Chapter 2
Literature Review

The primary objective of this review is to give the reader a cursory introduction to the human knee joint, its replacement with artificial materials, wear of these materials, and simulated testing methodologies. This chapter begins with a brief overview of the anatomy and physiology of the human knee, including its articulating bones, lubrication, degrees of freedom, and the loads and stresses experienced by the knee joint. A section discussing total knee replacements (TKR) follows, which explains the necessity for a TKR and reviews current commercially-available TKR (e.g., unique geometries, material makeup, theoretical and experimentally measured stresses, motions). The wear rates and mechanisms of TKR are subsequently discussed, specifically addressing UHMWPE wear and damage observed in \textit{ex vivo} implants and knee simulator studies, as well as the causes of the various types of wear. In addition, an overview of the simplified apparatuses that have been developed to simulate UHMWPE wear in TKR is included. The advantages and drawbacks of each device type are discussed. An outline of the research goals for this project concludes this chapter.

2.1 Human Knee Anatomy and Physiology

Knowledge of the anatomy and physiology of the natural knee is crucial in understanding the lubrication, motions, and stresses experienced by a TKR and its materials. Since the most common TKR have sizes and contours similar to those of the human knee, studying the natural knee can give invaluable insight into the performance and deterioration of TKR materials.

2.1.1 Anatomy of the Knee

The human knee joint is the articulation of three bones: the femur (i.e., thigh bone), patella (i.e., kneecap), and tibia (i.e., shinbone). The articulation of the three bones (i.e., the tibiofemoral and patellofemoral joints) forms the knee joint, which is enclosed by a fibrous joint capsule, lined by a synovial membrane. The synovial membrane encapsulates the joint cavity and excretes synovial fluid, which lubricates the
joint. The ends of the articulating bones are covered by articular cartilage that acts to cushion, distribute load, and, along with the synovial fluid, reduces friction in the joint. Figure 2-1 shows a generalized synovial joint with a fibrous capsule, synovial membrane, and joint cavity.

![Figure 2-1. Frontal Section of a generalized diarthrotic (synovial) joint (Tortora and Grabowski, 1993)](image)

In the knee, the distal end of the femur has a curved articular surface that is shaped somewhat like a “horseshoe” with the bend of the “horseshoe” in the front of the femur (Figure 2-2). The two ends of the femur extend backward, and are called the medial and lateral condyles. These surfaces articulate with the medial and lateral tibial condyles, forming the tibiofemoral joint, which flexes and extends the knee. Two fibrocartilagenous discs (i.e., menisci) lie between the tibial and femoral condyles to compensate for the incongruence of the articulating bones. Because, the distal end of the femur is curved and asymmetric in shape, the knee joint not only flexes and extends like a hinge, but it also slides and rotates during flexion. The degrees of freedom of the knee joint will be further explained in the following section.
2.1.2 Motions of the Knee

Human knee studies are excellent resources for predicting the motions and loads experienced by TKR. As mentioned previously, the knee joint does not perform exactly like a hinge during flexion and extension, but rather there are several motions involved between the articulating bones. The knee actually has six degrees of freedom:

- Flexion/extension (F/E) – rolling of the femur over the tibia,
- Anterior/posterior (AP) sliding – back and forth sliding of the tibia,
- Internal/external (i.e., tibial) rotation – rotation of the tibia about its long axis,
- Abduction/adduction – rotation of the tibia in the frontal plane,
- Medial/lateral sliding – side to side sliding of the tibia, and
- Axial compression – movement of the femur and tibia in their own long axes.

The three largest and most recognized movements of the knee (F/E, AP sliding, and tibial rotation) are shown visually in Figure 2-3. Knowledge of the flexion angle, AP sliding
distance, and the angle of tibial rotation is essential in replicating normal human activities in TKR and TKR materials wear testing.

![Figure 2-3](image)

**Figure 2-3. Secondary motions (AP sliding and tibial rotation) of the knee during extension for the cases of a stationary tibia (a) and a stationary femur (b) (Marieb, et al., 1995)**

Determining the accurate motions of the knee throughout one walking cycle has been the difficult goal of many researchers. Errors often arise in complex theoretical modeling or from skin and soft-tissue movement in experimental studies. Lafortune et al. (1992) reported the motions of the knee during one complete walking cycle. To avoid soft-tissue error during this study, traction pins were inserted directly into the femur and tibia of five subjects. Lafortune’s results have often been referenced and compared with simulator studies (Walker et al., 1997; DesJardins et al., 2000) and, hence, it is the goal of this project to reproduce these motion curves in the designed device as well. The F/E, AP sliding (i.e., drawer), and medial/lateral shift curves from Lafortune’s study are shown in Figure 2-4.
Figure 2-4. Flexion angle (A), medial/lateral shift (B), AP sliding/drawer (C), and axial displacement (D) for five different subjects. Each test was for one walking cycle from heel strike (H.S.) to toe-off (T.O.), back to heel strike. The thick line represents the average of the five. (Lafortune et al., 1992)

The two degrees of freedom not addressed in Figure 2-4 are tibial rotation and abduction/adduction. Wilson et al. (2000) performed a study concerning the passive movements of the knee during flexion. Results from this paper are seen in Figure 2-5, which shows tibial rotation and abduction/adduction as functions of flexion angle.
From Figures 2-4 and 2-5, it can be discerned that, in terms of displacement and angle, the three largest motions of the knee are F/E, AP sliding, and tibial rotation. To keep the device designed for this project simple and inexpensive, only these three degrees of freedom will be focused upon in the design.

### 2.1.3 Knee Joint Forces

To properly test TKR materials, knee simulators and wear testing devices should incorporate physiologically-correct loading of the knee. One of the earliest studies on the knee joint forces was done by J.P. Paul in 1965. His resulting joint force curve still remains one of the standard loading configurations for many knee simulators. Morrison (1970) followed this lead with a similar curve to Paul’s, with only slight differences due to different grouping of muscle forces and moments in his model. A comparison of the two force curves in terms of body weight is shown for one walking cycle in Figure 2-6.
The two curves in Figure 2-6 represent the total joint reaction force at the knee. The majority of this force is in the axial direction, due to the large influence of body weight and the quadriceps muscle force on the knee. Seireg et al. (1975) modeled the knee joint and its muscles somewhat differently than Paul or Morrison, but came up with a similar curve, with major differences only in the maximum joint forces. Seireg et al. (1975) divided the joint force into X, Y, and Z directions (Figure 2-7). For simplicity and cost-savings, the proposed device will only apply force in the axial direction; therefore, the concern is the force along the Z-axis shown in Figure 2-7. Burgess et al. (1996) reproduced the Seireg curve for axial force in the design of their six-station knee wear simulator. By constraining displacement, rather than force, in the other two directions (X and Y), they were able to produce results representative of *in vivo* wear.
Figure 2-7. Knee joint reaction and components, in terms of body weight, through one walking cycle (Seireg et al., 1975). Notice the large influence of the Z-component on the total resultant force.

It should be noted that all of the previously mentioned models and experiments for knee motions and forces have been performed with the average size person (approximately a 150-pound male) in mind. Additionally, most of the results are for cases of simple walking. While the majority of load bearing time in an individual’s life is during walking or standing, extreme cases such as stair climbing, running, or weight training could have a significant effect on TKR wear. The maximum forces during stair climbing, for instance, may be 50 percent higher than during level walking (Collier et al., 1991). In addition, the knee joint can flex up to 120° during stair climbing, compared to approximately 70° or less for level walking. By incorporating the flexibility of higher loading and large flexion angles within the current design, this project hopes to address the possibility of elevated wear during extreme cases. For additional information on the anatomy and physiology of the human knee, the reader is referred to Marieb (1995) or Tortora et al. (1997).
2.2 Osteoarthritis

Up to a half-million osteoarthritis (OA) sufferers world-wide undergo knee replacement surgery each year (AAOS, 1997). The most common reason for TKR is severe pain and immobility due to OA. Osteoarthritis, or degenerative joint disease, is characterized by the breakdown of articular cartilage (Arthritis Foundation, 2000). Particles of cartilage may break off and cause pain or inflammation in the knee joint. Over time, the cartilage may wear away entirely, resulting in bone-on-bone contact. Since bones, unlike cartilage, have many nerve cells, direct bone contact can be very painful to the OA sufferer.

In addition to the pain and swelling, the OA sufferer can experience a progressive loss of mobility (i.e., stiffness) at the knee joint. This is due to loss of the joint space, where the articular cartilage has completely worn away (Figure 2-8).

![Figure 2-8. Front (a) and side (b) x-ray views of an arthritic knee. The absence of joint space indicated by the arrows in the front view is due to a loss of articular cartilage (Joint Replacement Institute, 2000).](image)

The exact cause of OA is unknown. It is a progressive disease that normally effects the load-bearing joints (e.g., hip and knee) of people age 45 or older; therefore, it is commonly referred to as the general “wear and tear” of the joints (The Arthritis Society, 2000). Various medications are often recommended to reduce the swelling and pain of OA. Other treatments such as weight loss, braces, orthotics, steroid injections, and physical therapy may also help alleviate pain and restore function. However, since articular cartilage is avascular, or lacks a blood supply, repair and growth of adult
cartilage is minimal (Marieb, 1995). If the pain or immobility becomes too severe and other therapies do not alleviate the symptoms, TKR becomes necessary.

2.2 Total Knee Replacements

There are approximately 250,000 TKR performed each year in the United States alone (AAOS, 1997). Knee replacement is actually just a “resurfacing” of the knee joint. The distal end of the femur is covered with a metal femoral component, normally made of cobalt-chrome alloy (CoCr), and the top of the tibia is replaced with a metal tibial tray and an ultra high molecular weight polyethylene (UHMWPE) bearing surface. Often, the underside of the patella (i.e., kneecap) is also replaced with UHMWPE. Figure 2-9 shows a commercially-available TKR and a sketch of an implanted TKR.

2.3.1 TKR Component Shape and Size

There are numerous types of TKR commercially-available. There are high-conformity and low-conformity types, depending on the person and the desired degree of mobility. Orthopaedic surgeons often evaluate the condition of an individual’s posterior cruciate ligament (PCL) to determine whether a cruciate-retaining or a posterior-stabilizing (PS) TKR should be implanted. Some TKR models are cemented to the bone, while others are press fit (i.e., implanted without the use of bone cement). One of the
latest types of TKR is the mobile bearing implant, in which the UHMWPE bearing is not fixed to the tibial tray, but rather is able to slide and rotate, acting somewhat like the menisci of a healthy knee.

All of the aforementioned TKR are slightly different in shape and size; however, the majority of them are still manufactured using CoCr and UHMWPE. All TKR have femoral components that, like the femoral condyles, are not perfectly round, but rather their radii of curvature decrease toward the posterior aspect of the implant. In Figure 2-10(a), the changing radius of curvature in the left-side X-ray of an implanted TKR can be seen.

![Figure 2-10. Left (a) and front (b) view X-rays of an implanted TKR (Washington Orthopaedic and Knee Clinic, 2000).](image)

For simplicity sake and the capability of testing simple geometries instead of actual TKR components, it was desired to use circular CoCr discs rotating on UHMWPE bearings for the apparatus designed for this project. Hence, the radii of curvature of the femoral components of common and average-size TKR were of particular interest in the design phase of this project. Table 2-1 shows the femoral and tibial components’ radii of curvature for five common TKR from three major implant manufacturers. Further discussion, concerning the chosen radii of the designed device’s CoCr discs, will be subsequently covered in Chapter 3.
Table 2-1. Sagittal radii (mm) of some common TKR models. Notice the slight differences in radii for each type, as well as how the radius of curvature decreases towards the posterior of the femoral component (DesJardins et al., 2000).

<table>
<thead>
<tr>
<th>TKR name</th>
<th>Distal rad</th>
<th>Post. rad</th>
<th>Ant. rad</th>
<th>Post. rad</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kinemax Lo-Stress (Howmedica)</td>
<td>48</td>
<td>20</td>
<td>55</td>
<td>55</td>
</tr>
<tr>
<td>Kinemax Condylar (Howmedica)</td>
<td>48</td>
<td>20</td>
<td>60</td>
<td>70*</td>
</tr>
<tr>
<td>NexGen CR (Zimmer)</td>
<td>44</td>
<td>20</td>
<td>57</td>
<td>79</td>
</tr>
<tr>
<td>NexGen PS (Zimmer)</td>
<td>44</td>
<td>20</td>
<td>57</td>
<td>79</td>
</tr>
<tr>
<td>IBPS (J &amp; J)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PFC—curved tib. (J&amp;J)</td>
<td>32</td>
<td>20</td>
<td>44</td>
<td>44</td>
</tr>
</tbody>
</table>

* Flat at the bottom of the dish.

The radius of curvature for the tibial UHMWPE component also varies posteriorly and from TKR to TKR. To reduce machining costs, it was desired to use a flat tibial disc (radius = ∞) in the apparatus designed for this project. Although the radius of curvature of the UHMWPE will not play a part in the current design, the UHMWPE thickness was taken into consideration. Studies have shown that polyethylene thickness can significantly alter the maximum contact stresses and rates of wear of TKR. Based on finite element analysis, Collier et al. (1991) found that below 6 mm, both surface and Von Mises stresses rise considerably with decreasing thickness. They also concluded that at a thickness below 4 mm, a TKR loaded with four times body weight will exceed the yield strength of polyethylene (Collier, et al., 1991). Based on retrieved components, Wright et al. (1982) have recommended a minimum thickness of 6-8 mm to reduce fatigue failure and delamination. For most TKR and testing seen in the literature, the UHMWPE thickness used was between 10 and 12 mm. Further discussion of the UHMWPE thickness will be subsequently covered in Chapter 3.

2.3.2 Contact Stresses of TKR

Determining the accurate contact areas and stresses experienced by the UHMWPE of TKR may be one of the most difficult tasks of all TKR analyses. Several different methods for finding contact stress have been utilized in the literature, with varying results. Experimental methods, such as Fuji pressure-sensitive film (Collier et al., 1991), Fuji contact area film (Collier et al., 1991), and K-scan electronic pressure...
transducers (Harris et al., 1999), are often used to validate numerical and analytical methods. Although some studies have theoretically determined that the maximum contact stress exists a few millimeters below the surface of the UHMWPE, the greatest area of interest is in finding the contact stress at the surface.

Harris et al. (1999) examined the differences in calculation of measured contact areas using a K-scan sensor and Fuji pressure-sensitivity film. In another study, Fuji film was used to calculate the maximum contact stress, by observing the density of the developed pattern (Collier, et al., 1991). Bartel et al. (1986) examined the differences in maximum contact stress observed using an elasticity method and finite element analysis (FEA). Results from these studies are summarized in Table 2-2.

Table 2-2. Comparison of estimated contact areas and contact stresses using different methods. It should be noted that not all of the other conditions remain constant.

<table>
<thead>
<tr>
<th>Method</th>
<th>Reference</th>
<th>Load</th>
<th>Average Contact Area</th>
<th>Maximum Contact Stress</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elasticity</td>
<td>Bartel et al.</td>
<td>1.5 kN</td>
<td>N/A</td>
<td>18.0 MPa</td>
</tr>
<tr>
<td>FEA</td>
<td>Bartel et al.</td>
<td>1.5 kN</td>
<td>N/A</td>
<td>20.0 MPa</td>
</tr>
<tr>
<td>Fuji film</td>
<td>Collier et al.</td>
<td>2.8 kN</td>
<td>N/A</td>
<td>23.5 MPa</td>
</tr>
<tr>
<td>Fuji film</td>
<td>Collier et al.</td>
<td>0.7 kN</td>
<td>0.3 cm²</td>
<td>N/A</td>
</tr>
<tr>
<td>K-scan sensor</td>
<td>Harris et al.</td>
<td>3.6 kN</td>
<td>3.5 cm²</td>
<td>N/A</td>
</tr>
<tr>
<td>Fuji film</td>
<td>Harris et al.</td>
<td>3.6 kN</td>
<td>2.3 cm²</td>
<td>N/A</td>
</tr>
</tbody>
</table>

*a Result is maximum contact stress at the surface at full extension for one TKR.
*b Result is an average of 11 different TKR at 15° flexion.
*c Result is an average of 11 different TKR at full extension.
*d Result from one TKR at extension with a 10 mm thick UHMWPE insert.

As seen in Table 2-2, different experimental and theoretical methods produced varying results of contact area and contact stress. It should also be noted that the yield strength of UHMWPE is around 21 MPa. Based upon this fact, Bartel’s (1986) and Collier’s (1991) studies would suggest that the yield strength of the material is very nearly or is, in fact, being exceeded. However, studies addressing the dynamic contact stress during rolling have produced results that suggest the actual contact stresses are much lower. Waldman et al. (1997) predicted the rolling contact stress to be less than half of Bartel’s FEA result. For this project, it was desired to obtain a maximum contact stress in the range as reported in the literature. In Chapter 3, elasticity methods were
applied to determine the contact stress of the proposed CoCr disc on a flat block of UHMWPE.

2.3.3 TKR Kinematics and Friction

The primary goal of TKR manufacturers is to design implants that restore natural knee function. Orthopaedic companies would like their TKR to encounter the same kinematic motions and loading as a healthy knee. Unfortunately, this is not always the case. A study in 1996 showed that AP translation was detected in only three out of five patients tested with a Kinemax Modular Total Knee prosthesis (Long et al., 1996). Since most TKR have similar sizes and geometries as the natural knee, major differences in kinematics are thought to be due to the large increase in friction between CoCr and UHMWPE, compared to that of cartilage on cartilage of the human knee (Sathasivan et al., 1997). The natural knee joint, incorporating articular cartilage on articular cartilage with synovial fluid as a lubricant, is thought to have a coefficient of friction (COF), commonly represented as $\mu$, anywhere from 0.02 to as low as 0.004 (Kippers, 2000). To get an idea of how little friction there actually is in the human knee, consider that ice sliding on ice has a COF of around 0.05 (Kippers, 2000). Implanted TKR have been observed to have a COF around 0.07 (Reinholz, 2000). Although this is still rather low, it may be three or four times higher than that of the natural knee.

It can be assumed that a higher COF in TKR leads to less motion in the knee. Given the same degrees of flexion and joint force, a TKR will have considerably less AP translation and tibial rotation than that of the natural knee (Sathasivan et al., 1997). Consequently, it would take more force to move the TKR the same distances and angles as the natural knee. This premise was shown in a model developed by Sathasivan and Walker (1997). In their model, at 60° flexion and a given joint force, AP displacement and tibial rotation were 15 mm and 55°, respectively, with $\mu = 0.0$. With $\mu = 0.07$, the motions were reduced to only 10 mm of displacement and 40° of rotation (Sathasivan et al., 1997).

It has also been shown that the COF of TKR materials increases with increased wear. Reinholz et al. (2000) tested the coefficients of friction of polished CoCr disc
sliding and rotating on “new” and “used” UHMWPE. Their conclusion of an increased COF for the worn UHMWPE is shown in Figure 2-11.

![Figure 2-11. The coefficient of friction at a velocity of 140 mm/s and a load of 1000 N for a CoCr disc sliding on UHMWPE. Diluted calf serum was used as the lubricant. Ra refers to the surface roughness average of the UHMWPE (Reinholz et al., 2000).](image)

The apparatus for this project was designed assuming a COF similar to that observed in TKR in vivo. In Chapter 3, the predicted COF for the designed device is discussed. Predicting an accurate COF was necessary to determine the maximum forces and torques, and thus, the required motors and linear tables to be used in the design of the apparatus.

### 2.3.4 TKR Failure and Revision

Although some TKR designs have survival rates of better than 90% at 10 years (Godest et al., 2000), TKR do fail. The main type of failure observed clinically is due to wear of the UHMWPE (Collier et al., 1991). Wear of the UHMWPE is of considerable concern because it can lead to the generation of debris that may cause synovitis (i.e., joint swelling) and osteolysis (i.e., bone resorption) of the surrounding bone. In extreme cases, wear debris migration can cause bone necrosis (i.e., bone death) and loosening of the
tibial tray. Wear can also lead to misalignment or instability of the joint and can initiate fracture of the tibial component (Walker *et al.*, 2000).

Implant failure often leads to revision surgery. In cases absent of osteolysis or implant loosening, only the UHMWPE needs be replaced. Unfortunately, aseptic (i.e., non-toxic) loosening is the most common reason for revision, in which case, an entirely new TKR must be implanted. In these revisions, more bone must be removed from the patient and a larger TKR is inserted. Revision surgery is much more complex and technically more difficult, and thus, much riskier than first-time knee replacement. As well, revision surgery may require bone grafts, prolonged operating time, and an increased hospital stay, all of which lead to increased hospital costs (Lavernia *et al.*, 1995). There is also a good chance that the range of motion in the knee will be greatly decreased in comparison with the initial replacement (hipsandknees.com, Change ref format here 2000).

With the ever-increasing average human life span and implantation of TKR in younger patients, improved TKR materials and designs are needed to reduce the increasing number of revisions. One step in the development of new materials and designs is to look at the wear mechanisms and modes of failure observed in retrieved implants.

### 2.4 UHMWPE Wear

Retrieval studies can determine mechanisms of wear on TKR and indicate the wear performance of individual designs (Wasielewski *et al.*, 1994). By observing the wear characteristics of explanted TKR, the influence of wear on several factors, such as conformity, tibial rotation, AP sliding, UHMWPE thickness, etc., can be analyzed.

Historically, three types of wear have been identified on clinically retrieved knee implants: abrasive wear, adhesive wear, and fatigue wear (Wang *et al.*, 1996). Abrasive wear occurs by the ploughing of the UHMWPE surfaces whenever loose particles, such as bone cement and UHMWPE particles, are trapped between the articulating surfaces (Wang *et al.*, 1996). Adhesive wear occurs by the shearing of microscopic asperities on the UHMWPE surface, which can lead to what is referred to as pitting. The most common and most catastrophic type of TKR wear is fatigue wear, in which initiation and
propagation of surface or subsurface microcracks leads to surface cracking and subsurface delamination (Wang et al., 1996). For further information on wear mechanisms of TKR, refer to Walker et al. (2000) and Collier et al. (1991).

Many retrieval studies have analyzed implants with the aforementioned types of wear and attempted to classify the wear types with the specific type of TKR (Wang, 1998; Collier, 1991; Wasielewski, 1994). For instance, pitting is the most common type of surface damage associated with highly conforming TKR, while delamination most often occurs in lower conformity implants (Schmalzried et al., 1999). Studies have also examined the wear patterns and the UHMWPE orientation of retrieved TKR, attempting to associate the patterns with knee motions, such as tibial rotation and AP sliding (Blunn et al., 1991; Wasielewski, 1994). Retrieval studies, however, cannot predict the wear performance and wear rates of new or improved TKR materials; therefore, a screening device to test TKR materials and designs in vitro is essential to the evolution of a TKR.

2.5 Knee Simulators

The most complex wear testing devices are knee simulators, which test actual knee prostheses. Most knee simulators provide physiologic loading and can reproduce at least four degrees of freedom: F/E, AP sliding, tibial rotation, and abduction/adduction. For instance, the AMTI-Boston Knee Simulator (Watertown, MA), shown in Figure 2-12, has the following features and limitations (AMTI, 2000).

- Six station testing,
- Up to 134° of flexion,
- Up to 25 mm of AP sliding,
- Up to 20° of tibial rotation,
- Unconstrained abduction/adduction and medial sliding,
- Physiologic loading of up to 4500 N,
- Operating speeds of up to 2 Hz, and
- Temperature control.
Commercial knee simulators manufactured by MTS (Eden Prairie, MN) and Instron-Stanmore (Canton, MA) have similar features to the AMTI-Boston, with only minor differences in the maximum angles, distances, and loads.

![Three stations of the AMTI-Boston Six Station Knee Simulator, Model KS2-6-1000 (AMTI, 2000).](image)

Knee simulators have been shown to produce similar wear rates and wear patterns as seen with explanted TKR (LaBerge et al., 1990); however, simulators do have limiting factors. Knee simulators only test actual prostheses, which are rather expensive and can be difficult to obtain from the manufacturers. Consequently, to test a new TKR material in a simulator, complex and expensive machining would be required to shape the material into the proper geometric configuration. Secondly, knee simulators are very expensive, with some commercial models costing in excess of $200,000 (AMTI, 2000).

### 2.6 Simplified Wear Testing Devices

Recently, the goal of several researchers has been to find an inexpensive alternative to knee simulators to test knee replacement materials (Wang et al., 1999; Blunn et al., 1991). By incorporating a CoCr disc rotating and sliding on a flat UHMWPE block (Figure 2-13), these researchers hope to reproduce clinically-relevant wear mechanisms to ultimately predict the clinical performance of TKR materials,
without the need for actual knee prostheses or complex machining. The impetus for the design project undertaken in this research was largely based upon the groundwork of Wang and coworkers (1999) and Blunn and coworkers (1991).

![Diagram of CoCr disc on UHMWPE block](image)

**Figure 2-13. Simple schematic showing the loading and motions of the proposed configuration of a CoCr disc on an UHMWPE block.**

### 2.6.1 Howmedica Biaxial Line Contact Wear Machine (Wang *et al.*, 1999)

In 1999, Wang, in collaboration with colleagues at Howmedica (Rutherford, NJ), developed a “biaxial line-contact wear machine for the evaluation of implant bearing materials for total knee joint replacement” (Figure 2-14). The design rationale of these researchers was based on the apparatus being capable of simulating both geometry and motion of the knee joint (Wang *et al.*, 1999). Rather than incorporate a physiologically-correct loading curve, as well as for cost savings and simplicity, the researchers applied a static load of 1150 N. The designed apparatus (subsequently referred to as “Howmedica”) rotates a CoCr ring, loaded about an UHMWPE block, which oscillates about the load axis, to simulate tibial rotation (Wang *et al.*, 1999). AP sliding, abduction/adduction, and medial sliding were not included within the design criteria or end product. Specifications for the device are denoted in Table 2-3.
Validation of the “Howmedica” wear testing device was done by comparing test results obtained with the “Howmedica” apparatus with those produced by a commercially-available MTS six-station knee simulator (Eden Prairie, MN). Specifically, the following criteria were used in the assessment of the “Howmedica” apparatus:

(1) Observance of similar wear mechanisms in terms of morphology of wear surfaces and debris,

(2) Observance of similar wear rates, and

(3) Observance of similar wear rankings for a group of different materials tested (Wang et al., 1999).

Initial examination of the results obtained with usage of the “Howmedica” machine looked promising. Specifically, scanning electron microscope (SEM) micrographs showed similar wear surfaces, wear rates were in the same order of magnitude (≈ 30 % difference) as those observed with the knee simulator, and wear rates decreased with increasing polyethylene cross-linking and molecular weight, as anticipated. However, the lack of AP sliding and physiologically-correct loading raises doubts to these results. Excluding AP sliding prevents the CoCr disc from rolling over the UHMWPE. In pure sliding, contact stresses can be significantly higher than those determined estimated during rolling, thus, increasing wear rates (Waldman et al., 1997).
Conversely, static loading may actually decrease wear rates. Blunn et al. (1991) showed AP sliding with cyclic, dynamic loading resulted in greater deformation and wear of UHMWPE than AP sliding with only static loading. The apparatus designed by Blunn and coworkers (1991) is subsequently discussed in greater detail.

2.6.2 Cyclic Sliding Wear Testing Machine (Blunn et al., 1991)

Like the “Howmedica” machine, a CoCr disc loaded on an UHMWPE block was used in Blunn’s study (1991) (See Figure 2-13). The apparatus designed by Blunn and coworkers incorporated F/E and AP sliding with a cyclically-applied load. As stated by the researchers, the main design criteria were to show the influence of AP sliding on wear. Hence, tibial rotation was not included in the design. Several tests were conducted with varying conditions to compare sliding with rolling, and cyclic loading with static loading. Unlike Wang’s study (1999), Blunn and coworkers were only concerned with comparing results obtained on the designed apparatus, and not with independent simulator results. Based upon these studies, the authors concluded that pure sliding and cyclic loading are significantly more damaging to UHMWPE than rolling and static loading, respectively.

Although Blunn’s results may have some relevance in comparing different knee motions’ influences on wear, extrapolating these findings to predict the wear of different TKR materials and using this device to screen potential TKR materials may not be justified. One major limitation is the lack of tibial rotation, which eliminates the presence of any cross-shearing during testing. Without cross-shearing, molecular UHMWPE chains tend to orient themselves in one direction, strengthening the material in that direction, while weakening it in the direction perpendicular to the orientation (Wang et al., 1999). Along with the lack of tibial rotation, Blunn’s testing apparatus did not incorporate physiologically-correct loading. The cyclical loading was actually sinusoidal and applied at a frequency of 10 Hz. Physiologically-correct loading, previously discussed in Section 2.1.3, unlike a sine wave, has four peaks and cycles at a considerably lower frequency of approximately 1 Hz. Specifications for both of these simplified wear apparatuses along with testing parameters are summarized in Table 2-3.
Table 2-3. Specifications for two simplified wear testing devices

<table>
<thead>
<tr>
<th>Wear Testing Machine:</th>
<th>Cyclic Sliding Wear Testing Device</th>
<th>Biaxial Line Contact Wear Machine</th>
</tr>
</thead>
<tbody>
<tr>
<td>Authors:</td>
<td>Blunn et al. (1991)</td>
<td>Wang et al. (1999)</td>
</tr>
<tr>
<td>Femoral Component and Size:</td>
<td>CoCr disc 75 mm OD, 25 mm width</td>
<td>CoCr disc 71.9 mm OD, 25 mm width</td>
</tr>
<tr>
<td>Tibial Component and Size:</td>
<td>UHMWPE block 4 mm thickness</td>
<td>UHMWPE block 16 x 16 x 12.5 mm thickness</td>
</tr>
<tr>
<td>Applied Load:</td>
<td>2.2 kN (sinusoidal @ 10 Hz)</td>
<td>1.15 kN (static)</td>
</tr>
<tr>
<td>F/E:</td>
<td>Yes (angles not available)</td>
<td>Yes (0-60°)</td>
</tr>
<tr>
<td>AP sliding:</td>
<td>Yes (0-10 mm)</td>
<td>No</td>
</tr>
<tr>
<td>Tibial rotation:</td>
<td>No</td>
<td>Yes (0-30°)</td>
</tr>
<tr>
<td>Lubrication:</td>
<td>Distilled water</td>
<td>Bovine serum</td>
</tr>
</tbody>
</table>

2.7 Research Goals

Despite the drawbacks of the two apparatuses previously discussed, both studies (Wang et al., 1999; Blunn et al., 1991) demonstrated that a simplified machine consisting of a CoCr disc and a UHMWPE block could be used to produce wear and potentially predict the wear characteristics of current TKR materials. The impetus for the design project undertaken in this research was largely based on improving upon these previously designed apparatuses. Specifically, our design criteria included the incorporation of F/E, AP sliding and tibial rotation, as well as utilizing physiologically-correct loading, experienced in the knee joint.

The primary goal of this research was to design and construct a simplified wear testing apparatus for TKR materials, capable of quickly and inexpensively screening currently-available and new, alternative materials. By incorporating the knee’s most contributory motions and its complex loading curve, it was hoped that a device capable of reproducing wear characteristics similar to those seen with knee simulators and retrieved implants could be designed. A review of the literature has indicated physiologically-relevant loading and motions experienced by the knee and TKR, as well as devices previously constructed to simulate these motions. The apparatus designed for this research has taken into account the information gathered from the literature and is subsequently discussed in Chapter 3.