ASSESSMENT OF TOTAL KNEE REPLACEMENT PERFORMANCE USING MUSCLE-DRIVEN DYNAMIC SIMULATIONS

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by
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Abstract

The goal of a total knee replacement (TKR) procedure is to alleviate pain caused by the degradation of cartilage in the knee, and to restore normal knee function for an extended period of time. Modern TKR has proven to be a safe and effective means to restore the quality of life for patients, and the number of TKR procedures performed per year continues to rise. The motivation of this dissertation was the effective use and development of novel muscle-controlled dynamic simulations of TKR that can characterize knee joint loads.

The first study involved the use of a previously developed dynamic elastic foundation model of an “Oxford Rig.” The Oxford Rig is a commonly used cadaveric jig that simulates loaded knee flexion under quadriceps control. Undesirable collateral ligament laxities were introduced into the knee joint, and the model was utilized to predict the kinetic and kinematic effects of a surgical method, called joint line elevation, that is commonly used to address ligament laxity. The result of these simulations concurred with guidelines that have been qualitatively established through years of clinical experience.

The second study used the same Oxford Rig simulation to investigate an alternative surgical method that is used to address collateral ligament laxity, called femoral upsizing. To date, the mechanical implications of mismatching femoral and tibial component sizes has received little attention. The results of this study suggested that the magnitude of collateral ligament laxity should be indicative of the method used to address collateral ligament laxity. Specifically, if ligament laxity is mild it is suggested that femoral upsizing should be avoided, but if more moderate or severe ligament laxity is encountered, a combination of femoral upsizing and joint line elevation should be employed.

The third study augmented the Oxford Rig model to investigate the effects of changing the fixation point of linear displacement quadriceps actuators. A review of the literature revealed that many labs are using Oxford Rigs to investigate TKR
mechanics, but the location of the quadriceps actuator has varied from lab to lab. It was found that the fixation point of the quadriceps actuator had influence over the joint loads within the knee, but the kinematics of the knee joint were relatively unaffected by changes in actuator position.

The final study developed a novel methodology to predict tibiofemoral contact forces during gait. The accurate simultaneous prediction of muscle forces and tibiofemoral contact forces continues to be a challenging task for researchers, and is the goal of the “Grand Challenge Competition to Predict In Vivo Knee Loads,” hosted by the ASME Summer Bioengineering Conference. To address this challenge, we developed an dynamic elastic foundation model that employed a dual-joint paradigm to represent the knee joint, and incorporated in vivo kinematics and EMG data as inputs. The simulations provided kinematics, muscle forces, and tibiofemoral contact forces that compared well with data collected from an instrumented implant.

In conclusion, this dissertation described several novel computational simulations that were used to investigate TKR kinematics and kinetics during dynamic activities. These simulations have the potential to be used by surgeons to elucidate the effects of choices made within the operating room, and by implant designers to evaluate implant designs before they are put into production. Although TKR is effectively restoring the quality of life to many patients, the post operative function of the implants can be improved with computer models by optimizing surgical techniques and by streamlining the implant design cycle.
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Chapter 1

Introduction

1.1 Background

Total knee replacement (TKR) was first proposed in the nineteenth century as a means to alleviate pain from osteoarthritis. Since that time, continual advancements have been made to the surgical procedures, designs, and materials that are associated with TKR. In modern TKR, knee joint surfaces that have suffered cartilage degeneration are removed and replaced with metal and plastic implants. The diseased portion of the femur is replaced with a monolithic component that is composed of a metal alloy (e.g. cobalt-chrome). Resected tibial bone is replaced with a metallic baseplate, which holds a plastic bearing made of ultra high molecular weight polyethylene (UHMWPE). Finally, the diseased patellar bone is replaced with a UHMWPE implant. This design scheme relies upon soft tissues for constraint, and provides the tibiofemoral and patellofemoral joints the 6 degrees-of-freedom that they are afforded in a natural knee. Implants can be classified into two general groups, based on the treatment of the posterior cruciate ligament: cruciate-retaining (CR) knees, which maintain the natural PCL, and posterior substituting (PS) knees, which have no PCL, and instead rely on implant geometry to provide the contraint normally provided by the ligament.

The number of TKR recipients in the United States is on the rise, as modern implant designs have proven to be clinically effective with patient satisfaction rates between 90%-95% (Colizza et al., 1995; Emmerson et al., 1996; Ranawat et al.,
and implant lifespans lasting between 10–15 years (Scott and Volatile, 1986; Stern and Insall, 1992; Font-Rodriguez et al., 1997; Weir et al., 1996). The aging of the “baby boomer” population has coincided with the clinical success of the TKR. To this end, the number of TKR procedures performed in a given year has been steadily increasing. For example, the National Hospital Discharge Survey estimates that 300,000 TKR procedures were performed in the United States 2003. In a recent survey conducted by The American Academy of Orthopaedic Surgeons, it was estimated that 581,000 procedures were performed in 2010. This growth is expected to continue, as projections estimate that 3.48 million procedures will be performed in 2030 (Kurtz et al., 2007).

Patients over the age of 65 continue to represent the majority of the TKR patient population. Consequently, the needs of an elderly patient population has dictated the design goals of TKR implants. However, people between the ages of 45–64 now represent a substantial portion of the TKR patient population. This group of younger patients have a more active lifestyle, which requires the implant to have a longer lifespan and provide a larger range of motion. Several TKR designs have sought to meet these demands (Jones and Huo, 2006), but an “overly active” lifestyle has been cited as the second leading factor leading to revision knee surgery within the first two years following primary surgery (Heck et al., 1998).

The majority of TKR research relies upon physical methodologies, such as dynamic simulators, implant retrievals, and in vivo experiments. These methods have been employed since the introduction of TKR, and are largely responsible for the evolution of designs to their current form. Physical testing, however, is an expensive and time-consuming endeavor, which prevents these methods from being incorporated into an efficient design cycle. Although computer simulations have their own set of limitations, they have the potential to be used effectively within a design cycle because they are fast and relatively inexpensive. The utility of computational modeling of TKR has been recognized by the orthopaedic community, and number of computational TKR studies is growing rapidly.

Based on the excellent clinical history of TKR, it could be argued that there is little need for further research that is focused on the subject. However, the three most common causes for TKR revision surgeries are: 1) excessive polyethylene
wear, 2) loosening of components, and 3) knee joint instability (Sharkey et al., 2002), and all of these issues have the potential to be addressed through improvements in surgical procedures and implant design. Furthermore, a younger patient population has caused orthopaedic implant manufacturers to reconsider the design forms of TKR implants, and validation of new implant designs must be achieved in a timely and cost-effective manner. Finally, it should be noted that if positive outcome rates remained relatively static at 90%–95%, the number of individuals affected by unsatisfactory outcomes will continue to increase as the overall patient population grows.

1.2 Motivation and Rationale

The goal of any TKR procedure is to alleviate pain caused by osteoarthritis and to restore functionality of the lower limb for an extended period of time. The long-term success of a TKR is largely determined by three separate groups: 1) the manufacturers, who must design an effective implant; 2) the surgeons, who must implant the components with exacting precision; 3) the physical therapist and patient, who must be diligent in their rehabilitation efforts. If any one of these groups fails to do their job adequately, revision surgery may be required because of continued knee pain, a limited range of motion, or a shortened lifespan of the implant. Computational models offer a unique insight into TKR performance because they can estimate the location and magnitude of joint loads during a kinematic activity. These estimates are impossible to make in vivo, and have the potential to serve as predictive measures of long-term TKR success.

The motivation of this dissertation was the effective use and creation of novel muscle-controlled dynamic simulations of TKR that can characterize joint loads during everyday activities. The development of the current paradigm pairs a dynamic elastic foundation model with a popular muscle force optimization scheme, known as the computed muscle control (CMC) algorithm (Thelen et al., 2003; Thelen and Anderson, 2006). To date, CMC relies upon idealized joints to calculate muscle forces, and therefore cannot be used with contact-based joints. Our previous work developed such a computational model, which simulated the mechanics
of an “Oxford Rig,” a bench-top cadaveric test that creates knee flexion with a single quadriceps muscle force. The methods used in this model were used as a stepping stone for a second, more complex computational model, which included multiple muscles, and aimed to simultaneously simulate the kinematics of gait while predicting tibiofemoral contact loads.

1.3 Specific Aims

The specific aims of this dissertation were to:

1) Investigate several surgical methods that are currently used to address collateral ligament laxity in TKR patients. The effects of the choices made by a surgeon can be identified by examining the kinematics and kinetics of an Oxford Rig Simulation.

2) Develop a more realistic Oxford Rig model (i.e. one that includes more than one active muscle).

3) Develop a computational modeling paradigm that can incorporate estimations of articular contact forces into a muscle-driven simulation of any given \textit{in vivo} movement.

This dissertation consists of four separate studies, and describes the use and development of two muscle-driven dynamic computer models that were used to investigate several relevant issues in TKR. The first computer model simulates an “Oxford Rig,” a commonly used cadaveric rig that simulates a knee flexion, while the second model simulates the mechanics of \textit{in vivo} gait. Both of these models may be used to answer clinical questions, as they can be used to determine the effects of certain choices a surgeon must make during a total knee arthroplasty procedure. The models also have the potential to be used by implant designers, as they provide detailed characterizations of implant loads during a motion, and are not financially burdensome or overly time-consuming.

The Oxford Rig model was employed in the first, second, and third studies in
The Oxford Rig is a mechanical assembly that creates flexion and extension of a cadaveric knee by allowing the pelvis to translate on a set of vertical rails. The flexion and extension of the knee are controlled by quadriceps muscle forces. The use of the Oxford Rig is particularly attractive because the model topology permits 6 degrees-of-freedom for both the tibiofemoral and patellofemoral joints (Zavatsky, 1997). The rig is widely used in the orthopaedic community, as it effectively represents the kinematics of challenging everyday tasks, such as rising from a chair or performing a step-up task. The collection of experimental data from an Oxford Rig is a time-consuming task, which requires a surgeon to be on-site to ensure the proper placement of the implants. A computational model of an Oxford Rig could be of use to surgeons and implant designers. Implant placements and component geometries can be manipulated easily, while all other variables are held constant. A combination of variables can be manipulated without the need for a series of cadavers, which allows for the quick and effective creation of sensitivity studies that cannot be created with an experimental rig.

The first study used the Oxford Rig simulation to investigate the effects of joint line elevation in primary TKR. In the course of a total knee arthroplasty, a surgeon must properly balance the knee ligaments in flexion and extension to ensure that the knee joint provides stability throughout the entire range of motion. If the collateral ligaments are lax, the ligaments are out of balance, and the knee becomes unstable. A common method used to achieve ligament balance is joint line elevation, where the tibial component and femoral components are both moved proximally. Simulations of varying degrees of joint line elevation were run to examine the kinematic and kinetic effects on CR and PS knees. Recommendations were then given to aid in the decision making process for surgeons who encounter lax collateral ligaments during TKR.

The second study used the Oxford Rig simulation to investigate the effects of femoral component upsizing in revision TKR. Similar to the previous study, collateral ligament laxity can present a challenge to surgeons during the course of a revision surgery. The surgeon can restore ligament balance with joint line elevation or, alternatively, use an oversized femoral component. Simulations were run to compare the effects of joint line elevation and femoral upsizing in revision PS
knees. Again, recommendations were given to aid in the decision making process for surgeons who encounter lax collateral ligaments during revision TKR.

The third study examined the effects of changing the location of the quadriceps muscle actuator in Oxford Rigs. A review of the literature revealed that the model topology of various Oxford Rigs remains constant, but the fixation point of the quadriceps actuator varies from lab to lab. Simulations were run with the quadriceps actuator being affixed to the femur, pelvis, and ground reference frames. The differences between the kinematics and kinetics of the results were compared. Suggestions were made in regards to the validity of conclusions that can be made with Oxford Rigs, based on quadriceps actuator location.

The fourth study in this dissertation was primarily motivated by the “Grand Challenge Competition to Predict In Vivo Knee Loads,” hosted by the ASME Summer Bioengineering Conference. The organizers of this event acknowledge the importance of modeling of joint contact forces, and sought to motivate new modeling methods by creating this competition. Competitors were provided with an extensive set of in vivo data that was collected from a subject with an instrumented knee implant. The goal of the competition was to create a computational model that could faithfully reproduce the kinematics of gait while accurately predicting the medial and lateral tibiofemoral loads that were recorded with the instrumented implant. The model that was developed for the competition included 5 segments and 13 active muscles, and incorporated the CMC algorithm by employing a dynamic dual-joint modeling approach. The kinematic results of the model were compared to collected in vivo measurements and, the normalized tibiofemoral contact loads were compared to the results a previous study.

All of the studies within this dissertation consist of computational methods which estimate joint contact loads during everyday tasks. In this series of studies, these models have been tailored to examine specific research questions for one subject and employ a very limited variety of implant geometries. However, the methods presented in this dissertation can be augmented to address separate research questions in a relatively quick manner, and have the potential to be used in a pre-operative setting or within a manufacturer’s design cycle.
1.4 Overview

This thesis is organized to provide the reader with the necessary background information in Chapter 2, before discussing the four separate studies in Chapters 3–6. Chapter 2 provides a brief overview of the clinical record of TKR, and summarizes two of the most basic design forms of implants. An outline of the experimental studies that are relevant to TKR research is provided, and a more thorough review of computational models in biomechanics and TKR is presented. Finally, an exhaustive synopsis of research involving dynamic elastic foundation models of TKR is given. Chapters 3–6 are presented as separate studies, each with their own introduction, methods, results and discussion. Chapter 3 focuses on the effects of joint line elevation with simulations of an Oxford Rig. Chapter 4 also uses simulations of an Oxford Rig to investigate the effects of femoral upsizing. Chapter 5 analyzes the kinematic and kinetic changes to the knee joint that are associated with changing the fixation point of the quadriceps muscle actuator in an Oxford Rig. Chapter 6 details the development of a novel dual-joint modeling technique that effectively incorporated contact-based joint mechanics with the CMC algorithm. Chapter 7 attempts to recap the four separate studies, and unify the results in an effort to draw conclusions on how dynamic elastic foundation modeling can be predictive of TKR performance. Appendix A is a published article in The Journal of Orthopaedic Research, which is the result of a cooperative study conducted with a group from the Ohio State University. The manuscript investigates how systematic axial mal-alignment of TKR components affects the kinetics and kinematics of CR and PS knees in an Oxford Rig simulation. Appendix B is the supplementary material that was provided to The Journal of Orthopaedic Research for the manuscript in Appendix A. The material focuses primarily on the validation testing that was performed with mechanical and computational models of the Oxford Rig. Appendix C gives the reader further insight into the evolution of the dual-joint modeling paradigm, and contains results from normal gait and trunk-sway gait that were not included in Chapter 6.
2.1 Overview of Total Knee Replacement

Total knee replacement (TKR) was first proposed in the nineteenth century and attempted to simplify knee mechanics by limiting joint motion to rotation about a mediolateral axis. The hinged designs had poor clinical outcomes, and were quickly replaced with rotating hinge designs that allowed internal/external rotation to the joint. This improvement provided a more natural feel for the patient as well as improved performance in clinical follow-up studies (Walker and Sathasivam, 2000; Unwin and Walker, 1999). Flexion, extension, and internal/external rotation of the knee joint are the predominant knee motions; however, it has been suggested that translation and abduction/adduction should not be completely restricted, or the heavily constrained implant will generate large reaction forces at the implant-bone interface, causing components to loosen over time (McAuley and Engh, 2003; Nelson et al., 2003).

Because of the limited clinical success of mechanically hinged knees, Gunston introduced a prosthetic knee design that incorporated a metal-on-plastic bearing, which had recently gained popularity in total hip replacements (Andriacchi and Hurwitz, 1997). The design eliminated mechanical linkages from the joint, and instead relied upon soft tissues to provide hinge mechanics. The non-linked design requires subchondral bone and diseased cartilage to be cut away with a bone saw and replaced with either a metal component (femur) or a plastic bearing.
(tibia and patella) (Figure 2.1). This concept allows for translations and abduction/adduction of the knee to occur freely, which reduces stress and wear on the components and the bones to which they are attached (Freeman et al., 1973).

Figure 2.1. The knee before and after a non-linked TKR procedure. The diseased portion of the femur is removed and replaced with a metallic component. The diseased tibial bone is removed and replaced with a metallic baseplate and a plastic tibial bearing. Patellar bone replacement is not shown in this diagram. Source: www.mdconsult.com

Advancements in designs and materials have led to increased life spans for modern TKR implants. Femoral components and tibial baseplates are typically made of titanium or cobalt-chrome steel alloys, while tibial bearings and patellar buttons are usually made of ultra high molecular weight polyethylene (UHMWPE). State-of-the-art designs have increased the amount of conformity between the articulating surfaces so that the contact area between implants is increased and loads are more evenly distributed across the bearing surface (Insall, 1998).

Long-term clinical results can give a clear sense of the effectiveness of TKR designs. Several studies have shown patient satisfaction rates to be between 90%–95% (Colizza et al., 1995; Emmerson et al., 1996; Ranawat et al., 1997; Callaghan, 2001). Over 90% of prosthetic knee recipients claim significant reduction in pain for 10–15 years after surgery (Scott and Volatile, 1986; Stern and Insall, 1992; Font-Rodriguez et al., 1997; Weir et al., 1996). Because of these high rates of positive clinical outcomes, there has been a steady increase in the number of TKRs
performed, with some patients younger than 40 years old (Wickiewicz et al., 1983; Lonner et al., 2000) (Figure 2.2). In a recent survey conducted by the American Academy of Orthopaedic Surgeons, an estimated 581,000 TKR procedures were performed in 2010. The rise in the number of TKR procedures performed in a year is expected to continue, as Kurtz et al. (2007) projects an estimated 3.48 million procedures will be conducted in 2030.

![TKR Procedures in the United States 1996-2007](image)

**Figure 2.2.** Rates of TKR procedures performed in two-year intervals in the United States, per 10,000 people. Data are based on the results of the National Hospital Discharge Survey (2010).

Patients over the age of 65 continue to represent the majority of the TKR patient population. A typical elderly patient will desire a knee design that provides utmost stability to prevent the possibility of falling, and minimizes the amount of muscular effort required to accomplish simple tasks, such as walking, rising from a chair, or traversing stairs. These design goals have dictated the design of modern knee implants, which aim to increase the extensor mechanism moment arm. A trade-off exists, however, as methods used to maximize extensor mechanism efficiency limit the range of motion (ROM) of the implant.

Because more TKR patients are young and active, several TKR designs have been tailored to meet the increased functional demands placed on the implant.
(Jones and Huo, 2006). A properly sized and placed implant should provide roughly 120° of knee flexion (Sultan et al., 2003), but designs geared towards the younger patient population seek to increase the ROM of the implant beyond this amount so that patients can perform the variety motions associated with a physically active lifestyle. These designs also suffer from trade-offs, as conformity between the femoral and tibial components will help minimize wear and maximize implant lifespan; however, a TKR design with a large degree of conformity will not permit a large range of motion.

2.2 PCL Treatment

During a TKR operation, the surgeon must make the decision of whether to keep or remove the natural PCL. If the PCL is retained, a cruciate-retaining (CR) prosthesis must be implanted. Alternatively, if the PCL is removed, a posterior stabilizing (PS) prosthesis will be employed. There are several key design features to both types of implants, as well as inherent advantages and disadvantages with each design. These topics have been researched and documented extensively, but is beyond the scope of this literature review. The following is a very brief summary of the major differences in design goals of CR and PS implants.

2.2.1 Cruciate-Retaining Implants

If the PCL is kept intact during surgery, a CR model will be implanted. The first distinguishing feature of the CR knee is the profile of the tibial insert. It is relatively flat, allowing a large amount of freedom for anterior/posterior (A/P) translation and internal/external (I/E) rotation. The second distinguishing feature of a CR knee is the cutout that exists on the posterior sides of the femoral and tibial components. This gap allows room for the intact PCL to function. (Figure 2.3)

Despite the presence of the PCL, it is common for a CR knee be implanted such that anterior translation is further restricted. To do this, the implant can be positioned with the anterior edge of the tibial bearing uphill from the posterior edge, at a slope between 3°–10°. The slope purportedly provides resistance to
Figure 2.3. A CR implant is recognizable by the flat tibial bearing surface (dark green) and the cutout that exists on the posterior side of the femoral component (light green). The natural PCL prevents excessive anterior translation of the femur.

anterior translation, while providing a larger range of motion, especially in deep flexion (Walker and Sathasivam, 2000; Ostermeier et al., 2006; Yoshiya et al., 2005). The posterior slope is intended to keep the center of rotation of the knee from translating towards the patella. It is argued that this posterior positioning of the femur on the tibial plateau maximizes the extensor mechanism moment arm (Soudry et al., 1986), and therefore reduces the muscular force needed to provide a given knee extension moment.

The cruciate retaining (CR) knee model has enjoyed great clinical success. In 2000, almost two-thirds of the existing TKR patients in the world had received a CR implant (Walker and Sathasivam, 2000). The surgeons’ preference to use CR models can be attributed to the desire to keep the natural PCL intact if possible. The natural PCL is useful not only because it is a valuable proprioceptor (Attfield et al., 1996), but also because it prevents the femur from translating anteriorly off of the tibial plateau (Walker and Sathasivam, 2000).
2.2.2 Posterior Stabilized Implants

If the natural PCL is sacrificed during surgery, a posterior stabilized (PS) design is implanted. The model must physically limit A/P translation of the femoral component that an intact PCL would have otherwise provided. This is achieved with a cam-post mechanism, most easily recognized by a vertical intercondylar post emerging from the tibial bearing. The cam housing is located in the femoral intercondylar notch that is simply a gap in CR designs. (see Figure 2.4)

Figure 2.4. A PS implant is recognizable by the vertical intercondylar post emerging from the tibial bearing (dark blue). The cam housing, located on the posterior side femoral component (light blue), interacts with tibial post to prevent excessive anterior translation.

Several design characteristics have caused the PS knee to gain clinical popularity within recent years (Walker and Sathasivam, 2000). First, it has been shown that the range of flexion can be increased $10^\circ$–$15^\circ$ with the use of PS designs in lieu of a similar CR implant (Walker, 1980). It has also been argued that well-designed PS knees can provide a larger moment arm for the quadriceps than found in sloped CR knees (Walker, 1980). This makes everyday activities, such as rising from a chair or ascending a flight of stairs, easier for the patient.
The PS knee is easier for surgeons to implant with accurately and repeatability. Unlike CR implants, PS models are most effective when the tibial resection plane is made horizontally (Piazza et al., 1998). In the operating room, a horizontal cut is easier to make than one on a posterior slope. Because the PCL has been resected, it does need to be accounted for during tibial plateau cuts, and does not have to be considered during the ligament balancing process. The cam-post mechanism provides a controlled amount of A/P femoral translation as a function of flexion angle (termed femoral rollback), while CR knees have more variable A/P translation patterns because of varying posterior slope angles and PCL lengths (Wilson et al., 1996). It is for these reasons that Yoshiya et al. (2005) contends that PS knees provide more repeatable and desirable kinematics than CR knees.

2.3 Overview of Experimental Studies

Despite the clinical success of modern TKR (patient satisfaction in the range of 90%–95%, and implants lifespan in the range of 10–15 years) a positive outcome of a total knee arthroplasty cannot be guaranteed. Issues such as infection and arthrofibrosis are ever-present dangers that are inherent with any surgical procedure, however, the three most common causes for TKR revision surgeries are: 1) excessive polyethylene wear, 2) loosening of components, and 3) knee joint instability (Sharkey et al., 2002). These three issues have motivated a wide variety of research methods, all of which have led to steady improvements to component designs, materials, and surgical techniques.

The clinical success of modern TKR is attributable to the large of amount of research that has been devoted to the subject. A quick search for “total knee replacement” on Google Scholar yields over 329,000 results. The majority of TKR research relies upon physical methodologies, such as dynamic simulators, implant retrievals, and in vivo experiments. Conversely, a relatively small portion of TKR research employs computational methods, but this number is growing rapidly. This dissertation concerns computational modeling of TKR mechanics, so physical investigations will summarized briefly, and computational research will be explored in a more thorough manner.
2.3.1 Dynamic Functional Tests

The most straightforward dynamic simulators are used to investigate the constraint provided by the implant geometry (Thatcher et al., 1987; Klein et al., 2003). According to the American Society of Testing and Materials International (2005), these mechanical simulators prescribe anterior/posterior translations and internal/external rotations and record the subsequent reaction forces and torques provided by the implant geometry. Orthopaedic implant manufacturers commonly use “range of constraint” experiments to quickly estimate constraint forces. However, range of constraint tests do not consider the effects of soft tissues, nor are the kinematics intended to be representative of human motions, which limits the clinical utility of these simulators.

Other dynamic simulators are designed with the intent to replicate the kinematics of everyday tasks. Some of these simulators rely upon muscle forces to replicate the motion of the implants (Shaw and Murray, 1973; Rovick et al., 1991), while others apply forces directly to the femur or tibia, without implementing muscular forces (Walker and Hsieh, 1977; Pappas and Buechel, 1994; DiAngelo and Harrington, 1992). Over the last forty years, improvements have been made to the designs of dynamic simulators. Walker et al. (1997) created the Stanmore knee simulator, which has the proven ability to accurately reproduce clinically observed wear patterns. This jig became the gold-standard of the industry, and a variety of similar knee simulators are currently used to investigate implant performance during flexion/extension movements, including the Kansas knee simulator (Guess and Maletsky, 2005), the Johns Hopkins knee simulator (Maletsky and Hillberry, 2005), and the Purdue knee simulator (Maletsky and Hillberry, 2005).

The jig modeled in parts of this dissertation is the Oxford Rig (Bourne et al., 1978) (Figure 2.5). This versatile rig has been used to study a spectrum of questions regarding TKR mechanics (Singerman et al., 1999b,a, 1995a; Catani et al., 2009; Walker and Haider, 2003; More et al., 1993; Patil, 2005; D’Lima, 2000; D’Lima et al., 2001; D’Lima, 2001; Browne et al., 2005; Ezzet et al., 2001; Anglin et al., 2010; Mountney et al., 2008; Quentelier et al., 2008; Luyckx et al., 2009; Ferrari et al., 2003), as it can be used to re-create a variety of deep knee flexion exercises, such as riding a bicycle, rising from a chair, or climbing stairs.
Figure 2.5. An Oxford Rig, which was used for the purposes of validation of the computational model. Cadaveric elements of the rig were eliminated to reduce the number of assumptions that were made within the computational simulation. The femur and tibia were represented by steel rods, the quadriceps muscle was a steel cable, and the patellar tendon was a nylon strap. A universal joint was used at the hip, and the ankle was represented by a ball joint. Vertical translation of the pelvis sled created knee flexion.

The Oxford rig is especially appealing to investigators for a variety of reasons. First, the mechanical aspects of the rig are fairly simple, and include a U-joint hip, a ball-joint ankle, and a vertical slider used to vary the height of the pelvis. The hip joint allows flexion/extension and abduction/adduction. The ankle joint al-
lows flexion/extension, abduction/adduction, and internal/external rotation. The vertical slider allows the pelvis to move from a squat to stand position by applying a quadriceps force. This model topology provides the tibiofemoral and patellofemoral joints with six degrees-of-freedom (Zavatsky, 1997). The rig can be used with a mechanical representation of the lower limb, as shown in Figure 2.5, or it can be fitted with a cadaveric specimen to provide realistic soft tissue forces. It may also be fitted with transducers that can measure tibial rotation, contact force distribution, and relationships between quadriceps force and load (Fitzpatrick, 1989).

2.3.2 Implant Retrievals

Revision and post-mortem implant retrievals provide valuable information regarding wear and failure, as the implants are subjected to a wide variety of human motions throughout their lifespan. Throughout the history of TKR, this method has proven to be valuable for determination of implant failure and wear (Hood et al., 1983; Bartel et al., 1986; Wright et al., 1992; Wasielewski et al., 1994; Rao et al., 2002, 2010). However, these implants often take years to obtain, and design of the implant has already been mass-produced by the time of retrieval.

2.3.3 In Vivo Methods

The earliest possible time to test the functional performance of the implant in vivo is during surgery. A surgeon can roughly balance the collateral ligaments by ensuring that: 1) the femoral component does not readily translate off of the tibial plateau in extension, and 2) the knee joint can reach approximately 120° of flexion and does not feel overly “tight” in this pose (Mihalko et al., 2003; Sultan et al., 2003). Zalzal et al. (2004) and Skrinskas et al. (2003) have further refined this process by incorporating real-time software into balancing jigs. The results of both studies suggest that ligament tensioning devices can effectively reduce post-operative knee joint instability.

Post-operative in vivo research methods often involve measuring the kinematics of a subject performing everyday tasks and extracting TKR mechanics from this data. For example, many studies require patients to perform a knee flexion task,
such as squatting, kneeling, or walking on a treadmill (Delport et al., 2006; Haas et al., 2002; Oakeshott et al., 2003; Watanabe et al., 2004; Dennis et al., 2003, 2005). In these studies, moments about joints are calculated with inverse dynamics, and the kinematics of the implant are determined with the use of videofluoroscopy and 2-D to 3-D model registration techniques. These studies have been repeated with regularity for good reason. Results are sensitive enough to identify changes in kinematics of the prosthetic knee that caused by certain design features of the implant (Walker et al., 2002; Haas et al., 2002; Yoshiya et al., 2005; Oakeshott et al., 2003).

Research involving instrumented implants is rarely conducted because of the technical and financial challenges it presents. This research is further stifled by the difficulties associated with recruiting patients who are willing to receive such implants, and the small number of implant designs that have been paired with measurement devices. Despite these obstacles, several groups have successfully used instrumented implants to generate real-time information regarding knee joint loads during a variety of activities, (D’Lima et al., 2008; Varadarajan et al., 2008; Kaufman et al., 1996; Mundermann et al., 2008; Kirking et al., 2006). Measurement of instrumented TKR kinematics still requires fluoroscopic imaging techniques. Many advances have been made in this area (Hanson et al., 2007; Li et al., 2006; Suggs et al., 2008; Li et al., 2008), but determination of the implant kinematics for common tasks, such as over-ground walking, remains a challenging task because of the small capture volume of fluoroscopic cameras.

While all of the methods listed above have significantly improved the overall quality of TKR implants, all suffer from one or more of the following shortcomings, which prevent them from being incorporated effectively into the implant design cycle: 1) testing is time-consuming; 2) materials required for testing are financially burdensome; 3) results do not provide a complete characterization of the contact mechanics. Computer simulation is a potentially viable alternative method for determination of implant forces, as it does not suffer from the same physical, financial, and technical limitations of traditional investigative methods.


2.4 Simulation of Human Movement

The use and complexity of computational simulations of human motion has grown significantly in recent years. A good perspective of the evolution of musculoskeletal modeling can be traced through investigations of walking. Initial computer models used a simple inverted pendulum paired with a feedback controller to investigate the stability of bipedal locomotion (Hemami et al., 1973). Mochon and McMahon (1980) developed a planar ballistic walking model, consisting of three links (stance leg, thigh and shank of swing leg), which provided further understanding of the relationship between walking cadence and body posture during the swing phase of gait. Yamaguchi and Zajac (1990) created the first 3-D musculoskeletal model of walking with an 8 degree-of-freedom lower extremity model that had 8 muscles. Using an open-loop, trial-and-error adjustment process, it was determined that the muscle activation patterns used to make take a step were found to be somewhat comparable with EMG records. Anderson and Pandy (2001a) created one of the most complex models of gait with a 3-D 23 degree-of-freedom mechanical linkage, actuated by 54 muscles. Using a dynamic optimization routine, normal walking on level ground was simulated. As computational models have grown in complexity, the number of assumptions that are made throughout the development process inevitably increases. To this end, increasing degrees of caution must be exercised when interpreting the results of a complex simulation.

Computer models of human motion can be separated into two basic categories: 1) inverse dynamics and 2) forward dynamics. Both of these methods are currently used by those who simulate human motion, and both have provided estimations of muscle forces that are otherwise impossible to measure. The critical differences between these two models outlined in the following sections.

2.4.1 Inverse Dynamics

Inverse dynamic modeling is a computationally inexpensive technique, and can provide researchers with accurate and meaningful data, such as muscle forces and joint moments. When using inverse dynamics, the position, velocity, and acceleration of a segment are used to determine the forces within the muscle (Figure 2.6).
These measures rely on measurements of segmental kinematics and ground reaction forces made within an instrumented gait lab. The inverse dynamics method requires calculation of moments about each joint from measured kinematics, and muscle forces are determined by solving a static optimization problem at each instant during the movement.

During static optimization, an objective function must be used to distribute the muscle forces, because most joints in the body have many muscles crossing them, which makes them statically indeterminant (Cheng et al., 1990). Formulation of a well-defined objective function is not easily achieved, as physiological and environmental constraints are not always easily quantified. Crowninshield and Brand (1981) suggested that minimizing the sum of muscle stresses to the third power will result in muscle activity that maximizes endurance. Using a different approach, Anderson and Pandy (2001a) successfully simulated walking on level ground by employing the minimization of metabolic energy expenditure for each muscle per unit distance traveled. The task being performed can directly influence the choice of objective function. For example, it has been suggested that the performance criterion for a ballistic motion, such as rising from a chair, be a minimization of the time derivative of muscle force (Pandy, 1995).

**Figure 2.6.** A block diagram of the inverse dynamics technique. Measured motions and external forces are used to calculate moments about the joints of the body. Static optimization is used to determine the distribution of the muscle forces. This process is repeated at every instant of time during the activity. *Figure based on review by Pandy (2001).*
2.4.2 Forward Dynamics

Forward dynamic modeling is more computationally expensive than inverse dynamic techniques, as simulations are created by the numerical integration of differential equations of motion. (Figure 2.7). Unlike inverse dynamics, which determines muscle forces from measured motions and external forces, forward simulations rely upon muscle forces as input to produce motions. Forward dynamic routines are often used within an optimization scheme which calculates an objective function for the entire cycle of movement, and repeats the simulations thousands of times over until the objective function has been minimized. Alternatively, muscle excitations can be optimized at each time step in an effort to replicate the experimental kinematics as closely as possible (Neptune et al., 2000). These methods offer flexibility in regard to changes in physiology and goals of tasks. For example, knee simulations created using forward dynamic modeling can isolate and vary inputs, such as muscular and ligamentous forces, and initial joint angles and velocities (Piazza and Delp, 2001). The freedom to change inputs individually grants the researcher the ability to fine tune their research question. Yamaguchi et al. (1989) isolated the effects of thickening the patella relatively quickly and accurately by manipulating a single input parameter in a forward dynamic computer model.

![Figure 2.7](image_url)

**Figure 2.7.** A block diagram of the forward dynamics technique. Muscle activations or forces can be used as input, and the differential equations of motion of the musculoskeletal model are integrated forward in time to produce motions. *Figure based on review by Pandy (2001).*

As mentioned earlier, most joints in the body, including the knee, have muscle redundancy and are therefore statically indeterminate (Cheng et al., 1990). Regardless of the computational method used, the determination of individual muscle forces is a challenging and somewhat subjective task. Despite the inherent
differences between inverse and forward dynamics, Anderson and Pandy (2001b) suggest that both methodologies provide estimations of muscle forces during gait that are practically equivalent. For a more thorough summary of these techniques, the reader is referred to reviews performed by Pandy (Pandy, 2001) and Zajac (Zajac et al., 2003).

2.4.3 The Computed Muscle Control Algorithm

As outlined above, determination of muscle excitations that produce desirable kinematics in muscle-actuated dynamic models is a challenge. Thelen et al. (2003; 2006) developed the computerized muscle control (CMC) algorithm to address this task. CMC implements a two-step approach to faithfully recreate a human motion. First, a residual elimination algorithm is used to ensure consistency between ground reaction forces and whole body dynamics. This is accomplished by making small changes to the masses of the segments and the position of the center of mass of the torso. With the residual elimination completed, CMC then uses forward dynamics, paired with feedback and feedforward controls and static optimization, to drive the kinematics of a musculoskeletal model (Figure 2.8). CMC has been used to simulate cycling motions in a 3-degree-of-freedom 30-muscle model, and gait in a 21-degree-of-freedom 92-muscle model. The kinematic output of the simulations within 1° degree of experimental quantities, and muscle activations were similar to recorded EMG patterns. The use of CMC is growing in popularity, and the algorithm has been included as a module in the OpenSim project (Delp et al., 2007), an open source musculoskeletal modeling software package. The algorithm has recently been used to examine the mechanics and muscle contributions of crouch gait (Steele et al., 2010), running (Hamner et al., 2010), and sprinting (Chumanov et al., 2007).

While the use of the CMC algorithm continues to grow, it is not the only paradigm that has been used to effectively simulation human motions. CMC is unique because it employs an approach that is similar to a proportional-derivative controller, where optimization occurs at every time step. Other methods rely upon schemes that simulate a task to completion, and then base the optimization on the resulting kinematics and kinetics of the entire motion. This approach has been
Figure 2.8. A schematic of the CMC algorithm, after the residual algorithm has been completed. The desired accelerations are computed with a feedback controller, and static optimization computes a set of muscle excitations ($\vec{u}$) that achieve the desired accelerations. The algorithm is applied at every $\Delta t$ during a forward dynamic simulation. Source: Thelen and Anderson (2006)

used to accurately estimate individual muscle contributions of activities such as rising from a chair, pedaling, and running (Pandy, 1995; Zajac et al., 2002; Neptune et al., 2000). Despite the advances in computational technology, researchers must judiciously select and eliminate certain aspects of the human body that will be included in their model. In many cases the integrity of the knee kinematics is sacrificed. For example, the CMC algorithm makes the assumption that the knee can be represented with a one degree-of-freedom joint. While this assumption may not influence the results of, say, the investigation of plantarflexor muscle activations during walking, it is obviously not suitable for an investigation of total knee replacement kinematics.

2.5 Simulations of the Knee Joint

Computational modeling of the knee joint has grown in complexity in a manner similar to that of human motion modeling. Rigid body contact models were at the forefront of knee modeling techniques, as they were computationally inexpensive and provided valuable insight into the motions of the knee joint by assuming point contacts at the tibiofemoral and patellofemoral joints. Initially, these simulations were restricted to the sagittal plane, and the models eventually progressed to 3-D
representations of the knee. Because point contact is an unrealistic representation of tibiofemoral and patellofemoral loading, finite element analyses have been used to effectively examine the complex load distributions within the knee joint. Unfortunately, finite element analysis is computationally expensive, and investigations of joint loads are typically limited to static or quasi-static poses. In order to create models that can fully characterize knee loads, dynamic elastic foundation models were developed to estimate contact pressures quickly and cheaply. The most modern dynamic elastic foundation models include muscle force estimations in an effort to predict realistic knee kinematics and contact pressures.

2.5.1 Rigid Body Contact Models

The first rigid body contact models of the knee were restricted to sagittal plane tibiofemoral representations of the joint. Despite the limitations of 2-D modeling, these models were effectively used to gain knowledge of the knee joint that was otherwise difficult or impossible to attain with physical testing methods. Moeinzadeh et al. (1983) created one of the first 2-D tibiofemoral knee models to investigate how dynamic loading of the tibia affected the kinematics of the knee. The model predicted kinematics, contact points, and loads borne by the ligaments that were representative of a natural knee. Abdel-Rahman and Hefzy (1993) refined this 2-D tibiofemoral model to examine the results of a sudden impact on the knee joint in the sagittal plane. They found that the anterior fibers of the PCL and MCL are the primary constraints for a posterior forcing pulse when the knee is flexed between 20°–90° of flexion. This result coincided with clinical evaluations of PCL and MCL ruptures.

Other groups worked to incorporate the patellofemoral joint into their 2-D knee model. Yamaguchi and Zajac (1989) used such a model to examine the effects of changing the thickness and length of the patella. The group found that thickening the patella only increased the effective moment arm at flexions below 35°. However, it was found that lengthening of the patella significantly increased the effective moment arm at flexions greater than 25°. Kim and Pandy (1993) expanded their investigation of the knee joint by including a lower limb model that consisted of 4 segments, 8 muscles and 4 knee ligaments. This model was used to
investigate the muscle, ligament, and articular contact forces during a squatting activity. It was found that peak tibiofemoral loads were up to 8 times body weight, which were primarily caused by muscle exertion during the movement.

A three-dimensional model is required to completely characterize the complicated movements of the knee joint. Garg and Walker (1990) created one of the first 3-D rigid body contact models, and verified it by comparing predicted ligament lengths, tibiofemoral contact points, and patellofemoral contact points to previously published cadaveric data. The validated model was subsequently used to investigate how changes in CR tibial component conformity and tilt affected the range of motion of the knee. It was found that posterior tilting of the tibial tray increased the range of motion by $10^\circ$, and that less conformity between the femoral and tibial components led to an increased range of motion. Sathasivam and Walker (1997) developed a separate 3-D model of a prosthetic knee joint to examine the effects of friction within computational models. The simulations successfully replicated the kinematics of the Stanmore knee simulator, and it was found that major differences in knee kinematics could be observed when the coefficient of friction was manipulated. The overall results of this study showed that small changes to the geometry of highly conforming implants resulted in large changes to the contact path geometry, suggesting that such changes may result in unfavorable wear characteristics.

Piazza and Delp (2001) developed the first dynamic three-dimensional lower extremity computer model that included tibiofemoral and patellofemoral articulating surfaces. This model simulated a step-up task, which was chosen specifically because stair climbing is a challenging task for TKR patients (Andriacchi and Hurwitz, 1997). In this forward dynamic model, experimentally measured muscle activity, initial joint angles and velocities, and kinematics of the hip and ankle were used as inputs. The number of contact points at each articulation was allowed to vary freely, giving the model “on the fly” adjustability of contact calculations. This simulation tracked flexion/extension angles well, but over-estimated the amount of tibiofemoral translation that was measured clinically. These errors could be attributed to the fact that friction was not included in this model, nor was the posterior capsule modeled. A unique power of this model lies in the fact
that it is flexible enough to represent a variety of joints in the body.

2.5.2 Detailed Characterization of Contact Forces

Detailed analysis of contact mechanics in TKR is important because abnormal load distributions lead to deterioration of the polyethylene in a prosthetic joint. To elucidate the causes of implant failures, researchers need to identify the locations and magnitudes of contact forces within the joint. Although rigid body models have been proven to be useful in TKR research, they lack the ability to fully characterize implant loading. In order to gain a more clear sense of the forces within the joint, researchers have employed finite element analysis and elastic foundation methods.

2.5.2.1 Finite Element Analysis

Initial FE analyses of TKR implants were conducted in static poses. Bartel et al. (1985; 1986) performed static finite element analysis to predict surface and subsurface stresses in TKR during displacement-controlled conditions. In these two studies, the group investigated how the thickness and conformity of the implant changed the amounts of stress in the components. The results of the study suggested that the minimum thickness of a tibial bearing should be in the range of 8–10 mm, a standard that is still held today. Furthermore, it was suggested that the combination high stress and a transient contact area is likely the cause of surface damage that is observed clinically in tibial components.

Advances in computational power have allowed for the development of hybrid models that pair finite element analyses with quasi-static representations of kinematic tasks. Several groups have used a quasi-static, implicit, force-controlled, finite element analysis to investigate tibiofemoral loads during gait. For example, Godest et al. (2002) investigated how changes to the coefficient of friction affected the kinematics of implants during stance, and determined that the coefficient must be within the range of 0.01–0.07. In a separate study, Otto et al. (2001) analyzed the amount of axial rotation of a mobile bearing throughout gait. The simulations estimated high frictional torques produced at the mobile bearing, which may
explain why some mobile bearings fail to rotate during gait.

Several groups have created dynamic simulations that incorporate explicit FE models of the knee. Guess and Maletsky (2005) developed such a model utilizing a virtual prototyping software created by ADAMS (MSC Software Corporation; Santa Ana, CA). In this study, joint contact is modeled as a non-linear spring damper system. The goal was to develop and verify a three-dimensional model of a knee simulator so that knee loads and kinematics could be used as input to generate muscular force profiles for an identical experimental knee simulator. The group was able to predict muscle forces profiles for sagittal plane squatting simulations that were very accurate, however, this accuracy was substantially less for simulations of walking that were in 3-D. Halloran et al. (2005b) and Giddings et al. (2001) both developed dynamic FE models to mimic the mechanics of prosthetic knees in experimental simulators. Both simulations were able to simultaneously predict forces and kinematics that compared favorably to those found in the physical rigs. Although these models succeeded in their goals, they required many hours of computational time to produce results, and exemplified the potential utility for elastic foundation methods in TKR research.

2.5.2.2 Dynamic Elastic Foundation Modeling

Elastic foundation modeling can be readily paired with rigid body dynamics modeling (Blankevoort et al., 1991) and can also be used as a “cheap” alternative to finite element analysis. With a simple model of a cylinder in a half-pipe, Li et al. (1997) compared the contact pressures and patches that were created by an elastic foundation model and by a finite element model. The results of this experiment showed few differences, allowing Li to conclude that elastic foundation modeling has the potential to be an acceptable substitute for finite element analysis. In a more complex side-by-side analysis of TKR components, Halloran et al. (2005a) found that the tibiofemoral contact areas were the same between the two models, with the elastic foundation model overestimating peak contact pressures by 15%. While the elastic foundation estimations of pressure were high (and therefore conservative), it should be noted that the use of a elastic foundation model reduced computational time by 98%.
Accurate estimations of in vivo TKR joint loads relies upon simultaneous predictions of muscle forces (Herzog et al., 2003) and determinations of constraint present within the joint (Glitsch and Baumann, 1997). This is a technically challenging task, and has recently led to the development of dynamic elastic foundation models that include muscle forces and deformable articulating contact within the knee joint.

Caruntu and Hefzy (2004) employed Coons’ bicubic surface patches to model articular contact in a natural knee. With this approach, the normal stress was calculated between the patches that intersect using the following basic equation, which is based on Hooke’s Law:

\[ \sigma = K u \] (2.1)

where \( u \) is the penetration of the patch perpendicular to the surface, and \( K \) is the contact stiffness. The contact stiffness was calculated as follows:

\[ K = \frac{(1 - \nu)E}{(1 + \nu)(1 - 2\nu)t} \] (2.2)

where \( \nu \) is Poisson’s ratio of the UHMWPE, \( t \) is the thickness of the layer, and \( E \) is the modulus of elasticity. Caruntu and Hefzy applied non-linear force functions to the quadriceps tendon to simulate a knee extension. The resulting motion successfully recreated the ”screw home mechanism” of the knee, and the tibiofemoral contact forces in the medial compartment were found to be higher than those in the lateral compartment.

Moran et al. (2008) and Landon et al. (2009) modeled contact between implants with a rigid body spring model (Li et al., 1997), in which penetrations of spring nodes on the tibial component into the solid femoral component surface are computed. Computation of spring forces with this method is based on the following equation:

\[ F = kx + b\dot{x} \] (2.3)

where \( F \) is the contact force, \( k \) is the spring constant, \( x \) is the spring node penetration depth, \( b \) is the damping value, and \( \dot{x} \) is the penetration velocity along the surface normal. Distributions of the springs and values of \( k \) and \( b \) must be carefully determined, based on the properties of UHMWPE (Bartel et al., 1995;
Edidin et al., 2000; Piazza and Delp, 2001).

Moran et al. (2008) used this method to predict PS implant kinematics in a simulated tibiofemoral range of constraint test. The kinematics of the computational models compared favorably with the physical experiments, and it was suggested that compliance of the structure of the physical simulator led to small displacements that were not seen in the computational models. Landon et al. (2009) used the same method to predict tibiofemoral and patellofemoral kinematics and kinetics of a PS knee in an Oxford Rig that was controlled by a simple feedback controller. The locations of the tibiofemoral axes of rotation were used to compare kinematics to an experimental rig, and the results of the computational simulations compared well with the experimental tests. When the experimental rig performed a stand-squat-stand motion, the effects of frictional hysteresis were evident. This was not identifiable in the results of the simulation due to the fact that friction was not included in the model.

Pandy et al. (1998; 1998) created a dynamic model of the natural knee that included articular contact as well as many of the muscles and ligaments that cross the joint. In this knee model, the articulating surface of the femur is represented by a patchwork of polynomial surfaces that are representative of the bones digitized by Garg and Walker (1990). The surfaces of the tibia and patella are each represented by two planar surfaces. Specifically, the lateral tibial plateau slopes 7° posteriorly and 2° laterally; the medial tibial plateau slopes 2° posteriorly and 2° medially; and the medial and lateral facets of the patella are inclined at an angle of 130° to each other. Contact forces are calculated by geometrically determining the radii of the elliptical contact patch and the depth of penetration. A relationship between pressure and normal displacement can be established, and with Hooke’s law stiffness can be tuned to a value that is representative of cartilage. Finally, the magnitude of the resultant force can be determined by integrating the change in pressure over the contact area. With this model, the group was able to realistically simulate passive knee motions, such as extension and AP drawer tests, as well as perform tasks that consisted of maximum isometric flexions of the extensors and flexors.

Shelburne et al. (2004; 2005; 2006) conducted a series of studies using the knee
model created by Pandy et al. (1998) in conjunction with the optimization scheme presented by Anderson and Pandy (2001a). All of these studies consisted of a similar two-step procedure. First, a 3-D whole body musculoskeletal model was used with the optimization theory of Anderson and Pandy (2001a). The knee is represented as a 1 degree-of-freedom joint in this model. Optimization of the whole body model predicted body-segmental motions and corresponding leg muscle forces for one cycle of gait. Joint angles, ground reaction force data, and simulated muscle forces were then used as input for a lower limb model that incorporated the 3-D knee created by Pandy et al. (1998). In the lower limb model the patellofemoral and tibiofemoral joints each have 6 degrees-of-freedom. At each instant of the gait cycle, a static equilibrium problem is solved to determine the loads experienced within the knee joint. With this modeling paradigm, the group was first able to predict realistic loading patterns of the ACL during normal walking (Shelburne et al., 2004). In subsequent studies, the contributions of muscles forces, ligament forces, and ground reaction forces to tibiofemoral joint loads during normal gait were predicted (Shelburne et al., 2005, 2006). The timing and magnitudes of the peak tibiofemoral loads during stance were between 2.2–2.4 body weights, which was similar to those reported previously by

To predict tibiofemoral loads during gait, Kim et al. (2009) and Taylor et al. (2004) have relied upon a slightly different paradigm that implements inverse dynamics and Hertzian contact theory. The protocol for these experiments was very similar to the one used by Shelburne et al. (2004; 2005; 2006), with a few exceptions. Kim et al. used same whole body musculoskeletal model as described above, but Taylor et al. used the musculoskeletal model created by Heller et al. (2001). The Heller knee joint is represented by a ball joint, not a one degree-of-freedom joint. Muscle forces were predicted ahead of time with a whole-body model, but neither group used the optimization theory of Anderson and Pandy (2001a) to predict muscle forces. Instead the groups made calculations of joint moments from inverse dynamics and decomposed the moments into muscle forces using an optimization scheme. Kim et al. minimized the sum of the squares of the muscle activations, while Taylor et al. minimized the sum of the square of the muscle stresses. Both groups then used the resulting muscle forces as input into a
3-D knee model that represented the femoral condyles with spheres and the tibial plateau with a flat surface. Hertzian contact theory was then used to calculate tibiofemoral loads. Kim et al. compared predicted tibiofemoral loads to results from a 68 kg subject who was wearing an instrumented implant. Average RMS errors for medial, lateral, and total tibiofemoral loads were $140 \pm 40$ N, $115 \pm 32$ N, and $183 \pm 45$ N respectively. Taylor et al. found peak resultant tibiofemoral loads between the range of 2.97–3.33 body weights, which was slightly higher than the range of 2.7–2.8 bodyweights reported by Hardt et al. (1978), and substantially higher than the values reported by Hurwitz et al. (1998) and Schipplein et al. (1991).

The research team at the University of Florida, headed by B.J. Fregly, has focused their research on the development of dynamic elastic foundation models for TKR. The group first developed an algorithm for the determination of contact pressures for both natural and prosthetic knees (Fregly et al., 2003). The contact model placed a bed of springs on the tibial surface and assumed the femur to be rigid. Estimations of contact pressures ($p$) for natural knees were derived from plane strain elasticity theory, and were calculated with the following formula (Blankevoort et al., 1991):

$$p = -\frac{(1 - \nu)E}{(1 + \nu)(1 - 2\nu)} \cdot \ln \left[1 - \frac{d}{h}\right]$$

where $\nu$ is Poisson’s ratio of the UHMWPE, $E$ is the modulus of elasticity, $d$ is the deflection of the spring and $h$ is the layer thickness. The group employed a separate equation for prosthetic knees because UHMWPE provides less deformation than natural cartilage. Therefore, the following formula was derived from plane strain elasticity theory for TKR implants (Blankevoort et al., 1991):

$$p = \frac{(1 - \nu)E}{(1 + \nu)(1 - 2\nu)} \frac{d}{h}$$

The material properties of the UHMWPE were modeled in linear and non-linear fashions. The validity of this code was tested by simulating static tibiofemoral pressure experiments with cadaver knees and CR implants. Results of all simula-
tions predicted average contact pressure well, but only the linear material model of UHMWPE could accurately predict peak contact pressures.

With a validated elastic foundation model in place, the group has worked to create elastic foundation models that are paired with dynamic models of the following: wear simulators (Zhao et al., 2008), normal gait (Bei and Fregly, 2004; Fregly et al., 2005; Hamilton et al., 2005), fast, slow, wide and toe-out gait (Zhao et al., 2007), and stair ascent (Fregly et al., 2005; Hamilton et al., 2005). To simulate all of these activities a similar modeling approach was used. The tibia was held in a fixed position while the flexion-extension, internal-external rotation and anterior-posterior translation of the femoral component were prescribed, based on data that was collected through fluoroscopy. This allowed the tibiofemoral joint to have 3 degrees-of-freedom (superior-inferior translation, medial-lateral translation, and varus-valgus rotation), which were predicted via dynamic simulation. The patellofemoral joint was not considered in these models. The vertical ground reaction force was scaled to be between 0.25-3.0 body weights, and the loading of the femoral component differed between studies. In some cases, the femur was loaded axially off-center to produce a 70% medial – 30% lateral load split (Bei and Fregly, 2004; Fregly et al., 2005) or on-center to produce a 50% medial – 50% lateral load split (Fregly et al., 2005). Instead of relying upon approximations of femoral load distributions, the most one study used outputs from an instrumented implant (Zhao et al., 2007).

The results of these studies were an encouraging sign that dynamic elastic foundation models are well-suited for TKR research. Depending on the activity, simulations took between 10-30 minutes to run. In the normal gait study, the resulting overall contact forces and torques were realistic, but the contact patches were too large, which resulted in small contact pressures (Bei and Fregly, 2004). The experiments that modeled a wear simulator (Zhao et al., 2008) and both normal gait and stair ascent (Fregly et al., 2005; Hamilton et al., 2005) were extended to tribological examinations of the polyethylene. It was found that predicted damage areas, volumes, and maximum wear depth was close to the patterns seen in retrieved implants (Fregly et al., 2005; Zhao et al., 2008), and that the coincidence of highest crossing intensity with greatest tribological intensity occurs in the lat-
eral compartment (Hamilton et al., 2005). Finally, the study that modeled various types of gait (Zhao et al., 2007) correlated knee adduction torques with medial compartment loads, which has the potential to be an important surrogate measure for medial compartment loading during gait.

In an effort to emphasize the need for further research and to underscore the dangers of using dynamic elastic foundation models without caution, Fregly et al. (2008) performed a sensitivity study with their model. Gait data was collected from a patient with an instrumented implant, and the methods described above were used to create a tibiofemoral model. The group locked all of the degrees-of-freedom of the tibiofemoral joint and systematically introduced errors to the kinematic inputs of the model. It was found that very small changes in prescribed kinematics created substantially large errors in knee load calculations. These errors were mostly due to errors introduced to superior/inferior translation and varus/valgus rotation. The result of this study suggest that accuracy on the order of milliradians and microns is needed to estimate contact mechanics directly from \textit{in vivo} measurements. Therefore estimated axial loads and load splits (not prescribed kinematics) should be used as for a dynamic model.

2.5.2.3 Surrogate Modeling

In an effort to make their simulations even faster, the University of Florida group has aimed to create a surrogate articular contact model of TKR. Surrogate modeling has been successfully used in other areas of engineering, and involves replacing a computationally expensive model with a cheap one, which is constructed from a cloud of data points from the original model. The end result is a fast and efficient model that can be used within optimization schemes which require thousands of iterations. In this case, the surrogate models are ultimately be paired with a muscular force optimization scheme.

Lin et al. (2009) first investigated the plausibility of TKR surrogate model by investigating a 2-D model. In this study, the planar motion of a Stanmore simulator was re-created with a dynamic elastic foundation model. The kinetics and kinematics from 9 unique simulations of were collected and used as input for the surrogate model. The surrogate model was able to reproduce contact forces and
motions, and reduced computational time from 13 minutes to 13 seconds for one cycle. Given the success of the 2-D surrogate model, the group continued to expand this paradigm by creating a surrogate model of a 3-D motion produced in a Stanmore simulator (Lin, 2010). Again, the results from the surrogate model compared well with the dynamic elastic foundation modeling results, and computational time was reduced from 17 minutes to 5 seconds per simulation.

To estimate *in vivo* TKR contact loads during gait, Lin et al. (2010) paired their surrogate model with an optimization scheme that predicted muscle forces. The model consisted of 3 segments and included 11 muscles and one patellar ligament, and the knee had 12 degrees-of-freedom. Activations of the muscles were grouped such that 8 static optimization problems needed to be solved at each instant of the gait cycle. These optimizations were posed such that penalty functions would be imposed if excessive muscle force or contact forces were implemented as a solution. Furthermore, a 4 part constraint function was used to ensure that resulting muscle forces did not create knee joint moments that varied greatly from the moment calculated from inverse dynamics. A subject with an instrumented implant was used to collect the *in vivo* data for this experiment, and the predicted kinematics of the knee joint closely matched those recorded within the gait lab. Tibiofemoral loads predicted by the model also closely matched those of the instrumented implant, with RMS errors of less than 15 N. The optimization scheme of this study was relatively burdensome, so simulations of a full gait cycle took 90 minutes to create. This was still a vast improvement over the 32 hours required to run the simulation without surrogate modeling.

### 2.6 Summary

The number of TKR patients in the United States continues to increase because of high satisfaction rates (90%–95%) (Colizza et al., 1995; Emmerson et al., 1996; Ranawat et al., 1997; Callaghan, 2001) and long implant lifespans (10–15 years) (Scott and Volatile, 1986; Stern and Insall, 1992; Font-Rodriguez et al., 1997; Weir et al., 1996). Despite this success, there is still a need for improvement in implant design. A variety of investigative methods have been developed for the
purposes of TKR research. Computational simulations are unique because they can provide estimates of measurements that are otherwise impossible to obtain, and they can be effectively incorporated into an efficient design cycle. Dynamic elastic foundation models (summarized in Table 2.1) are particularly attractive because they provide researchers with detailed characterizations of TKR kinematics and joint loads during everyday tasks. Through the use and development of these models, the function and wear of TKR implants can be further improved.

**Table 2.1.** A table summarizing the dynamic elastic foundation studies that were described above. Due to page size limitations, the following numbering scheme will be used to describe the methods used to control the simulations: 1) Prescribed muscle forces 2) Prescribed external forces 3) Prescribed motions 4) Simple feedback controller 5) Prescribed muscle forces determined from an optimization of a separate model 6) Muscle forces determined from inverse dynamics and optimization 7) Muscle forces determined from optimization of a surrogate model

<table>
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<tr>
<th>Study</th>
<th>Model Type</th>
<th>Joint(s) Modeled</th>
<th>Knee DOF</th>
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3.1 Introduction

Successful implantation of a total knee replacement (TKR) requires proper balancing of ligament tensions to reduce sensations of instability in the knee (Bottros et al., 2006; Figgie et al., 1986; Insall et al., 1983; Martin and Whiteside, 1990). Implants are usually designed with the expectation that the gaps between cut bone surfaces will be equal in flexion and extension in order to recreate the physiological axis of knee rotation (Bottros et al., 2006). It is often the case in patients with collateral ligament laxity that the flexion gap is larger than the extension gap. A common strategy for managing such a gap incongruity is to increase the tibial bearing thickness (thus reducing the flexion gap) and shift the femoral component proximally on the femur (to maintain the extension gap). This technique results in an elevated joint line, so called because the tibiofemoral articular interface is moved proximally relative to its original natural position.

Many in vivo investigations of joint line are based on populations of patients
who have undergone revision TKR surgery (Bellemans, 2004; Hofmann et al., 2006; Laskin, 2002; Partington et al., 1999). In cases of revision, a cruciate-substituting implant is used, as the posterior cruciate ligament (PCL) must be resected. During primary TKR procedures, however, either cruciate-retaining (CR) or posterior substituting (PS) designs may be selected, both of which are subject to joint line elevation. In a radiographic study of 200 primary TKR recipients (both PS and CR), it was shown that elevations in joint line were not significantly affected by the choice of implant design (Snider and MacDonald, 2009). The average joint line elevation found across all knees was more than 4 mm, with elevations as high as 18 mm.

The joint line is often elevated to compensate for excessive collateral ligament laxity. Although several authors have advocated maintaining the natural joint line with primary TKR components (Figgie et al., 1986; Singerman et al., 1995b; Cope et al., 2002; Sato et al., 2007; Wyss, 2006), the quantitative effects of elevation on joint mechanics have not been clearly characterized. Factors such as variability in implant designs, patient populations, and surgical procedures make it difficult to quantify the sensitivity of mechanical variables to changes in joint line elevation with in vivo methods.

A variety of in vitro knee motion simulators have been created to investigate TKR kinematics during flexion tasks (Martin and Whiteside, 1990; Singerman et al., 1995b; Maletsky and Hillberry, 2005; Miller et al., 1997, 1998; Most et al., 2005). The Oxford Rig (Figure 3.1) is a widely used means for cadaveric testing that can simulate a deep flexion task while permitting six degrees-of-freedom each at the tibiofemoral and patellofemoral articulations (Zavatsky, 1997). Kessler et al. (2009) used an Oxford Rig to investigate the influence of femoral component positioning upon joint kinematics in CR TKR, but did not consider systematic elevations of the joint line. In another study, Emodi et al. (1999) used an Oxford Rig to investigate the effects of shifts in joint line in CR TKR and found that elevation of the joint line resulted in increased strain in the anterolateral bundle of the PCL (the only bundle for which strain gauge attachment was feasible in vitro). The range of motion of the knee in this study was limited to only 105° of knee flexion and kinematics of the CR TKR were defined only in terms of the locations
of tibiofemoral contact. Contact points were measured using pressure-sensitive films that have been shown to have the potential to alter joint mechanics (Wu et al., 1998). Emodi et al. concluded that PS designs should be used if joint line elevation is unavoidable; however PS knees were not tested in their study.

Figure 3.1. The Oxford Rig, which consists of seven segments connected by five joints. The patellofemoral and tibiofemoral articulations have six degrees-of-freedom each.

The purpose of this study was to examine the effects of joint line elevation in PS and CR TKR using a computational simulation of an Oxford Rig test. Finite helical axes were calculated to characterize how the kinematics of the tibiofemoral joint were affected by changes in joint line elevation. The location of the axis of rotation is a clinically relevant measure that has been related to quadriceps extensor mechanism efficiency (Mahoney et al., 2002). Many daily activities require deep
knee flexion and the computational modeling approach employed in the present study permitted assessment of contact forces and joint motions at higher knee flexion angles than has been possible in previous in vitro tests of the effects of joint line elevation. The effects of joint line elevation were evaluated independently of other factors, as the CR and PS versions of the same implant design could be placed on the bones in a perfectly controlled manner and soft tissue properties could be maintained across simulations.

3.2 Materials and Methods

A forward-dynamic simulation of an Oxford Rig was used to simulate a loaded knee flexion performed under quadriceps control. The seven-segment model was created using the SIMM/Dynamics Pipeline (MusculoGraphics, Inc.; Santa Rosa, CA, USA) and SD/FAST (Parametric Technologies, Inc.; Needham, MA, USA) software packages. Segment lengths and inertial characteristics of the femur and tibia were computed for a male subject 180 cm tall with body mass of 75 kg using relationships for predicting anthropometric values published by Winter (Winter, 1990) (Table 3.1). The pelvis was modeled as a 30 kg mass to approximate half-body weight. Small values were assigned to the moments of inertia of the pelvis, but were inconsequential because the pelvis does not rotate in an Oxford Rig. Approximate masses and moments of inertia for the implants were estimated from component dimensions and material properties. As with previous instances of the Oxford Rig, a ball joint was used to represent the ankle and the hip was modeled as a universal joint. The pelvis was allowed to translate vertically such that the hip was located directly superior to the ankle (Figure 3.1). This topology has been shown to permit six degrees-of-freedom in the tibiofemoral articulation (Zavatsky, 1997). The patellofemoral articulation also had six degrees-of-freedom, creating a complete knee model with 12 degrees-of-freedom.

Musculoskeletal dimensions and force-generating properties were based on previously published values as follows. The model included the right half of a pelvis, a right femur, a right tibia, a patella, one simplified quadriceps muscle (vastus intermedius), and four ligament groups (posterior cruciate, patellar, lateral collateral,
Table 3.1. Masses and moments of inertia used within the simulated cadaver model and the mechanical leg model. Values for the simulated cadaver were computed for a 180cm tall male with a body mass of 75 kg, based on the anthropometric values of Winter (1990). Values of masses of the mechanical leg were measured with a balance, and moments of inertia were calculated after performing pendulum experiments with the segments.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Simulated Cadaver</th>
<th>Mechanical Leg</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mass (kg)</td>
<td>Mass (kg)</td>
</tr>
<tr>
<td>Pelvis</td>
<td>30</td>
<td>10.07</td>
</tr>
<tr>
<td>Femur</td>
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<td>1.475</td>
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<td>0.2</td>
</tr>
<tr>
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<td>0.025</td>
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<td>Tibia</td>
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<td>1.37</td>
</tr>
<tr>
<td>Tibial component</td>
<td>0.2</td>
<td>0.2</td>
</tr>
</tbody>
</table>

and medial collateral). Other muscles that pass posteriorly to the knee (semitendinosus, semimembranosus, short and long heads of biceps femoris, and medial and lateral gastrocnemius) were included, but generated only small passive force of less than 5 N. Ligaments were modeled as springs with quadratic force-deformation relationships. Force-generating properties for these muscles were based on the model presented by Delp et al. (1990). The PCL was modeled using ten fibers, characterized by unique attachment points and slack lengths chosen such that the PCL began to develop force at 80° flexion when natural tibiofemoral motions were applied (Makino et al., 2006; Kurosawa et al., 1985). Patellar ligament and collateral ligament properties were those specified for a model used to simulate a step-up task in TKR by Piazza and Delp (2001). Cylindrical wrapping surfaces were used to prevent the quadriceps tendon from passing through the anterior aspect of the femoral component and to prevent the medial collateral ligament from passing through the distal femur.

Contact forces between articulating surfaces were computed using a custom-written implementation of a rigid-body-spring-model (Landon et al., 2009; Li et al., 1997) in which forces depended on the interpenetration of contacting implant surfaces. Surface geometries were derived from CAD representations of size 3 Scorpio (Stryker Orthopaedics; Mahwah, NJ, USA) PS and CR TKR implants. Nodes rep-
resenting spring elements were placed approximately 2 mm apart on the bearing surfaces of the tibial and patellar components. The femoral component was represented as a collection of trimmed parametric surface patches. Contact forces were determined from the penetration of these nodes into femoral component surfaces using a Kelvin-Voigt model (Landon et al., 2009; Wineman and Rajagopal, 2000):

\[ F = kx + b\dot{x} \]  

(3.1)

where \( F \) is the contact force, \( k \) is the spring constant, \( x \) is the spring node penetration depth, \( b \) is the damping value, and \( \dot{x} \) is the penetration velocity along the surface normal. Values for \( k \) depended on the area represented by each spring node such that \( k \) was equal to the area multiplied by \( 2.568 \times 10^{11} \text{ N} \cdot \text{m}^{-3} \) (Moran et al., 2008). The value used for \( b \) was \( 200 \text{ N} \cdot \text{s} \cdot \text{m}^{-1} \) (Piazza and Delp, 2001; Moran et al., 2008). Frictional contact forces were not modeled.

A proportional-derivative feedback controller was used along with a simplified 2-D model of the Oxford Rig to determine the quadriceps force required to lower the pelvis of the 3-D model in a controlled manner. Use of a 2-D model (which represented the hip, knee and ankle with revolute joints) permitted estimation of the required knee extension moment without consideration of the accelerations produced by articular contact forces. The initial pelvis height of the 2-D model corresponded to \( 20^\circ \) of knee flexion. At each 10 ms timestep, errors in the vertical position and velocity of the pelvis taken between the 2-D and 3-D models were calculated, assuming that the 2-D pelvis was desired to move downward with a constant velocity of \( 3.9 \text{ cm} \cdot \text{s}^{-1} \). This rate of descent in the 2-D model resulted in position that corresponded to \( 120^\circ \) of knee flexion in 10 s. The slow velocity was chosen to reduce inertial effects and maintain stability during the motion. Position and velocity errors were multiplied by gains of \( 60 \text{ s}^{-2} \) and \( 2 \text{ s}^{-1} \) (arrived at by trial-and-error) and added to the expected acceleration of the 2-D pelvis (0 m-s\(^{-2}\)) to find the 3-D pelvis acceleration that would attempt to track the 2-D model. This acceleration was used with closed-form equations of motion to compute the desired knee extensor moment. By assuming a quadriceps moment arm that linearly decreased from 4 cm to 2 cm as knee flexion changed from \( 20^\circ \) to \( 120^\circ \) (Draganich et al., 2002), the required force to be applied along the line of
action of vastus intermedius was determined.

The geometry of the femur and tibia were derived from a previously published lower-extremity model (Delp et al., 1990), and the default positions of the femoral and tibial components relative to bony landmarks on the femur and tibia of were established according to the manufacturers' surgical guidelines. The same default positions were used for CR and PS designs. Instances of collateral ligament laxity were simulated by increasing the slack length of the medial and lateral collateral ligaments by 3 mm and 6 mm. Concomitant joint line elevation for both the CR and PS designs was modeled by effectively thickening the tibial component by 3 mm and 6 mm, while shifting the femoral component proximally by 3 mm and 6 mm relative to its default position. This range of joint line elevations brackets the average of PS and CR TKR observed by Snider and MacDonald (2009).

Simulation results were post-processed to compute axes of tibiofemoral rotation and contact characteristics. Using the method of Spoor and Veldpaus (1980), displacement matrices were decomposed to identify the finite helical axis between the femur and tibia for rotations of at least 10°. The location of this helical axis in the femoral frame and was characterized using the intersection of the three-dimensional axis with a sagittal plane passing through the transepicondylar midpoint (Figure 3.2). Effective tibiofemoral contact points were located throughout the simulation by separately calculating the center of pressure across all contacting nodes on the medial and lateral tibial bearing surfaces. Knee flexion angles were determined using Euler angle decomposition of the tibiofemoral rotation matrix according to the convention proposed by Grood and Suntay (1983).

Model predictions of TKR mechanics were tested by performing Oxford Rig evaluations of a mechanical leg model. The femur and tibia of the mechanical leg were aluminum rods upon which size 3 Scorpio PS implants were mounted. The patellar button was mounted on a 2.54 cm wide by 0.2 cm thick strip of nylon webbing (Jontay Distributor, Waycross, GA, USA). The medial and collateral ligaments were represented by rubber bands. Trials using the physical rig were begun with the knee flexed to approximately 30° and proceeded to extend until a flexion angle of 90° was reached; this occurred in about 35 s. A linear actuator under displacement control was connected to a steel cable that pulled on the patella along
Figure 3.2. Projections of the axis of tibiofemoral rotation onto the mid-knee plane for CR and PS knees with the default joint line elevation. Axis of rotation locations for angles between 20°–120° are shown with respect to a coordinate system attached to the femoral component.

A line of action directed along the mechanical axis of the femur. Computational simulations were performed after assigning inertial properties that corresponded to those of the mechanical leg (Table 3.1). The rubber bands were modeled as linear springs with experimentally determined slack lengths of 5.9 cm and a spring constants of 34.6 N/m. Passive muscles were not included in these computational simulations as they were not modeled in the mechanical leg.

The motions of the mechanical leg femur, tibia, and patella were tracked using clusters of four reflective markers affixed to each segment. Marker locations were tracked using three Eagle cameras (Motion Analysis Corp.; Santa Rosa, CA, USA) at 100 Hz, and marker location residuals determined for this volume during calibration were approximately 0.1 mm. Lengths of the medial and lateral collat-
eral ligaments, varus/valgus angles, and internal/external angles of the knee were calculated. Root mean squared errors for these measures were computed between the simulated and mechanical leg tests.

### 3.3 Results

In the case of default joint lines, the axis of tibiofemoral rotation for PS designs was found to translate more than it did for CR designs, for which the axis was relatively stationary (Figure 3.2). At flexion angles higher than 70° the axis of rotation for the PS design translated posteriorly to a substantial degree, while the CR axis of rotation exhibited much less movement (Figure 3.2). Movement of the axis of rotation with flexion was much more sensitive to elevation of the joint line for CR designs than for PS (Figure 3.3).

![Figure 3.3](image-url)

**Figure 3.3.** Path of the axis of tibiofemoral rotation during the knee flexion in the femoral frame for CR (left) and PS (right) knees. Results for joint line elevations of 0 mm, 3 mm, and 6 mm are shown for both CR and PS designs.

Locations of the tibiofemoral contact points for CR and PS implants exhibited responses to changes in joint line elevation that corresponded to the shifts in the axis of rotation (Figure 3.4). For the default joint line position of the CR knee, contact points moved little between 20° and 120° flexion. The CR knee exhibited increasing posterior movement of the contact points (femoral rollback) as the joint line was elevated further. In contrast, the PS knees exhibited similar degrees of
rollback for all instances of joint line elevation. None of the CR knees exceeded 120° of knee flexion during the 10 second simulation. In the cases of the neutral and 3 mm elevation, a terminal knee flexion of 119° was achieved. For the 6 mm elevation, the femur translated off of the posterior lip of the tibial implant at 109° knee flexion. The PS knees for all joint line elevations achieved a minimum of 120° knee flexion.

Joint kinetics were also affected by the extent of joint line elevation. The tibiofemoral contact forces in the CR knee with 6 mm of joint line elevation were 1000 N greater than the contact force for the PS design in deep flexion (Figure 3.5).

Conversely, tibiofemoral contact forces were relatively unaffected by joint line elevation for PS knees. The loads required of the vastus intermedius were similar for both CR and PS designs (Figure 3.6), but in the case of a 6 mm joint line elevation for the CR design, the required quadriceps load in deepest flexion (109°) was reduced by 500 N when compared the PS design for the same flexion angle.

Loads carried by the PCL increased with increasing joint line elevation (Figure 3.7). In the case of 6 mm of joint line elevation, individual fibers of the anterolateral bundle of the PCL became taut at between 55° and 60° of flexion and achieved maximal strains ranging between 7.6%–11.0% at 109° of flexion.

The PS implant motions for the mechanical leg as measured in the physical Oxford Rig were similar to those predicted using the computational simulation. The average RMS difference for the length of the medial and lateral collateral ligaments was 1.60 mm and 0.93 mm, respectively. The average RMS error was found to be 0.25° for the varus/valgus angle and 1.03° for the internal/external knee angle.

### 3.4 Discussion

The results of the present study show that joint line elevation that accompanies CR TKR may raise forces in the PCL (Figure 3.7), increase tibiofemoral contact forces (Figure 3.5), and induce excessive femoral rollback (Figures 3.3 and 3.4). In contrast, joint kinematics and kinetics were relatively unaffected by joint line elevation.
Figure 3.4. Locations of centers of pressure on the medial and lateral tibial condyles in 10 degree increments between 20° and 120° of knee flexion for CR (left) and PS (right) designs at 0 mm, 3 mm, and 6 mm of joint line elevation. A superior view of a right tibial component is shown.

in PS TKR. The simulated Oxford Rig tests gave accurate representations of knee replacement mechanics while permitting perfectly reproducible implant placements and soft tissue characteristics across different joint line elevation scenarios.
Figure 3.5. Tibiofemoral condylar contact loads and for CR and PS knees for 0 mm, 3 mm, and 6 mm of joint line elevation plotted versus knee flexion angle.

![Tibiofemoral Contact Loads](image)

Figure 3.6. Required quadriceps force for CR and PS knees for 0 mm, 3 mm, and 6 mm of joint line elevation plotted versus knee flexion angle.

![Quadriceps Force](image)

The kinematics simulated for the implant design considered in the present study compare favorably to those measured in previous in vivo and in vitro studies. In our simulations the axis of rotation remained stationary in the early stages of flexion for both the Scorpio CR and PS designs. Previous investigations of Scorpio CR kinematics in Oxford Rig cadaver simulations (Browne et al., 2005; D’Lima et al., 2001) and in patients performing weight-bearing knee bends (Kessler et al., 2007) also have shown minimal femoral rollback between 20° and 70° flexion for this design. These authors attributed this behavior to the implant design, which
Figure 3.7. PCL forces for CR knees plotted versus knee flexion angle.

features a constant distal femoral radius from 15° of hyperextension to 75° of flexion. At flexion angles higher than 70° the femoral cam engaged the tibial post in our simulations of PS TKR, and this cam-post interaction promoted substantial femoral rollback at higher flexion angles. These findings agree with a previous cinefluoroscopic study that showed that PS implants exhibit a larger amount of posterior femoral rollback than their CR counterparts during in vivo knee bend activities (Haas et al., 2002).

The results for our simulated Oxford Rig tests using CR implants were similar to the results of physical Oxford Rig tests performed by Emodi et al. (1999). Joint line elevation was found to increase femoral rollback by similar magnitudes in both studies. Like Emodi et al., we determined that joint line elevation reduced the demands on the quadriceps in deep flexion, most likely due to improved extensor mechanism efficiency caused by increased rollback. The authors also found that the anterolateral PCL began to develop force at 42° of flexion, which is a slightly lower value than the 55° predicted by our simulation. The loads experienced by the PCL exceeded the ultimate strength of the PCL structure found by Kennedy et al. (1976). However, Emodi found strain magnitudes of 15% in these fibers in deepest flexion, which compared very favorably to the 11% we predicted. Furthermore, this model lacks meniscofemoral components of the PCL as well as other soft tissues, which would bear some of the loads experienced solely by the PCL in our model. While Emodi et al. performed in vitro tests using different CR implants,
similarities between the results of the two studies are encouraging signs that computational modeling may be an effective means for obtaining valuable information that is available from cadaveric Oxford Rig tests in less time and reduced cost.

In addition to considering CR implants, we were able to evaluate the effects of joint line elevation in PS TKR. The results suggest that in cases in which joint line elevation is unavoidable, PS TKR affords more consistent joint mechanics that are likely to produce a more stable and durable outcome. When joint line is elevated, even by as much as 6 mm, PS TKR implant kinematics are relatively unaffected and tibiofemoral contact forces remain unchanged. These findings correspond to the clinical recommendations of Scuderi and Insall (1989), who advocated that PS designs can be implanted with joint line elevations of up to 10 mm with no adverse effects.

A number of limitations should be considered when interpreting the results of this computational model. Frictional forces were not included in the contact force algorithm, and this choice may have been responsible for minor differences between measured and simulated axes of rotation that became apparent at low flexion angles. Frictional forces have been shown to be determinative of TKR motions (Sathasivam and Walker, 1997), and should be taken into account in future modeling efforts. The Oxford Rig paradigm does not typically include hamstrings forces, which can also alter the tibiofemoral kinematics of the knee. Other muscles and ligaments were represented by muscle-tendon actuators with simplified paths, and many soft-tissue structures, such as the patellar retinaculum, were not included in the model. As one would expect, the simulation results were sensitive to the slack lengths and attachment locations of the models PCL fibers. The PCL properties in the model were assigned in accordance with the study done by Makino et al. (2006), which caused the PCL to engage at a specific angle when natural knee motions were applied. As such, this model represents only one knee, and the simulation results cannot be said to be representative of a broad range of anatomical variation, and may not be indicative of PCL properties of a knee with TKR implants. The simulations of the present study were carried out for a particular implant design, and may not generalize to other manufacturers implant designs.

Results of this study suggest that elevations of the joint line in CR TKR may
have adverse affects on joint kinematics and articular contact forces. Specifically, elevation of the joint line increase PCL loads, leading to potentially excessive femoral rollback and large loads on the tibial bearing. These changes in joint mechanics may lead to instability and increased implant wear. Our simulation results support the recommendations of previous authors that PS designs should be implanted when flexion-extension gap imbalances are evident (Emodi et al., 1999; Scuderi and Insall, 1989). The computer simulation of an Oxford Rig test described in this paper permits quantification of the changes in implant performance that result from joint line elevation. Similar simulations could be created in the future to examine the influence of other factors, such as implant design variations and component malalignment.
Chapter 4

Mechanical Consequences of Addressing Collateral Ligament Laxity Through Joint Line Elevation or Femoral Component Upsizing in Revision TKR

4.1 Introduction

Proper management of knee joint stability is critical to the success of a total knee replacement. Freeman and Insall (1977) initially popularized the method of balancing the flexion and extension gaps as an effective technique to provide stability within the knee joint throughout the range of motion. During revision knee surgery, knee laxity is frequently encountered. Collateral ligament laxity often contributes to inequality between the flexion and extension gaps, with the flexion gap typically being larger than the extension gap. Insertion of a thicker tibial polyethylene insert to fill the flexion space usually requires elevation of the joint line by a proximal shift of femoral component position to avoid extension tightness (Bottros et al., 2006).

When technically possible, restoration of the pre-operative joint line in revision
knee surgery is desirable to achieve optimal knee kinematics and function (Martin and Whiteside, 1990; Figgie et al., 1986; Bryan and Rand, 1982). Patients with aggregate mean scores of joint line elevation exceeding 8 mm have been found to also exhibit significant reductions in Knee Society Clinical Rating Scores (Partington et al., 1999). It is important to note, however, that this absolute value is not the only variable associated with satisfactory scores as many of the patients in this study that had elevation of the joint line beyond 8 mm had excellent knee scores due to the excellent knee stability obtained by elevating the joint line. The unfavorable clinical outcomes associated with an elevated joint line in some patients may be due to the relatively depressed position of the patella. The kinematics of a depressed patella may lead to impingement on the front of the tibial component, leading to anterior knee pain and limited range of motion (Yoshii et al., 1991).

An alternative, and often concomitant, strategy for limiting the magnitude of joint line elevation is to increase the antero-posterior sizing of the femoral component (Bottros et al., 2006). Currently, most surgeons employ a combination of maximal femoral component upsizing before consideration of joint line elevation to achieve joint stability during revision knee surgery. Unfortunately, current prosthetic design dimensions do not provide enough compensation with component upsizing. Femoral component upsizing helps reduce collateral ligament slack in flexion and would reasonably be expected to influence patellofemoral mechanics and articular loads, but the mechanical implications of upsizing have received little attention to date.

The purpose of the present study was to use a computer simulation of a loaded knee flexion in order to evaluate changes in knee mechanics that follow from increasing degrees of collateral ligament laxity that are addressed by (1) elevation of the joint line; or (2) femoral component upsizing with reduced degrees of joint line elevation. The use of a computer model permits specification of ligament laxity and eliminates variation due to surgical technique and anatomy that have the potential to confound benchtop tests or in vivo studies.
4.2 Materials and Methods

A musculoskeletal model of the lower extremity was created in order to simulate both intraoperative ligament balancing and a loaded knee flexion activity that is similar to that which occurs in cadaveric knee simulators. Segment lengths and bony landmarks on the femur and tibia were based on the lower extremity model of Delp et al. (1990). Ligaments were modeled as springs with quadratic force-deformation relationships, and slack lengths and attachment sites of collateral ligaments were based on the TKR model presented by Piazza and Delp (2001). Two ligament segments were used to represent each collateral ligament and four segments were used to represent the posterior capsule. Properties of the posterior capsule were approximated based on those described by Blankevoort and Huiskes (1996) and Sperber and Wredmark (1998). Slack lengths of the medial and lateral collateral ligaments were increased by 3 mm, 6 mm, and 9 mm to simulate collateral ligament laxity. A cylindrical wrapping surface fixed in the tibial coordinate system was used to prevent penetration of the medial collateral ligament into the bone surface of the medial tibial epicondyle. Similar cylinders fixed in the femoral coordinate system prevented penetration of the posterior capsule into medial and lateral condyles. The models used in these simulations were created using the SIMM/Dynamics Pipeline (Musculographics, Inc.; Santa Rosa, CA, USA) and SD/FAST (Parametric Technologies; Needham, MA, USA) software packages.

Forward-dynamic simulations of a flexion gap test were created by placing the knee in 90° of flexion while locking the tibia in place and applying a 20 Nm hip flexion moment to distract the knee joint (Figure 4.1). Flexion gaps were calculated by averaging the perpendicular distances between the edges of posterior femoral condyles and the medial and lateral depressions of tibial plateau. The extension gap test was simulated by positioning the knee in full extension and applying 50 N distracting forces in equal and opposite directions along the mechanical axes of the femur and tibia (Figure 1). Extension gap distances were calculated by averaging the perpendicular distances from the medial and lateral distal femoral condyles to the tibial plateau.

Implant placements were based the results of the simulations of intraoperative gap balancing. Gap distances were determined for both flexion and extension tests
Figure 4.1. Schematic representations of flexion and extension gap simulations. Simulated laxities in the collateral ligaments caused the flexion gaps to be larger than the extension gaps.

and cutting planes were established on the leg model according to the manufacturers surgical protocol for Size 3 Triathlon (Stryker Orthopaedics; Mahwah, NJ, USA) Total Stabilized (TS) total knee replacements. This procedure was first done with normal collateral ligament slack lengths to establish baseline implant positions. In cases of collateral ligament laxity, the femoral component was shifted in the proximal direction, and the tibial bearing thickness was increased to properly balance the flexion/extension gaps. This process was repeated with Size 4 femoral components paired with Size 3 tibial bearings. Implant placements were affected by the use of Size 4 femoral components because of the larger distal femoral radius of the implant (Table 4.1).

Table 4.1. A summary of the implant placements, based on flexion and extension gap simulations. Size 3 and 4 femoral components were implanted on knees with 3, 6, and 9 mm of collateral ligament slack length. The additional tibial bearing thicknesses and proximal femoral component shifts are shown.

<table>
<thead>
<tr>
<th>Femoral Component Size</th>
<th>Additional Ligament Slack Length (mm)</th>
<th>Additional Tibial Bearing Thickness (mm)</th>
<th>Proximal Femoral Component Shift (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>3</td>
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<td>4.69</td>
</tr>
</tbody>
</table>

To examine the mechanical effects of the femoral component upsizing strategy,
a forward-dynamic simulation was created to simulate a loaded knee flexion under quadriceps control. The rigid body model used in the knee flexion models included the right half of a pelvis, a right femur, a right tibia, a patella, and TS TKR components. The framework of the model was based on an Oxford Rig (Biden, 1981) which replaces the hip joint with a universal joint, substitutes the ankle joint with a ball joint, and restricts the pelvis to vertical translations with respect to the ankle. This topology provides six degrees-of-freedom for each of the tibiofemoral and patellofemoral joints (Zavatsky, 1997). Segment masses and inertial properties of the femur and tibia for a 75 kg, 180 cm tall male were calculated based on anthropometric values established by Winter (1990). Inertial characteristics of the TKR components were calculated from component dimensions and material properties. A single simplified quadriceps muscle controlled the lowering of the pelvis, and had attachment points and a line of action that mimicked the vastus intermedius. Other muscles that cross the knee joint (semitendinosus, semimembranosus, short and long heads of biceps femoris, and medial and lateral gastrocnemius) remained passive and generated only small forces during simulations. The same ligaments used in the intra-operative knee model were implemented in this model, and the patellar ligament developed by Piazza and Delp (2001) was also included. The properties of the wrapping surfaces from the previous model were kept constant, and a wrapping cylinder was added in the femoral frame to constitute the boundary between the vastus intermedius and the femoral component.

Contact between tibiofemoral and patellofemoral articulating surfaces was modeled with custom-written software that implemented a rigid-body-spring-model (Li et al., 1997). Instances of tibiofemoral contact and patellofemoral contact were modeled by placing deformable beds of springs on the tibial and patellar implants, while the femoral component was considered to be rigid. Penetration depths and velocities of individual spring nodes allowed for the calculation of localized contact forces using a Kelvin-Voigt model (Wineman and Rajagopal, 2000). Frictional forces were not considered. For a more thorough explanation of the methods used for computational modeling of contact, including experimental validation, the reader is referred to Moran et al. (2008) and Landon et al. (2009).

The quadriceps force required to lower the pelvis in a controlled manner was
determined with a two-dimensional (2-D) representation of the Oxford Rig. This simplified version of the three-dimensional (3-D) model implemented revolute joints at the hip, knee, and ankle, and consequently eliminated accelerations caused by contact forces. A proportional-derivative controller was used to control the vertical position of the pelvis in the 2-D model. The pelvis was initially positioned at 0.875 m above the ankle, which corresponded to 15° of knee flexion. In order to reach 125° of knee flexion in 10 seconds, the desired downward velocity of the pelvis was 3.9 cm/s. Based on the desired motion of the 2-D pelvis, the controller calculated errors in position and velocity for each 10 ms step in time. These errors were multiplied by position and velocity gains of 60 s\(^{-2}\) and 2 s\(^{-1}\), respectively, to determine the desired acceleration of the pelvis. The knee extension moment arm of the quadriceps actuator used to compute the desired actuator force was prescribed as a function of knee flexion angle (Draganich et al., 1987), and a closed-form set of equations of motion yielded the required quadriceps force in the 2-D model. These calculated forces were applied during the 3-D forward-dynamic simulation, and the pelvis descended in the desired controlled manner.

Kinematics of the tibiofemoral joint were defined in terms of finite helical axis locations. The helical axes were calculated for each frame of simulation output by using the method of Spoor and Veldpaus (1980). Displacement matrices were constructed from the 4-by-4 homogeneous transformation matrices between the femoral and tibial coordinate systems. Knee angles were determined using Euler angle decomposition of the tibiofemoral rotation matrix according to the convention proposed by Grood and Suntay (1983). Pairs of displacement frames, separated by 10° of knee flexion, were utilized to calculate helical axis locations. The 10° of knee flexion separation ensured that a sufficient amount of motion occurred for the displacement algorithm to reliably identify an axis of rotation. Displacement matrices were decomposed to determine a point and vector along the three-dimensional finite helical, both in the femoral coordinate system. For each time frame, two-dimensional representations of the axes of rotation were defined by locating the point where the three-dimensional axis intersected a sagittal plane positioned at the knee center (Figure 4.2).

Simulated knee joint motions were visualized using SIMM and OpenSim soft-
Figure 4.2. Projection of the axis of tibiofemoral rotation onto the mid-knee plane for a size 3 femoral component with 0 mm of joint line elevation. Axis of rotation locations for angles between 20°-120° are shown with respect to a coordinate system attached to the femoral component.

To assess the risk of interference between the patellar and tibial components, the minimum distance between the tibial post and the patellar dome was computed for each simulation (Figure 4.3). This distance varied between simulations and in some cases the tibial post and the patellar component surfaces intersected. No contact forces were calculated between the tibial and patellar implant surfaces, however, and these components thus were permitted to pass freely through one another during the simulation.

4.3 Results

The required quadriceps force during deep flexion (knee flexion angle between 90° and 125°) was reduced when excessive laxity was imposed and a larger femoral component was selected (Figure 4.4). When compared to a knee with default ligament slack lengths at 125° flexion, use of the Size 3 femoral component required additional quadriceps force that ranged from 99 N to 824 N (Table 4.2). This additional force depended on the degree of collateral ligament laxity that was simulated. The use of the Size 4 femoral component required less quadriceps
Figure 4.3. Sagittal plane view of the knee in deep flexion. The minimum distance between the tibial post and the patellar component were computed during simulated knee flexion.

force than the default case (unaltered ligament slack lengths and Size 3 femoral component) when 3 mm and 6 mm of additional collateral ligament slack length were imposed, but when slack lengths were increased by 9 mm, 150 N of additional quadriceps force was required.

At flexion angles between 20° and 90°, the required quadriceps forces for knees with Sizes 3 and 4 femoral components were similar and forces were generally higher than for the knee with normal ligament properties (Figure 4.4). Differences in required quadriceps force between the default case and the increased laxity cases were greatest at approximately 75° of knee flexion (Table 4.2). At this knee flexion angle, use of a Size 3 femoral component required quadriceps forces that were higher by magnitudes between 70 N and 427 N. Similarly, use of a Size 4 femoral component required additional quadriceps forces that ranged between 79 N and 348 N. Knees with Size 4 femoral components exhibited axes of rotation
that shifted less due to ligament laxity than knees with Size 3 implants (Figure 4.5).

Table 4.2. A table showing the changes in required quadriceps force at 75° and 120° of knee flexion.

<table>
<thead>
<tr>
<th></th>
<th>75° Required Quad Force (N)</th>
<th>% Extra Force</th>
<th>120° Required Quad Force (N)</th>
<th>% Extra Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Size 3-0 mm</td>
<td>1705</td>
<td>N/A</td>
<td>4953</td>
<td>N/A</td>
</tr>
<tr>
<td>Size 3-3 mm</td>
<td>1776</td>
<td>4.1</td>
<td>5052</td>
<td>2</td>
</tr>
<tr>
<td>Size 3-6 mm</td>
<td>2010</td>
<td>17.9</td>
<td>5324</td>
<td>7.5</td>
</tr>
<tr>
<td>Size 3-9 mm</td>
<td>2132</td>
<td>25</td>
<td>5776</td>
<td>16.6</td>
</tr>
<tr>
<td>Size 4-3 mm</td>
<td>1784</td>
<td>4.6</td>
<td>4699</td>
<td>-5.1</td>
</tr>
<tr>
<td>Size 4-6 mm</td>
<td>1942</td>
<td>13.9</td>
<td>4943</td>
<td>-0.2</td>
</tr>
<tr>
<td>Size 4-9 mm</td>
<td>2054</td>
<td>20.4</td>
<td>5102</td>
<td>3</td>
</tr>
</tbody>
</table>

For cases of 6 mm and 9 mm of additional collateral ligament laxity, knees with upsized femoral components deviated less from the default case axis of rotation path than did knees with Size 3 femoral components. In the case of 3 mm of collateral ligament laxity, Size 3 knees followed the default case axis path very closely, while the upsized knees exhibited a small amount of deviation.

Visualizations of segment orientations at 120° of knee flexion showed that the distance between the tibial post and the patellar dome decreased as collateral ligament slack length was increased (Figure 4.6).

The default-case knee with no additional collateral ligament slack length exhibited 3.6 mm of patellotibial clearance. As collateral slack lengths were increased by 3 mm and 6 mm, knees with Size 3 femoral components exhibited decreasing patellotibial clearances of 2.3 and 0.1 mm, respectively. It was found that 1.7 mm of patellotibial penetration occurred in the 9 mm case. Knees with upsized femoral components exhibited larger clearances between the tibial post and the patella. Distances of 4.1 mm, 2.1 mm, and 0.5 mm were measured for cases of 3 mm, 6 mm, and 9 mm of additional collateral ligament slack length.

4.4 Discussion

The results of this study suggest that the magnitude of collateral ligament laxity should inform the choice of the method used to balance the flexion and extension
gaps. In the cases of mild collateral ligament laxity (3 mm), the use of a same-sized femoral component, paired with a thicker tibial insert yielded knee kinematics and kinetics that were most consistent with a knee that had no collateral ligament laxity (Figures 4.4 and 4.5). An upsized femoral component yielded slight differences in the location of the axis of rotation, but this change in kinematics reduced quadriceps forces in deep flexion. In cases of more severe collateral ligament laxity (6 mm and 9 mm), femoral upsizing reduced the amount of joint line elevation in the knee (Table 4.1). It is likely that the diminished elevation of the joint line yielded quadriceps loads and axes of rotation that were more consistent those of a knee with no collateral laxity (Figures 4.4 and 4.5).

Elevation of the joint line systematically reduced the minimal clearance between the patellar button and the post of the tibial component (Figure 4.6). In the case of 9 mm of collateral ligament laxity, contact occurred between the patellar insert and the tibial post when a same-sized femoral component was employed. However, when the femoral upsizing technique was employed, the contact between the patella and the tibial post was avoided. This result supports the conclusion that femoral upsizing should be used in more severe cases of collateral ligament laxity.

Previous studies have suggested that the outcome of a revision total knee replacement is not significantly affected by a joint line elevation, as long as the elevation does not exceed 8 mm (Figgie et al., 1986; Partington et al., 1999). The results of the current study are in partial agreement with this finding. It was found that mild amounts of collateral ligament laxity (3 mm) did not have much influence on knee joint kinetics or kinematics, regardless of which surgical approach used. It should be noted, however, that the threshold of allowable joint line elevation may be less than 8 mm. Collateral ligament laxities of 6 mm and 9 mm led to substantial increases in quadriceps forces and alterations of the axis of rotation.

The patellotibial kinematics findings from our simulation study are similar to those of a previous study performed by Yoshii et al. (1991). In both cases it was demonstrated that elevations of joint line led to impingement of the patella on the tibial component. These kinematics are unfavorable because, as the patellotibial clearance diminishes, the lever arm of the extensor mechanism is compromised. This loss of leverage requires the quadriceps muscles to generate larger forces, but
in addition the muscle might also operate on sub-optimal portions of its force-length curve (Figgie et al., 1986).

Certain limitations are associated with the computer simulation approach used in the present study. The Oxford Rig modeled in the simulation is itself a cadaver simulation of a knee flexion task. In some respects, the Oxford Rig test does not faithfully represent an in vivo movement: the pelvis is situated directly over the ankle and there is no forward flexion of the trunk. This creates excessive knee extension moments which may not be experienced by a patient during a knee-bend activity. The movement is also performed under the control of a single quadriceps actuator, rather than under the influence of multiple muscles, including knee flexors. Finally, only one type of implant was considered in this study, and other implant geometries may yield different results. Although this model suffers from these limitations, we feel that it is an effective means to gauge overall trends that result from the different surgical protocols used to address collateral ligament laxity.

The current study only addresses one scenario of flexion/extension gap incongruity. There are other instances of gap imbalances which require different surgical approaches (Bottros et al., 2006). The computational simulation approach used in the present study could be implemented in future studies to inexpensively and rapidly gauge the effectiveness of other proposed surgical treatments, such as component downsizing, femoral augmentation, and soft tissue release.

The results of these simulations suggest that thickening of the tibial component may be an effective method to manage cases of mild collateral laxity (3 mm) provided the knee is able to achieve terminal extension. In the case of moderate to severe collateral ligament laxity (6 mm, 9 mm), it seems advisable to upsize the femoral component to as much as is possible to minimize the degree of joint line elevation. While the results of the present study suggest that upsizing may help to address collateral laxity, better options are needed for managing more severe cases such as severe compromise of the posterior capsule and collateral ligaments. In such cases, joint line elevation and upsizing alike may be required to achieve joint stability but may lead to excessive loading of the quadriceps in flexion. These increased forces have implications for clinical symptoms and may
potentially promote implant wear and loosening of the patellofemoral articulation. Similar concerns are raised by the reduced amounts of patellotibial clearance and even impingement that may occur for large degrees of collateral ligament laxity.
Figure 4.4. Required quadriceps force for Size 3 and Size 4 femoral components for 3 mm (top), 6 mm (middle), and 9 mm (bottom) of collateral ligament laxity. All figures include required quadriceps forces of a default case Size 3 knee with no simulated additional laxity. Forces are plotted versus knee flexion angle.
Figure 4.5. Axis of rotation locations for Size 3 and Size 4 femoral components for 3 mm (top), 6 mm (middle), and 9 mm (bottom) of collateral ligament laxity. All figures show axes of rotation for a default case Size 3 knee with no simulated additional laxity. Axis locations are plotted with reference to a coordinate system attached to and aligned with the femoral component.
Figure 4.6. Patellotibial clearances (in mm) for same-sized and upsized femoral components with 3 mm, 6 mm, and 9 mm of collateral ligament laxity at 120° knee flexion. A negative number implies contact, but no forces between implants were simulated or applied in the simulations.
Chapter 5

Position of the Quadriceps Muscle Actuator Influences Knee Loads During Simulated Squat Testing

5.1 Introduction

The Oxford Rig knee simulator provides an indication of total knee replacement function during everyday activities that occur under quadriceps load, such as rising from a chair or climbing stairs. The jig simulates a squatting task by employing a universal joint at the hip, a ball joint at the ankle, and pelvis stage that slides on vertical rails. This model topology provides six degrees-of-freedom at the tibiofemoral and patellofemoral joints (Zavatsky, 1997). Over the years, many investigators have published studies of knee mechanics using variants on the original Oxford Rig design (Quentelier et al., 2008; Singerman et al., 1994, 1995a; Bourne et al., 1978; Walker and Haider, 2003; D’Lima, 2000, 2001). All of these design iterations have maintained 12 degrees-of-freedom for the knee joint, but the fixation point of the actuator intended to stand in for quadriceps muscles has often been varied. While the actuator in the original Oxford Rig was attached to the femur (Bourne et al., 1978), quadriceps actuators have also been fixed to the moving pelvis stage (Quentelier et al., 2008) and to the ground (Singerman et al., 1995a, 1994).
The use of two “back of the envelope” static free-body diagrams effectively demonstrates that kinetic differences will occur based on the fixation point of the quadriceps actuator (Figure 5.1). In the diagram on the left, the quadriceps actuator is internal to the system (placed on the femur or pelvis). The weight of the pelvis \((W)\) acts downwards, and the femur and tibia are considered to be massless. A horizontal reaction force is imposed by the vertical rails of the Oxford Rig \((R_{hor})\), and a large reaction forces occurs at the ankle joint \((R_{ankle})\). This large reaction force at the ankle causes a substantial moment about the knee joint \((M_{knee})\). This system changes when the quadriceps actuator is attached to the ground and pulls upward \((F_{quad})\), as seen in the diagram on the right. The upward pulling force of the quadriceps actuator reduces the amount of downward force caused by pelvis weight, which must be borne by the reaction force at the ankle. Subsequently, the moment about the knee is reduced. While these “back of the envelope” free-body diagrams demonstrate clear differences of static poses of the Oxford Rig, it is unclear how changes in actuator position will affect the kinematics and kinetics of the TKR implants throughout the entire squatting motion.

The purpose of the present study was to use a forward-dynamic computer simulation to vary the fixation point of the quadriceps actuator and examine how TKR kinematics and kinetics are affected. Simulations of loaded knee flexions were carried out with the actuator mounted on the femur segment, on the pelvis stage, and fixed to the ground while pulling in two different directions- vertical and horizontal (Figure 5.2).

5.2 Methods

We based our computational model on the Oxford Rig within our laboratory (Figure 5.3). This rig employs a mechanical leg, consisting of steel rods for the femur and tibia. Custom-made aluminum mounting blocks attach to the steel rods and are fitted with Scorpio PS (Stryker Orthopaedics) Size 3 TKR components. The patella mounts on a 2.54 cm wide by 0.2 cm thick strip of nylon webbing. A custom-made aluminum swivel connects the patellar tendon to a nylon coated steel cable, which attaches to a linear actuator (Northern Tool, Burnsville, MN). Medial and
Figure 5.1. Two free-body diagrams that are representative of an Oxford Rig. When the quadriceps actuator is mounted on the femur or pelvis (left), the reaction force at the ankle is relatively large, which causes a substantial moment about the knee joint. When the quadriceps actuator is mounted to the ground reference frame and pulled vertically (right), the reaction force at the ankle is decreased and the resulting moment about the knee becomes smaller.

lateral collateral ligaments are represented by rubber bands.

A CAD model of the rig was created in SolidWorks (Concord, MA), which was used to determine the inertial properties of the segments (Table 5.1). The mass of the pelvis was varied in the simulations between 10 and 20 kg. Because the femur and tibia consisted of several parts that were made of a variety of materials, the moments of inertia about the x and z axes were double-checked with a simple pendulum experiment. The mass of each segment was determined and the distance from the center of mass to the axis of the pendulum was calculated. The segment was allowed to swing freely for 20 cycles, and the average period of a swing was calculated. This process was repeated 3 times for each segment and an average period was determined. The moment of inertia \( I \) was then calculated with the following formula:

\[
T = 2\pi \cdot \left[ \frac{I}{(mgL)} \right]^{1/2}
\]
Figure 5.2. Models of Oxford Rigs with the quadriceps actuator fixed to (A) the femur and (B) the pelvis stage. Actuators were also attached to the ground with actuator displacement in the (C) vertical and (D) horizontal directions. Knee flexion occurred as the pelvis stage was lowered under quadriceps control.

where $T$ is experimentally determined period of pendulum sway, and $L$ is the distance from the center of mass to the axis of the pendulum.

The ten-segment forward dynamic model was created using the SIMM / Dy-
Figure 5.3. The Oxford Rig in the Penn State Biomechanics Lab. The femur and tibia are represented by steel rods, and the collateral ligaments are represented with rubber bands. A Size 3 Scorpio PS TKR implant is used within the rig.

...namics Pipeline (MusculoGraphics, Inc., Santa Rosa, CA) and SD/FAST (Parametric Technologies, Inc., Needham, MA) software packages. As with previous instances of the Oxford Rig, a ball joint was used to represent the ankle and the hip was modeled as a universal joint. The pelvis was allowed to translate vertically such that the hip was located directly superior to the ankle (Figure 5.3). This topology has been shown to permit six degrees-of-freedom in the tibiofemoral articulation (Zavatsky, 1997). The patellofemoral articulation also had six degrees-
Table 5.1. Masses and moments of inertia used within the model. Values of masses were measured with a balance, and moments of inertia for the tibia and femur were calculated after performing pendulum experiments with the segments.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mass (kg)</th>
<th>Moments of Inertia (kg·m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>10 or 20</td>
<td>I_{xx,yy,zz} = 0.001</td>
</tr>
<tr>
<td>Femur</td>
<td>1.475</td>
<td>I_{xx,zz} = 0.121711</td>
</tr>
<tr>
<td></td>
<td></td>
<td>I_{yy} = 0.001844</td>
</tr>
<tr>
<td>Femoral component</td>
<td>0.2</td>
<td>I_{xx,yy,zz} = 0.001</td>
</tr>
<tr>
<td>Patella</td>
<td>0.025</td>
<td>I_{xx,yy,zz} = 0.001</td>
</tr>
<tr>
<td>Patellar component</td>
<td>0.025</td>
<td>I_{xx,yy,zz} = 0.001</td>
</tr>
<tr>
<td>Tibia</td>
<td>1.37</td>
<td>I_{xx,zz} = 0.108282</td>
</tr>
<tr>
<td></td>
<td></td>
<td>I_{yy} = 0.001713</td>
</tr>
<tr>
<td>Tibial component</td>
<td>0.2</td>
<td>I_{xx,yy,zz} = 0.001</td>
</tr>
</tbody>
</table>

of-freedom, creating a complete knee model with 12 degrees-of-freedom. For details regarding the validation of this computational model, the reader is referred to Appendix B.

The model included one quadriceps cable, and three ligament groups (patellar, lateral collateral, and medial collateral). The “muscle properties” that were used to represent the quadriceps cable (steel) and patellar tendon (nylon webbing) were modeled with quadratic force-deformation curves that were based on their material properties. The collateral ligaments were modeled as springs with linear force-deformation relationships, which were determined experimentally. Cylindrical wrapping surfaces were used to prevent the quadriceps tendon from passing through the anterior aspect of the femoral component and to prevent the medial collateral ligament from passing through the distal femur.

All iterations of the model were controlled with a massless simulated linear actuator. Actuators were placed on the femur, on the pelvis stage (oriented vertically), in the ground reference frame (oriented vertically over the hip and ankle joints), and in the ground reference frame (oriented horizontally, in line with the hip joint when the knee was flexed 30°). The velocity of the actuators were tuned so that knee flexions from 30° to 100° were simulated over a 20 s interval. This slow knee flexion motion was intentionally modeled in an effort to minimize unwanted inertial effects. Varied demands on the knee extensors were simulated by placing 10 and 20 kg masses at the pelvis stage. The quadriceps cable routing,
with respect to the femur, was consistent across simulations.

Contact forces between articulating surfaces were computed using a custom-written implementation of a rigid-body-spring-model (Landon et al., 2009; Li et al., 1997) in which forces depended on the interpenetration of contacting implant surfaces. Surface geometries were derived from CAD representations of size 3 Scorpio (Stryker Orthopaedics; Mahwah, NJ, USA) PS TKR implants. Contact forces were determined from the penetration of these nodes into femoral component surfaces using a Kelvin-Voigt model (Landon et al., 2009; Wineman and Rajagopal, 2000):

\[ F = kx + b\dot{x} \]  \hspace{1cm} (5.2)

where \( F \) is the contact force, \( k \) is the spring constant, \( x \) is the spring node penetration depth, \( b \) is the damping value, and \( \dot{x} \) is the penetration velocity along the surface normal. Values for \( k \) depended on the area represented by each spring node such that \( k \) was equal to the area multiplied by \( 2.568 \times 10^{11} \text{ N} \cdot \text{m}^{-3} \) (Moran et al., 2008). The value used for \( b \) was \( 200 \text{ N} \cdot \text{s} \cdot \text{m}^{-1} \) (Piazza and Delp, 2001; Moran et al., 2008). Frictional contact forces were not modeled.

Based on the orientation of the linear quadriceps actuator, the knee did not always flex from 30° to 100° in a linear manner. All results were based on knee flexion angle, rather than time. Knee flexion angles were determined using Euler angle decomposition of the tibiofemoral rotation matrix according to the convention proposed by Grood and Suntay (1983).

### 5.3 Results and Discussion

Large quadriceps forces that increased with knee flexion were required when the actuator was mounted on the femur or the pelvis stage, while different quadriceps forces were required for actuators affixed to the ground (Figure 5.4). When the muscle actuator was placed on the femur and pelvis, similar quadriceps forces ranging between 550 N to 2400 N were required to create the squatting motion. Mounting the actuator on the ground segment, however, significantly reduced required quadriceps force and quadriceps force did not consistently increase with flexion. A vertical orientation of the actuator led to a relatively constant required
quadriceps force between 150–200 N. Orienting the actuator horizontally initially resulted in no reduction of required quadriceps force, but as the knee flexed and the pelvis descended a reduction of required quadriceps force was noted. Only the results from the 20 kg simulation are shown, but proportional changes in quadriceps forces and contact forces occurred when pelvis masses were changed from 10 kg to 20 kg.

![Image of required quadriceps forces plotted as a function of knee flexion angle.](image)

**Figure 5.4.** Required quadriceps forces plotted as a function of knee flexion angle. Femur-based and pelvis-based quadriceps actuators required increasing quadriceps force as flexion continued, while ground-based actuators required substantially less force.

No significant differences in tibiofemoral or patellofemoral contact forces were found between simulations with femur-based and pelvis-based actuators (Figures 5.5 and 5.6). When the muscle actuator was placed on the femur and pelvis, tibiofemoral forces ranged between 1000–2100N, and patellofemoral contact forces ranged between 650–4400 N. Similar to required quadriceps force, simulations with grounded actuators had tibiofemoral and patellofemoral contact forces with reduced magnitudes. All simulations, with the exception of the one with a vertically oriented quadriceps force, encountered fluctuations in contact forces around 45° of knee flexion. This was due to the fact that the posterior component of the quadriceps force caused the patella to resist transitioning from the anterior flange of the femoral component to the posterior condyles.
Figure 5.5. Tibiofemoral contact forces plotted as a function of knee flexion angle. Femur-based and pelvis-based quadriceps actuators caused increasing contact forces as flexion continued, while ground-based actuators generated less reaction forces between implants.

Despite large differences in loading of the knee joint, the method of actuation did not substantially affect TKR implant kinematics during knee flexion simulations. Internal/external rotation patterns all followed the same basic movement, progressing from roughly $-2^\circ$ at $30^\circ$ of knee flexion, to a neutral pose at $100^\circ$ knee flexion (Figure 5.7). Differences in internal/external rotation between simulations was less than $1^\circ$. Further similarities in kinematics between the simulations can be suggested by the timing of the cam-post mechanism contact. In order for cam-post mechanism contact to occur, the knee must be flexed deeply enough to expose the cam to the post, and the femur must be translated anteriorly far enough on the tibial plateau for the two design elements to intersect. In the two simulations with grounded actuators, initial cam-post interaction occurred at approximately the same knee flexion angle, between $71^\circ$–$73^\circ$ (Figure 5.8). This was a very light bump, however, and it drove the femur posteriorly on the tibial plateau, causing a drop-off of cam-post forces. Full engagement of the cam-post mechanism was similar for all simulations, and occurred between $85^\circ$–$87^\circ$ of knee flexion.
Figure 5.6. Patellofemoral contact forces plotted as a function of knee flexion angle. Similar to required tibiofemoral contact force plots, femur-based and pelvis-based quadriceps actuators caused increasing contact forces as flexion continued, while ground-based actuators generated less reaction forces between implants.

5.4 Conclusions

A growing number of studies are employing the Oxford Rig to study TKR mechanics. Although this study did not address a specific clinical question, the results will be useful for tailoring future knee simulators to varied design constraints and scientific questions. For example, space limitations may prevent the actuator from being mounted on the femur, but our results show that mounting the actuator on the pelvis stage produces substantially the same knee loading and motions. It was found that when the quadriceps actuator is grounded, muscle and contact forces were reduced, perhaps making this choice less suitable for studying internal knee forces. Knee joint kinematics, however, were largely unaltered by this reduced loading. This suggests that useful investigations of implant motions, if not forces, may be possible with less powerful ground-fixed actuators. Finally, efforts should be made to document the fixation point within publications, as they are often omitted from the literature.
Figure 5.7. Internal/external knee angle plotted as a function of knee flexion angle. All simulations exhibited internal/external knee angles progressing from roughly $-2^\circ$ at $30^\circ$ of knee flexion, to a neutral pose at $100^\circ$ knee flexion.

Figure 5.8. Cam-post mechanism forces plotted as a function of knee flexion angle. Full engagement of the cam-post mechanism was similar for all simulations, and occurred between $85^\circ$–$87^\circ$ of knee flexion.
Chapter 6

Dual-Joint Modeling for the Estimation of TKR Contact Forces During Normal Gait

6.1 Introduction

Since the introduction of total knee replacements (TKR) in the 1970s, prediction of wear characteristics of implants has been a primary concern of clinicians and manufacturers alike. The American Academy of Orthopaedic Surgeons estimates that 581,000 total knee replacements were performed in the United States in 2010, a substantial increase from the National Hospital Discharge Survey’s (2003) estimate of 300,000 performed in 2003. A considerable sector of this population consists of young and active patients, who demand an implant that has a large the range of motion and a superior lifespan. To this end, detailed knowledge of joint loading characteristics within total knee implants is becoming increasingly critical. Determination of the forces applied to joint replacement components during human movement is a challenging goal, however, and is one that is not readily met.

Traditionally, researchers have relied upon in vivo experimental methods to elucidate the causes and methods of implant failure. The limitations that arise when dealing with live subjects are considerable. It is very difficult to put a prospective implant design into a human, and the variability of human subject anatomy and
surgical techniques leads to subjective comparisons of design performance. Aside from in vivo techniques, methods such as bench-top tests and implant retrievals have also improved the overall quality of TKR implants. However, all suffer from the following shortcomings which prevent them from being incorporated into an effective design cycle: 1) testing is time-consuming; 2) materials required for testing are financially burdensome; 3) results do not provide a complete characterization of the contact mechanics.

Computer simulation is a potentially viable alternative method for determination of implant forces, as it does not suffer from the same physical, financial, and technical limitations of traditional investigative methods. Several groups have predicted tibiofemoral loads with muscle-driven dynamic elastic foundation models. Shelburne et al. (2004; 2005; 2006), Taylor et al. (2004), and Kim et al. (2009) have all employed a two-phase approach to create their models. In phase one, a whole body model with an idealized knee joint (such as a pin joint or a ball joint) was used to estimate muscle forces during the entire gait cycle. Distribution of muscle forces were estimated using static optimization schemes or inverse dynamics. In phase two, the resulting kinematics and muscle forces were used as input for an elastic foundation model of a 12 degree-of-freedom knee joint. The resulting tibiofemoral loads were predicted with a quasi-static system of equations. Lin et al. (2010) developed an entirely different approach, in which a surrogate knee contact model was paired with a custom-made optimization routine to simultaneously determine muscle forces and tibiofemoral loads. While all of these methods have proven to be successful in predicting tibiofemoral loads, they either neglect the dynamics of a motion by using a quasi-static approach, or they are not readily replicated for a variety of implant geometries.

The computed muscle control (CMC) algorithm (Thelen et al., 2003; Thelen and Anderson, 2006) provides an attractive framework for the purposes of simulating TKR mechanics during gait. CMC uses forward dynamics, paired with feedback and feedforward controls, and static optimization to determine muscle excitations. The excitations are then used to create muscle forces, which ultimately drive the kinematics of the model. CMC has the ability to simultaneously simulate kinematics and muscle forces for tasks such as pedaling and walking (The-
len et al., 2003; Thelen and Anderson, 2006), and has been used in other studies to investigate muscle contributions during motions such as crouched gait (Steele et al., 2010), running (Hamner et al., 2010), and sprinting (Chumanov et al., 2007). The usefulness of CMC has led to it becoming a module within the OpenSim project (Delp et al., 2007). Previous implementations of CMC required idealized joints such as revolutes and ball joints in order to track measured motions. Models incorporating articular contact, in which contact forces and generalized accelerations depend upon changes in position (i.e. relative positions of implants), have thus far not been successfully integrated into this scheme.

The purpose of the present work is to develop a novel CMC-based paradigm that is specifically suited for the prediction of articular contact forces within the knee. The methods described are applied to the estimation of tibiofemoral forces, but could straightforwardly be adapted to determine patellofemoral forces or contact forces within other joints. If simulation-based estimates of articular contact loads are to be judged sufficiently valid for use in altering designs or surgical techniques, those estimates must be tested against experimentally measured forces. Therefore, the results of these simulations will be compared to results obtained from an instrumented implant in a previous study (Zhao et al., 2007).

6.2 Materials and Methods

Experimental data collection

The experimental data for this study was provided by the organizers of the “Grand Challenge Competition to Predict In Vivo Knee Loads,” hosted by the ASME Summer Bioengineering Conference. A brief summary of the data collection methods will be outlined below:

Data was collected from a single adult male (age, 80 years; mass 67 kg; height 1.72 m; 8 months subsequent to TKR surgery). Institutional review board approval and patient informed consent were obtained. The subject had 14 sets of electrodes placed on their skin to collect surface electromyography (EMG) data. EMG data for the following muscles was recorded: semimembranosus, biceps femoris long head, vastus medialis, vastus lateralis, rectus femoris, gastrocnemius medial head,
gastrocnemius lateral head, tensor fascia latae, tibialis anterior, peroneus longus, soleus, adductor magnus, gluteus maximus, and gluteus medius.

The subject was fitted with 43 retro-reflective surface markers for use with a video motion analysis system (Motion Analysis Corporation, Santa Rosa, CA), and was asked to perform static trials (for motion analysis calibration), maximum isometric trials (for EMG calibration) and to walk in an instrumented gait lab with a normal gait, trunk swaying gait, trunk thrusting gait, and gait with walking poles, all at self-selected speeds. During the walking trials ground reaction data (AMTI Corporation, Watertown, MA), motion analysis data, EMG data, and instrumented knee implant data were all simultaneously recorded.

Musculoskeletal Model Development

Marker data from static trials were used as locations for bony landmarks, which enabled the creation of a properly scaled whole-body model in OpenSim (Delp et al., 2007). Masses and inertial characteristics of the segments were computed for the subject using relationships for predicting anthropometric values published by Zatsiorsky (Zatsiorsky and Seluyanov, 1985; Zatsiorsky, 2002) (Table 6.1). The inverse kinematics subroutine within OpenSim was utilized to generate a set of full-body joint angles that re-created the motions recorded as marker trajectories with Motion Analysis. The full-body model was reduced to a five-segment model of the lower limb and exported to SIMM (MusculoGraphics, Inc.; Santa Rosa, CA, USA) so that a dynamic musculoskeletal model, complete with equations of motion, could be created with Dynamics Pipeline (MusculoGraphics, Inc.; Santa Rosa, CA, USA) and SD/FAST (Parametric Technologies; Needham, MA, USA). The model included the following segments: pelvis, right femur, right tibia, patella, and foot.

The five-segment model included 13 active muscles that crossed the knee joint with force generating properties and attachment points appropriately scaled from the model presented by Delp et al. (1990) (Figure 6.1). The active muscles used in the model were as follows: biceps femoris long head (BFLH), biceps femoris short head (BFSH), gastrocnemius lateral head (GL), gastrocnemius medial head (GM), gracilis (GRAC), rectus femoris (RF), sartorius (SAR), semimembranosus
Table 6.1. A summary of the masses and moments of inertia for the segments within the model. Anthropometric values were based on the relationships established by Zatsiorsky (1985; 2002) for a male subject that had a mass of 67 kg and a stature of 1.72 m.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mass (kg)</th>
<th>Moments of Inertia (kg·m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>8.008</td>
<td>Ixx,yy = 0.067</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Izz = 0.054</td>
</tr>
<tr>
<td>Femur</td>
<td>10.151</td>
<td>Ixx,zz = 0.206</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = .042</td>
</tr>
<tr>
<td>Femoral component</td>
<td>0.200</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patella</td>
<td>0.024</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patellar component</td>
<td>0.025</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Tibia</td>
<td>3.103</td>
<td>Ixx,zz = .0397</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = .006</td>
</tr>
<tr>
<td>Tibial component</td>
<td>0.200</td>
<td>Ixx,yy,zz = .001</td>
</tr>
</tbody>
</table>

(SM), semitendinosus (ST) tensor fascia latae (TFL), vastus intermedius (VI), vastus lateralis (VL), and vastus medialis (VM).

The following passive ligament groups were modeled as springs with quadratic force-deformation relationships: medial collateral ligament (MCL), lateral collateral ligament (LCL), posterior cruciate ligament (PCL), patellar ligament (PAT), and two patellofemoral “ligaments” used to represent the soft tissues surrounding the anterior portion of the knee joint. The PCL was modeled using ten fibers, with attachment points and slack lengths chosen such that the PCL develops force at 80° flexion when natural tibiofemoral motions are applied (Makino et al., 2006; Kurosawa et al., 1985). The knee model created by Piazza and Delp (2001) provided guidelines for the creation of the PAT, LCL and MCL. Cylindrical wrapping surfaces were used to prevent the quadriceps tendon from passing through the anterior aspect of the femoral component and to prevent the MCL and LCL from passing through the distal femur. The soft-tissue “ligaments” were attached to the patella at the midline on the medial and lateral sides, and were attached to the femur at the medial and lateral bone surfaces along the trans-epicondylar axis. The slack lengths of these ligaments were tuned such that they would not generate more than 20 N of force during gait.

CAD models of CR TKR implants were reverse engineered (Geomagic, Inc.; Research Triangle Park, NC, USA) from point cloud data, and were placed on their respective bones based on post-surgery CT data (Lin et al., 2010). Approximate masses and moments of inertia for the implants were estimated from component
Figure 6.1. The five segment model included 13 active muscles and 4 ligament groups. Two non-physical patellofemoral ligaments were used to represent the soft tissues surrounding the anterior portion of the knee joint.

dimensions and material properties (Table 6.1).

**Dual-Joint Modeling Paradigm Development**

Computation of a dynamically consistent set of knee muscle forces is problematic when implementing a knee joint that incorporates contact forces. In our case, a 12-degrees-of-freedom joint represents the knee, and tibiofemoral forces, patellofemoral forces, and muscle forces are all dependent upon the positions and velocities of the femur, tibia, and patella all influence but . A method such as Computed Muscle Control (Thelen et al., 2003; Thelen and Anderson, 2006) cannot be applied for the determination of knee contact forces because it depends
upon the determination of required joint torques to determine muscle forces during static optimization. In the case of CMC, the knee joint is represented as a 1-degree-of-freedom joint, whose kinematics are based upon knee flexion angle, allowing for easy computation of required joint torques about the knee. In the case of a 12-degrees-of-freedom knee, the relative positions and velocities of the femur, tibia, and patella are influential to the kinematics of the knee, and therefore make calculation of required joint torque impossible. This problem was solved in the present study through the use of a dual-joint modeling approach in which two different models are used in conjunction with each other to represent the knee.

Model 1 is based on a ball-jointed knee that is used to compute the muscle forces that track measured knee motions by employing a proportional-derivative controller (for computing desired knee accelerations), inverse dynamics (for determining required knee torque), and static optimization (for distributing muscle forces) (Figure 6.2). At any time in the simulation when $t = n \times 0.005$ s (where $n$ is an integer), the pelvis, hip, and ankle motions (generated with inverse kinematics) were prescribed, and the translations of the tibiofemoral joint are locked into place, allowing the knee to only rotate in the $x$, $y$, and $z$ axes. At this point the simulation controls the knee with a proportional derivative (PD) feedback controller. The positions and velocities of the knee flexion angle are subtracted from the expected positions and velocities of the knee flexion angle. The errors in position and velocity are multiplied by gains ($k_p = 750s^{-2}$ and $k_v = 25s^{-1}$), which were determined through trial-and-error. These correctional accelerations are then added to the expected acceleration of the knee flexion to produce a desired acceleration. Desired acceleration, measured ground reaction force, and measured free moment are used as inputs for an inverse dynamics calculation, which produces the required flexion/extension torque about the knee joint.

EMG signals collected during the gait trials were trimmed to one gait cycle, rectified, and filtered with a second order low-pass Butterworth filter with a cutoff frequency of 6 Hz. The magnitudes of the excitation were normalized to the maximum isometric contraction trials. The resultant EMG signals had patterns that were representative of gait, but the magnitudes of the signals did not seem reasonable. Consequently, the magnitudes of the EMG signals were readjusted so
Figure 6.2. A block diagram for “Model 1” in which desired knee acceleration was tracked with a feedback controller. The required torques were calculated with inverse dynamics, and muscle moment arms were determined based on the pose of the the same knee joint in an Oxford Rig simulation. Recorded EMG data provided preferred activation envelopes of ±10% for the output of the static optimization.

that maximum magnitudes matched those determined by Perry (1992). It should be noted that EMG signals were collected for 8 (BFLH, GL, GM, RF, SM, TFL, VL, VM) of the 13 muscles within the model. Simulated EMG data of the VI were based on the average of the VL and VM. Simulated EMG data for the BFSH, GRAC, SAR, and ST were based on those determined by Perry (1992) during gait.

Muscle activations were optimized using a subroutine in the Dynamics Pipeline, which attempted to match muscle activations within a ±10% envelope of the measured EMG signals while balancing required torques from the PD controller. Muscle moment arms necessary for static optimization were stored in lookup tables created from the output of a simulated Oxford Rig test which employed the same lower limb model and implant geometry (Landon et al., 2009; Thompson et al., 2011). The resulting muscle forces are passed as inputs into the forward dynam-
ics calculation (performed with SIMM/Dynamics Pipeline (MusculoGraphics, Inc.; Santa Rosa, CA) and SD/FAST (PTC, Inc.; Needham, MA)).

Model 2 replaces the ball-jointed knee with a joint that is based on articular contact and has 12-degrees-of-freedom. This model is used to integrate forward in time with a variable time step integrator for the duration of the Model 1 time step, $\Delta t = 0.005$ s (Figure 6.3). During these instances in time, the positions, velocities, and accelerations of the pelvis, hip, and ankle motions were prescribed. No effort was made to control the model aside from applying constant muscle activations that were determined in Model 1. Muscle and ligament forces were subject to change based on changes in length and velocity, and contact forces differed based on changes in the relative positions and velocities of implants.

Figure 6.3. A Block diagram for “Model 2.” If $t \neq n \times 0.005$, where $n$ is an integer, the activations for all muscles remain constant. The muscle forces, ligament forces, and damping torques are based upon the position and velocity of the lower limb segments. The free moment, ground reaction force, and non-knee kinematics are used as inputs to the forward dynamics engine. The resulting positions and velocities of the tibiofemoral and patellofemoral joints determine the articular contact forces.
Contact forces in Model 2 were determined using a rigid body spring model approach (Li et al., 1997), in which arrays of spring nodes are placed on the tibial and patellar contact surfaces. Both components were represented by a single NURBS surface and nodes representing spring elements were placed approximately 2 mm apart on the surfaces. Contact forces were computed from the penetration and penetration velocity of each spring node into a patchwork of 3 NURBS surfaces representing the femoral component (Landon et al., 2009) using the following Kelvin-Voigt model:

\[ F = kx + b\dot{x} \]  

(6.1)

where \( F \) is the contact force, \( k \) is the spring constant, \( x \) is the spring node penetration depth, \( b \) is the damping value, and \( \dot{x} \) is the penetration velocity along the surface normal. Values for \( k \) depended on the area represented by each spring node such that \( k \) was equal to the area multiplied by \( 2.568 \times 10^{11} \text{ N} \cdot \text{m}^{-3} \) (Moran et al., 2008). The value used for \( b \) was \( 200 \text{ N} \cdot \text{s} \cdot \text{m}^{-1} \) (Moran et al., 2008; Piazza and Delp, 2001). Light rotational dampers were employed within the patellofemoral and tibiofemoral joints to prevent chattering of the implants. Equal and opposite resulting torques were applied to both bodies within the joint and were in the range of 2 Nm. Frictional contact forces between implants were not modeled.

Preliminary simulations tracked knee flexion well and provided qualitatively acceptable muscle activation patterns. However, estimates of muscle forces and tibiofemoral contact forces were unrealistically high. In order to reduce the contractile forces of the muscle for a given activation, maximum isometric forces were uniformly decreased. All other input parameters for the model were kept constant, and the resulting motions and contact load estimations were recorded.

### 6.3 Results

Initial simulations estimated total tibiofemoral loads (medial load + lateral load) to exceed 4.5 body weights during stance (Figure 6.4). Uniform decreases in maximum isometric force of the muscles did not change the kinematics of the knee during gait, but effectively decreased the tibiofemoral loads within the joint. Previous studies have suggested that maximum total tibiofemoral loads should be
between 2.0 and 2.5 body weights (Shelburne et al., 2006; Zhao et al., 2007), which corresponded with simulations that employed 50% of the maximum isometric forces suggested in the Delp model (1990).

![Figure 6.4](image)

**Figure 6.4.** Total tibiofemoral contact loads plotted against percent gait cycle. Maximum isometric forces were decreased to simulate an older subject.

Simulations of the three gait trials predicted two peaks in total tibiofemoral loads during stance (Figure 6.5). The peaks ranged in magnitudes between 2.5–3.3 body weights. During this time, medial compartment loads peaked between 2.0–2.6 body weights (Figure 6.6), and lateral compartment loads were between 0.3–1.0 body weights (Figure 6.7). Total tibiofemoral loads decreased during the initial swing and mid-swing phases of gait to about 0.5 body weights and the majority of these loads came from the medial compartment. An increase in contact forces occurred in both compartments during terminal swing.

The dual-joint modeling paradigm was able to track the knee flexion angle well for the three gait trials (Figure 6.8), while providing qualitatively acceptable muscle activations (Figure 6.9). The mean RMS error for knee flexion angle for the three gait trials was 2.53°, with the largest deviations occurring during early stance. Similarly, the largest deviations from the 10% EMG envelopes occurred during early stance.
Figure 6.5. Total tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.

6.4 Discussion

In this study, we sought to develop a CMC-based paradigm that can be used quickly and effectively to predict tibiofemoral contact forces of unique implant geometry during gait. We achieved this goal by implementing a dual-joint modeling paradigm, which employs 1) a ball-jointed knee, inverse dynamics, and a PD controller to control knee flexion angle and 2) a forward dynamics simulation paired with a rigid body spring model to predict contact forces. This simulation incorporates co-contraction of the muscles spanning the knee joint by incorporating EMG signals into the static optimization of muscle forces. Further, simulation output isolates the magnitude of medial and lateral compartment loads and also permits the investigation of the locations and distributions of contact forces at any point in the simulation (Figure 6.10 and 6.11). Generating simulations of a single gait cycle based on experimental data required 50 minutes of computer time and did not require the time-consuming creation of a surrogate contact model (Lin, 2010; Lin et al., 2010).

The changes made to the maximum isometric forces were not made without
Figure 6.6. Medial compartment tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.

reason. Leg muscles in the default SIMM model are based on a 37 year-old male (Delp et al., 1990) citep Brand 1983 and Wickeiwicz, but our subject was XX years old. A final estimate of 50% maximum isometric force seemed to be reasonable for an older subject, as Thom et al. showed that peak muscle power of an old male population (age: 73.9 ±3.8 years) is about 45% of that of young men (age: 26.5 ±4.1 years) (Thom et al., 2005).

Estimations of medial and lateral tibiofemoral contact forces for three separate gait trials were comparable to the magnitudes and patterns of the instrumented implant used by Zhao et al. (Zhao et al., 2007). In the medial compartment, the double-peaked pattern of forces can be seen in the simulations and instrumented implant (Figure 6.6), but the magnitudes of simulated contact forces were approximately 0.5-1.0 body weights higher than the instrumented implant throughout gait. The lateral compartments in the simulations and instrumented implant exhibited a relatively flat profile throughout the gait cycle (Figure 6.7). In contrast to the medial compartment, the simulated contact forces were between 0-0.5 body weights lower than the instrumented implant. When the medial and lateral forces are summed, the timing of the double-peaked profiles of the simulations and the
Figure 6.7. Lateral compartment tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.

The instrumented implant match well (Figure 6.5). Differences of 1.0 body weights can be seen in the first peak, while the magnitudes of the second peak are about 2.5 body weights for the simulations and the instrumented implant. It is encouraging that we see the same general patterns between simulations and the instrumented implant, and it is likely that some differences in magnitude and timing of contact forces may be accounted for by differences in implant geometry and gait.

There are several areas in which this model could be improved. It is important to note that only knee flexion angles were tracked in these simulations. This decision was made for two reasons. First, the static optimization of muscle forces provided unrealistically high co-contraction when it was asked to solve for torques in the the x, y, and z axes. Second, it was determined that the changes in internal/external rotations and varus/valgus knee angles (calculated through inverse kinematics) were not accurate enough for the purposes of tracking. The model had difficulty tracking kinematics and activation patterns during early stance. This is a challenging portion of the gait cycle to control, as small errors in knee angle change the position of the foot with respect to the ground. The position of the ground reaction force remains unaffected, which causes deleterious modifications.
Figure 6.8. Knee flexion angles plotted against percent gait cycle. The three simulations of gait are represented with solid lines, while the corresponding experimental values are represented with dashed lines of the same color.

to knee moment calculations during inverse dynamics.

The results of this study demonstrate the utility of a dual-joint modeling approach for predicting articular contact forces within a joint. To date, the CMC algorithm relies upon idealized joints, and therefore cannot be used to estimate contact forces. This paradigm addresses this issue and has the ability to predict the magnitude and location of contact forces that cause wear. The dual-joint modeling paradigm is still in its infancy, but it has the potential to be a fast and cost-effective means to determine contact forces within any joint in the body.
Figure 6.9. 13 muscle activations and preferred activation envelopes for one simulation, plotted against percent gait cycle. The muscle activations are represented by the solid black line, while the preferred activation envelopes (established by adding ±10% to the measured EMG signals) are shown as gray bands.
Figure 6.10. A breakdown of total tibiofemoral contact forces (green) into medial (blue) and lateral (yellow) loads for one simulation can be seen at the top. Below are pictorial representations of contact forces, based on a color-coding scheme, at 0%, 20%, 40%, 60%, 80%, and 100% of the gait cycle. A close-up of one of these plots can be seen in Figure 6.11.
Figure 6.11. An close-up overhead view of a right tibial component. The gray dots represent the bed of springs that was placed on the surface of the implant. Colored dots represent active springs, and the magnitude of force on that spring is approximated by a color-coding scheme. When viewed sequentially throughout a gait cycle, these pictures create a movie that is descriptive of the dynamic changes in location and magnitude of tibiofemoral forces.
Conclusions

7.1 Summary

This dissertation consisted of four separate studies that used muscle-driven dynamic simulations to investigate TKR performance during challenging tasks (Oxford Rig simulations) and everyday activities (dual-joint modeling of gait). Although these studies investigated different research questions, they all involved the simultaneous prediction of kinematics and joint contact forces within the knee joint. In the course of this dissertation, a novel computational methodology was developed that has the potential to be used by clinicians, manufacturers, and fellow researchers.

A dynamic computer simulation of an Oxford Rig was used to quantify the kinematic and kinetic effects of joint line elevation in primary TKR (Chapter 3). It is not uncommon for a TKR patient to have excessive laxity in their collateral ligaments, which causes knee instability. A surgeon can address this issue by elevating the joint line, which consists of thickening the tibial component and moving the femoral component proximally. Several groups (Emodi et al., 1999; Scuderi and Insall, 1989) have advocated the use of PS implants when joint line elevation is necessary, but neither group quantified the effects of systematic joint line elevations in CR and PS knees. The results of this study showed that systematic increases in joint line elevation negatively affected the kinematics of CR knees and increased the contact forces that were imposed on the implant. Conversely, PS
implants provided predictable kinematics and consistent tibiofemoral joint forces, regardless of the degree of joint line elevation. Although our results do not significantly change the methods that have been established with clinical experience, the results of this study bolstered our confidence that a computational simulation of an Oxford Rig could be effectively used to investigate TKR performance.

The same computational model of the Oxford Rig was used to examine the effects of femoral component upsizing in revision TKR (Chapter 4). Excessive collateral ligament laxity is especially common in revision total knee arthroplasty. As shown in the previous study, a surgeon has the option to elevate the joint line to ensure stability of the knee. An alternative, and often concomitant, strategy for limiting the joint line elevation is to increase the size of the femoral component. The mechanical implications of mismatching femoral and tibial component sizes has received little attention to date, and this research question was well-suited for our Oxford Rig simulation. The results of this study suggested that the magnitude of collateral ligament laxity should be indicative of the method used to address collateral ligament laxity. Specifically, if ligament laxity is mild (3 mm or less) it is suggested that femoral upsizing should be avoided, but if more moderate or severe ligament laxity is encountered (more than 3 mm), a combination of femoral upsizing and joint line elevation should be employed.

The Oxford Rig model was augmented to investigate the effects of changing the fixation point of the quadriceps actuator (Chapter 5). A review of the literature revealed that many labs are using Oxford Rigs to investigate TKR mechanics, but the location of the quadriceps actuator has varied from lab to lab. This results of this study showed that quadriceps forces were significantly reduced when the quadriceps actuator was connected to the ground frame of reference. A vertical orientation of the ground-based actuator reduced required quadriceps more than an actuator that was horizontally oriented. There were minimal changes to required quadriceps force when the actuator was attached to the pelvis and femur segments. The kinematics of the Oxford Rig were consistent across all simulations. This study suggested that researchers who use Oxford Rigs with ground-based actuators should limit their studies to kinematic investigations of TKR performance.

The final study in this dissertation developed a novel methodology to predict
tibiofemoral contact forces during gait (Chapter 6). This work was motivated by the Grand Challenge Competition to Predict In Vivo Knee Loads, hosted by the ASME Summer Bioengineering Conference, and has received a podium presentation for the proceedings in June, 2011. Although other groups have created simulations that estimate tibiofemoral contact loads during gait, there is a need for further development of such models. Our model attempted to expand the capabilities of the CMC algorithm by employing a dual-joint paradigm. The model incorporated \textit{in vivo} EMG data as inputs so that qualitatively realistic muscle actuations patterns were used throughout gait. The simulation results showed that knee flexion angle was tracked well, and predicted tibiofemoral loads were similar to those found in previous studies (Zhao et al., 2007).

7.2 Novelty and Utility of Work

Modern TKR has an excellent clinical history, and continues to be used as an effective means to alleviate pain that is caused by cartilage degeneration in the knee. The success of TKR is largely attributable to the massive amount of research that has been devoted to the subject. While research methods, such as experimental dynamic simulators, implant retrievals, and \textit{in vivo} experiments, have traditionally been used to investigate TKR performance, the utility of computational modeling has begun to gain traction within the orthopaedic community. A small portion of TKR research is based on computational methods, and only a handful of research groups have sought to develop muscle-driven dynamic simulations that can provide detailed characterizations of joint loads. Accurate predictions of muscle forces and joint contact loads during a kinematic activity continues to be a challenging objective, and this dissertation represented a small step towards the realization of this goal.

The Oxford Rig simulation was successfully used to investigate the methods used to address collateral ligament laxity. The methods used to control the Oxford Rig were novel, in that the 3-D model was approximated with a 2-D sagittal plane representation of the rig. This method served as a predecessor for the methods developed within the dual-joint modeling paradigm. The results of these simulations
were not ground-breaking, as they ultimately agreed with previous recommenda-
tions that were based on years of clinical experience. It should be noted, however,
that the patient anthropometry, implant geometry, and component placement can
be changed easily in these simulations. Therefore, these simulations can be easily
tailored to address a wide spectrum of research questions, and may be used by
manufacturers who have limited amounts of clinical experience or by surgeons who
do not perform many TKR procedures in a given year.

The dual-joint modeling paradigm was developed as an effort to create a simu-
lation of gait with a muscle-controlled dynamic elastic foundation model. To date,
no research group has used such a strategy to simultaneously predict kinematics,
muscles forces, and joint contact forces. The results of this study suggest that a
dual-joint modeling approach may permit the CMC algorithm to function with
joints based on articular contact. Ideally, the method would be further developed
and incorporated with the CMC module in OpenSim, making it freely available to
the public.

7.3 Limitations

The Oxford Rig simulations in this dissertation had several limitations. A com-
mon shortcoming for all of the simulations was a lack of surface friction between
implants. Sathasivam and Walker (1997) demonstrated the sensitivity of knee
models to surface friction, and the kinematics of our implants may have been
slightly skewed by its absence. Simulation results were also sensitive to the slack
length and attachment points of the PCL fibers. The PCL was modeled to mimic
the behaviors that were observed in a natural knee study performed by Makino
et al. (2006). Therefore, the PCL properties are only representative of one spe-
cific knee, and do not represent a broad range of anatomical variation. In some
respects, the Oxford Rig test does not faithfully represent an in vivo movement, as
the pelvis is situated directly over the ankle and there is no forward flexion of the
trunk. This creates unrealistically large knee extension moments which may not
be experienced by a patient. The movement is also performed under the control of
a single quadriceps actuator, and the effects of co-contraction are not considered.
Finally, only one set of anthropometric measures was used to represent a subject in these studies, and the implant geometries were limited to a select number of designs and sizes.

The dual-joint modeling simulations also had several limitations. For several reasons, only knee flexion angles were tracked in these simulations. Because small errors in knee joint angles compound themselves during early stance, the model had difficulty tracking kinematics and activation patterns during this portion of gait. The data provided by the organizers of the Grand Challenge Competition to Predict In Vivo Knee Loads was not necessarily well-suited for our needs. No anthropometric data was given for the subject, aside from stature and mass, and the EMG data was unreliable at times. The implant geometries were provided as a reverse engineered mesh. Therefore, approximations of articulating surfaces were made with polysurface drapes so that NURBS surfaces could be employed with our contact algorithm. Finally, like the Oxford Rig simulations, friction was not modeled between implants, and the same PCL model was used in this model.

7.4 Future Work

Both of the models presented in this dissertation could benefit from similar improvements which would make their results more accurate. As noted in the limitations, the inclusion of surface friction between implants has a substantial influence on the kinematics of the implants, and should therefore be included in future modeling efforts. Although great care was taken to model the PCL that was used in our models, it may not be representative of a PCL in an elderly patient who suffers from osteoarthritis. The accuracy of the kinematics of the knee may benefit from a re-modeling of the ligament.

The simulations in this dissertation did not take excessive amounts of time to complete (between 50 minutes to 4 hours); however, there are several areas that could be improved to reduce the time required to create a simulation. The preprocessing of implant surface geometries is a tedious and time-consuming task. Future work should attempt to minimize the number of surfaces that are used to represent implant geometries. This would not only reduce the amount of time required for
preprocessing, but it would reduce the number of surfaces that need to be analyzed within the code, which would lead to faster simulations. To date, no efforts have been made to parallelize the simulation codes, which are “embarrassingly parallelizable” at points. The computational time needed to run simulations has the potential be reduced drastically if graphics processing cards were expertly utilized.

The use of computers within the operating room is becoming more popular, as computer-assisted navigational tools are being utilized to ensure exact implant placement. With a computer in the surgical theater, a surgeon could potentially make patient-specific measures of bone geometries or ligament laxity, and use this as input into a computational simulation. Such a system could provide real-time surgical guidelines for unforeseen choices that must be made, such as joint line elevation and femoral upsizing.

Finally, simulations such as the ones developed in this dissertation, could be of use in the field of orthopaedic rehabilitation. Therapists could elucidate of instability within the knee joint on a patient-by-patient basis, which may reduce the chances of a slip or fall. Computational models could also be used to determine the root cause of knee stiffness, which could lead to the creation of effective patient-specific rehabilitation protocols.
Appendix A

Biomechanical Effects of Total Knee Arthroplasty Component Malrotation

A.1 Introduction

Modern total knee arthroplasty (TKA) is generally considered to be a safe and cost-effective operation to alleviate pain and restore function in patients who suffer from osteoarthritis (OA) (Insall and Kelly, 1986; Ranawat, 1986). Despite the clinical success of TKA, some patients have difficulty performing important activities of daily living such as kneeling, squatting, and rising from a chair (Weiss et al., 2002) and experience quadriceps weakness after surgery (Mizner et al., 2005; Stevens et al., 2003). In some patients, preoperative/postoperative decreases in quadriceps strength can be as high as 60% (Stevens et al., 2003). Some TKA patients also experience stiffness, limited motion, and instability in the knee due to improperly managed collateral ligaments (Fehring and Valadie, 1994).

The success of TKA depends on many factors, including prosthesis design, the preoperative condition of the joint, surgical technique, and postoperative rehabilitation. Error in surgical technique has been suggested to be the most common cause for revision TKA (Stulberg et al., 2002). Incorrect positioning or orientation of the implant can lead to accelerated wear and loosening of the implant and sub-
optimal functional performance (Stulberg et al., 2002). Despite the importance of component alignment to postoperative outcome, there is significant variability associated with femoral component rotational alignment (Jenny and Boeri, 2004; Siston et al., 2008, 2005). Siston et al. (2005) found alignment errors for femoral component rotation ranging from 13° internal rotation to 16° external rotation. It is not uncommon for errors to occur in tibial component rotational alignment in the transverse plane (Akagi et al., 2005; Ikeuchi et al., 2007; Siston et al., 2006), with component rotational alignment ranging from 44° internal rotation to 46° external rotation (Siston et al., 2006). The large variability in component rotational alignment may be due to the fact that small linear errors in identifying anatomic landmarks translate into large rotational errors (Siston et al., 2007).

Variability in femoral and tibial component rotational alignment can lead to improper joint kinematics and pain. Poor rotational alignment of the femoral and tibial components is a major cause of patellofemoral complications (Clayton and Thirupathi, 1982; Dennis et al., 1992), with a strong correlation between combined femoral and tibial component internal rotation and the severity of patellofemoral complications (Berger et al., 1998). Rotation of the femoral component of 5° from the transepicondylar axis has been reported to alter tibiofemoral kinematics and patellar tracking (Miller et al., 2001). It has been suggested that malrotation of the tibial component will lead to impingement of the polyethylene, which could lead to loss of motion and wear (Dalury, 2001). A study of patients with anterior knee pain found that the average tibial component alignment was 6.2° internal rotation, compared with 0.4° external rotation for patients without pain (Barrack et al., 2001). Despite previous reports of internal component rotation being associated with more severe complications than external rotation (Berger et al., 1998; Barrack et al., 2001; Rhoads et al., 1990), the biomechanical reasons for these complications remain largely unknown.

It remains unknown how variability in rotational alignment may impact TKA patients ability to post-operatively perform activities that they deem important. Therefore, the purpose of this study was to determine how the variability in femoral and tibial rotational alignment affects the mechanics of the knee joint. To address this issue, we created a forward-dynamic simulation of an Oxford Rig that simu-
lates a knee flexion under quadriceps control. This rig mimics the knee kinematics that occur when rising from a chair, or climbing stairs, or riding a bicycle (Zavatsky, 1997). Forward-dynamic simulations enable cause-effect relationships to be identified and are well-suited for performing “what if?” studies (Delp et al., 2007), wherein the rotation of a prosthetic component can be precisely changed over a large range of values while holding all other parameters constant. Establishing ranges of “acceptable” and “unacceptable” rotational alignment for the femoral and tibial components may provide orthopaedic surgeons with guidelines that may ensure more successful postoperative functional outcomes.

A.2 Methods

A forward-dynamic simulation of an Oxford Rig was used to simulate a loaded knee flexion performed under quadriceps control. The Oxford Rig uses a ball joint to represent the ankle, a universal joint to represent the hip, and permits the pelvis to translate vertically with respect to the ankle joint. This model topology provides 6 degrees-of-freedom to both the tibiofemoral and patellofemoral joints. A seven-segment model was created using the SIMM/Dynamics Pipeline (MusculoGraphics, Inc.; Santa Rosa, CA) and SD/FAST (Parametric Technologies, Inc.; Needham, MA) software packages. Segment lengths and inertial characteristics of the femur and tibia were computed for a male subject 180 cm tall with body mass of 75 kg and used relationships for predicting anthropometric values published by Winter (1990). Small values were assigned to the moments of inertia of the pelvis, but were inconsequential because the pelvis does not rotate in an Oxford Rig. Approximate masses and moments of inertia for the implants were estimated from component dimensions and material properties (Table A.1).

Internal forces that influenced kinematics in the model were created by one simplified quadriceps muscle (vastus intermedius), and four ligament groups (patellar, lateral collateral, medial collateral, and, when present, posterior cruciate). Force-generating properties of muscle were based on the model presented by Delp et al. (1990). The PCL was modeled using ten fibers, characterized by unique attachment points and slack lengths as described by Makino et al. (2006). Ligaments
Table A.1. Masses and moments of inertia used within the model. Values were based computed for a 180cm tall male with a body mass of 75 kg, based on the anthropometric values of Winter (1990).

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mass (kg)</th>
<th>Moments of Inertia (kg·m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>30</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Femur</td>
<td>6.93</td>
<td>Ixx,zz = 0.1134</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = 0.001</td>
</tr>
<tr>
<td>Femoral component</td>
<td>0.2</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patella</td>
<td>0.025</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patellar component</td>
<td>0.025</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Tibia</td>
<td>3.22</td>
<td>Ixx,zz = 0.0543</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = 0.001</td>
</tr>
<tr>
<td>Tibial component</td>
<td>0.2</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
</tbody>
</table>

were modeled as springs with quadratic force-deformation relationships specified by Piazza and Delp (2001). Other muscles (semitendinosus, semimembranosus, short and long heads of biceps femoris, and medial and lateral gastrocnemius) were included, but remained passive and only generated forces of less than 5 N. Wrapping surfaces were used to prevent muscles and ligaments from passing through bony surfaces of the femur and tibia.

Contact forces between articulating surfaces were computed using a custom written implementation of a rigid-body-spring-model (Landon et al., 2009), in which forces depended on the interpenetration of contacting implant surfaces. Surface geometries were derived from CAD representations of size 5 Scorpio (Stryker Orthopaedics; Mahwah, NJ) cruciate-retaining (CR) and posterior-stabilized (PS) implants.

A proportional-derivative feedback controller was used in conjunction with a simplified planar model of the Oxford Rig to determine the quadriceps force required to lower the pelvis in a controlled manner. We chose proportional-derivative control over other control methods because we believe it is more easily extendable to complex models with multiple muscles requiring static optimization in future studies. The simplified 2-D sagittal plane model was necessary so that estimations of the required knee extension moment could be made without needing to consider the accelerations produced by articular contact forces in a 3-D contact-based model. For both the 2-D and 3-D models, the initial pelvis height corresponded to
20° of knee flexion. A rate of descent of the pelvis of 7.8 cm s\(^{-1}\) was chosen so that the final pose of the 2-D model bent the knee to 120° of flexion in 5 s. A slow rate of descent was chosen to reduce inertial effects and maintain stability during the motion. Within the feedback loop, position and velocity errors of the 3-D model in relation to the 2-D model were multiplied by gains of 60 s\(^{-2}\) and 2 s\(^{-1}\) (arrived at by trial-and-error). The resulting acceleration terms were then added to the expected acceleration (0 ms\(^{-2}\)) to find the corrected 3-D pelvis acceleration that was used to track the descent of the 2-D pelvis. The corrected acceleration and closed-form equations of motion for the simplified 2-D model were used to compute the desired knee extensor moment for the 3-D model. By assuming a quadriceps moment arm that linearly decreased from 4 cm to 2 cm as knee flexion changed from 20° to 120° (Draganich et al., 1987), the required force to be applied along the line of action of vastus intermedius was determined.

The default positions of the TKA assembly were established according to the manufacturers surgical guidelines, relative to bony landmarks on the femur and tibia. Identical default positions with a 0° cut on the tibial plateau were used for CR and PS designs. Based on our previous studies (Siston et al., 2005, 2006), femoral rotational alignment was varied between 15° external rotation and 15° internal rotation in 5° increments, and tibial rotational alignment for the CR implant was varied between 20° external rotation and 20° internal rotation in 5° increments. Due to its more highly constrained design, the simulation failed for the PS implant at some of the more extreme tibial rotational alignments. Therefore, tibial rotational alignment of the PS implant was only varied between 15° external rotation and 15° internal rotation in 5° increments. Combinations of these rotations provided 63 simulations for the CR implant design and 49 simulations for the PS implant design, for a total of 112 simulations. We analyzed the effects of these rotations on quadriceps muscle force, collateral ligament forces, and varus/valgus knee angle using the convention of Grood and Suntay (1983), with bone-embedded coordinate systems. Additionally, we analyzed tibiofemoral anterior/posterior translation as defined by the centers of pressure for the two implant designs using implant-embedded coordinate systems.
A.3 Results

The femoral component alignment had a greater effect on most variables of interest than the tibial component alignment or choice of implant design. Internal rotation of the femoral component yielded more undesirable results than external rotation of the femoral component.

Quadriceps Muscle Force:

Internally rotated femoral components required the highest forces of the lumped quadriceps in deep flexion of the simulated squat. The combination of internally rotated CR femoral components and internally rotated tibial components caused increases in the required quadriceps force, with a maximum value of 5470 N (Figure A.1). The combination of externally rotated femoral components with any rotation of tibial components produced nominal changes in quadriceps force, with a minimum value of 4770 N. In all cases, the quadriceps forces were the same until approximately 80° of flexion.

![Figure A.1. Effect of femoral and tibial component rotational alignment on quadriceps force at 120° knee flexion for the CR implant design. The quadriceps demand was greatest for internal femoral and tibial component rotations.](image-url)
Collateral Ligament Force:

Force in the medial collateral ligament (MCL) was only generated for internal femoral component rotations of 10° or greater, while force in the lateral collateral ligament (LCL) was only generated for external femoral component rotations of 10° or greater. This was true for both the CR and PS implant designs, regardless of tibial component rotation. When the femoral component rotation was varied with the tibial component held fixed at 0°, the maximum MCL forces were approximately 475 N at 90° of flexion for the CR implant and approximately 700 N at 120° of flexion for the PS implant (Figure A.2). The maximum LCL force occurred at 20° flexion for both implants, with values of approximately 75 N for the CR implant and approximately 100 N for the PS implant (Figure A.3). When both the tibial and femoral component rotations were varied in the CR implant, internally rotating the tibial component increased the MCL force, but only when the femoral component was at 15° internal rotation.

Figure A.2. Effect of femoral component rotation on MCL force for the CR and PS implant designs. The MCL force is higher for the PS implant design than for the CR implant design and is zero for the neutrally aligned condition.

Varus/Valgus Knee Angle:
Figure A.3. Effect of femoral component rotation on LCL force for the CR and PS implant designs. The LCL force is higher for the PS implant design than for the CR implant design and is zero for the neutrally aligned condition.

The varus/valgus knee angle for both the CR and PS implant designs was similarly affected by variations in femoral and tibial rotational alignment and was more sensitive to variations in femoral component rotation than to variations in tibial component rotation. Externally rotated femoral components induced a varus alignment throughout flexion, with a maximum value of 15° varus, while internally rotated femoral components induced a valgus alignment, with a maximum value of 15° valgus (Figure A.4). Variations in tibial component rotation induced varus/valgus angles of no more than 1°.

Tibiofemoral Anterior/Posterior Translation:

Based on the motion of the center of pressure of the tibiofemoral joint, anterior/posterior translations for both the CR and PS implant designs were more sensitive to variations in tibial alignment than to variations in femoral alignment. An externally rotated tibial component permitted greater anterior translation than an internally rotated tibial component (Figures A.5 and A.6). For the CR implant, the medial condyle consistently exhibited anterior translation, while the lateral contact point exhibited a posterior translation only when the tibial component was
Effect of femoral component rotation on varus/valgus knee angle for the CR implant design. Externally rotated femoral components induced a varus alignment throughout flexion while internally rotated femoral components induced a valgus alignment throughout flexion.

internally rotated (Figure A.5). For the PS implant, all combinations of component rotation exhibited posterior translation. Posterior translations of the medial condyle appeared to be lessened in the cases of external tibial component rotation (Figure A.6). Overall, the PS implant exhibited more posterior translation than the CR implant, which was relatively stationary throughout flexion.

We also analyzed the effects of component alignment variability on patellar tracking and the screw-home motion. These data can be found as supplementary material online, along with video clips of some of the Oxford Rig simulations.

A.4 Discussion

Despite the reported high variability in rotational alignment of the femoral and tibial components in TKA, the biomechanical effects of this variability on functional tasks remained unknown. To our knowledge, no study has systematically investigated the effects of varying both the femoral and tibial components together, nor has any study investigated the effects of component alignment on muscle force or ligament forces during a simulated functional task. Interestingly, we found that
Figure A.5. Effect of femoral and tibial component rotation on center of pressure locations on the medial and lateral tibial condyles between 20° and 120° knee flexion for the CR implant design. The anterior translation is greatest for external tibial component rotation. A superior view of a right tibial component is shown.

Our variables of interest were affected differently by femoral rotation and tibial rotation. Femoral rotation had a greater effect on quadriceps forces, collateral ligament forces, and varus/valgus kinematics, while tibial rotation had a greater effect on anterior/posterior translations.

The advantage of a computational simulation of a single subject is that we can determine the effects of component alignment within the same “person and exclude the effects of variables such as weight, height, bony geometry, ligament properties, and component size. By holding these other variables constant and using a small number of implant designs, which is a common approach for computer simulation studies of TKA (Piazza and Delp, 2001; Piazza et al., 1998), we were able to perform the large number of simulations (112) required for our study and elucidate the effects of variability in component alignment and implant design. However,
Figure A.6. Effect of femoral and tibial component rotation on center of pressure locations on the medial and lateral tibial condyles between 20° and 120° knee flexion for the PS implant design. The anterior translation is greatest for external tibial component rotation. A superior view of a right tibial component is shown. Tibial component rotations between 15° external rotation and 15° internal rotation are shown due to complications with simulating tibial malrotations of greater than 15°.

despite these advantages, it is important to note that implant designs with different tibiofemoral articular conformity, mobile bearing designs, and designs with a medial or lateral pivot would likely influence knee joint forces and kinematics, especially anterior/posterior translation, and could lead to results that are different than the present study. In particular, the Scorpio implant used in this study has femoral condyles which have a single radius of curvature, unlike many other CR implant designs which have separate centers of curvature for the medial and lateral condyles in the coronal plane. An extension of our current work would be to determine the sensitivity of our results to subject characteristics as well as implant geometry.
There are some potential limitations of the Oxford Rig computer model. The Oxford Rig simulates a perfect up-down movement with the hip directly above the ankle and does not simulate trunk flexion. This configuration places a much higher demand on the quadriceps muscle and ligaments than what would be seen in a true squat. The pelvis is only permitted to translate vertically, the foot is permitted to plantarflex a large amount during knee flexion, and the only muscle in the model that carries any force is the lumped quadriceps muscle. Additionally, we assumed material properties and attachment points for the ligaments in the model based on values from the literature, even though there is considerable variability in reported values. However, the objective of the current study was not to determine the actual values for muscle and ligament forces, but instead to determine the effect of variability in component alignment on our variables of interest. The Oxford Rig is an established clinical model and is a commonly used biomechanical testing device (Zavatsky, 1997; Kessler et al., 2009). Combining the proven capabilities of the Oxford Rig with the advantages of forward-dynamic computer simulations was an effective means to perform our parametric investigation of the effects of component alignment variability. The quadriceps force required to perform the squatting motion was greater for the internally rotated femoral components than for the externally rotated femoral components by as much as 500 N for knee flexion angles greater than 80°. Increased quadriceps strength leads to improved functional performance (Hurley and Scott, 1998; Slemenda et al., 1997). Since OA patients and TKA patients experience significant quadriceps weakness (Mizner et al., 2005; Stevens et al., 2003), component rotation which increases the required quadriceps force could make it more difficult for patients to kneel, squat, or rise from a chair.

Large amounts of internal femoral component rotation may be detrimental to the MCL. Errors in femoral rotational alignment have been shown to contribute to an imbalanced soft tissue envelope surrounding the knee and lead to instability and a limited range of motion (Anouchi et al., 1993; Fehring, 2000; Olcott and Scott, 1999). For both CR and PS knees, we found internal femoral component rotation of 15° created forces within the MCL that are above the published yield point of 453 N (Kennedy et al., 1976). Regardless of whether the MCL would rupture, the larger MCL forces associated with internal femoral component rotation would
likely be perceived as stiffness by TKA patients, which may inhibit post-operative function. The increase in LCL force induced by external rotation of the femoral component may be due not only to external rotation of the tibia with respect to the femur, but a combination of rotation and translation. In the model, when the femoral component is externally rotated, it appears the femur is positioned more medially on the tibia in early flexion, and then translates laterally as the knee is flexed, which may explain why LCL force is greatest in extension and decreases with flexion.

Our results of a simulated functional task show a strong correlation between component alignment and tibiofemoral kinematics and are consistent with previous studies. Internally rotated femoral components have been shown to induce dynamic valgus in the knee (greater valgus orientation with knee flexion) (Miller et al., 1998; Anouchi et al., 1993). We found that an externally rotated tibial component biases the position of the medial condyle posteriorly, which is a potential explanation for the paradoxical anterior femoral translation which has been observed previously following TKA (Siston et al., 2006; Dennis et al., 1998). Additionally, we found more posterior femoral rollback for the PS implant than for the CR implant, which is consistent with a previous in vivo study of knee bend activities (Haas et al., 2002). Our results for patellofemoral kinematics also agreed with those of previous studies (Miller et al., 2001; Barrack et al., 2001; Anouchi et al., 1993), which found that internally rotated femoral components tilt and displace the patella more medially compared with neutrally aligned femoral components. Details can be found in the supplementary material.

In order to test the accuracy of the motions predicted using the simulated Oxford Rig, we performed two validation studies. First, measurements were made using a physical Oxford Rig and compared to simulations that were tailored to match its characteristics (Table B.1). The physical model held a mechanical linkage fitted with a TKA rather than a cadaver specimen. After correcting for minor errors in registration of the components in the physical model, simulated collateral ligament lengths and knee angles (abduction/adduction and internal/external rotation) were found to agree with measured values with average RMS errors between simulation and experiment of approximately 1.29 mm and 0.90°. Second, because
we are using ligament properties from the literature, we performed a sensitivity analysis of the effect of variations in our ligament parameters. We found that increasing the stiffness of the MCL, LCL, and PCL or decreasing the slack length of the MCL led to an increase in ligament forces, but had no effect on varus/valgus knee angle or the required quadriceps force. Decreasing the slack length of the PCL led to an increase in ligament force and a decrease in quadriceps force, but had no effect on varus/valgus knee angle. Although ligament forces increased with increased stiffness or decreased slack length, the forces increased in a predictable manner and exhibited the same trends as the ligament parameters chosen for our simulations. These validation studies give us confidence in our simulation results. Full details of these tests are provided in the supplementary material.

Our findings may support the tendency for orthopaedic surgeons to bias the femoral component into external rotation (Siston et al., 2005) and avoid malrotation of the tibial component. Internal rotations of the femoral component of more than 10° place a greater demand on the quadriceps and induce higher forces in the MCL, which may translate into joint stiffness, while external alignment appears to have some advantages, such as less quadriceps demand. External rotations of the tibial component permit more anterior femoral translation. These results suggest that the combination of internally rotated femoral components and externally rotated tibial components may inhibit function in TKA patients following surgery. Future studies may capitalize on the predictive capabilities for forward dynamic simulations, and may help us understand the relationship between surgical parameters and the ability to perform important functional activities. Such investigations may be key steps towards ensuring patient satisfaction and improving post-operative performance.

A.5 Acknowledgments

Funding for this research was provided by the National Science Foundation Graduate Research Fellowship program to Julie Thompson. The authors thank Stryker Orthopaedics for supplying the implant geometry and Ajit Chaudhari for his assistance.
Appenidx  
B  

Supplementary Material to the  
Biomechanical Effects of TKR  
Malrotation  

B.1 Patellar Kinematics  

Patellar Tilt:  
The patellar tilt for the CR and PS implant designs was similarly affected by  
variations in femoral and tibial rotational alignment, but was much more sensitive  
to variations in femoral component alignment than to variations in tibial compo- 
nent alignment. Internally rotated femoral components induced a medial patellar  
tilt, while externally rotated femoral components induced a lateral patellar tilt.  
These results are consistent with previous studies that have reported that inter- 
nally rotated femoral components tilt the patella more medially compared with  
the neutral alignment (Miller et al., 2001; Rhoads et al., 1990). For all component  
rotations, the patella tilted laterally with increasing flexion.  

Patellar Medial/Lateral Translation:  
The patellar medial/lateral translation for the CR and PS implant designs was  
similarly affected by variations in femoral and tibial rotational alignment. In early  
flexion, the patellar medial/lateral translation was more sensitive to variations in
femoral component alignment than to variations in tibial component alignment. Internally rotated femoral components induced a medial patellar location, while externally rotated femoral components induced a lateral patellar location (Figure B.1). These results are consistent with previous studies that have reported that internally rotated femoral components displace the patella medially compared with the neutral alignment (Rhoads et al., 1990; Anouchi et al., 1993).

**Figure B.1.** Effect of femoral and tibial component rotational alignment on patellar M/L location at 20° knee flexion for the CR implant design. (Positive values indicate a lateral location). Internally rotated femoral components induced a medial patellar location while externally rotated femoral components induced a lateral patellar location.

In mid-flexion, the patellar medial/lateral translation was affected by variations in both femoral and tibial component alignment. Internally rotated femoral components and externally rotated tibial components induced a more medial patellar location while externally rotated femoral components and internally rotated tibial components induced a more lateral patellar location. In high flexion, externally rotated femoral and tibial components induced a more medial patellar location while internally rotated femoral and tibial components induced a more lateral patellar location (Figure B.2).
B.2 Tibiofemoral Kinematics

Screw-Home Motion:

The screw-home motion (internal rotation of the tibia with respect to the femur during flexion) was more sensitive to variations in femoral component alignment than to variations in tibial component alignment. For both the CR and PS implants, externally rotated femoral components exhibited the natural screw-home motion while internally rotated femoral components reversed the screw-home motion (Figure B.3). Previous work has reported a reduced screw-home motion following TKA (Siston et al., 2006). These results provide a potential explanation for why this paradoxical motion has been observed in knees following TKA.
Figure B.3. Effect of femoral and tibial component rotational alignment on screw-home motion for the CR implant design. Internal femoral component rotations reversed the natural screw-home motion.

B.3 Validation Testing

Model predictions of TKA mechanics were first tested by performing Oxford Rig evaluations of a mechanical leg model. The femur and tibia of the mechanical leg were steel rods upon which size 3 Scorpio PS implants were mounted. The patellar button was mounted on a 2.54 cm wide by 0.2 cm thick strip of nylon webbing (Jontay Distributor, Waycross, GA). Medial and lateral collateral ligaments were modeled with rubber bands. Trials using the physical rig were started with the knee flexed to approximately 30° and continued until a flexion angle of 90° was exceeded in about 35 s. A linear actuator under displacement control was connected to a steel cable that pulled on the patella along a line of action directed along the mechanical axis of the femur. Ten trials were performed for each femoral rotation between 10° external rotation and 10° internal rotation, while the tibial component remained neutrally aligned. The motions of the femur, tibia, and patella of the mechanical leg were tracked using clusters of four reflective markers affixed to each segment. Marker locations were tracked using three Eagle cameras (Motion Analysis Corp.; Santa Rosa, CA) running at 100 Hz.

Six points on both the tibial component and femoral component were chosen as
reference points, used to align the computational model with the Motion Analysis data. The fiduciary points were chosen based on two characteristics: 1) easily identifiable by the naked eye, and 2) easily reached with a wand during static Motion Analysis trials. They can be seen as yellow diamonds in Figure B.4

**Figure B.4.** Effect of femoral and tibial component rotational alignment on patellar M/L location at 20° knee flexion for the CR implant design. (Locations of the fiduciary points for the femoral and tibial components.

Twelve Motion Analysis trials were performed in which the tip of a retro-reflective wand was placed at each fiduciary point. This process established the location of the fiduciary points in the global frame of reference. Within each trial, technical frames of reference were also established by creating coordinate systems that have an origin within the femoral and tibial clusters. The components and the clusters were rigidly attached to the femur and tibia; therefore, we were able to make transformations between the ground and the technical frames. The CAD coordinate systems for the femoral and tibial components provided anatomical frames of reference. Through a final series of transformations, the fiduciary points were placed in the anatomical reference frame. The locations of the fiduciary points were then compared with the locations of the same points in the CAD model and the average of the errors between the fiduciary points and the CAD model were calculated to be 1.25 mm for the femur and 2.19 mm for the tibia.

Because of the blunt tip of the wand and human error, slight errors occurred during the fiduciary trials. To determine if certain trials had more errors than others, one fiduciary point at a time was systematically removed, and the average
of the errors between the fiduciary points and the CAD model was re-evaluated. Removing one fiduciary point for both the femur and the tibia provided a substantial decrease in the errors between the rig and the computational model, lowering the average error to 0.90 mm for the femur and 1.36 mm for the tibia.

Computational simulations of the Oxford Rig were performed after assigning inertial properties that corresponded to those of the mechanical leg (Table B.1). Small values were assigned to the moments of inertia of the pelvis, but were inconsequential because the pelvis does not rotate in an Oxford Rig. Passive muscles were not included in these computational simulations, as they did not exist in the mechanical leg. Internal/external rotations, varus/valgus angles, and collateral ligament lengths were calculated as a function of knee angle and were compared between the Motion Analysis results from the physical rig and the output of the computational simulations of the same rig.

Table B.1. Masses and moments of inertia used within the model. Values of masses were measured with a balance, and moments of inertia were calculated after performing pendulum experiments with the segments.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mass (kg)</th>
<th>Moments of Inertia (kg m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>10.074</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Femur</td>
<td>1.475</td>
<td>Ixx,zz = 0.121711</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = 0.001844</td>
</tr>
<tr>
<td>Femoral component</td>
<td>0.2</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patella</td>
<td>0.025</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Patellar component</td>
<td>0.025</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
<tr>
<td>Tibia</td>
<td>1.37</td>
<td>Ixx,zz = 0.108282</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Iyy = 0.001713</td>
</tr>
<tr>
<td>Tibial component</td>
<td>0.2</td>
<td>Ixx,yy,zz = 0.001</td>
</tr>
</tbody>
</table>

The physical rig results demonstrated the same patterns as the computational model, but suffered from offsets. A program in MATLAB was created to perturb the rotation matrix of the transformation matrices for the tibia and femur. An exhaustive search of combinations of rotations between $-6^\circ$ and $6^\circ$ were tested for the x, y, and z Euler angles of the femur and tibia. The sum of the RMS error for the measures of interest (internal/external rotation, varus/valgus angle, MCL length, and LCL length) was used as the objective function. This methodology yielded a femoral x-y-z rotation of $3^\circ$, $-4^\circ$, $4^\circ$ and a tibial x-y-z rotation of $3^\circ$, 3$^\circ$.
5°, 6°. After the corrective rotations were made, the measures of interest did not suffer from the same offsets as seen previously (Figures B.5–B.8), and RMS errors were reduced, as shown in Tables B.2 and B.3.

![Figure B.5](image-url)  
**Figure B.5.** Effect of femoral component rotation on tibiofemoral internal / external rotation angle. Externally rotated femoral components induced an external rotation while internally rotated femoral components induced an internal rotation. Dashed lines correspond to the computational simulations, while solid lines correspond to the physical rig.

**Table B.2.** A table outlining the RMS errors for LCL and MCL ligament lengths without any corrective rotations, and their values after the the femoral x-y-z rotation of 3°, −4°, 4° and a tibial x-y-z rotation of 3°, 5°, 6° were applied.

<table>
<thead>
<tr>
<th>LCL Length</th>
<th>MCL Length</th>
<th>RMS Errors (mm)</th>
<th>RMS Errors (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Before Correction</td>
<td>After Correction</td>
<td>Before Correction</td>
</tr>
<tr>
<td>Neutral:</td>
<td>4.8519</td>
<td>0.9355</td>
<td>8.7592</td>
</tr>
<tr>
<td>5° Ext:</td>
<td>3.9542</td>
<td>1.4666</td>
<td>6.7678</td>
</tr>
<tr>
<td>5° Int:</td>
<td>4.7025</td>
<td>0.9995</td>
<td>6.527</td>
</tr>
<tr>
<td>10° Ext:</td>
<td>4.7994</td>
<td>1.0785</td>
<td>8.5534</td>
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<tr>
<td>10° Int:</td>
<td>3.2066</td>
<td>1.9568</td>
<td>7.2761</td>
</tr>
</tbody>
</table>
Figure B.6. Effect of femoral component rotation on varus/valgus knee angle. Externally rotated femoral components induced a varus alignment throughout flexion while internally rotated femoral components induced a valgus alignment throughout flexion. Dashed lines correspond to the computational simulations, while solid lines correspond to the physical rig.

Figure B.7. Effect of femoral component rotation on LCL length. Externally rotated femoral components led to a greater LCL length throughout flexion than internally rotated femoral components. Dashed lines correspond to the computational simulations, while solid lines correspond to the physical rig.
Figure B.8. Effect of femoral component rotation on MCL length. Internally rotated femoral components led to a greater MCL length throughout flexion than externally rotated femoral components. Dashed lines correspond to the computational simulations, while solid lines correspond to the physical rig.

Table B.3. A table outlining the RMS errors for varus / valgus and internal / external rotation angle without any corrective rotations, and their values after the femoral x-y-z rotation of 3°, −4°, 4° and a tibial x-y-z rotation of 3°, 5°, 6° were applied.

<table>
<thead>
<tr>
<th>Varus/Valgus Angle</th>
<th>Internal/External Rotation</th>
<th>RMS Errors (deg)</th>
<th>RMS Errors (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before Correction</td>
<td>After Correction</td>
<td>Before Correction</td>
<td>After Correction</td>
</tr>
<tr>
<td>Neutral:</td>
<td>0.7453</td>
<td>0.2592</td>
<td>4.8548</td>
</tr>
<tr>
<td>5° Ext:</td>
<td>0.8943</td>
<td>0.1749</td>
<td>4.1502</td>
</tr>
<tr>
<td>5° Int:</td>
<td>0.4502</td>
<td>0.7248</td>
<td>3.1845</td>
</tr>
<tr>
<td>10° Ext:</td>
<td>1.2265</td>
<td>0.3566</td>
<td>4.6806</td>
</tr>
<tr>
<td>10° Int:</td>
<td>0.7029</td>
<td>1.0966</td>
<td>5.2382</td>
</tr>
</tbody>
</table>

B.4 Ligament Characteristics Sensitivity Study

A sensitivity analysis was performed to assess how our results would change with variations in ligament parameters, specifically changes in stiffness and slack length.

For the first part of the analysis, we varied ligament stiffness by adjusting the force-length curves in the model. We compared the default MCL stiffness with a doubled MCL stiffness and halved MCL stiffness and found that there were no changes in the knee varus/valgus angle or required quadriceps force for neutral
alignment of the implants. We then performed the same analysis for a 15° internal femoral component alignment, and again found no change in varus/valgus angle or quadriceps force. Next, we varied the LCL stiffness and compared the default LCL stiffness with a doubled LCL stiffness for a 15° external femoral component alignment, which yielded no change in varus/valgus angle or quadriceps force. Finally, we varied the PCL stiffness and compared the default PCL stiffness with a doubled PCL stiffness for a 15° internal femoral component alignment, which again yielded no change in varus/valgus angle or quadriceps force. We did, however, find that the peak MCL force increased by approximately 450 N, peak LCL force increased by approximately 30 N, and peak PCL force increased by approximately 150 N for doubled MCL, LCL, and PCL stiffnesses, respectively.

For the second part of the analysis, we varied ligament slack lengths. First, we decreased the slack length of the PCL by 20% and compared this case to the default PCL slack length for a 15° internal femoral component alignment. We found that peak MCL force increased by approximately 350 N, peak PCL force increased by approximately 2800 N, and peak quadriceps force decreased by approximately 300 N for a 20% decrease in PCL slack length. The varus/valgus knee angle, however, did not change. We then decreased the slack length of the MCL by 5% and compared this case to the default MCL slack length for a 15° internal femoral component alignment. We found that peak MCL force increased by approximately 1400 N for a 5% decrease in MCL slack length, but there was no change in quadriceps force or varus/valgus angle.

Although ligament forces increased with an increased stiffness or decreased slack length, and the required quadriceps force decreased for a decrease in PCL slack length, the forces varied in a predictable manner and exhibited the same trends as the results from our original choice of ligament parameters. However, the large change in magnitude of the ligament forces illustrates the need to more accurately measure ligament properties.
Appendix C

Supplement to Dual-Joint Modeling for the Estimation of TKR Contact Forces During Normal Gait

C.1 Additional Normal Gait and Trunk Sway Results

The results presented in Chapter 6 were based on a model that was created to represent the experimental data collected from an 80 year old man walking with normal gait. In this model, the maximum isometric forces of all models were reduced by 50% to represent the degradation of muscle strength that is present in an elderly subject. Originally, simulations were run with maximum isometric forces that were provided in the model developed by Delp et al. (1990). The tibiofemoral contact forces and knee flexion tracking results are presented in Figures C.1–C.4.

A vast amount of experimental data was provided by the organizers of the “Grand Challenge Competition to Predict In Vivo Knee Loads.” During data collection, the subject was asked to walk with various forms of gait, including trunk swaying gait. To date, no one has published the tibiofemoral loading patterns that are associated with trunk swaying gait. Because of this fact, the organizers of the event asked competitors to model such a gait pattern, so that results could
not be tailored to previously published data. The tibiofemoral contact forces and knee flexion tracking results for a subject with “100% isometric muscle force” are presented in Figures C.5–C.8.

![Figure C.1. Total tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. These simulations were run without the reduction in maximum isometric muscle force, as modeled in Chapter 6. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.](image)
Figure C.2. Medial compartment tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. These simulations were run without the reduction in maximum isometric muscle force, as modeled in Chapter 6. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.

Figure C.3. Lateral compartment tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of normal gait. These simulations were run without the reduction in maximum isometric muscle force, as modeled in Chapter 6. Results from an instrumented implant from a separate study (Zhao et al., 2007) are plotted with a dashed red line.
Figure C.4. Knee flexion angles plotted against percent gait cycle. The three simulations of normal gait are represented with solid lines, while the corresponding experimental values are represented with dashed lines of the same color. These simulations were run without the reduction in maximum isometric muscle force, as modeled in Chapter 6.

Figure C.5. Total tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of trunk sway gait. Results from an instrumented implant from a separate study of normal gait (Zhao et al., 2007) are plotted with a dashed red line.
Figure C.6. Medial tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of trunk sway gait. Results from an instrumented implant from a separate study of normal gait (Zhao et al., 2007) are plotted with a dashed red line.

Figure C.7. Lateral tibiofemoral contact loads plotted as a function of percent gait cycle for three simulations of trunk sway gait. Results from an instrumented implant from a separate study of normal gait (Zhao et al., 2007) are plotted with a dashed red line.
Figure C.8. Knee flexion angles plotted against percent gait cycle. The three simulations of *trunk sway gait* are represented with solid lines, while the corresponding experimental values are represented with dashed lines of the same color.


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Vita
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Michael Hast graduated from the Pennsylvania State University with his B.S. in Mechanical Engineering in 2004. After graduation, Michael worked as an Associate Project Engineer for Stryker Orthopaedics, in Mahwah, New Jersey. While working for Stryker, Michael helped develop surgical tools and implants for the Triathlon Total Knee System, and headed the Mobile Bearing Implant Development team.

Michael left Stryker Orthopaedics in August of 2005 to return to the Pennsylvania State University to pursue a Ph.D. in Mechanical Engineering. There, he continued to focus his research on total knee replacements under the mentorship of Dr. Stephen Piazza. His master’s thesis (2007), *Analysis of Total Knee Replacements Using Dynamic Computer Simulations of Controlled Knee Flexions*, developed the Oxford Rig model that is used throughout this dissertation. Michael has presented his work at several academic conferences, including oral presentations at the North American Congress of Biomechanics conference, and the American Society of Biomechanics conference. He has been named second author on two publications (Landon et al., 2009; Thompson et al., 2011), and plans to publish the work presented in this dissertation shortly.

During his time at University Park, Michael served as a teaching assistant for several courses within the Mechanical Engineering department, and was a member of a team that is currently developing energy harvesting devices that are to be used in conjunction with trans-tibial prosthetic legs. After completing his dissertation, Michael has accepted a position as an expert biomechanical engineer with AR-CCA, a biomechanical, forensic engineering and litigation consulting firm located in Penns Park, Pennsylvania.