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Contact Stress in a Whole Joint Bioreactor: Intrinsic Levels & Augmentation by External Loading

A Thesis submitted in partial satisfaction of the Requirements for the degree Master of Science

in

Bioengineering

by

Allison Margaret Deppe

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2007
The thesis of Allison Margaret Deppe is approved

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Chair

University of California, San Diego

2007
Dedication

This thesis is dedicated to my family and friends who have given me innumerable bits of encouragement, lots of laughter, loving support and prayers. I would have never reached this point with my sanity without you.

*Hebrews 11:1*
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A whole joint bioreactor was recently described as an *ex vivo* apparatus for studying the biomechanical regulation of cartilage metabolism in knee joints [17]. Stimulation of chondrocyte lubricant biosynthesis by continuous passive motion (CPM) of knee joints was hypothesized to be due to cartilage responses to compressive and shear stresses [18].

In the whole joint bioreactor setup, compressive stress within the joint at the cartilage surfaces in the passive state of different degrees of flexion is unknown. The objective of this study was to assess the distribution and magnitude of contact stress...
within knee joint preparations. The results show significant effects of angle and load magnitude on contact stress and force within the joint. Near full extension, the anterior regions of the LFC exhibited high stress levels, and these generally correspond to regions found previously to respond biologically to CPM. These results provide biomechanical measures and methods to test the role of such cartilage contact stresses in a whole joint bioreactor.

Previously, no methods were available to apply loads externally to manipulate cartilage compressive stress in the whole joint bioreactor. The objective of this study was to examine the effects of axial and patellar loading as possible biomechanical stimulation methods for use in joint-scale bioreactors. Contact stress magnitude and location and overall joint force were increased significantly by both axial and patellar loading. These results suggest that such an approach could also be used to study effects of mechanical loading of knee joints with repair procedures, such as implanted tissue grafts, under controlled loading conditions.
Chapter 1: Background

1.1 Introduction

Articular cartilage is a specialized connective tissue that covers the bones in synovial joints. Normally it functions as a low-friction, wear-resistant and load-bearing material [23]. Injury to articular cartilage in disease states, such as osteoarthritis, causes severe joint pain and morbidity. Furthermore, damaged has a low propensity for self-repair [16]. These qualities are the cause of numerous joint problems, which has motivated the research into many different regeneration or repair techniques.

This background will focus on (1) current and experimental strategies for cartilage repair and regeneration, (2) the role of loading in the regulation of articular cartilage metabolism, (3) techniques used to model joint kinematics and kinetics, and (4) some of the experimental tools utilized in assessing contact stress.

1.2 Treatments of articular cartilage defects

Due to the poor ability for self-repair in cartilage, there is a great need for reliable repair techniques in order to regain the functionality of the tissue. Generally, treatment protocols are defined by the size of the cartilage injury needing repair.

In patients with relatively small lesions, a variety of methods are used to restore the cartilage. Options include implanting autografts or utilizing techniques that enhance the tissue’s natural repair process. The use of autografts is limited due to the small amount of cartilage available in the body for transplantation. The impact at the donor site must also be considered. In addition, the mechanical integrity of tissue from
a lower weight bearing region may not withstand the stresses at the new higher-loading location [25]. Treatments that have been examined include generating microfractures in the subchondral bone, or the introducing growth factors. Microfractures are created by using various methods so that mesenchymal stem cells from the bone marrow migrate into the defect and begin the repair process. Results from such studies have been varied with the size of the defect, the age of the patient and several other factors [19]. However, even when cartilage-like tissue is created, the mechanical properties are typically inferior than that of the native tissue [25]. Growth factors, such as insulin-like growth factor-I (IGF-I), have been injected in order to increase chondrocyte proliferation and matrix deposition. Again, results have been variable, and additionally, osteophytes have been induced to form [19]. More studies are needed in this area before such treatments can be considered effective in the long term.

In patients with larger lesions, few repair strategies are available. Complete joint replacement can provide relief from painful symptoms. However the limited lifetime of the implant makes such treatment problematic for younger patients who may subsequently face difficult revision surgery. Joint replacement has been reliably successful in older patients (those who are 65 years of age and older), leaving the need for a reliable treatment for a large segment of the population [19].

Another option is the in vitro growth of the tissue from a seed of host cells from the patient, or so called autologous transplant. The controlled environment of bioreactors provides numerous advantages to growing tissues. It has been calculated that in order to resurface the entire knee would require an implant approximately 5 cm
in diameter ranging from 1-5 mm thick [25], indicating that efficient methods for growth need to be established in order to make this treatment option feasible. The structure and function of the engineered tissue can be significantly impacted by the compressive and shear stress it encounters [25]. Therefore, the amount of loading applied to engineered tissue must be carefully regulated in order to generate tissue with mechanical properties similar to those of native cartilage tissue.

A novel approach to generating mature articular cartilage tissue is to assemble components to form a whole joint and then guide the articulating joint structure to mature in a joint-scale bioreactor. A major step toward this is to be able to maintain and regulate native joints in bioreactors, extending what is known about the response of cartilage to external parameters. Nugent-Derfus et al. [17] developed a whole joint bioreactor that maintained the health and viability of the articular cartilage of an immature bovine stifle joint. Furthermore, they were able to compare the biochemical response of joints maintained under static conditions and ones subjected to continuous passive motion (CPM). PRG4 secretion was found to be dependent on the amount of contact the site had with opposing articular cartilage.

Overall, the current strategies seek to fill the defect with a cartilage-like material that can bear loads while still maintaining a low-friction surface. However, most of the techniques still fail to generate cartilage that has the same biomechanical properties in the long term. Not only has repair tissue been found to have roughly half the tensile strength of intact cartilage, it may experience up to three times the strain of adjoining native cartilage [8]. Thus, in order to achieve durable cartilage, factors that cause or induce biomechanical changes must be studied further.
1.3 Loading induced changes in cartilage metabolism

Articular cartilage adapts to have distinct biochemical compositions under dynamic and static loading conditions [9, 13, 14, 18]. As the mechanical properties of the cartilage are heavily dependent upon the biochemical makeup of both the cartilage and its immediate environment, the application of loads on developing cartilage could be crucial to guide the attainment of appropriate load-bearing characteristics.

Mechanical loading of the joint can cause changes in oscillatory fluid flow, hydrostatic pressure, potential gradients, interstitial pH, cell-matrix interactions, as well as a cell deformation [9, 13, 28]. In order to examine regulatory factors, differences between responses to static and dynamic loading of cartilage explants are often used. Because the experimental setup for the loading conditions generally alters the boundary conditions (vs. free swelling explants), the static loading condition is used as the control for the dynamic loading [14]. Such studies have found that the mechanical stimuli not only regulate the synthesis and assembly of the load bearing constituents aggrecan and collagen but also proteoglycan-4 (PRG4) that mediates lubrication [18].

Moreover, the stimulatory effect caused by the dynamic loading is frequency dependent, causing an upregulation of proteoglycans and proteins at frequencies greater than 0.001 Hz [9, 13], but not at lower frequencies. Furthermore, an equine in vivo study suggested that changes in synovial fluid induced by joint loading influenced chondrocytes to increase proteoglycan synthesis and reduce proteoglycan breakdown [28].
1.4 Contact area and force measurement techniques
The determination of physiologically appropriate loads on cartilage has been hampered by the difficulty in directly measuring contact stress and forces acting across joints \textit{in vivo}. Therefore, several different approaches have been used to assess the stresses and forces present at the tibio-femoral joint. Most studies can be divided into types, one that is phenomenological and another that is anatomically based. Phenomenological models are at a gross scale, interested mainly in the overall response of the joint. In contrast, anatomically based models are created to study the various components of the joint by including accurate geometries and material properties [1].

The approaches can also be separated based on whether they address kinematic or kinetic aspects. Kinematic models analyze only the motion of the joint, and often solve for contact properties as a function of time. In contrast, kinetic models relate the motion of the joint to the loading parameters.

1.4.1 Kinematic Models
The analysis of the contact areas within the joint can be useful in testing for correlations between contact stress distributions and the development of degenerative changes in the cartilage. Additionally, such analysis can help to understand how existing or proposed surgical procedures may affect the performance and long-term health of the joint.

Ihn et al. [12] studied the differences in the contact area and pressure distribution for an intact human knee, compared to those with a partial and complete meniscectomy. The knees from five amputees were obtained and the skin and
subcutaneous tissues were removed, leaving the joint capsule intact. The knee was then mounted in full extension between two jigs attached to an axial load transmitting machine. Pressure sensitive film was inserted between the tibia and femoral condyle. Loading ranged from 30kg to 300kg which was thought to be the physiological range of daily activity [12].

The results showed that the contact pressure was evenly distributed over the intact medial contact area, and the lateral contact area had only a small (finger-print size) stress concentration. Additionally, in joints with an intact meniscus the medial contact area was always larger than the lateral by an average ratio of 1.36:1. This difference between contact areas was less for knees with either partial or complete meniscectomy. In particular, partial and total meniscectomy both decreased the overall contact area and increased the size of large stress concentrations.

Freeman and Pinskerova [5] reviewed the movement of the contact area over the motion of the human knee. In the context of this review, and unlike the previous study, ‘contact area’ was used to describe the area of direct tibiofemoral contact, excluding meniscal contact. Therefore, if the menisci carry load, this contact area would be smaller than the total load-bearing area. The numerous studies they reviewed showed certain discrepancies in the movement of the contact area with flexion. For instance, laterally, between 30 and 60 degrees of flexion, the contact area has been reported to move anywhere from 1mm anteriorly to 15mm posteriorly. Freeman offers a potential explanation for the discrepancies in the lateral measurements by suggesting that from 0 to 10 degrees posterior rolling of the condyle is probably obligatory; from 10 to 90 degrees the amount of rolling is variable; and afterwards the movement is
again obligatory. Meanwhile, medial condyle movement remains paradoxical; the contact area shifts posteriorly 2-8mm between 0 and 30 degrees and then remains stationary, while the condyle shifts only 1.5mm between 0 and 120 degrees of flexion. The explanation may lie in the shape of the femur and tibia in the range from full extension to 30 degrees flexion. The authors go on to explain that in extension there is a gap between the tibia and femur posteriorly, but in flexion the gap is anterior, thereby giving the appearance of shifting and then stationary contact area even though the condyle continues to move backward throughout the motion.

Contrastingly, non-invasive techniques, particularly the use of MRI, have also been utilized in order to assess the contact areas [20]. The inter-observer and intra-observer coefficients of variation were fairly low for the estimation of both the medial and lateral contact area, indicating that the assessment of contact area through the use of MRI is a fairly precise method. In normal patients, the size of the medial contact area decreased slightly between full extension and 60 degrees flexion, and the centroid moved posteriorly 8mm. The size of the lateral compartment contact area did not change significantly over the motion, but the centroid did move posteriorly 5.2mm [20]. Again it was found that the medial contact area was larger than the lateral [29]. Furthermore the contact areas were shown to shift to the posterior portion of the femoral condyles during knee flexion.

1.4.2 Kinetic Models

Kinetic models can be subdivided into quasi-static or dynamic models. The former analyzes the joint at a series of flexion angles by solving equilibrium equations. The results are then compiled in order to provide a response over the full
range of motion. However, such models do not account for the dynamic inertial loads of the joint. The dynamic models, on the other hand, solve the differential equations of motion in order to obtain the forces and parameters of motion [1].

One such quasi-static, kinetic model is presented by Thambyah et al. [26]. In this study, gait analysis data was combined with the lines of action and moment arms of the major structures in a human knee, established in a previous study [11]. The model included a closed quasi-static system of external forces and moments balance with internal stabilizing structures while ignoring muscle co-contractions. The anatomical orientation of the force bearing structures were simplified to lie only in the sagittal plane and to consist of only two lines of action, namely the patellar ligament and hamstrings tendon.

The results from this study report that there are three distinct peaks in the compressive force curve (Figure 1-1A). The first peak occurs soon after the heel strike and was approximately three times body weight (BW). The second peak occurred approximately at the onset of the single limb stance phase with a magnitude about 1BW.

Finally the third peak, which occurred just prior to the toe-liftoff, reached about 3BW. Additionally, the shear force has both a posteriorly-directed and anteriorly-directed component for a portion of the gait cycle (Figure 1-1B). The initial posteriorly-directed shear force is consistent with the necessary reaction inside the joint to prevent excessive sliding of the femur over the tibia. While the method used in this study could be suitable for analyzing large numbers of subjects, it introduces
numerous simplifications that prevent it from achieving high accuracy for an individual patient.

Dynamic simulations required numerically solving the complicated equations inherent in a 3D dynamic model. In one study, the joint was simulated by modeling the femoral condyles as two rigid partial spheres, the tibial plateau as a rigid plane and the ligaments as linear-elastic springs, but the menisci were not included [1]. Through a series of rigorous transformations, the solutions under various dynamic impulses were found. Although there is little experimental data to which to compare the results of the model, the contact points of the femoral condyle both move posteriorly as in previous kinematic models [5, 20, 29]. As the menisci were not present in this model, no simulations with axial loading could be presented, thus severely limiting the number of experimental studies to which it could be compared.

Another model goes one step further in that they represent the articulating surface as deformable [2]. This model compared an anatomically accurate knee, based on MRI and CT data, as well as an artificial knee based upon the manufacturer’s specifications. The model was stable for both small and large strain contact models with linear or nonlinear material properties. Static contact pressures predicted in the artificial knee case were compared with experimental measurements under various loads; the linear model was able to match both the peak and average contact pressures. Additionally, the natural knee contact model can predict average pressures, contact areas, and contact forces within 10% when compared to pressure data obtained in other studies. However, the model is limited in that it does not have menisci, nor does it represent any muscles, ligaments or external forces.
Other groups have developed different methods in which to study and observe the forces seen within the knee. D’Lima et al. were interested in tibiofemoral forces as they pertained to total knee arthroplasty [3]. In their study they implanted an instrumented tibial prosthesis into a patient. The prosthesis then was used to measure postoperative forces occurring in the joint during activities such as standing, walking, climbing stairs and rising from a chair. While the geometry and material properties of the prosthesis are inherently different than a natural human knee, the results obtained can not only help develop better prosthesis, but also approximate the magnitudes of forces for a wide range of activities, which would be extremely laborious to model individually using numerical methods. Other models that were similar, found the force throughout the gait cycle to have three distinct peaks (Figure 1-2). The shapes of the curves were found to be quite different than other model predictions, but could be accounted for by the differing shape of the prosthesis from a natural joint. The magnitude of the peak forces exhibited in walking, at the sixth postoperative week, reached a maximum of approximately 2.5 BW.

Another group developed a robotic system in order to conduct biomechanical tests on knee joints ex vivo [6]. The premise was to devise a system which allows for a hybrid control of displacement of the knee as well as the force/moment applied to it without restricting the natural motion of the joint. Unlike many contact surface studies, which only load the joint axially, this system offers much more complex and physiologically simulating loading patterns. Since the system stabilized under a combined compressive and cyclic anterior load after a few cycles of preconditioning, the system was determined to be effective. However, the position control was too slow
to simulate physiological situations as well as the position control could not be perfectly controlled when the load difference was greater than 5 N or 0.2 Nm in moment. Although this system has limitations, it could provide a valuable platform for future biomechanical studies without relying so heavily on mathematical assumptions.

1.4.3 Pressure Measuring Devices
Many *ex vivo* studies of knee joints utilize pressure sensitive films. However, the effect of such films on the contact mechanics, the sensitivity, accuracy as well as the limitations of the film need to be considered. The validity of the contact area measurements and the pressure magnitudes obtained from the film were studied by Liau et al [15]. They compared an artificial joint and a finite model and found that the measurement of contact area in the artificial joint by using Fuji film is overestimated its actual contact area by 1.2–2.4% while the measurement of contact pressure underestimated its actual contact pressure by 8–14%. Another study reported that the Fuji film underestimated the contact area by as much as 36% [10]. Many of the discrepancies between the Fuji film and experimental models were attributed to the perturbation caused by insertion of the film itself. It was suggested that for a natural joint, the film was stiffer than the articular cartilage around it, concentrating the stress and leading to an overestimation of the actual contact pressure. The reverse would occur in an artificial joint, where the film was softer than to the surrounding metallic components, thereby shielding the stress underestimating the contact pressure [15]. Related to this, the contact pressure changes significantly when the film is inserted into an articular joint due to the finite thickness of the film (0.3mm) [15].
Additionally, pressure sensitive film cannot measure dynamic loads, as the resulting pattern would only reflect the maximum historical value of the local load. Pressure sensitive films also are hampered by the limited range of each film type, creating problems with either saturation when forces greater than the maximum value of the film is exceeded, or by sensitivity difficulties when the film does not record any pressures below the lower threshold of the film [10, 15].

Digital sensors have also been used in joint research studies. They function because the application of pressure results in a change in the resistance of this sensing unit in inverse proportion to the applied pressure [7]. Inherent benefits from such a system include its ability to measure dynamic loads and be reused numerous times, depending on loading conditions [10].

The functionality and use of Fuji film was compared to the digital K-scan system (Tekscan Inc., South Boston, MA, USA), and it was found that the later enabled an “easy, reproducible measurement of contact area at different loads and flexion angles” [10]. Moreover the K-scan system also had a smaller standard deviation than that observed in the corresponding Fuji film tests. Another study found that Tekscan sensors were on average 2.5 times more accurate than Fuji film [7]. One possible draw back to the digital sensors in general, however, is the possible “crimping” of the sensor around dual axes with very high radii of curvature. Crimping could lead to large pressure spikes in the output.

1.5 Aims of Thesis
The overall aims of this were work were to (1) assess the ex vivo distribution and magnitude of contact stress within an immature bovine stifle joint, (2) examine the
effects of axial and patellar loading on the stress distribution and (3) determine if there is a correlation between the force within the joint and the cartilage biochemical response.
Figure 1-1. (A) Average tibiofemoral bone-on-bone contact forces (compressive) and (B) average shear forces. Both are plotted against percent gait cycle. One standard deviation is shown for each of the averaged curves. Adapted from [26].
Figure 1-2. Tibial forces as a function of the percent gait cycle averaged over six representative steps. No significant differences in mean or peak tibial forces could be detected between values recorded with or without shoes. Adapted from [3].
Chapter 2: Experimental Study

2.1 Introduction

A whole joint bioreactor was recently described by Nugent-Derfus et al as an \textit{ex vivo} apparatus for studying the biomechanical regulation of cartilage metabolism in bovine stifle joints [17]. Stimulation of chondrocyte lubricant biosynthesis by continuous passive motion (CPM) of stifle joints was hypothesized to be consistent with cartilage responses to compressive and shear stresses [18]. In the whole joint bioreactor setup, compressive stress in the passive state of different degrees of flexion is unknown.

The effect of externally applied loads on the contact stress within test joints in the bioreactor is also unknown. \textit{In vivo}, the joint experiences a wide range of loading patterns. Experiments conducted on cadaveric joints often employ an axial load [12]. The Oxford knee rig has also been established as a reliable model for human knee studies, as it includes the six degrees of freedom necessary in knee flexion [31]. Versions of this model have been used to passively measure the patellar load during flexion [4]. This study applied two different types of external loads on a stifle joint fixed on a modified continuous passive motion (CPM) machine: axial, via pneumatic actuators aligned with the axis of the tibia and femur, and patellar, by applying a force to the patella.

Direct determination of the contact stress in joints \textit{in vivo} has also proven challenging. Thus, studies have attempted to estimate these forces through mathematical models [1, 2, 26]. Additionally, a real time \textit{in vivo} measurement of loading within a knee replacement [3] has been successful technically, but does not
necessarily provide an indication of the biomechanics of an intact human knee. *Ex vivo* measurements of the contact stress have been conducted using pressure sensitive film [10, 12] or digital sensors[10] in cadaveric joints. In this study, a digital sensor was pulled through the static joint at a constant rate while recording the load along its path.

The objectives of the present study were to (1) assess the *ex vivo* distribution and magnitude of contact stress within an immature bovine stifle joint at different degrees of flexion, (2) examine the effects of axial and patellar loading on the stress distribution at selected flexion angles, and (3) determine if there is a correlation between the contact stress distribution within the joint and the cartilage biochemical response observed in previous studies.

### 2.2 Experimental Methods

#### 2.2.1 Joint Isolation

Immature (1-3 weeks) bovine stifle joints were harvested and prepared for experimentation essentially as described previously [17], except with maintenance of the patella and patellar tendon for some of the samples. Briefly, joints were obtained from an abattoir. The soft tissues surrounding the joint were removed, and the femur and tibia were cut with a bone saw such that the total length of the extended joint was approximately 25 cm. The bone marrow was removed with a curette and the internal shafts were washed with hydrogen peroxide. The joint tissues remaining were the proximal femur and distal tibia bones, articular cartilage, menisci, cruciate and collateral ligaments. For patellar loading experiments, the patella and patellar tendon were also maintained.
2.2.2 **Contact pressure measurements**

A digital sensor (Flexiforce B201, Tekscan, South Boston, MA) was modified to allow measurement of contact stress within the joint, and conditioned and calibrated before use. The sensor was glued with silicone adhesive (RTV 108, GE Silicones, Waterford, NY) onto one long strip of plastic (0.002” thick polyester, Artus Corp., Englewood, NJ) as a backing material, and another, similar in thickness to the sensor (0.0075”) and cut to be contiguous with the sensor, was glued next to the sensor, and lastly topped with a third long strip of plastic (0.0015”) (Figure 2-1). The overall length of the modified sensor was 0.35m (length), 14mm (width), with an overall thickness, including the silicone, of 0.6mm. The sensor was conditioned by applying and removing 20 lbs (90N) to the sensing area (9.5mm diameter, 70 mm² area) 10 times which was sufficient to precondition the sensor before use, per the manufacturer’s instructions. The sensor was then calibrated with at least ten known weights (0-25 N) achieving a linear fit ($R^2>0.992$) in the force and stress (0-600 kPa) range expected for the test. Control experiments in the knee joint preparation with an additional backing of 0.25-0.75mm did not significantly change the measurements within the joints, indicating that the geometry of the modified sensor had a negligible effect on measurements, consistent with the relatively thick layers (~10-20mm total) of articular cartilage that is soft compared to the sensor material.

Each condyle was analyzed for contact stress in three anatomical aspects. Relative to an anterior-posterior centerline along the maximum curvature of the condyle; the first was defined as the area lateral to the centerline, the second was the area medial to the centerline, and the third aspect was on the centerline (Figure 2-
2.A,B). In this way, three aspects were defined on each condyle (M1, M2, M3 on the medial condyle and L1, L2, L3 on the lateral condyle). Additionally, anterior and posterior points of reference were marked with a pen. The anterior point was taken to be 9.5mm proximal to the menisci at 0 degrees of flexion. The posterior point was taken to be the border of the articular cartilage of the proximal region of the condyle (Figure 2-2.C). At the conclusion of the tests below, the joint was disarticulated and the length along the maximum curvature, between the anterior and posterior points, was measured on each condyle.

The knee joint was secured into a specialized instrument to allow application of flexion, essentially as described previously [17]. The cut ends of the bones were secured into custom designed polysulfone bone clamps and attached to an adapted CPM device (FlexMate K500, BREG, Vista, CA). The joint flexion angle on the CPM was adjusted to that desired, as measured with a goniometer, within the instrument range of -10 to 120 degrees of flexion. Measurements were conducted at flexion angles ranging from 0 to 45 degrees.

Data was acquired while pulling the sensor along the surface of the femoral condyle. The sensor was inserted into the joint anteriorly, proximal to the meniscus, on the medial side and restrained laterally at one of the aspects of interest (e.g. M1) by medial-lateral guides. The sensor was attached on the posterior end to a timing belt via poly ribbon and on the anterior end with a clamp. Starting at a fixed position indexed to the anterior region, the sensor was pulled through the joint at a fixed rate by a motor (NSH-12R, Bodine Electric) with sampling at 960 Hz. Each measurement was repeated two additional times, and the data from the three trials were averaged. The
test protocol was repeated for each of the remaining aspects of the medial and lateral femoral condyles (M2, M3, L1, L2 and L3). Load measurements were normalized to sensor area to estimate stress.

2.2.3 Variation in Intrinsic Passive Loading with Flexion Angle

The effect of flexion angle on contact stress was assessed by making measurements in knees (n=3) positioned passively at 0, 15, 30 and 45 degrees of flexion (Table 2-1).

2.2.4 Effect of Active Loading Applied Axially to the Femur and Tibia

The effect of axial loading on contact stress was assessed by making measurements in knees (n=3) positioned at 0 and 45 degrees of flexion, with applied axial loads of 0, 45, 90 and 135N (Figure 2-3). For application of load, pneumatic actuators at the interface between the CPM frame and the bone clamp were used, with adjustment of a pressure regulator (Parker, Cleveland, OH) to achieve the desired load, as measured with in-line a 2000lb (8900 N) capacity load cell (Honeywell, Morristown, NJ).

2.2.5 Effect of Active Loading Applied to the Patella

The effect of loading at the patella, simulating quadriceps action, was assessed by making measurements in knees positioned at 0 and 45 degrees of flexion, with applied load magnitudes of 0, 14, 24, 45N (Figure 2-4). For application of load, two stainless steel screws (M6) were set ~5mm deep into the cortical bone of opposing sides of the anterolateral patella with their axes at a ~90 degree angle. Four attachments of polypropylene ribbon were secured to each screw, with the two lateral ribbons secured to provide stability and the two medial ribbons attaching to a free
weight pulley system, with the ribbons from the two screws converging at a single free weight. The line of action of the patellar load was directed to be similar to the natural direction from the quadriceps insertion point to that at the greater trochanter, based on measurements taken at the time of joint dissection. In these studies, data was not collected in the medial aspect of the lateral condyle (L2) because of obstruction by the patellar tendon.

2.2.6 Data Reduction
Acquired data was reduced to rectangular maps, representing the data from all aspects of each condyle. Data sets were mapped back to their physiological position (M1, M2, M3, L1, L2, L3) and each data set was reduced by averaging to a set of 1000 points, distributed uniformly from the anterior to posterior borders of the joint. Outliers evident as data points that spiked to saturation level in a fraction of a second, were replaced by linearly interpolated values from data sets as necessary (<1.5% of total data). For each joint, the three repeated data sets for each loading condition and aspect were averaged.

A 2D spatial map of stress for each sample and an average of all samples within an experimental group was then created. Data analysis was done using Matlab 7.0 (Mathworks, Natick, MA) to automatically interpolate between the three reduced data sets and mapped onto the entire joint surface, with an arbitrary color indicating the level of normal stress. Spatial maps were created for each condyle for each angle and loading condition.

For quantitative and statistical analyses, the data sets were further reduced into six anterior – posterior (A-P) regions for each aspect relative to the margins of the
articular surface. Thus, 18 points per condyle were generated, each representing the average value around that area (Figure 2-5). The location of peak stresses on the condyle was determined for intrinsic loading conditions. Additionally the contact stresses were converted to average forces for each of the 18 locations by multiplying the average stress by the area of the region. The forces in these locations were then summed for each condyle in order to estimate the net force on the MFC and on the LFC, and then overall for the net force on the entire joint. (See Appendix C for more detailed description).

2.2.7 Statistical Analysis
Statistical analyses were performed using Systat 10.2 software (Systat Software, San Jose, CA). Data are given as mean ± SEM.

2.2.7.1 Effects on Net Forces
For passively loaded joints, the effects of flexion angle on net force overall, as well as on net force on each condyle (LFC and MFC), were determined by repeated measures ANOVA (flexion angle as repeated factor) and planned comparisons between load (overall and for each condyle) at each flexion angle with Bonferroni correction comparisons.

Similarly, for each of the externally loaded configurations, i.e., axial and patellar, the effects of axial load amplitude and flexion angle on net force overall, as well as on net force on each condyle (LFC and MFC), were determined by repeated measures ANOVA (flexion angle of 0° or 45° and external load amplitude). The effect of external load amplitude at each flexion angle was assessed by linear regression.
2.2.7.2 Effects on Regional Stress and Peak Stress Position

To directly compare external loading methods, the effects on regional stress levels of no load, 45N external axial load, and 45N external patellar load were compared at 0° and 45° flexion. The stress averaged over each of the five most anterior of the six defined A-P regions in each aspect (regions 2-6, aspects 1-3, 15 in total) were analyzed, since the most posterior region was not in contact at the experimental flexion angles. The effects of load (intrinsic, axial, patellar) and flexion angle (0° and 45°) on regional stress were assessed independently by a repeated measures ANOVA (A-P region and aspect as repeated factors). For the case of patellar loading, the most medial aspect of the lateral condyle (L2) was obstructed by the patellar tendon so that only ten regions could be compared. The effect of load type was determined by repeated measures ANOVA (A-P region and aspect as repeated factors) on the passive intrinsic load and external loading by axial and patellar mechanisms at 45N.

The effect of load amplitude – intrinsic, axial and patellar – on peak stress location was assessed by linear regression at each flexion angle for each condyle.

2.2.7.3 Relationship of Contact Stress to Previously Determined Biosynthetic Response

The contact stress results were used to compare the pattern of stress amplitude with the patterns of PRG4 biosynthesis observed previously [17]. There, for flexion between 10° and 46°, the region described as continuously stimulated was that extending from the border of regions 5 and 6 (anteriorly) to the ~border of regions 4 and 3 (posteriorly), where the contact point when the joint in flexion was at 45 degrees marked the beginning of the continuously stimulated region described in the previous study [17]. The remaining area was then divided roughly into thirds to create the other
two duty regions (Figure 2-19). The region of cartilage that was never stimulated was slightly smaller than the other two regions. The A-P regions were grouped into different groups depending on their respective duty cycles. Within each duty cycle both the force and PRG4 secretion data were averaged and reduced so that there were only three data points per duty cycle (Figure 2-19).

The effect of duty cycle on average forces was assessed by repeated measures ANOVA (duty cycle and aspect as repeated factors) and planned comparisons with Bonferroni correction comparisons. Univariate regressions were determined for the average contact stresses at 10 degrees of flexion to static PRG4 secretion rates (-CPM) as well as the time average contact stresses between 10 and 46 degrees to dynamic PRG4 secretion (+CPM) and the ratio of PRG4 secretion (+CPM/-CPM).
Table 2-1. Summary of experiments. Shaded boxes represent data sets that exclude L2, the most medial aspect of the lateral condyle.

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Figure 2-1. Flexiforce sensor A201 (A), schematic of plastic strips adhered to sensor (B), with silicone adhesive in between each layer, and sensor after modifications (C).
Figure 2-2. Schematic of layout of aspects (A) on lateral and medial femoral condyles (LFC and MFC), aspects marked on specimen (B), and anterior and posterior reference points on MFC (C).
Figure 2-3. Axial Loading schematic at 0 degrees of flexion (A), at 45 degrees of flexion (B), anterior view of axial loading (C), and lateral view of loading at 45 degrees of flexion (D).
Figure 2-4. Patellar loading schematic at 0 degrees of flexion (A), at 45 degrees of flexion (B), anterior view of patellar loading (C), and lateral view of loading at 45 degrees of flexion (D).
Figure 2-5. Sample of spatial map divided into regions with 18 points representing the average values for that region (six per aspect) used for statistical analysis.
2.3 Results

2.3.1 Effects on Net Forces

2.3.1.1 Effect of Angle

The forces at the joint surface in the passive state of flexion depended markedly on the flexion angle. The net force on the cartilage surfaces varied significantly with flexion angle (p<0.001, Figure 2-6.A). A Bonferroni corrected post-hoc comparison illustrated this significance between 15 degrees and 30 degrees (p<0.05) and 45 degrees (p< 0.01). Analysis of forces on each condyle confirmed the dependence within samples on angle overall (p<0.01, Figure 2-6.B), with a trend toward a higher force in the LFC than the MFC (p=0.08) and also interactive effects of condyle and angle with the LFC affected more than the MFC (p=0.08).

2.3.1.2 Effect of Axial Load

With applied axial load at 0º and 45º flexion, the forces at the joint surface depended markedly on the external load magnitude (Figure 2-7). ANOVA indicated that the net force on the cartilage surfaces varied significantly with axial load (p<0.05), without independent (p=0.12) or interactive (p=0.82) effects of flexion angle. Univariate regression indicated a significant a positive correlation of axial load and net force on the cartilage surface at 45 degrees ($r^2=0.99$, p<0.005, Figure 2-7) but not at 0 degrees ($r^2=0.74$, p=0.20, Figure 2-7). The average coefficient of variance (CV) was 67% at 0 degrees and 36% at 45 degrees.

Analysis of forces on each condyle (Figure 2-8) confirmed the dependence on axial load (p<0.01) with interactive effects of condyle (MFC or LFC, p<0.05). Univariate regression indicated a significant a positive correlation of axial load and net
force on the cartilage surface on both the MFC and LFC at 45 degrees (r²=0.90, p<0.01, and r²=0.97, p<0.05, respectively) but not on either the MFC (r²=0.52, p=0.71) or LFC (r²=0.94, p=0.15) at 0 degrees. The average CV at 0 degrees was 48% and 84% for the MFC and LFC respectively, and at 45 degrees was 61% and 73% for the MFC and LFC respectively.

2.3.1.3 Effect of Patellar Load

The net force at the joint surfaces under applied patellar load depended markedly on the external load magnitude (p<0.001) but not flexion angle (p=0.09) with an interactive effect (p<0.01, Figure 2-9). At 0° flexion, univariate regression indicated a significant positive correlation of net force and patellar load (r²=0.90, p<0.001). However, at 45° flexion, there was no detectable change in force with load (r²=0.97, p=0.14). The average coefficient of variance (CV) was 9% at 0 degrees and 31% at 45 degrees.

Analysis of forces on each condyle (Figure 2-10) confirmed the dependence on patellar load (p<0.001) with significant interaction between load and condyle (MFC or LFC, p<0.005). Univariate regression indicated a significant a positive correlation of axial load and net force on the cartilage surface on the LFC at 0 degrees (r²=0.99, p<0.001, Figure 2-10) but not on the MFC at 0 degrees (r²=0.48, p=0.10) or either the LFC (r²=0.39, p=0.14) or the MFC (r²=0.70, p=0.54) at 45 degrees. The average CV at 0 degrees was 23% and 17% for the MFC and LFC respectively, and at 45 degrees was 69% and 31% for the MFC and LFC respectively.
2.3.2 **Effects on Regional Stress and Peak Stress Position**

2.3.2.1 **Effect of Intrinsic Passive Loading**

The regional stress over the joint surface under intrinsic loading varied markedly with flexion angle (p<0.001) and condyle (p<0.01) with significant interaction (p<0.05), as well as with A-P region (p<0.001) and an interactive effect of aspect and condyle (Figures 2-11, 2-12). Contact stress differed significantly between region 2 and regions 3 and 4 (p<0.001) as well as between region 6 and regions 3, 4 and 5 (p<0.001, p<0.001, p<0.01 respectively).

Furthermore, the location of peak stress shifts significantly with increasing flexion angle towards the posterior of the LFC (p<0.001), but not on the MFC (p=0.06, Figure 2-13). A post hoc analysis on the LFC peak stress location reveals that at 0 and 45 degrees they are significantly different (p<0.05) with a trend towards varying significantly at 15 and 30 degrees of flexion (p=0.055, Figure 2-13).

2.3.2.2 **Effect of Axial Load**

Under applied axial loading, the regional stress over the joint surface varied significantly load magnitude (p<0.001) and flexion angle (p<0.001) with a tendency for an effect of condyle (p=0.07, Figure 2-14), with marked interactive effects between condyle and region (p<0.001), condyle and angle (p<0.005), aspect and region (p<0.001), as well as aspect, region, and other factors (each, p<0.001).

Under applied axial loading the location of peak stress does not shift significantly with increasing flexion angle on either the MFC (p=0.49) or the LFC (p=0.86, Figure 2-15), nor is there any significant shift with increasing axial load for the MFC or LFC (p=0.89, 0.88, respectively).
2.3.2.3  **Effect of Patellar Load**

Under applied patellar loading, the regional stress over the joint surface varied significantly with load magnitude ($p<0.01$), condyle ($p<0.001$) and A-P region ($p<0.001$, Figure 2-16), with a trend for an independent effects of angle ($p=0.06$) and interactive effects of between condyle and region ($p<0.001$), condyle and angle ($p<0.001$), aspect and region ($p<0.01$), as well as aspect, region, and other factors (each, $p<0.001$). Contact stress differed significantly between region 2 and regions 4 and 6 ($p<0.001$, $p<0.05$ respectively) as well as region 4 and region 5 ($p<0.001$) and region 6 and regions 3, 4 and 5 ($p<0.001$).

Under applied patellar loading the location of peak stress shifts significantly with increasing flexion angle on the LFC ($p<0.005$) but not the MFC ($p=0.15$, Figure 2-17), nor is there any significant shift with increasing patellar load for the MFC or LFC ($p=0.82$, 0.16, respectively).

2.3.2.4  **Effect of Load Type**

Comparison of contact stress pattern between the states of intrinsic passive loading, 45N axial load, and 45N patellar load indicated marked variations in amplitude and position. Regional stress varied significantly with flexion angle ($p<0.01$), condyle ($p<0.005$), and A-P region ($p<0.001$) (Figure 2-18), with significant interaction between condyle and angle ($p<0.01$). Contact stress differed significantly between region 2 and region 3 ($p<0.005$) as well as between region 6 and regions 3, 4 and 5 ($p<0.001$, $p<0.001$, $p<0.01$ respectively). There were no significant differences in regional contact stress distribution with regards to the type of loading protocol ($p=0.51$) at 45N.
2.3.3 **Relationship of Contact Stress to Previously Observed Biosynthetic Response**

The average regional forces at 10 degrees varied significantly with the three artificially superimposed duty cycles (p<0.001) with significant interaction with angle (p<0.001) and condyle (p<0.05). A corrected post hoc analysis showed a significant difference between the continuous and intermittent (p<0.05) and never (p<0.001) duty cycles as well as between intermittent and never regions (p<0.001, Figure 2-19).

Average regional contact stresses on the MFC at 10 degrees of flexion correlated significantly with the static (-CPM) PRG4 secretion. Univariate regression indicated a significant positive correlation of PRG4 secretion with force on the cartilage surface on the MFC ($r^2=0.88$, p<0.001) but not on the LFC ($r^2=0.22$, p=0.21,) or when analyzed together ($r^2=0.06$, p=0.33, Figure 2-20).

The time average regional contact stresses between 10 and 45 degrees of flexion correlated significantly, as determined with a univariate regression, with the ratio of PRG4 secretion levels in stimulated (+CPM) and static (-CPM) culture systems, ($r^2=0.41$, p<0.01, Figure 2-21.B), but did not correlate significantly directly with the PRG4 secretion levels in the stimulated culture system ($r^2=0.08$, p=0.28, Figure 2-21.A)
Figure 2-6. Net surface force on (A) entire joint surface (■) and (B) medial (●) and lateral (■) femoral condyles, solid (−) and dashed (−−) respectively, at discrete angles in a joint under passive, intrinsic loading. Mean ± SEM, (n=3).
Figure 2-7. Total surface force measured in instrumented knee joints at 0 and 45 degrees flexion under varying axial loads. Data are represented as 0° (●) and 45° (■) with linear regression lines dashed (---) and solid (--) respectively. Data are shown as mean ± SEM, (n=3).
Figure 2-8. Net surface force on medial and lateral femoral condyles measured in instrumented knee joints at 0 and 45 degrees flexion under varying axial loads. Data are represented as medial (○,●) and lateral (□,■) and 0° (○,□) and 45° (●,■) respectively. Linear regression lines are shown. Data are shown as mean ± SEM, (n=3).
Figure 2-9. Total surface force measured in instrumented knee joints at 0 and 45 degrees flexion under varying patellar loads. Data are represented as 0° (●) and 45° (■) respectively with linear regression lines dashed (- -) and solid (–) respectively. Data are shown as mean ± SEM, (n=3).
Figure 2-10. Net surface force on medial and lateral femoral condyles measured in instrumented knee joints at 0 and 45 degrees flexion under varying patellar loads. Data are represented as medial (○,●) and lateral (□,■) and 0° (○,□) and 45° (●,■) respectively. Linear regression lines are shown. Data are shown as mean ± SEM, (n=3).
Figure 2-11. Average spatial distribution of contact stress on medial and lateral condyles (MFC, LFC) at discrete angles in a joint under passive, intrinsic loading represented as a color scale contour mapping onto the joint regions, (n=3, from experiment #1).
Figure 2-12. Average spatial distribution of contact stress on medial and lateral condyles (MFC, LFC) at 0 and 45 degrees in a joint under passive, intrinsic loading represented as a color scale contour mapping onto the joint regions, (n=6, all from experiment #2).
Figure 2-13. Shift of peak pressure anterior-posterior (A-P) position on medial (●) and lateral (■) femoral condyles, solid (−) and dashed (− −) respectively, at discrete angles in a joint under passive, intrinsic loading. Mean ± SEM, (n=3).
Figure 2-14. Average spatial distribution of contact stress on lateral and medial femoral condyles (LFC, MFC) at 0 and 45 degrees in a joint under varying axial loads (0, 45, 90, 135 N) represented as a color scale contour mapping onto the joint regions, (n=3).
Figure 2-15. Shift of peak pressure anterior-posterior (A-P) position in a joint under external axial loading on the medial (○,●) and lateral (□,■) femoral condyles and 0° (○,□) and 45° (●,■) respectively. Mean ± SEM, (n=3).
Figure 2-16. Average spatial distribution of contact stress on medial and lateral condyles (MFC, LFC) at 0 and 45 degrees in a joint under varying patellar loads (0, 14, 24, 45 N) represented as a color scale contour mapping onto the joint regions, (n=3). Data was not recorded for the most medial aspect of the lateral condyle, as it was obstructed by the patellar tendon.
Figure 2-17. Shift of peak pressure anterior-posterior (A-P) position in a joint under external patellar loading on the medial (○,●) and lateral (□,■) femoral condyles and 0° (○,□) and 45° (●,■) respectively. Mean ± SEM, (n=3).
Figure 2-18. Comparison of average spatial distribution of contact stress on lateral and medial femoral condyles (LFC, MFC) at 0 and 45 degrees in a joint under intrinsic and external loads (axial and patellar, 45N) represented as a color scale contour mapping onto the joint regions. (n=3). Data was not recorded for the most medial aspect of the lateral condyle, as it was obstructed by the patellar tendon.
Figure 2-19. Lateral and medial femoral condyles were divided into sub-regions that continuously (con), intermittently (int), or never (nev) slide against opposing tissues during CPM stimulation. PRG4 secretion by cartilage disks from freshly harvested (A) or bioreactor cultured (with or without CPM stimulation) (B) joints, represented as a logarithmic color scale contour mapping onto the joint regions. Composite spatial maps of tibiofemoral contact stress represented as a color scale contour mapping onto the joint regions at 10 degrees of flexion (C) and time average of stress from 10-46 degrees of flexion (D). (A) and (B) adapted from [17].
Figure 2-20. Regional correlation of static PRG4 secretion and contact stress at 10 degrees when medial and lateral femoral condyles (MFC, (●, – - -), and LFC, (■, – – ), respectively) are analyzed separately and together (—).
Figure 2-21. Regional correlation of net contact stress between 10-45 degrees and (A) PRG4 secretion levels after dynamic simulation (+CPM), and (B) ratio of dynamic (+CPM) and static (-CPM) PRG4 secretion levels. Data and regression lines are marked as medial and lateral femoral condyles (MFC, ●, – – –), and LFC, (■, – – ), respectively and both together (––).
2.4 Discussion

The tibiofemoral force and the distribution of contact stress was determined under intrinsic conditions in addition to several magnitudes of external loading either axial or via the patellar tendon. The resulting data showed a significant effect of angle, aspect, and load magnitude, which corresponded to similar findings in the literature of other types of joints, most frequently human. The alteration of contact stress magnitude and location by both axial and patellar loading provide a way to test the role of such stress in a whole joint bioreactor. Such an approach could also be used to study effects of mechanical loading of knee joints with repair procedures, such as implanted tissue grafts, under controlled conditions.

It is important to note that while the information ascertained from this study is significant in regards to adding insight to parameters of a novel articular joint model, claims should not be overstated. In particular, the study data should not be used to characterize contact stress properties of the human knee. Therefore, this experimental protocol must be limited to discussion in bovine, and possibly all pastural animals, or to be discussed as a potential platform for the maturation of tissue engineering constructs. Additionally, factor of animal was significant in some cases, therefore is a possibility that other trends may not be fully determined with the current sample size. Furthermore, as in any study design, inherent limitations in the data collection and reduction process were present. Slight phase shifts might have been introduced as the motor and data acquisition software were not automated, and thus the end points of the data acquisition could not be known with extreme precision (±1mm). Automation of
the system would help reduce any spatial variation due to phase shifts in the A-P direction.

The results from this study correspond with many findings in the human knee found in literature. The dependence of force on angle is consistent with literature obtained from previous studies of both human and animal knee biomechanics [1, 3, 24, 26], where the measured or calculated tibiofemoral contact force is a function of angle. The significant difference of peak stress position on the LFC with regards to angle corresponds to literature that tracks the shift of the centroid of the contact area as a function of angle [5, 20]. The inconsistent tracking in the centroid of the medial condyle also has been observed in human studies [5]. While there were no significant differences between medial and lateral condyle net forces, there were differences in the contact stress pattern, which correspond with previously established differences in compartment contact stresses demonstrated in human knee joints [1].

The force distribution patterns on the defined aspects were expected to vary significantly, which was one reason why several aspects were tested in the first place. The data, however, does not validate this assumption, but rather varies greatly. Some possible explanations for this are that the experimental procedure was not sufficient to keep the sensor in the same path within a joint, or that the anatomical reference points used to mark the aspects were not sufficient to observe the spatial variation in the medial to lateral direction. It could also be that the regions being compared are too large. While it was shown that the A-P region had sufficient divisions to observe these differences, perhaps more divisions were necessary to observe aspect differences. It is possible that there simply is not a significant aspect variation observable in our model.
The summed joint forces show a linear correlation with increasing load, most drastically at 45 degrees, which agrees with predicted values from a simplified free body diagram (Figures 2-7, 2-9) (See Appendix A). Thus the free body diagrams can then be used to estimate the entire joint force in future experiments.

Comparison between the intrinsic case and the axial and patellar loads under the same external loading did not show a difference between the different loading protocols. Although the magnitude of the load does play a factor, it does not appear that the method upon which the external loading is applied significantly effects the stress distribution. Thus, when applying external load to a whole joint bioreactor, it does not matter if it is applied via the patella or tibia and femur. However, the patellar loading had a lower coefficient of variance than the axial loading, and thus it could be used to generate a more targeted biochemical response.

The spatial maps of contact stress reveal several interesting characteristics that cannot be observed by numbers alone. More than just simply being a function of angle, the contact stress patterns are strikingly different. At 0 degrees the stress appears to be more widely distributed, and due the presence of distinct crescent shaped patterns in several joints, it appears that the bulk of the contact stress is focused on the menisci. At 45 degrees, though, the general trend is that the contact stress becomes more focused in the center of the condyle, and the distinct crescent shaped stress patterns are lacking. This suggests that the majority of the contact stress is then being borne by the contact of femoral articular cartilage on tibial articular cartilage.

It was of particular interest to compare already obtained biochemical response results [17] to the contact stress values recorded during this study. While many
relationships between biochemical response and biochemical response of cartilage have been performed on the cellular or explant level, comparison of contact stress to biochemical response on a whole joint level may prove very insightful. While individually the stress distribution and PRG4 secretion varied with duty cycle, the two did not correlate to each other over both condyle surfaces under static conditions (-CPM). The high contact stress appears to correspond to the low PRG4 secretion in the continuous region of the LFC, but there were still moderately high stresses in the analogous region of the MFC, but with a much higher PRG4 secretion than on the LFC. However, under dynamic conditions (+CPM), a significant correlation was present between the summed force on the cartilage surface and the ratio of dynamic to static secretion levels. The possible mechanisms would govern this relationship are still ambiguous. Overall, it remains unclear if the change in PRG4 secretion by the stimulus of CPM is due to the presence of dynamic stimulation rather than the absence of static loads, or perhaps something more complex.

As a whole, the experimental setup presented here has been able to simulate the changes in the magnitude and position of stress as a function of angle. It has gone further than other sources in literature in that it has given insight into further trends that might be apparent. Without background information into how the stresses inside an immature bovine stifle joint occur in vivo, it is impossible to determine how the removal of the soft tissues alters the stress distribution. As in vivo data is difficult to acquire, this study has shown that if the change in stress distribution could be known, the application of external loads could alter that distribution, perhaps in a manner that would make the in vitro setup more closely mimic the in vivo patterns.
This study can be built upon in several ways. A combined study with the biochemical response combined with external loading would be necessary to verify the trends that are roughly appearing in this study. Furthermore, the ultimate goal will be to attempt a more accurate physiological loading model. This may be done with special attempts to preserve soft and connective tissues of the joint and to include a more physiological dynamic stimulation of the joint with axial or patellar loading.
Appendix A: Free Body Diagram

A.1. Introduction

To provide a prediction of the effects of patellar and axial loading obtained experimentally were within reason, free body diagrams (FBD) were constructed. In the formulation of the FBD, several assumptions were made. (1) The problem was reduced to a two dimensional problem. (2) The tibia and femur were treated as rigid rectangular beams with constant structural properties. (3) A cartilage layer with constant properties was also included. (4) The CPM mount produces flexion of the femur and tibia about the axis of the knee. (5) The displacement in the y direction is not constrained at the tibia-femur interface and will not affect the contact force if the contact surface is assumed to be flat. This last assumption was made in this analysis, although it is not the case inside the joint, particularly when one considers the presence of menisci. However, due to the angle at which the patellar force acts, the vertical forces will be much smaller, and any effect of these vertical forces and on the curvature of the joint will have negligible contributions to the contact force when compared to the axial component.

A.2. Axial Loading

In the axial loading setup, the bone clamps are not restricted from motion in the axial direction of both the femur and tibia.

A.2.1. 0 Degrees of Flexion

At zero degrees of flexion the axial forces are aligned. Thus, the net force within the joint is equal to the force applied at the ends. Consequently the tibiofemoral force equals the applied force. Here, the cartilage and bones could compress axially until equilibrium is reached. (Figure A-1)

\[ \sum F_x = 0 \]  
\[ F_{FT} = F_{APP} \]  

Equation A-1  
Equation A-2

A.2.2. 45 Degrees of Flexion

At any flexion angle, \( \alpha \), this problem becomes a bit more involved, but the system is still determinate (Figure A-2.A). Summing forces in both the x and y-directions as a whole provides the reaction forces \( R_T \) and \( R_F \)

\[ \sum F_x = 0 \]  
\[ F_{APP} \cos \alpha - F_{APP} \cos \alpha + R_T \sin \alpha - R_F \sin \alpha = 0 \]  
\[ R_T = R_F \]  

Equation A-3  
Equation A-4

Assuming there is no y-directed force transmission at the joint, e.g. due to the low friction nature of cartilage,

\[ \sum F_y = 0 \]
\begin{align*}
F_{APP} \sin \alpha + F_{APP} \sin \alpha - R_T \cos \alpha - R_F \cos \alpha &= 0 & \text{Equation A-5} \\
R_T = F_{APP} \tan \alpha & & \text{Equation A-6}
\end{align*}

Looking at the tibia or femur individually, and knowing Ra, the tibial force can be determined (Figure A-2.B).
\[ \sum F_x = 0 \]  
\[ F_{APP} \cos \alpha + R_T \sin \alpha - F_{FT} = 0 \]  
\[ F_{FT} = F_{APP} \cos \alpha \left( 1 + \tan^2 \alpha \right) \]  
\[ \text{Equation A-7} \]
\[ \text{Equation A-8} \]

A plot of the relationship between the applied and calculated forces as a function of angle is presented in Figure A-3.

**A.3. Patellar Loading**

For the patellar loading system bone clamps that hold the joint in the CPM machine restrict motion in all directions within the two dimensions, thereby creating a support that could feel reaction forces in both the x and y directions as well as a moment.

**A.3.1. 0 Degrees of Flexion**

At zero degrees, the tibial contact force is assumed to act normal to the contact area and thus solely in the x-direction. While moment and vertical reaction forces may exist at the bone clamp-joint interface, it is not necessary to determine them in order to solve for the tibial force.

The external forces acting on the system only include $F_{APP}$, the force acting on the patella (Figure A-4.B). The problem can then be separated into two force balances, one in the x-direction and one in the y-direction (Figure A-4.C).

Analyzing Figure A-4.C allows us to divide the figure into several different segments as shown in Figure A-4.D. $L_T$ and $L_F$ are the lengths of the tibia and femur respectively, $L_{TI}$ is the distance from the tip of the tibia to the insertion point of the patellar tendon and $L_C$ is the combined thickness of articular cartilage in the joint.

At the point of external load, there will be a discontinuity in the axial load. Since the joint is over constrained, the overall joint cannot change length such that the amount that the external force causes the tibia to stretch will be compensated by compression in the cartilage and femur (Figure A-4.E). Mathematically, this can be summarized by
\[ \delta_T + \delta_F = \delta_{\text{total}} = 0 \]
where
\[ \delta = \frac{FL}{AE} \]
\[ \text{Equation A-9} \]
\[ \text{Equation A-10} \]

$F$= external force, $l$ is the length of upon which the force acts, $A$ is the cross sectional area and $E$ is the Young’s Modulus of the material. Thus, assuming a constant cross sectional area for the tibia and femur, the constraint equation becomes
\[
\frac{R_x}{A E_{\text{bone}}} + \frac{F_{FT}}{A} \left( \frac{L_T + L_F - L_{TI}}{E_{\text{bone}}} - \frac{L_C}{E_{\text{cartilage}}} \right) = 0, \quad \text{Equation A-11}
\]

and can be arranged to
\[
R_x = -F_{FT} \left( \frac{L_T + L_F - L_{TI}}{L_{TI}} - \frac{L_C E_{\text{bone}}}{E_{\text{cartilage}} L_{TI}} \right). \quad \text{Equation A-12}
\]

The term in parenthesis only contains anatomical and material properties that do not depend on load, and thus can be grouped as a constant term, \(K_1\).
\[
\left( \frac{L_T + L_F - L_{TI}}{L_{TI}} - \frac{L_C E_{\text{bone}}}{E_{\text{cartilage}} L_{TI}} \right) = \text{Const.} = K_1 \quad \text{Equation A-13}
\]

The second term in \(K_1\) is dominant. Based upon estimates for \(E_{\text{bone}} \sim 19 \text{ GPa}, E_{\text{cartilage}} \sim 0.5 \text{ MPa}, \) which fall within ranges found in literature \([22]\ [27]\ [21]\), and approximations for \(L_C \sim 2 \text{ cm} \) and \(L_{TI} \sim 10 \text{ cm}, K_1 \) is approximately
\[
K_1 \approx \frac{-L_C E_{\text{bone}}}{E_{\text{cartilage}} L_{TI}} \approx -7.6 \times 10^3 \quad \text{Equation A-14}
\]

The forces can then be summed in the axial direction to solve for the tibial force.
\[
\sum F_x = 0 \\
R_x - R_{x_F} + P_x = 0 \quad \text{Equation A-15}
\]
\[
F_{FT} = \frac{P_x}{1 + K_1} \quad \text{Equation A-16}
\]

**A.3.2. 45 Degrees of Flexion**

At 45 degrees, a similar system occurs (Figure A-5.A). It is still over constrained, and the additional constraint of no net displacement in the x-direction is necessary, thus the constraint equation is still valid. However, the displacement of the femur is no longer due to the bone’s compressive stiffness, but rather a complex state of stress including components affected by the flexibility of the femur as shown in Figure A-5B.
\[
\Delta \approx \frac{F_{FT} L_F^3}{3EI} \quad \text{Equation A-17}
\]

Overall, a similar method could be used to solve the problem at 45 degrees, and all that would change is that there would be a different constant term, \(K_2\), instead of \(K_1\). The tibial force would still be positively related to, and proportional to, the load although by a different factor.
\[
F_{FT} \propto \frac{1}{K_2} \quad \text{Equation A-18}
\]
A.4. Discussion

The free body diagram models predict a proportional relationship between external load and the tibiofemoral force inside the joint. The differences in the constraints placed on the bone clamps for the axial and patellar experiments cause the tibiofemoral force to be related to axial loading only as a function of angle, while the patellar loading requires knowledge of the geometry of the joint and its material properties.
<table>
<thead>
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<th>Description</th>
<th>Unit</th>
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</thead>
<tbody>
<tr>
<td>$\alpha$</td>
<td>flexion angle</td>
<td>degrees</td>
</tr>
<tr>
<td>$F_{\text{APP}}$</td>
<td>applied axial force</td>
<td>N</td>
</tr>
<tr>
<td>$F_{\text{FT}}$</td>
<td>normal tibiofemoral force</td>
<td>N</td>
</tr>
<tr>
<td>$F_{\text{V}}$</td>
<td>shear tibiofemoral force</td>
<td>N</td>
</tr>
<tr>
<td>$M_{\text{F}}$</td>
<td>moment at femur-bone clamp interface</td>
<td>N*m</td>
</tr>
<tr>
<td>$M_{\text{T}}$</td>
<td>moment at tibia-bone clamp interface</td>
<td>N*m</td>
</tr>
<tr>
<td>$R_{\text{F}}$</td>
<td>reaction force at femur</td>
<td>N</td>
</tr>
<tr>
<td>$R_{\text{T}}$</td>
<td>reaction force at tibia</td>
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<table>
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<tr>
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<th>Description</th>
<th>Unit</th>
</tr>
</thead>
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<td>m</td>
</tr>
<tr>
<td>$\delta_{\text{F}}$</td>
<td>change in length of femur at 0 degrees</td>
<td>m</td>
</tr>
<tr>
<td>$\Delta$</td>
<td>change in length of tibia at 45 degrees</td>
<td>m</td>
</tr>
<tr>
<td>$A$</td>
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<td>m$^2$</td>
</tr>
<tr>
<td>$E_{\text{bone}}$</td>
<td>Young’s modulus of bone</td>
<td>Pa</td>
</tr>
<tr>
<td>$E_{\text{cartilage}}$</td>
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<td>Pa</td>
</tr>
<tr>
<td>$F_{\text{APP}}$</td>
<td>external patellar force</td>
<td>N</td>
</tr>
<tr>
<td>$F_{\text{FT}}$</td>
<td>normal tibiofemoral force</td>
<td>N</td>
</tr>
<tr>
<td>$I$</td>
<td>moment of inertia of femur</td>
<td>m$^3$</td>
</tr>
<tr>
<td>$L_{\text{C}}$</td>
<td>combined thickness of articular cartilage on</td>
<td>m</td>
</tr>
<tr>
<td></td>
<td>tibia and femur</td>
<td></td>
</tr>
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<td>$L_{\text{F}}$</td>
<td>length of femur</td>
<td>m</td>
</tr>
<tr>
<td>$L_{\text{IT}}$</td>
<td>length of tibia to the patellar tendon insertion point</td>
<td>m</td>
</tr>
<tr>
<td>$L_{\text{T}}$</td>
<td>length of tibia</td>
<td>m</td>
</tr>
<tr>
<td>$P$</td>
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</tr>
<tr>
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</tr>
<tr>
<td>$P_{y}$</td>
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<td>reaction force at femur</td>
<td>N</td>
</tr>
<tr>
<td>$R_{\text{XT}}$</td>
<td>reaction force at tibia</td>
<td>N</td>
</tr>
<tr>
<td>$S$</td>
<td>normal force of patella on femur</td>
<td>N</td>
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**Nondimensional**

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<thead>
<tr>
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<th>Description</th>
<th>Unit</th>
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<tr>
<td>$K_{1}$</td>
<td>proportionality constant at 0 degrees</td>
<td></td>
</tr>
<tr>
<td>$K_{2}$</td>
<td>proportionality constant at 45 degrees</td>
<td></td>
</tr>
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</table>
Figure A-1. Free body diagram for axial loading at 0 degrees, where $F_{\text{APP}}$ is the applied external force acting axially along the femur and tibia, and $F_{\text{FT}}$ is the tibiofemoral force.
Figure A-2. Free body diagram for axial loading at 45 degrees, (A) entire system, and (B) tibial component, where $F_{\text{APP}}$ is the applied external force acting axially along the femur and tibia, $F_{\text{FT}}$ is the normal component of the tibiofemoral force, $F_{\text{V}}$ is the tangent component of the tibiofemoral force, $R_T$, $M_T$ and $R_F$, $M_F$ are the reaction forces and moments supported at the junction of the tibia and femur with the bone clamp respectively and $\alpha$ is the angle of flexion.
Figure A-3. Effect of flexion angle, $\alpha$, on the relationship between the applied axial force, $F_{\text{APP}}$, and resultant tibiofemoral force, $F_{\text{FT}}$. 
Figure A-4. Free body diagram for patellar loading at 0 degrees. (A) System geometry before applied patellar load, (B) applied and reaction forces, (C) x and y-components of FBD, (D) x-FBD split at point of action of $P_x$, and (E) displacements of respective sections, where $F_{APP}$ is the applied external force acting on the patella parallel to the femur, $F_{FT}$ is the tibiofemoral force, $R_x_T$ and $R_x_F$ are the x-components of the reaction forces at the tibia and femur junction to the bone clamp respectively, $P_x$ is the component of the applied force acting in the x-direction on the tibia.
Figure A-5. Free body diagram for patellar loading at 45 degrees. (A) Forces and reactions on system, (B) schematic of displacement at tibia and femur interface where $F_{\text{APP}}$ is the applied external force acting on the patella parallel to the femur, $F_{\text{FT}}$ is the normal component of the tibiofemoral force, $R_{\text{T}}$ and $R_{\text{F}}$ are the axial-components of the reaction forces at the tibia and femur junction to the bone clamp respectively, and $\alpha$ is the angle of flexion.
Appendix B: Variation in Knee Size

B.1. Introduction

Some of the test groups appeared to have distinctly different intrinsic contract stress distributions. It was hypothesized that the age, and thus size of the immature joint might play a role in these differences. Without knowing the exact age of each of the animals from which the immature stifle joints were obtained, the maturity of the tissue was assumed to be correlated to the size of the joint [30].

B.2. Methods

Several different anatomical measurements were taken on each of the joints: Overall medial-lateral width of tibial plateau, anterior-posterior lengths of each tibial plateau, length of medial and lateral femoral condyles (Figure B-1.A), and the femoral head width (Figure B-1.B). A simple correlation was conducted between these measurements and experimental results such as intrinsic medial and lateral net forces at both 0 and 45 degrees as well as total joint force at both angles for all nine joints.

B.3. Results

Univariate regression indicated a significant a positive correlation of tibial plateau width and the sizes of the tibial plates ($r^2=0.88$, $r^2=0.85$, $p<0.001$, Figure B-2.A) as well as the length of the medial condyle ($r^2=0.60$, $p<0.01$, Figure B-2.B), but these did not correlate to the overall femoral head width ($p=0.76$). Based on Pearson correlation coefficients, there appeared to be moderate negative correlation between the lateral force at 0 degrees and the medial tibial plateau, but a univariate regression indicated there was no significance ($r^2=0.10$, $p=0.40$, Figure B-3.A) nor was the positive correlation with femoral head width significant ($r^2=0.29$, $p=0.08$, Figure B-3.B). The overall joint force at zero degrees follows the same relative trends as the lateral force but again without significance ($r^2=0.11$, $p=0.39$, and $r^2=0.38$, $p=0.13$, respectively, Figure B-3).

B.4. Discussion

Several different anatomical measurements were made as it was not known which value would be most characteristic of the joint size. Due to the high correlation between the medial-lateral width of the tibial plateau and the anterior-posterior lengths, it is possible to make only one of these measurements and infer the others. The lack of correlation between the lateral femoral condyle length or the femoral head width to the other anatomical measurements suggests that there is much more variation in these features in developing joints. Thus, these values would not be suitable alone to describe the joint size.

Overall, there were not many high correlations between joint size and the experimental results. This fact leads to the conclusion that although joint size most likely contributed to animal variability, but it was not the dominant source. Thus, in
the attempt to explain the variability in the different experiments, more investigation must be focused on variation in the experimental setup and methods.
Figure B-1. Schematic of the anatomical measurements on the (A) tibial plateau – tibial plateau width (−) and medial and lateral tibial plate anterior-posterior lengths (---), and (B) femoral head width (−).
Figure B-2. Correlations in anatomical measurements of bovine stifle joint, (A) tibial plateau width to medial (●) and lateral (■) anterior-posterior length and (B) tibial plateau width to medial condyle length (○).
Figure B-3. Correlations in net force values and knee size. (A) Tibial plateau width and the force on the lateral condyle (■) and the entire joint (●), and (B) femoral head width with and the force on the lateral condyle (■) and the entire joint (●).
Appendix C: Data Reduction

C.1. Data Collection

Data for each trial was taken at 960Hz. A trial consisted of a single pass through the joint. For each experimental group, six trials were conducted – three in the posterior- anterior direction and three in the anterior –posterior direction. The two directions did not provide the same data; the anterior-posterior direction consistently provided higher peak loads than the opposite direction. After some pilot tests, it was noted that this was due to, based on visual observations, the shear stress from the sensor induced in the joint, which caused the joint to move slightly in the same direction as the sensor movement. In the anterior-posterior direction this motion, particularly at 0 degrees, caused the knee to bind more, even in the presence of synovial fluid, whereas in the opposite direction, the sensor pulled the joint upwards. The later motion was visibly less the first, presumably because it counteracted against gravity. Because of this discrepancy, it was decided to use only the posterior-anterior data sets as the anterior-posterior data sets would overestimate the normal loads within the joint. A single data set consists of a column of time and the corresponding loads in grams. Data sets were assembled into a series of spread sheets: one file per angle, per joint.

C.2. Filtering

Each data set was plotted as a scatter plot in Excel to graphically determine if there were any outlying points. Although outlying points would likely be averaged out and not be significant in the spatial maps, it was necessary to remove them so that the peak stresses could be more accurately determined. Two types of filtering were done 1) when notes were made at the time of the experiment that the sensor was touching one of the rods at one end or the other and 2) when the sensor behaved erratically or saturated for a small set of data points.

The first sometimes occurred when the upper rod was placed too close to the joint. In order for the sensor to travel the entire condyle length, it might have to leave the joint and then wrap around the upper or lower road for some distance. If this was the case, the data set would generally behave normally and then go to zero before building up again. Any data after the sensor read zero was then replaced with a set of zeros of the same length (Figure C-1.A). Therefore, the data set could be normalized and still cover the same distance as other data sets.

The second type of filtering was necessary when it was determined graphically that there was an erratic spike or a stray point at was inconsistent with the data surrounding in it. Spikes such as these were most frequently determined by a rapid saturation of the sensor in a fraction of a second and then returning to measurable values just as quickly. Data point(s) that were physically inconsistent with the rest of the data points were replaced by taking the average of the unaffected points before and after it (them) (Figure C-1.B).
C.3. Data Reduction

Although the original data was a function of time, the data are presented as a function of distance after time is normalized and multiplied by velocity. The three posterior-anterior data sets were to be averaged to yield one final data set. Since the length of the data sets was controlled manually and the speed of the motor was varied slightly in between trials, the data for each trial was not identical. In order to get an average of the three trials, the data sets needed to be reduced to a subset of points corresponding to 1000 equidistant locations. A Matlab code was introduced to do this as well as to convert the load measurements into stress values.

A set of 1000 evenly distributed points were defined as the “desired final position data.” For each final desired position, the trial position data sets were searched for the closest match without going past the desired position. The index of this trial position was then used to assign the corresponding trial load with the correct final desired position. This was done for all three aspects. When all 1000 loads were compiled from each of the individual trials, they were averaged and then converted to stress by the following equation.

\[
\text{Data}(kPa) = \text{Load}(g) \times \left( \frac{1kg}{1000g} \right) \times 9.81m/s^2 \times \left( \frac{1}{\text{sensor area}} \right) \times \left( \frac{1kPa}{1000Pa} \right)
\]

A matrix of averaged data for both medial and lateral for all angles and loading conditions for each animal were output and saved in an excel file. For the statistical analysis that compared the entire surface area, the data sets were divided into six regions and the stresses were averaged in these areas. The division was not uniform, but rather the first division was an average of 250 points and the remaining 5 regions were an average of 150 data points. This was done because the first region – the most posterior, contained little spatial variation. Therefore, with this division scheme, there was a higher resolution of regions where there was more variation in stress.

C.4. Plotting

Spatial plots were created by first averaging the data sets between animals in Excel. These matrixes were then imported into another Matlab script, which spatially arranged the data sets. Aspect 3 was assigned an x-coordinate of zero, and aspects 1 and 2 were assigned +0.5 for the MFC and -0.5 for the LFC such that all joints are plotted as right joints. In this way the horizontal axis was also normalized, enabling the comparison of joints of different sizes. A filled contour map was then created, which interpolated the data in between the three sets of data.

For lateral femoral condyle under the patellar mechanism, no data was taken for aspect 2 due to the obstruction by the patellar tendon. For spatial plots of this condyle, aspect 3 was not plotted at x=0. Instead, it was plotted at x=0.16. This was necessary because aspect 3 overlaps with both aspect 1 and aspect 2, and in the absence of aspect 2, it was necessary to extend the location upon which aspect 3 was plotted in order to represent that two-thirds of the data was collected.
Figure C-1. Data filtration (A) type 1, where experimentally noted artifacts were removed from the beginning or end of a data set and (B) type 2, where erratic spikes in the data set are replaced by linearly interpolated data.
Appendix D: Dynamic Motion and Belt Tension

D.1. Purpose

The experimental design that was used for the experiments called for the dynamic pull through of the sensor through the stifle joint by a timing belt attached to a motor. This verification experiment was conducted to determine if the tension on the belt and thus the sensor as well as the dynamic motion of the sensor had an effect on the output of the sensor.

D.2. Methods

The joint was harvested and loaded into the CPM device and the sensor was prepared as described earlier (in Chapter 2). The sensor inserted into the joint and fixed in the medial and lateral direction by the lateral guides. The sensor was hooked up to the timing belt and motor. The starting point was determined such that the sensor would pause at a location within the joint where there was a substantial amount of load. The motion of the sensor was broken into five segments that are summarized in Table D-1.

Table D-1. Tension and motor parameters for each time segment.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Time (s)</th>
<th>Belt Tension</th>
<th>Motor</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>0-3</td>
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<td>On</td>
</tr>
<tr>
<td>B</td>
<td>3-6</td>
<td>On</td>
<td>Off</td>
</tr>
<tr>
<td>C</td>
<td>6-9</td>
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<td>Off</td>
</tr>
<tr>
<td>D</td>
<td>9-12</td>
<td>On</td>
<td>Off</td>
</tr>
<tr>
<td>E</td>
<td>12-15</td>
<td>On</td>
<td>On</td>
</tr>
</tbody>
</table>

D.3. Results

Removing tension does reduce the sensor readings as can be seen by the drop in stress between region B and C in Figure D-1. Tension does affect the loads recoded in the knee by an average factor of 4.2 kPa. Furthermore, in this test putting tension back on the system caused a slightly higher average final stress of 9.7 kPa compared to the original average stress of 8.5 kPa.

The static versus dynamic measurements show very little differences. The sensor readings plateau during each static period (regions B,C,D in Figure D-1). Furthermore, when the static portions of the curve (regions B,C,D) were removed, the resulting shape closely matched the dynamic controls (Figure D-2).

D.4. Discussion

It is not primarily the tension in the direction of the belt that causes this shift because the original and final readings still go to zero even when there is tension on
the sensor. It does come into play inside the joint however, as the curvature of the joint causes the sensor to deflect, thus yielding an extra load in the direction perpendicular to the condyle surface, and thus a high pressure picked up by the sensor. This deflection is a function of the angle of the sensor over the condyle in respect to the horizontal guides as well as the amount of tension placed on the system.

Furthermore, in this test putting tension back on the system caused a slightly higher final stress than the original (1.25 kPa). Much of this can be attributed to the attempt to add the tension quickly and thus the belt was jerked tight, more so than would be done in normal experiments.

Efforts were made to keep the amount of tension approximately the same for all the tests by using the same weights and relying on a consistent coefficient of friction between the weights, motor and countertop. Additionally, an attempt was made to minimize the deflection of the sensor within the joint by aligning the top horizontal rod directly over the tibiofemoral contact surface at each angle. Caution must be exercised, however, in comparing quantitative data to other magnitudes found in literature as stress in this experiment could be biased towards the high side.

As there were no sudden changes in sensor readings at the cusp between dynamic and static portions of the curve and that within each static section, the values are fairly constant, it can be reasonably inferred that dynamic and static measurements in this setup result in the same measurement readings.
Figure D-1. Average contact stress in one aspect during time course of belt tension experiment. The sensor was dynamic in sections (A,E) and static in (B-D). The sensor and belt had tension in sections (A,B,D,E) and no tension in section (C).
Figure D-2. Comparison of stress measurements along a single aspect between the dynamic sections (A,E) of the experimental data (●) and the dynamic control (■).
Appendix E: Sensor Thickness Test

E.1. Purpose

The purpose of this experiment was to determine the effect of inserting a sensor of finite thickness into the joint. By inserting the sensor in the joint, the joint is distracted and therefore the load read by the sensor could potentially be higher. The objective of this experiment was to determine if the sensor reading significantly increased with increasing thickness, and if so, to create a relationship to linearly extrapolate to a load value that would correspond to a zero-thickness sensor.

E.2. Method

The joint was harvested and loaded into the CPM device as described earlier in Chapter 2. The sensor inserted into the joint and fixed in the medial and lateral direction by the lateral guides. The sensor was pulled through the joint three times at 0 and 45 degrees. The sensor was removed from the joint, the thickness at the sensor head measured and another plastic strip was added with silicone adhesive. This process was repeated at least three times until the final sensor thickness was at least twice the original thickness. Data was converted into stress and the peak stresses for each trial were determined.

E.3. Results

The peak stresses as a function of sensor thickness are shown in Figure E-1. While there were significant differences between peak stresses at the different angles (p<0.001) and animals (p<0.001), the thickness of the sensor did not have a significant effect (p<0.38).

E.4. Discussion

One of the reasons that could explain why stress did not increase significantly was that as thickness was that by adding thickness to the sensor it became stiffer. Thus, the sensor head could no longer wrap around regions of sharp curvature that might have caused a higher stress. Another reason could be that the sensor thickness is relatively much less than the combined thickness of the cartilage. As the sensor thickness was doubled in the experiment, and the thickness still did not have a significant effect, it was determined that for the purposes of our experiments, the effect of inserting the sensor in the joint was negligible. Therefore, no extrapolation to a zero thickness sensor was conducted, but rather data was used as originally output.
Figure E-1. Average peak contact stresses from measurements in a single aspect at 0° (●, ■) and 45° (○, □).
Appendix F: Supplementary Figures

Figure F-1. Average spatial distribution of contact stress in joint #1 of experiment #1 on medial and lateral condyles (MFC, LFC) at discrete angles in a joint under passive, intrinsic loading represented as a color scale contour mapping onto the joint regions.
Figure F-2. Average spatial distribution of contact stress in joint #2 of experiment #1 on medial and lateral condyles (MFC, LFC) at discrete angles in a joint under passive, intrinsic loading represented as a color scale contour mapping onto the joint regions.
Figure F-3. Average spatial distribution of contact stress in joint #3 of experiment #1 on medial and lateral condyles (MFC, LFC) at discrete angles in a joint under passive, intrinsic loading represented as a color scale contour mapping onto the joint regions.
Figure F-4. Average spatial distribution of contact stress for joint #1 of experiment #2a on lateral and medial femoral condyles (LFC, MFC) at 0 and 45 degrees in a joint under varying axial loads (0, 45, 90, 135 N) represented as a color scale contour mapping onto the joint regions.
Figure F-5. Average spatial distribution of contact stress for joint #2 of experiment #2a on lateral and medial femoral condyles (LFC, MFC) at 0 and 45 degrees in a joint under varying axial loads (0, 45, 90, 135 N) represented as a color scale contour mapping onto the joint regions.
Figure F-6. Average spatial distribution of contact stress for joint #3 of experiment #2a on lateral and medial femoral condyles (LFC, MFC) at 0 and 45 degrees in a joint under varying axial loads (0, 45, 90, 135 N) represented as a color scale contour mapping onto the joint regions.
Figure F-7. Average spatial distribution of contact stress in joint #1 of experiment #2b on medial and lateral condyles (MFC, LFC) at 0 and 45 degrees in a joint under varying patellar loads (0, 14, 24, 45 N) represented as a color scale contour mapping onto the joint regions. Data was not recorded for the most medial aspect of the lateral condyle, as it was obstructed by the patellar tendon.
Figure F-8. Average spatial distribution of contact stress in joint #2 of experiment #2b on medial and lateral condyles (MFC, LFC) at 0 and 45 degrees in a joint under varying patellar loads (0, 14, 24, 45 N) represented as a color scale contour mapping onto the joint regions. Data was not recorded for the most medial aspect of the lateral condyle, as it was obstructed by the patellar tendon, and data was missing for the most lateral aspect of the medial condyle at 0° and 24N load.
Figure F-9. Average spatial distribution of contact stress in joint #1 of experiment #2b on medial and lateral condyles (MFC, LFC) at 0 and 45 degrees in a joint under varying patellar loads (0, 14, 24, 45 N) represented as a color scale contour mapping onto the joint regions. Data was not recorded for the most medial aspect of the lateral condyle, as it was obstructed by the patellar tendon.
References


