Nm/deg
	Total	1.04	±	0.63	Nm/deg
4.0-4.5 Nm	female	1.01	±	0.59	Nm/deg
	male	1.07	±	0.44	Nm/deg
	Total	1.04	±	0.51	Nm/deg
4.5-5.0 Nm	female	1.13	±	0.77	Nm/deg
	male	1.00	±	0.41	Nm/deg
	Total	1.06	±	0.61	Nm/deg

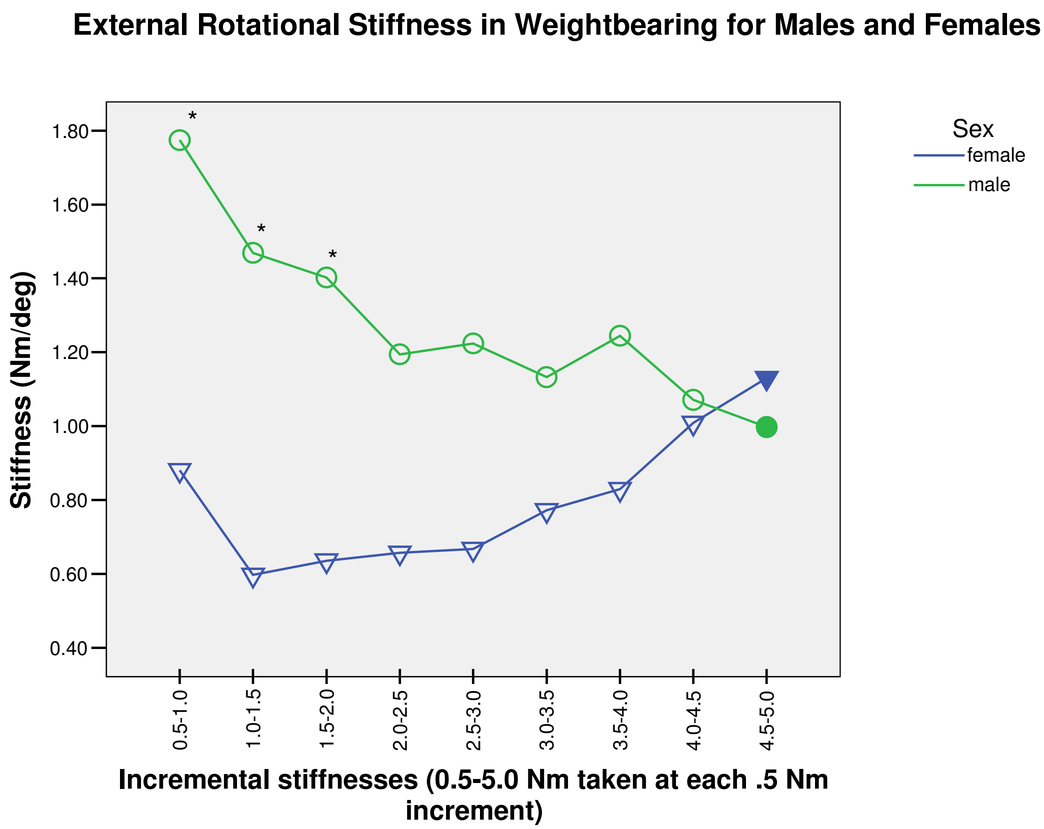


Figure 36: Profile plot for male and female external rotation (WB). * indicates a significant difference between males and females (occurring from 0.5-2.0 Nm applied torque). Solid triangles indicate incremental stiffness constants significantly differ from initial stiffness in female displacement (occurring from 4.5-5.0 Nm of applied torque). Solid circle indicates stiffness significantly less than initial stiffness for males.

CHAPTER V

DISCUSSION

Objective 1

The primary findings of objective 1 were that across the entire loading cycle, valgus and external rotational loading stiffness were the most reliable measures (tables 2 & 7). In the varus and internal rotation conditions, acceptable reliability ($ICC_{2,k} > .70$) was seen only in the early phase of loading (up to 2 Nm and 3 Nm, respectively).

The lower reliability in the upper ranges of varus and internal rotation loading may be due to various neuromuscular protective responses. Although every attempt was made to insure muscle passivity in the NWB conditions, it is possible that subjects may have reacted to loads differently. It is reasonable to assume that if subjects were indeed guarding, they would do so in ways that would diminish reliability greatly. Additionally, these directions often felt the most "unnatural" to the subjects, thus the potential for more neuromuscular guarding.

The occurrence of greater reliability early in the loading cycle is important to the discussion of male/female stiffness differences that occur early in loading. Where reliability is decreased due to increased error, meaningful differences in stiffness are more difficult to detect, and the ones found should be viewed in that light.

Due to the inability to locate other studies that have reported within day or between day measurement consistency, the current reliability results are not compared to previous work.

Objective 2

The primary finding of objective 2 is a consistency in moment by displacement curves produced from these data and those seen in other studies (Bryant & Cooke, 1988; Hsu et al., 2006; Markolf et al., 1976). The largest deviation from previous work is that for several loadings, there is a brief phase (0 – 1 Nm) of increased slope to the curve (figures 23 & 24). This suggests increased stiffness at the onset of loading. As detailed in the methods, the stiffnesses calculated at the beginnings of the loading curves, however, contain highly aberrant stiffness values when compared to previous literature.

For example, even in full extension, the highest reported values for varus/valgus stiffness in the existing literature are on the order of 15 Nm/deg (Markolf et al., 1976). Valgus and varus values in this study for the first three levels of incrementally derived stiffness were on the order of minimum 20 Nm/deg (at 20 degrees knee flexion) and ranged as high as 280 Nm/deg. In internal and external rotation, existing studies report stiffnesses on the order of 0.31 to 2.5 Nm/deg for any condition (Louie & Mote Jr., 1987; Markolf et al., 1981). In internal and external rotation up to .5 Nm of applied torque, this study found unnaturally high stiffnesses on the order of 6 Nm/deg up to 220 Nm/deg. This is the reason that statistical analyses were only conducted at points beyond these aberrant early incrementally measured stiffnesses.

The increased slope at the beginning of the loading curve may not therefore be representative of true torsional joint stiffness. More likely, it is the result of some guarding by the subjects (cadaveric studies lack these regions of high initial stiffness) (Hsu et al., 2006; Markolf et al., 1976), or possibly the result of a measurable torque with very little measurable displacement.

It is likely that during the early phase of loading there was a small amount of soft tissue compression. This early applied torque in combination with the soft tissue compression that would result in no bony movement could likely be a contributing cause to these aberrantly high stiffnesses. Thus, the decision to eliminate these points from the analyses was made.

Qualitatively, there appears to be a stiffness difference between males and females for each condition when their torque by displacement curves are superimposed (figures 23-28). Though not the focus of this study, the greater displacement under equal torque application (i.e., greater laxity for females) is consistent with studies finding laxity differences between males and females (Rosene & Fogarty, 1999).

Although the qualitative comparisons of the various torque by displacement curves looking only at the overall stiffness are suggestive of the changes in overall laxity commonly reported, it is important to also consider the behavior of the joint across the loading cycle. Objective 3 will attempt to quantify these changes.

Objective 3

The stiffness values found in this study are generally in agreement with existing literature. Differences in stiffness between males and females in specific conditions are added to the literature in this study. For convenience, the mean stiffness results of this study are compared with existing findings in table 14. The table represents a quick comparison with representative studies that most closely mimic the conditions tested in this study and include male/female stiffness reporting if available. For more detail of previous research results, see table 1.

Table 14: Comparison of current stiffness values with previous research

Displacement:	Current Stiffness		95% CI		Existing Stiffness	Reference
	(Nm/deg)		Lower	Upper	(Nm/deg)	
Valgus	Female	1.88*	1.60	2.15	3.50 in 1-2 deg flex	(Bryant&Cooke,1988)
	Male	2.32	2.05	2.59	(combined sex)	in-vivo
Varus	Female	1.85*	1.47	2.23	2.94 in 1-2 deg flex	(Bryant&Cooke,1988)
	Male	2.68	2.28	3.07	(combined sex)	in-vivo
Int Rotation	Female	0.53	0.42	0.64	.79 at 15 deg flex	(Hsu et al, 2006)
	Male	0.52	0.41	0.64	1.06 at 15 deg flex	cadaveric
Ext Rotation	Female	0.40	0.34	0.45	.79 at 15 deg flex	(Hsu et al, 2006)
	Male	0.42	0.37	0.48	1.06 at 15 deg flex	cadaveric
Int Rotation	Female	1.56	1.24	1.87	.18 at 20 deg flex	(Markolf et al, 1981)
(WB)	Male	1.56	1.24	1.89	(Combined sex)	cadaveric
Ext Rotation	Female	0.80*	0.55	1.05	.18 at 20 deg flex	(Markolf et al, 1981)
(WB)	Male	1.28	1.03	1.53	(Combined sex)	cadaveric

Valgus

Valgus coupled with external tibial rotation has been previously identified as a possible mechanism for ACL injury (Fung & Zhang, 2003; Olsen et al., 2004). A decreased valgus stiffness for females was observed in this study. On average, females had 19% less overall stiffness in valgus displacement. Of note is the fact that females were less stiff than males principally in the early phase of loading (1.5-3.5 Nm) by as much as 54% (figure 31). This follows a general pattern seen in the data of females starting with less initial stiffness and "catching up" as displacement increases.

If females are less stiff than males early in an imparted displacement, the musculature and joint position could be potentially different than for males when applied torques are increasing. This would give rise to different needs and strategies for controlling potentially injurious motion. This is a seemingly mechanical deficit for females in valgus and would underscore the need for alternate strategies of neuromuscular control in order to augment stiffness in females.

Such strategies have been suggested in that females activate their muscles sooner in response to perturbing activities (Carcia, Shultz, Granata, Gansneder, & Perrin, 2004; Shultz et al., 2001). As such perturbations have been reported to include component of valgus and external rotations (Schmitz, Shultz, Kulas, Windley, & Perrin, 2004), the findings of females activating their muscles sooner in response to a perturbation may be a neuromuscular adaptation to the lesser initial joint stiffness for females demonstrated in the current study.

Varus

Female knees were less stiff in varus displacement (figure 32). In varus, females exhibited a mean overall stiffness that was 31% less than males' stiffness. The similar pattern seen in valgus of less initial stiffness than males and increasing later in the loading phase was observed. A male - female difference was found from 1.5-2.5 Nm of applied torque and was a great as 55% (figure 32). Coupled with the noted valgus stiffness deficit in females, it appears that varus/valgus stiffness for males is greater as hypothesized, particularly in the early phase of loading. The increase in female stiffness in later displacement supports the use of a breakpoint analysis and two distinct phases of stiffness (Markolf et al., 1976), though the amount of applied torque at which this specifically occurs is difficult to determine in existing literature due to the discrepancies previously described in choice of terminal stiffness.

Internal Rotation

For internal rotation there was a significant interaction of sex and incrementally derived stiffness. Both sexes exhibited increasing stiffness at points in the loading with females starting with significantly less initial stiffness (figure 33). This difference only existed through 1.0 Nm of applied torque but reached 91%. During the loading phase the males experienced significant increases only at the 4.5-5.0 Nm loading whereas females experienced significant increases at the 1.5-2.0 Nm loading and the 3.5 – 4.0 Nm loading.

Similar increases are noted in current research (Hsu et al., 2006) at about 1.7 Nm of applied torque. Hsu's breakpoint was chosen qualitatively and applies to cadaveric knees as opposed to the in vivo measures taken in this study. Additionally, the current study found two breakpoints for females and one for males. Also found presently was one specific range of applied torque where male knees were significantly stiffer than female knees in internal rotation, with no overall main effect for sex.

Given this range of decreased female knee IR stiffness, and coupled with a similar pattern in external rotation, it is illustrated that females have stiffness that is considerably less than males in the transverse plane in smaller displacements. However females increase in stiffness as loading increases, equaling the stiffness of males.

Because of lower stiffness early in displacement, it is possible that the strategies for arresting injurious motion are different between males and females. The ACL serves as a restraint of tibial internal rotation (Andersen & Dyhre-Poulsen, 1997; Nielsen, Ovesen, & Rasmussen, 1984). Similar to Valgus, if the tibia is more internally rotated at low load levels, this may predispose the ACL to greater strains at lower loadings. Additionally, it has been demonstrated that females generate less volitional stiffness in response to an internal rotational perturbation (Wojtys, Huston, Schock, Boylan, & Ashton-Miller, 2003). This combination of decreased passive stiffness found in the current study in internal rotation combined with previous finding of females generating less volitional stiffness may be a contributing factor in the ACL injury sex bias.

External Rotation

External rotational stiffnesses for males and females were very similar to the internal rotation stiffnesses, females having a mean stiffness 49% less than that of males through 1.0 Nm of applied torque (figure 34). In external rotation as in internal rotation, female knees increased in stiffness soon after 1.0 Nm of applied torque (increases at the 2.0 -2.5 Nm loading and the 3.5 – 4.0 Nm loading) and equaled that of male knees later in displacement. While this again supports a breakpoint analysis, it shows a discrepancy between studies of where that breakpoint lies (Hsu et al., 2006). Of note is the lagging female knee joint stiffness in the transverse plane. As discussed, female knee joint stiffness catches up to male knee joint stiffness later in the loading cycle. Thus the breakpoint approach to determining joint stiffness may need to be specific to sex. What remains unknown is the effect of lesser stiffness for females in small internal and external rotational displacements.

Internal Rotation in Weightbearing

During weight bearing, internal rotation stiffness was unchanging in males while females experienced significant increases in stiffness at the 3.0 -3.5 Nm loading and the 4.0 – 4.5 Nm loading (figure 35). It should be noted, however, that these changes are noted in ranges where reliability was poor (at points beyond 2.5 Nm of applied torque where $ICC_{2,k} < .50$). Because reliability decreased with increased applied torque, differences between males and females at higher loads are less sensitive to statistical testing. It is possible that given greater reliability, more could be said about the male/female comparison in internal rotational stiffness during weightbearing.

External Rotation in Weightbearing

Females were seen to have less overall stiffness in this condition, with a mean stiffness that was 38% less than that of males and was as low as 50% less early in the imparted displacement (figure 36). The early stiffness deficit for females was similar to that of the valgus condition and persisted from 1.5 Nm to 3.0 Nm of applied torque. Similar to other conditions, the potential exists for female knee joints to be in different positions and in need of different strategies for control once applied torques and forces increase.

Limitations

Although great care was taken to control all aspects of the study, several limitations of the reported data exist. Most importantly, application of torques was not time standardized. This limitation may have led to some of the early aberrant stiffnesses witnessed in the data. Every attempt was made, however, to apply torques evenly and quasi-statically. Though the application was not time standardized, the load was as evenly applied through the displacement as possible.

Also, no control was made for hormone levels in females day to day. Ligamentous laxity has been shown to be affected by fluctuations in hormone levels (Shultz, Kirk, Johnson, Sander, & Perrin, 2004). These fluctuations in joint laxity could greatly affect the reliability of incrementally derived stiffness measurement. Testing reliability in males only may help reveal if sex hormone fluctuations influenced reliability.

Finally, the efforts undertaken to control for neuromuscular effects on stiffness included verbal instructions, practice trials for acclimation of subjects, and visual observation on the part of the researcher. However, the use of EMG was not possible, thus no control could be made for level of muscular activation.

Conclusions

By examining stiffness statistically and at increments during displacement in varus/valgus and internal/external rotational loading, this study offers a more comprehensive assessment of sex differences for knee joint stiffness across a loading cycle. For valgus, varus, and external rotation in weightbearing, female knees exhibited less stiffness overall. Perhaps more revelatory, however, is the interaction that was found for all conditions for sex and applied torque.

Female knee joint stiffness appears to be less than male knee joint stiffness early in the loading phase of these displacements. In some instances, (internal rotation, external rotation, and internal rotation in weightbearing) female knee joint stiffness increases during displacement to “catch up” and equal that of male knee joints.   Therefore, to look at mean stiffness over the entire loading phase may not expose fully a potential factor of ACL injury in females.

This difference, both as an overall difference, and as one exhibited at the initial part of displacement, appears to be mechanical. Research that examines potential anatomical and hormonal causes for these stiffness deficits in females is needed in order to understand the risk factor further. In addition, research into possible strategies utilized by females to control knee joint position and stiffness in functional tasks is needed both to mitigate risk and to understand further the mechanisms whose breakdown adds to the increased injury risk that seems to exist in females.

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APPENDIX A

Main effects and interaction for Valgus loading from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	2842.266	1	2842.266	494.472	.000
Sex	31.945	1	31.945	5.557	.024
Error	206.931	36	5.748		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
Incremental Stiffness	Sphericity Assumed	101.080	16	6.318	6.226	.000
	Greenhouse-Geisser	101.080	2.338	43.229	6.226	.002
	Huynh-Feldt	101.080	2.580	39.176	6.226	.001
	Lower-bound	101.080	1.000	101.080	6.226	.017
Inc. Stiff. * Sex	Sphericity Assumed	62.298	16	3.894	3.837	.000
	Greenhouse-Geisser	62.298	2.338	26.643	3.837	.020
	Huynh-Feldt	62.298	2.580	24.145	3.837	.016
	Lower-bound	62.298	1.000	62.298	3.837	.058
Error(stifflvl)	Sphericity Assumed	584.428	576	1.015		
	Greenhouse-Geisser	584.428	84.176	6.943		
	Huynh-Feldt	584.428	92.886	6.292		
	Lower-bound	584.428	36.000	16.234		

APPENDIX B

Main effects and interaction for Varus loading from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	3045.726	1	3045.726	280.023	.000
Sex	101.894	1	101.894	9.368	.004
Error	358.931	33	10.877		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
appdmmt	Sphericity Assumed	65.434	16	4.090	6.595	.000
	Greenhouse-Geisser	65.434	4.384	14.927	6.595	.000
	Huynh-Feldt	65.434	5.292	12.365	6.595	.000
	Lower-bound	65.434	1.000	65.434	6.595	.015
appdmmt * Sex	Sphericity Assumed	27.151	16	1.697	2.736	.000
	Greenhouse-Geisser	27.151	4.384	6.193	2.736	.027
	Huynh-Feldt	27.151	5.292	5.131	2.736	.019
	Lower-bound	27.151	1.000	27.151	2.736	.108
Error(appdmmt)	Sphericity Assumed	327.439	528	.620		
	Greenhouse-Geisser	327.439	144.663	2.263		
	Huynh-Feldt	327.439	174.629	1.875		
	Lower-bound	327.439	33.000	9.922		

APPENDIX C

Main effects and interaction for Internal Rotation from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	89.325	1	89.325	184.475	.000
Sex	.000	1	.000	.001	.978
Error	16.463	34	.484		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
factor1	Sphericity Assumed	12.458	8	1.557	21.942	.000
	Greenhouse-Geisser	12.458	2.281	5.461	21.942	.000
	Huynh-Feldt	12.458	2.526	4.932	21.942	.000
	Lower-bound	12.458	1.000	12.458	21.942	.000
factor1 * Sex	Sphericity Assumed	2.711	8	.339	4.775	.000
	Greenhouse-Geisser	2.711	2.281	1.189	4.775	.008
	Huynh-Feldt	2.711	2.526	1.073	4.775	.006
	Lower-bound	2.711	1.000	2.711	4.775	.036
Error(factor1)	Sphericity Assumed	19.304	272	.071		
	Greenhouse-Geisser	19.304	77.555	.249		
	Huynh-Feldt	19.304	85.878	.225		
	Lower-bound	19.304	34.000	.568		

APPENDIX D

Main effects and interaction for External Rotation from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	54.403	1	54.403	511.260	.000
Sex	.064	1	.064	.604	.443
Error	3.618	34	.106		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
factor1	Sphericity Assumed	2.655	8	.332	29.454	.000
	Greenhouse-Geisser	2.655	1.839	1.444	29.454	.000
	Huynh-Feldt	2.655	1.996	1.330	29.454	.000
	Lower-bound	2.655	1.000	2.655	29.454	.000
factor1 * Sex	Sphericity Assumed	.647	8	.081	7.172	.000
	Greenhouse-Geisser	.647	1.839	.352	7.172	.002
	Huynh-Feldt	.647	1.996	.324	7.172	.002
	Lower-bound	.647	1.000	.647	7.172	.011
Error(factor1)	Sphericity Assumed	3.065	272	.011		
	Greenhouse-Geisser	3.065	62.522	.049		
	Huynh-Feldt	3.065	67.870	.045		
	Lower-bound	3.065	34.000	.090		

APPENDIX E

Main effects and interaction for Internal Rotation in weightbearing from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	766.360	1	766.360	196.862	.000
Sex	.004	1	.004	.001	.975
Error	128.465	33	3.893		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
factor1	Sphericity Assumed	25.126	8	3.141	8.416	.000
	Greenhouse-Geisser	25.126	2.459	10.220	8.416	.000
	Huynh-Feldt	25.126	2.752	9.130	8.416	.000
	Lower-bound	25.126	1.000	25.126	8.416	.007
factor1 * Sex	Sphericity Assumed	18.745	8	2.343	6.279	.000
	Greenhouse-Geisser	18.745	2.459	7.624	6.279	.001
	Huynh-Feldt	18.745	2.752	6.811	6.279	.001
	Lower-bound	18.745	1.000	18.745	6.279	.017
Error(factor1)	Sphericity Assumed	98.523	264	.373		
	Greenhouse-Geisser	98.523	81.133	1.214		
	Huynh-Feldt	98.523	90.816	1.085		
	Lower-bound	98.523	33.000	2.986		

APPENDIX F

Main effects and interaction for External Rotation in weightbearing from SPSS:

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	349.279	1	349.279	142.427	.000
Sex	18.764	1	18.764	7.652	.009
Error	83.380	34	2.452		

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
factor1	Sphericity Assumed	4.078	8	.510	2.951	.004
	Greenhouse-Geisser	4.078	2.472	1.649	2.951	.047
	Huynh-Feldt	4.078	2.760	1.478	2.951	.041
	Lower-bound	4.078	1.000	4.078	2.951	.095
factor1 * Sex	Sphericity Assumed	8.857	8	1.107	6.410	.000
	Greenhouse-Geisser	8.857	2.472	3.582	6.410	.001
	Huynh-Feldt	8.857	2.760	3.209	6.410	.001
	Lower-bound	8.857	1.000	8.857	6.410	.016
Error(factor1)	Sphericity Assumed	46.975	272	.173		
	Greenhouse-Geisser	46.975	84.061	.559		
	Huynh-Feldt	46.975	93.829	.501		
	Lower-bound	46.975	34.000	1.382		

APPENDIX G

Table of ICC2,K for each condition (including TMS, EMS, BMS):

Valgus:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
1.50	2.99	±	1.69	2.63	±	1.79	0.82	0.75	0.64	0.93	5.13	0.72
2.00	2.49	±	1.60	2.10	±	1.31	0.80	0.71	0.75	0.71	3.55	
2.50	2.27	±	1.45	1.91	±	1.09	0.65	0.86	0.64	0.86	2.43	
3.00	2.12	±	1.16	1.64	±	0.76	0.59	0.74	1.18	0.54	1.38	
3.50	2.18	±	1.47	1.54	±	0.58	0.40	1.14	2.08	0.93	1.59	
4.00	1.81	±	1.05	1.84	±	1.02	0.87	0.38	0.00	0.26	1.88	
4.50	1.76	±	0.73	1.60	±	0.67	0.55	0.49	0.13	0.31	0.67	
5.00	1.67	±	0.50	1.67	±	0.67	0.74	0.34	0.00	0.15	0.55	
5.50	1.82	±	0.88	1.75	±	0.58	0.71	0.48	0.03	0.26	0.86	
6.00	1.80	±	0.80	1.64	±	0.52	0.68	0.45	0.12	0.22	0.68	
6.50	1.61	±	0.66	1.74	±	0.61	0.79	0.29	0.08	0.14	0.67	
7.00	1.49	±	0.57	1.90	±	0.78	0.74	0.39	0.81	0.17	0.76	
7.50	1.92	±	0.96	2.02	±	0.91	0.78	0.45	0.05	0.32	1.43	
8.00	1.97	±	1.07	2.18	±	1.36	0.92	0.38	0.22	0.22	2.76	
8.50	2.02	±	1.11	2.39	±	1.28	0.78	0.60	0.71	0.51	2.35	
9.00	1.93	±	1.01	2.67	±	1.51	0.70	0.82	2.75	0.70	2.61	
9.50	2.59	±	2.38	2.82	±	1.62	0.66	1.39	0.26	2.16	6.16	

Varus:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
1.50	2.50	±	1.47	2.67	±	1.61	0.82	0.68	0.11	0.75	4.00	0.45
2.00	2.35	±	2.04	2.53	±	1.17	0.56	1.34	0.13	1.72	3.80	
2.50	2.14	±	0.89	1.62	±	0.62	0.74	0.46	1.10	0.21	0.96	
3.00	2.03	±	0.84	1.74	±	0.94	0.87	0.34	0.32	0.18	1.40	
3.50	2.11	±	1.02	1.76	±	0.50	0.72	0.54	0.50	0.28	1.02	
4.00	1.98	±	0.76	1.66	±	0.54	0.63	0.46	0.41	0.23	0.64	
4.50	1.94	±	0.65	1.69	±	0.69	0.67	0.39	0.25	0.22	0.68	
5.00	2.19	±	0.97	1.55	±	0.54	0.72	0.52	1.66	0.22	1.00	
5.50	2.23	±	0.88	1.86	±	0.78	0.64	0.53	0.55	0.36	1.02	
6.00	2.06	±	0.73	1.69	±	0.58	0.00	0.73	0.55	0.44	0.44	
6.50	2.09	±	0.84	1.75	±	0.70	-0.48	1.02	0.47	0.72	0.49	
7.00	2.30	±	0.86	2.00	±	0.88	-0.06	0.90	0.37	0.78	0.74	
7.50	1.96	±	0.57	1.89	±	0.87	0.06	0.84	0.02	0.52	0.55	
8.00	2.20	±	0.88	1.76	±	0.30	-0.14	0.94	0.79	0.46	0.40	
8.50	2.39	±	0.80	2.12	±	0.77	0.83	0.33	0.28	0.17	1.06	
9.00	2.26	±	0.87	1.81	±	0.66	0.83	0.36	0.79	0.15	1.05	
9.50	2.88	±	1.38	1.81	±	0.47	0.33	1.13	4.57	0.80	1.32	

Internal Rotation:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
0.50	0.35	±	0.23	0.38	±	0.18	0.88	0.69	0.00	0.01	0.08	0.43
1.00	0.34	±	0.16	0.41	±	0.19	0.85	0.65	0.01	0.01	0.05	
1.50	0.32	±	0.11	0.40	±	0.16	0.72	0.56	0.02	0.01	0.03	
2.00	0.40	±	0.14	0.43	±	0.17	0.91	0.68	0.00	0.00	0.04	
2.50	0.44	±	0.13	0.58	±	0.20	0.15	0.09	0.07	0.03	0.03	
3.00	0.52	±	0.13	0.68	±	0.52	0.41	0.24	0.09	0.11	0.19	
3.50	0.67	±	0.43	0.51	±	0.09	-0.65	-0.45	0.08	0.12	0.07	
4.00	0.64	±	0.11	0.68	±	0.29	0.42	0.24	0.01	0.04	0.06	
4.50	0.63	±	0.10	0.62	±	0.18	0.18	0.11	0.00	0.02	0.02	

External Rotation:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
0.50	0.45	±	0.35	0.34	±	0.22	0.86	0.69	0.05	0.02	0.15	0.64
1.00	0.37	±	0.22	0.26	±	0.12	0.77	0.68	0.04	0.01	0.05	
1.50	0.33	±	0.15	0.30	±	0.10	0.90	0.83	0.00	0.00	0.03	
2.00	0.36	±	0.18	0.31	±	0.10	0.74	0.67	0.01	0.01	0.04	
2.50	0.37	±	0.14	0.34	±	0.12	0.77	0.71	0.00	0.01	0.03	
3.00	0.44	±	0.12	0.35	±	0.09	0.42	0.39	0.04	0.01	0.02	
3.50	0.45	±	0.12	0.38	±	0.11	0.61	0.57	0.02	0.01	0.02	
4.00	0.49	±	0.14	0.45	±	0.12	0.39	0.36	0.01	0.01	0.02	
4.50	0.51	±	0.10	0.52	±	0.16	0.26	0.25	0.00	0.02	0.02	

Internal Rotation in weightbearing:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
0.50	1.56	±	1.15	1.24	±	0.96	0.92	0.33	0.36	0.16	2.08	0.40
1.00	1.31	±	0.72	1.22	±	0.60	0.86	0.27	0.03	0.11	0.78	
1.50	1.39	±	0.52	1.53	±	0.81	0.60	0.51	0.07	0.27	0.66	
2.00	1.03	±	0.35	1.47	±	0.63	0.72	0.33	0.66	0.09	0.43	
2.50	1.13	±	0.34	1.55	±	0.86	-0.07	0.89	0.63	0.44	0.41	
3.00	1.19	±	0.52	1.98	±	1.39	0.49	0.99	2.19	0.71	1.50	
3.50	1.54	±	0.46	2.69	±	3.26	0.27	2.78	4.65	4.56	6.29	
4.00	1.39	±	0.36	2.03	±	1.26	-0.07	1.30	1.44	0.88	0.82	
4.50	1.56	±	0.52	2.35	±	1.02	-0.14	1.09	2.17	0.71	0.61	

External Rotation in weightbearing:

Torque	Day 1 Mean		SD	Day 2 Mean		SD	ICC 2,K	SEM	TMS	EMS	BMS	AVG ICC
0.50	1.93	±	1.78	1.17	±	0.98	0.81	0.77	2.30	0.58	3.54	0.77
1.00	1.49	±	1.34	1.02	±	0.67	0.76	0.65	0.88	0.41	1.84	
1.50	1.50	±	1.25	0.90	±	0.54	0.76	0.61	1.42	0.31	1.54	
2.00	1.07	±	0.71	0.86	±	0.45	0.88	0.24	0.17	0.07	0.65	
2.50	1.13	±	0.94	0.89	±	0.44	0.70	0.51	0.23	0.25	0.84	
3.00	0.96	±	0.45	0.84	±	0.37	0.90	0.14	0.06	0.03	0.31	
3.50	1.21	±	0.80	0.90	±	0.28	0.55	0.54	0.38	0.22	0.50	
4.00	1.05	±	0.44	0.87	±	0.28	0.76	0.22	0.13	0.05	0.22	
4.50	0.95	±	0.28	0.98	±	0.41	0.81	0.18	0.00	0.04	0.20	