Biomechanical Response of the Knee Complex to a Non-Linear Spring-Loaded Knee Joint Orthosis

Christine Dailey Walck
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by

Christine Dailey Walck

A Dissertation Submitted to the College of Engineering Department of Mechanical Engineering in Partial Fulfillment of the Requirements for the Degree of Doctor of Science in Mechanical Engineering

Embry-Riddle Aeronautical University
Daytona Beach, Florida
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DEDICATION

This dissertation is dedicated to my husband and family.

To my husband, Gregory Walck: your endless support and kindness has allowed me to continue my educational journey without compromise. You have been my constant source of support and encouragement during the challenges of this dissertation and life. I am truly thankful for having you in my life as my husband and friend.

To my family: each day I think of you and how much you all believe in me. You all have always loved me unconditionally and your good examples have taught me to work hard for the things that I aspire to achieve. Without you and your selfless acts to support me, I would not be writing this. It is because of you that I was able to overcome the burdens I was given to bear. You took it on yourselves to make my burdens your own and to see me through them. For this, I thank you sincerely.

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Abstract

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Year: 2019

The use of knee orthosis, specifically on the tibiofemoral joint, has become the preferred treatment method for rehabilitation and injury prevention. Recent orthotic designs have proposed the use of springs to store energy during motion and provide assistance to the lower extremity. However, the biomechanical influence of orthotics at the knee joint has not been quantified. As such, this study aims to quantify the biomechanical response of the tibiofemoral joint complex to a non-linear spring-loaded knee joint orthosis (KJO). Joint angles, moments, and forces obtained from two dynamic trials were applied to a newly developed computational musculoskeletal model, and a static equilibrium problem was solved at each instant during the squat cycle, with and without a non-linear spring-loaded KJO, to find individual muscle forces of the lower extremity. The KJO was seen to increase the gluteus maximus muscle force while decreasing the soleus muscle force throughout the squat cycle. Due to the increased activation occurring in the gluteus maximus and the decrease in the rectus femoris during the brace-on descent, the knee joint axis moved in a less anterior direction then in the brace-off descent. As a result, the pelvis translated in a more posterior direction due to the tension supplied by the gluteus maximus and the ease of the soleus. Furthermore, hip
and knee flexion were decreased in the upright position during the brace-on conditions. Results suggest that the KJO could be used as a performance tool to encourage a more balanced synergy that employs the posterior chain musculature versus a quadriceps dominant strategy while preventing hyperextension tendencies. In addition, the model created in this study paired with the inverse dynamics approach, could contribute to the current knowledge of biomechanical response estimations in clinical movement and force production analysis aiding physical therapist and orthopedic specialist in proscribing the most appropriate KJO to a specific patient.
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Definition of Terms

Abduction is the movement of a limb away from the midline of the body.

Adduction is a movement of a limb toward the midline of the body.

Anatomical plane a hypothetical plane used to transect the human body, in order to describe the location of structures or the direction of movements.

Anatomical position is the description of the body in a specific stance. In the anatomical position, the body is upright, standing with feet flat on the ground, palms facing forward, and arms by the side.

Angular movement refers to movement that produces an increase or decrease in the angle between the adjacent bones. It includes: flexion, extension, abduction and adduction.

Anterior refers to the front of the human body.

Bubble-level is an instrument designed to indicate whether a surface is horizontal (level) or vertical (plumb).

Co-contraction is the simultaneous activation of antagonist muscles around a joint) provides the nervous system with a way to adapt the mechanical properties of the limb to changing task requirements—both in statics and during movement.

Condyles are rounded protuberance at the end of some bones which form an articulation with another bone.

Coronal plane divides the body into anterior and posterior sections.

Dynamometry is the measurement of external force production by a human subject in a specific exercise.

Femur commonly called the thigh bone; it’s the largest, longest and strongest bone in the body. The round knobs at the distal end of the bone are called condyles.
Fibula is a long, thin bone in the lower leg on the lateral side, and runs alongside the tibia from the knee to the ankle.

Functional knee brace is designed to provide stability for unstable knees during activities of daily living or vocational or avocational activities.

Inferior is a term used to refer to what is below something.

Isokinetic pertains to a concentric or eccentric contraction of a muscle in which the speed and tension are constant throughout the range of lengthening or contracting.

Isokinetic dynamometry is a passive device which resists applied forces and controls the speed of exercise at a predetermined rate. Such dynamometers generally provide a record of applied force throughout a joint range of motion.

Jarring effect is a result from the user attempting to extend past a predetermined locking angle limit set by the user/therapist. This attempt creates a hard stop which causes significant pain and potentially causes further injury to the knee.

Lachman is a clinical test used to diagnose the integrity of a patient’s ACL.

Lateral refers to the side farthest from the body’s midline (opposite of the medial side)

Ligaments Groups of fibrous tissue connecting two bones to form a joint.

Lysholm Knee Scoring Scale is a questionnaire that evaluates the outcomes of knee ligament surgery.

Medial refers to the side closest to the midline of the body

Muscle insertion is the point or mode of attachment of a skeletal muscle to the bone or other body part that it moves.

Muscle origin is the point or mode of attachment of a skeletal muscle to the bone or other body part that is the fixed attachment.
OpenSim is an open source platform designed to let users develop models of MSK structures and to create dynamic simulations of a wide variety of movements.

Osteoarthritis is also referred to as degenerative arthritis or as degenerative joint disease. It caused by aging and wear and tear of cartilage.

Patella also known as the kneecap; is a thick, circular-triangular bone which articulates with the femur. The patella covers and protects the anterior articular surface of the knee joint.

Posterior refers to the back of the human body.

Pronation is the rotation of the forearm and hand so that the palm faces downward or toward the back.

Prophylactic knee brace is intended to prevent or reduce the severity of knee injuries in contact sports.

RAND/UCLA Approach primarily as an instrument to enable the measurement of the overuse and underuse of medical and surgical procedures.

Rehabilitation knee brace is designed to allow protected and controlled motion during the rehabilitation of injured knees.

Sagittal plane divides the body into right and left sections.

Superior is a term used to refer to what is above something.

Supination is the rotation of the forearm and hand so that the palm faces forward or upward.

Tegner Activity Level Scale is a graduated list of activities of daily living, recreation, and competitive sports. The outcomes evaluate performance and activity restrictions both before and after surgery.

Tendon is a strong band of tissues; associate muscles with bones.

Tibia The larger bone comprising the shin located medial to the fibula. The tibia runs from the knee to the ankle. The tibia is commonly called the shin bone.
<table>
<thead>
<tr>
<th>Term</th>
<th>Description</th>
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<tbody>
<tr>
<td>Transverse plane</td>
<td>divides the body into superior and inferior sections.</td>
</tr>
<tr>
<td>Unloader knee brace</td>
<td>is specifically designed to reduce the pain and disability associated with osteoarthritis of the medial compartment of the knee.</td>
</tr>
<tr>
<td>Visual analogue scale</td>
<td>is unidimensional measure of pain intensity that tries to measure a characteristic or attitude that is believed to range across a continuum of values and cannot easily be directly measured.</td>
</tr>
<tr>
<td>Wilcoxon signed-rank test</td>
<td>is a non-parametric statistical hypothesis test used to compare two related samples, matched samples, or repeated measurements on a single sample to assess whether their population mean ranks differ (i.e. it is a paired difference test).</td>
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# List of Acronyms

<table>
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<tr>
<td>AAOS</td>
<td>American Academy of Orthopedic Surgeons and the American Academy</td>
</tr>
<tr>
<td>ACL</td>
<td>Anterior Cruciate Ligament</td>
</tr>
<tr>
<td>ACLD</td>
<td>Anterior Cruciate Ligament Deficient</td>
</tr>
<tr>
<td>ACLR</td>
<td>Anterior Cruciate Ligament Reconstruction</td>
</tr>
<tr>
<td>ADP</td>
<td>Adenosine Piphosphate</td>
</tr>
<tr>
<td>ASIS</td>
<td>Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>ATP</td>
<td>Adenosine Triphosphatase</td>
</tr>
<tr>
<td>BF</td>
<td>Bicep Femoris</td>
</tr>
<tr>
<td>BW</td>
<td>Body Weight</td>
</tr>
<tr>
<td>CA</td>
<td>Cylinder Axis</td>
</tr>
<tr>
<td>CK</td>
<td>Creatine Kinase</td>
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<td>Central Nervous System</td>
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<tr>
<td>COM</td>
<td>Center of Mass</td>
</tr>
<tr>
<td>COP</td>
<td>Center of Pressure</td>
</tr>
<tr>
<td>DF</td>
<td>Dorsiflexion</td>
</tr>
<tr>
<td>DOMS</td>
<td>Delay Onset of Muscle Soreness</td>
</tr>
<tr>
<td>EF</td>
<td>Extension Facet</td>
</tr>
<tr>
<td>FEA</td>
<td>Flexion-Extension Axis</td>
</tr>
<tr>
<td>FF</td>
<td>Flexion Facet</td>
</tr>
<tr>
<td>FPS</td>
<td>Frame per second</td>
</tr>
<tr>
<td>GMD</td>
<td>Gluteus Medius</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Full Form</td>
</tr>
<tr>
<td>--------------</td>
<td>-----------</td>
</tr>
<tr>
<td>GMX</td>
<td>Gluteus Maximus</td>
</tr>
<tr>
<td>JCF</td>
<td>Joint Contact Force</td>
</tr>
<tr>
<td>KAFOs</td>
<td>Knee-Ankle-Foot Orthoses</td>
</tr>
<tr>
<td>KE</td>
<td>Kinetic Energy</td>
</tr>
<tr>
<td>KJO</td>
<td>Knee Joint Orthosis</td>
</tr>
<tr>
<td>L</td>
<td>Lateral</td>
</tr>
<tr>
<td>LCL</td>
<td>Lateral Collateral Ligament</td>
</tr>
<tr>
<td>M</td>
<td>Medial</td>
</tr>
<tr>
<td>MCL</td>
<td>Medial Collateral Ligament</td>
</tr>
<tr>
<td>MSK</td>
<td>Musculoskeletal</td>
</tr>
<tr>
<td>MVC</td>
<td>Maximum Voluntary Contraction</td>
</tr>
<tr>
<td>NICE</td>
<td>National Institute for Health and Care Excellence</td>
</tr>
<tr>
<td>NSAID</td>
<td>Nonsteroidal Anti-inflammatory Drug</td>
</tr>
<tr>
<td>OA</td>
<td>Osteoarthritis</td>
</tr>
<tr>
<td>OARSI</td>
<td>Osteoarthritis Research Society International</td>
</tr>
<tr>
<td>PCL</td>
<td>Posterior Cruciate Ligament</td>
</tr>
<tr>
<td>PE</td>
<td>Potential Energy</td>
</tr>
<tr>
<td>PF</td>
<td>Plantarflexion</td>
</tr>
<tr>
<td>PFS</td>
<td>Patellofemoral Syndrome</td>
</tr>
<tr>
<td>R</td>
<td>Right</td>
</tr>
<tr>
<td>RCT</td>
<td>Randomized Controlled Trials</td>
</tr>
<tr>
<td>RF</td>
<td>Rectus Femoris</td>
</tr>
<tr>
<td>RICE</td>
<td>Rest, Ice, Compression and Elevation</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>---------------------------------------</td>
</tr>
<tr>
<td>RMS</td>
<td>Root Mean Square</td>
</tr>
<tr>
<td>ROM</td>
<td>Range of Motion</td>
</tr>
<tr>
<td>SCOKJ</td>
<td>Stance-Control Orthotic Knee Joint</td>
</tr>
<tr>
<td>SOL</td>
<td>Soleus</td>
</tr>
<tr>
<td>SR</td>
<td>Systematic Review</td>
</tr>
<tr>
<td>SS</td>
<td>Subject-Specific</td>
</tr>
<tr>
<td>ST</td>
<td>Semitendinosus</td>
</tr>
<tr>
<td>TAF</td>
<td>Tibial Articular Facet</td>
</tr>
<tr>
<td>TEA</td>
<td>Transepicondylar Axis</td>
</tr>
<tr>
<td>VGRF</td>
<td>Vertical Ground Reaction Force</td>
</tr>
<tr>
<td>VL</td>
<td>Vastus Lateralis</td>
</tr>
<tr>
<td>VM</td>
<td>Vastus Medialis</td>
</tr>
<tr>
<td>WB</td>
<td>With Brace</td>
</tr>
<tr>
<td>WOB</td>
<td>Without Brace</td>
</tr>
</tbody>
</table>
Structure of Dissertation

The dissertation consists of six chapters. This structure gives an outline of the research and the sequence of the dissertation. The objective of this dissertation was to investigate the biomechanical responses of the knee complex to a non-linear spring-loaded knee joint orthosis (KJO). In addition, the musculoskeletal (MSK) model created in this study paired with the inverse dynamics approach, could contribute to the current knowledge of biomechanical response estimations in clinical movement and force production analysis aiding physical therapist and orthopedic specialist in prescribing the most appropriate KJO to a specific patient.

Chapter (1) gives a background of the knee anatomy and skeletal muscle make-up and physiology, including activation and contraction sequence and muscle mechanics. Chapter 1 introduces the three types of muscle contractions and concludes with an explanation on the production of movement.

Chapter (2) is a clinical review highlighting the controversy behind knee braces based on their designs and effectiveness through clinical evidence. Chapter 2 begins with a discussion on common knee injuries and problems; including a discussion on nonsurgical treatments. The chapter concludes with a detailed review on the current state of knee braces.

Chapter (3) explains the benefits of subject-specific testing and includes materials and methods used to investigate the biomechanical responses of the knee complex to a non-linear spring-loaded KJO. It also explains the combination of a subject-specific MSK model with a measured kinematic three-dimensional motion capture analysis subjected to
kinetic parameters. Chapter 3 finishes by reviewing OpenSim’s analysis pipeline which was used for this study.

**Chapter (4)** summarizes the main finding of this study, including joint range of motion, joint torque and individual muscle forces. This chapter introduces graphs that compare the brace-off trial conditions to the brace-on trial conditions. Chapter 4 also compares the results from this study to current literature as well as electromyography data as a validation tool.

**Chapter (5)** provides a discussion that reviews the results presented in Chapter 4 while offering an explanation as to how the knee brace affects each parameter of interest. Chapter 5 reveals that the results presented in Chapter 4 disagree with the original hypothesis which stated the non-linear spring response could lower the quadriceps during knee extension. This chapter ends by identifying the brace may be more suitable as a performance enhancement tool rather than a quadriceps assistive device.

**Chapter (6)** discusses the limitations found within this study including limitations found in materials, the MSK model, and within the analysis pipeline. Chapter 6 also considers future work that can be established from this study’s results. This chapter ends by stating specific contributions this study makes to the field of biomedical engineering well as the medial field in general.
Chapter 1

Skeletal Muscle Structure and Function

1.1 Knee Anatomy

The knee is the most complicated and largest joint in the human body. Its main biomechanical role is to allow locomotion with minimal energy exertion required by its muscle groups. It transmits, absorbs and redistributes forces created during the activities of daily life; causing the knee to bear enormous weight and pressure loads while providing flexible movement and stability. Such demands result in the knee being the most vulnerable part in the body. For example, during gait the knee supports 1.5 times the person’s body weight (BW); climbing stairs requires the knee to support about 3-4 times BW; and squatting requires about 8 times BW [1].

To unravel the biomechanics of the knee complex, its structure in a biomechanical manner using arthrokinematics described in three dimensions based on a series of planes and axes will be investigated. There are three planes of motion that pass through the human body. These planes are illustrated in Figure 1 as the sagittal plane, the coronal (frontal) plane, and the transverse (horizontal) plane. The sagittal plane lies vertically, dividing the body into right and left parts. The frontal plane also lies vertically but divides the body into anterior and posterior parts. The transverse plane lies horizontally and divides the body into superior and inferior parts.
There is an axis associated with each plane. An axis is a straight line around which an object rotates about. Therefore, movement at a joint takes place in a plane about an axis. The three axis of rotation are (1) the sagittal axis which passes horizontally from posterior to anterior and is formed by the intersection of the sagittal and transverse planes, (2) the frontal axis which passes horizontally from left to right and is formed by the intersection of the frontal and transverse planes, and (3) the vertical axis which passes vertically from inferior to superior and is formed by the intersection of the sagittal and frontal planes.

At the individual joint level, movement occurs in all three planes not solely in one. Therefore, the movement described in each plane about its respected axis is considered a gross movement [3]. With this in mind, it can be said that flexion/extension movements are described in the sagittal plane; abduction/adduction movements are described in the coronal plane; and rotation movements are described in the transverse plane.
described in the coronal plane; and internal/external rotation and pronation/ supination movements are described in the transverse plane.

The knee is often described in the sagittal plane because its main range of motion (ROM) is flexion and extension created by its joints.

1.1.1 Joints

Joints are formed where two bones meet. The knee consists of two joints: the tibiofemoral joint which connects the femur to the tibia; and the patellofemoral joint which joins the patella to the femur. These two joints work together to form a modified hinge joint that allows the knee to bend and straighten, but also to rotate slightly and move from side to side. This type of joint is classified as a synovial joint. The four bones that form these two joints are the tibia, patella, femur, and fibula.

Figure 2 Schematic of the knee with callouts for each bone that is attached to the knee [1].
When a person sits, the tibia and femur barely interact with each other. However when standing, these two bones lock together to form a stable unit. For this unit to work together as a hinge, their surfaces must articulate. This is done through the rounded protuberance at the end of the femur called the femoral condyle. The femoral condyle is made up of the lateral and medial condyle as seen in Figure 3. The lateral condyle is the more prominent and is the broader both in its antero-posterior and transverse diameters. The medial condyle is the longer than the lateral condyle and, when the femur is held with its body perpendicular, projects to a lower level. When the femur is in its natural oblique position, however, the lower surfaces of the two condyles lie practically in the same horizontal plane [4]. The condyles are in part responsible for how the knee moves especially in flexion and extension. Its influence is described in detail in section 1.5.4. However, many other factors control how a joint moves, such as ligaments which can restrict movement and help control stability.

Figure 3 Schematic of the knee in flexion. Schematic highlights the femoral condyles and the tibial plateau with a detailed view of the tibiofemoral surfaces and the patellofemoral surfaces [5].
1.1.2 Ligaments

Bones are attached by strong tough bands that aren’t particularly flexible called ligaments. There are four well known ligaments within the knee and they are the most important structures in controlling its stability. The four ligaments are two collateral ligaments and two cruciate ligaments. The collateral ligaments consist of the medial collateral ligament (MCL) which attaches the medial side of the femur to the medial side of the tibia, and the lateral collateral ligament (LCL) which attaches the lateral side of the femur to the lateral side of the fibula. These two ligaments limit the sideways motion of the knee.

The two cruciate ligaments are the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL) which attaches the tibia and the femur at the center of the knee. The ACL is located deep inside and in front of the PCL. The ACL limits rotation and anterior motion of the tibia while the PCL limits posterior motion of the tibia and is location behind the ACL.
1.1.3 Tendons

The knee is also home to two major tendons—the quadriceps tendon and patellar tendon. Tendons are elastic tissue structures that connect muscles to bones and are technically part of the muscle itself, serving as stabilizers to the knee. Tendons also transmit muscle force to the bone resulting in movement. This force transmission will be explained in detail throughout the upcoming sections.

The quadriceps tendon specifically connects the quadriceps muscle of the thigh to the superior patella and works with the quadriceps to extend the leg. The patella is also connected at its inferior to the tibia creating a pulley-like mechanism for improving angle of pull. This mechanism results in greater mechanical advantage in knee extension. This pulley changes the direction of an applied force to help support the work of the quadriceps muscles during the contraction that results in the knee extension. This is demonstrated in Figure 5 where \( F_Q = F_P \) and the pulley represent the patella.
1.1.4 Muscles

To describe any skeletal muscle in the body, the most basic description is its location. Traditionally, the muscle is named or described based on where it begins (its origin) and where it ends (its insertion). There are two main muscle groups in our lower extremity which keep the knee well aligned and moving correctly. These two groups are identified as the quadriceps and hamstrings.

The quadriceps is a collection of four muscles found on the anterior of the thigh as shown in Figure 6. Although they are four individual muscles they all share a common purpose; they are responsible for straightening the knee; bringing a bent knee to a straight position. In other words, when the quadriceps muscles contract they cause the knee to straighten and when the quadriceps relax, the knee bends. The quadriceps collection consists of the rectus femoris, vastus intermedius (located under the rectus femoris), vastus medialis, and the vastus lateralis. The rectus femoris and vastus intermedius give
the central power to extension working through the patella as a level or pulley described earlier. While the vastus medialis and the vastus lateralis assist in the extensor role, they also give peripheral support to both movement and the joint itself [7]. Figure 6 shows the origin and insertion point for these muscles. Notice that each muscle is inserted by a common tendon on the tuberosity of the tibia.

The hamstring is a group of muscles located in the posterior compartment of the thigh and controls the knee when moving from a straight position to a bent position. The hamstrings include semimembranosus, semitendinosus, biceps femoris long head and biceps femoris short head. These muscles are shown in Figure 7.

Figure 6 Origin and insertion locations points for the quadriceps muscle collection [8].
1.2 Skeletal Muscle Structure

A skeletal muscle is a working system made of many components invested in a large network of connective tissue able to produce motion and force through a mechanical outcome called contractions. The outcome is produced by a functional unit comprised of two discrete units. Specifically, the muscle belly and tendon that connect the belly to the bone. The muscle belly consists of fibers and connective tissue which play a huge role in the muscle’s ROM and its ability to produce contractions [10].

The muscle fiber is a long cylindrical, multinucleated cell that is filled with smaller rod-like units of filaments. Figure 8 magnifies the view of a belly and the organization of the filaments composing such muscle.
Figure 8 Magnified view of a belly and the organization of the filaments composing such muscle. Figure highlights myofibrils and its subunits sarcomeres [10].

Figure 8 also magnifies a single myofibril. Myofibrils are composed of subunits called sarcomeres. Each sarcomere contains two types of myofilaments (also referred to as contractile proteins): actin and myosin. The structure of the myosin-containing filament generates tension during muscle contraction while actin-containing filament regulates the tension generated. Thus, the sarcomeres are the functional unit of contraction. The actin myofilaments are anchored at both ends of the sarcomere at the Z-line and project into the interior of the sarcomere where they surround the thicker, less abundant myosin myofilament. The amount of contractile proteins within the sarcomere is strongly related to a muscle’s contractile force [10].
Contraction results from the formation of cross-bridges between the myosin and actin myofilaments, causing the actin chains to “slide” on the myosin chain. Contraction of a whole muscle is equal to the sum of singular contraction events occurring within the individual sarcomere. The sliding theory is represented in Figure 9. It illustrates that when the muscle is contracted the actin filaments slide closer together and when the muscle is relaxed they slide further away [10].

Figure 9 Schematic structure of the myosin-actin interaction during muscle contractions [10].

1.2.1 Muscle Architecture

Muscle architecture is the physical arrangement of muscle fibers relative to the axis of force generation at the macroscopic level. This is what determines a muscle's mechanical function. Fiber arrangement can be classified under four main groups as seen in Figure 10: circular, convergent, parallel and pennate muscles.
Circular muscle structures appear circular in shape and are normally sphincter muscles which surround an opening, a natural body passage, or an orifice which relaxes as required by normal physiological functioning such as the mouth. Convergent muscle structures occur when the muscles origin is wider than the point of insertion. This fiber arrangement allows for maximum force production. An example is the pectoralis major. Convergent muscles are also sometimes known as triangular muscles [11].

Parallel muscles have fibers that extend parallel to the tendon. These muscles are normally long and cause large movements. They are not very strong but have good endurance. Examples include sartorius and sternocleidomastoid [10, 11]. Within the parallel group are three main categories: strap, fusiform, or fan-shaped. These muscles are more spindle shaped, with the muscle belly being wider than the origin and insertion. Examples are biceps brachii and psoas major [11].

Pennate muscles have a large number of muscle fibers per unit making them very strong, but these muscle tire easily. Within the pennate group there exist the unipennate, bipennate, and multipennate muscles.

Unipennate have fibers that are oriented at a single angle relative to the tendon, while muscles that are oriented on two sides of the tendon are termed bipennate. A multipennate muscle is when the fibers are oriented at several angles. The multipennate is the most general category of them all [10]. Each muscle fiber arrangement is shown in Figure 10.
Figure 10 Muscle skeletal fiber arrangement: fusiform, convergent, circular, multipennate, bipennate, unipennate, and parallel [11].

Each muscle has architectural properties. These properties consist of mass, muscle length, fiber length, pennation angle, physiological cross-sectional area (PCSA), and fiber to muscle length ratio ($L_f/L_m$ Ratio). Muscle architectural values are the best predictors of muscle function, and the most important architectural properties are muscle fiber length
and PCSA, because muscle excursion and velocity are directly proportional to muscle
fiber length while isometric muscle force is directly proportional to muscle PCSA [12].
The architectural properties are defined for the quadriceps and hamstrings in Table 1 and
Table 2.

Furthermore, as discussed in Chapter 3, these properties are of utmost importance
when developing models of MSK structures and creating dynamic simulations of
movement.

Table 1

Architectural Properties of the Quadriceps Collection [13].

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Mass (g)</th>
<th>Muscle Length (cm)</th>
<th>Fiber Length (cm)</th>
<th>Pennation Angle (°)</th>
<th>PCSA (cm²)</th>
<th>Lf/Lm Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vastus Lateralis</td>
<td>110.55 ± 43.33</td>
<td>36.28 ± 4.73</td>
<td>7.59 ± 1.28</td>
<td>13.93 ± 3.49</td>
<td>13.51 ± 4.97</td>
<td>0.21 ± 0.03</td>
</tr>
<tr>
<td>Vastus Medialis</td>
<td>375.85 ± 137.18</td>
<td>27.34 ± 4.62</td>
<td>9.94 ± 1.76</td>
<td>18.38 ± 6.78</td>
<td>35.09 ± 16.14</td>
<td>0.38 ± 0.11</td>
</tr>
<tr>
<td>Vastus Intermedius</td>
<td>171.86 ± 72.89</td>
<td>41.20 ± 8.17</td>
<td>9.93 ± 2.03</td>
<td>4.54 ± 4.45</td>
<td>16.74 ± 6.91</td>
<td>0.24 ± 0.04</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>239.44 ± 94.83</td>
<td>43.90 ± 9.85</td>
<td>9.68 ± 2.30</td>
<td>29.61 ± 6.89</td>
<td>20.85 ± 7.17</td>
<td>0.22 ± 0.04</td>
</tr>
</tbody>
</table>

Note: Where PCSA = Physiological Cross-sectional Area and Lf/Lm Ratio = Fiber Length / Muscle Length. Muscles were taken from each of 21 formaldehyde-fixed human lower extremities (mean age ± standard deviation, 83 ± 9 years; male:female ratio, 9:12; height, 168.4 ± 9.3 cm; weight, 82.7 ± 15.3 kg)
Table 2

Architectural Properties of the Hamstring Collection [13].

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Mass (g)</th>
<th>Muscle Length (cm)</th>
<th>Fiber Length (cm)</th>
<th>Pennation Angle (°)</th>
<th>PCSA (cm²)</th>
<th>Lf/Lm Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Femoris LH</td>
<td>113.37 ± 48.53</td>
<td>34.73 ± 3.65</td>
<td>9.76 ± 2.62</td>
<td>11.58 ± 5.50</td>
<td>11.33 ± 4.75</td>
<td>0.28 ± 0.08</td>
</tr>
<tr>
<td>Biceps Femoris SH</td>
<td>59.79 ± 22.62</td>
<td>22.39 ± 2.50</td>
<td>11.03 ± 2.06</td>
<td>12.33 ± 3.61</td>
<td>5.06 ± 1.69</td>
<td>0.49 ± 0.07</td>
</tr>
<tr>
<td>Semitendinosis</td>
<td>99.74 ± 37.81</td>
<td>29.67 ± 3.86</td>
<td>19.30 ± 4.12</td>
<td>12.86 ± 4.94</td>
<td>4.82 ± 2.01</td>
<td>0.65 ± 0.11</td>
</tr>
<tr>
<td>Semimembranosus</td>
<td>134.31 ± 57.56</td>
<td>29.34 ± 3.42</td>
<td>6.90 ± 1.83</td>
<td>15.09 ± 3.43</td>
<td>18.40 ± 7.53</td>
<td>0.24 ± 0.06</td>
</tr>
<tr>
<td>Gluteus Maximus</td>
<td>547.24 ± 162.17</td>
<td>26.95 ± 6.42</td>
<td>15.69 ± 2.57</td>
<td>21.94 ± 26.24</td>
<td>28.17 ± 11.05</td>
<td>0.62 ± 0.22</td>
</tr>
</tbody>
</table>

Note: Where PCSA = Physiological Cross-sectional Area and Lf/Lm Ratio = Fiber Length / Muscle Length. Muscles were taken from each of 21 formaldehyde-fixed human lower extremities (mean age ± standard deviation, 83 ± 9 years; male:female ratio, 9:12; height, 168.4 ± 9.3 cm; weight, 82.7 ± 15.3 kg)

1.3 Skeletal Muscle Physiology

Skeletal muscle is formed by a structure-function relationship. This relationship is best described through muscle physiology itself. To recap, muscles are composed of building blocks called sarcomeres and are designed for specific function such as the output of a large excursion. So how does the muscle know when to start this output? The answer is: where skeletal muscle physiology begins.

Physiology of a muscle can be broken down into three main parts: 1) activation and contraction sequence of muscles, 2) muscle mechanics and 3) muscle fiber types and motor units. Due to the present research goals, only parts 1) and 2) will be discussed. Part 3) involves a detailed description of how muscle fibers are activated by their composite nerves and can be found in reference [13].
1.3.1 Activation and Contraction Sequence

The activation and contraction sequence begins with a process known as excitation and contraction (EC) coupling. EC coupling is a microscopic chain of events that causes a neural activation signal to culminate in the muscle contraction. The details within these events are beyond the scope of this research but in short, a signal is sent from the nervous system through the peripheral nerve which arrives at the neuromuscular junction, the interface between muscle and nerve. The nerve signals the muscle to “do its thing.” The muscle replies with a chemical reaction of calcium ions. The release of calcium ions in the region of the myofilaments results in muscle contraction whereas the uptake of calcium ions results in muscle relaxation [13].

This part of the activation and contraction sequence can be thought of as a cycle where the conversion of chemical energy to mechanical energy occurs. One cycle consists of activation, contraction, and relaxation resulting in a muscle twitch, a small local muscle contraction. If a second excitation should occur before the first cycle is completed the process is then termed temporal summation. Furthermore, if a train of excitations or activation pulses is delivered to the muscle, separated in time by different amounts, the result is then termed as a tetanic contraction and the force generated is quite different. The faster the train of excitations occurs, the higher the resulting force will be. This is because there is less time for relaxation to occur. Frequency, \( f \), can be related to excitations or pulses through the equation below.

\[
f = \frac{1}{\text{interpulse}}
\]  

(1)
Therefore, muscle force varies as a function of activation frequency; serving as a method in which the central nervous system (CNS) can use to alter muscle force. If high forces are required, the CNS can deliver high-frequency pulses; conversely, if low forces are required the CNS can deliver low-frequency pulses [13].

1.3.2 Muscle Mechanics

Thus far it has been shown that force can be altered through fiber lengths and activation frequency. Thus muscle force can be considered a function of two variables: length and velocity.

1.3.2.1 Length-Tension Curve

Length-tension relationship reflects the fact that tension generation in skeletal muscles is a direct function of the magnitude of overlap between the actin and myosin filaments. The length-tension curve is shown in Figure 11 and is generated by maximally stimulating a skeletal muscle at a variety of discrete lengths and measuring the tension generated at each length (zero velocity) [13].

By investigating the curve, it is seen that too much or too little overlap leads to sub-optimal tension being developed but where the overlap is “just right,” maximal tension is developed [14]. In other words, when the muscle is at the optimum length, the number of active cross bridges is the greatest. When the muscle is stretched beyond this length, the number of active cross bridges decreases because the overlap between the actin and myosin fibers decreases. As the muscle becomes shorter than the optimum length the thin filaments at opposite ends of the sarcomere first begin to overlap one another and interfere with each other's movements. This results in a slow decrease in tension as the sarcomeres get shorter. Then, as the sarcomeres get shorter the thick
filaments come into contact with the Z lines and the decrease in tension with decreasing length becomes even steeper [14].

Figure 11 Length-tension curve illustrates the sliding-filament model and the cross-bridge cycle [14].

1.3.2.2 Force-Velocity Curve

A normal muscle’s aim is to produce an amount of force through contraction that matches the weight load being lifted. As seen from the activation-contraction sequence, that contraction speed is faster for light loads, and slower for heavier loads. This is confirmed through the force-velocity curve that describes the force generated by a muscle as a function of velocity under conditions of constant load. Such relationship gives a
further insight into two of the three types of contractions, specifically concentric contractions and eccentric contractions. Each type is further discussed in section 1.4 but for now, a typical force-velocity curve is shown in Figure 12.

![Force - Velocity Curve](image)

Figure 12 Force-velocity curve shows the relationship of muscle force production [15].

The force-velocity curve shows that for concentric contractions the force is greater at a lower velocity. As the velocity increases the contractile force deceases. When velocity is equal to zero, a peak isometric force occurs. If the load becomes greater than this peak, the muscle is required to lengthen (eccentric contraction). As the lengthening velocity increases the force will also increases. This is because the muscle fibers are contracting as they are lengthening.
1.3.2.3 Length-Tension-Velocity Curve

Neither constant length nor constant velocity of contraction at a constant force is common during normal activities [13]. In fact, muscle force changes with length and velocity. The two separate curves, length-tension and force-velocity, share a common point shown in Figure 13. This point is where the maximum isometric tension occurs (at \( L_o \) when velocity equals zero resulting in a maximum tension, \( P_o \)).

Figure 13 Hypothetical muscle length-force-velocity surface for skeletal muscle. The shaded regions represent a “slice” of the surface at either constant length or velocity [13].

Figure 13 is the hypothetical muscle length-force-velocity surface for skeletal muscle. Shaded regions represent a “slice” of the surface at either constant length or
velocity. A “slice” of the surface at constant length is simply a force-velocity curve measured at that length while a “slice” of the surface at constant velocity is simply a length-tension curve measured at that velocity.

By examining muscle length-force-velocity behavior, it can be concluded that when muscle velocity is high, the force will be low no matter what the sarcomere length is. Therefore, at high velocities the length is not significant. Only at low concentric velocities does the sarcomere length become significant at modifying the force. These relationships are important in neuromotor control as we attempt to understand how muscle actions can be responsible for observed external movements [13].

1.4 Types of Contraction

As mentioned previously, there are three ways in which a muscle can contract. Also, as shown previously, a muscle’s contractile force is dependent on its contraction velocity and its fiber length. Thus a muscle can be shortened (concentric contraction), lengthened (eccentric contraction) or it can remain static (isometric contraction) [16].

Figure 14 shows the color-coded tension-velocity curve. The resting length, also known as isometric contraction, occurs when the velocity is equal to zero (blue) and serves as a transition point between the eccentric contraction (a negative contraction velocity shown in red) and the concentric contraction (a positive contraction velocity shown in green).
1.4.1 Concentric Contraction

A concentric contraction is the result of a muscle that has actively shortened. This action can be easily seen and understood through weight training examples. In weight training, the concentric contraction occurs when the muscles become shorter developing a tension force. This typically results from lifting a weight up, against gravity [18].

For example, the concentric motion during a bicep curl occurs during the rising phase; bringing the weight from the hanging position to the curled position. This is illustrated in Figure 15.
23

Figure 15 Concentric motion during a bicep curl occurs during the rising phase [19].

This is very similar to most free-weight exercises. Some other examples are the:

- Shoulder press: occurs when the weight is pressed above the head.
- Push-up: when the arms push the body upward and away from the ground surface.
- Leg Extension: when the weight raises and the leg extends.

1.4.2 Isometric Contraction

Isometric contraction results when the muscle is actively held at a fixed length. This type of contraction requires the muscle to exert a force large enough to statically hold the load or resistance. An example of an isometric contraction would be carrying an object in front of the body. The force generated during an isometric contraction is wholly dependent on the length of the muscle while contracting. This is shown in the force-length curve in Figure 12 and Figure 14. The force-length relationship indicates that muscles generate the greatest force when at their resting (ideal) length [18].
Looking back at the bicep curl example used in the concentric contraction example, it can be said that the muscle contracts isometrically if the athlete pauses, producing no movement. This is shown in Figure 16.

![Figure 16 Isometric contraction occur when the athlete statically holds the weight in a given position [19].](image)

Interestingly the isometric contraction, sometimes called the Paused Isometric, is frequently used in the conventional bodybuilding community. Essentially, at a specific point in a movement you pause and simply hold the weight in a position for time before continuing on with the movement. This can be useful for sticking point training - a weak point in your ROM, or to exhaust the glycogen supplies in the muscle tissue [18].

Another example of an isometric contraction occurs at the top of a leg extension movement where the quadriceps muscles contract isometrically. This occurs when the athlete holds for a count or pauses after the knee is in the extended position before lowering the weight. Although the athlete’s quadriceps is not actively shortening, they are contracting to hold the weight with the knee fully extended.
1.4.3 Eccentric Contraction

An eccentric contraction is produced when a muscle is lengthened while under tension. During an eccentric movement the muscle is fully activated however it is forced to lengthen due to the high external load. This type of contraction typically results from lowering the weight [18].

Examine once more the bicep curl. The eccentric contraction occurs during the lowering phase. This is shown in Figure 17.

Figure 17 Eccentric contraction occurs during the lowering phase of the bicep curl [19].

Referring back to the force-velocity graph in Figure 12 and Figure 14, the eccentric contractions produce more force than either isometric or concentric contractions. Maximum eccentric strength is estimated to be between 1.5 and 2.0 times maximum concentric strength [10]. Some other examples of an eccentric contraction can be found in the same examples provided for concentric contraction. The contractions however occur in different phases of the movements.
- Shoulder press: the eccentric motion occurs as the weight descends back to the starting position.
- Push-up: when the arms lower the body back toward the floor.
- Leg Extension: quadriceps muscles eccentrically contract as the weight lowers and the leg bends.

Eccentric contractions are currently a very popular area of study for three reasons. First, is due to the fact that much of a muscle's 'normal' activity occurs while it is actively lengthening; so eccentric contractions are physiologically common. Second, muscle injury and soreness are also selectively associated with eccentric contraction. Third, muscle strengthening may be greatest using exercises that involve eccentric contractions. Due to these reasons, eccentric contractions may have very important applications therapeutically to strengthen muscle [16].

1.5 Introduction to Production of Movement

Thus far it has been shown that within the MSK system, muscles generate force. We now might ask ourselves, what role do muscles play in the production of movement? The answer is somewhat complex however if we investigate the components of movement, the system of movement itself becomes more attainable.

Movement is enabled by three components, specifically muscles, tendons, and bones. Their interaction, the movement system, consists of a series of events where muscles generate and transmit force to bones via tendons. If sufficient force is produced, the bone will move. Within the movement system is the muscle-tendon unit. This unit is formed from muscle fibers and fascicles that exit the muscle belly forming the white glossy tendon. The tendon is designed to transmit muscle force to the bone and slide
during movement. The tendon is usually thought of as a rigid linkage structure; however this is not the case. In reality, tendons are relatively compliant tissue structures. Its compliancy forms its own set of mechanical properties which in turn affect the muscle’s length and overall production of strength [13].

Previously in section 1.1.3, it was described that tendons have elastic and compliant tissue structures that connect muscles to bones and are technically a part of the muscle itself serving as stabilizers to the knee. Now how a tendon affects muscle properties and trends will be investigated, as well as the role they play in transmitting the force to the bone, creating movement.

1.5.1 Mechanical Properties of the Tendon

In addition to tendons being compliant and elastic they also vary greatly in size, length and thickness, throughout the body. Therefore, tendons properties will be discussed before investigating how they affect or interact with the muscle.

Since tendons vary so much in size, length, and thickness, when seeking relations, it is convenient to normalize deformation and load of the tendon in question. The term for normalized deformation is strain which is the change in length relative to its original length and is expressed as:

\[
\varepsilon = \frac{l - l_o}{l_o}
\]  

(2)

Where:

\(l\): length relative

\(l_o\): original length
Normalized load is termed as stress and is expressed as:

\[ \sigma = \frac{F}{A} \]  

(3)

Where:

- \( \sigma \): stress
- \( F \): force
- \( A \): cross-sectional area

Strain and stress are valuable pieces of information that ultimately yield a tendon’s elasticity. The elasticity of tendon is expressed through Young's modulus which is the ratio of the stress acting on a substance to the strain produced.

\[ E = \frac{\sigma}{\varepsilon} \]  

(4)

Where:

- \( E \): Young's modulus
- \( \sigma \): stress
- \( \varepsilon \): strain

Recall that muscle force generation is sensitive to its length and more so, its velocity. Therefore, if during a muscle contraction, a tendon’s compliance (elasticity) is significantly high, the tendon will strain allowing the muscle to deform further. This is
where the tendon starts to affect a muscle’s properties. Table 3 lists the components of movement along with their associated Young’s Modulus value as well as a list of comparable, more well-known objects.

<table>
<thead>
<tr>
<th>Production of Movement’s Component</th>
<th>Young’s Modulus for Production of Movement’s Component</th>
<th>Well-known Object for Comparison</th>
<th>Young’s Modulus for Well-known Object for Comparison</th>
</tr>
</thead>
<tbody>
<tr>
<td>Passive Muscle being stretched</td>
<td>10 kPa</td>
<td>Rubber</td>
<td>20 kPa</td>
</tr>
<tr>
<td>Tendon</td>
<td>1 GPa</td>
<td>Softwood such as Pine</td>
<td>0.6 GPa</td>
</tr>
<tr>
<td>Bone</td>
<td>20 GPa</td>
<td>Hardwood such as Walnut</td>
<td>15 GPa</td>
</tr>
</tbody>
</table>

It has now been shown that a tendon’s compliance affects the muscle, yet the magnitude of tendon deformation depends on its own stress-strain properties. Therefore, the previous isometric and isotonic assumptions used in the study of production of force aren't truly valid in the current study of movement. Isometric and isotonic assumptions help us understand muscle force parameters but are limiting in the study of production of movement because movement involves muscle and tendon properties together not individually [13].

Because muscle and tendon properties interact, muscle activation is never truly maximal or performed at a constant length or under a constant load and typically cannot be measured directly. Instead the muscle and the tendon must form the muscle-tendon
unit and find a common ground in which both can perform optimally. The following section will describe the muscle-tendon properties as a unit rather than separate identities which will then allow optimal parameters to be found.

1.5.2 Muscle-Tendon Properties

Since muscles generate force based on their length-tension and force-velocity properties resulting in tendon deformation and the magnitude of tendon deformation depends on its own stress-strain properties, we must consider both muscle and tendon as a unit rather than separately. The muscle-tendon unit is unique and distinct from the properties of either the muscle or the tendon alone. The flow chart shown in Figure 18 represents the interaction between a muscle and tendon during contraction.

Figure 18 Interaction between muscles and tendons during contraction [13].
This interaction begins with a muscle activation resulting in a force generation based on the muscle’s length-tension and force-velocity properties. Tendons then deform under the force of the muscle contraction based on its strain-stress properties. This process continues during an ‘isometric’ contraction until both muscle and tendon velocities equal zero. The process shows that the muscle shortens at the expense of the tendon lengthening. Therefore, tendons can be seen in the way as mechanical springs. The schematic shown in Figure 19 represents a muscle contracting “isometrically” while in series with a tendon. It further illustrates that the muscle shortens at the expense of the tendon. Note that the tendon or spring may strain to different extents due to differences in intrinsic compliance.

Figure 19 Schematic model of a muscle contracting “isometrically” while in series with a tendon. The muscle shortens at the expense of the tendon [20].
So far the way a force is transmitted from muscle through tendon has been discussed. However, it was also said that a tendon connects muscle to bone. This brings up the next step in the movement system, which is to investigate how bones react to force and what their limiting factors might be.

1.5.3 Joint Components and Movement

To recap, force is generated at the muscle and transmitted, via tendon to the bone. If the force transmitted is strong enough then the affected bone will move. This force transfer can be seen as a mechanical system and therefore can be quantitatively described using the concept of torque [13]. Torque can then be said to be the result of the interaction between muscle and joint. Torque is equal to the cross product of applied force and its moment arm. Both force and moment arm are vectors dependent on direction and magnitude and are expressed as follows.

\[ \tau = \vec{r} \times \vec{F} \]  

(5)

\[ \tau = |r| |F| \sin(\theta) \]  

(6)

\[ r = d \times \sin(\theta) \]  

(7)
Where:

F: force

θ: angle of application

τ: torque

r: perpendicular distance from the muscle’s angle of application to the center of rotation (joint axis).

d: distance

By applying the concept of torque to the MSK system, it can be stated that torque represents “strength.” There are three strategies for changing the magnitude of strength: 1) changing the force itself, 2) changing the length of the moment arm, and/or 3) changing the angle between the two [13]. These are illustrated in Figure 20.
Figure 20a & b) illustrates that torque increases when either force or moment arm is increased. This is also shown in the torque equation since torque is directly proportional to both force and moment arm. c) Illustrates that torque increases as the angle between application of force and axis of rotation approaches 90 degrees.

Figure 20a) shows that torque is increased when force is increased. This is also seen from the torque equation since torque and force are directly proportional. Torque is also directly proportional to the moment arm. However, the latter may not be seen as easily. Take a look at Figure 20b). It illustrates that a muscle with a short moment arm will produce a larger angular excursion. However, if that same muscle lengthened its moment arm, the angular excursion imposed on the joint would decrease. This is expected as it follows the principles of basic geometry.
To better understand the moment arm concept, think of opening a paint can using a screw driver. If the screw driver is a stubby one then it takes much more effect to lift the lid than if the screw driver was a long one. Figure 20c) illustrates that torque increases as the angle between the application of force and the point of rotation approaches 90 degrees. Again, this concept can be seen directly from the torque equation.

At the joint angle where torque is at a maximum, the muscle force and joint moment arm are involved to varying degrees but neither muscle nor joint properties need to be maximized. This is because when measuring “strength” interaction between muscle and joint properties is really being measured.

1.5.4 Joint moment of the knee

Torque is often synonymous with moment. Moment is defined as the turning effect produced by a force, which can be thought of as the rotational analog to linear force (turning force), and is calculated by multiplying the perpendicular force by the distance from the pivot (or axis of rotation) [21]. Therefore, to calculate moment, one must know the appropriate axis of rotation.

Earlier it was stated that the knee’s articulating surfaces affect how the knee moves throughout its ROM. As of today, how we track the knee’s axis of rotation during its ROM is still a debatable topic. In the late 19th century the knees axis of rotation was described using “instant center of rotation” (ICR) shown in Figure 21. Because the surfaces of the condyles are not perfect circles, the axis of rotation was said to move as the knee flexes and extends rather than stay fixed at its center (Figure 22). The ICR model was useful because it linked the shape of the condyles to the motion characteristics of the knee [22].
Figure 21 Successive instantaneous centers can be mapped as the knee moves from flexion to extension. This is known as the instant center pathway. In the normal knee, the pathway is semicircular located in the femoral condyle [23].

Figure 22 Path of instantaneous center of rotation (PICR) of the knee. Line drawn perpendicular from the instantaneous center to the joint surface is normally parallel to the joint surface, indicative of a sliding motion between surfaces [23].
This principle was reanalyzed by Rudolf Fick in 1911 using three, rather than two, dimensions. Fick concluded that the flexion-extension axis of the knee did not lie in the sagittal plane but was offset by several degrees. This offset orientation of the flexion-extension axis would result in a single, fixed axis, rather than an instant center [22].

In 2003, it was concluded that the knee has four independent axes: patella, posterior condylar, distal condylar, and longitudinal axes. The axes combine to produce the characteristic helical motion of the knee [22]. The derivation of the flexion-extension axis was done using three-dimensional imaging technology, which led to the development of a model of the tibiofemoral joint with three independent axes of motion, about which the knee moves during different kinematic events. These three axes include the posterior condylar axis, distal condylar axis, and the longitudinal axes shown in Figure 23 and Figure 24.

Figure 23 Diagram represent two flexion axes of the knee. Abbreviations: EF, extension facet; FF, flexion facet; TAF, tibial articular facet; M, medial; L, lateral [22].
Several models have attempted to integrate the structure and function of the knee, ranging from a hinge at its most simplistic, to a complex roll-glide mechanism [22]. The four-bar linkage theory was one attempt shown in Figure 25. It was developed to integrate the bony geometry and ligament restraint in the knee. The theory defines the four rigid links as the ACL and PCL and the bony structure of the femur and tibia. This model married cruciate ligament isometry with the roll-glide pattern of knee motion providing an explanation the appearance of roll back of the femoral condyles on the tibial plateau, which is seen on plain radiographs and fluoroscopy. However, to describe the cruciates as isometric may be an oversimplification [22].
In 2015, an investigation was conducted to calculate the flexion-extension axis (FEA) of the knee through in-vivo knee kinematics data, and then compared it with two major anatomical axes of the femoral condyles: the transepicondylar axis (TEA) defined by connecting the medial sulcus and lateral prominence, and the cylinder axis (CA) defined by connecting the centers of posterior condyles [24]. After analyzing knee kinematics data on 20 healthy subjects acquired during weight-bearing condition using bi-planar x-ray imaging and 3D-2D registration techniques the author of the study, Lei Ren, concluded that the CA is closer to the FEA compared with the TEA; it can better serve as an anatomical surrogate for the functional knee axis Figure 26.
Figure 26 Positions and orientations of the transepicondylar axis (TEA), cylinder axis (CA), and the computed flexion-extension axis (FEA). A. Coronal view. B. Sagittal view. C. Transverse view [24].

Therefore, the knee axis of rotation is not easily found and is still up to debate regarding the best representation. The axis of rotation is however one of importance for calculating joint moments. Section 3.3.2.2 describes how the knee axis of rotation is modeled for the current study.

1.5.5 Energy

One theme that emerges from the discussion of movement is energy. The strategy used to conserve energy involves locomotion; to move from one place to another using generated propulsive force. The speed, efficiency, and endurance of locomotion are established through the combination of muscle mechanics and skeletal form.

By now, it can be seen that the MSK system appears to be designed so that energy is not wasted. In fact, because tendons can absorb and store elastic strain energy like a spring, its compliance can serve as an energy storage site to make the gait cycle itself more energy efficient [20]. In other words, energy is stored and released by tendons. The
muscle-tendon unit can then be said to transfer kinetic energy (KE) to potential energy (PE) where the metabolic cost is associated with activating the muscle and making the spring stiff. Changes in either mass or stiffness is referred to as tuning, to operate at a desired frequency. Such tuning occurs in humans during locomotion as well. This is a capability within the nervous system called recruitment and allows an individual to attempt to move in a manner that minimizes energy expended or its metabolic cost.

To predict passive and active muscle forces in biomechanical simulations, Hill-type muscle models shown in Figure 27 are commonly used. Such models are considered macroscopic because they predict muscle forces on an organ level and as a one-dimensional force output which is applied to skeletal models between origin and insertion points. Typically, Hill-type muscle models consist of three elements: a contractile element incorporating force–length and force–velocity dependencies, a serial and a parallel elastic element in diverse configurations. Various extensions account for physiologically observable effects, such as contraction history effects, recruitment patterns of slow-and fast twitch fibers, high frequency oscillation damping, or force in eccentric contractions [25]. Hill’s model and equation are discussed further in Chapter 3.
Figure 27 Three element Hill type muscle model as it relates to the force-length and force-velocity curves. $F^T$: Tendon Force, $F^M$: Muscle Force, CE: Contractile Element, and two non-linear spring elements, SE: Series Element, PE: Parallel Element [26].

As a closing remark to Chapter 1, future studies should exploit the concept of muscle energy storage in locomotion to benefit patients with movement disorders. However, this study will investigate the biomechanical response of the knee complex to a non-linear spring-loaded knee joint orthosis to quantify its effects in energy savings through muscle offloading during a specific patient’s flexion and extension parameters.
Chapter 2

Clinical Relevance

Now that the skeletal muscle structure and function of the knee has been reviewed, some common knee injuries that are associated with bracing as a potential treatment will be discussed. The controversy that exists amongst such treatment will also be highlighted by reviewing the effectiveness of knee orthopedics through clinical evidence.

2.1 Common Knee Injuries

Knee injuries account for 19 to 23% of all MSK injuries and are classified as either direct stress injuries, repeated stress injuries or as pathological syndromes of the knee [1, 27]. Below are a couple of common injuries from each category using known definitions and prognosis.

2.1.1 Direct Stress Injuries

2.1.1.1 Knee Sprains

When considering a direct stress injury, also classified as a traumatic injury, knee sprains are considered the most common. A knee sprain is when one or multiple knee ligaments are torn or stretched beyond its normal range. Each type of sprain is classified by a grading system: grade 1 describes a mild injury that causes only microscopic tears in the ligament; grade 2 describes a moderate injury in which the ligament is partially torn; and grade 3 describes a severe injury in which the ligament is completely torn through and surgery is required. According to Harvard Medical School of Health, the four major ligament sprains are listed below [28]:
• **Anterior cruciate ligament (ACL) sprain**—The ACL typically sprains during a sudden stop; a twist, pivot or change in direction at the joint; an extreme hyperextension; or a direct impact to the lateral aspect of the knee or lower leg. Although these injuries are seen among athletes in sports such as football, basketball, rugby, wrestling, gymnastics and/or skiing, 75% of ACL injuries occur without any direct contact with another player. Still, no matter how the sprain occurs, in most cases a full recovery for a grade 1 or 2 can take up to 4 to 12 months.

• **Posterior cruciate ligament (PCL) sprain** — A PCL sprain is caused from a direct impact to the anterior aspect of the knee, such as hitting the knee on the dashboard in a car crash or landing hard on a bent knee during sports. In athletes, PCL injuries are most common among those who play football, basketball, soccer and/or rugby. Like the ACL sprain, the PCL sprain can take up to 4 to 12 months to fully recover.

• **Medial collateral ligament (MCL) sprain** — The MCL can be torn or sprained by a direct sideways blow to the lateral aspect of the knee or lower leg. The kind of blow that can happen in football, soccer, hockey and/or rugby. The MCL can be injured by a severe knee twist or a fall that moves the shin, lower leg, laterally, away from the upper leg. If the sprain is a grade 1 or 2, the recovery time typically falls between 2 to 4 weeks.

• **Lateral collateral ligament (LCL) sprain** — An LCL is the least likely knee ligament to be sprained because most LCL injuries are caused by a blow to the
medial aspect of the knee during contact sports and that area is usually shielded by the opposite leg. However, if an LCL sprain occurs at a grade 1 or 2, recovery time for typically falls between 2 to 4 weeks.

The knee’s ligaments are closely related and therefore, when one ligament suffers a serious sprain there is a good chance that another may also be injured [1, 27]. For example, because the MCL helps to protect the ACL from certain types of extreme knee forces, the ACL can become vulnerable to injury when the MCL is torn. In more than half of moderate (grade 2) or severe (grade 3) MCL sprains, the ACL is also sprained. The known prognosis for knee sprains according to the Harvard Medical School of Health is about 90% of people with ACL injuries and 80% with PCL injuries can expect a full recovery after proper treatment and a good rehabilitation program involving orthopedic [29, 30]. However, as a long-term complication, some patients with Grade 3 ACL and/or PCL injuries may eventually develop symptoms of OA in the injured knee. For almost all MCL sprains and most LCL sprains have an excellent prognosis [28].

2.1.1.2 Meniscus Tear

Another common traumatic knee injury is a torn meniscus. The meniscus is a rubbery, C-shaped disc that cushions the knee. Each knee has two menisci; one lateral and one medial. These cushions prevent excess wear and tear inside the knee joint. When these disks are injured or torn, a patient has several options for treatment. Treatments can include nonsurgical solutions, surgery to repair the meniscus, or surgery to remove the meniscus. Such treatments are described below, again, according to the Harvard Medical School of Health [28]:
• Nonsurgical — includes temporary bracing of the injured knee and a rehabilitation regime to keep the muscles surrounding the knee strong while the knee is not bearing as much weight. This approach is most effective for small tears (5 millimeters or less) near the edge of the meniscus, where healing is usually good, or for people who are not good candidates for surgery.

• Reparative surgery — used if the tear is large (1 to 2 centimeters) but involves part of the meniscus where there is enough blood supply for healing. This is shown in Figure 28.

![Figure 28](image)

Figure 28 shows a surgery that repairs the meniscus tear using sutures [28].

• Partial meniscectomy surgery (surgery to remove part of the meniscus) — used if the tear involves part of the meniscus where blood flow and healing is poor. The surgeon may trim away ragged edges along the tear to allow the joint to move smoothly. A partial meniscectomy is shown in Figure 29.
Figure 29 shows a partial meniscectomy which is a surgery to trim away ragged edges along the tear to allow the joint to move smoothly [28].

For most patients the prognosis for a torn meniscus is very good. However, if the meniscus has been repaired surgically or partially removed, physical therapy will be required before resuming normal athletic activities. Although, most patients feel very satisfied with the results of their surgery, some patients eventually develop arthritis in the injured knee. Typically, arthritis develops many years after the injury and is due to either the damage done to the joint sustained from the original meniscus tear or from surgery itself. These surgeries often take away some or all of the cushioning effect of the meniscus [28].

2.1.2 Repeated Stress Injuries

Unlike direct stress injuries, repeated stress injuries occur over time caused by an overuse of the joint [27]. Such injuries are also classified as repetitive knee injuries. The most common are patellofemoral syndrome (PFS) known as runner’s knee, and knee tendonitis known as jumper’s knee [1, 31].

2.1.2.1 Patellofemoral Syndrome

PFS or Runner’s Knee is a disorder characterized by thinning, softening, and breaking down of the cartilage located under and around one or both patellae. PFS is
frequently diagnosed in adolescents and young adults with a greater frequency in females participating in high impact sports; specifically sports that involve running, jumping, sudden stopping and/or twisting that place abnormally high stress on the knee. Although PFS is the most common knee pathology, the direct cause of it is unclear [31, 32].

PFS’s leading symptoms include dull, aching pain in the anterior aspect of the knee, a grating or grinding feeling when the knee is bending and an increase in pain during walking. PFS is difficult to treat in the short term; however, it usually improves in the long term. Treatment options include weight loss; supportive knee braces to help protect and stabilize the knee joint and improve alignment of the kneecap; and physical therapy including rehabilitation exercises meant to stretch and strengthen the muscles of the knee. However, if these treatments fail to reduce or eliminate symptoms and pain, surgery may be necessary [31].

2.1.2.2 Knee Tendonitis

Jumper’s knee is a tendon injury seen in athletes at the point where the tendon attaches to bone. This means that it involves either the patellar tendon at the lower edge of the kneecap or the quadriceps tendon at the upper edge of the kneecap [28]. Jumper's knee refers to functional stress overload due to jumping or the level of a certain activity is increased too quickly. Overload can also occur through overuse meaning a particular body motion is repeated too often.

The cause of knee tendonitis is unknown but is a common MSK disorder affecting both recreational and elite athletes potentially leading to disability lasting several months [33]. Most treatments include 20 minutes of ice per day as well as resting the area for up to a month. Depending on the location and severity of tendonitis, temporary bracing and a
rehabilitation program may be necessary to regain strength, motion, and function. The length of time for rehabilitation varies depending on the type and severity of tendonitis however with proper treatment the current prognosis for the affected tendon usually recovers completely. Whereas an incomplete rehabilitation or a hasty return to activity can slow the healing process or lead to re-injury [28].

2.1.3 Pathological Knee Syndrome

Pathological syndromes are diseases that affect the knee joint [1]. The most common amongst these is osteoarthritis (OA), also known as the degenerative joint disease. Such disease is also the most common form of arthritis in the United States, affecting 15.8 million Americans [6, 34]. Furthermore, in the United States, approximately 9% of individuals aged 60 years and older suffer from knee OA [35] and more than 15% of women over the age of 80 [34].

Although there is no cure for OA, individualized treatment programs including non-surgical management, drug induced interventions and surgical options can help limit the loss of function, reduce pain, and maintain joint mobility [34]. Drug induced and surgical treatment options are beyond the scope of this work however, non-surgical management which can involve rest and physical therapy with biomechanical interventions such as bracing will be discussed. It should be understood that rest management is to be used in moderation. Merely a few days of muscle disuse can substantially reduce skeletal muscle mass and strength leading to long term muscle atrophy [36]. As we learned in Chapter 1, this can become dangerous as muscle provides support and stability to joints as well as unload joints and the surrounding cartilage.
In 2014 the Osteoarthritis Research Society International (OARSI) developed concise, patient-focused, evidence-based, expert consensus guidelines for non-surgical knee OA treatment options based off the RAND/UCLA approach; a methodology for measuring expert opinion to reach a classification for appropriateness of each treatment modality. A summary of treatments voted “Appropriate,” organized by clinical sub-phenotype are explained in detail throughout OARSI’s published guidelines [37]. Below, displayed in Figure 30, is the summary diagram of the appropriate treatments for specific OA types paired with the rationale behind the recommended appropriateness shown in Table 4.

Figure 30 Diagram of the appropriate treatments [37].
Table 4

Intervention recommendation and rationale summary provided by Osteoarthritis Research Society International [37].

<table>
<thead>
<tr>
<th>Treatment</th>
<th>Recommendation</th>
<th>Rationale</th>
<th>Quality of Evidence</th>
</tr>
</thead>
<tbody>
<tr>
<td>Exercise (Land-based &amp; Water-based)</td>
<td>Appropriate</td>
<td>Results were generally positive among land-based exercise type and did not significantly favor any specific exercise regimens. A 2007 SR investigating water-based exercise in knee and hip OA found small to moderate short-term benefits for function and quality of life, but only minor benefits for pain.</td>
<td>Good</td>
</tr>
<tr>
<td>Strength Training</td>
<td>Appropriate</td>
<td>Strength training programs primarily incorporate resistance-based lower limb and quadriceps strengthening exercises. Both weight-bearing and non-weight bearing interventions were included, as well as group and individual programs. Participants experienced similarly significant improvement with each of these programs.</td>
<td>Good</td>
</tr>
<tr>
<td>Biomechanical Intervention</td>
<td>Appropriate</td>
<td>Interventions are recommended as directed by an appropriate specialist. A 2011 SR and three recent RCTs evaluated the effectiveness of knee braces/sleeves in conservative management of knee OA. One review suggested that knee braces/sleeves were effective in decreasing pain, joint stiffness, and drug dosage and improved physical function with insignificant adverse events. The conclusions were limited due to the heterogeneity and poor quality of available evidence.</td>
<td>Fair</td>
</tr>
</tbody>
</table>

In 2018, the National Institute for Health and Care Excellence (NICE) published their own guidelines for management of OA, represented in Figure 31 below. However, shortly after this publication the OARSI conducted a pragmatic randomized controlled
trial implementing core NICE guidelines for OA in primary care with a model consultation (MOSAICS) that resulted in no evidence of benefits on the primary outcome of physical functioning at six months [38].

![Management of Osteoarthritis flow chart](image)

Figure 31 Management of Osteoarthritis flow chart [39]

The field of rehabilitation is one of controversial and subjective reports due to the heterogeneity and poor quality of available evidence [37]. In the following section and subsections, the controversy is seen to grow as common knee braces and their proposed interventions are reviewed.
2.2 Common Knee Braces

It is not surprising that the list of injuries and problems that can occur in the knee is extensive. As mentioned previously, the knee is required to bear large amounts of weight while remaining mobile. It is even less surprising to read the estimates of knee injuries account for up to 60% of all sport injuries, with the ACL accounting for almost half of these injuries. Such injuries can result in high healthcare costs, as an ACL injury is often associated with surgery, long and costly rehabilitation, differing degrees of impairment, and potential long-term consequences such as OA [40].

Undergoing surgery or noninvasive treatment methods, rehabilitation will most likely be necessary to allow the knee to regain normal mobility, no matter the injury sustained. In recent years, knee braces have become a more popular rehabilitation intervention to treat a variety of knee pain and injuries. The current market sales forecast for knee braces was valued at 1.4 billion USD in 2014 and expected to grow to over 2 billion USD in 2020 [41] This growth is largely due to the number of people with arthritis and other knee problems turning to knee braces for support or pain relief.

It is recognized by Medical Coverage Guidelines and Policies along with many others that knee braces can be divided into four categories: (1) Prophylactic — braces intended to prevent or reduce the severity of knee injuries in contact sports; (2) Functional — braces designed to provide stability for unstable knees during activities of daily living or vocational or avocational activities; (3) Unloader— braces specifically designed to reduce the pain and disability associated with OA of the medial compartment of the knee; and (4) Rehabilitation — braces designed to allow protected and controlled motion during the rehabilitation of injured knees [42-44]. Each type of brace is designed
to assist in a primary manner. The following sub-sections will describe each brace’s principle purpose and the latest studies surrounding its effectiveness and performance.

2.2.1 Prophylactic Knee Brace

Prophylactic knee braces are designed to prevent or reduce the severity of ligamentous injuries to the knee without reducing knee mobility [45]. This type of bracing is meant to offer protection to the MCL against valgus knee stresses, while shielding the knee complex to avoid re-injury and is therefore commonly used by athletes who play contact sports such as football. Conversely there is scarce evidence confirming the brace’s efficacy and the consensus on its effectiveness has not been established [44].

In 2000, Scott A. Paluska, M.D. and Douglas B. McKeag, M.D., M.S. out of the University of Pittsburgh Medical Center stated a prophylactic knee brace may offer a subjective sense of protection, but it is unable to protect the MCL during a direct lateral impact. Paluska and McKeag went on to state that the brace, at best, offers limited resistance to lateral knee impact and provides little meaningful rotational stress protection and at worst, can generate increased forces that augment associated injuries to the medial knee [44]. However, in 2009, after a literature review was conducted to provide a synopsis of the current understanding of knee bracing based on subjective and objective publications, it was announced no conclusions of this brace can be made due to limited evidence. The review did reveal that the limited evidence could be a result of fear of performance hindrance [40]. Many skilled players in football have voiced the concern that the brace will limit speed and agility. Therefore, these players tend to avoid routine brace wear in fear of performance limitations [44].
In 2011, five different prophylactic knee braces designed by the DonJoy brace company, were evaluated by five different therapists. Trial participants included 24 healthy subjects (14 men and 10 women) between the age of 18 and 22 with no prior history of lower limb injury and brace use. The trial included five different braces: 1) Hinged Trupull Advanced System; 2) Hinged ‘‘H’’ Buttress knee brace; 3) Butrs patella knee brace; 4) Drytex Lateral Patella knee brace; and 5) Drytex economy–hinged knee brace. The test protocol consisted of dynamic balance, jumping performance, proprioception, coordination, and maximal force. Results suggested brace 2 and 5 were the more effective choices. The study concluded that prophylactic knee braces can be used to enhance proprioception, coordination, maximal force, and balance but it is important to choose a proper brace based on the individual’s fitness level [46].

In 2016, Katie A. Ewing and associates investigated the changes in lower-limb muscle function with prophylactic knee bracing in athletes during landing. It was determined through motion analysis and subject-specific MSK modeling that the hamstrings and vasti muscles produce greater flexion and extension torques and greater peak muscle forces with a K300 mx Pod Orthotic brace. These findings suggest this type of prophylactic brace could be more effective in providing stability to the knee joint by increasing the active stiffness to the hamstrings and vasti muscles. Furthermore, results suggested that prophylactic brace usage is less important in MCL injury prevention than it is in strength training, conditioning, technique refinement and flexibility [44, 47].

Also in 2016, Jonathan K. Sinclair and associates investigated the effects of bracing on pain symptoms and patellofemoral loading in recreational athletes during a two week trial. Kinematics of the lower extremity, patellofemoral loading, and self-
reported knee pain scores were obtained during sport specific tasks including jog, cut, and single leg hop. Results showed significant reductions in the run and cut movements for peak patellofemoral force/pressure and in all movements for the peak knee abduction moment when wearing the DonJoy TriZone prophylactic knee brace. Significant improvements were also shown in pain reduction. Therefore, it was concluded that for recreational athletes who suffer from patellofemoral pain should be advised to utilize knee bracing as a conservative method to reduce pain symptoms [32].

Again in 2016, Jonathan K. Sinclair investigated the effects of knee bracing. However, this time knee joint kinetics and internal/external rotations in netball players was investigated. Results showed that there were no differences (p > 0.05) in knee joint kinetics during sport specific movements. Yet the internal/external rotation ROM was significantly (p < 0.05) reduced when wearing the DonJoy TriZone prophylactic knee brace in all movements. Interestingly, the subjective ratings of stability revealed that the netball players felt that the knee brace improved knee stability in all movements and did not hinder performance [45].

Based on the literature and above examples, it is clear that additional well-designed studies are needed to identify the proper role for prophylactic braces. Such a role should agree with the current evidence and common clinical recommendations [44].

2.2.2 Functional brace

The Functional brace is designed to provide stability by reducing rotation in the knee during activities of daily living and vocational or avocational activities. It is intended to provide support to an already injured knee and is most commonly prescribed to treat an ACL injury, or for those who need additional support after ACLR surgery
In some cases, the functional brace can be used to support mild to moderate PCL or MCL instability. However, after an extensive literature review spanning from 1980 to 2013 performed by Sean Smith and associates, it was found that current functional brace technologies do not sufficiently restore normal biomechanics to the anterior cruciate ligament deficient (ACLD) knee or protect the reconstructed ACL and does not improve long-term patient outcomes [48].

In 2003, Dan K. Ramsey and Gunnar Nemeth examined the neuromuscular response to functional bracing relative to anterior tibial translations in vivo using EMG data with simultaneous recorded skeletal tibiofemoral kinematics from four ACLD patients. Results suggested that the DonJoy Legend functional brace may only offer proprioceptive feedback rather than a mechanical stabilization [49]. However in 2005, it was found that the DonJoy Goldpoint functional brace increased peak abductor moments in ACLD knees and reduced bilateral kinetic symmetry in the coronal plane. It was also found that the brace increased peak moments and impulses of the abductors and extensors in ACLR knees. Such findings came from calculated three-dimensional joint moments and angular impulses at the knee using measured kinematic and kinetic data from 15 ACLD and 15 ACLR patients. Such finding led the team to conclude that functional bracing can be and should be recommended for both ALCD and ACLR patients [50].

In 2006, an article published in The American Journal of Sports Medicine also revealed promising results after conducting a study involving 257 skier-employees with ACLR whom wore CTi2 Innovation Sports functional knee braces and 563 skier-employees with ACLR whom did not. During a one-year screening, the braced skiers showed significantly higher preseason rates of grade 2 or higher Lachman and pivot-shift
tests (braced, 29% and 22%, respectively; non-braced, 11% and 10%, respectively; P = .05). Furthermore, the injury rate in the non-braced skiers was 9% while the injury rate in the braced skiers was 4%. For the injuries requiring surgery, the injury rate in non-braced skiers was 4% (25/563), and the injury rate in braced skiers was 1% (3/257). This study lists many limitations but still recommends functional bracing for ACLR skiers with evidence of increased laxity because without a brace, ACLR skiers are estimated to be almost 3 times more likely to sustain a subsequent knee injury [51].

In 2013, to provide insight to athletes who might refrain from wearing knee braces because of the fear of performance hindrance, Niyousha Mortaza and associates examined the effect of a customized functional knee brace upon the functional and isokinetic performance of ACLD and healthy subjects. Functional test consisted of crossover hop and single leg vertical jump whereas the isokinetic test was performed using Biodes Multi-joint System 3 dynamometer. The team found that neither of the examined groups showed sufficient negative or positive effect on knee performance, force control, or force generation capacity of the knee joint. However, they did find small positive effect of the brace on the peak torque and average power in the ACLD group indicating that functional braces can reduce the bilateral asymmetry in ACLD patients [52]. Despite the results, most patients who use functional knee braces still report subjective improvements that may exceed objective measurements of knee stability, pain attenuation, performance enhancement and confidence during athletics which may impart a false sense of confidence [44].

In 2016, a systematic review published in the Journal of Sport Rehabilitation resulted in inconsistent evidence of functional knee bracing on joint-position
improvement after ACLR even though physicians often provide a functional knee brace to the patients’ returning to physical activity in order to assist their lack of balance ability and to reduce the risk of retear of the reconstructed ACL [53]. However today’s physicians, just like Paluska, Scott A., and Douglas B. McKeag back in 2000, claim that functional knee braces deserve consideration as a component of the treatment and rehabilitation for ligamentous knee instability because they offer some control, as seen in laboratory tests using cadavers or surrogate leg models that demonstrated limitations of tibial rotation and anteroposterior translation, of external knee rotation and anteroposterior joint translation [44]. Though, Paluska, Scott A., and Douglas B. McKeag also stated that these effects rapidly diminished during physiologic stress loads. As we can see, more research is needed to provide evidence on the biomechanical effects of a functional knee brace that define its correct application [48, 53].

2.2.3 Unloader (off-loader) Knee Brace

The unloader knee brace, sometimes called off-loader or patellofemoral knee brace, is specifically designed to reduce the pain and disability associated with OA of the medial compartment of the knee and is often prescribed to patients suffering from unicompartmental knee OA. It is called the “unloader” because it is said to unload the compressive forces on the medial compartment to improve tibiofemoral alignment by bracing the knee in the valgus position [54-56]. Some researchers refer to the brace as a mechanical intervention designed to reduce pain, improve physical function, and possibly slow disease progression improving the quality of life [54]. However, the mechanical efficacy provided by the brace is lacking in evidence as shown in the discussion below.
In 2000, Brain P. Self and associates reported that a custom Monarch valgus loading knee brace designed for patients with medial compartment OA reduced the varus moment demonstrating the biomechanical function of the brace in five subjects. The brace consisted of a unilateral structure with a lateral double axis hinge. Additionally, a lateral condylar pad with a variable pressure air bladder was incorporated allowing the patient to control the amount of desired pressure. The team’s suggested results came after analyzing results from calculated forces and moments at the knee of five subjects diagnosed with medial compartment OA during gait trials. Calculations were based off data collected by a three-dimensional video-based motion analysis system and from force plate information. The data showed that the custom Monarch brace specifically reduced the varus moment at 20% and 25% of stance and that the valgus force measured with the brace remained fairly constant throughout the first 80% of the stance phase [57].

In 2009, Ramsey and Russell provided a general survey of the evidence documented in the scientific literature concerning the efficacy of unloader knee braces for improving symptomatology associated with painful disabling medial compartment knee OA. The survey collectively showed evidence that off-loader braces are effective in mediating pain relief in conjunction with knee OA and malalignment. This suggests that bracing should be the first treatment used before joint realignment or replacement surgery is considered. However, the survey concluded that more work is needed to substantiate bracings long-term benefits, given that patient compliance is an issue [54, 58]. Patients reported that the brace is difficult to wear for extended periods because of the degree of force imparted on the limb to alter alignment. More evidence is also needed to effectively understand the kinematic effects imposed by the brace in order to have a more
concise comparison across investigations. At the time of this survey, kinematic evidence showed highly controversial results. Amongst such results, the positive ones showed significant decrease in varus angulation of the knee; a reduction in peak adduction moments during gait and stair stepping; and condylar separation of the medial compartment with the use of bracing. The negative results discussed in the survey concluded no substantial changes in the femoral-tibial angle or joint space; no significant change in the adduction moment; and no apparent gait adaptations were observed. A summary of the survey’s results regarding pain, function, and instability as well as kinematic and kinetic data provided by Ramsey and Russell [54].

In 2013, a pilot study investigating the effect of varus and valgus adjustments to a commercially available OAsys (Össur) unloader knee brace on knee biomechanics during gait in populations with normal knee alignment found no differences between the braced and non-braced conditions in peak or mean knee adduction moments nor in the EMG parameters including total muscle activation and mediolateral-directed co-contraction ratio results. The materials and methods used in this pilot study consisted of seven-camera VICON motion analysis system synced with AMTI force plates and 16-channel tethered, double differential EMG sensors. The collected data was then exported into a subject-specific MSK model which revealed results that did not support the use of unloader braces in subjects with normally aligned knees, particularly if the purpose of the brace is to reduce the knee adduction moment and/or alter muscle activation patterns to reduce compartmental knee load [59].

However, in 2014, Richard J. Steadman and associates conducted a literature search to review the current state of unloading braces on knee mechanics, clinical impact,
and long-term disease progression. They found through a PubMed MEDLINE database including articles published since 1980, that the brace does improve OA symptoms in parameters such as pain, instability, and quadriceps strength. However, it was also found that the current literature suggests a debate regarding the effectiveness in biomechanical parameters and the hindrance of disease progression [55, 56].

Also in 2014, a study investigating the effects of bracing on knee adduction moment during stair negotiation was performed on thirty male and female participants with painful patellofemoral OA. Participants ascended and descended a seven-step staircase both with and without an unloading knee brace. The joint moments were calculated through inverse dynamics techniques by combining kinematic data form motion analysis and GRF data while pain was assessed using analogue scale. The study resulted in the braced condition showing reduced maximal flexion angle moment, the total range of motion at the knee and the internal peak knee extension moment in the sagittal plane when compared to the unbraced condition. However, there was no difference in pain reported between conditions for either stair ascent or descent. Out of speculation, the authors suggested the mechanism explaining subtle changes in knee kinematics and kinetics was related to a greater perception of joint stability with use of the brace [60].

In 2015, a pilot study conducted by a team from Australia evaluated the immediate and four-week effects of a DonJoy unloader knee brace on knee-related symptoms and performance-based function in people with knee OA after ACLR gave positive results [61]. The study showed that the individuals involved immediately felt more confident wearing both types of braces; unadjusted brace providing sagittal plane
support, and adjusted brace providing sagittal plane support with varus/valgus readjustment. Following the four-week brace intervention, individuals again felt that the allocated brace improved knee confidence, perceived task difficulty and stability suggesting that the unloader knee brace, adjusted or unadjusted, has the potential to improve knee-related symptoms associated with knee OA after ACLR. This pilot study concluded that the unloader brace has the potential to improve knee-related symptoms associated with knee OA after ACLR [61]. The pilot study came to its results using methods of Wilcoxon signed-rank tests.

Yet another review which assessed the benefits and harms of braces in the treatment of patients with knee OA showed that wearing a brace may result in little to no difference in reducing pain or improving knee function. Furthermore, the study showed no difference in the quality of life after 12 months. However, the evidence at this time was graded a low-quality. This review concluded that the optimal choice for an orthosis remains unclear, and long-term implications are lacking [35].

Although it seems that the unloader brace reduces pain associated with OA in the knee, discomfort and side effects have been reported such as thrombophlebitis of the lower limbs. Furthermore, the lack of evidence in the braces biomechanical support and tibiofemoral alignment correction suggest more randomized clinical trials concerning bracing in knee OA are still necessary [35, 55, 56]. However, with the number of OA patients predicted to increase as the population ages, a reduction of patient morbidity for this widespread chronic condition in combination with this treatment modality is still believed to have a positive impact on health care costs and the economic productivity and quality of life of the affected individuals [54, 62].
2.2.4 Rehabilitation Knee Brace

The rehabilitation brace, sometimes referred to as a postoperative brace is prescribed in an attempt to decrease the recovery period or rehabilitation protocol by restraining the motion of the knee in some way [63]. This type of brace is also used to protect a reconstructed/repaired ligament and allow for early controlled motion [29]. However, just as seen in prophylactic, functional, and unloader knee braces, the true efficacy of the rehabilitation brace has yet to be fully evaluated or understood as studies show mixed results to its usefulness [64, 65].

In 2001, a study conducted by Eve Möller and associates showed absolutely no difference in the rehabilitation progress or healing results after two years when comparing patients who wore a postoperative DonJoy L.E.S. rehabilitation knee brace versus those who did not [63]. The study included 62 patients who participated in the rehabilitation program either with or without the postoperative brace for six weeks following bone-tendon-bone Anterior Cruciate Ligament Reconstruction (ACLR). Within the non-braced group, one patient experienced a partial rupture of the graft after a new trauma one year after surgery. There were four surgeons involved. Although each surgical method was said, by literature, to yield the same stability, two of the surgeons performed an arthroscopic ACLR while the other two used a modified technique. During each follow-up, evaluations of laxity, muscle and functional performance, ROM and knee circumference, and Lysholm score, Tegner activity level, and visual analogue scale was conducted. Results yielded no significant difference in knee laxity, neither in muscle and functional performance, nor in ROM and knee circumference between the groups.
However, the Tegner activity score showed significant difference after six months in favor of group A but then no difference at year two [63].

In 2007, a systematic review of bracing following ACLR was preformed using electronic databases AMED, Cinahl, Cochrane database, Embase, Medline (via Ovid), Physiotherapy Evidence Database (PEDro) and Pubmed from their inception to August 2006. After the review, there appeared to be no significant longer-term differences in clinical outcomes between patients who wore postoperative knee braces and those who did not [64]. Unsatisfied by these and other negative results, many scientists, including design engineers from Horton’s Orthotic Lab Inc. and NASA as well as engineers from the University of Auckland and Queens University, found it of upmost importance to improve the brace design in such ways that truly allow the injured patient to regain their mobility in a more energy-efficient manner; to be effective in assisting with building muscle strength during everyday use of the knee brace; and to reduce the overall rehabilitation timeline [65, 66].

Horton’s Orthotic Lab Inc. and NASA took a crack at a new design with their Stance-Control Orthotic Knee Joint (SCOKJ) brace aimed for people with weak quadriceps muscles due to polio, spinal cord injuries, and other conditions such as unilateral leg paralysis. SCOKJ’s offers benefits which replicate the knee’s natural gait through the use of a selectively lockable joint that operates in three distinct but complementary modes: free motion or automatic stance control, for walking, and manual lock, for standing. Their design concept was built upon the traditional knee-ankle-foot orthoses (KAFOs) which either locked or unlocked knee joints depending on the level of stability required [67]. The SCOKJ entered the commercial realm back in 2002.
To investigate the biomechanical and energetic effects of stance-control orthoses, a study was performed on nine nondisabled adults who walked with KAFOs that incorporated the Horton SCOKJ in all three modes [66]. The study in general showed that the kinematics for the auto and unlocked modes were more similar than for the auto and locked modes indicating that the SCOKJ allows subjects to walk with a more normal gait pattern. Surprisingly and against the researchers’ hypothesis, it was found that the gait in the auto mode did not yield lower oxygen cost compared with the locked mode [66]. However, it should be noted that this study was based on persons with no known knee injury and therefore these persons may have greater muscle function and strength and greater joint ROM than a typical KAFO user probably would have. In turn, this may also cause an error in data collection since this may have influenced a number of aspects of the study, including control of the ankle joint, walking speed, and step length variability.

While in 2017, the University of Auckland and Queens University designed a soft-stop knee brace for rehabilitation from ligament injuries specifically for ACLD patients to provide resistive spring-damper forces at the ROM limits and to decrease the chance of jarring. The new brace was tested using a Vicon motion capture system to assess the kinematic parameters resulting in positive feedback. The dampers increased the ROM by up to 6° and furthermore the subject indicated that he felt more comfortable with the soft limits created by the resistive forces and more confident during a full ROM exercise. Unfortunately, the brace is in a redesign phase due to the stress and cyclic failure of the dampers [65] and is not currently available for public use.

Again, rehabilitation braces are meant to limit movement or lock the knee in a given position after surgery. However, in 2003 a prospective controlled study examined
subjects who underwent ACL Bone-Patellar Tendon-Bone reconstruction. The study’s results suggested that the application of a rehabilitation brace in full extension during the first postoperative week is more effective in recovering full extension mobility of the operated knee than the application of the rehabilitation brace locked from $0^\circ$ to $90^\circ$ [30]. During this study both groups followed the same rigorous, accelerated rehabilitation protocol. Also, at the beginning of the second post-operative week the brace was unlocked ($0^\circ$ to $120^\circ$) and then removed at the beginning of the third week for both groups. Each patient was evaluated pre and post-operatively with bubble-level heel height difference measurements (HHD) and KT 1000 arthometric assessment at the fourth postoperative month. Preoperative bubble-level HHD measurements for both groups were similar however the four and eight-week follow-ups showed that the extension of the operated knees of Group B was greater than in Group A. However, the KT 1000 arthometric scores showed no difference between the two groups.

After reading the reviews current surveys of each brace, it can be clearly seen that more evidence-based trials are needed to end the controversial topic and to provide a worthwhile solution to a large problem; a problem that is imposing a significant financial burden on the healthcare system. Therefore, the following Chapter will discuss a possible method for evaluating knee braces based on biomechanical responses of a subject-specific’s joint kinematics and muscle response throughout a squat movement to a non-linear spring-loaded KJO.
Chapter 3

Methodology

To investigate the biomechanical responses of the knee complex to a non-linear knee joint orthosis (KJO), a measured kinematic 3-D motion capture analysis subjected to kinetic parameters during two separate trials (brace-off and brace-on) collected at Embry-Riddle Aeronautical University’s Motion Capture Laboratory was used to drive a subject-specific (SS) MSK model developed within OpenSim simulation software. The results from this type of analysis gave promising insight into the KJO’s performance when compared to its designed parameters specifically its effect on joint range of motion (ROM) and muscle force.

The following sections highlight the methodology used for data collection followed by the technical development of the modeling and simulation protocol.

3.1 Trial Movement

As seen throughout Chapter 2, orthopedic companies including the one in this trial, suggest that KJO provide assistance to the quadriceps muscles during knee extension without limiting ROM. To investigate this, a squat exercise ranging from 0° knee flexion to no less than 115° knee flexion was performed during two trials: 1) brace-off: when the subject performs the movement without the KJO, and 2) brace-on: when the subject performs the movement while wearing the KJO.

The squat exercise was chosen because it is an integral part of strength and conditioning programs designed to increase lower body strength, muscle mass, and bone mineral density [68]. It is also appropriate and commonly used in knee rehabilitation settings [69]. Furthermore, the squat is considered one of the best exercises for improving
quality of life because of its ability to recruit multiple muscle groups in a single maneuver [70]. Thus, the study of muscle forces during this activity is of importance in the design of internal prostheses and orthosis [69].

The primary joint and muscle actions that occur during the squat exercise can be divided into two phases: 1) eccentric (lowering), and 2) concentric (lifting). During the eccentric phase, the hip extensors, knee extensors, and plantarflexor muscles contract as they lengthen. These same muscles are activated in the concentric phase but now shorten as they contract [71]. During this movement the quadriceps femoris and gluteus maximus are referred to as target muscles while the hamstrings muscles and the soleus act as synergists. Figure 32 highlights the target muscles in red and the synergist muscles in purple.

Figure 32 Target muscles (red) and synergist muscles (purple) during a squat exercise.
Scientific research has established a force-velocity relationship, as discussed in Chapter 1, which describes the impact of external forces on a muscle’s concentric and eccentric shortening velocities [72]. This identifies cadence as a great influencer on inertial forces experienced by the MSK system throughout a movement. (i.e. A faster cadence will result in greater inertial forces per Newton’s second law [68]. However, this is not the only way to influence the net force experienced by the MSK system. Specifically during a squat, inertial forces can be increased by increasing the knee flexion angle. It has been shown that the vertical component of GRF is highest during a deep squat when compared to a semi and half-squat exercises [73].

Therefore, with these techniques on how to influence the inertial force magnitude and with efforts to keep the subject safe, the deep body weight squat performed at a comfortable steady cadence for five repetitions was chosen [70]. The deep body weight squat is defined as knee flexion being greater than 100 degrees and performed for about 3 seconds. This depth and cadence ensures the subject remains balanced throughout the movement while providing large enough GRFs that the dynamic force plate will read the acceleration values. A five-minute minimum rest period between trials was mandated to ensure fatigue was not a factor [74].

3.1.1 Subject

A healthy male (71.6 kg; 1803 mm tall) who was familiarized with squat techniques, signed the written consent form and volunteered to participate in the present study. The subject, free of all known MSK disorders, performed five repetition squats without the KJO (brace-off) and five repetition squats with the KJO (brace-on). The
subject performed each trial barefooted to limit the effects of shoes on the ground reaction forces as seen in Figure 33.

For the present study, the “free of all known MSK disorders” condition was mandated. This is because joint properties vary between individuals, particularly in the case of injury or pathology and can have significant effect on joint angles, and internal loads, as well as muscle moment arms [75].

Furthermore, this subject-specific or single-subject study was chosen due to the driving huge interest in 'precision' medicine. Recognition that physicians need to take individual variability into account is being more and more popular [76]. Classical clinical trials harvest a handful of measurements from thousands of people which led to mistreating many. However the recognition of individual variability, with the well-known limitations of classic clinical trials to predict patient outcomes, has shed light on the growing need to refocus on single person studies. Single-subject clinical trials use observations from a single patient to establish efficacy or side effect profiles. The
renewed interest in single-subject clinical trials stems from an observed lack of universality in response to interventions and a greater focus on the individual with the emergence of precision medicine and treatment [77].

Physicians have long done single-subject clinical trials these in an ad hoc way. For instance, a doctor may prescribe one drug for hypertension and monitor its effect on a person's blood pressure before trying a different one. But few clinicians or researchers have formalized this approach into well-designed trials — usually just a handful of measurements are taken, and only during treatment [77].

Transforming routine patient care into massively parallel single-subject clinical trials requires a shift in the mindset of regulatory agencies, researchers, physicians, and pharmaceutical companies. For these reasons, the method described in the upcoming sections could be utilized allowing physicians and orthopedic specialist to move toward individual treatments by proscribing the most effective KJO for a specific patient.

3.1.2 Non-Linear Spring-Loaded Knee Joint Orthosis

The non-linear spring-loaded KJO used in this study is shown in Figure 34. The brace was custom fitted to the subject and is considered to be capable of providing the objectives of both a functional and rehabilitation knee brace.

This KJO features a Varying Radius Spring (VRS) technology which allows for the non-linear spring system to operate through stored energy. In other words, the VRS stores energy created by a bending moment during flexion and then releases said energy during extension. This technology claims that the release of stored energy may result in offloading the quadriceps muscular activation, requiring less exerted energy from the wearer during extension.
To achieve this, it is assumed the VRS system’s lower hinge of the brace applies a load to the spring element that gradual increases the input force required to bend the spring as the angle of knee flexion increases. As seen in Figure 35 the spring engages in a linear behavior from 0° to 15° ROM in knee flexion then becomes non-linear.

![Figure 34 Custom 3-D printed non-linear spring-loaded knee joint orthosis used in this study.](image)

Figure 34 Custom 3-D printed non-linear spring-loaded knee joint orthosis used in this study.

![Figure 35 Force versus angle results for bending test of VRS technology.](image)

Figure 35 Force versus angle results for bending test of VRS technology.
3.2 Measured Kinematic and Kinetic Data

Kinematic data was captured using the VICON Motion Capture System while kinetic data was collected using an AMTI OR6-6-OP force plate located under the subject’s right foot. An additional AMTI filler plate was utilized to level the ground located under the left foot as shown in Figure 36. Muscle activation was measured using Delsys EMG surface sensors placed on the subject’s right Gluteus Maximus (GMX), Gluteus Medius (GMD), Semitendinosus (ST), Bicep Femoris (BF), Vastus Lateralis (VL), Vastus Medialis (VM), Rectus Femoris (RF), and Soleus (SOL). VICON NEXUS’s proprietary software for motion analysis synced all data in the time domain.

Figure 36  Embry-Riddle Aeronautical University’s Motion Capture Laboratory.

3.2.1 Experimental Kinematic Description

Embry-Riddle Aeronautical University’s Motion Capture Lab is outfitted with a VICON Motion Capture System consisting of four Vero v2.2 cameras mounted on
tripods and 12 Vantage V5 (eight wide lens and four standard lens) mounted along the ceiling’s parameter. Cameras were properly calibrated before each trial in coherence with VICONs calibration wand recommendations [78]. Furthermore, the camera’s frame rates were synced to the lesser of the two cameras (i.e. 330 Hz). Parameters for each of these cameras are listed in Table 5.

Table 5

VICON camera parameters for the V5 and v2.2 models.

<table>
<thead>
<tr>
<th>Model</th>
<th>V5</th>
<th>v2.2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolution (MP)</td>
<td>5</td>
<td>2.2 (2048 x 1088)</td>
</tr>
<tr>
<td>Max Frame Rate (Hz)</td>
<td>420 @ 5MP</td>
<td>330 @ 2.2MP</td>
</tr>
<tr>
<td>Max Frame Rate (Hz)</td>
<td>2000</td>
<td>330</td>
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<tr>
<td>On-Board Marker Processing</td>
<td>Yes</td>
<td>-</td>
</tr>
<tr>
<td>Standard Lens</td>
<td>12.5 mm</td>
<td>6.5-15.5 mm Varifocal</td>
</tr>
<tr>
<td>Wide Lens</td>
<td>8.5 mm</td>
<td>6.5-15.5 mm Varifocal</td>
</tr>
<tr>
<td>Minimum Standard FOV (H x V)</td>
<td>48.5 x 41.2</td>
<td>44.1 x 23.6 (tele)</td>
</tr>
<tr>
<td>Minimum Wide FOV (H x V)</td>
<td>63.5 x 55.1</td>
<td>98.1 x 50.1 (wide)</td>
</tr>
<tr>
<td>Camera Latency</td>
<td>4.7 ms</td>
<td>3.6 ms</td>
</tr>
<tr>
<td>Power</td>
<td>PoE</td>
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<tr>
<td>Max Power Consumption</td>
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</tr>
<tr>
<td>Dimensions (mm) (H x W x D)</td>
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<td>83 x 80 x 135</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td></td>
<td>0.57</td>
</tr>
</tbody>
</table>
To capture the movement and achieve quality results, 22 reflective markers were placed on the leg ipsilateral of the right KJO and pelvis in sets of three or more to define each body segment [79].

Specifically, the markers were placed on the left and right (R) anterior superior iliac spine (ASIS), V. Sacral, R upper lateral thigh, R anterior and posterior thigh, R medial thigh, R lower lateral thigh, R upper lateral and medial knee, R lower lateral and medial knee, R anterior and posterior tibia, R lower lateral and medial tibia, R upper lateral and medial tibia, R lateral and medial ankle, R heel, and R toe tip. This marker set defined the pelvis and the right femur, tibia, talus, calcaneus, and toe body segments.

![Marker placements on test subject and the KJO during brace-on trial.](image)

3.2.2 Ground Reaction Forces

A ground reaction force (GRF) consists of three components: one vertical ground reaction force (VGRF) and two shear forces. The VGRF is the sum of gravitational and inertial forces acting on a subject in the direction of motion. The gravitational force remains constant throughout the movement since it is only dependent on mass whereas
the inertial force is variable since it depends on the total mass of the system plus its acceleration. The VGRF measurement gives insight into how the foot sole bears the support forces during the movement performance [68]. The shear forces act parallel to the ground and brakes or accelerates forward motion of the center of mass. Such external forces give an indication of the total force experienced by the subject during a movement [72]. This concept is illustrated in Figure 38. The GRF point of application is called the center of pressure (COP) and is the most important factor that affects joint moments and forces during lifting [68].

GRF and the COP are measured by a force plate. The force plate uses instrumented pedestals that contain transducers located at the corners of the plate to measure plate’s deformation. With the known force plate geometry and pedestal locations, the forces and moments acting on the plate can then be determined.

Figure 38 Coordinates of a typical ground reaction force (GRF) measurement system [80].
AMTI's OR6-6-OP-2K-CTY 2000 series force plate specifically assesses external forces along three perpendicular axes and moments about the associated axes as well as the average COP with an accuracy of a fraction of a millimeter. This plate has a high sensitivity and low crosstalk less than 2% on all channels; a sampling rate of 1000 Hz; a 10 V maximum excitation; and is designed for analysis of athletic performance, running and jumping. Detailed specifications are shown in Table 6.

To give a level, stable platform and nonbiased environment a filler plate was paired collinearly to the force plate as seen in Figure 39.
### Table 6

AMTI Force Plate specifications for OR6-6 series.

<table>
<thead>
<tr>
<th>OR6-6 Series Specifications</th>
<th>1000</th>
<th>2000</th>
<th>4000</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fz Capacity, lb (N)</td>
<td>1000</td>
<td>2000</td>
<td>4000</td>
</tr>
<tr>
<td></td>
<td>(4450)</td>
<td>(8900)</td>
<td>(17800)</td>
</tr>
<tr>
<td>Fx, Fy Capacity, lb (N)</td>
<td>500</td>
<td>1000</td>
<td>2000</td>
</tr>
<tr>
<td></td>
<td>(2225)</td>
<td>(4450)</td>
<td>(8900)</td>
</tr>
<tr>
<td>Mz Capacity, in*lb (Nm)</td>
<td>5000</td>
<td>10,000</td>
<td>20,000</td>
</tr>
<tr>
<td></td>
<td>(600)</td>
<td>(1100)</td>
<td>(2300)</td>
</tr>
<tr>
<td>Mx, My Capacity, in*lb (Nm)</td>
<td>10,000</td>
<td>20,000</td>
<td>40,000</td>
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<tr>
<td></td>
<td>(1100)</td>
<td>(2300)</td>
<td>(4500)</td>
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<tr>
<td>Fz Natural Frequency, Hz</td>
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<td>1000</td>
<td>1000</td>
</tr>
<tr>
<td>Fx, Fy Natural Frequency, Hz</td>
<td>400</td>
<td>550</td>
<td>800</td>
</tr>
<tr>
<td>Fz Sensitivity, ( \mu V/[V*Nlb] )</td>
<td>0.75</td>
<td>0.38</td>
<td>0.19</td>
</tr>
<tr>
<td></td>
<td>(0.17)</td>
<td>(0.08)</td>
<td>(0.04)</td>
</tr>
<tr>
<td>Fx, Fy Sensitivity ( \mu V/[V*Nlb] )</td>
<td>3.0</td>
<td>1.5</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td>(0.67)</td>
<td>(0.34)</td>
<td>(0.17)</td>
</tr>
<tr>
<td>Mz Sensitivity, ( \mu V/[V<em>N</em>lb] )</td>
<td>0.38</td>
<td>0.19</td>
<td>0.09</td>
</tr>
<tr>
<td></td>
<td>(3.38)</td>
<td>(1.69)</td>
<td>(0.85)</td>
</tr>
<tr>
<td>Mx, My Sensitivity, ( \mu V/[V<em>N</em>lb] )</td>
<td>0.18</td>
<td>0.09</td>
<td>0.05</td>
</tr>
<tr>
<td></td>
<td>(1.59)</td>
<td>(0.79)</td>
<td>(0.39)</td>
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<tr>
<td>Height, in (mm)</td>
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<td>3.25</td>
<td>4.0</td>
</tr>
<tr>
<td></td>
<td>(82.5)</td>
<td>(82.5)</td>
<td>(102)</td>
</tr>
<tr>
<td>Weight, lb (kg)</td>
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<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(18)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Top Plate Material</td>
<td>composite</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
3.2.3 Electromyography (EMG)

Muscles and tendons are the interface between the CNS and the articulated body segment. To estimate muscle force, the level of EMG activity can be used and then, through comparison with other EMG signals, the role of each muscle in the movement can be assessed. In other cases, the force-length and force-velocity properties of muscle, as well as the EMG activity, are qualitatively considered in the estimation of forces [81].

Muscle activation in the current research was measured using the Delsys Inc Wireless System. A trigger module (Delsys, Natick, MA) used to synchronize the EMG data with VICON’s motion data, was connected directly into the Lock (VICON, Denver, CO) via RCA connection, as seen in Figure 40.

A Trigno Avanti Docking Station (Delsys, Natick, MA), also seen in Figure 40, interfaces with the trigger module allowing for the wireless EMG sensors to send a digital signal to the computer running at an RF frequency band of 2400-2483 MHz. The docking station maintains a range of 40 meters, but for the current study it is limited to the range
of the Delsys Trigno Mini EMGs at 20 meters; the EMG sensors have a sampling rate of 1926 samples per second.

Figure 40 Delsys Trigno Avanti EMG docking station and trigger module interface (Delsys, Natick, MA).

Eight Delsys Electromyography (EMG) surface sensors were placed on the subjects right Gluteus Maximus (GMX), Gluteus Medius (GMD), Semitendinosus (ST), Bicep Femoris (BF), Vastus Lateralis (VL), Vastus Medialis (VM), Rectus Femoris (RF), and Soleus (SOL) as described in Figure 41. Proper preparation of the tests subjects’ leg which included a shaved surface removed of excess dirt or skin using alcohol wipes and tape ensured each EMG sensor connection. These particular muscles were chosen based on the KJO functionality and the muscles activated during a squat exercise.
To normalize the activation data a Maximum Voluntary Contraction (MVC) Test using EMGworks Acquisition (Delsys, Natick, MA) was utilized. This required a MVC
for each muscle. The contraction movements used for the lower extremity muscles previously described are listed in Table 7.

Table 7

Maximum voluntary contraction (MCV) movements used for each muscle.

<table>
<thead>
<tr>
<th>Maximum Voluntary Contraction Movements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gluteus Maximus</td>
</tr>
<tr>
<td>Gluteus Medius</td>
</tr>
<tr>
<td>Semitendinosus</td>
</tr>
<tr>
<td>Bicep Femoris</td>
</tr>
<tr>
<td>Vastus Lateralis</td>
</tr>
<tr>
<td>Vastus Medialis</td>
</tr>
<tr>
<td>Rectus Femoris</td>
</tr>
<tr>
<td>Soleus</td>
</tr>
</tbody>
</table>

Out of the systems mentioned above, the VICON vero 2.2 cameras has the lowest sampling rate of 330 Hz therefore, each system will sync in the time domain at 330 Hz. (VICON V5 cameras: 420 Hz; AMTI force plate: 1000 Hz; and Delsys EMG sensors: 1928 Hz)
The above kinematics and kinetic data was used as inputs into the MSK simulated model. The following sections will discuss the technical development of the modeling and simulation protocol.

3.3 Three-Dimensional Musculoskeletal Modeling: OpenSim

For the present study, OpenSim (NCSSR, Stanford, Palo Alto, CA) was used for three-dimensional MSK modeling and dynamic simulations. OpenSim is an open source project allowing researchers full access to all models and incorporating a repository for sharing of newly developed models. OpenSim’s models are made up of seven parts: bodies, joints, constraints, forces, markers, contact geometry, and controllers [83]. These parts are shown in Figure 42. The first five will be reviewed as they pertain to this research.

*Bodies* are rigid segments that represent the skeletal (bone) part of the model. The bodies are interconnected by *joints* that permit motion. There are several types of joints to describe the connection such as biological joints, specifically the knee joint which uses splines to describe the translation of the tibia with respect to the femur as a function of knee flexion; or weld joints that fuse two bodies together and has no coordinates. There are also, amongst several others, pin joints which have one coordinate about the common Z-axis; ball joints which have three rotational coordinates; and slider joints which contain one coordinate along common X-axis. *Constraints* are then put in place to control or limit the motion of the bodies. Constraints can be applied in translation and/or rotation or act as a coordinate coupler constraint which relates the generalized coordinate of a given joint to any other coordinates in the model.
Connected to the body is muscle. Muscles are modeled as a specialized force element described as a line segment between an origin on one body segment and insertion on another. In some cases, intermediate via points are included to represent the wrapping of muscle around other structures like bones, other muscles, or retinacula [79]. Muscle force is depended on the muscle model structure. This concept is defined more in the next section. In addition to muscle forces, a variety of other forces which represent external applied forces, passive-dampers, and controlled linear and torsional actuators can be defined [83]. Also rigidly connected to bodies are markers referred to as virtual markers as opposed to experimental markers, whose location is expressed in local coordinates. In order to perform inverse kinematics, the virtual marker set must match the experimental marker set used to collect motion capture data. This can be seen in Figure 51 and later explained in section 3.3.2.

![Figure 42 Opensim Musculoskeletal components visualization](image)

Figure 42 Opensim Musculoskeletal components visualization [84].
This type of MSK modeling provides a non-invasive means to study human movement and predict the effects of interventions on exercises such as the squat exercise used in the current study. The goal of this study is to use OpenSim’s GAIT 2392 MSK model to predict the impacts of a KJO during knee flexion and extension. Specifically looking at ROM, joint moments, and individual muscle force comparison between two trials: brace-off and brace-on by means of an inverse dynamics analysis. The inverse flow used is shown in Figure 50 and will be covered in detail in section 3.3.

This particular OpenSim GAIT 2392 MSK model was chosen because it was designed for simulations of movements which are leg dominated and best represents joints and muscle groups that are of most interest to this study without adding complexity of unrelated parts such as the lumbar or upper extremities [83].

3.3.1 Hill’s Muscle Model Structure

Most models, including the one used by OpenSim, use Hill’s hyperbolic equation to characterize muscle performance. Such equation was once thought to be an empirical one that lacked precision in predicting velocities at high and low loads for many decades however it is now recognized as more than just descriptive.

In 2013, Chun Y. Seow wrote a review published in the Journal of General Physiology that described the connection between the historic Hill equation illustrating the relationship between shortening velocity and load to characterize muscle performance, and the kinetics of myosin cross-bridge cycle based on the latest findings on myosin motor interaction with actin filaments within the structural confines of a sarcomere. Such connection is illustrated through a comparison between a normalized form of the Hill equation and an equation derived from actomyosin kinetics. This
comparison, with associating assumptions, is described in detail throughout Chun Y. Seow’s review [85]. Therefore, the following section will review the model’s elements without defending its accuracy.

The Hill type model utilized within OpenSim consists of three elements that predict passive and active muscle forces (Figure 43). More specifically, the contractile element (CE) predicts the active muscle force and incorporates force-length and force-velocity dependencies. Maximum isometric force, the optimal muscle-fiber length and the pennation angle are parameters that describe the specific muscle characteristics. The parallel elastic element (PEE) represents the elastic structures surrounding the muscle and predicts the passive force depending on the CE length. The serial elastic element (SEE) represents tendon where its force is equal to the sum of PEE and CE forces.

Figure 43 Relationship between the 3-element Hill type model and anatomical muscle. The parallel elastic element (PEE) represents the elastic structures surrounding the muscle and predicts the passive force depending on the CE length. The serial elastic element (SEE) represents tendon where its force is equal to the sum of PEE and CE forces.
3.3.2 OpenSim’s Musculoskeletal Model GAIT2392

The current research references OpenSim’s GAIT 2392 model; one of the core models in OpenSim developed mainly by Darryl Thelen from the University of Wisconsin-Madison and Ajay Seth, Frank C. Anderson and Scott L. Delp from the University of Stanford. The GAIT 2392 model was adopted from the Delp model and was designed for simulations of movements which are leg dominated featuring 23-degree-of-freedom and 92 musculotendon actuators to represent 76 muscles in the torso and lower extremities including pelvis, femur, tibia, fibula, talus, foot, and toes [86]. This model is shown in Figure 44.

Figure 44 OpenSim’s musculoskeletal model GAIT 2392 featuring 23-degree-of-freedom and 92 musculotendon actuators to represent 76 muscles in the torso and lower extremities including pelvis, femur, tibia, fibula, talus, foot, and toes.
For each body segment there is a fixed reference frame as seen in Figure 45. OpenSim describes these frames as follows:

- **Pelvis**: The pelvic reference frame is fixed at the midpoint of the line connecting the two anterior superior iliac spines;

- **Femur**: The femoral frame is fixed at the center of the femoral head

- **Tibia**: The tibial frame is located at the midpoint of the line between the medial and lateral femoral epicondyles

- **Patella**: The patellar frame is located at the most distal point of the patella

- **Talus**: The talar frame is located at the midpoint of the line between the apices of the medical and lateral malleoli

- **Calcaneus**: The calcaneal frame is located at the most interior, lateral point on the posterior surface of the calcaneus

- **Toe**: The toe frame is located at the base of the second metatarsal

Furthermore, models of the hip, knee, and ankle, subtalar, and metatarsophalangeal joints define the relative motions of these segments.
3.3.2.1 Joint modeling

The primary joints namely the hip, knee, and ankle joints are described as follows per OpenSim documentation [83]:

**Hip Joint**

The hip joint is characterized as a ball-and-socket joint. The transformation between the pelvic and femoral reference frame is thus determined by successive rotations of the femoral frame about three orthogonal axes fixed in the femoral head.

**Knee Joint**

In the Gait 2392 model, the insertions of the quadriceps on the tibia are modeled as moving points in the tibial frame. The geometry for determining knee moments and kinematics in the sagittal plane is seen in Figure 46. From these kinematics, the moment
of the quadriceps force about the instant center of knee rotation can be computed [86].

The Gait2392 model does not model the knee as a conventional 1 DOF hinge (pin) but rather as a custom joint. This custom joint’s effective center of rotation changes as a function of flexion (the one generalized coordinate) and has functions for translations in the X-Y plane of the joint (tx, ty).

Figure 46 Geometry for determining knee moments and kinematics in the sagittal plane. $\theta_k$ is the knee angle; $\phi$ is the patellar ligament angle; $\beta$ is the angle between the patella and the tibia; $F_q$ is the quadriceps force; and $l_{pl}$ is the length of the patellar ligament [86].

The femoral condyles are represented as ellipses, and the tibial plateau is represented as a line segment. The transformation from the femoral reference frame to the tibial reference frame is specified such that the femoral condyles remain in contact with the tibial plateau throughout the range of knee motion. The tibiofemoral contact point
depends on the knee angle and is specified according to data reported by [87]. This concept is shown in Figure 47.

![Figure 47 Schematic of the human knee joint. Due to the rolling and sliding of the non-circular femoral condyles (oval fixed in the femur, parent P) on the tibia plateau (body, B), the joint does not operate as a simple pin. In this model, the tibia has one rotational degree-of-freedom, $\theta$, but translates in the plane of rotation (x,y) plane of the joint [88].](image)

**Ankle, Subtalar, and Metatarsophalangeal Joints**

The ankle, subtalar, and metatarsophalangeal joints are modeled as frictionless revolute joints (as seen in Figure 48). When displayed within the simulation, the axes produce realistic motion of the ankle and subtalar joints (i.e. the bone surface models do not collide or disarticulate), but exhibit unrealistic motion of the metatarsophalangeal joint (i.e. the phalanges separate from the metatarsals). To fix this problem, the
metatarsophalangeal axis is rotated by – eight degrees on a right-handed vertical axis to minimize disarticulation of the joint.

Figure 48 Ankle, subtalar (ST), and metatarsophalangeal (MTP) joints are modeled as revolute joints with axes oriented as shown (Delp et al., 1990).

3.3.2.2 Muscle-tendon modeling

The paths or the lines of action of the muscle-tendon actuators in the lower extremity portion of the model are defined based on anatomical landmarks. Each muscle-tendon path is represented by a series of line segments. Muscles can be sufficiently described as a single segment from origin to insertion. However, if the muscle wraps over bone or is constrained by retinacula, intermediate via points known as “wrapping points” are introduced to represent the muscle path more accurately. For instance, because the quadriceps tendon wraps over the distal femur when the knee is flexed beyond 80°, additional wrapping points are defined for the knee flexion angles greater than 80
degrees. This will allow the quadriceps tendon to wrap over the bone, instead of passing through it [83].

Strength properties, specifically maximum contraction forces, are based primarily on cadaver muscle cross-sections while scaled to better reflect Anderson and Pandy's model and the joint torque-angle relationships measured on healthy, living subjects. This method of "strength scaling" is outlined in Anderson and Pandy and the Yamaguchi [89, 90]. The scaling is uniformly applicable for all muscles at a joint except for bi-articular muscles as they span two joints. Furthermore, in some cases Anderson and Pandy's muscle strength parameters were used instead, as they are more physiologically accurate [83]. Included in Appendix A are the isometric muscle forces from Gait2392, Delp1990, and Carhart2000, along with the scale factors provided by OpenSim.

3.3.2.3 Customized GAIT 2392 model

To accommodate the current study OpenSim's GAIT 2392 model was simplified by removing the torso and the left femur, tibia, fibula, talus, foot, and toes (Figure 49). These simplifications were made to accommodate the single force plate data, avoiding misrepresentation of the left external forces. This resulted in a model containing six body segments (keeping the Pelvis, and the right Femur, Tibia, Talus, Calcaneus, & Toes) and 43 muscles (originally 12 body segments and 76 muscles). Such simplifications did not change the MSK original model or structure of the pelvis and/or the right lower extremity (femur, tibia, fibula, talus, foot, and toes).
3.4 Inverse Analysis: Pipeline SimTRACK

OpenSim offers a modeling and simulation framework through a standardized pipeline called Simtrack that allows the user to run dynamic simulations using experimental data to estimate parameters such as joint moments and muscle force. The tailored pipeline used for this research is shown in the Figure 50.

Starting with a generic model (i.e. customized gait 2392) and experimental static kinematics as inputs, the pipeline will utilize four main tools: 1) scale tool: scales the generic model according to the participant’s anthropometrics and experimental static kinematic trial; 2) inverse kinematics (IK) tool: calculates joint kinematics using an experimental dynamical trial; 3) inverse dynamics (ID) tool: estimates general muscle forces through an ID approach; and 4) static optimization (SO) tool: an extension of the

Figure 49 Custom GAIT 2392 model featuring six body segments (keeping the Pelvis, and the right Femur, Tibia, Talus, Calcaneus, & Toes) and 43 muscles (originally 12 body segments and 76 muscles).
ID tool that further resolves general forces into individual muscle forces utilizing the residual reduction algorithm (RRA) module.

Figure 50 SimTrack (inverse analyses) pipeline from OpenSim [86].

3.4.1 Step 1: Scaling

The Scale Tool scales the generic model (i.e. customized GAIT 2392) according to the participant’s anthropometrics and experimental static kinematic trial. OpenSim allows the user to achieve this through a combination of manual scaling and/or measurement-based scaling. Manual scaling uses segment lengths that are known from medical imaging information such as CT scans or MRIs whereas measurement-based scaling uses the distances between experimental markers measured during a static trial compared to the equivalent distances between virtual marker pairs. Measurement-based scaling is shown in Figure 51 and is the method of choice for this study.
Figure 51 Experimental marker positions (dark blue) and virtual markers, placed on a model in anatomical correspondence (pink) [83].

Next, the scale tool will scale the generic mass and inertia values, segment lengths, tendon slack length, and optimal muscle fiber length to “best” match the anthropometrics of the subject using body mass and the experimental marker trajectories as captured during a static trial with the subject in the upright squat-ready position. In other words, the scale tool will “best match” the static pose that minimized a sum of weighted squared errors of markers and/or coordinates.

The default, unscaled version of the GAIT 2392 model represents a subject that is 1.8 meters in height; weighting 75.16 kg. To match the subject in the current study, the model was scaled to 1.803 meters in height and 71.6 kg of mass with the static kinematic file filtered with a low pass filter at 6 Hz. The virtual marker set loaded on the generic
custom model (Figure 52) matched that of the experimental marker placement described earlier in section 3.2.1. The body segments were defined using marker pairs as described in Table 8 and Table 9.

Figure 52 Custom marker model used in OpenSim; L=left, R=right, ASIS=anterior superior iliac spine, med=medial, lat=lateral.
Table 8

Body segments defined using marker pairs and scale factors for brace-off trial.

<table>
<thead>
<tr>
<th>Body Name/Measurement Used</th>
<th>Marker pairs</th>
<th>Applied Scale Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis/ Pelvis</td>
<td>R.ASIS, L.ASIS</td>
<td>1.066</td>
</tr>
<tr>
<td>Femur_R/ Femur_R</td>
<td>R.ASIS, R.Upper.Knee.Lat</td>
<td>1.048</td>
</tr>
<tr>
<td>Tibia_R/ Tibia_R</td>
<td>R.Upper.Knee.Lat, R.Ant.Lat</td>
<td>1.050</td>
</tr>
<tr>
<td>Talus_R/ Unassigned</td>
<td>n/a, n/a</td>
<td>1.0</td>
</tr>
<tr>
<td>Calcn_R/ Calcn_R</td>
<td>R.Heel, R.Toe.Tip</td>
<td>0.936</td>
</tr>
<tr>
<td>Toes_R/ Unassigned</td>
<td>n/a, n/a</td>
<td>1.0</td>
</tr>
</tbody>
</table>

Table 9

Body segments defined using marker pairs and scale factors for brace-on trial.

<table>
<thead>
<tr>
<th>Body Name/Measurement Used</th>
<th>Marker pairs</th>
<th>Applied Scale Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis/ Pelvis</td>
<td>R.ASIS, L.ASIS</td>
<td>1.068</td>
</tr>
<tr>
<td>Femur_R/ Femur_R</td>
<td>R.ASIS, R.Upper.Knee.Lat</td>
<td>1.092</td>
</tr>
<tr>
<td>Tibia_R/ Tibia_R</td>
<td>R.Upper.Knee.Lat, R.Ant.Lat</td>
<td>1.008</td>
</tr>
<tr>
<td>Talus_R/ Unassigned</td>
<td>n/a, n/a</td>
<td>1.0</td>
</tr>
<tr>
<td>Calcn_R/ Calcn_R</td>
<td>R.Heel, R.Toe.Tip</td>
<td>0.933</td>
</tr>
<tr>
<td>Toes_R/ Unassigned</td>
<td>n/a, n/a</td>
<td>1.0</td>
</tr>
</tbody>
</table>
3.4.2 Step 2: Inverse Kinematics (IK)

IK computes joint coordinates (joint angles) that best replicate the marker position history. In other words, the IK tool goes through each time step (frame) of motion and computes generalized coordinate values which positions the model in a pose that “best matches” the experimental model. Mathematically, the "best match" is expressed as a weighted least squares problem, whose solution aims to minimize both marker and coordinate errors [83]. This is expressed mathematically through the weighted least squares equation which is solved by IK and is shown below.

\[
\min_q \left[ \sum_{i \in \text{markers}} w_i \|X_i^\text{exp} - X_i(q)\|^2 + \sum_{j \in \text{unpr} \text{scribed \ coords}} w_j (q_j^\text{exp} - q_j)^2 \right] \tag{8}
\]

Where:

- \( q \): vector of generalized coordinates being solved for,
- \( x_i^\text{exp} \): experimental position of marker \( i \),
- \( x_i(q) \): position of the corresponding model marker
- \( q_j^\text{exp} \): experimental value for coordinate \( j \). *Prescribed coordinates are set to their experimental values.
- \( w_i's \): marker weights
- \( \omega_j's \): coordinate weights

Experimentally-measured kinematic data (from the dynamic trial) is used as an input into the IK tool. Total root mean square (RMS) and maximum marker errors are
reported as “messages” after the tool has finished running. Errors indicate how much the experimental markers differ from the virtual markers and are an indication for the quality of the scaling process. These error values, according to OpenSim should be less than 2-4 cm. However, these guidelines vary depending on the nature of the model and the motion being examined [83]. Nevertheless, the values can be used to guide changes in marker placement and possibly led to a rerun of the scale tool.

A motion file containing the generalized coordinate trajectories (joint angles and/or translations) computed by IK is provided as an output file and later used as an input file to the inverse dynamics tool. This motion file provides a visual feedback as shown in Figure 53.

![Computed IK results from experimental motion capture data.](image)

**Figure 53** Computed IK results from experimental motion capture data.

### 3.4.3 Complimentary Step: Inverse Dynamics (ID)

Based on kinematics describing the movement of a model and kinetics (GRF applied as a point force of the right calcaneus) applied to the model, the ID tool
determines the generalized net forces and torques at each joint responsible for a given movement by mathematically solving the equations of motion [83]. The classic equation of motion is shown in (9. In this equation the left hand side is made up of knowns gathered from the motion of the model while the right hand side is unknown generalized forces and torques.

\[ M(q)\ddot{q} + C(q, \dot{q}) + G(q) = \tau \]  

(9)

Where:

- \( q, \dot{q}, \ddot{q} \): vectors of generalized positions, velocities, and accelerations, respectively;
- \( M \): system mass matrix;
- \( C \): vector of Coriolis and centrifugal forces;
- \( G \): vector of gravitational forces;
- \( \tau \): vector of generalized forces.

The dynamic kinematic input file has an option of a low pass filter at 6 Hz meant to mitigate noise and errors associated with data collection. Since noise is amplified by differentiation the calculated forces and torques will also be noisy without a filter. For that reason, the filter was opted for this study.

It is worth mentioning that the moment calculated from the ID tool uses a moment arm that is different than the muscle moment arm described in section 1.5.3. The method is the same however the moment arm, \( r \), differs. The muscle moment arm, as
stated previously, is the perpendicular distance from the muscle’s angle of application, \( \theta \), to the center of rotation (joint axis). Whereas the moment arm in ID is the perpendicular distance from the external force’s point of application (reaction at the foot) to the joint center of rotation. The ID moment arm is shown in Figure 54 highlighting the knee joint moment.

![Diagram of knee joint with forces](image)

Figure 54 Moments and forces calculated in part by a force plate located under the foot in the sagittal [88].

3.4.4 Step 3a: Static Optimization (SO)

Static Optimization (SO) is an extension of the ID tool which further resolves the general forces into individual muscle forces that generate the accelerations seen in the input kinematics [83]. The muscle forces are resolved by minimizing the sum of squared (or other power) muscle activations. At each time step the tool will update the model
coordinates, external loads, and calculate the muscle forces and activations needed to
generate the model's accelerations or in other words to satisfy Newton’s 2nd law.

\[ F = m X a \]  \hspace{1cm} (10)

Where:

\( F \): force;
\( m \): mass;
\( a \): acceleration

More specifically, the tool uses the known motion of the model to solve the
equations of motion for the unknown generalized forces (e.g., joint torques) constrained
by force-length-velocity properties.

\[ \sum_{m=1}^{n} [a_m f(r_m^0, l_m, v_m)] r_{m,j} = \tau_j \]  \hspace{1cm} (11)

while minimizing the objective function:

\[ J = \sum_{m=1}^{n} (a_m)^p = \tau_j \]  \hspace{1cm} (12)
Where:

- \( n \): number of muscles in the model;
- \( a_m \): activation level of muscle \( m \) at a discrete time step;
- \( F_m^0 \): its maximum isometric force;
- \( l_m \): its length;
- \( v_m \): its shortening velocity;
- \( f(F_m^0, l_m, v_m) \): its force-length-velocity surface;
- \( r_{m,j} \): its moment arm about the \( j \)th joint axis;
- \( \tau_j \): the generalized force acting about the \( j \)th joint axis; and
- \( P \): a user defined constant.

Note that for static optimization \( f(F_m^0, l_m, v_m) \) computes the active fiber force along the tendon.

However, due to experimental errors and modeling assumptions, kinematics is not always dynamically consistent with the ground reaction forces, indicating the existence of residual forces. If this is the case, residual forces will be needed to drive the model to track the given kinematics; satisfying Newton’s second law. This means that the force term in Equation (10 really has two components: expected force plus residual force (Equation 13). To add residuals to the model, the RRA module is used. The RRA module and process is described in section 3.3.5.
\[ F_{expected} + F_{Residual} = m \times a \]  \hspace{1cm} (13)

Where:

- \( F_{expected} \): expected force
- \( F_{Residual} \): residual force
- \( m \): mass
- \( a \): acceleration

The static optimization tool allows the user to use either IK results or RRA results (RRA will be discussed in the upcoming section) as an input file. As seen in ID, if the user chooses IK data as the input, a filter should be applied. Filtering the kinematics will spline-fit the motion data and filter the coordinate positions, reducing noise in the accelerations [83]. However, if the user inputs RRA results, filtering is not needed.

If the model's actuators (muscles) cannot generate sufficient forces to satisfy the model's coordinate accelerations, static optimization will fail. The static optimization tool will also produce a set of results to help with troubleshooting, but these outputs are not valid [83]. To solve this error the user will need to run the RRA module to add sufficient actuators. This tool is discussed in the next section.

3.4.5 Step 3b Residual Reduction Algorithm (RRA)

When SO fails the model may need refinement. Model refinement is achieved through residual reduction using the residual reduction algorithm (RRA) module. The tool will find a new set of model parameters that reduces the residual forces creating an adjusted model that can be used in SO. The tool setup requires kinematic data with a low pass 6 Hz filter option; a tracking task file that specifies which coordinates the tool will
track and the corresponding tracking weight. The tracking file can also be used to specify any constraints on the RRA actuators; a drop down option allowing the user to define which body center of mass (COM) the RRA will adjust (i.e. the pelvis); an actuator input file which contains residual and/or reserve actuators; as well as an external load motion file applied to the model as a point or body force.

To clarify, residuals actuators are placed on the base segment and whose force accounts for the errors (kinematics that do not balance with experimental GRFs) in the model whereas, actuators placed on the coordinates are called reserve actuators and are used the drive the model when the model excludes its muscles (i.e. RRA) [83].

After each run of the RRA module, a new adjusted model file will be loaded where the COM of the pelvis has been automatically adjusted. The RRA module also provides a set of estimated changes to the mass and the COM for all of the segments which would further reduce the residuals needed to match the experimental kinematic and force data. However, these suggested changes are left up to the user to change and are not automatically done.

The user can perform several iterations of running RRA and adjusting the model until the residual forces and tracking errors have stabilized. Once the last iteration is complete and the user is satisfied with the results, the adjusted model will satisfy Newton’s 2nd law and can be used as an input into the static optimization tool.

In the current study, the residuals were high due to the high degree of flexion required by the deep squat. As noted 3.3.2, this model was originally designed for gait analyses where the hip is not required to reach such extreme flexion. To minimize the residuals and to satisfy Newton’s 2nd law, the RRA module was run twice before a
suitable adjusted model that the SO tool would accept, was produced. The residual error report and results will be further discussed in Chapter 4.

3.4.6 Validation & Verification

Validation and verification was done using OpenSim’s protocols [75]. However, even with the OpenSim experts, growing community, and over 25 years of experience in developing MSK modelling and simulation in the field of movement science, a well-defined verification and validation process for new models is still lacking. Such a shortcoming allows a negative implication in the clinic and a wider impact on healthcare. Furthermore, this can also negatively impact to fidelity of the MSK models.

For this reason, I have chosen to implement an open source model that has already been tested and validated to help create a custom model and simulation that answers my research question. In Chapter 5, I will further validate my process and results through comparisons with independent experiments and other models, however, this approach can be problematic. Direct validation is only possible with invasive methods and these cannot be applied in a clinical setting. Nevertheless, good quality experimental data can give first indications about which muscles are active providing a starting point that can be compared to EMG excitations [75].

Although EMG excitations are not directly comparable with muscle force estimations in the magnitude of force, EMG muscle excitations provide a validation tool for the temporal characteristics of muscles estimated by mathematical models. This approach has been used in several studies [91]. However, when using this validation technique the limitations of EMG need to be recognized while interpreting results. The current study will address EMG limitations in Chapter 4.
Chapter 4

Results

Joint angles, moments, and forces obtained from dynamic trials were applied to an adjusted custom GAIT 2392 model, and a static equilibrium problem was solved at each instant during the squat cycle to find individual muscle forces as described in Chapter 3. Position errors for each of the model’s generalized coordinates and the average residual values (Figure 63 and Figure 64) were used to tune the model.

Direct comparisons between the brace-off and brace-on trials are made to provide insight into how the KJO affects knee biomechanics and muscle forces. Comparisons of the brace-off trial are also made with relevant literature for model verification and validation. A summary of average joint position and moments can be found in Table 11 while a summary of muscle forces can be found in Table 13.

The comparative graphs presented below follow OpenSim’s sign convention in the sagittal plane as follows: ankle dorsiflexion (DF) and plantarflexion (PF) are positive and negative values respectively; knee extension (E) and knee flexion (F) are positive and negative values respectively; and hip extension and flexion are negative and positive values respectively.

The subject was asked to perform each squat at a self-selected cadence and therefore the time domain between each trial is not identical. For this reason, Table 10 identifies the trial movement from the starting (upright) position to full knee flexion (bottom) to end (upright) position in the time domain represented in seconds.
Table 10

Identifies a position within the squat movement paired with its corresponding time (s) for each repetition. This is done for both brace-off and brace-on trials.

<table>
<thead>
<tr>
<th>Position in the Squat Cycle</th>
<th>Time (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Squat 1</td>
</tr>
<tr>
<td>-----------------------------</td>
<td>---------</td>
</tr>
<tr>
<td><strong>Brace-off</strong></td>
<td></td>
</tr>
<tr>
<td>Start: Max Knee Extension</td>
<td>7</td>
</tr>
<tr>
<td>60° knee Flexion (descent)</td>
<td>8.1</td>
</tr>
<tr>
<td>90° knee Flexion</td>
<td>8.5</td>
</tr>
<tr>
<td>Max Flexion</td>
<td>9.4</td>
</tr>
<tr>
<td>60° knee Flexion (ascent)</td>
<td>10.3</td>
</tr>
<tr>
<td>Total Squat time</td>
<td>3.7</td>
</tr>
<tr>
<td><strong>Brace-on</strong></td>
<td></td>
</tr>
<tr>
<td>Max Extension</td>
<td>3.4</td>
</tr>
<tr>
<td>60° knee Flexion (descent)</td>
<td>4.1</td>
</tr>
<tr>
<td>90° knee Flexion</td>
<td>4.5</td>
</tr>
<tr>
<td>60° knee Flexion (ascent)</td>
<td>6.0</td>
</tr>
<tr>
<td>Max Flexion</td>
<td>5.1</td>
</tr>
<tr>
<td>Total Squat time</td>
<td>3.3</td>
</tr>
</tbody>
</table>

4.1. Joint Range of Motion

The average magnitude in sagittal plane angular movement suggests that the KJO affects the ankle, knee, and hip joint ROM, specifically at the extremes (i.e., when the
knee is fully extended at the neutral upright position and when the knee is fully flexed at the bottom of the squat). Such changes are described in the subsections below.

4.1.1 Ankle ROM

The ankle complex is comprised of the talocrural and subtalar joints. This study is particularly interested in the former since the articulation is of the tibia and fibula with the talus. Normal talocrural ROM is 20° dorsiflexion and 50° plantar flexion. However, a higher degree of mobility at this joint is required to facilitate balance and control in both the ascent and descent of the deep squat exercise [70].

Literature reported that for a squat the average ROM for DF with extended knee is 8.23° ± 5.29° (males) and 11.54° ± 5.58° (females) [92]. While an average ROM for DF at 60° knee flexion was reported to be 20° ± 6° and at 80° knee flexion was reported to be 40° ± 10° [93].

The average PF seen in this study was 1° ± 1.33° and 2.5° ± 1.30° without and with the KJO respectively. It was also observed that minimal to no change occurred between trials in DF at 60° knee flexion (average 22.42° ± 2.15° WOB, 25.49° ± 1.80° WB) during descent or at full knee flexion (average 39.8° ± 0.75° WOB, 39.4° ± 1.10° WB). These results suggest that the KJO has a small influence on PF and almost no influence on DF.

These angles suggest that the subject in the current study has a large anterior weight shift at the pelvic, pushing the knee over the toes during descent. Such weight shift is possible to see in the recorded videos and could have been induced by instructions given to the subject. To maintain balance, the subject was asked to keep contact between the heel of the foot and the floor.
Figure 55 Differences between brace-off (red) and brace-on (dashed-blue) ankle joint range of motion in the sagittal plane. Results show a 1.5° increase in PF at the top of the squat (Average 0.98° ± 1.33° WOB; 2.5° ± 1.30° WB), a 0.40° decrease in DF at the bottom of the squat (average 39.8° ± 0.75° WOB, 39.4° ± 1.10° WB).

4.1.2 Knee ROM

The knee joint consists of the tibiofemoral joint, which carries out sagittal plane angular movement from 1.7° to 0° of extension and from 0° to approximately 160° of flexion [70, 93, 94]. The subject in the current study performed each brace-off squat at an average knee flexion of 133° ± 1.61° and each brace-on squat at an average of 125° ± 4.00°. It is possible that this limitation was due to the physical size and placement of the brace on the leg and not a consequence of the non-linear spring influence.

During both trials, the subject was observed to have a slight degree of knee flexion in the upright position specifically on average 4.5° ± 1.10° WOB and 0.5° ± 1.25° WB. This suggests that the KJO creates a 4° ± 0.15° reduction in knee flexion.
during the upright position encouraging a more neutral behavior. No change in either trail was observed in knee flexion ranges seen in day to day activities [95].

![Average Knee ROM](image)

Figure 56 Differences between brace-off (red) and brace-on (dashed-blue) knee joint range of motion in the sagittal plane. Results show a 4° reduction in knee flexion at the top of the squat (Average 4.5° ± 1.1° WOB, 0.5° ± 1.25° WB) and a 8° reduction in knee flexion at the bottom of the squat (Average 133° ± 1.61° WOB, 125.1° ± 4.0° WB).

4.1.3 Hip ROM

The hip articulation is between the head of the femur and the acetabulum of the os coxae, carrying out flexion and extension in the sagittal plane. It has been reported that the mean hip range of motion during squatting is to be 95° ± 27° of flexion [70]. More specifically, the average hip flexion ROM reported in males is 113.87° ± 9.76° and 120.34° ± 7.70° in females [92].

Both reports match the hip ROM observed in this study where the average hip flexion during the brace-off and brace-on trials was an average of 110° ± 1.04° and 103°
\[ \pm 2.13^\circ \] respectively. Such results show that the KJO decreases hip flexion at the bottom of the squat by 6%.

Interestingly, hip flexion at full knee extension was \(4^\circ \pm 3.10^\circ\) and \(0.33^\circ \pm 1.15^\circ\) during brace-off and brace-on conditions respectively. This suggests that the pelvis experiences a posterior translation during brace-on conditions furthering suggesting that the KJO could prevent hyperextension conditions during the upright position.

![Figure 57](image)

Figure 57 Differences between brace-off (red) and brace-on (dashed-blue) hip joint range of motion in the sagittal plane. Results show a 3.5° reduction in hip flexion in the upright position (Average \(4^\circ \pm 3.10^\circ\) WOB, \(0.3^\circ \pm 1.15^\circ\) WB) and a 6.6° difference in hip flexion at the bottom of the squat (Average \(110^\circ \pm 1.04^\circ\) WOB, \(103.5 \pm 2.13^\circ\) WB) at flexion.

4.2 Joint Moments

The generalized net torques determined through inverse dynamics explained in section 3.3.3, show the greatest torque moment occurring at the knee when compared to the ankle and hip joints for both trials as seen in Figure 58 and Figure 59.
Figure 58 Average ankle, knee, and hip joint moments comparison in the sagittal plane during the brace-off trial.

Figure 59 Average ankle, knee, and hip joint moments comparison in the sagittal plane during the brace-on trial.
Again, looking at the average magnitude at maximum knee extension and flexion, results suggest that the non-linear KJO affects the torque produced at each joint. The changes at the extremes are seen and described in the following subsections followed by a detailed discussion in Chapter 5.

4.2.1 Ankle Moment

The ankle complex contributes significant support and aids in power generation during squat performance. However, kinetic data of the ankle during squatting is limited because most studies have focused on the biomechanics of the knee, hip, and spine [70].

This study observed a 4.8 Nm difference in torque at the ankle joint during full knee extension and a 9.8 Nm difference at full knee flexion during the brace-on trial when compared to the brace-off trial.

![Average Ankle Moment](image)

Figure 60 Brace-off (red) and Brace-on (blue) ankle joint moment comparison in the sagittal plane. 4.8 Nm decrease in torque at full knee extension (Average -21.1 ± 10.81 Nm WOB, -16.2 Nm ± 6.82 WB) and a 9.8 Nm decrease in torque at full knee flexion (-22.4 ± 5.05 Nm WOB, -12.6 ± 6.02 Nm WB)
4.2.2 Knee Moment

Assisting the tibiofemoral joint is the patellofemoral joint, a gliding joint in which the patella slides over the trochlear surfaces of the femur during knee flexion and extension. This provides additional mechanical leverage in extension because of a greater force arm, as well as reducing wear on the quadriceps and patellar tendons from friction against the intercondylar groove [70].

This trend agrees with the data shown in Figure 61 where torque increases as the subject descends into the deep squat producing the largest values at maximum knee flexion with small variance between trials.

![Average Knee Moment](image)

Figure 61 Brace-off (red) and Brace-on (blue) knee joint moment comparison in the sagittal plane. 0.57 Nm decrease in torque at full knee extension (Average -10.8 ± 5.29 Nm WOB, -11.4 ± 4.13 Nm WB). 5.81 Nm increase in torque at full knee flexion (Average 77.0 ± 2.72 Nm WOB, 82.9 ± 2.25 Nm WB).
4.2.3 Hip Moment

During the squat, hip torque increases in conjunction with hip flexion, with maximal torque occurring near or at the bottom of the squat [70]. This trend was observed as hip torque magnitude increased from 6.5 Nm at full knee extension to 56.2 Nm at the bottom of the squat during brace-off conditions. This was also the case for brace-on, as the torque magnitude increased from 1.9 Nm to 36.8 Nm.

Comparing the trials, is it seen in Figure 62 that the KJO reduces the torque by 4.7 Nm during the upright position and by 19.4 Nm at the bottom of the squat.

Figure 62 Brace-off (red) and Brace-on (blue) hip joint moment comparison in the sagittal plane. 4.7 Nm decrease at full knee extension (Average -6.5 ± 2.87 Nm WOB, -1.9 ± 1.9 Nm WB) and 19.4 Nm decrease at full knee flexion (Average -56.2 ± 6.26 Nm WOB, -36.8 ± 1.59 Nm WB).
Table 11

A summary of average joints positions (deg) and joint moments (Nm).

<table>
<thead>
<tr>
<th>Joint</th>
<th>Position (deg) at Full Knee Extension (Start Position)</th>
<th>Position (deg) at 60° Knee flexion (Descent)</th>
<th>Position (deg) at Full Knee Flexion (Bottom Position)</th>
<th>Position (deg) at 60° Knee flexion (Ascent)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Average (WOB)</td>
<td>Average (WB)</td>
<td>Average (WOB)</td>
<td>Average (WB)</td>
</tr>
<tr>
<td>Ankle</td>
<td>0.98 ± 1.33 (PF)</td>
<td>2.52 ± 1.30 (PF)</td>
<td>22.42 ± 2.15 (DF)</td>
<td>25.49 ± 1.80 (DF)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td>4.45 ± 1.10 (F)</td>
<td>0.50 ± 1.25 (F)</td>
<td>52.58 ± 3.41 (F)</td>
<td>58.18 ± 3.78 (F)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>4.0 ± 3.10 (F)</td>
<td>0.33 ± 1.15 (F)</td>
<td>32.28 ± 1.83 (F)</td>
<td>37.62 ± 3.76 (F)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Joint Moment (Nm) at Full Knee Extension (Start Position)</td>
<td>Joint Moment (Nm) at 60° Knee flexion (Descent)</td>
<td>Joint Moment (Nm) at Full Knee Flexion (Bottom Position)</td>
<td>Joint Moment (Nm) at 60° Knee flexion (Ascent)</td>
</tr>
<tr>
<td></td>
<td>Average (WOB)</td>
<td>Average (WB)</td>
<td>Average (WOB)</td>
<td>Average (WB)</td>
</tr>
<tr>
<td>Ankle</td>
<td>21.1 ± 10.81</td>
<td>16.2 ± 6.82</td>
<td>4.71 ± 4.08</td>
<td>19.44 ± 5.08</td>
</tr>
<tr>
<td>Knee</td>
<td>10.78 ± 5.29</td>
<td>11.4 ± 4.13</td>
<td>46.02 ± 6.76</td>
<td>54.4 ± 5.32</td>
</tr>
<tr>
<td>Hip</td>
<td>6.53 ± 2.87</td>
<td>1.87 ± 1.90</td>
<td>16.94 ± 1.73</td>
<td>19.44 ± 2.77</td>
</tr>
</tbody>
</table>
4.3 Residual errors

Residual reduction algorithm (RRA) module is primarily intended for gait analysis. However, the algorithm still provides a solution that satisfies Newton’s 2\textsuperscript{nd} Law and allows the squat model to support the lumbar without the addition of left pelvis and lumbar muscle actuators.

To evaluate the RRA results, OpenSim recommends the thresholds values shown in Table 12. As stated, the peak residual forces should typically be in the range of 0-25 N and the RMS residuals between 0-10 N.

Table 12
Threshold values used to evaluate RRA results for full-body simulations of walking and running

<table>
<thead>
<tr>
<th>Thresholds:</th>
<th>GOOD</th>
<th>OKAY</th>
<th>BAD</th>
</tr>
</thead>
<tbody>
<tr>
<td>MAX Residual Force (N)</td>
<td>0-10 N</td>
<td>10-25 N</td>
<td>&gt; 25 N</td>
</tr>
<tr>
<td>RMS Residual Force (N)</td>
<td>0-5 N</td>
<td>5-10 N</td>
<td>&gt; 10 N</td>
</tr>
<tr>
<td>MAX Residual Moment (Nm)</td>
<td>0-50 Nm</td>
<td>50-75 Nm</td>
<td>&gt;75 Nm</td>
</tr>
<tr>
<td>RMS Residual Moment (Nm)</td>
<td>0-30 Nm</td>
<td>30-50 Nm</td>
<td>&gt;50 Nm</td>
</tr>
<tr>
<td>MAX pErr (trans, cm)</td>
<td>0-2 cm</td>
<td>2-5 cm</td>
<td>&gt;5 cm</td>
</tr>
<tr>
<td>RMS pErr (trans, cm)</td>
<td>0-2 cm</td>
<td>2-4 cm</td>
<td>&gt;4 cm</td>
</tr>
<tr>
<td>MAX pErr (rot, deg)</td>
<td>0-2 deg</td>
<td>2-5 deg</td>
<td>&gt;5 deg</td>
</tr>
<tr>
<td>RMS pErr (rot, deg)</td>
<td>0-2 deg</td>
<td>2-5 deg</td>
<td>&gt;5 deg</td>
</tr>
</tbody>
</table>

However, the size of residuals will depend on the type of motion being studied. For example, residuals for high-speed activities like sprinting will typically be larger than walking [83]. Residuals will also be larger if there are external forces unaccounted for, such as subject’s left GRFs.
The models errors presented in this study are on average within these guidelines however some values are larger than the maximum threshold especially during deep knee flexion. Residual force and moment errors are shown in Figure 63 and Figure 64 below. The position errors for the hip, knee, and ankle for both models were less than or equal to 0.1, 0.005, and 0.04 degrees respectively which are within the threshold guidelines.

Figure 63 Brace-off model residual forces in the x-direction (blue) and y-direction (red) and the torque about the z-axes (green).
Due to the large error values seen in the direction of motion (i.e. y-direction) results found during knee flexion angles less than 25° or greater than 60° could be misleading. These errors agree with the ones found in a study using the OpenSim Gait 2392 scaled MSK model to calculate the tibiofemoral joint contact forces (JCFs) in six subjects for five squat repetitions of squats. Tibiofemoral JCFs of 0.8-3.2 times body weight (BW) were measured. While the MSK simulations underestimated the measured knee JCFs at low flexion angles, an average error of less than 20% was achieved between approximately 25°–60° knee flexion. With an average error that behaved almost linearly with knee flexion angle, an overestimation of approximately 60% was observed at deep flexion. Such data indicates that loading estimations from this particular MSK gait model at both high and low knee joint flexion angles should be interpreted carefully [96].
For these reasons, the follow subsections will look at both extremes with caution however focus on the internal forces seen at 60° knee flexion during both the descent and ascent phase.

4.4 Muscle Forces

Muscle force control is an important determinant of joint stability. It allows us to understand joint movement which can lead to improvements for healthy subjects and athletes through specific training programs [97].

The following comparative graphs describe muscle force as it reacts to brace-off and brace-on conditions during five squat repetitions. Discussions on these effects are further discussed in Chapter 5.

During a closed-chain exercise like the squat exercise, the primary muscles acting about the knee are the quadriceps femoris, which carry out concentric knee extension, as well as eccentrically resisting knee flexion. The primary muscles also act as a flexor to the hip [69]. While the synergist muscles include the soleus and the hamstrings which behave paradoxically and co-contract with the quadriceps. This synergistic action has important implications for enhancing the integrity of the knee joint in squat performance. Specifically, the hamstrings exert a counter-regulatory pull on the tibia, helping to neutralize the anterior tibiofemoral shear imparted by the quadriceps [70]. In addition to this, the hamstrings act as a hip extensor [69].

With these actions in mind, the following comparative muscle force graphs are placed into two groups: 1) primary target muscles including the rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), gluteus maximus (GMX), and gluteus medius (GMD) and 2) synergistic muscles including the long head of the biceps femoris.
(BF), semitendinosus (ST), and soleus (SOL). A final bar graph comparison is shown in Figure 65 and Figure 66.

4.4.1 Primary Target Muscles

4.4.1.1 Rectus Femoris

During the brace-off trial the RF was observed to increase during repetitions three through five. Unlike the brace-on trial where the muscle force remained fairly consistent in magnitude throughout all five squat repetitions. Overall, an increase was observed at full knee extension (average 12.8 N WOB, 14.2 N WB) and at full knee flexion (average 16 N WOB, 17.22 N WB) during the brace-on trial when compared to the brace-off trial.

At 60° knee flexion during descent the KJO was seen to decrease the RF force by 13 N (average 45.7 WOB, 32.6 N WB). However, during the ascend phase at 60° knee flexion the KJO was seen to have almost no effect (average 31.3 WOB, 32.7 N WB).

4.4.1.2 Vastus Lateralis

Again, it was observed that the force increased during the last three repetitions for the brace-off trial yet remained relatively consistent in magnitude during the brace-on trial. At the extremes, the KJO was observed to increase the VL force (average 12.6 N WOB, 15.4 N WB) at full knee extension and at full knee flexion (average 20.3 N WOB, 37.7 N WB).

Specifically, two peaks are noticed for every squat repetition: a max peak and a semi-max peak. The max VL peak occurred at 90° knee flexion of each squat while the semi-max peak occurred during the initial rise from the squat. On average the max peak was 128 N WOB and 134.3 N WB showing that the KJO increased the peak spike muscle
force by 4.7%. On average for the semi-peak was 33.2 N WOB and 42.8 N WB suggesting the KJO increased the muscle force by 29%.

Following this trend the KJO also increased the VL force at 60° knee flexion during both the descent (average 123 N WOB, 127 N WB) and ascent (average 27.6 N WOB, 32.3 N WB).

4.4.1.3 Vastus Medialis

As seen in both the RF and VL force profiles, the VM force also increased during the last three repetitions for the brace-off trial while remaining relatively constant in magnitude during the brace-on trial. At the extremes, the KJO was observed to increase the VM force at full knee extension (average 7.54 N WOB, 10.36 N WB) and at full knee flexion (average 13.4 N WOB, 22.44 N WB).

Again, the two peaks were noticed for every squat repetition. On average the max peak was 64.24 N WOB and 67.3 N WB suggesting that the KJO increased the max peak by 4.6%. On average for the semi-peak was 18.8 N WOB and 24.2 N WB suggesting the KJO increased the muscle force by 29%.

However, little to no change was seen at 60° knee flexion during the descent, (average 59.5 N WOB, 60.5 N WB) or the ascent (average 15.4 N WOB, 18.6 N WB) between brace-on and brace-off conditions.

4.4.1.4 Gluteus Maximus

The GMX is a powerful hip extensor, acting eccentrically to control squat descent and concentrically to overcome external resistance on the ascent. Given its attachment at the iliotibial band, the GMX is also thought to play a role in stabilizing knee and pelvis
during squatting. GMX has been shown to produce a peak hip extensor force at approximately 90° and below [70]. This agrees with the data collected in this study.

The current data showed that the force increased during the decent and hit an average maximum of 5.37 N WOB, 5.69 N WB at the bottom of the squat. At the top of the movement or at full knee extension the GMX was observed to increase (average 2.08 N WOB, 5.41 N WB) during the brace-on trial when compared to the brace-off trial.

Although this trend is consistent with the forces found at 60° knee flexion the change in increase seen is small. At 60° knee flexion the KJO increases the GMX by 1.3 N (descent average 5.9 N WOB, 7.2 N WB and ascent average 4.5 N WOB, 5.8 N WB).

4.4.1.4 Gluteus Medius

The GMD is also a primary hip muscle during the squat exercise yet no significant change in average force was seen between trials at full knee extension (average 10.9 N WOB, 11.5 N WB) or at full knee flexion (average 7.66 N WOB, 7.96 N WB). It was however, noticed that both force profiles have a decrease in max peaks during the first two repetitions for brace-off conditions and during the second and third repetitions for brace-on conditions which could be due to CNS learning. This is further discussed in Chapter 5.

Again, looking at 60° knee flexion during descent, results show that the KJO decreased the GMD force by 1.9 N (average 13.8 N WOB, 11.9 N WB). However during the ascend phase at 60° knee flexion the KJO was seen to do the opposite; increasing the GMD force by 2 N (average 11.9 N WOB, 13.9 N WB).
4.4.2 Synergistic Muscles

During the squat the hamstrings function both as hip extensors and knee flexors, and therefore their length remains fairly constant throughout performance, allowing for a fairly consistent force output. However, peak hamstrings activity has been shown to occur anywhere between 10° and 70° of flexion, with the lateral hamstrings producing greater activity than the medial hamstrings [70].

4.4.2.1 Long Head Biceps Femoris

A consistent trend of force production magnitude was seen throughout each repetition for the long head BF. However, during the brace-on trial it was observed that the magnitude becomes more stabilized, reducing its peaks by 36.96%. Furthermore, the brace-on trial showed a 5.32% increase at full knee extension and a 31% decrease at full knee flexion when compared to the brace-off trail.

At 60° knee flexion during descent the KJO was seen to increase the BF force by 2.5 N (average 9.2 N WOB, 11.7 N WB). However during the ascend phase at 60° knee flexion the KJO was seen to have almost no affect the BF force by 0.4 N (average 8.3 N WOB, 8.7 N WB).

4.4.2.2 Semitendinosus

The ST showed a small overall contribution of muscle force production throughout the movement for both trials with an average force of 5.07 N WOB, 5.1 N WB at full knee extension and 5.5 N WOB, 5 N WB at full knee flexion. This small contribution was also seen at 60° knee flexion during the descent and ascent phase showing an average 4.2 N WOB, 4.7 N WB and 4.5 N WOB, 4.3 N WB respectively.
4.4.2.3 Soleus

The SOL is a pure plantar flexor with proximal attachments at the tibia and fibula and distal attachments at the calcaneus [70]. It is an underestimated muscle in the squat exercise yet produces a significant amount of force. On average the soleus produced 171.8 N WOB, 65 N WB at full knee extension and 66 N WOB, 50 N WB at full knee flexion.

More interesting is the amount of force reduced during brace-on conditions versus the brace-off. The maximum peak force occurred at the bottom of the squat with an average force of 277.6 N WOB and 94.2 N WB which suggests that the brace decreases the peak muscle force by 66%. At the extremes the KJO reduces the force by 62% at full knee extension and by 24.24% at full knee flexion.

Such trend was also seen at 60° knee flexion during both descent and ascent where the KJO was seen to decrease the SOL force by 67.4 N (average 123.5 N WOB, 56 N WB) and by 3.3 N (average 47 N WOB, 43.7 N WB) respectively.
Figure 65 Comparison of the average muscle force found at 60 degree knee flexion during the descent. Gluteus medius (GMD), semitendinosus (ST), biceps femoris (BF), gluteus maximus (GMX), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), soleus (SOL).

Figure 66 Comparison of the average muscle force found at 60 degree knee flexion during the ascent. Gluteus medius (GMD), semitendinosus (ST), biceps femoris (BF), gluteus maximus (GMX), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), soleus (SOL).
Table 13
A summary of average individual muscle forces (N) at 60° knee flexion during descent and ascent for brace-off (WOB) and brace-on (WB) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Average (WOB)</th>
<th>Average (WB)</th>
<th>Average (WOB)</th>
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4.5 Electromyographic Response

EMG responses of the RF, VL, VM, GMX, GMD, long head BF, ST, and SOL using a root mean square analysis were recorded. These recordings were used as a validation technique when comparing trends to the models activation and force production calculated from SO.

For both trials and each muscle, the EMG response was normalized to its associated MVC and graphed. These graphs can be found in Appendix B. Highlighted below are the GMX, BF and SOL. EMG results were filtered using root mean square whereas the model’s activation results, found in Appendix B, from OpenSim’s SO was filtered at a low pass 6 Hz.
4.5.1 Gluteus Maximus

The GMX activation is seen in Figure 67 to increase during brace-on conditions which may indicate a balanced synergy. Such synergy is explained in detail throughout Chapter 5. Such results follow the trends seen in both the GMX calculated activation and force profiles.

![GMX EMG Data](image)

Figure 67 GMX normalized EMG activation data for brace-off (red) and brace-on (blue) trials.

4.5.2 Long Head Bicep Femoris

The activation observed during the brace-off trial showed significantly higher spikes occurring at the bottom of the squat, whereas the activation seen in the brace-on trial stayed fairly level without drastic spikes shown in Figure 68. This same trend was noticed in the BF activation and force production trend produced by the model and seen in Appendix B.
Figure 68 BF normalized EMG activation data for brace-off (red) and brace-on (blue) trials.

4.5.3 Soleus

The SOL activation is reduced during the brace-on trial indicating the KJO has an impact on this muscle. The subject’s normalized activation is shown in Figure 69. The model’s activation trend appears to be similar to the EMG data however exhibits no peaks for the brace-on trial. The EMG peaks observed during the first two repetitions of the brace-on trial could be caused by an error in the sensor or the subject becoming unbalanced. This will be discussed further in Chapter 5 and 6.
Figure 69 SOL normalized activation data for brace-off (red) and brace-on (blue) trials.
Chapter 5

Conclusions and Discussion

It was assumed that the non-linear spring technology in the KJO releases stored energy and in turn it offloads the quadriceps muscular activation, requiring less exerted energy from the wearer during extension. This assumption resulted in the following hypothesis: A non-linear spring response could lower the quadriceps force during knee extension. However, based on the results presented in Chapter 4, this hypothesis appears incorrect.

Instead, the results presented in Chapter 4 suggest that the non-linear spring-loaded KJO has a performance influence on the subject’s movement patterns and force production during a deep squat exercise rather than an assistive influence. This influence is now discussed in the following subsections.

5.1 Qualitative Analysis of the Movement

A normal squat pattern will result in the knee axis translating in an anterior direction during the descent, and a posterior direction during the ascent. If an individual biases their quadriceps during the movement, then this axis will move even further anteriorly. This is because they are required to shift their center of mass forward, to remain over the base of support and maintain balance.

If more activation is occurring in the GMX during the descent, then the knee joint axis will still move in an anterior direction, but it will be less. As a result, the pelvis will translate in a more posterior direction as the individual will feel more balanced in this position due to the tension supplied by the GMX.
This phenomenon was illustrated in the squat pattern performed during both brace-off and brace-on conditions. In the brace-off condition, it is possible to see the knee joint axis continued to translate anteriorly, as the subject completes the descent portion of the squat. In the brace-on condition, the axis moves forward in the beginning of the descent, and then can be seen to stabilize, as the pelvis moves posteriorly with a stronger hip abduction to complete the squat. This is likely due to the increased tension and balance provided by the GMX in the brace-on condition.

It is important to remember that at about 60° of knee flexion, the tension supplied by the quadriceps is neutralized by the posterior chain musculature, (i.e. hamstrings and gluteals). At angles deeper than 60°, the posterior chain, primarily GMX, will control the movement. The hallmark feature of this movement synergy will be a more posteriorly oriented pelvis at angles below 60° of knee flexion, whereas a quadriceps dominant strategy will bring the pelvis into a more anteriorly oriented position.

One additional effect of a quadriceps dominant synergy will be increased SOL activation to stabilize and control the tibia on the talus due to the anteriorly displaced center of mass. A synergy that employs more posterior chain musculature will reduce SOL activation because the individual will have their weight shifted in a more posterior direction.

Furthermore, during the squat exercise many athletes bounce at the bottom just before raising. This bounce is meant to increase acceleration and therefore increase the force. This increased acceleration is seen to help the athlete return to a standing position. This inherent movement was observed in the current study. It was also observed that the
KJO seemed to limit this bounce from occurring and in turn decreased the muscle spikes seen in the brace-off trial.

The KJO also affected the posture in the upright position. Large decreases were seen in the brace-on condition at full knee extension. This suggests that the brace is functioning as a performance training tool that aids in maintaining a neutral position. In other words, the KJO may have a performance influence through hyperextension prevention and balance synergy encouragement.

5.2 Analysis of Muscle Torque and Force Production

Sagittal plane balance during a deep squat is known to be enhanced by proximal posterior chain activation, (i.e. GMX). This will enable the subject to produce a posterior biased weight shift during the squat. The central nervous system (CNS) will always adopt the most readily available strategy to keep the subject balanced.

If afferent input from the GMX to the somatosensory cortex indicates that it is fully functioning, then less demand will be placed on the SOL to achieve balance in an anterior – posterior direction, and tension supplied by this muscle may be reduced. This could explain the reduction in SOL muscle force and the increase in GMX muscle force seen in the brace-on trial.

The eccentric and concentric phases of the squat task are controlled primarily by the RF and GMX. At about 60° of knee flexion, the leverage supplied by the RF is minimized, and the torque that it is able to supply is subsequently reduced. This enables the muscles of the posterior thigh (GMX, BF, and ST) to control the task. Of the three, the GMX has the cross-sectional area and leverage to manage the movement most effectively. It is plausible that the changes in proprioception discussed early will bring
more GMX activity into the movement synergy, especially as it relates to extending the hip, in both the negative and positive phases.

It is interesting to note that as GMX activity increases, both BF and ST decrease during the brace-on condition. BF and ST are both weak extensors of the hip, so it stands to reason that if GMX is not functioning optimally, their activity and contribution to the task will be increased, as was noted in the brace-off condition.

However, further research is needed to make a final conclusion on this trend. It is important to evaluate a combination of knee flexion and dorsiflexion throughout the squat exercise to get an accurate representation of tibial position in the sagittal plane. This is because better posterior chain activation will enable the subject to squat deeper, with better balance. This will cause the dorsiflexion angle to open up and the knee flexion angle to become smaller.

Additional research involving KJO without the non-linear spring should also be considered. This will give insight on the protagonist that provides the performance enhancement benefit; the brace or the non-linear spring technology. Additional research should also investigate the effects KJOs have on the opposite knee. Many players returning to the field wearing a KJO experience an injury in the healthy knee. This lends the question; do KJO negatively affect the opposite knee? The discussion on supplementary research possibilities are further discussed in Chapter 6.

The performance of the RF during the brace-on condition was also interesting because during the first repetitions of the squat, the RF muscle force did not change substantially compared to the brace-off condition. However, during the last three repetitions, it appeared that there was a reduction in RF force generation.
This may be due to the predominant neuromotor strategy at the subcortical level that biases neural programming to the quadriceps to perform the squat task. The KJO may alter proprioception provided by mechanoreceptors in the joint. The neuroplastic properties of the CNS will allow it to accommodate to the change in proprioception afforded by the KJO, and alter efferent output to reduce force supplied by the RF to complete the task.

In a sense, the CNS is learning and changing its outputs based on alteration of the inputs. It may be useful to attempt several trials with more repetitions to enable this “accommodating or CNS learning effect” to take place to get a more accurate representation of RF activity. Future studies like this amongst others are presented later in Chapter 6.
Chapter 6

Limitations and Future Work

6.1 Limitations

OpenSim’s GAIT 2392 model is one of the most widely used models in the literature however it comes with a few limitations [79]. The limits affecting the current study are listed below.

1) The GAIT 2329 model was designed for simulations of movements which are leg dominated. This model also best represents the joints and muscle groups that are of most interest to this study without adding complexity of unrelated parts such as the lumbar or upper extremities. However, it was also developed for gait analysis making it difficult to analyze movements that require higher degrees of flexion. Specifically, angles greater than 120° knee flexion and ±90° pelvic tilt.

This introduces a trade space of sorts. To minimize this space and reduce error, the RRA module was used to apply residual actuators at the pelvis. This gave the model added strength to achieve the required flexion angles without adding the complexity of additional parts to the model.

2) Joint angle measurements during motion analysis are subject to error caused by kinematic crosstalk. Kinematic crosstalk is when one joint rotation (e.g., flexion) is interpreted as another (e.g., abduction) [98]. The measurement of the ‘screw-home’ motion of the human knee, in which axial rotation and extension are coupled, is especially prone to errors due to crosstalk. This suggests that the measurement of screw-home rotation may be strongly influenced by errors in the location of the flexion axis [98].
To minimize this crosstalk, the model located the knee joint center of rotation using markers placed just above and below the knee joint. The model then computed the knee angles and moments about the instant center of knee rotation.

3) Even with the efforts of defining accurate muscle paths in the lower extremity through wrapping points, there are some muscles that still pass through the bones or deeper muscles with extreme hip flexion and extension, and thus yield unrealistic moment arms. This is seen in the interior GMX which passes through the ischial tuberosity when the hip degree of flexion is beyond 60°. Furthermore, when hip flexion reaches beyond 80° the superior and the middle components of the GMX pass through the deeper muscles.

Noticing these limitations, the goal of this study, to compare parameters of ROM and muscle force between two models (brace-off and brace-on), is not impaired. However it is noted that the joint moments and muscle forces during the individual trials are subject to error during hip flexion/extension greater than 60-80°.

Additional limitations which possibly results in error were seen in the SimTrack pipeline. The specific limitations for each tool as it relates to the current study are listed below:

**Limitations with Scaling**

1.) Subjective influence of the user - The scale tool allows the user to intervene if the marker errors are too high by adjusting the markers on the model to fit the
placement of the experimental static trial. The tool also allows for markers to be excluded. This process is highly subjective regarding the user’s decisions [83].

2.) Enhanced maximum isometric forces – During the scaling process the maximum isometric forces stay the same. In other words, they are not automatically changed according to the subjects’ characteristics. This change is left up to the user [83]. However, if the user manually strengthens or weakens a muscle’s maximum isometric force, the relation to the tendon stiffness stays the same. This can be problematic. For example, a great enhancement in maximum isometric force of a muscle means unrealistic tendon stiffness deeming it no longer physiological [79].

Limitations with Inverse Kinematics

1) Recommended error values of are based off gait analyses and not a deep squat; leaving this study unsure of what acceptable errors are.

2) In an effort to decrease the error values, inaccurate changes to scaling and marker placements is possible due to the subjective behavior behind the user’s edits.

Limitations of Static Optimization

1) An under-actuated model is the product of muscles not being strong enough to generate the required forces and/or model coordinates not having any actuation whatsoever (muscles or other actuators). In the custom model, there are no muscles to control lumbar flexion/extension requiring the lower limb muscles to control trunk motion, without much success. This problem was diverted by adding actuators to the pelvis using the RRA module.
Limitations to RRA

1) Reserve and residuals must be tuned carefully. Reserve actuators can contribute a significant amount of the total joint torque; joint torque in which muscles are preferred to provide. Setting these values based off one subject is subjective and may lead to skeptical results. The current study did not overly tune the parameters; instead parameters suggested by OpenSim’s GAIT 2392 Model were used and the same parameters were applied to both brace-on and brace-off conditions.

Limitations Associated with Electromyography

A limitation in the EMG methodology specifically to this study was due to motion artifact. Motion artifact is caused by the relative movement of the sensor with respect to the underlying skin over the muscle of interest. It can result from a direct impact to the sensor or to the body or a rapid movement of the body segment to which the sensor is attached. Motion artifact is particularly problematic during dynamic contractions or vigorous activities.

Motion artifact was seen in the brace-on trial where the RF EMG sensor and KJO conflicted with one another. Specifically, the KJO was placed just below the belly of the RF. After the KJO was tighten it created a distributed load on the leg interfering with the sensor. This led to a false muscular activation of the RF.

Furthermore it was observed that the values of the normalized muscular activation for the quadriceps muscles and the SOL had noticeably lower activations when compared to the other muscles. This result was possibly caused by the MVC exercises. The individual exercises were meant to isolate each muscle and therefore activating them as
much as the subject physically could. The results from the MVC observed were significantly higher than that of the data collected during the squatting motion leading to a misrepresentation of activation.

Limitations regarding the sensors technology also exist. Delsys recognizes the limitations while offering unique solutions. Some of these are listed below:

1) Surface sensors are susceptible to error because skin moves during movement. This can cause muscle crosstalk or in other words, the signal reported may include activation from a neighboring muscle. These muscle crosstalk signals become superimposed on the EMG signal from the muscle of interest, and their presence distorts the amplitude and timing of the EMG data. To combat this error Delsys sensors are designed with an interface and inter-electrode distance that has been demonstrated to offer the optimal cross-talk suppression while maintaining the EMG signal amplitude.

2) Physiological noise originates from tissues other than muscles that generate electrical signals, such as EKG signal. It can be reduced by properly locating the EMG sensor further away from the source of the noise, if possible.

6.2 Future Work

Based on the results, the next study should take a step back. As mentioned in Chapter 5, the evaluation of KJOs without non-linear spring technology could determine if the results presented in the current study stem from the non-linear spring or from the brace itself.

Additional studies should also include evaluating what impacts a KJO has on the opposite knee. This is important to understand because many players returning to the
field wearing KJOs experience injuries in the healthy knee. Raising the question, do KJO negatively affect the opposing knee?

Future studies should include applying the current methodology to a larger population; identifying which injuries and which post-surgical patients would be a “best fit” for this KJO amongst other braces used in current treatment protocols; and investigating the effect of the KJO during normal gait and during high impact movements such as running, skipping, and jumping.

Such studies will allow for a more complete understanding as to how the non-linear spring technology affects the biomechanics of the knee. This will also minimize anomalies that are associated with subject-specific testing.

6.3 Contributions and Wider Impact

To the best of the researchers’ knowledge, the novel work within this Ph.D. dissertation is the first in biomechanical modeling focusing on the biomechanical effects of a non-linear spring-loaded KJO to the knee complex during human squatting. It then successfully analyzed the differences in joint angles, moments, and internal muscle forces between two squat trials. The focus lied on comparing two computational models against each other to give better insight on how a specific non-linear spring profile affected a person’s movement and force profile throughout a squat cycle. The simulations used inverse kinematics and dynamics as well as static optimization providing an alternative non-invasive method to the traditional in-vivo method for quantifying internal muscle forces.

This study is one of the very few works using recent technology in mathematical modeling to estimate internal muscle forces independently of EMG measurements.
Instead, this study used simultaneously captured EMG data parallel to the kinematics and ground reaction forces which were used in the modeling process. This is important as EMG measurements are the only validation tool able to present muscle activity non-invasively. Most studies which estimated muscle forces and validated the results to EMG used EMG data in their estimation process. Furthermore, some studies do not collect their own EMG data but instead use available EMG profiles from the literature.

Therefore, these models and the inverse dynamic process can be used to help understand joint kinematics as well as muscle activations and internal forces; thereby contributing to the current knowledge of biomechanical response estimations in clinical movement analysis. This knowledge will further aid physical therapist and orthopedic specialist in prescribing the most appropriate KJO to a specific patient. Furthermore, such process is also well-suited for performing “what if?” studies allowing a variety of KJO studies to be conducted. This shows the usefulness of such approach in biomechanical modeling as well as in the medical field of orthotics.

The models presented here will be made available on the OpenSim’s website. This will provide a squat model for others to begin their research with as well as enable other OpenSim users to compare their data with this study. This will further contribute to the biomedical field of engineering by developing a data pool for muscle force trends and estimations for a healthy adult during the squat cycle. This is very important as it motivates the interaction between different laboratories and encourages them to upload their data which in turn develops a strong community for estimating internal muscle forces in clinical squat analysis.
Through conducting this study, important limitations and uncertainties of the current model to estimate muscle activations and forces was uncovered. These results showed that this model is not yet at the stage to be implemented into a clinical routine processing. Sharing of this study will show the limitations with the current model’s restrictions involving deep knee flexions which may encourage others to create a model that can withstand such extreme angles accessible for the implementation into the clinical deep squat analysis. The results did however showed both computational models were successful during semi-squatting angles, which suggest they are suitable for the purpose of measuring patients, as the main population experience knee flexion during sitting and standing from a chair.
## Appendix A

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Figure 70 Ankle ROM comparison between brace-on (WB) conditions to brace-off conditions.

Figure 71 Knee ROM comparison between brace-on (WB) conditions to brace-off conditions.
Figure 72 Hip ROM comparison between brace-on (WB) conditions to brace-off conditions.
Calculated Joint Moment Data

Figure 73 Ankle joint moment comparison between brace-on (WB) conditions to brace-off conditions.

Figure 74 Knee joint moment comparison between brace-on (WB) conditions to brace-off conditions.
Figure 75 Ankle joint moment comparison between brace-on (WB) conditions to brace-off conditions.
Position errors

**Position Errors for Each of the Model's Generalized Coordinates During the Trial - Ankle Dorsiflexion Angle**

Figure 76 Ankle dorsiflexion angle position errors for the model's generalized coordinates are at or below 0.1° (0.002 radians).

**Position Errors for Each of the Model's Generalized Coordinates During the Trial - Hip Flexion Angle**

Figure 77 Hip flexion angle position errors for the model's generalized coordinates are at or below 0.04° (0.0075 radians)
Figure 78 Knee extension-flexion angle position errors for the model's generalized coordinates are at or below 0.005° (0.0001 radians).
EMG SO Results

Figure 79 BF muscle activation results for the brace-off (red) and brace-on (blue) calculated by OpenSim.

Figure 80 SOL muscle activation results for the brace-off (red) and brace-on (blue) calculated by OpenSim.
Figure 81 GMX muscle activation results for the brace-off (red) and brace-on (blue) calculated by OpenSim.
EMG Results

Figure 82 Rectus femoris activation results for the brace-off (red) and brace-on (blue). Results indicate interference between the KJO and the EMG sensor.

Figure 83 Vastus lateralis activation results for the brace-off (red) and brace-on (blue). Brace-off (red) shows signs of higher activation, whereas brace-on (blue) shows to be more consistent throughout the trial, without larger spikes in activation.
Figure 84 Vastus medialis activation results for the brace-off (red) and brace-on (blue). Activation results indicate the KJO had minimal impact on the vastus medialis.

Figure 85 Gluteus maximus activation results for the brace-off (red) and brace-on (blue). Brace-on (blue) shows higher level of activation in comparison to brace-off (red).
Figure 86 Gluteus medius activation results for the brace-off (red) and brace-on (blue). Brace-off displayed higher activation than seen in brace-on.

Figure 87 Semitendinosus activation results for the brace-off (red) and brace-on (blue). Activation levels remain fairly consistent suggesting KJO has minimal impact on the activation in the semitendinosus.
References:


