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KNEE BIOMECHANICS IN OBESE FEMALE DURING LOCOMOTION

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BMI	=	Body mass index
PVGRF	=	Peak vertical ground reaction force
PEKADM	=	Peak external knee adduction moment
KAAI	=	Knee adduction angular impulse
SS	=	Self selected speed
CV	=	Constant velocity
FR	=	Froude velocity
Nor	=	Normal group
OW	=	Overweight group
OB I	=	Obesity I group
OB II	=	Obesity II group

CHAPTER I

INTRODUCTION

Obesity is a global issue of metabolic disease (Vincent et al., 2012). The prevalence of musculoskeletal problem is correlated with rising obesity population (Vincent et al., 2012). The major of osteoarthritis risk factor in load-bearing joint such as knee, hip, and ankle and non load-bearing joint such as hand is strongly associated with obese women (Berenbaum and Sellam, 2008;Blagojevic et al., 2010;Russell and Hamill, 2011;Vincent et al., 2012). The body mass index (BMI) significantly correlates with the incident risk of radiological and symptomatic knee osteoarthritis that was shown in each BMI increase by 1 kg/m^2 above 27 is associated with a 15% increased risk which cause a reduction of fatigue durability of articular cartilage (Berenbaum and Sellam, 2008;Blagojevic et al., 2010;Landinez-Parra et al., 2011). Moderate to normal loading (10-20 MPa) is thus useful to keep cartilage healthy and functional, whereas excessive mechanical load, as for instance overweight impact may cause chondrocyte death and damage in the matrix. Overweight may act in two different ways to cause osteoarthritis. First, it is due to the increased magnitude of force transmitted across the joint inducing cartilage rupture, meniscal tears and therefore osteoarthritis. Second, systemic factor should be included, possibly a cartilage growth factor or bone growth factor that may accelerate cartilage break-down causing osteoarthritis (Landinez-Parra et al., 2011).

Biomechanical factors such as kinetic and kinematic conditions during walking should be carefully considered for the treatment of cartilage disease in weight bearing joints (Koo et al., 2011). Increased mechanical loads related to obesity appear as a logical explanation to the increased risk of osteoarthritis in weight-bearing joints with destruction of the cartilage and ligaments. Repeated overloading of a normal joint can induce abnormalities in chondrocyte and osteoblast behavior, as well as alterations in the extracellular matrix. In addition, obesity is associated with an increase in bone mass, which can lead to increased rigidity of the subchondral bone, thereby promoting rupture of the cartilage matrix (Berenbaum and Sellam, 2008;Domitrovic et al., 2011). The matrix alterations are related, at least in part, to activation of mechanoreceptors at chondrocyte surface, which triggers a cascade of deleterious intracellular events, caused of osteoarthritis (Berenbaum and Sellam, 2008). Knee hyperextension at heel strike during walking for obese subjects may influence the tibiofemoral contact regions more in the medial than lateral compartment. The locations of the thickest cartilage in the medial condyles of femur were significantly correlated with the knee flexion/extension angle at heel strike, while the locations in the lateral condyles were not associated with flexion/extension angle. The medial and lateral cartilage thickness variations in the knee are influenced by the peak knee adduction moment during normal walking (Koo et al., 2011). There is growing interest in the role of the external knee adduction moment in the pathogenesis of knee pain and osteoarthritis. The external knee adduction moment tends to adduct the knee into a varus position and is significantly correlated with disease severity (Foroughi et al., 2009). Tibia angle had a significant impact on the knee adduction moment. Increasing the tibia angle moves the knee medially decreasing the knee adduction moment arm and produces a similar effect as medial thrust or changes in and increasing trunk sway caused reductions in the first peak of the knee adduction angle. Additionally, toeing in and increasing trunk sway caused reductions in the first peak of the knee adduction moment. The external moments were calculated by taking into account the moment about joint center and inertial forces. An external moment is equal and opposite to the internal moment created by muscles, soft tissues and joint contact forces (Kakihana et al., 2004). However, it is unclear whether the difference of the external knee adduction moment and other biomechanics parameters between obese and normal healthy subjects.

Modified footwear and orthosis have been investigated as potential conservative management of knee osteoarthritis (Erhart et al., 2008;Erhart et al., 2008;Fantini Pagani et al., 2010;Jenkyn et al., 2011;Radzimski et al., 2012). More specifically, lateral wedging has been used with goal of reducing symptoms associated with medial knee osteoarthritis, hypothetically by reducing the peak external knee adduction moment. (Russell and Hamill, 2011;Radzimski et al., 2012). If specific footwear interventions are associated with decreased external knee adduction moment during walking and other physical activity, these may be useful toward the management of symptoms of patients with medial knee osteoarthritis and potentially to reduce the risk of future osteoarthritis in people who are at increased risk, such as those with knee injuries or obese.

There is growing interest in the role of the external knee adduction moment in the pathogenesis of knee pain and osteoarthritis (Foroughi et al., 2011;Koo et al., 2011). The external knee adduction moment tends to adduct the knee into a varus position and is significantly correlated with disease severity (Sheehan and Gormley, 2013). Moreover, the thickness of the medial tibial cartilage is related with the peak knee adduction moment and knee adduction moment impulse that represents the dynamic knee joint load during walking (Henriksen et al., 2012). However, the effectiveness of lateral wedge insoles on knee joint load parameters such as the external knee adduction moment and knee adduction moment impulse in various body mass index is unclear.

1.1 Research Questions

1. Is there the difference of gait kinetic and kinematic due to the external knee adduction moment between obese, overweight, and normal healthy subjects?

2. Is there the correlation between anthropometry, biomechanical parameters during walking, muscle strength, and female BMI level?

3. Does orthosis effect on gait kinetic and kinematic due to the external knee adduction moment between obese, overweight, and normal healthy subjects?

1.2 Objective

1.2.1 General Objective

To compare knee biomechanics between obesity and normal healthy female subject during locomotion

1.2.2 Specific Objectives

1. To compare gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) between obesity female and normal healthy subject during stance phase of gait cycle.

2. To study the correlation between gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment), and anthropometry and female BMI levels.

3. To study the effect of orthosis on gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) in obesity female and normal healthy subject.

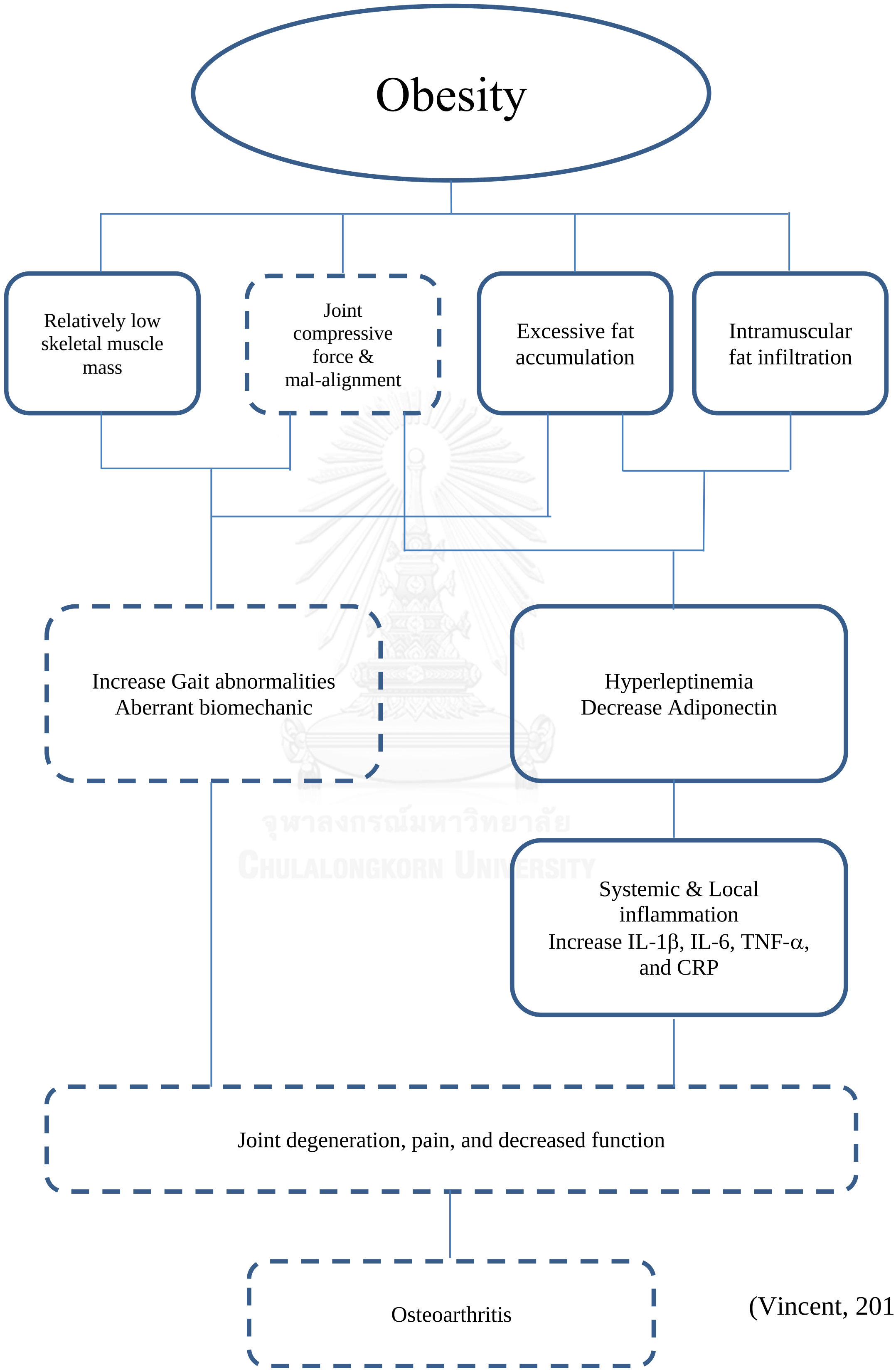
1.3 Hypothesis

1. There is no different of gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) between obesity female and normal healthy subject during stance phase of gait cycle.

2. There is no correlation between gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment), and anthropometry AND female BMI levels.

3. There is no effect of orthosis on gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) in obesity female and normal healthy subject.

1.4 Conceptual framework



(Vincent, 2012)

CHAPTER II

LITERATURE REVIEW

2.1 Obesity

Obesity is a global issue of metabolic disease (Vincent, 2012) that is fundamentally a problem of energy balance irrespective of the under lying social, cultural, behavioural and genetic determinants (Trayhurn, 2007). Differences between intake and expenditure are primarily buffered by changes in the amount of lipid (triacylglycerols) deposited in the specialized fuel storage organ, white adipose tissue (white fat) (Trayhurn, 2007). Adipose tissue biology is now recognized as a major endocrine and signaling organ. In addition to fuel storage, white adipose tissue can act as a thermal insulator and protect other organs from mechanical damage. Two further features of the tissue should be highlighted in multiple depots in the body, both subcutaneously and internally, and clusters of adipocytes may also be located adjacent to, or embedded in, other organs such as the lymph nodes and skeletal muscle. A second important feature is that adipose tissue is not made up simply of mature adipocytes, which store the lipid, but contains a variety of other cells (e.g. fibroblasts, endothelial cells, macrophages) which constitute around 50% of the total cellular content. From the functional viewpoint, it is, if course, the mature adipocytes those are the pivotal cells within adipose tissue (Trayhurn, 2007).

White adipocytes are major secretory cells, making adipose tissue a key endocrine organ. Indeed, adipose tissue is the largest endocrine organ in most human and certainly so in the overweight and obese that results the large secretory organ has the potential to impact broadly on the body as a whole. The most important secretion from adipocytes is fatty acids, of which there is net release at periods of negative energy balance. In addition to fatty acids, several other lipid moieties are released by fat cells; these included prostanoids, which are synthesized by the tissue, and cholesterol and retinol, which are not synthesized but rather are stored and subsequently released. In addition, certain steroid hormone conversions can take place within white adipocytes (Trayhurn, 2007).

The new component of the secretome of adipocytes is the wide range of protein factors and signals that are released. These adipokines or adipocytokines now number in excess of 50 different molecular entities. The adipokines are highly diverse in terms of protein structure and physiological function. They include classical cytokines, growth factors and proteins of the alternative complement system; they also include proteins involved in the regulation of blood pressure, vascular haemostasis, lipid metabolism, glucose homeostasis and angiogenesis (Trayhurn, 2007).

Leptin is the adipokine which has received most attention. Its discovery in 1994, as the product of the Ob gene in the genetically obese (ob/ob) mouse, was the pivotal discovery which led to the realization that adipose tissue is a critical endocrine organ. Leptin is an essential signal from adipocytes to the hypothalamus in the control of appetite and energy balance. Indeed, without functional leptin, severe obesity ensues as in the ob/ob mouse and the human homologues that have been described. Leptin is in practice a pleiotropic hormone, its functions extending far wider than appetite and energy balance to encompass a multiplicity of actions, including acting as a signal in reproduction and immunity (Trayhurn, 2007).

Much attention has also been focused on adiponectin, which is a hormone produced exclusively by adipocytes. In contrast to most adipokines, and leptin in particular, the expression and circulating levels of adiponectin fall in obesity. A number of roles are attributed to adiponectin, including the modulation of insulin sensitivity and vascular function, as well as anti-inflammatory action. Adiponectin, like leptin, is a powerful example of an adipocyte derived hormone which interacts with other organs and a wide range of physiological systems and metabolic processes (Trayhurn, 2007).

2.2 Inflammation and obesity

A number of inflammation-related proteins are released by white adipocytes, as well as adiponectin, and these include cytokines, chemokines and acute phase proteins. In addition to these factors, several other inflammation-related adipokines are recognized, including leptin and the angiogenic protein, vascular endothelial growth factor. In obesity, the production of many of these adipokines increases markedly and the tissue is in effect inflamed (Trayhurn, 2007).

One of the most important recent developments in obesity research is the emergence of the concept that obesity is characterized by chronic mild inflammation paralleling the situation with other diseases. The basis for this view is that the circulating level of several cytokines and acute phase proteins, it is considered that the expanded adipose tissue mass contributes, either directly or indirectly, to the increased production and circulation levels of inflammation-related factors in obesity. In other words, the state of inflammation in adipose tissue in obesity leads to an increased production and release of inflammation-related factors (Trayhurn, 2007).

Close links, and even similarities, between adipocytes and immune cells are increasingly evident. The inflammatory state of adipose tissue in obesity has been highlighted by recent reports demonstrating that there is extensive infiltration of the tissue by macrophages in the obese. The arrival of macrophages is thought to lead to a considerable amplification of the inflammatory state in white fat, through the cytokines and chemokines that they secrete (Trayhurn, 2007).

The current view is that the inflammatory state of obesity plays a key causal role in the development of type 2 diabetes and the metabolism syndrome (which includes artherosclerosis, hypertension and hyperlipidaemia) associated with obesity. A central hypothesis is that it is the increases in obesity that lead to the associated diseases. The reduction in adiponectin in the obese is thought to be of particular significance in view of the anti-inflammatory effect of this adipokine. Alterations in fatty acid flux have also been implicated (Trayhurn, 2007).

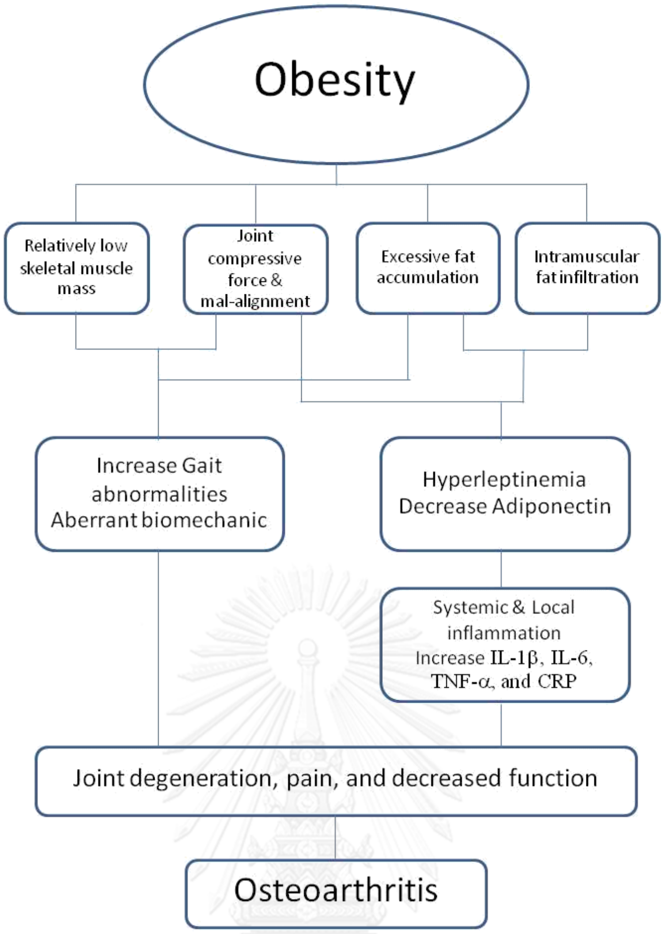


Figure 1 Obesity – specific mechanisms of OA pathophysiology (Vincent, 2012)

2.3 Obesity – specific mechanisms of osteoarthritis pathophysiology

Although numerous pathways contribute to the onset of osteoarthritis, obesity – specific mechanisms include relative loss of muscle mass and strength over time, mechanical stress, and systemic inflammation. Excessive adipose tissue compresses load-bearing joints and creates an inflammatory environment within tissue and joints. Obesity induces abnormal joint loads and leads to adverse changes in the composition, structure, and properties of articular cartilage. With increased body weight, both muscle mass and fat mass increase, yet the volume of muscle mass remains relatively low and inadequate to match the loads placed upon it (Vincent, 2012).

When strength is normalized for body mass, obese persons have lower muscle strength than the normal-weight counterparts, including the quadriceps and lumbar muscle groups. People who are obese attempt to compensate for muscle weakness and instability by altering gait patterns and adopting different body transfer patterns to move excessive weight. With inadequate lower limb strength, less absorption of the impact forces on weight-bearing joint occurs. Repetitive forces damage articular cartilage. Joint misalignment in the load-bearing joints may occur with increased body segment girth, altered posture, skeletal muscle strength imbalance, or weakness of muscles that control joint motion (Hinman, 2010).

Over time, the cumulative effects of excessive body fat, in addition to mechanical loading and aberrant joint motion, contribute to the osteoarthritis pathophysiology and onset of inflammation and pain(Sowers, 2010). Low-grade systemic is now considered a hallmark of obesity and manifests as elevations in interleukin (IL)-1β, IL-6, tumor necrosis factor (TNF)-α, and the acute phase reactant C-reactive protein (CRP)(Schrager, 2007). The biomarkers might link obesity with the onset and progression of osteoarthritis(Vincent, 2012). The local inflammation response in the synovial fluid of joints afflicted by osteoarthritis includes elevations of IL-1β. Systemic levels of IL-1β, IL-6, TNF-α, and CRP also increase with the presence of hip or knee osteoarthritis. Five-year prospective evidence indicates that elevated levels of TNF-α or CRP can predict the progression of osteoarthritis (Spector, 1997; Sharif, 2000). Chronically high IL-6 levels are predictive of knee osteoarthritis during a 10-year period (Livshits, 2011). The IL-1β protein content of the vastus lateralis is 34% higher and quadriceps strength is 40% lower in obese persons with osteoarthritis compared with persons without osteoarthritis (Levinger, 2011).

Inflammation is mediated by the activities of several adipokines, such as adiponectin and leptin (McNulty, 2011). Leptin modulates food intake by acting on neural pathways in the hypothalamus and brainstem. Although the specific mechanisms underlying adipokines action in osteoarthritis are not fully know, recent evidence suggests that excessive leptin levels may activate cellular pathways that contribute to cartilage break down (McNulty, 2011). Normally, leptin activates expression of growth factors and production of extracellular matrix in cartilage and can upregulate matrix metalloproteinases and IL-1, both of which contribute to nitric oxide production and subsequent chondrocyte apoptosis and cartilage breakdown (Lago, 2008). Leptin also is found in cartilage and osteophytes in person with osteoarthritis. Hyperleptinemia occurs locally in the human osteoarthritis joint. the combined influence of pain and worsening inflammation in untreated obesity likely contributes to an elevated risk for functional impairment in obese, older adults. Adiponectin is a hormone secreted by adipocytes. Although produced in relatively low concentrations compared with the concentration found in plasma, this hormone could be found in the synovial fluid of osteoarthritic joints and likely is derived from the infrapatellar fat pad and synovium (Vincent, 2012). Conflicting evidence indicates that adiponectin can be proinflammatory (triggering IL-6 and nitric oxide production) or anti-inflammatory (unregulating inhibitors of metalloproteinases) (Hu, 2011).

2.4 Knee joint

2.4.1 Bones of the knee joint

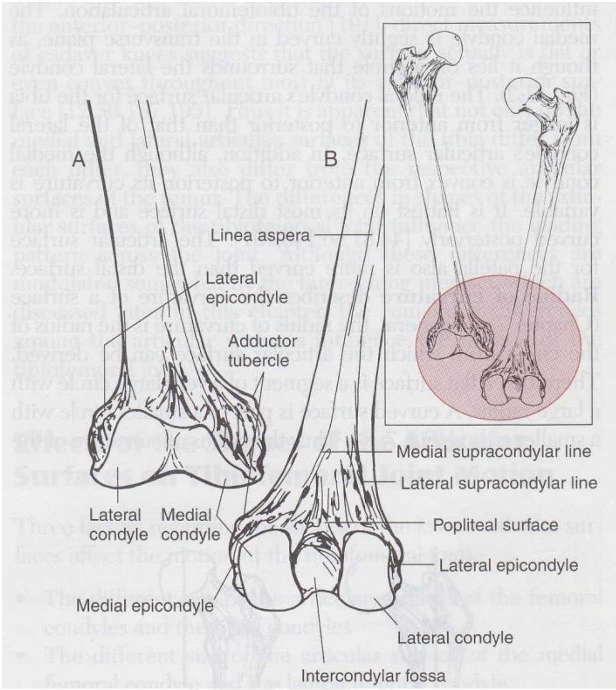


Figure 2 Bones of the knee joint (Oatis, 2009)

The knee joint is composed of the distal femur, proximal tibia, and the patella. The shaft of the femur has three surfaces, anterior, medial, and lateral. The medial and lateral surfaces are separated from each other posteriorly by the linea aspera, the prominent posterior crest that gives rise to much of the quadriceps femoris muscle. The linea aspera splits distally, contributing to the medial and lateral supracondylar lines and demarcating a posterior direction and widens medially and laterally to form the medial and lateral supracondylar lines. The supracondylar lines terminate in the expanded distal end of the femur, which provides the articular surfaces for the knee joint (Oatis, 2009).

2.4.1.1 The distal end of femur

The distal end of the femur consists for two large condyles that are continuous with each other anteriorly but are separated by an intercondylar notch posteriorly. The anterior portions of the articular surface of both medial and lateral condyles combine to provide articulation for the patella. Although this patellar surface is continuous with the rest of the articular surfaces of the medial and lateral condyles, it distinguished from the tibiofemoral articular surfaces by a very slight mediolateral groove. The articular surface for the patella is concave in the medial-lateral direction with a distinct longitudinal groove through its midlind. It is convex in a superior-inferior direction. The anterior surface of the lateral condyle, which articulates with the patella, extends farther anteriorly than the anterior surface of the medial condyle, forming a buttress against lateral dislocation of the patella (Oatis, 2009).

The medial and lateral condyles are separated from one another by the intercondylar fossa on their distal and posterior surfaces where they articulate with the tibia. The medial and lateral walls of the intercondylar fossa provide attachments for the posterior cruciate ligament and anterior cruciate ligament, respectively. The surfaces of the two condyles are quite different from one another, which helps explain the complex motions of the tibiofemoral articulation (Oatis, 2009).

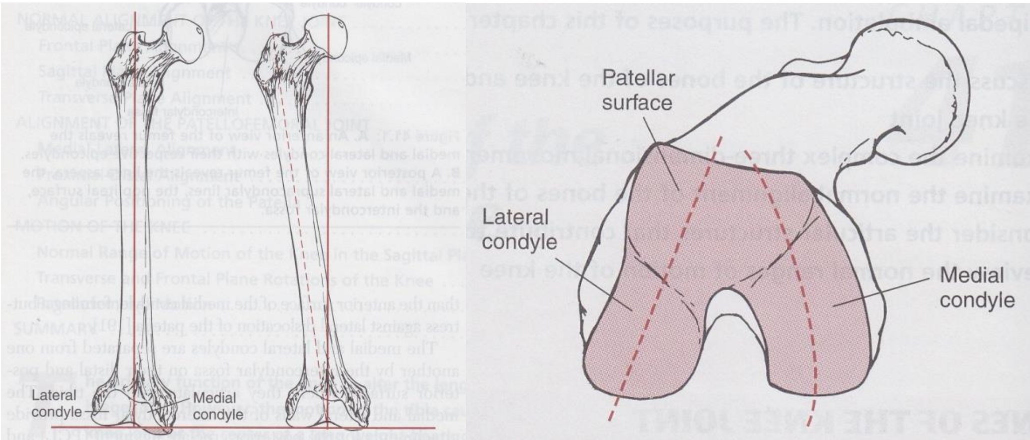


Figure 3 The lateral and medial condyle of femur (Oatis, 2009)

The medial condyle extends farther distally than the lateral condyle. However, because in the normal knee the two condyles lie on the same horizontal plane, the shaft of the femur forms a slight angle with the vertical (Figure 3). The proximal surface of the medial condyle is marked by the adductor tubercle, a palpable landmark where the adductor magnus attaches. The medial aspect of the medial femoral condyle offers an easily palpated apex known as the medial epicondyle (Oatis, 2009).

The shape and size of the tibiofemoral articular surface of the medial condyle distinguish it from the lateral condyle and influence the motions of the tibiofemoral articulation. The medial condyle is slightly curved in the transverse plane, as though it lies on a circle that surrounds the lateral condyle (Figure 3). The medial condyle’s articular surface for the tibia is longer from anterior to posterior than that of the lateral condyle’s articular surface. In addition, although the medial condyle is convex from anterior to posterior, its curvature is variable. It is flattest on its most distal surface and is more curved posteriorly. The articular surface for the patella also in more curved than the distal surface. In general, the radius of curvature is the radius of the circle from which the articular surface can be derived. Therefore, a flat surface is a segment of a very large circle with a large radius. A curved surface is part of a smaller circle with a smaller radius. Thus the radius of curvature of the medial condyle is greatest distally and is smaller on its posterior surface. This asymmetry in curvature contributes to the complex motion between the femur and tibia (Figure 2.4).

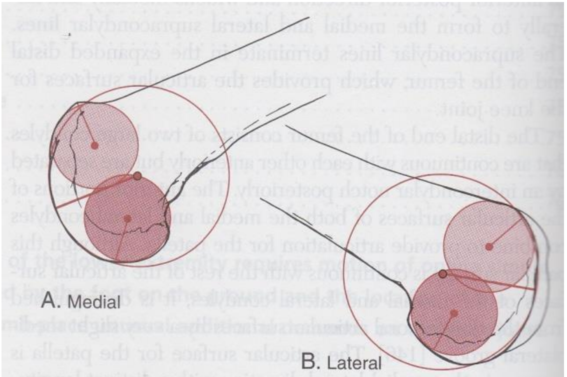


Figure 4 The radii of curvature of the lateral and medial condyle of femur (Oatis, 2009)

The lateral condyle's articular surface for the tibia projects posteriorly, more in the sagittal plane than the medial condyle. Like the medial condyle, the articular surface presents variable curvatures and, like the medial condyle, is flattest distally. The lateral femoral condyle is flatter distally than the medial condyle and hence has a larger redius of curvature. In the frontal plane, both condyles are slightly convex, but the lateral condyle is flatter than the medial one. The lateral aspect of the lateral condyle forms a prominent projection, the lateral epicondyle which is an important palpable landmark. The knee joint's axis of flexion and extension passes approximately through the lateral and medial epicondyles (Chyrchill, 1998).

2.4.1.2 The proximal tibia

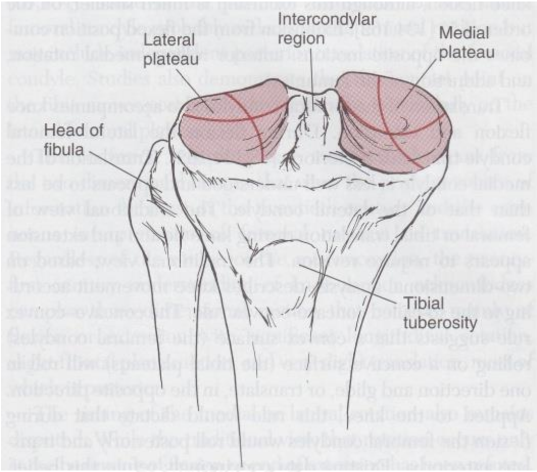


Figure 5 The proximal tibia (Oatis, 2009)

The tibia is the second longest bone of the body, exceeded only by the femur. It is characterized by an expanded proximal end that consists of medial and lateral condyles, or plateaus, separated by a nonartuculating intercondylar region (Figure 5). The nonarticulating region is roughened and consists of an intercondylar eminence and smooth intercondylar areas anterior and posterior to the eminence. Medial and lateral intercondylar tubercles, or spines, project proximally from the eminence. The intercondylar region provides attachment for the medial and the lateral menisci and the ACL and PCL. The anterior surface of the proximal tibia is marked by the tibia tuberosity, readily palpated since it is covered by only skin and the infrapatellar bursa. Just distal to the lateral tibial plateau and lateral to the tibial tuberosity in another tubercle, the tubercle of the lateral condyle of the tibia. A facet for the head of the fibula is located on the inferior surface of the lateral condyle. The facet faces laterally, distally, and slightly posteriorly (Oatis, 2009).

The articular surfaces of the proximal tibia for the femoral condyles consist of medial and lateral facets on the tibia plateaus. The proximal articular surfaces of the tibia are considerably smaller than the respective articular surfaces on the femur. Additionally, the articular surface on the medial tibial plateau is larger than the articular surface of the lateral tibia plateau, decreasing the stress applied to the medial tibial plateau which bears more force than the lateral plateau in upright stance (Ateshian, 1991; Riegger-Krugh, 1998).

The tibia's medial articular surface is slightly concave. However, it has a very large radius of curvature, indicating that it is relatively flat. The shape of the lateral articular surface is more variable. It is concave in the medial-lateral articular surface is more variable. It is concave in the medial-lateral direction but, like the femur, the lateral tibial plateau is flatter than the medial plateau. Although some authors report that the lateral articular surface also is concave in the anterior-posterior direction, direct measurements of cadaver knees suggests that the surface actually is flat or even convex throughout most of its anterior-posterior surface. Thus it is apparent that not only do the medial and lateral articular surfaces of the tibia differ from each other, they also differ from the respective articular surfaces of the femur. The differences in shapes of the articular surfaces of the tibiofemoral joint influence the loading pattern across the joint. The remaining differences among the articular surfaces influence the motion of the tibiofemoral joint. (Oatis, 2009)

2.4.1.3 Effects of the shapes of the articular surfaces on tibiofemoral joint motion

Three different factors regarding the shapes of the knee's articular surfaces affect the motion of the tibiofemoral joint such as the different size of the articular surfaces of the femoral condyles and the tibial condyles, the articular surface of the medial femoral condyle and the lateral femoral condyle, and the variation in curvature from anterior to posterior in all of articular surfaces. Each of these factors has a different impact on the motion that occurs at the tibiofemoral joint, and together they help to explain the complex three-dimensional motion that occurs during flexion and extension of the knee (Oatis, 2009).

2.4.1.3.1 Tibiofemoral motion

The complex shapes and incongruities of the tibiofemoral joint surfaces contribute to intricate three-dimensional movement of the femur and tibia during knee flexion and extension. The classic view of tibiofemoral motion is based on two-dimensional analyses that suggest that knee flexion begins with lateral rotation of the femur and concomitant anterior gliding of up to 2 cm (Oatis, 2009).

More recent three-dimensional analyses confirm the three-dimensional nature of the tibiofemoral movement during flexion and extension but provide more precise measurements of the frontal and transverse plane movements. The lateral rotation of the femur with respect to the tibia accompanies knee flexion reaching approximately 20 degrees of the lateral rotation as the knee moves from full extension to at least 90 degree of flexion (Figure 6). In addition, femoral abduction with respect to the tibia also occurs with knee flexion, although this excursion in much smaller, on the order of 5 degree. Extension from the flexed position combines the opposite motions: anterior rolling, medial rotation, and adduction of the femur (Oatis, 2009).

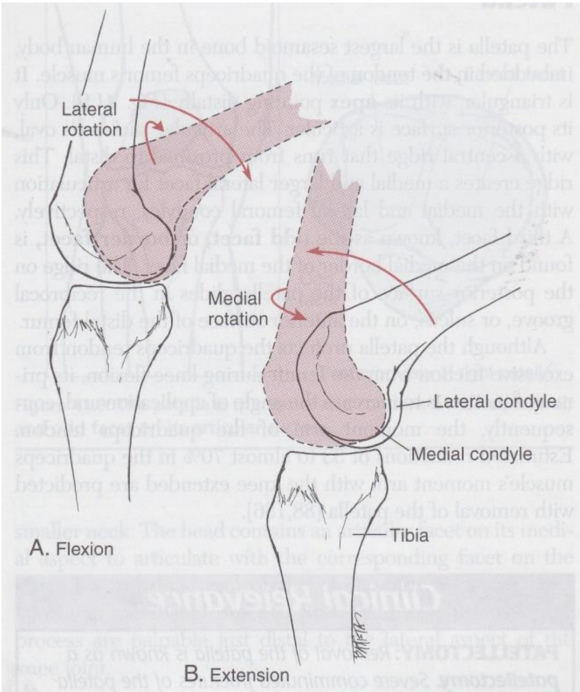


Figure 6 Lateral and medial rotation of femur during knee flexion and extension motion
(Oatis, 2009)

Translation of the femoral condyles also accompanies knee flexion and extension. During flexion the lateral femoral condyle translates posteriorly. Translation of the medial condyle is less well understood and appears to be less than that of the lateral condyles. The traditional view of femoral or tibial translation during knee flexion and extension appears to require revision. The traditional view, based on two-dimensional analysis, describes knee movement according to the concave-convex rule. This rule would dictate that during flexion the femoral condyles would roll posteriorly and translate anteriorly. Existing data convincingly refute this belief. Using three-dimensional imaging techniques, investigators consistently demonstrate substantial posterior translation of the lateral femoral condyle during knee flexion. Some of the posterior translation of the lateral femoral condyle probably reflects lateral femoral rotation but may include independent posterior translation of the femoral condyle. The contact point on the tibia moves posteriorly during flexion, particularly on the lateral-femoral plateau (Oatis, 2009).

The confusion regarding knee motion likely stems from the two-dimensional images that were the primary source of information for most of the twentieth century and the erroneous interpretation of femoral rotation as translation. Regardless of the source of the misconception, the twenty-first century understanding of knee motion acknowledges complex three-dimensional motion that consists mainly of flexion or extension with significant longitudinal rotation, slight frontal plane motion, and very slight translation, most of which is posterior (Oatis, 2009).

The timing of the medial or lateral rotation also remains disputed. While the traditional view that rotation occurs only at the beginning of flexion or end of extension has been refuted, some investigators suggest that there is an initial rotation at the beginning of flexion (or end of extension), which then ceases until at least 45 degree of flexion. Others suggest that the rotation continues smoothly through at least the first 90 degree of the motion. The screw home mechanism describes the final medial rotation of the femur as the knee reaches full extension. Whether this is a distinct femoral movement or the continuation of the femoral rotation throughout the range is unresolved (Oatis, 2009).

Despite the existing controversies, the tibiofemoral movement during knee flexion and extension exhibits characteristic components. First, during flexion, as the femur rolls into flexion, it rotates laterally with respect to the tibia. Conversely, the femur rotates medially as it rolls into extension. Second, contact between the femur and tibia migrates posteriorly on the tibia during flexion and anteriorly during extension. Finally, there appears to be some anterior-posterior translation between the tibia and femur during some portions of flexion and extension, although this translation may be small (Oatis, 2009).

The knee does not function as a pure hinge joint. It allows significant motion about the three axes, medial-lateral, anterior-posterior and longitudinal. Although the motion about the medial-lateral axis far exceeds the motion about the other two axes, all of the motions play a significant role in the function of the tibiofemoral articulation. In addition, the tibiofemoral joint allows translation along all three axes. Although only the anterior-posterior gliding that limited by the cruciate ligaments is well described, there is potential for a small amount of medial and lateral translation and slight distraction of joint along its long axis. Therefore, the motion of the tibiofemoral joint is an example of a joint with six degree of freedom (DOF), allowing rotation about, and translation along, three axes (Figure 2.7).

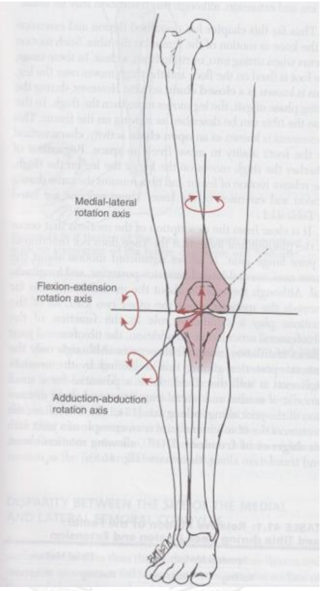


Figure 7 Knee rotation and translation about 3 axes and 6 degree of freedom. (Oatis, 2009)

2.4.1.4 Patella

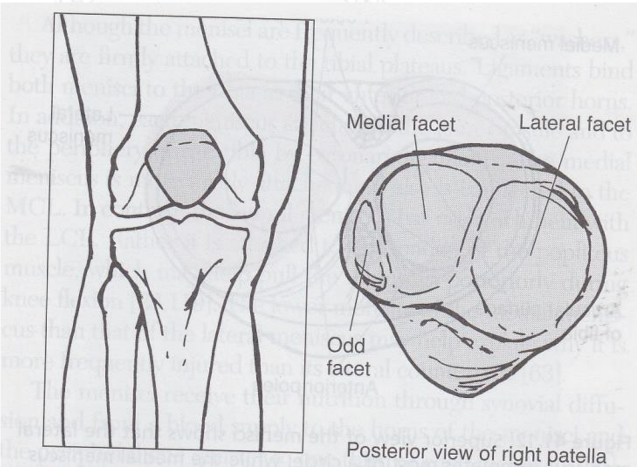


Figure 8 Patella (Oatis, 2009)

The patella is the largest sesamoid bone is the human body imbedded in the tendon of the quadriceps femoris muscle. It is triangular, with its apex pointing distally (Figure 8). Only its posterior surface is articular. The articular surface is oval with a central ridge that runs from proximal to distal. This ridge creates a medial and larger lateral facet for articulation with the medial and lateral femoral condyles, respectively. A third facet, known as the odd facet, or border facet, is found on the medial border of the medial facet. The ridge on the posterior surface of the patella glides in the reciprocal groove, or sulcus, on the anterior surface of the distal femur (Oatis, 2009).

Although the patella protects the quadriceps tendon from excessive friction from the femur during knee flexion, its primary function is to increase the angle of application and, consequently, the moment arm of the quadriceps tendon. Estimated reductions of 33 to almost 70% in the quadriceps muscle's moment arm with the knee extended are predicted with removal of the patella (Oatis, 2009).

2.4.2 Articular structures of the knee

The knee joint complex consists of the tibiofemoral and patellofemoral articulations. The proximal tibiofibular joint has an indirect effect on the knee as it functions to absorb motion at the foot and ankle. Although the tibiofemoral joint is often described as a hinge joint, it is more precisely a combination of hinge and pivot joints and is sometimes called a modified hinge joint. The patellofemoral joint is a gliding joint. The tibiofemoral and patellofemoral articulations share the same supporting structures but also exhibit unique features and motions. The following describes the functionally relevant characteristics of the articular cartilage, the menisci, and the non-contractile supporting structures of the entire knee joint complex (Oatis, 2009).

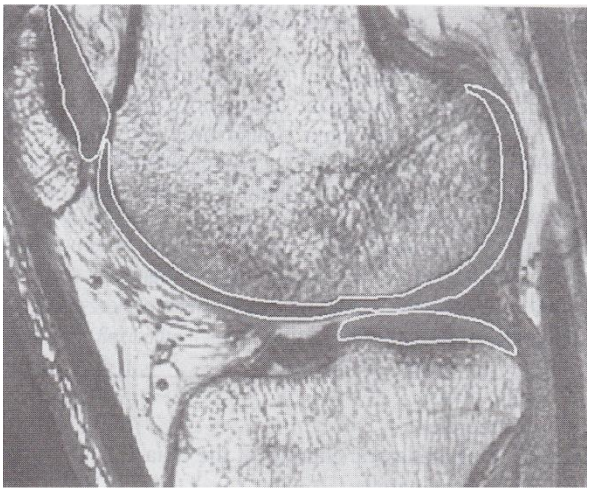


Figure 9 Knee joint cartilage. (Oatis, 2009)

2.4.2.1 Organization of the trabecular bone and articular cartilage found in the knee

The distal femur, proximal tibia, and patella all demonstrate trabecular bone whose organization is correlated with the forces and stresses applied to each bone. The organization observed in each bone suggests that the bones develop according to the forces applied to them and that each bone is specialized to sustain very large loads (Oatis, 2009).

The knee joint also possesses the thickest articular cartilage found anywhere in the body, even thicker than that found in the hip joint (Figure 9). Average thicknesses of between 2 and 3 mm. are reported for the patellar and tibial surfaces, with only slightly less on the distal femur. Peak thicknesses of approximately 6 mm and reported on the patella and tibia. The presence of such thick articular cartilage provides further evidence that these articulations sustain large forces. Thick articular cartilage allows considerable deformation of the articulating surface as well. The curvature of the patella in the superior and inferior directions is larger than the patellar surface of the femur. The compliance of the thick articular cartilage on the patella and the tibia helps improve the congruity between the articulating surfaces of the patellofemoral and tibiofemoral joint. Improved congruence increases the area of contact and thus reduces the stress applied to the surface. The knee exhibits additional specializations that appear designed to help minimize the stress across the tibiofemoral joint, namely, the menisci (Oatis, 2009).

2.4.2.2 Menisci

The two menisci are fibrocartilaginous discs seated on the medial and lateral tibial plateaus. The medial meniscus is larger in diameter that the lateral meniscus, consistent with the larger medial tibial plateau (Figure 10). The menisci cover more than 50% of the tibia plateaus, with the lateral covering a greater percentage of the plateau than the medial meniscus. As a result, there is more direct contact between the femur and tibia in the medial joint compartment than in the lateral compartment (Oatis, 2009).

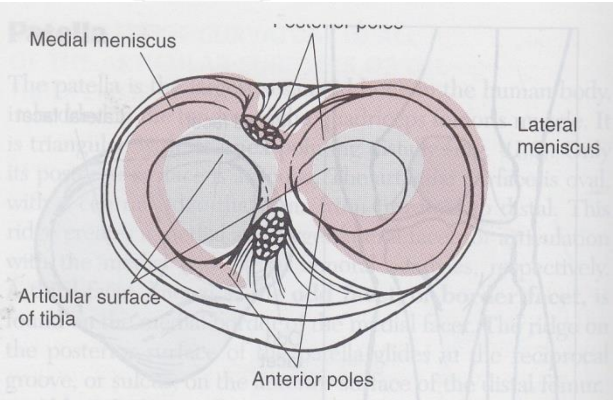


Figure 10 Superior view of the menisci. (Oatis, 2009)

When viewed from above, each meniscus forms part of a circle, the medial meniscus completes approximately a half circle, while the lateral forms almost a complete circle. The anterior and posterior ends of the arcs in each meniscus are known as anterior and posterior poles, or horns. The poles of the lateral meniscus are close to each other, while the poles of the medial meniscus are close to each other, while the poles of the lateral meniscus are close to each other, while the poles of the medial meniscus are farther apart. Viewed in the frontal plane, each meniscus is wedge shaped, thicker on its periphery and thin centrally, creating a concave surface for the femoral condyles (Figure 11).

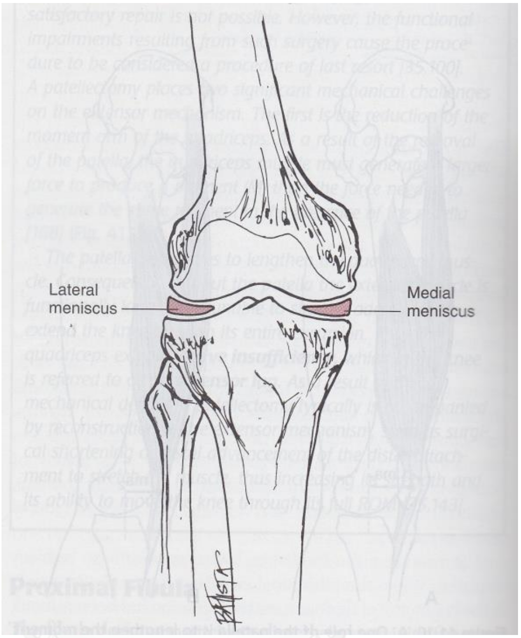


Figure 11 Front view of the menisci. (Oatis, 2009)

Although the menisci are frequently described as washers, they are firmly attached to the tibial plateaus. Ligaments bind both menisci to the tibia at their anterior and posterior horns. In addition, each meniscus attaches to the joint capsule and to the periphery of the tibia by coronary ligaments. The medial meniscus is more firmly attached and also is connected to the medial collateral ligament. In contrast, the lateral meniscus has no attachment with the lateral collateral ligament. Rather it is attached to the tendon of the popliteus muscle, which may help pull the meniscus posteriorly during knee flexion. The lower mobility of the medial meniscus than that of the lateral meniscus may help explain why it is more frequently injured than its lateral counterpart (Oatis, 2009).

The menisci receive their nutrition through synovial diffusion and from a blood supply to the horns of the menisci and the peripheral one quarter to one third of each meniscus. Therefore, lesions along the periphery demonstrate healing and even regeneration of meniscus-like tissue. The periphery of the menisci also appears to have sensory innervation that may extend into the more central portion of the discs. The sensory function appears to be mostly proprioceptive (Oatis, 2009).

Several functions are ascribed to the menisci, including; shock absorption, knee joint lubrication, and stabilization. However, the primary function of the menisci is to increase the contact area between the femur and tibia, thereby reducing the stress sustained the articular cartilage (Oatis, 2009).

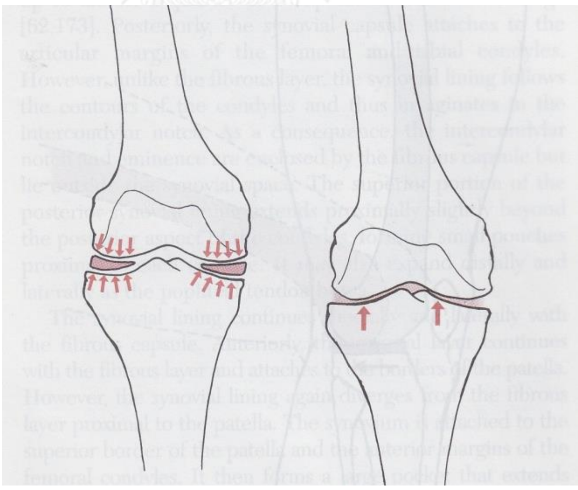


Figure 12 The menisci increase the contact area between tibia and femur. (Oatis, 2009)

Each meniscus is concave on its superior surface but relatively flattened inferiorly, reflecting the shapes of the femoral condyle and tibial plateau contacting it. Without the menisci, contact between the differently curved femoral condyle and the tibia plateau occurs over a very small area, leading to large stresses applied to the bones (Figure 12). The addition of a meniscus between the femoral condyle and tibial plateau approximately doubles the area of contact between the femur and tibia. As a result, the menisci significantly reduce the stress between the femur and tibia. Conversely, removal of a meniscus increases the stresses applied to the tibial plateau and femoral condyle. The stresses increase as more meniscal tissue is removed (Oatis, 2009).

The complex motion between the femur and tibia applies similarly complex loads to the menisci lying between the two long bones. These forces cause the menisci to deform and to glide on the tibia during knee motion. The motion of the menisci is consistent with their role as washers between the two bone surfaces. The menisci move in concert with the rolling femoral condyles (Figure 2.13). As the femur rolls posteriorly in knee flexion, the menisci are pushed posteriorly ahead of the rolling condyles. Similarly, they glide anteriorly ahead of the anteriorly rolling condyles during knee extension.

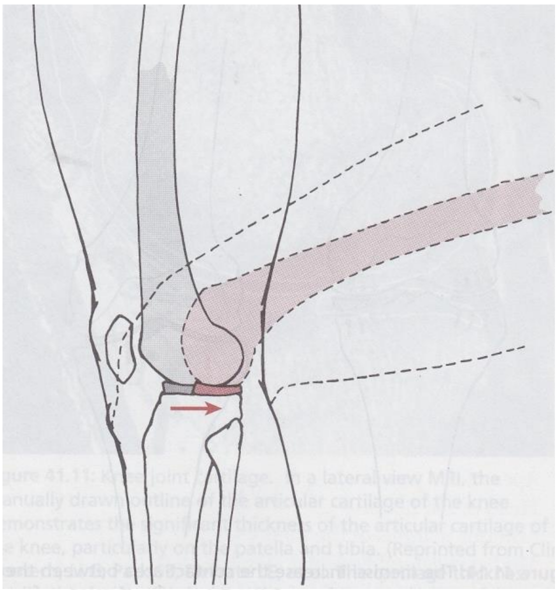


Figure 13 The menisci glide posteriorly with knee flexion and anteriorly with knee extension (Oatis, 2009)

The lateral meniscus moves farther than the medial meniscus because the latter is stabilized by attachments to the medial knee joint capsule, collateral ligament, and the tibial plateau by the coronary ligaments. Because the menisci remain attached at their poles as they slide posteriorly and anteriorly on the tibia, they also undergo considerable distortion in shape. This strain may contribute to eventual tears (Oatis, 2009).

2.4.3 Normal alignment of the knee joint

Alignment of the knee is affected by the alignment of the hip, ankle, and foot. This interaction is the result of the knee’s location between the ground on which the subject stands and the superimposed weight of the head, arms, trunk, and opposite lower extremity. Misalignment of the knee can result from malalignment of the hip, ankle, or foot joints, from muscle imbalances, and from abnormal loads on the knee joint. Conversely, there is evidence that knee joint deformities cause abnormal stresses on the joint and can lead to joint degeneration. Accurate identification of malalignments of the knee and associated deformities of adjacent joint is an essential part of a thorough musculoskeletal evaluation (Oatis, 2009).

2.4.3.1 Frontal plane alignment

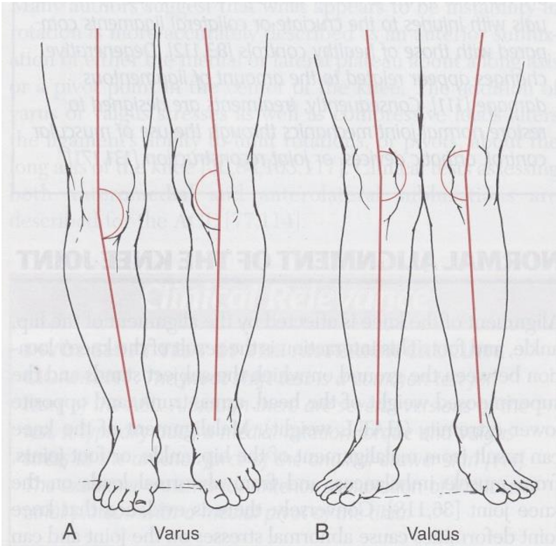


Figure 14 Varus (A) and valgus (B) deformity of knee. (Oatis, 2009)

The unique angulation of the knee joint in the frontal plane is considered a hallmark of bipedal ambulation. The medial femoral condyle extends father distally than the lateral femoral condyle. However, in the normal joint alignment, the distal surfaces of the two condyles lie on the same horizontal plane. Consequently, the shaft of the femur projects laterally from the vertical, putting the knees and feet are closer together than hip joints in normal erect standing. Frontal plan alignment is described by the term varus and valgus. Valgus is the alignment in which the angle between the proximal and distal segments opens laterally. In varus alignment, the angle opens medially (Figure 14).

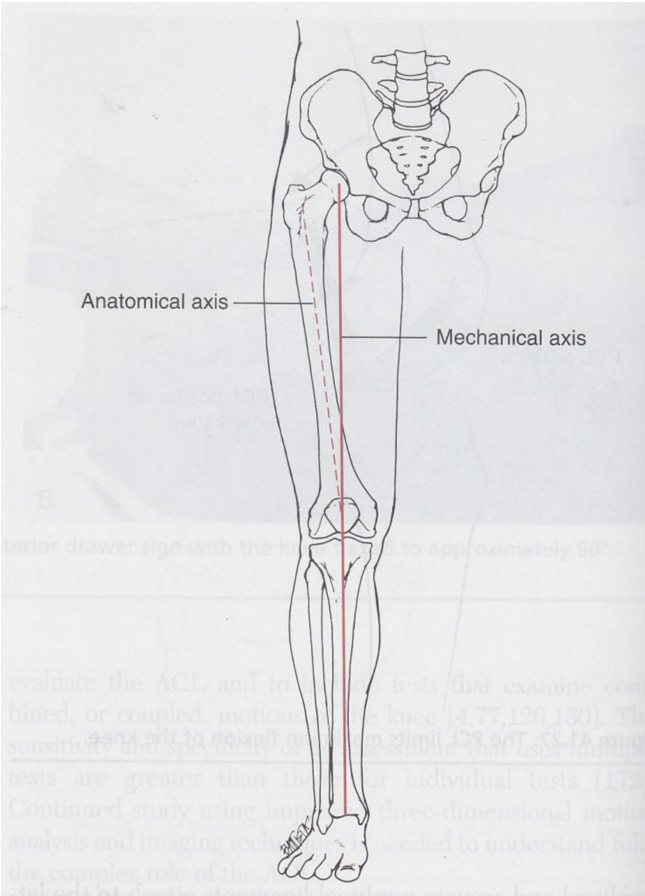


Figure 15 The anatomical and mechanical axes of knee. (Oatis, 2009)

The exact value of valgus or varus of the knee depends on the method of measurement. The measure can be made using the anatomical or the mechanic axes of the knee (Figure 15). The anatomical method uses the long axes of the femur and the tibia. The mechanical method uses the mechanical axes of the lower extremity. Using the anatomical axes, varus alignment is abnormal in adults and is often associated with degenerative joint disease. However, using the mechanical axes, the normal alignment of the knee is in approximately 2 degrees of varus. Location of the mechanical axes requires radiographic assessment. Therefore, normal frontal plane alignment measured in a physical examination is slight valgus. Normal valgus alignment of the knee results in a narrower base of support during stance, requiring less lateral shift to keep the body's center of mass over its base of support during single-limb stance and gait (Figure 16).

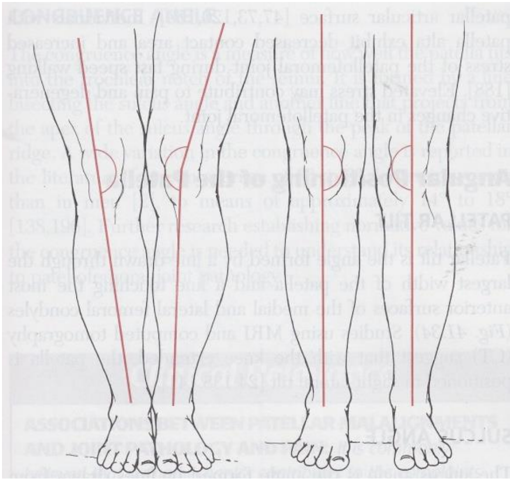


Figure 16 Valgus of knee allows the feet to be closed. (Oatis, 2009)

2.4.3.2 Sagittal plane alignment

Normal erect standing posture of the knee in the sagittal plane consists of a vertically aligned femur and tibia, together forming a 180 degrees angle. However, hyperextension of the knee in standing can occur, and the associated postural alignment is known as genu recurvatum (Figure 17). It frequently results from muscle imbalances at the ankle or knee. Such a posture applies increased stress to the posterior joint capsule of the knee and to the anterior cruciate ligament.

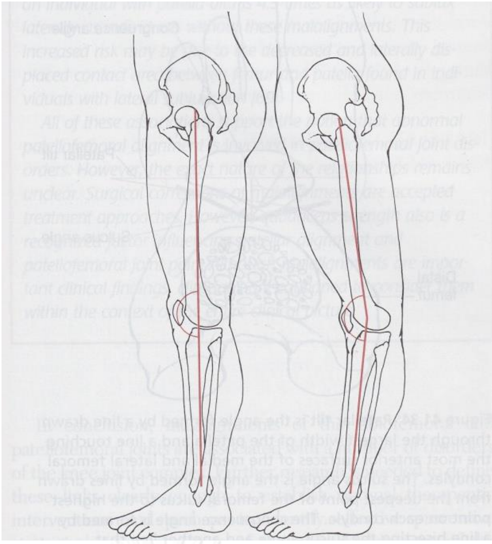


Figure 17 Genu recurvatum of knee. (Oatis, 2009)

2.4.3.3 Transverse plane alignment

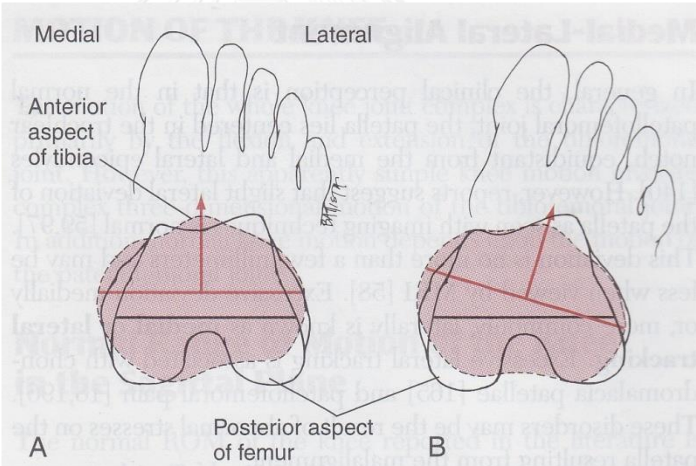


Figure 18 Transverse plane alignment (A) normal and (B) lateral version of the knee. (Oatis, 2009)

In a knee free from joint pathology, the tibial plateaus and femoral condyles are aligned so that with the knee extended, the transverse axes of the proximal tibia and distal femur are parallel. This is described as 0 degree of version of the knee (Figure 18). In individuals with osteoarthritis of the knee, 5 degrees of lateral tibial version with respect to the femur(Oatis, 2009).

2.4.4 Motion of the knee

The motion of the whole knee joint complex is characterized primarily by the flexion and extension of the tibiofemoral joint. However, this apparently simple knee motion involves complex three-dimensional motion of the tibiofemoral joint. In addition, normal knee motion depends upon the motion of the patellofemoral joint (Oatis, 2009).

The normal range of motion of knee is 132 – 144 degrees flexion and 0 – 1.6 degrees hyperextension. However, hyperextension occurs commonly in young children, and then gradually disappears in adolescence. In adults, age and gender appear to have little effect on knee range of motion, but obesity in negatively associated with knee flexion range of motion (Oatis, 2009).

Knee excursions during gait range from almost complete extension (approximately 1 degree in midstance) to 65-75 degrees in midswing. However, many common activities of daily living require more knee flexion. Stair ascent and descent use between 90 and 110 degree of flexion, rising from a chair requires approximately 90 degrees, getting in and out of a bath tub requires approximately 130 degree, and squatting can use up to 165 degrees (Oatis, 2009).

2.4.4.1 Transverse and frontal plane rotations of the knee

The knee joint allows, actually requires, medial and lateral rotation and abduction and adduction. However, there are limited and varied data describing the normal available range of motion in subjects without knee pathology. The challenge in establishing normative values of transverse and frontal plane motion of the knee stems from the technical difficulty of quantifying three-dimensional motion of such small excursions. In addition, these motions are significantly smaller than flexion and extension but are directly influenced by the position of the knee in the sagittal plane. Mean values of total medial and lateral rotation excursion vary from 12 – 80 degree when the knee is flexed. Despite this wide variation in reported means, studies consistently demonstrate a significant decrease in total rotation excursion when the knee is extended. Peak medial and lateral rotation also are approximately equal to one another. However, much less rotation occurs during normal locomotion. The total rotations during locomotion range from approximately 8 – 15 degree, with medial rotation occurring in the swing phase of gait. Frontal plane motion such as abduction, and adduction are less than rotation that approximately 10 – 20 degree and 5 degree during gait (Oatis, 2009).

2.4.5 Forces and Moments on the Tibiofemoral joint

The muscle force is a major contributor to the joint reaction force sustained by any joint. Once the muscle force is determined at a joint, static equilibrium equations can be used to calculate the joint reaction forces at the joint. Since muscle loads are a major contributor to joint reaction forces, it is not surprising that considerably higher loads of up to several times body weight are reported during activities such as walking = 3.03, jogging = 12.4, lifting = 2.12, squatting = 7.6, and ascending stairs = 4.25 (Oatis, 2009).

The joint reaction force at a joint frequently is reported in terms of its components of its components of axial, or compressive, loading as well as its shear forces in the anterior-posterior and medial-lateral directions. The compressive loads at the knee are far greater than the shear forces. The joint reaction force is of particular interest because it is regarded as an important contributing factor in the development of osteoarthritis. The knee joint is one of most common weight-bearing joints affected by osteoarthritis, and knee osteoarthritis is a leading cause of disability in aging adult. Therefore, it is important for the clinician to recognize the relationship between joint and muscle forces and their possible associations with osteoarthritis (Oatis, 2009).

The tibiofemoral joint exhibits 6 degrees of freedom and thus sustains forces and moments along and about the medial-lateral, anterior-posterior, and longitudinal axes. Moments about the medial-lateral and anterior-posterior axes are particularly relevant clinically. Moments about the medial-lateral axis tend to produce flexion or extension. An internal extension moment produced by the quadriceps balances the external flexion moment exerted by the ground reaction force during a squat. In the frontal plane, during normal locomotion the ground reaction force applies and external adduction moment on the knee during mid-stance. This adduction moment increases the forces applied to the medial tibial plateau and femoral condyle. The adduction moment increases in individuals with varus alignment of the knee and is associated with degenerative changes of the medial side of the knee joint, medial compartment knee osteoarthritis (Oatis, 2009).

In contrast, an individual who lacks adequate hip and knee joint stabilization in the frontal plane may sustain large external abduction moments are associated with medial knee pain and tears of the anterior cruciate ligament (Oatis, 2009).

One important element in linking joint forces and moments with subsequent joint degeneration is the area over which the force is applied. The ability of a joint to sustain joint reaction forces depends not only on the magnitude of the reaction force, but also on its location and how it is dispersed across the joint surface. The in congruity of the articular surfaces of the tibiofemoral joint directly affects the contact area of the knee and, consequently, the stress applied to the tibial surfaces. The normal medial compartment of the knee bears more of the joint reaction force than the lateral compartment. However, the overall articular surface is greater on the medial side of the joint than on the lateral surfaces. The difference over which tibial condyle sustains larger stress. The magnitudes of peak stress vary from 4 to 7 MPa at the hip during level walking. Additional research is needed to characterize the stresses at the knee in individuals with and without pathology (Oatis, 2009).

Obesity is a significant risk factor for knee osteoarthritis. This finding is logical, since body weight is a contributor to the compression forces on the knee. However, other factors including lower extremity alignment and walking patterns also affect the stresses at the knee (Oatis, 2009).

2.5 The gait cycle, the basic unit of gait

Gait is a cyclical movement that, once begun, possesses very repeatable events that continue repetitively until the individual begins to stop the motion. The steady-state movement of normal locomotion is composed of a basic repeating cycle, the gait cycle (Figure 19).

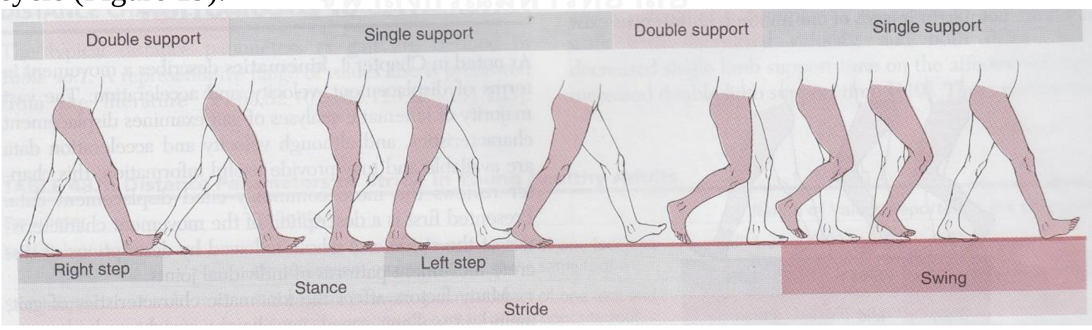


Figure 19 The gait cycle. (Oatis, 2009)

The cycle is traditionally defined as the movement pattern beginning and ending with ground contact of the same foot, the gait cycle begins when the right the heel of foot contacts the ground and ends when it contacts the ground again. Thus a gait cycle consists of the time the reference foot is on the ground called stance and the time it is of the ground called swing. The movement of both limbs that occurs during the gait cycle is known as the stride (Oatis, 2009).

The stance phase of gait makes up approximately 60% of the gait cycle, so that the remaining 40% consists of the swing phase. The gait cycle with respect to the right limb is slightly out of phase with the gait cycle of the left limb. At contact on the right, the left limb is just ending its stance phase. At approximately 10% of the gait cycle on the right, the left limb leaves the ground and begins its swing phase, returning to the ground at approximately 50% of the gait cycle of the right limb. Thus the gait cycle is characterized by two brief periods, each lasting approximately 10% of the gait cycle, in which both limbs are in contact with the ground. These are periods of double limb support, and the remaining cycle consists of single limb support (Oatis, 2009).

The stance phase can be divided into smaller periods associated with specific functional demands and identified by distinct events (Figure 20). The period immediately following ground contact is known as contact response, or weight acceptance, and ends when the whole foot flattens on the ground. During contact response, the limb absorbs the shock of impact and becomes fully loaded. The foot flat event that ends contact response occurs at approximately 15% of the normal gait cycle. It is important to recognize that loading response includes double limb support and continues into single limb support. The period following loading response is midstance, also known as trunk glide, since during this period the trunk glides over the fixed foot, moving from behind the stance foot to in front of it. Heel off ends trunk glide at approximately 40% of the gait cycle and begins terminal stance, which ends at 50% of the gait cycle when contralateral ground contact occurs. The final stage of stance from 50% to 60% of the gait cycle, is preswing and is characterized by double limb support. It ends with toe off (Oatis, 2009).

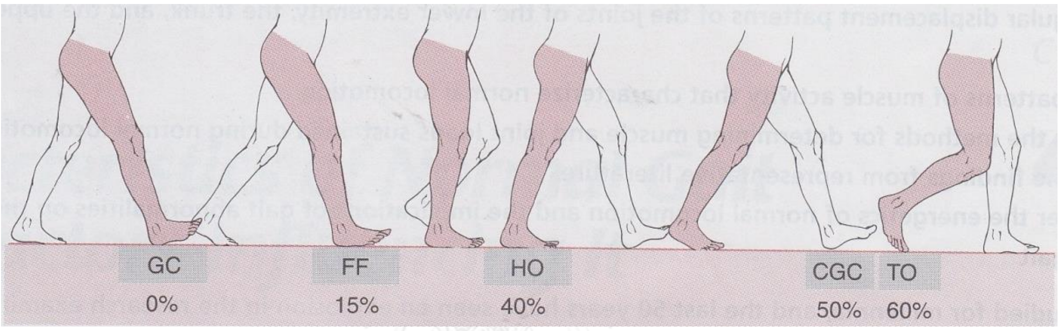


Figure 20 The stance phase of gait cycle. (Oatis, 2009)

The swing phase also is divided into early, middle, and late period, although it lacks distinctive events to delineate these phases (Figure 21). Early swing continues from 60% to approximately 75% of the gait cycle and is characterized by the rapid withdrawal of the limb from the ground. Midswing continues until approximately 85% of the gait cycle and consists of the period in which the swing limb passes the stance limb. Late, or terminal, swing finds the swing limb reaching toward the ground, preparing for contact (Oatis, 2009).

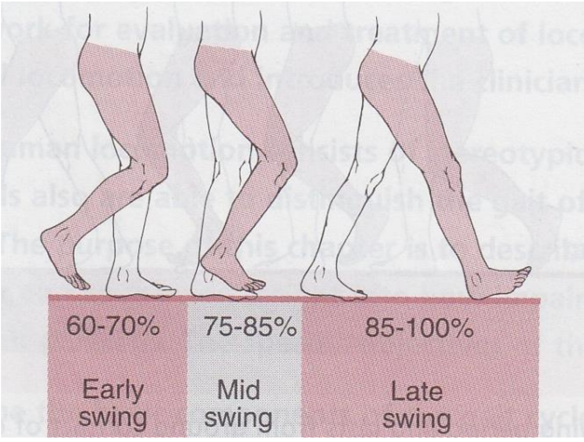


Figure 21 The swing phase of gait cycle. (Oatis, 2009)

Although the differences are small among ambulators without pathology, the right and left limb movements are not mirror images of one another. Differences exist in timing and movement patterns, in muscle activity, and in the loads applied to each limb. Asymmetry appears greatest at slower walking speeds. When evaluating the gait patterns of individuals with asymmetrical impairments, clinicians must remember that small asymmetries in gait are normal, particularly when walking slowly (Oatis, 2009).

Consideration of the basic functional tasks of the swing and stance phase of gait provides a framework for characterizing the movements in each phase of gait. While the overriding goal of locomotion is forward progression, the stance and swing phase contribute to that goal in different ways. The stance phase has three tasks in locomotion: providing adequate support to avoid a fall, absorbing the shock of impact between the limb and the ground, and providing adequate forward and backward force for forward progress. The basic tasks of the swing phase are safe limb clearance, appropriate limb placement of the next contact, and transfer of momentum (Oatis, 2009).

2.5.1 Kinematics of Locomotion

Kinematics describes a movement in terms of displacement, velocity, and acceleration. The vast majority of kinematic analyses of gait examines displacement characteristics, and although velocity and acceleration data are available and may provide useful information. Many factors affect the kinematic characteristics of gait, including walking speed, age, height, weight or body mass index, strength and flexibility, pain, and aerobic conditioning (Oatis, 2009).

2.5.1.1 Temporal and distance parameters of a stride

A stride consists of the movement of both limbs during a gait cycle and contains two steps. A step is operationally defined as the movement of a single limb from ground contact of one limb to ground contact of the opposite limb (Figure 22). There is considerable difference in step and stride characteristics among subjects and even among trials of the same subject. Despite this normal variability, these parameters are capable of distinguishing between individuals with and without impairments (Oatis, 2009).

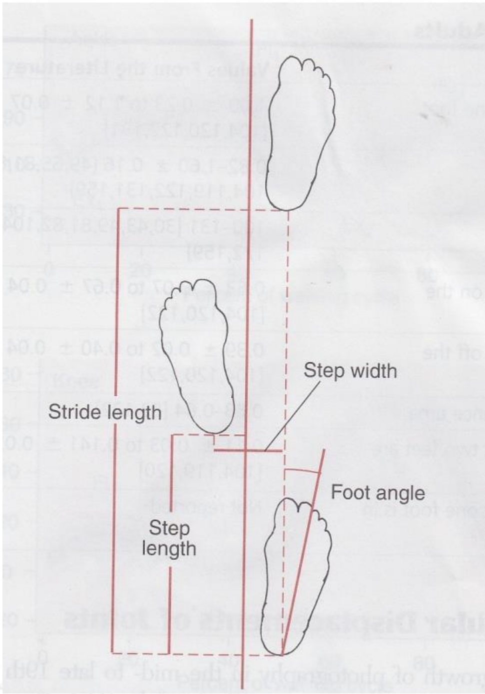


Figure 22 The temporal and distance parameters of stride. (Oatis, 2009)

The typical distance parameters of gait are defined in Table 1. Stride and step lengths depend directly upon standing height, so measures of absolute step or stride length, although frequently reported, are difficult to interpret. These measures can be normalized by standing height or lower extremity length to compare values from different individuals. Estimates of normalized stride length vary from approximately 60 to 110% of standing height. Judge et al. report a mean step length of 74 ± 4% of leg length in young healthy adults. Step width and foot angle are less frequently reported but provide and indication of the size of the base of support (Oatis, 2009).

Table 1 Distance characteristics of the stride (Oatis, 2009)

Parameter	Definition	Range of Values Reported in the Literature
Stride length	The distance between ground contact of one foot and the subsequent ground contact of the same foot	1.33 ± 0.09 to 1.63 ± 0.11 m
Step length	The distance between ground contact of one foot and the subsequent ground contact of the opposite foot	0.70 ± 0.01 to 0.81 ± 0.05 m.
Step width	The perpendicular distance between similar points on both feet measured during two consecutive steps	0.61 ± 0.22 to 9.0 ± 3.5 m.
Foot angle	Angle between the long axis of the foot and the line of forward progression	5.1 ± 5.7 to 6.8 ± 5.6 degree

Walking speed affects swing and stance time differently. Increased walking speed decreases the overall duration of the gait cycle, but the decreases the overall duration of the gait cycle, but the decreases in cycle duration results in a greater decrease in stance time than in swing time, double limb support time decreases, and single limb support time increases. The difference between running and walking is the absence of a double limb support phase in running. The ratio between swing and stance time increases toward 1 with increasing walking speed (Oatis, 2009).

Table 2 Temporal characteristics of the stride. (Oatis, 2009)

Parameter	Definition	Range of Values Reported in the Literature
Stride time	Time in second from ground contact of one foot to ground contact of the same foot	1.00 ± 0.23 to 1.12 ± 0.07 sec.
Speed	Distance/time, usually reported in m/sec	0.82 to 1.60 ± 0.16 sec.
Cadence	Steps per minute	100 to 131 steps/min.
Stance time	Time in seconds that the reference foot is on the ground during a gait cycle	0.63 ± 0.07 to 0.67 ± 0.04 sec.
Swing time	Time in seconds that the reference foot is off the ground during a gait cycle	0.39 ± 0.02 to 0.40 ± 0.04 sec.
Swing/stance ratio	Ratio between the swing time and the stance time	0.63 to 0.64
Double support time	Time in seconds during the gait cycle that two feet are in contact with the ground	0.11 ± 0.03 to 0.141 ± 0.03 sec.
Single support time	Time in seconds during the gait cycle that one foot is in contact with the ground	Not reported

2.5.1.2 Angular displacements of joints

The sagittal plane motions of the joints of the lower extremity are the most thoroughly studied and best understood, at least in part because sagittal plane motions are the largest and easiest to measure. In contrast, frontal and transverse plane motions of the joints of the lower extremities and the three-dimensional motion of upper extremities and trunk are less frequently studied. Joint displacement data reveal intra- and inter-subject variability in all planes, although the variability is greater in the frontal and transverse planes than in the sagittal plane and across subjects than between cycles of a single individual. The smaller excursions in the frontal and transverse planes are particularly sensitive to differences in measurement procedure, which accounts for some of the increased variability of these motions.

Despite the variability in magnitudes of the movements, the patterns and sequencing of joint movements in gait are remarkably consistent across trials and across subjects (Oatis, 2009).

2.5.1.2.1 Sagittal plane motions

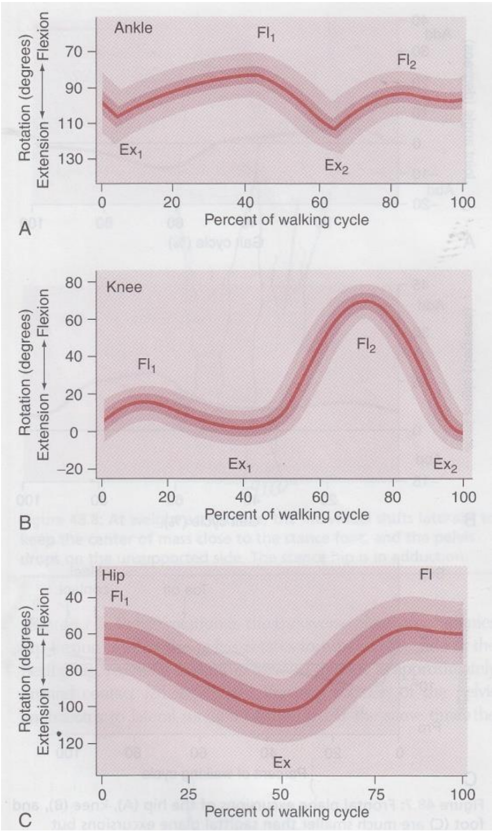


Figure 23 The sagittal plane excursions of the ankle, knee, and hips. (Oatis, 2009)

The classic studies by Murray remain the foundation for understanding sagittal plane motion of the lower extremity (Figure 2.38). More-recent studies confirm the overall patterns of motion for the hip, knee, and ankle, although there is variation in the reports maximal joint positions (Oatis, 2009).

The knee exhibits a slightly more complex movement pattern, landing in extension, albeit usually a few degrees shot of maximum extension, at ground contact. The knee flexes 10 to 20 degree immediately after contact, reaching maximum flexion at about 15% of the gait cycle when the subject achieves foot flat. At foot flat the knee begins to extend and reaches maximum extension at about 40% of the gait cycle as the heel rises from the ground. Flexion of the knee begins again and reaches a maximum of approximately 70 degree in midswing (approximately 75% of the gait cycle). Knee extension resumes, and the knee reaches maximum knee extension just before ground contact(Oatis, 2009).

2.5.2 Kinetics of Locomotion

Kinetics examines the forces, moments, and impulse generated during a movement and, in the case of locomotion, includes the moments generated by the muscles, the forces applied across joints, and the mechanical power and energy generated.

2.5.2.1 Joint moments and reaction forces

As indicated in the preceding sections, gait consists of complex cyclical movements occurring in a coordinated sequence that is controlled by muscle activity, In addition, gait entails the repetitive impact loading of both lower extremities in each gait cycle. Thus it is easy to recognize that normal locomotion produces large forces between the foot and the ground, requires significant muscle forces, and generates large joint reaction forces. Much impairment in gait is related to an individual's inability to generate sufficient muscular support or to sustain the large reaction forces of gait (Oatis, 2009).

2.5.2.2 Dynamic equilibrium

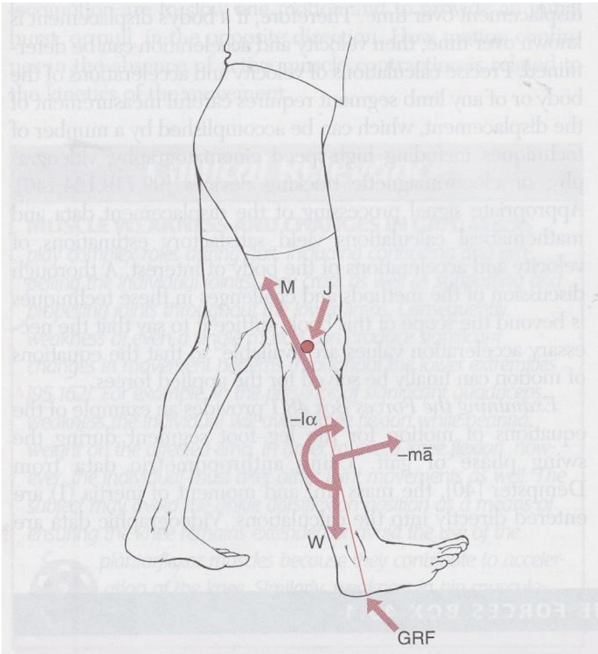


Figure 24 Free body diagram of the leg-foot segment during stance induces the forces. (Oatis, 2009)

Solutions to the conditions of dynamic equilibrium, also known as equations of motion, require knowledge of several parameters, including mass and moment of inertia. Theoretically, the equations of motion in dynamic equilibrium can be used to calculate a body's acceleration from all of the forces on the body. This approach is useful to determine the response of an airplane or rocket to an applied force. However, in the case of human movement, where forces cannot be measured directly, the equations of motion are used more often to determine the forces on the body when the accelerations are known. This approach, known as inverse dynamics, allows estimation of the forces on the human body and requires direct determination of the acceleration. Application of inverse dynamics in static equilibrium is straight forward because the accelerations are zero (Oatis, 2009).

The acceleration is the change of velocity over time, and velocity is the change in displacement over time. Therefore, if a body's displacement is known over time, then velocity are accelerations can be determined. During stance phase, the external force on the foot is applied call the ground reaction force (Figure 24). The direction and magnitude of the ground reaction force must be known to solve the equations of motion during stance and can be measured directly by force plate (Oatis, 2009).

There is co-contraction of the hamstrings and quadriceps during late swing and early stance and sometimes at the transition from late stance to early swing as well. Ligaments also apply significant loads to the knee joint during gait. Thus there is more than one structure applying force at the knee joint, producing a dynamically indeterminate system.

Moments in the transverse and frontal planes also are reported and appear to be important in the mechanics and pathomechanics of locomotion. However, fewer consensuses exist regarding the magnitude and even the pattern of these moments. Moments in the frontal and transverse planes are smaller than those in the sagittal plane, and smaller moments are more sensitive to measurement errors, including the location of the joint axes and the kinematics of the movements (Oatis, 2009).

The external frontal plane moment at the knee, known as the knee adduction moment, has been implicated in the development and progression of knee osteoarthritis as well as the pain and disability associated with knee osteoarthritis (Figure 25). Treatments to reduce the adduction moment and reduce pain include lateral shoe wedges, knee braces that apply a valgus stress to the knee, and surgical osteotomy to realign the knee joint (Oatis, 2009).

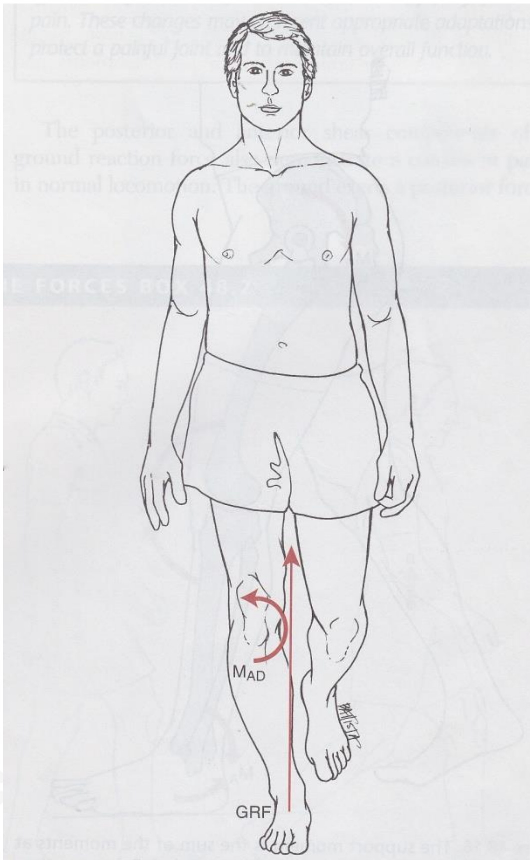


Figure 25 Adduction moment on the knee in gait. (Oatis, 2009)

2.5.2.3 Ground reaction forces

With every stride, each foot applies a load to the ground and the ground pushes back, applying a ground reaction force to each foot. The magnitude and direction of this ground reaction force changes throughout the stance phase of each foot and is directly related to the acceleration of the body’s center of mass. The center of mass of the body rises and falls as the individual moves from double support when the center of mass is low to single support when the center of mass is high. Similarly, the center of mass moves from side to side as the individual passes from stance on the right to stance on the left. The ground reaction force is measured directly by force plates imbedded in the walking surface(Oatis, 2009).

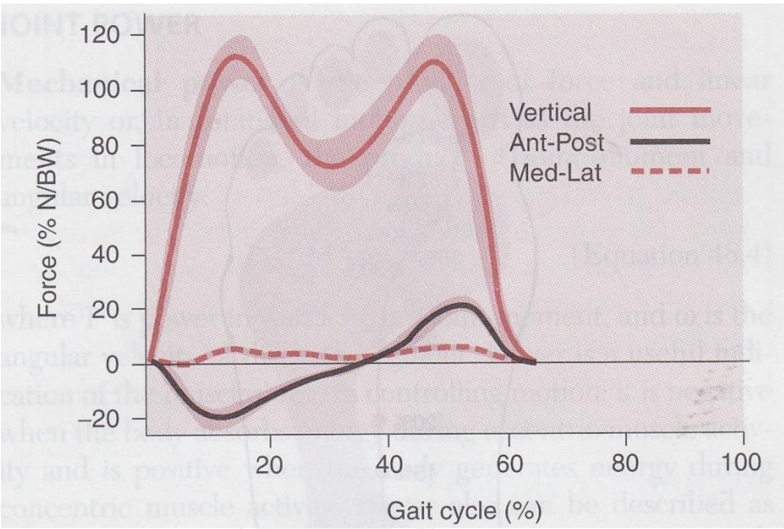


Figure 26 Ground reaction force during gait. (Oatis, 2009)

The ground reaction force typically is described by a vertical force as well as anterior-posterior and medial-lateral shear forces. The vertical ground reaction force of one foot is characterized by a double-humped curve (Figure 26). The two peaks are greater than 100% of body weight and occur when the body accelerates upward. The valley between the peaks is less than 100% of body weight and occurs during single limb support. The vertical ground reaction force also is characterized by a brief but high peak just following ground contact, which reflects the impact of loading(Oatis, 2009).

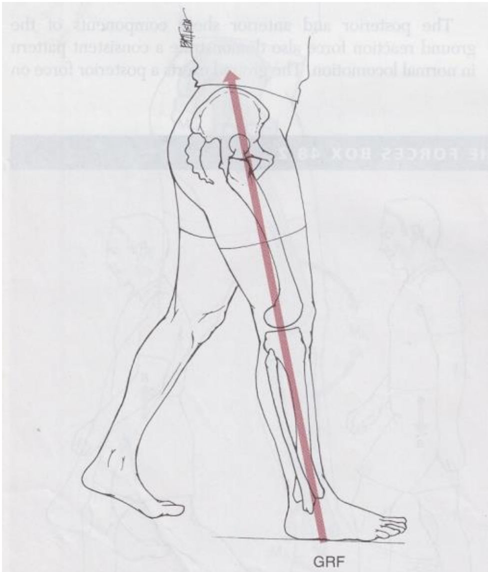


Figure 27 Ground reaction force vector during gait. (Oatis, 2009)

The ground reaction force vector is the sum of the individual components of the ground reaction force. Whether described as a single force vector or as three individual components, the ground reaction force generates external moments on the joints of the body in all three planes (Figure 27). Realistic computation of joint moments and forces during gait must include the three components of the ground reaction force or the force vector(Oatis, 2009).

The location of the ground reaction force with respect to the foot indicates the path of the center of pressure through the foot. In the normal foot, the center of pressure progresses in a relatively straight line from the posterior aspect of the plantar surface of the heel through the midfoot and onto the forefoot where it deviates medially onto the plantar surface of the great toe (Figure 28).

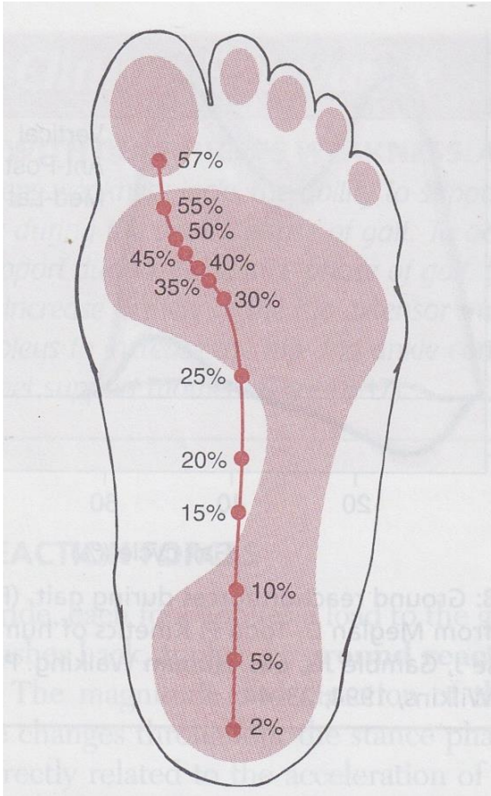


Figure 28 Progression of the center of pressure during locomotion. (Oatis, 2009)

2.5.3 Factors that influence parameters of gait

Factors affect to gait performance are gender, speed, and age. Although most observers would report the difference of gait pattern between male and female. Women walk with higher cadences than men and shorter strides. Yet when the distance characteristics of the gait cycle are normalized by height, females demonstrate a similar or slightly larger stride length(Oatis, 2009).

A study directly comparing 99 males and females of similar ages reports statistically different joint kinematics, although these differences are on the order of 2 – 4 degree, and the clinical significance of these differences is negligible(Kerrigan, 1998). The same study also reports that female exhibit a statistically greater flexion moment is preswing with increases in power absorption or generation at the hip, knee, and ankle. A similar study found no gender differences in flexion or adduction moments at the knee during stance (Kerrigan D. C., 2000). These studies suggest that while slight differences exist in some kinetic variables of gait, none of these differences are enough to explain the higher incidence of knee osteoarthritis in women(Oatis, 2009).

Gait speed affects several parameters of gait performance. The temporal and distance characteristics of gait, cadence, step length, and stride length increase with increased walking speed and decrease with decreased speed. Increased speed appears to increase the variability of some temporal and spatial gait parameters such as step width. Angular excursions generally increase with walking speed, but changes vary with the joint and the direction of motion. Increases in joint excursions at the proximal joints are related to the increase in stride length associated with increased speeds (Oatis, 2009).

2.6 Lateral wedge insole

Figure 29 Laterally wedged insoles. (Russell, 2011)

Modified footwear and orthosis have been investigated as potential conservative management of knee osteoarthritis (Radzimski, 2011, Pagani, 2010, Erhart, 2008, Jenkyn, 2011, and Erhart, 2008). More specifically, lateral wedging has been used with goal of reducing symptoms associated with medial knee osteoarthritis, hypothetically by reducing the peak external knee adduction moment. (Radzimski, 2011, and Russell, 2011).

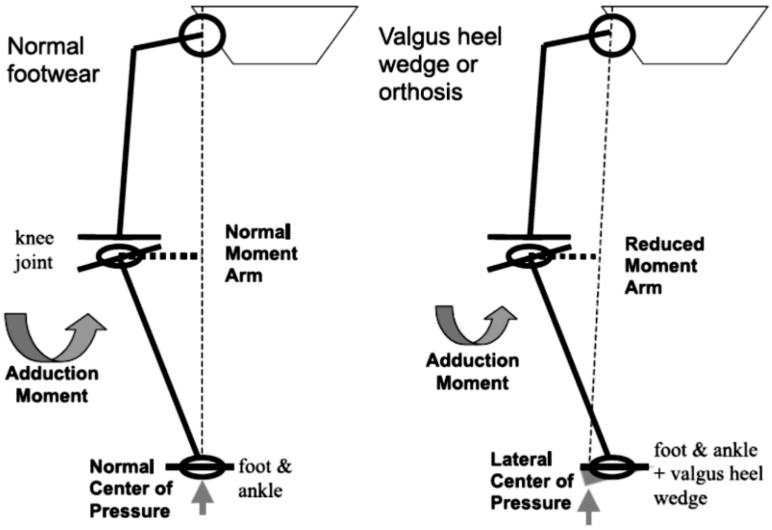


Figure 30 Laterally wedged insoles induce center of pressure lateral shift for reduce adduction moment of knee (Maly, 2002)

Laterally wedged insoles (Figure 29) are designed to laterally shift the center of pressure under the foot and shift the mechanical axis of the limb such that the moment arm of the external knee adduction moment decreases and the load distribution shifts towards the lateral compartment of the knee joint (Figure 30). The center of pressure shift may decrease the peak medial compartment load and its surrogate measure, the peak external knee adduction moment. Even small (1 mm.) shifts in the center of pressure under the foot can significantly decrease the external knee adduction moment and peak medial compartment load (Russell, 2011).

CHAPTER III
MATERIALS AND METHODS

This dissertation was divided into three studies for finding the answers for three research questions included:

1. Is there the difference of gait kinetic and kinematic due to the external knee adduction moment among obese, overweight, and normal healthy subjects?

2. Is there the correlation among anthropometry, biomechanical parameters during walking, and female BMI level?

3. Does orthosis effect on gait kinetic and kinematic due to the external knee adduction moment among obese, overweight, and normal healthy subjects?

3.1 Study 1

3.1.1 Objective of Study

To compare gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) among obese, overweight, and normal healthy subject during stance phase of gait cycle.

3.1.2 Hypothesis

There was no different of gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) among obese, overweight, and normal healthy subjects during stance phase of gait cycle.

3.1.4 Design: Cross-sectional design

Setting: Department of Physical Therapy,

Faculty of Allied Health Sciences,

Chulalongkorn University

Tools: 3-D Motion analysis system and Force transducer (BERTEC)

3.1.5 Subject:

Eighty Thai females aged between 18 to 40 years, participated in this study. The sample size was calculated by software G*Power version 3.0.10. All of them were explained about objective, process of testing, the equipment components and safety features. Subjects were informed about procedures and signed informed consent in accordance with the Ethics Review Committee for Research Involving Human Research Subjects, Health Science Group, Chulalongkorn University.

Inclusion criteria:

- BMI of subjects were $\geq$ 18.5 kg / m^2
- Free of lower extremity injuries affecting gait and without knee pain
- Had never been previously diagnosed with knee osteoarthritis, experience, or sought medical treatment for knee osteoarthritis related symptoms.
- weight change < 2.5 kg net change during the previous 3 months (Browning, 2009)
- Foot morphology was normal
- Static knee alignment axis in frontal plane (valgus and varus) lesser than 10 degree (Vanwanseele, 2009)
- Sedentary lifestyle (Messier, 2011)

  - sitting activity > 4 hrs/day

- no or irregular exercise

- Physical activity (Holliday, 2011)
  - work standing < 1 hr/day
  - lifting 25 kg < 10 times/week or 50 kg < 1 time/week
  - kneeling or squatting < 1 hr/day
- Hormone estrogen (Holliday, 2011)
  - Normal menstruation cycle
  - Non using estrogen replacement therapy
  - Non using contraceptive pill
  - Non pregnancy
  - No history of hysterectomy
- No history of metabolic - disease (Sturmer, 2000; Holliday, 2011)
  - Hypertension : Systolic / Diastolic blood pressure > 160/95 mm.Hg. or Use of anti-hypertensive drug
  - Diabetes Mellitus : History of Diabetes or use of oral antidiabetics or insulin
  - Heart disease
  - Hypercholesterolemia : Serum Cholesterol level > 6.2 m.mol./L or Use of anti-hyperlipidemic drugs
- No history of kidney problem (Holliday, 2011)
- No history of cancer (Holliday, 2011)
- No history of Thyroid disease (Holliday, 2011)

Exclusion criteria:

- Cannot complete the testing protocol

The World Health Organization (WHO), the International Association for the Study of Obesity (IASO), The International Obesity Task Force (IOTF) and the Steering Committee of the Regional office for the Western Pacific Region of WHO (WPRO) recommended that adult overweight could be specified in Asia when the BMI exceeded 23.0 kg/m^2 and that obesity should be specified when the BMI exceeded 25.0 kg/m^2. (WHO/IASO/IOTE, 2000) This Asian Classification was adopted in this study. Eighty subjects were divided to 20 subjects for each 4 groups by BMI classification into normal, overweight, obesity I and obesity II, respectively.

Table 2 Classification of obesity for Asian adults (WHO/IASO/IOTE, 2000)

Classification	BMI (kg/m^2)
Underweight	< 18.5
Normal	18.5 – 22.9
Overweight	23 – 24.9
Obesity I	25 – 29.9
Obesity II	$\geq$ 30

3.1.6 Testing Protocol

Before all else, subjects were asked to fill in a questionnaire form to answer about the general history, history of illness and exercise activity. The information was used to screen subjects before subsequent tests. All of practice and testing trials were performed in the Biomechanics Laboratory at Department of Physical Therapy, Faculty of Allied Health Sciences, Chulalongkorn University. The test procedure was read carefully to all subjects by researcher. Then subjects who were included criteria were informed about the aim of study, experimental procedure and possible risks before signing the consent form.

3.1.6.1 Subject preparation

Each subject was determined anthropometry and reduced anxiety by explaining the detail of the test and the involved equipment. Body composition and body weight were measured by body composition analyzer that using bioelectrical impedance analysis (BIA) principle to show percent body fat, fat free mass, and visceral fat of each subjects. Standing height was measured by height meter in centimeter. BMI of each subjects were calculated by own body weight in kilogram divided by the square of height in meter. Leg length without shoes (the distance from the greater trochanter to the ground during quiet standing) and circumference were determined by measuring tape in centimeter. Knee width (width between medial and lateral femoral condyles), and ankle width (width between medial and lateral malleoli) of subjects were measured(Lai, 2008). The waist to hip ratio was using a measuring tape to measure the circumference of hips at the widest part of buttocks. Then measure waist at the smaller circumference of natural waist. To determine the ratio, divided waist measurement by hip measurement. The data were corrected and used to compare for subjects.

The clinical practice guideline of the Royal College of Orthopaedic Surgeons of Thailand was used to screening physical examination for subjects who without knee osteoarthritis in to this study that was adapted from the American College of Rheumatology (The Royal College of Orthopaedic Surgeons of Thailand, 2012). The guideline has established clinical criteria for diagnosing primary osteoarthritis of the knee included symptom knee pain and finding ostephytes in knee X-ray image with at least 1 of the following 3 criteria: 50 years of age or older, morning stiffness lasting less than 30 minutes, and crepitus. The range of motion of back, hips, knee, and ankle in all direction were determined.

All subjects were requested to wear shorts, and tight fitting T- shirt. Elastic stockinet was rolled over the thigh and leg sections of each lower limb to minimize the vibration of the muscles in walking(Lai, 2008). To track the motions, spherical retro-reflective markers were placed on anatomical landmarks and segment that following the Helen Hayes marker set by researcher who had over 10 years experience identifying anatomical landmarks (Wilken, 2011).

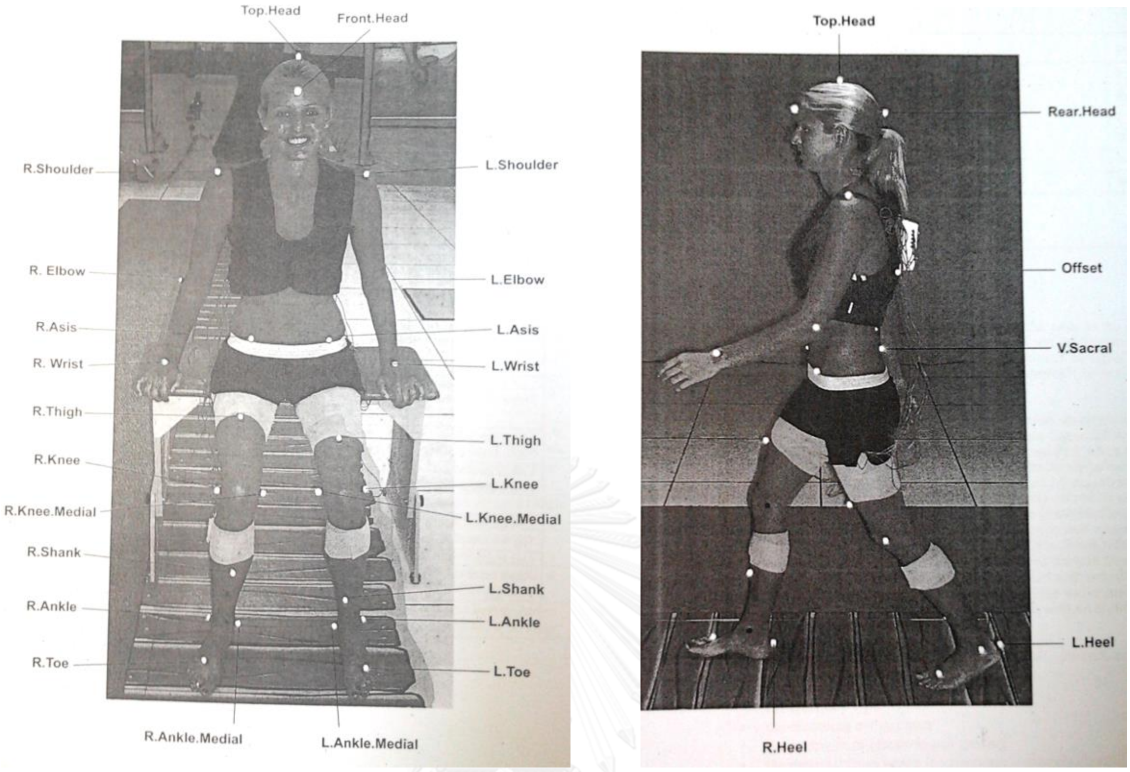


Figure 31 Helen Hayes Maker Set Static and Walking Trials

Table 3 Helen Hayes Marker Set with static trials marker list

Description	Cortex Marker name	Static	Walking			Placement
			Lower body	Full body	Full body with head	
Top of the head	Top. Head				+	On the center top of the head, in line with the back and front markers
Back of the head Front of the head	Rear. Head Front. Head				+	On the front and back of the head at the same height above the floor
Left shoulder Right shoulder	L. Shoulder R. Shoulder			+	+	Tip of the Acromion Process
Left elbow Right elbow	L. Elbow R. Elbow			+	+	Lateral Epicondyle of the Humerus
Left wrist Right wrist	L. Wrist R. Wrist			+	+	Centered between the Styloid Processes of the Radius and Ulna
Offset	Offset	+		+	+	Right trunk
Left ASIS Right ASIS	L. ASIS R.ASIS	+	+	+	+	Anterior Superior Iliac Spine
Sacrum	V.Sacral	+	+	+	+	Superior Aspect at the L5-Sacral interface
Left thigh wand Right thigh wand	L.Thigh R.Thigh	+	+	+	+	On lower thigh below the mid-point, for greatest visibility by all cameras
Left lateral knee Right lateral knee	L.Knee R.Knee	+	+	+	+	Along the flexion/extension axis of rotation at lateral femoral condyle
Left shank wand Right shank wand	L.Shank R.Shank	+	+	+	+	On lower shank below the midpoint, for greatest visibility by all cameras
Left lateral ankle Right lateral ankle	L.Ankle R.Ankle	+	+	+	+	Along the flexion/extension axis of rotation at lateral malleolus
Left heel Right heel	L.Heel R.Heel	+	+	+	+	Posterior Calcaneus at same hight from floor as toe marker
Left toe Right toe	L.Toe R.Toe	+	+	+	+	Center of the foot between the 2nd and 3rd metatarsals
Left medial ankle Right medial ankle	L.Ankle.Medial R.Ankle.Medial	+				Along the flexion/extension axis of rotation at medial malleolus
Left medial knee Right medial knee	L.Knee.Medial R.Knee.Medial	+				Along the flexion/extension axis of rotation at medial femoral condyle

3.1.7 Equipments and experimental set up

Temporo-spatial gait, 3-dimension kinetic and kinematic data were collected using an eight camera Motion Analysis System (Motion Analysis Corp., Santa Rosa, CA) that have flash rate of 120 Hz. The cameras were synchronized with the BERTEC force platform, which was set to have a sampling frequency of 1,200 Hz. Cortex version 2.5 was software created by the Motion Analysis corporation for use with their Motion Analysis motion capture system that combines 3 major functions into a single software package such as calibration of capture volume, tracking and indentifying marker locations in calibrated 3D space and post processing tools for tracking, editing, and preparing the data for other packages. Then the data were processed with clinical gait evaluation module by OrthoTrak 6.6 Gait Analysis Software. Static and dynamic calibrations were performed before experimental trial.

Subjects were asked for perform static trial to system detect the reflective markers. Then, four medial marker of knees and ankles were removed for prepare walking trial.

Subjects were allowed to walk barefoot at 3 speed such as the self-selected speed, standard speed of 1.24 m/s, and a Froude (FR) velocity that base on subject's leg length (length in centimeters from the greater trochanter to the floor while subject was standing) and g = 9.8 m/s^2, can calculate by following equation (Russell, 2011, and Wilken, 2011).

$$FR\ velocity = 0.04 \times \sqrt{g \times leg\ length}$$

Subjects were required to walk along 10 m. walkway. Subjects were given enough time to warm up and became familiar with each velocity. The sequence of walking velocity for each subject start with self selected velocity then the order of the other velocity was randomized prior to the test. Five minute rest between trials was performed. Five successful trials of a velocity condition were collected for each subject (Cofre, 2011). Successful trials were defined as the trials which subject walk with complete contact with the force plate.

3.1.8 Main Outcome Measures

The data were corrected from right side of lower extremity included;

Spatio-temporal parameter:

- step length	- step width
- stride length	- cadence
- double support time	- support time
- non support time	- velocity

Kinematics: knee joint angle

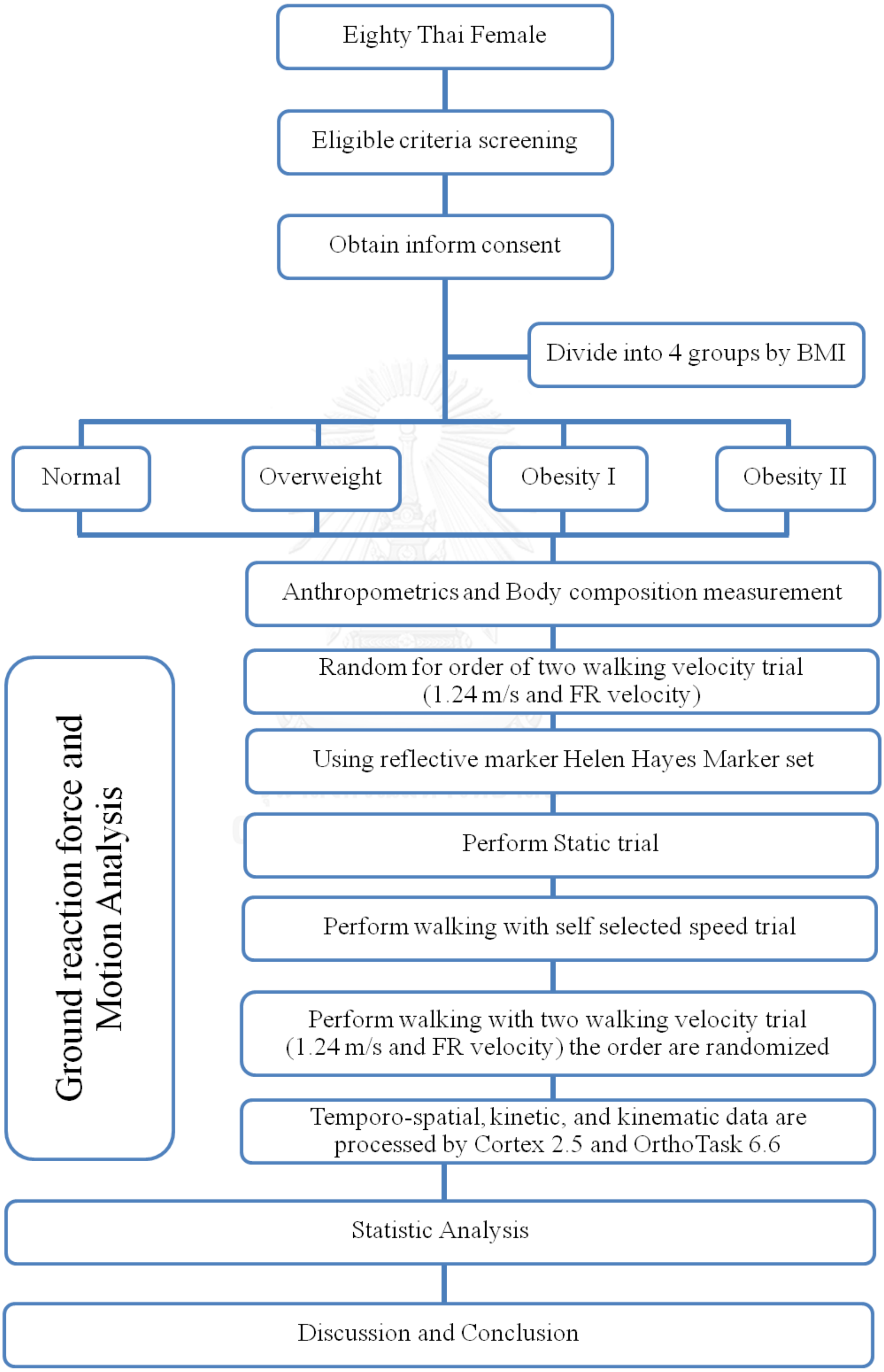
Kinetics: peak vertical ground reaction force, peak external knee adduction moment, and knee adduction angular impulse

3.1.9 Statistical Analysis

Cortex version 2.5 and OrthoTrak Gait Analysis version 6.6 software was used to define gait parameters and estimate kinetic and kinematics data. Statistical analysis was performed with the SPSS 17.5 for Window. The level of significance was set at p-value $\leq$ 0.05. Subject characteristic data were expressed as mean $\pm$ standard deviation (SD) value. Investigation data were expressed as mean $\pm$ standard error of mean (SEM) value by average from five complete trials. The normal distribution of data was tested by Kolmogorov-Smirnov test. The difference of mean between each

BMI groups were test by two-way ANOVA and Bofferoni Post Hoc test. The correlation of anthroometric, biomechanics parameter and BMI were determined by Pearson correlation test.

3.1.10 Experimental Design



3.2 Study 2

3.2.1 Objective of Study

To study the effect of orthosis on gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) in obese, overweight, and normal healthy subject.

3.2.2 Hypothesis

There was effect of orthosis on gait tempero-spatial data, kinetic and kinematic data (external knee adduction moment) in obese, overweight, and normal healthy subject.

3.2.3 Design**:** Cross-sectional design

Setting**:** Department of Physical Therapy,
Faculty of Allied Health Sciences,
Chulalongkorn University

Tools**:** 3-D Motion analysis system
(Motion Analysis Corp., Santa Rosa, CA)
Force transducer (BERTEC)
Shoe (Reebok^{TM} - Model V53843)
Lateral wedge insole (5^{o} and 70 Shore hardness, Unique material formulation SureStep-Control^{TM} EVA - Salfordinsole^{TM}, UK)

3.2.4 Subject:

Eighty Thai females aged between 18 to 40 years, participated in this study. The sample size was calculated by software G*Power version 3.0.10. All of them were explained about objective, process of testing, the equipment components and safety features. Subjects will be informed about procedures and signed informed consent in accordance with the Ethics Review Committee for Research Involving Human Research Subjects, Health Science Group, Chulalongkorn University.

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  - Normal menstruation cycle
  - Non using estrogen replacement therapy
  - Non using contraceptive pill
  - Non pregnancy
  - No history of hysterectomy
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3.2.5 Testing Protocol

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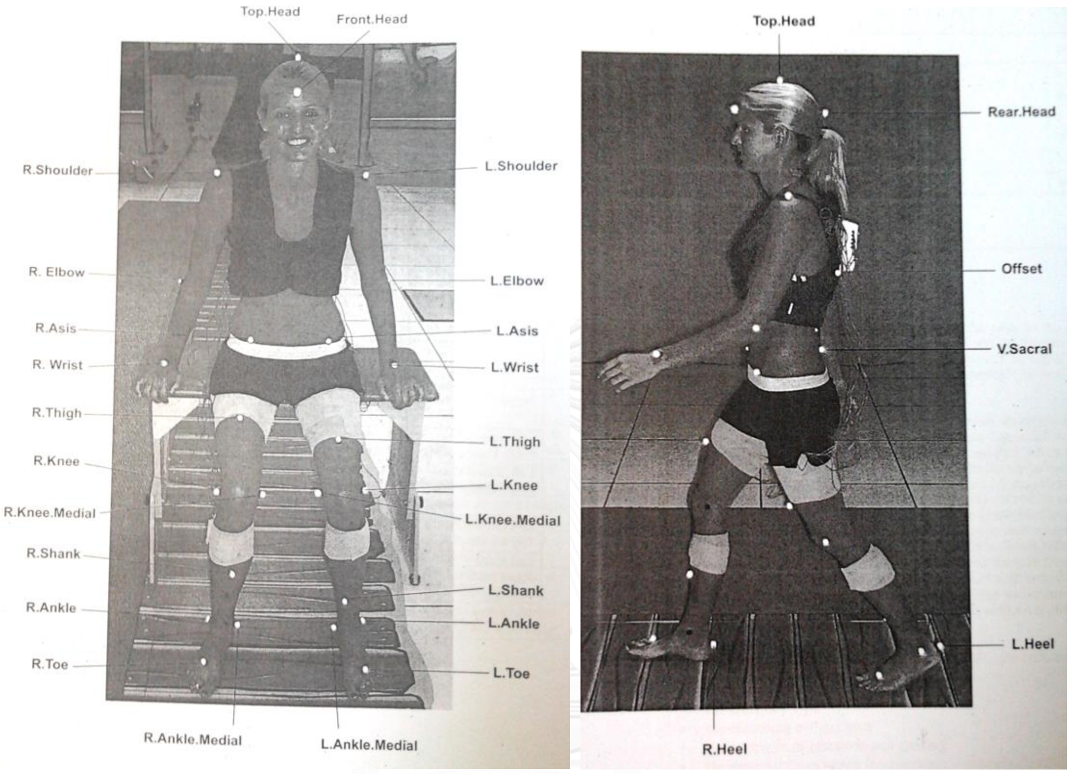

Figure 32 Helen Hayes Mater Set Static and Walking Trials

Table 5 Helen Hayes Marker Set with static trials marker list

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Sacrum	V.Sacral	+	+	+	+	Superior Aspect at the L5-Sacral interface
Left thigh wand Right thigh wand	L.Thigh R.Thigh	+	+	+	+	On lower thigh below the mid-point, for greatest visibility by all cameras
Left lateral knee Right lateral knee	L.Knee R.Knee	+	+	+	+	Along the flexion/extension axis of rotation at lateral femoral condyle
Left shank wand Right shank wand	L.Shank R.Shank	+	+	+	+	On lower shank below the midpoint, for greatest visibility by all cameras
Left lateral ankle Right lateral ankle	L.Ankle R.Ankle	+	+	+	+	Along the flexion/extension axis of rotation at lateral malleolus
Left heel Right heel	L.Heel R.Heel	+	+	+	+	Posterior Calcaneus at same hight from floor as toe marker
Left toe Right toe	L.Toe R.Toe	+	+	+	+	Center of the foot between the 2nd and 3rd metatarsals
Left medial ankle Right medial ankle	L.Ankle.Medial R.Ankle.Medial	+				Along the flexion/extension axis of rotation at medial malleolus
Left medial knee Right medial knee	L.Knee.Medial R.Knee.Medial	+				Along the flexion/extension axis of rotation at medial femoral condyle

3.2.5.2 Equipments and experimental set up

Temporo-spatial gait, 3-dimension kinetic and kinematic data were collected using an eight camera Motion Analysis System (Motion Analysis Corp., Santa Rosa, CA) that have flash rate of 120 Hz. The cameras were synchronized with the BERTEC force platform, which was set to have a sampling frequency of 1,200 Hz. Cortex version 2.5 was software created by the Motion Analysis corporation for use with their Motion Analysis motion capture system that combines 3 major functions into a single software package such as calibration of capture volume, tracking and indentifying marker locations in calibrated 3D space and post processing tools for tracking, editing, and preparing the data for other packages. Then the data were processed with clinical gait evaluation module by OrthoTrak 6.6 Gait Analysis Software. Static and dynamic calibrations were performed before experimental trial.

Subjects were asked for perform static trial to system detect the reflective markers. Then, four medial marker of knees and ankles were removed for prepare walking trial.

Subjects were allowed to walk barefoot, shoe, and shoe with lateral wedge insole that the order of shoe condition was randomized. The speed of walking were set in 3 velocity such as the self-selected speed, standard speed of 1.24 m/s, and a Froude (FR) velocity that base on subject's leg length (length in centimeters from the greater trochanter to the floor while subject was standing) and g = 9.8 m/s^2, can calculate by following equation (Russell, 2011, and Wilken, 2011).

$$FR\ velocity = 0.04 \times \sqrt{g \times leg\ length}$$

Subjects were required to walk along 10 m. walkway. Subjects were given enough time to warm up and became familiar with each velocity. The sequence of walking velocity for each subject start with self selected velocity then the order of the other velocity was randomized prior to the test. Five minute rest between trials was performed. Five successful trials of a velocity and shoe condition were collected for each subject (Cofre, 2011). Successful trials were defined as the trials which subject walk with complete contact with the force plate.

3.2.6 Main Outcome Measures

The data were corrected from right side of lower extremity included;

Spatio-temporal parameter:

- step length	- step width
- stride length	- cadence
- double support time	- support time
- non support time	- velocity

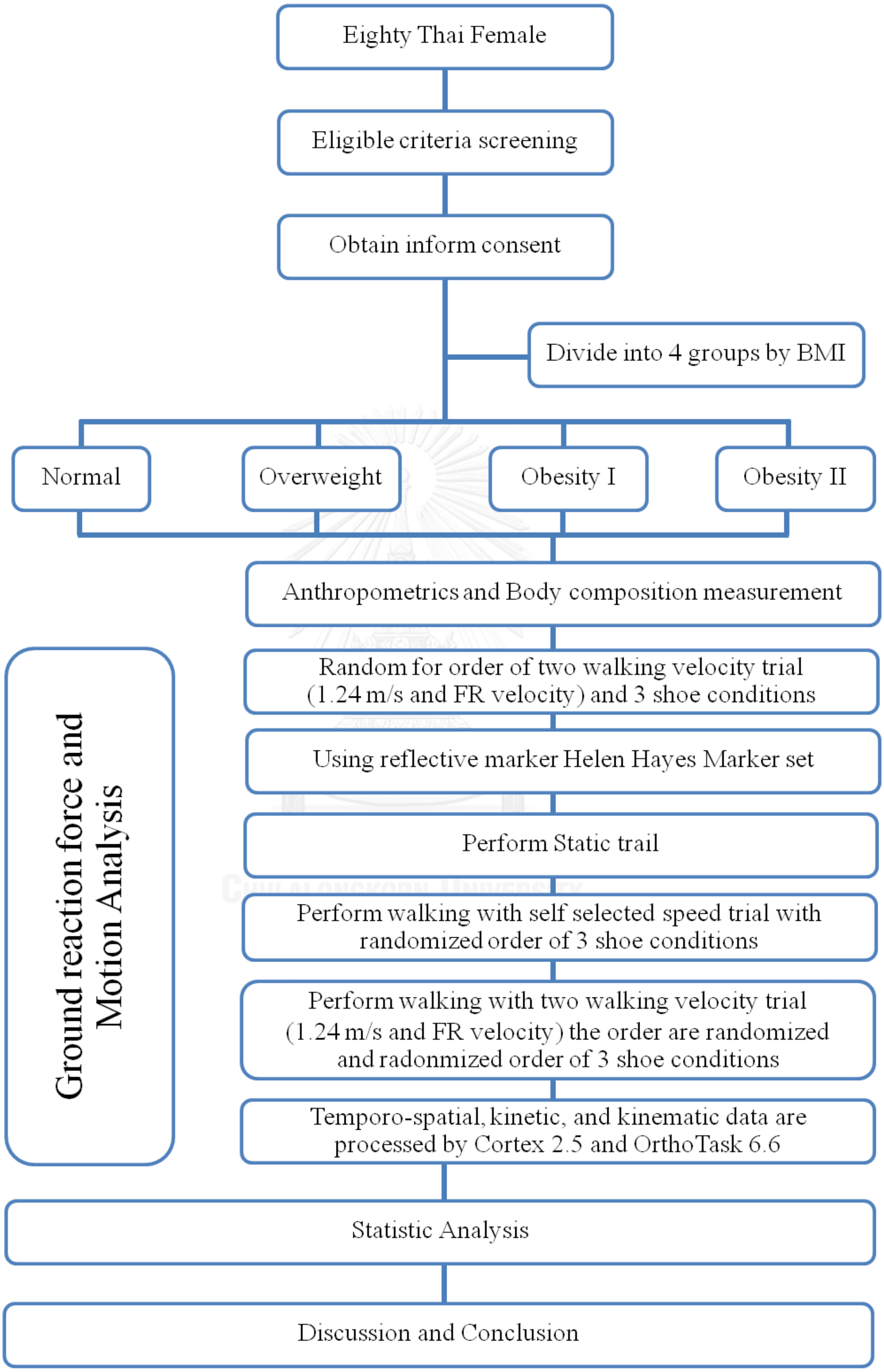
Kinematics: knee joint angle

Kinetics: vertical ground reaction force, peak external knee adduction moment and knee adduction angular impulse.

3.2.7 Statistical Analysis

Cortex version 2.5 and OrthoTrak Gait Analysis version 6.6 software was used to define gait parameters and estimate kinetic and kinematics data. Statistical analysis was performed using SPSS 17.5 for Windows. The level of significance was set at p-value $\leq$ 0.05. Subject characteristic data were expressed as mean and standard deviation (SD) values. Investigation data were expressed as mean and standard error of mean (SEM) values by averaging the five complete trials of each participant. The normal distribution of data was tested using the Kolmogorov-Smirnov test. The differences in the means between the walking conditions of each BMI level were test using the Repeated ANOVA and Bonfferoni post-hoc test. (Adouni and Shirazi-Adl, 2013)

3.2.8 Experimental Design



CHAPTER IV

RESULTS

4.1 Demographic and anthropometric characteristics

The 80 participant's demographic and anthropometric characteristics were presented in table 6. There were no significant differences between groups as regards age, height, and leg length. As expected, there were significant differences between the groups in that the weight (*p = 0.000*), body mass index (*p = 0.000*), waist circumference (*p = 0.000*), hip circumference (*p = 0.000*), waist-hip ratio (*p = 0.000*), percent body fat (*p = 0.000*), fat mass (*p = 0.000*), lean body weight (*p = 0.000*) and visceral fat (*p = 0.000*) were higher in the overweight, obese I and obese II groups than the normal group, respectively. Moreover, the knee circumference (*p = 0.000*) and Q angle (*p = 0.003*) showed an increase among the normal, overweight, obese I and obese II groups, respectively.

4.2 Study 1

The Spatiotemporal variables of each walking velocity were shown in table 21 and 22. When walking with self selected speed we found that there was significantly difference in the step width between normal (9.95 ± 0.70) and obesity II (12.28 ± 0.47) groups (*p* = 0.034). Moreover, the support time (normal = 59.37 ± 0.33; obesity II = 61.35 ± 0.30) and non-support time (normal = 40.63 ± 0.33; obesity II = 38.65 ± 0.30) were significantly difference between normal and obesity II groups (*p* = 0.000). In addition, the support time (overweight = 60.07 ± 0.38;

obesity II = 61.35 ± 0.30) and non-support time (overweight = 39.93 ± 0.38; obesity II = 38.65 ± 0.30) were significantly difference between overweight and obesity II groups (p = 0.030). As the self selected speed, there were significantly difference between normal and obesity II in the support time (normal = 59.43 ± 0.45; obesity II = 61.16 ± 0.51) and non-support time (normal = 40.57 ± 0.45; obesity II = 38.84 ± 0.51) during walk with constant velocity (1.24 m/s) (p = 0.039). When participants walk with Froude (FR) velocity, there were significantly differences between obese II (38.71 ± 0.30) compare with normal (40.35 ± 0.40) and overweight (40.22 ± 0.32) groups in non-support time (p = 0.003 and 0.008, respectively) in figure 33 - 36.

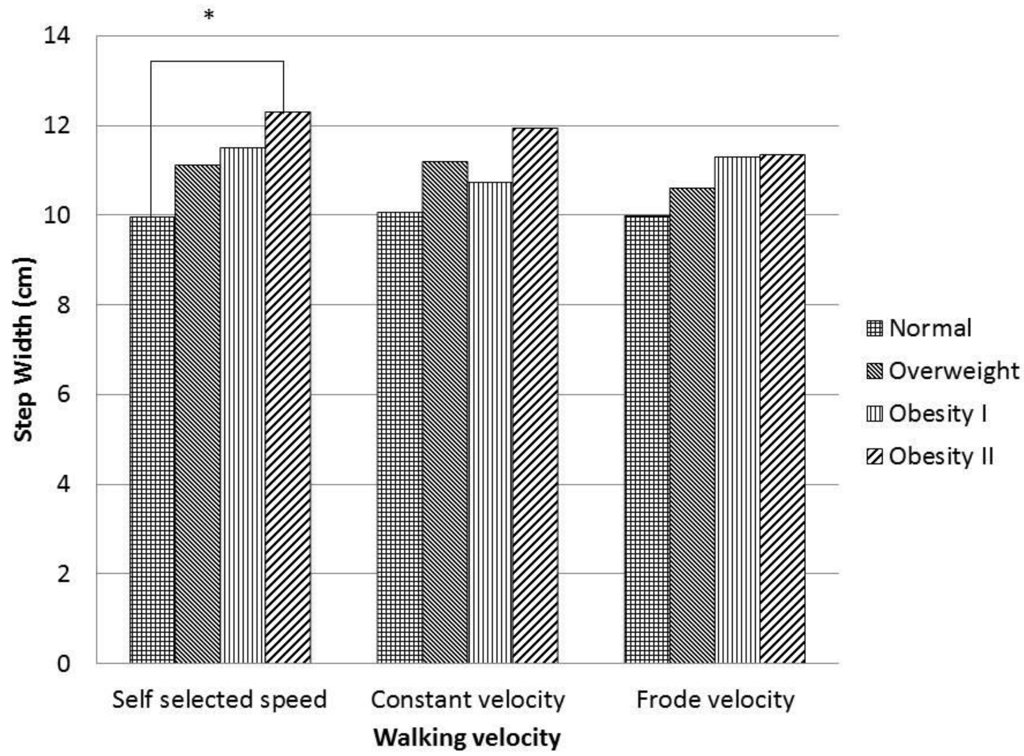
The kinetics and kinematic variables were shown in table 23 and 24. We found that there were significant differences of the peak external knee adduction moment and the knee adduction moment impulse between difference BMI groups. When walk with self selected speed we found that there was significantly difference in the peak external knee adduction moment (Nm/w) (figure 37) between normal (0.334 ± 0.031) and obesity I (0.222 ± 0.022) groups (p = 0.021). In addition, there were significantly differences in the peak external knee adduction moment (Nm) (figure 39) when compare with obesity II (28.089 ± 2.905) to normal (15.645 ± 1.738; p = 0.027), overweight (19.501 ± 2.020; p = 0.006), and obesity I (22.886 ± 2.377; p = 0.002). There was significantly difference in the knee adduction angular impulse between normal (0.112 ± 0.015) and obesity I (0.083 ± 0.009) groups (p = 0.015) (figure 38).

During walking with constant velocity (1.24 m/s), we found that that there was significantly difference in the peak external knee adduction moment (Nm/w) between normal (0.352 ± 0.033) and obesity I (0.214 ± 0.024) groups (p = 0.005). In addition, there were significantly differences in the peak external knee adduction moment (Nm) when compare with obesity II (21.821 ± 2.232) to normal (14.584 ± 1.329; p = 0.019), overweight (16.494 ± 1.076; p = 0.022), and obesity I (18.205 ± 1.541; p = 0.004). There was significantly difference in the knee adduction angular impulse between normal (0.120 ± 0.015) and obesity I (0.074 ± 0.010) groups (p = 0.024).

When participants walk with Froude (FR) velocity, that there was significantly difference in the peak external knee adduction moment (Nm/w) between normal (0.325 ± 0.033) and obesity I (0.225 ± 0.026) groups (p = 0.025). In addition, there were significantly differences in the peak external knee adduction moment (Nm) when compare with obesity II (25.306 ± 2.223) to normal (17.664 ± 2.128; p = 0.013), overweight (20.722 ± 2.344; p = 0.019), and obesity I (20.835 ± 2.301; p = 0.026). There was significantly difference in the knee adduction angular impulse between normal (0.115 ± 0.015) and obesity I (0.076 ± 0.011) groups (p = 0.048).

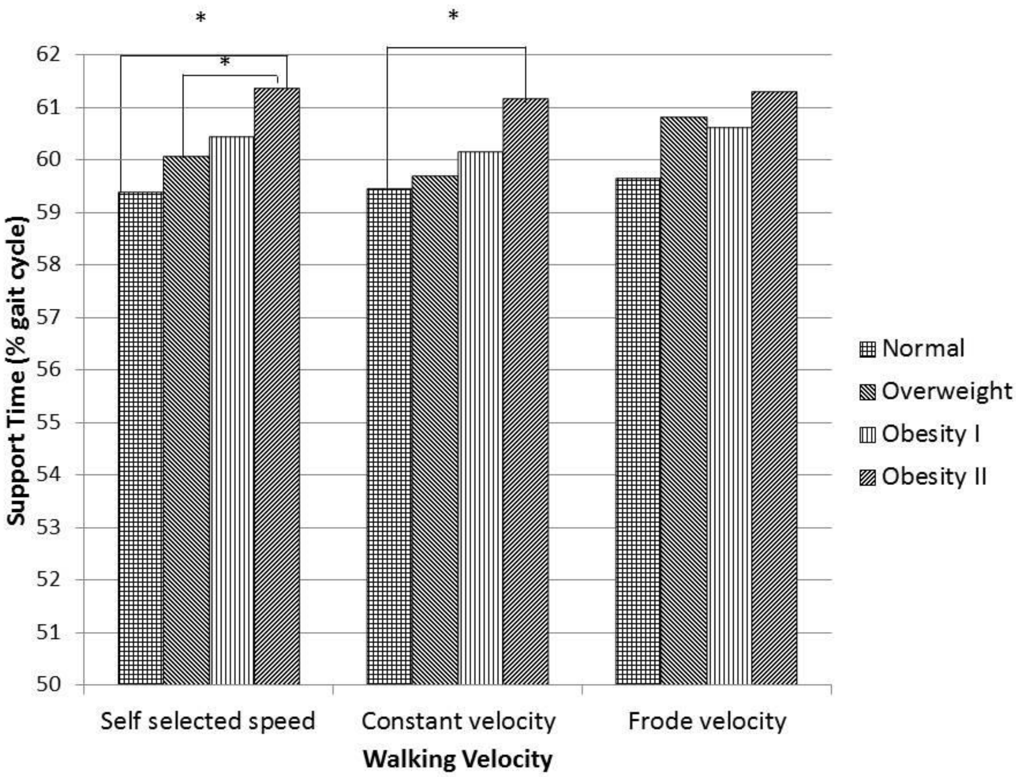
There was no significantly difference in the peak knee flexion angle, the peak vertical ground reaction force when walking with the three experimental speeds.

The table 7 to 20 shown the correlation between demographic parameter especially BMI to the temporo-spatial and biomechanical parameters. We found that BMI correlate with step width (r = 0.298), support time (r = 0.427), non support time (r = - 0.426), double support time (r = 0.271), and peak external knee adduction moment (r = 0.309). Moreover, we found that the peak external knee adduction moment was correlated with weight (r = 0.349), right knee width (r = 0.261), right foot width (r = 0.304), right foot length (r = 0.225), waist circumference (r = 0.242), hip circumference (r = 0.338), right knee joint circumference (r = 0.337), right Q angle (r = 0.342), lean body weight (r = 0.368), visceral fat (r = 0.305), and peak knee flexion angle (r = -0.353).



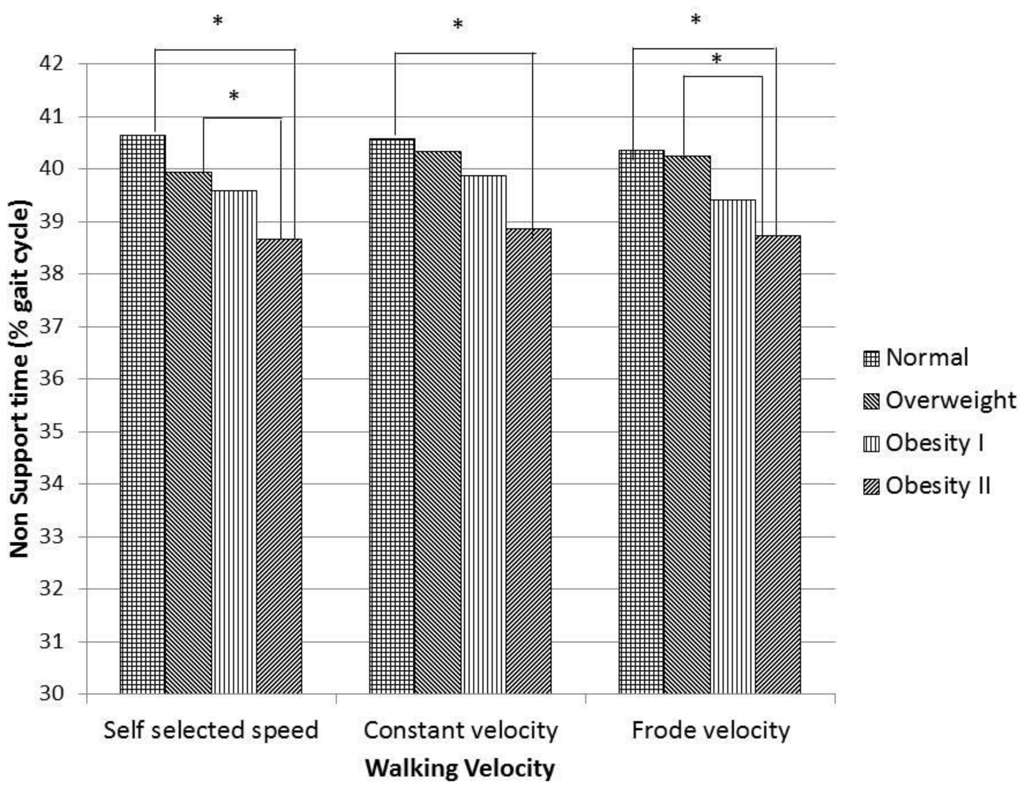
* significantly different between BMI groups (*p < 0.05)*

Figure 33 Step width (cm) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



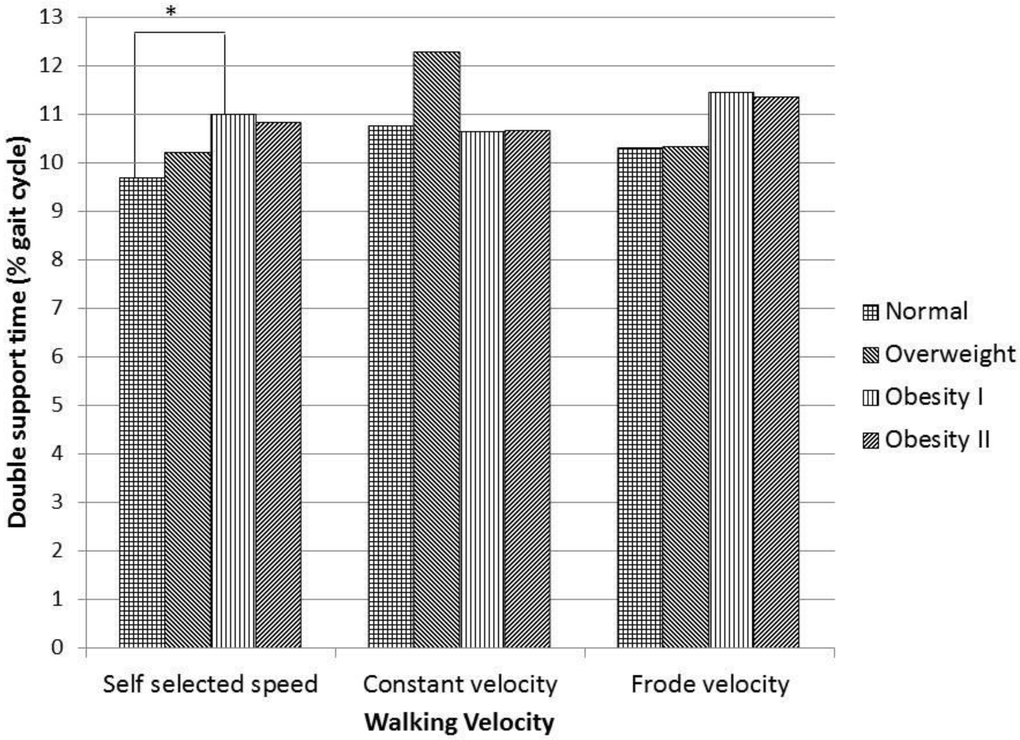
* significantly different between BMI groups (*p < 0.05)*

Figure 34 Support time (% gait cycle) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



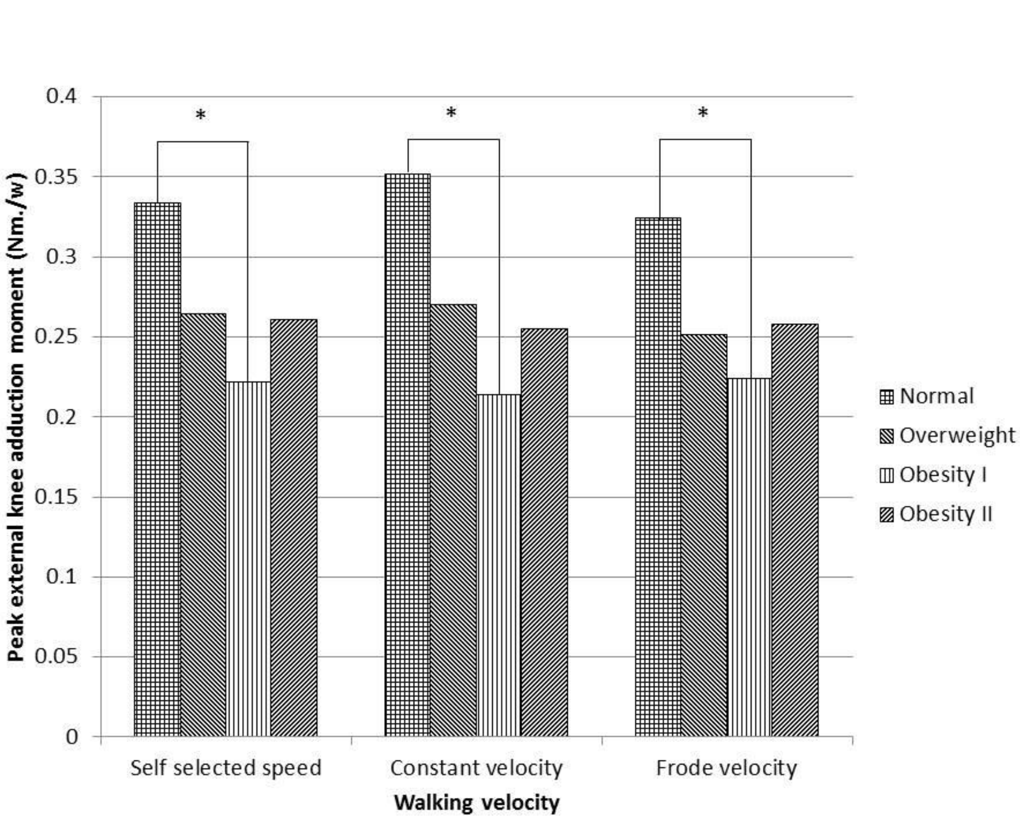
* significantly different between BMI groups (*p < 0.05)*

Figure 35 Non support time (% gait cycle) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



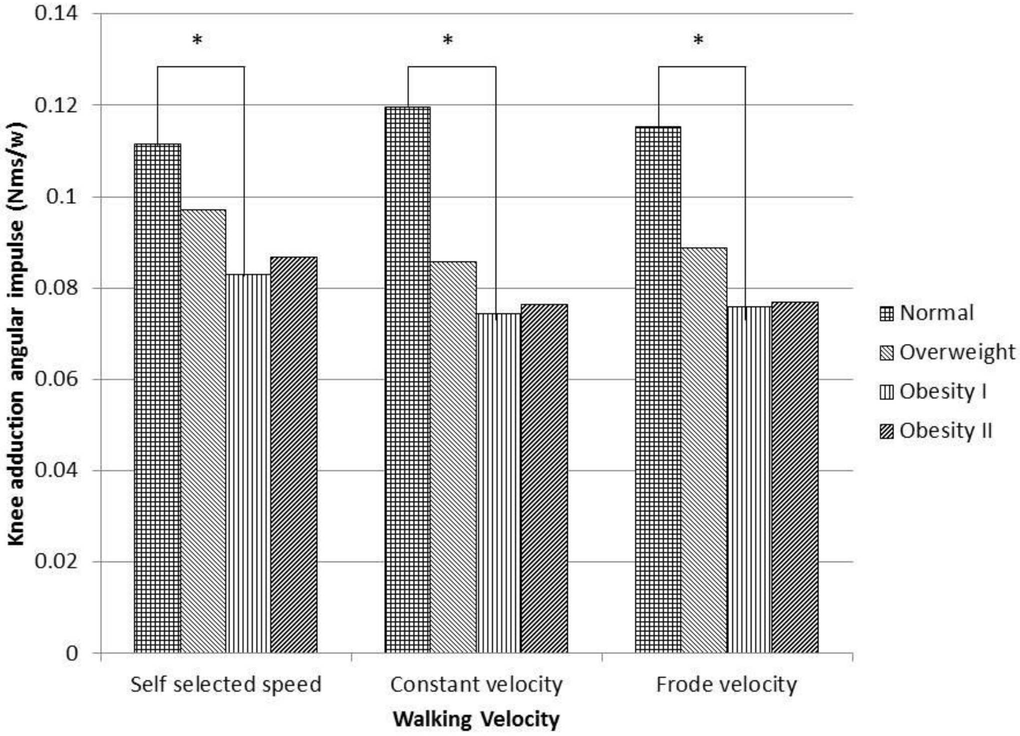
* significantly different between BMI groups (*$p < 0.05$)*

Figure 36 Double support time (% gait cycle) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



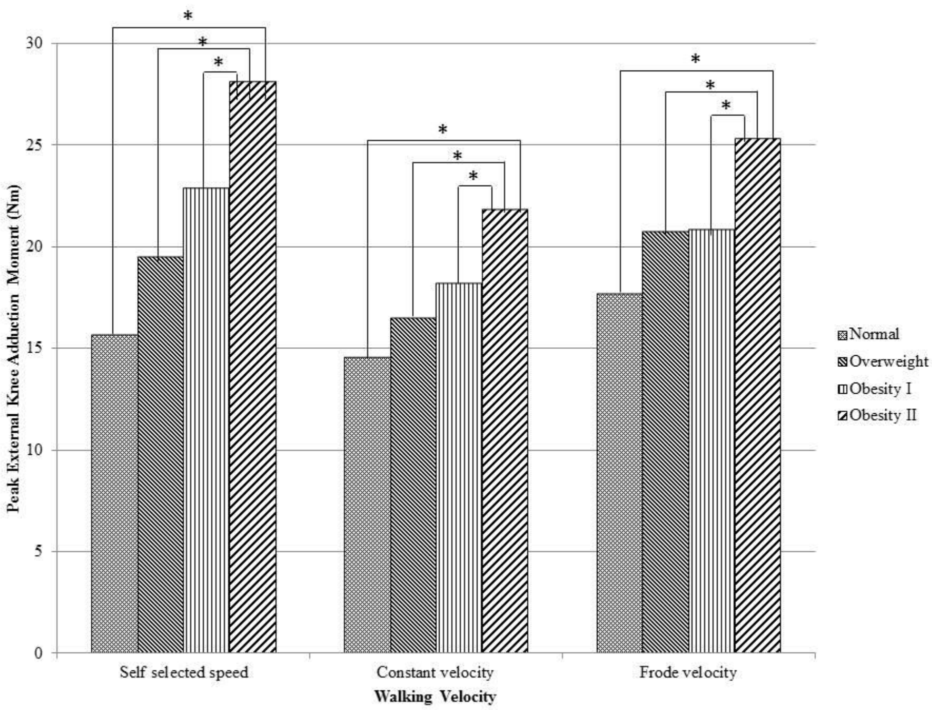
* significantly different between BMI groups (*p < 0.05)*

Figure 37 Peak external knee adduction moment (Nm./w) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



* significantly different between BMI groups ($p < 0.05$)

Figure 38 Knee adduction angular impulse (Nms/w) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.



* significantly different between BMI groups (*p < 0.05)*

Figure 39 Peak external knee adduction moment (Nm) in the four groups during walking at self-selected speed, constant velocity and Froude velocity.

Table 6 Demographic and anthropometric characteristics of eighty female participants.

Parameter	Normal (n = 20)		Overweight (n = 20)		Obese I (n = 20)		Obese II (n = 20)		*P* value
	Mean	**SD**	**Mean**	**SD**	**Mean**	**SD**	**Mean**	**SD**	
Age (Years)	24.5	3.43	26.6	5.71	29.25	6.13	26.85	6.23	0.065
Weight (kg.)	51.1	2.89	59.94	4.1	69.33	6.37	86.99	9.88	**0.000***
Height (cm.)	159.59	4.29	157.93	5.07	157.43	5.75	160.7	6.13	0.199
BMI ($kg.m^{-2}$)	20.12	1.21	23.96	0.66	27.87	1.35	33.68	3.81	**0.000***
Right leg length (cm.)	81.4	3.62	79.3	4.24	78.58	4.09	80.91	2.88	0.064
Waist circumference (cm.)	69.93	4.67	78.08	4.33	85.58	4.74	97.04	8.76	**0.000***
Hip circumference (cm.)	90.05	3.8	98.2	3.88	103.18	4.83	114.73	6.79	**0.000***
Waist Hip ratio	0.78	0.04	0.8	0.06	0.83	0.05	0.85	0.07	**0.000***
Right knee joint circumference (cm.)	33.69	1.39	36.93	1.7	37.75	2.39	41.93	2.88	**0.000***
Right Q angle (degrees)	10.35	1.09	11.3	1.78	11.55	1.96	12.45	1.9	**0.003***
Percent body fat (%)	28.05	3.86	30.95	3.29	34.83	3.91	37.81	4.83	**0.000***
Lean body weight (kg.)	36.71	2.03	41.35	3.01	45.08	3.74	53.87	5.13	**0.000***
Visceral fat (kg.)	2.73	0.99	4.98	0.9	8.5	1.38	14.73	4.99	**0.000***

* significantly different between BMI groups ($p < 0.05$)

Table 7 Correlations between parameters during gait.

Parameters		BMI	Age	Weight	Height
BMI	Pearson Correlation	1	.190	.945**	.008
	Sig. (2-tailed)		.092	.000	.945
Age	Pearson Correlation	.190	1	.062	-.344**
	Sig. (2-tailed)	.092		.583	.002
Weight	Pearson Correlation	.945**	.062	1	.326**
	Sig. (2-tailed)	.000	.583		.003
Height	Pearson Correlation	.008	-.344**	.326**	1
	Sig. (2-tailed)	.634	.002	.125	.000
Right leg length	Pearson Correlation	-.049	-.330**	.177	.728**
	Sig. (2-tailed)	.000	.969	.000	.002
Right knee width	Pearson Correlation	.818**	.018	.876**	.314**
	Sig. (2-tailed)	.000	.373	.000	.001
Right ankle width	Pearson Correlation	.362**	-.139	.487**	.411**
	Sig. (2-tailed)	.000	.983	.000	.001
Right foot width	Pearson Correlation	.409**	-.021	.527**	.395**
	Sig. (2-tailed)	.030	.438	.000	.000
Right foot length	Pearson Correlation	.281*	-.100	.468**	.621**
	Sig. (2-tailed)	.012	.377	.000	.000
Waist circumference	Pearson Correlation	.916**	.180	.899**	.101
	Sig. (2-tailed)	.000	.110	.000	.374
Hip circumference	Pearson Correlation	.908**	.083	.949**	.272*
	Sig. (2-tailed)	.000	.466	.000	.015
Waist Hip ratio	Pearson Correlation	.491**	.232*	.404**	-.171
	Sig. (2-tailed)	.000	.940	.000	.011
Right knee joint circumference	Pearson Correlation	.849**	-.020	.894**	.276*
	Sig. (2-tailed)	.000	.702	.000	.181
Right Q angle	Pearson Correlation	.391**	-.072	.409**	.108
	Sig. (2-tailed)	.000	.525	.000	.340
Percent body fat	Pearson Correlation	.767**	.099	.742**	.063
	Sig. (2-tailed)	.000	.381	.000	.576
Lean body weight	Pearson Correlation	.855**	.047	.941**	.402**
	Sig. (2-tailed)	.000	.678	.000	.000
Visceral fat	Pearson Correlation	.975**	.259*	.902**	-.040
	Sig. (2-tailed)	.000	.020	.000	.725

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 8 Correlations between parameters during gait (continued).

Parameters		BMI	Age	Weight	Height
Peak knee flexion angle	Pearson Correlation	.077	.300**	-.022	-.255*
	Sig. (2-tailed)	.496	.007	.849	.022
Peak VGRF	Pearson Correlation	-.106	.009	-.181	-.261*
	Sig. (2-tailed)	.350	.936	.109	.020
PEKADM (normalized)	Pearson Correlation	-.191	-.107	-.178	.008
	Sig. (2-tailed)	.090	.344	.114	.946
KAAI (normalized)	Pearson Correlation	-.315**	-.038	-.295**	.012
	Sig. (2-tailed)	.004	.738	.008	.913
PEKADM (non normalized)	Pearson Correlation	.309**	-.058	.349**	.178
	Sig. (2-tailed)	.005	.610	.001	.114
KAAI (non normalized)	Pearson Correlation	.042	.000	.087	.153
	Sig. (2-tailed)	.710	.999	.443	.176
Step Width	Pearson Correlation	.298**	-.017	.291**	.034
	Sig. (2-tailed)	.007	.878	.009	.764
Velocity	Pearson Correlation	-.163	-.011	-.135	.065
	Sig. (2-tailed)	.149	.926	.232	.568
Stride length	Pearson Correlation	-.138	.083	-.061	.234*
	Sig. (2-tailed)	.223	.462	.592	.037
Cadence	Pearson Correlation	-.090	-.044	-.110	-.091
	Sig. (2-tailed)	.427	.701	.333	.425
Support Time	Pearson Correlation	.427**	.176	.394**	-.053
	Sig. (2-tailed)	.000	.118	.000	.640
Non Support	Pearson Correlation	-.426**	-.176	-.394**	.053
	Sig. (2-tailed)	.000	.117	.000	.640
Step length	Pearson Correlation	.018	-.028	.077	.195
	Sig. (2-tailed)	.874	.809	.498	.084
Double support time	Pearson Correlation	.271*	.163	.245*	-.052
	Sig. (2-tailed)	.015	.148	.029	.648

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 9 Correlations between parameters during gait (continued).

Parameters		Right leg length	Right knee width	Right ankle width	Right foot width	Right foot length
BMI	Pearson Correlation	-.049	.818**	.362**	.409**	.281*
	Sig. (2-tailed)	.663	.000	.001	.000	.012
Age	Pearson Correlation	-.330**	.018	-.139	-.021	-.100
	Sig. (2-tailed)	.003	.872	.220	.854	.377
Weight	Pearson Correlation	.177	.876**	.487**	.527**	.468**
	Sig. (2-tailed)	.116	.000	.000	.000	.000
Height	Pearson Correlation	.728**	.314**	.411**	.395**	.621**
	Sig. (2-tailed)	.000	.097	.315	.051	.000
Right leg length	Pearson Correlation	1	.188	.114	.223*	.404**
	Sig. (2-tailed)	.056	.000	.000	.000	.000
Right knee width	Pearson Correlation	.188	1	.424**	.387**	.400**
	Sig. (2-tailed)	.268	.000	.000	.000	.000
Right ankle width	Pearson Correlation	.114	.424**	1	.561**	.478**
	Sig. (2-tailed)	.041	.000	.000	.000	.000
Right foot width	Pearson Correlation	.223*	.387**	.561**	1	.525**
	Sig. (2-tailed)	.005	.001	.000	.000	.000
Right foot length	Pearson Correlation	.404**	.400**	.478**	.525**	1
	Sig. (2-tailed)	.000	.000	.000	.000	
Waist circumference	Pearson Correlation	-.022	.771**	.399**	.419**	.375**
	Sig. (2-tailed)	.845	.000	.000	.000	.001
Hip circumference	Pearson Correlation	.107	.843**	.443**	.504**	.405**
	Sig. (2-tailed)	.347	.000	.000	.000	.000
Waist Hip ratio	Pearson Correlation	-.200	.300**	.166	.108	.157
	Sig. (2-tailed)	.275	.000	.000	.000	.000
Right knee joint circumference	Pearson Correlation	.116	.859**	.449**	.457**	.404**
	Sig. (2-tailed)	.907	.000	.008	.071	.005
Right Q angle	Pearson Correlation	-.058	.323**	.297**	.223*	.290**
	Sig. (2-tailed)	.608	.003	.007	.047	.009
Percent body fat	Pearson Correlation	.067	.677**	.230*	.227*	.252*
	Sig. (2-tailed)	.554	.000	.041	.043	.024
Lean body weight	Pearson Correlation	.193	.803**	.522**	.579**	.479**
	Sig. (2-tailed)	.087	.000	.000	.000	.000
Visceral fat	Pearson Correlation	-.045	.795**	.286*	.336**	.244*
	Sig. (2-tailed)	.693	.000	.010	.002	.029

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 10 Correlations between parameters during gait (continued).

Parameters		Right leg length	Right knee width	Right ankle width	Right foot width	Right foot length
Peak knee flexion angle	Pearson Correlation	-.283*	-.069	-.191	-.108	-.278*
	Sig. (2-tailed)	.011	.544	.090	.341	.013
Peak VGRF	Pearson Correlation	-.201	-.121	-.048	-.207	-.188
	Sig. (2-tailed)	.073	.285	.673	.065	.095
PEKADM (normalized)	Pearson Correlation	.023	-.191	-.012	-.005	-.002
	Sig. (2-tailed)	.837	.090	.914	.968	.989
KAAI (normalized)	Pearson Correlation	.027	-.260*	-.131	-.035	-.033
	Sig. (2-tailed)	.810	.020	.247	.756	.773
PEKADM (non normalized)	Pearson Correlation	.125	.261*	.252*	.304**	.225*
	Sig. (2-tailed)	.269	.019	.024	.006	.045
KAAI (non normalized)	Pearson Correlation	.107	.076	.067	.207	.150
	Sig. (2-tailed)	.344	.501	.556	.066	.184
Step Width	Pearson Correlation	-.061	.205	.174	.117	.039
	Sig. (2-tailed)	.590	.068	.123	.299	.734
Velocity	Pearson Correlation	-.014	-.147	.057	.056	.110
	Sig. (2-tailed)	.901	.193	.616	.620	.331
Stride length	Pearson Correlation	.207	.000	-.009	.065	.277*
	Sig. (2-tailed)	.065	.997	.939	.564	.013
Cadence	Pearson Correlation	-.138	-.142	.060	.018	-.045
	Sig. (2-tailed)	.222	.207	.600	.874	.692
Support Time	Pearson Correlation	-.120	.396**	.218	.137	.111
	Sig. (2-tailed)	.287	.000	.052	.225	.326
Non Support	Pearson Correlation	.120	-.396**	-.218	-.137	-.111
	Sig. (2-tailed)	.287	.000	.052	.225	.326
Step length	Pearson Correlation	.186	.092	.018	.165	.230*
	Sig. (2-tailed)	.099	.416	.873	.144	.040
Double support time	Pearson Correlation	-.169	.212	.095	.066	.072
	Sig. (2-tailed)	.133	.059	.400	.559	.526

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 11 Correlations between parameters during gait (continued).

Parameters		Waist circumference	Hip circumference	Waist Hip ratio	Right knee joint circumference
BMI	Pearson Correlation	.916**	.908**	.491**	.849**
	Sig. (2-tailed)	.000	.000	.000	.000
Age	Pearson Correlation	.180	.083	.232*	-.020
	Sig. (2-tailed)	.110	.466	.038	.860
Weight	Pearson Correlation	.899**	.949**	.404**	.894**
	Sig. (2-tailed)	.000	.000	.000	.000
Height	Pearson Correlation	.101	.272*	-.171	.276*
	Sig. (2-tailed)	.807	.386	.078	.320
Right leg length	Pearson Correlation	-.022	.107	-.200	.116
	Sig. (2-tailed)	.000	.000	.009	.000
Right knee width	Pearson Correlation	.771**	.843**	.300**	.859**
	Sig. (2-tailed)	.000	.000	.056	.000
Right ankle width	Pearson Correlation	.399**	.443**	.166	.449**
	Sig. (2-tailed)	.000	.000	.107	.000
Right foot width	Pearson Correlation	.419**	.504**	.108	.457**
	Sig. (2-tailed)	.006	.003	.294	.001
Right foot length	Pearson Correlation	.375**	.405**	.157	.404**
	Sig. (2-tailed)	.001	.000	.163	.000
Waist circumference	Pearson Correlation	1	.855**	.721**	.812**
	Sig. (2-tailed)		.000	.000	.000
Hip circumference	Pearson Correlation	.855**	1	.262*	.883**
	Sig. (2-tailed)	.000		.019	.000
Waist Hip ratio	Pearson Correlation	.721**	.262*	1	.328**
	Sig. (2-tailed)	.000	.000	.003	.000
Right knee joint circumference	Pearson Correlation	.812**	.883**	.328**	1
	Sig. (2-tailed)	.000	.000	.028	.000
Right Q angle	Pearson Correlation	.394**	.375**	.231*	.411**
	Sig. (2-tailed)	.000	.001	.039	.000
Percent body fat	Pearson Correlation	.732**	.698**	.430**	.693**
	Sig. (2-tailed)	.000	.000	.000	.000
Lean body weight	Pearson Correlation	.816**	.906**	.312**	.834**
	Sig. (2-tailed)	.000	.000	.005	.000
Visceral fat	Pearson Correlation	.890**	.858**	.502**	.795**
	Sig. (2-tailed)	.000	.000	.000	.000

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 12 Correlations between parameters during gait (continued).

Parameters		Waist circumference	Hip circumference	Waist Hip ratio	Right knee joint circumference
Peak knee flexion angle	Pearson Correlation	.081	-.018	.188	.019
	Sig. (2-tailed)	.477	.871	.094	.868
Peak VGRF	Pearson Correlation	-.123	-.240*	.051	-.167
	Sig. (2-tailed)	.278	.032	.654	.138
PEKADM (normalized)	Pearson Correlation	-.220*	-.156	-.211	-.132
	Sig. (2-tailed)	.050	.167	.061	.241
KAAI (normalized)	Pearson Correlation	-.328**	-.245*	-.295**	-.244*
	Sig. (2-tailed)	.003	.029	.008	.029
PEKADM (non normalized)	Pearson Correlation	.242*	.338**	-.010	.337**
	Sig. (2-tailed)	.031	.002	.929	.002
KAAI (non normalized)	Pearson Correlation	-.002	.118	-.172	.098
	Sig. (2-tailed)	.988	.298	.126	.385
Step Width	Pearson Correlation	.308**	.287**	.187	.302**
	Sig. (2-tailed)	.005	.010	.096	.006
Velocity	Pearson Correlation	-.137	-.136	-.061	-.106
	Sig. (2-tailed)	.224	.228	.592	.350
Stride length	Pearson Correlation	-.137	-.075	-.168	-.080
	Sig. (2-tailed)	.226	.509	.137	.479
Cadence	Pearson Correlation	-.061	-.110	.051	-.063
	Sig. (2-tailed)	.593	.332	.654	.580
Support Time	Pearson Correlation	.491**	.402**	.367**	.349**
	Sig. (2-tailed)	.000	.000	.001	.002
Non Support	Pearson Correlation	-.491**	-.402**	-.367**	-.349**
	Sig. (2-tailed)	.000	.000	.001	.002
Step length	Pearson Correlation	-.016	.082	-.142	.038
	Sig. (2-tailed)	.891	.470	.209	.736
Double support time	Pearson Correlation	.269*	.267*	.136	.151
	Sig. (2-tailed)	.016	.017	.231	.182

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 13 Correlations between parameters during gait (continued).

Parameters		Right Q angle	Percent body fat	Lean body weight	Visceral fat
BMI	Pearson Correlation	.391**	.767**	.855**	.975**
	Sig. (2-tailed)	.000	.000	.000	.000
Age	Pearson Correlation	-.072	.099	.047	.259*
	Sig. (2-tailed)	.525	.381	.678	.020
Weight	Pearson Correlation	.409**	.742**	.941**	.902**
	Sig. (2-tailed)	.000	.000	.000	.000
Height	Pearson Correlation	.108	.063	.402**	-.040
	Sig. (2-tailed)	.555	.528	.100	.672
Right leg length	Pearson Correlation	-.058	.067	.193	-.045
	Sig. (2-tailed)	.003	.000	.000	.000
Right knee width	Pearson Correlation	.323**	.677**	.803**	.795**
	Sig. (2-tailed)	.036	.029	.000	.001
Right ankle width	Pearson Correlation	.297**	.230*	.522**	.286*
	Sig. (2-tailed)	.169	.018	.000	.001
Right foot width	Pearson Correlation	.223*	.227*	.579**	.336**
	Sig. (2-tailed)	.010	.032	.000	.076
Right foot length	Pearson Correlation	.290**	.252*	.479**	.244*
	Sig. (2-tailed)	.009	.024	.000	.029
Waist circumference	Pearson Correlation	.394**	.732**	.816**	.890**
	Sig. (2-tailed)	.000	.000	.000	.000
Hip circumference	Pearson Correlation	.375**	.698**	.906**	.858**
	Sig. (2-tailed)	.001	.000	.000	.000
Waist Hip ratio	Pearson Correlation	.231*	.430**	.312**	.502**
	Sig. (2-tailed)	.000	.000	.000	.000
Right knee joint circumference	Pearson Correlation	.411**	.693**	.834**	.795**
	Sig. (2-tailed)	.000	.002	.000	.001
Right Q angle	Pearson Correlation	1	.346**	.365**	.330**
	Sig. (2-tailed)		.002	.001	.003
Percent body fat	Pearson Correlation	.346**	1	.486**	.753**
	Sig. (2-tailed)	.002		.000	.000
Lean body weight	Pearson Correlation	.365**	.486**	1	.791**
	Sig. (2-tailed)	.001	.000		.000
Visceral fat	Pearson Correlation	.330**	.753**	.791**	1
	Sig. (2-tailed)	.003	.000	.000	

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 14 Correlations between parameters during gait (continued).

Parameters		Right Q angle	Percent body fat	Lean body weight	Visceral fat
Peak knee flexion angle	Pearson Correlation	.059	.041	-.032	.088
	Sig. (2-tailed)	.601	.719	.776	.437
Peak VGRF	Pearson Correlation	-.011	-.134	-.184	-.074
	Sig. (2-tailed)	.926	.238	.102	.516
PEKADM (normalized)	Pearson Correlation	.133	-.247*	-.130	-.174
	Sig. (2-tailed)	.240	.027	.249	.122
KAAI (normalized)	Pearson Correlation	-.007	-.308**	-.248*	-.293**
	Sig. (2-tailed)	.953	.005	.027	.008
PEKADM (non normalized)	Pearson Correlation	.342**	.146	.368**	.305**
	Sig. (2-tailed)	.002	.197	.001	.006
KAAI (non normalized)	Pearson Correlation	.167	-.026	.117	.044
	Sig. (2-tailed)	.138	.818	.301	.699
Step Width	Pearson Correlation	.139	.147	.305**	.240*
	Sig. (2-tailed)	.218	.192	.006	.032
Velocity	Pearson Correlation	.141	-.032	-.151	-.186
	Sig. (2-tailed)	.212	.781	.183	.099
Stride length	Pearson Correlation	-.075	.023	-.095	-.123
	Sig. (2-tailed)	.510	.841	.400	.275
Cadence	Pearson Correlation	.181	-.002	-.127	-.123
	Sig. (2-tailed)	.109	.986	.263	.276
Support Time	Pearson Correlation	.119	.378**	.341**	.416**
	Sig. (2-tailed)	.295	.001	.002	.000
Non Support	Pearson Correlation	-.119	-.378**	-.341**	-.416**
	Sig. (2-tailed)	.295	.001	.002	.000
Step length	Pearson Correlation	-.024	.130	.039	.008
	Sig. (2-tailed)	.835	.252	.730	.943
Double support time	Pearson Correlation	.087	.208	.232*	.237*
	Sig. (2-tailed)	.444	.064	.039	.034

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 15 Correlations between parameters during gait (continued).

Parameters		Peak knee flexion angle	Peak VGRF	PEKADM (normalized)	KAAI (normalized)	PEKADM (non normalized)	KAAI (non normalized)
BMI	Pearson Correlation	.077	-.106	-.191	-.315**	.309**	.042
	Sig. (2-tailed)	.496	.350	.090	.004	.005	.710
Age	Pearson Correlation	.300**	.009	-.107	-.038	-.058	.000
	Sig. (2-tailed)	.007	.936	.344	.738	.610	.999
Weight	Pearson Correlation	-.022	-.181	-.178	-.295**	.349**	.087
	Sig. (2-tailed)	.849	.109	.114	.008	.001	.443
Height	Pearson Correlation	-.255*	-.261*	.008	.012	.178	.153
	Sig. (2-tailed)	.015	.076	.959	.919	.342	.429
Right leg length	Pearson Correlation	-.283*	-.201	.023	.027	.125	.107
	Sig. (2-tailed)	.532	.284	.118	.031	.011	.361
Right knee width	Pearson Correlation	-.069	-.121	-.191	-.260*	.261*	.076
	Sig. (2-tailed)	.071	.327	.593	.107	.025	.683
Right ankle width	Pearson Correlation	-.191	-.048	-.012	-.131	.252*	.067
	Sig. (2-tailed)	.244	.037	.751	.569	.013	.115
Right foot width	Pearson Correlation	-.108	-.207	-.005	-.035	.304**	.207
	Sig. (2-tailed)	.006	.185	.979	.928	.073	.105
Right foot length	Pearson Correlation	-.278*	-.188	-.002	-.033	.225*	.150
	Sig. (2-tailed)	.013	.095	.989	.773	.045	.184
Waist circumference	Pearson Correlation	.081	-.123	-.220*	-.328**	.242*	-.002
	Sig. (2-tailed)	.477	.278	.050	.003	.031	.988
Hip circumference	Pearson Correlation	-.018	-.240*	-.156	-.245*	.338**	.118
	Sig. (2-tailed)	.871	.032	.167	.029	.002	.298
Waist Hip ratio	Pearson Correlation	.188	.051	-.211	-.295**	-.010	-.172
	Sig. (2-tailed)	.894	.135	.220	.026	.003	.407
Right knee joint circumference	Pearson Correlation	.019	-.167	-.132	-.244*	.337**	.098
	Sig. (2-tailed)	.964	.807	.332	.863	.002	.123
Right Q angle	Pearson Correlation	.059	-.011	.133	-.007	.342**	.167
	Sig. (2-tailed)	.601	.926	.240	.953	.002	.138
Percent body fat	Pearson Correlation	.041	-.134	-.247*	-.308**	.146	-.026
	Sig. (2-tailed)	.719	.238	.027	.005	.197	.818
Lean body weight	Pearson Correlation	-.032	-.184	-.130	-.248*	.368**	.117
	Sig. (2-tailed)	.776	.102	.249	.027	.001	.301
Visceral fat	Pearson Correlation	.088	-.074	-.174	-.293**	.305**	.044
	Sig. (2-tailed)	.437	.516	.122	.008	.006	.699

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 16 Correlations between parameters during gait (continued).

Parameters		Peak knee flexion angle	Peak VGRF	PEKADM (normalized)	KAAI (normalized)	PEKADM (non normalized)	KAAI (non normalized)
Peak knee flexion angle	Pearson Correlation	1	.014	-.363**	-.311**	-.353**	-.343**
	Sig. (2-tailed)		.904	.001	.005	.001	.002
Peak VGRF	Pearson Correlation	.014	1	.170	.076	.072	.006
	Sig. (2-tailed)	.904		.132	.501	.528	.956
PEKADM (normalized)	Pearson Correlation	-.363**	.170	1	.918**	.840**	.876**
	Sig. (2-tailed)	.001	.132		.000	.000	.000
KAAI (normalized)	Pearson Correlation	-.311**	.076	.918**	1	.673**	.911**
	Sig. (2-tailed)	.005	.501	.000		.000	.000
PEKADM (non normalized)	Pearson Correlation	-.353**	.072	.840**	.673**	1	.857**
	Sig. (2-tailed)	.001	.528	.000	.000		.000
KAAI (non normalized)	Pearson Correlation	-.343**	.006	.876**	.911**	.857**	1
	Sig. (2-tailed)	.002	.956	.000	.000	.000	
Step Width	Pearson Correlation	.013	.106	-.051	-.108	.080	-.020
	Sig. (2-tailed)	.907	.348	.654	.341	.482	.863
Velocity	Pearson Correlation	.250*	.155	.027	-.058	-.008	-.092
	Sig. (2-tailed)	.025	.169	.814	.612	.941	.416
Stride length	Pearson Correlation	.046	.187	-.002	.038	-.026	.021
	Sig. (2-tailed)	.686	.096	.983	.736	.822	.850
Cadence	Pearson Correlation	.206	.099	.064	-.047	.037	-.069
	Sig. (2-tailed)	.067	.381	.573	.679	.747	.540
Support Time	Pearson Correlation	-.044	.145	-.129	-.128	.051	.019
	Sig. (2-tailed)	.701	.198	.255	.257	.654	.869
Non Support	Pearson Correlation	.043	-.145	.129	.128	-.051	-.019
	Sig. (2-tailed)	.702	.198	.255	.258	.656	.869
Step length	Pearson Correlation	.080	.070	-.081	-.097	-.034	-.085
	Sig. (2-tailed)	.480	.535	.475	.394	.768	.453
Double support time	Pearson Correlation	-.139	-.023	-.163	-.091	-.068	-.005
	Sig. (2-tailed)	.219	.838	.148	.420	.551	.967

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 17 Correlations between parameters during gait (continued).

Parameters		Step Width	Velocity	Stride length	Cadence
BMI	Pearson Correlation	.298**	-.163	-.138	-.090
	Sig. (2-tailed)	.007	.149	.223	.427
Age	Pearson Correlation	-.017	-.011	.083	-.044
	Sig. (2-tailed)	.878	.926	.462	.701
Weight	Pearson Correlation	.291**	-.135	-.061	-.110
	Sig. (2-tailed)	.009	.232	.592	.333
Height	Pearson Correlation	.034	.065	.234*	-.091
	Sig. (2-tailed)	.505	.990	.050	.241
Right leg length	Pearson Correlation	-.061	-.014	.207	-.138
	Sig. (2-tailed)	.042	.158	.855	.189
Right knee width	Pearson Correlation	.205	-.147	.000	-.142
	Sig. (2-tailed)	.133	.838	.807	.839
Right ankle width	Pearson Correlation	.174	.057	-.009	.060
	Sig. (2-tailed)	.422	.732	.471	.975
Right foot width	Pearson Correlation	.117	.056	.065	.018
	Sig. (2-tailed)	.709	.219	.001	.688
Right foot length	Pearson Correlation	.039	.110	.277*	-.045
	Sig. (2-tailed)	.734	.331	.013	.692
Waist circumference	Pearson Correlation	.308**	-.137	-.137	-.061
	Sig. (2-tailed)	.005	.224	.226	.593
Hip circumference	Pearson Correlation	.287**	-.136	-.075	-.110
	Sig. (2-tailed)	.010	.228	.509	.332
Waist Hip ratio	Pearson Correlation	.187	-.061	-.168	.051
	Sig. (2-tailed)	.008	.351	.580	.531
Right knee joint circumference	Pearson Correlation	.302**	-.106	-.080	-.063
	Sig. (2-tailed)	.152	.602	.417	.284
Right Q angle	Pearson Correlation	.139	.141	-.075	.181
	Sig. (2-tailed)	.218	.212	.510	.109
Percent body fat	Pearson Correlation	.147	-.032	.023	-.002
	Sig. (2-tailed)	.192	.781	.841	.986
Lean body weight	Pearson Correlation	.305**	-.151	-.095	-.127
	Sig. (2-tailed)	.006	.183	.400	.263
Visceral fat	Pearson Correlation	.240*	-.186	-.123	-.123
	Sig. (2-tailed)	.032	.099	.275	.276

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 18 Correlations between parameters during gait (continued).

Parameters		**Step Width**	**Velocity**	**Stride length**	**Cadence**
Peak knee flexion angle	Pearson Correlation	.013	.250*	.046	.206
	Sig. (2-tailed)	.907	.025	.686	.067
Peak VGRF	Pearson Correlation	.106	.155	.187	.099
	Sig. (2-tailed)	.348	.169	.096	.381
PEKADM (normalized)	Pearson Correlation	-.051	.027	-.002	.064
	Sig. (2-tailed)	.654	.814	.983	.573
KAAI (normalized)	Pearson Correlation	-.108	-.058	.038	-.047
	Sig. (2-tailed)	.341	.612	.736	.679
PEKADM (non normalized)	Pearson Correlation	.080	-.008	-.026	.037
	Sig. (2-tailed)	.482	.941	.822	.747
KAAI (non normalized)	Pearson Correlation	-.020	-.092	.021	-.069
	Sig. (2-tailed)	.863	.416	.850	.540
Step Width	Pearson Correlation	1	.005	-.152	.079
	Sig. (2-tailed)		.963	.177	.488
Velocity	Pearson Correlation	.005	1	.485**	.821**
	Sig. (2-tailed)	.963		.000	.000
Stride length	Pearson Correlation	-.152	.485**	1	-.040
	Sig. (2-tailed)	.177	.000		.722
Cadence	Pearson Correlation	.079	.821**	-.040	1
	Sig. (2-tailed)	.488	.000	.722	
Support Time	Pearson Correlation	.111	-.196	-.064	-.080
	Sig. (2-tailed)	.329	.081	.570	.483
Non Support	Pearson Correlation	-.111	.196	.064	.080
	Sig. (2-tailed)	.329	.082	.570	.483
Step length	Pearson Correlation	-.142	.377**	.721**	-.066
	Sig. (2-tailed)	.208	.001	.000	.562
Double support time	Pearson Correlation	.084	-.243*	-.013	-.211
	Sig. (2-tailed)	.456	.030	.907	.061

**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).

Table 19 Correlations between parameters during gait (continued).

Parameters		Support Time	Non Support	Step length	Double support time
BMI	Pearson Correlation	.427**	-.426**	.018	.271*
	Sig. (2-tailed)	.000	.000	.874	.015
Age	Pearson Correlation	.176	-.176	-.028	.163
	Sig. (2-tailed)	.118	.117	.809	.148
Weight	Pearson Correlation	.394**	-.394**	.077	.245*
	Sig. (2-tailed)	.000	.000	.498	.029
Height	Pearson Correlation	-.053	.053	.195	-.052
	Sig. (2-tailed)	.273	.273	.081	.113
Right leg length	Pearson Correlation	-.120	.120	.186	-.169
	Sig. (2-tailed)	.000	.000	.572	.054
Right knee width	Pearson Correlation	.396**	-.396**	.092	.212
	Sig. (2-tailed)	.095	.095	.990	.342
Right ankle width	Pearson Correlation	.218	-.218	.018	.095
	Sig. (2-tailed)	.101	.100	.127	.444
Right foot width	Pearson Correlation	.137	-.137	.165	.066
	Sig. (2-tailed)	.247	.247	.043	.376
Right foot length	Pearson Correlation	.111	-.111	.230*	.072
	Sig. (2-tailed)	.326	.326	.040	.526
Waist circumference	Pearson Correlation	.491**	-.491**	-.016	.269*
	Sig. (2-tailed)	.000	.000	.891	.016
Hip circumference	Pearson Correlation	.402**	-.402**	.082	.267*
	Sig. (2-tailed)	.000	.000	.470	.017
Waist Hip ratio	Pearson Correlation	.367**	-.367**	-.142	.136
	Sig. (2-tailed)	.001	.001	.734	.171
Right knee joint circumference	Pearson Correlation	.349**	-.349**	.038	.151
	Sig. (2-tailed)	.216	.216	.315	.399
Right Q angle	Pearson Correlation	.119	-.119	-.024	.087
	Sig. (2-tailed)	.295	.295	.835	.444
Percent body fat	Pearson Correlation	.378**	-.378**	.130	.208
	Sig. (2-tailed)	.001	.001	.252	.064
Lean body weight	Pearson Correlation	.341**	-.341**	.039	.232*
	Sig. (2-tailed)	.002	.002	.730	.039
Visceral fat	Pearson Correlation	.416**	-.416**	.008	.237*
	Sig. (2-tailed)	.000	.000	.943	.034

**. Correlation was significant at the 0.01 level (2-tailed).

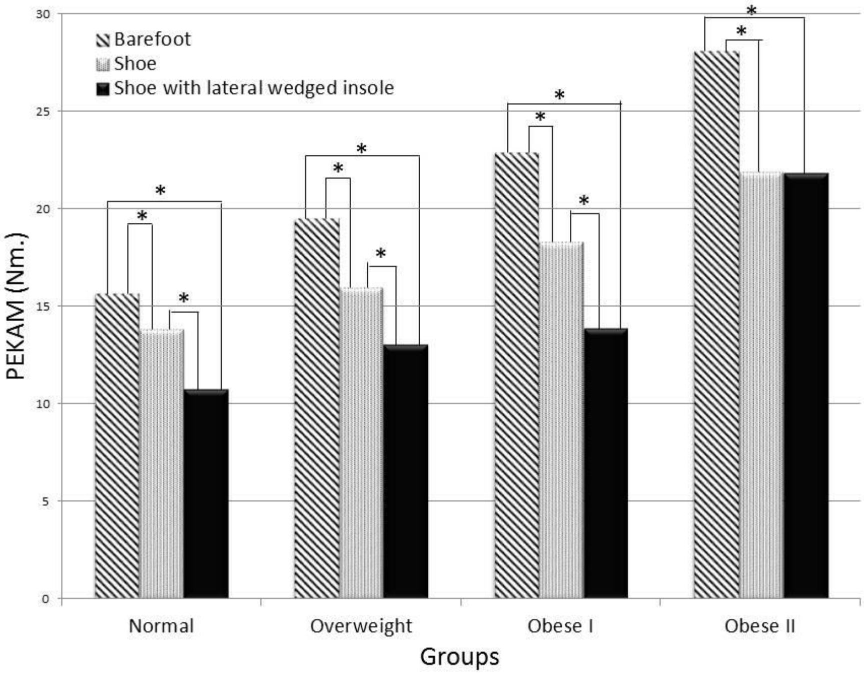
*. Correlation was significant at the 0.05 level (2-tailed).

Table 20 Correlations between parameters during gait (continued).

Parameters		Support Time	Non Support	Step length	Double support time
Peak knee flexion angle	Pearson Correlation	-.044	.043	.080	-.139
	Sig. (2-tailed)	.701	.702	.480	.219
Peak VGRF	Pearson Correlation	.145	-.145	.070	-.023
	Sig. (2-tailed)	.198	.198	.535	.838
PEKADM (normalized)	Pearson Correlation	-.129	.129	-.081	-.163
	Sig. (2-tailed)	.255	.255	.475	.148
KAAI (normalized)	Pearson Correlation	-.128	.128	-.097	-.091
	Sig. (2-tailed)	.257	.258	.394	.420
PEKADM (non normalized)	Pearson Correlation	.051	-.051	-.034	-.068
	Sig. (2-tailed)	.654	.656	.768	.551
KAAI (non normalized)	Pearson Correlation	.019	-.019	-.085	-.005
	Sig. (2-tailed)	.869	.869	.453	.967
Step Width	Pearson Correlation	.111	-.111	-.142	.084
	Sig. (2-tailed)	.329	.329	.208	.456
Velocity	Pearson Correlation	-.196	.196	.377**	-.243*
	Sig. (2-tailed)	.081	.082	.001	.030
Stride length	Pearson Correlation	-.064	.064	.721**	-.013
	Sig. (2-tailed)	.570	.570	.000	.907
Cadence	Pearson Correlation	-.080	.080	-.066	-.211
	Sig. (2-tailed)	.483	.483	.562	.061
Support Time	Pearson Correlation	1	-1.000**	-.015	.473**
	Sig. (2-tailed)		.000	.895	.000
Non Support	Pearson Correlation	-1.000**	1	.015	-.473**
	Sig. (2-tailed)	.000		.895	.000
Step length	Pearson Correlation	-.015	.015	1	-.046
	Sig. (2-tailed)	.895	.895		.688
Double support time	Pearson Correlation	.473**	-.473**	-.046	1
	Sig. (2-tailed)	.000	.000	.688	

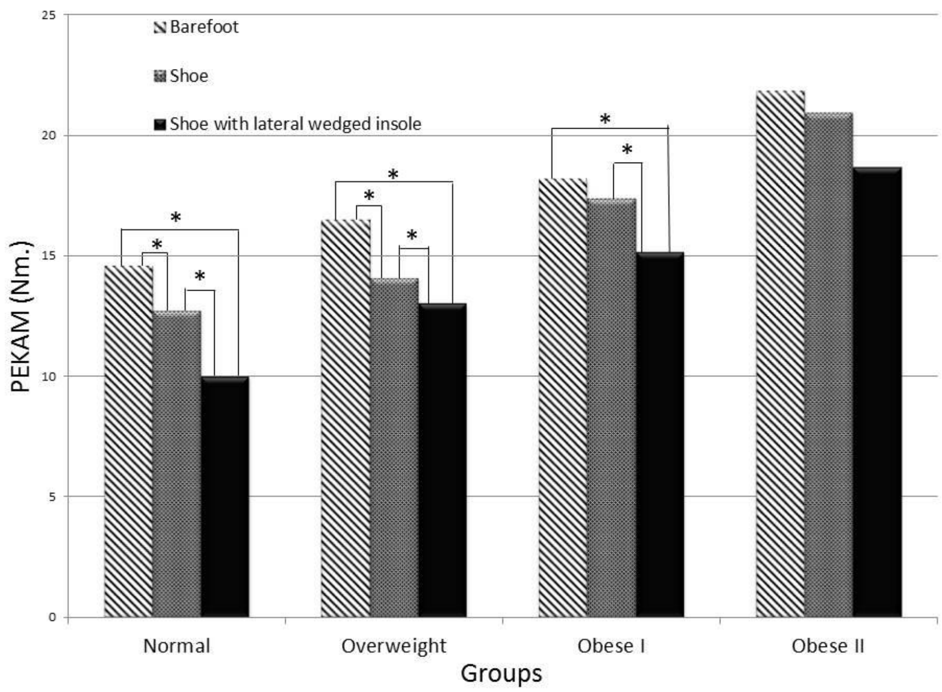
**. Correlation was significant at the 0.01 level (2-tailed).

*. Correlation was significant at the 0.05 level (2-tailed).



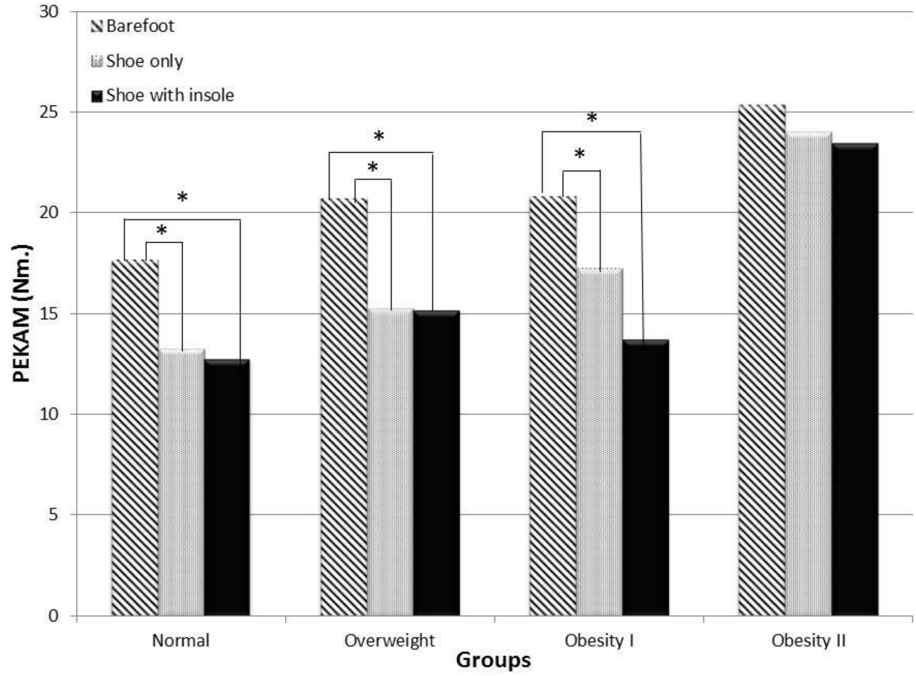
*** significantly different between walking conditions (*p < 0.05)***

Figure 40 Peak external knee adduction moment (Nm) in the three conditions during walking at self-selected speed.



* significantly different between walking conditions (*p < 0.05)*

Figure 41 Peak external knee adduction moment (Nm) in the three conditions during walking at constant velocity (1.24 m/s).



* significantly different between walking conditions ($p < 0.05$)

Figure 42 Peak external knee adduction moment (Nm) in the three conditions during walking at Froude velocity (FR).

4.3 Study 2

The influence of the shoe and lateral wedge insoles on knee kinetics was compared among the four experimental groups to examine how the insole impacted the knee joints. Table 25 presents the knee kinetics of the four body mass index groups during walking at self-selected speed (SS). We found that there was significant difference in the peak external knee adduction moment in all body mass index groups during walking with shoes compared with barefoot, and shoes + insoles compared with barefoot. However, there was no significant different for the peak external knee adduction moment of shoes (21.88 ± 2.36) compared with shoes + insoles (21 ± 1.97) walking in only the obese II group (*p = 1.000*) as presented in Figure 34. Moreover, there was no significant difference in PVGRF, and knee adduction angular impulse between the 3 walking conditions in all BMI groups (figure 40).

Table 26 presents the knee kinetics of the four body mass index groups during walking at constant velocity (1.24 m/s). We found that there was significant difference in the peak external knee adduction moment in the normal and overweight groups during walking with shoes compared with barefoot, shoes + insoles compared with barefoot and shoes compared with shoes + insoles. There was a significant difference in the peak external knee adduction moment during walking with shoes + insoles compared with barefoot (*p = 0.043*) and shoes compared with shoes + insoles (*p = 0.046*) in the obese I group as shown in Figure 35. However, there were no significant differences in all kinetic parameters during the 3 walking conditions for the obese II group(figure 41).

Table 27 presents the knee kinetics of the four body mass index groups during walking at Froude velocity. We found that there was significant difference in the peak external knee adduction moment in the normal, overweight and obesity I groups during walking with shoes compared with barefoot, and shoes + insoles compared with barefoot. There was a significant difference in the peak external knee adduction moment during walking with shoes + insoles compared with barefoot (p = *0.025*, *0.014*, *0.023*) and shoes compared with barefoot (p = *0.036*, *0.018*, *0.049*) in the normal, overweight and obese I groups, respectively. As shown in Figure 36. However, there were no significant differences in all kinetic parameters during the 3 walking conditions for the obese II group (figure 42).

CHAPTER V

DISCUSSION

This study focuses on knee biomechanics in difference BMI level women and the effect of lateral wedge insole on knee biomechanics that related with knee osteoarthritis risk factors. We found that the gait pattern and knee joint load parameters were changed when the BMI was increased. The lateral wedge insole could reduce the knee joint load such as the peak knee adduction moment in normal, overweight, and obesity I groups. However, there was no effect of lateral wedge insole to decrease the peak knee adduction moment in obesity II group.

The result of our study showed that the different BMI level in woman had effect to gait patterns during each walking speed. The present study was used motion analysis to measure spatiotemporal, kinetic, and kinematic parameters over a wide range of BMI for Asia Pacific region population. Our findings were consistent with previous studies that higher BMI level tend to have wider step width, longer stance time and lesser non-support time while walking compared with normal group that indicated to change the gait pattern for dynamic posture stabilization. (McGraw et al., 2000;Lai et al., 2008;Koo et al., 2011). Walking efficiency has been shown to be negatively correlated with body adiposity, rather than body weight. The higher energy cost associated with walking in obese to increased internal work resulting from a larger moment of inertia created by disproportionately heavier limbs. The upper body anterior tilt, greater vertical displacement of center of mass and extraneous movements resulting from greater limb reduce the mechanical efficiency of walking in obese, thereby increasing the metabolic cost of the task. The wider step width in obese during walking was suggested to improve the dynamic stability of body that been shown to raise the metabolic cost of walking. As the wider step width, the dynamic stability was maintained by change the gait pattern such as longer stance phase duration, shorter swing phase and higher double support time (Wearing et al., 2006).

The knee flexion was meant to improve the absorption of additional impact imposed by heavier weight (Koo et al., 2011). Furthermore, peak knee flexion angles during the stance phase have been reported as being lower and a smaller range of knee motion in obese people has been reported (Vartiainen et al., 2012) (Beijersbergen et al., 2013). However, from our study we found that there was no significant in the peak knee flexion angle during stance phase due to all groups and walking velocities. It might be reflected that the joint load absorption effect from knee flexion mechanics during stance phase of walking was no difference in difference BMI level female. The higher knee joint load such as peak knee adduction moment person would be prone to induce articular cartilage injury that cause of the knee joint osteoarthritis. The previous studies found that obesity individuals walk with similar sagittal plan motion of knee and hip, but altered frontal plane of hip such as pelvic drop kinematics compared to non obese individuals at walking speed 1.25 m/s, which was close to the reported preferred walking speed for obese adults 1.29 m/s (Lerner et al., 2014). At fast walking speed (1.5 m/s), obese individuals walk with both sagittal and frontal plane kinematic alterations that more extended knee during loading response phase and greater pelvic obliquity during single support phase (Lerner et al., 2014). The hip and ankle joint motion should be investigated in the further study.

The peak VGRF was the cause of external load to act on lower extremity joint especially knee joint during walking. The increased forced passing across the articular surfaces may be prone to musculoskeletal pathology that induced cartilage damage cause of osteoarthritis (Sheehan and Gormley, 2013) (Freedman Silvernail et al., 2013). From our study found that there was no significantly difference in the peak VGRF during walking all velocities that cause of normalizing with body weight. However, the articular architecture of all groups may be same in female participant groups. The raw vertical ground reaction force may better represent the risk of joint injury than normalized peak vertical ground reaction force (Freedman Silvernail et al., 2013).

The knee adduction moment reflects the dynamic joint load on the medial compartment, and was a predictor for the risk of progression for the medial compartment osteoarthritis (Maly et al.) (Hunt and Bennell, 2011); (Miyazaki et al., 2002). The medial and lateral cartilage thickness variations in the knee were influenced by the peak knee adduction moment during normal walking (Koo et al., 2011). The joint surfaces of obese were exposed to increase loading. Therefore, for adults who were obese the articular surfaces of the knee were subjected to greater absolute forces, and as a result may be prone to musculoskeletal pathology (Sheehan and Gormley, 2013). Investigations of joint loading in the knee typically normalize the knee adduction moment to global measures of body size such as body weight and height to allow comparison between individuals. However, such measurements may not reflect the knee architecture that affects the force acting upon the articular surface. The recent studies found that normalized peak external knee adduction moment were not sensitive to osteoarthritis severity; however, the non-normalization technique was superior at distinguishing between osteoarthritis severities (Robbins et al., 2011;Robbins et al., 2013;Brisson et al., 2015). In the present study, we present the results both of normalized and non-normalized peak external knee adduction moment. Our study found that the peak external knee adduction moment (Nm/w) was least in obesity I group for all walking velocities. However, the peak external knee adduction moment (Nm) was highest in obesity II groups comparing with the others in figure 33. As previous studies, that shown increased peak knee adduction (Lai et al., 2008) (Freedman Silvernail et al., 2013) have also been reported for adults with excess body mass (Sheehan and Gormley, 2013). From the previous studies, the body mass index (BMI) significantly correlates with the incident risk of radiological and symptomatic knee osteoarthritis that was shown in each BMI increase by 1 kg/m^2 above 27 was associated with a 15% increased risk which cause a reduction of fatigue durability of articular cartilage (Berenbaum and Sellam, 2008;Blagojevic et al., 2010;Landinez-Parra et al., 2011). A 5-unit increase in body mass index was associated with an 35% increased risk of knee osteoarthritis (Jiang et al., 2011). Mechanical stress resulting from a high body mass index (BMI) was known to be a risk factor for the development of knee osteoarthritis and better understanding of the positive effect of obesity on osteoarthritis development (Jiang et al., 2011). The result of our study that might confirmed the higher BMI level in female to be risk of articular cartilage damage caused of knee osteoarthritis.

The knee adduction angular impulse was the product of the average of briefly applied force, times the radius of a rigid body, times the brief time period, that was the time integral of the knee adduction moment over the stance phase. The knee adduction angular impulse provides a quantitative, albeit surrogate, measure of how the medial compartment was loaded during the entire stance phase of gait (Henriksen et al., 2012). From our study we found that the normalize knee adduction moment impulse was least in obesity I group that should be not reflect the knee joint load. However, the non-normalized knee adduction angular impulse was investigated and there was no significantly difference in all groups during walking with all velocities.

The EMG and the direction of the center of pressure in the foot during stance phase measurement was not investigated since it was out of the scope of this study. Further study should be investigate the effects of lateral wedge insoles to reduce knee joint load and their clinical effects.

The purpose of the second study was to determine the effect the lateral wedge insole could have in decreasing the kinetic risk factors for knee osteoarthritis in normal, overweight, obese I and obese II women. The peak external knee adduction moment was decreased as an effect of shoes and shoes with lateral wedge insoles in the self-selected speed walking condition that represented the natural walking of the participant. However, the peak external knee adduction moment did not change in the obese II group during walking at constant velocity ($p = 1.000$). Additionally, the lateral wedge insoles had no effect on knee adduction angular impulse during walking.

Recent studies, have reported lateral wedge insoles could reduce the peak external knee adduction moment in normal and obese participants with knee osteoarthritis (Weinhandl et al.;Kerrigan et al., 2002;Reilly et al., 2006;Russell and Hamill, 2011;Hinman et al., 2012;Barrios et al., 2013). The peak external knee adduction moment was the result of the ground reaction force multiplied by the moment arm from the knee joint center. The peak external knee adduction moment during the stance phase gait has been characterized both as a determinate and a surrogate for dynamic medial knee joint load. The peak external knee adduction moment reflects the medial-to-lateral joint load distribution and has been associated with lower limb varus alignment, medial osteoarthritis disease severity, and medial-to-lateral bone mineral density ratio. The peak external knee adduction moment during the stance phase potentially captures the maximal medial joint load experienced at any one instance in time. Efforts have been directed toward developing and testing interventions that lower peak external knee adduction moment with the ultimate goal of modifying disease course in medial tibiofemoral osteoarthritis. The knee adduction angular impulse was the product of the average of briefly applied force, times the radius of a rigid body, times the brief time period (delta t), that was the time integral of the knee adduction moment over the stance phase. The knee adduction angular impulse provides a quantitative, albeit surrogate, measure of how the medial compartment was loaded during the entire stance phase of gait. Laterally wedge insoles were one of the gait modifications potentially slowing down progression of medial knee osteoarthritis by inducing the lateral shift of the center of pressure in the foot. This shifts the frontal plane ground reaction force vector towards the knee joint center. Together with the more vertically oriented ground reaction force, this serves to reduce the moment arm of the knee adduction moment cause of the central mechanism explaining the load-reducing effect of lateral wedge insoles (Weinhandl et al.;Hinman et al., 2012).

In a recent study, we there to be no effect of lateral wedge insoles on the peak external knee adduction moment in the obese II group ($p = 1.000$ in self selected speed and $p = 0.930$ in constant velocity). As can be seen from the mechanism of the action of the lateral wedge insoles on knee joint load, the degree of the lateral wedge insoles could have effect on reducing the knee to ground reaction force lever arm by increasing the lateral shift of the centre of pressure (Hinman et al., 2012). The 5-degree lateral wedge insole could reduce the knee joint load parameter among normal subjects (Jones et al., 2013). However, in the obese II group, the 5 degree lateral wedge insole might be not appropriate for reducing knee joint load by reduction of lever arm. A recent study recommended a 4 – degree lateral wedge insole could reduce the knee joint load and the higher lateral wedging degree could further reduce peak external knee adduction moment but be less comfortable (> 8 degrees) among the normal group (Tipnis et al., 2014). From our findings, we suggest that the degree of lateral wedge insoles should be of concern in applying a higher body mass index for reducing the knee joint load. However, no significant difference was found for the knee adduction angular impulse representing the whole stance phase knee joint load of all body mass index level groups in both walking velocity conditions in this study. We suggest that the degree of the lateral wedge insoles and the length of time for their application in effectively reducing the knee joint load be further studied in the future.

Investigations of joint loading in the knee typically normalize the knee adduction moment to global measures of body size such as body weight and height to allow comparison between individuals. However, such measurements may not reflect the knee architecture that affects the force acting upon the articular surface. Recent studies found that normalized peak external knee adduction moment were not sensitive to osteoarthritis severity; however, the non-normalization technique was superior at distinguishing between osteoarthritis severities (Robbins et al., 2011;Robbins et al., 2013;Brisson et al., 2015). In the present study, we present the results both of normalized and non-normalized Peak external knee adduction moment.

The subtalar joint movement and the direction of the center of pressure in the foot during stance phase measurement was not investigated since it was out of the scope of this study. Further study should be investigate the acute effects of lateral wedge insoles to determine long-term use and their clinical effects.

CHAPTER VI

CONCLUSION

The healthy overweight and obese participants had changed gait pattern and knee biomechanics during walking in all type of velocities. The knee adduction moment was higher in obese participants that represent the higher knee joint load to be induced joint cartilage damage cause of osteoarthritis. The peak external knee adduction moment was decreased by the effect of shoes and shoes with lateral wedge insoles in the self-selected speed walking condition for normal, overweight, and obese I participants. However, the peak external knee adduction moment did not have any effect upon the obese II group. Additionally, there was no effect of lateral wedge insoles on knee adduction angular impulse in all groups. The degree of lateral wedge insoles for reducing knee joint load among obese II individuals was of interest. The long-term effect and length of time for using lateral wedge insoles while preventing or delaying knee osteoarthritis onset among various body mass index levels requires further research.

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APPENDICES

APPENDIX A
INFORMED CONSENT

หนังสือแสดงความยินยอมเข้าร่วมการวิจัย

ทำที่...

วันที่..............เดือน...........................พ.ศ. ....

เลขที่ ……………………

ข้าพเจ้า ซึ่งได้ลงนามท้ายหนังสือนี้ ขอแสดงความยินยอมเข้าร่วมโครงการวิจัย ชื่อโครงการวิจัย ชีวกลศาสตร์ของข้อเข่าขณะเดินในหญิงอ้วน ชื่อผู้วิจัย นางสาวชิดชนก นุตาลัย ที่อยู่ที่ติดต่อ 42/56 หมู่บ้านพาร์คราชาวดี พระราม 2 ซอย 3 ถนนพระราม 2 บางมด จอมทอง กรุงเทพ 10150 โทรศัพท์ 080-073-7269

ข้าพเจ้า **ได้รับทราบ**รายละเอียดเกี่ยวกับที่มาและวัตถุประสงค์ในการทำวิจัย รายละเอียดขั้นตอนต่างๆ ที่จะต้องปฏิบัติหรือได้รับการปฏิบัติ ความเสี่ยง/อันตราย และประโยชน์ซึ่งจะเกิดขึ้นจากการวิจัยเรื่องนี้ โดยได้อ่านรายละเอียดในเอกสารชี้แจงผู้เข้าร่วมการวิจัยโดยตลอด และ**ได้รับคำอธิบาย**จากผู้วิจัย **จนเข้าใจเป็นอย่างดี**แล้ว

ข้าพเจ้าจึง**สมัครใจ**เข้าร่วมในโครงการวิจัยนี้ ตามที่ระบุไว้ในเอกสารชี้แจงผู้เข้าร่วมการวิจัย โดยข้าพเจ้ายินยอม เข้ารับการตรวจวัดสัดส่วนร่างกาย แต่งตัวด้วยชุดสำหรับเก็บข้อมูล และ ติด reflective marker บนผิวหนังตามส่วนต่างๆ ของร่างกาย เดินเท้าเปล่า เดินสวมรองเท้า และ เดินสวมรองเท้าพร้อมแผ่นสอดในรองเท้า ด้วยความเร็วในการเดินที่กำหนด 3 ความเร็ว รวมทั้งสิ้นเดิน 45 รอบ เมื่อเสร็จสิ้นการวิจัยแล้วข้อมูลที่เกี่ยวข้องกับผู้มีส่วนร่วมในการวิจัยจะถูกทำลาย

ข้าพเจ้ามีสิทธิ**ถอนตัว**ออกจากการวิจัยเมื่อใดก็ได้ตามความประสงค์ **โดยไม่ต้องแจ้งเหตุผล** ซึ่งการถอนตัวออกจากการวิจัยนั้น จะไม่มีผลกระทบในทางใดๆ ต่อข้าพเจ้าทั้งสิ้น

ข้าพเจ้าได้รับคำรับรองว่า ผู้วิจัยจะปฏิบัติต่อข้าพเจ้าตามข้อมูลที่ระบุไว้ในเอกสารชี้แจงผู้เข้าร่วมการวิจัย และข้อมูลใดๆ ที่เกี่ยวข้องกับข้าพเจ้า ผู้วิจัยจะ**เก็บรักษาเป็นความลับ** โดยจะนำเสนอข้อมูลการวิจัยเป็นภาพรวมเท่านั้น ไม่มีข้อมูลใดในการรายงานที่จะนำไปสู่การระบุตัวข้าพเจ้า

หากข้าพเจ้าไม่ได้รับการปฏิบัติตรงตามที่ได้ระบุไว้ในเอกสารชี้แจงผู้เข้าร่วมการวิจัย ข้าพเจ้าสามารถร้องเรียนได้ที่คณะกรรมการพิจารณาจริยธรรมการวิจัยในคน กลุ่มสหสถาบัน ชุดที่ 1 จุฬาลงกรณ์มหาวิทยาลัย ชั้น 4 อาคารสถาบัน 2 ซอยจุฬาลงกรณ์ 62 ถนนพญาไท เขตปทุมวัน กรุงเทพฯ 10330 โทรศัพท์ 0-2218-8147, 0-2218-8141 โทรสาร 0-2218-8147 **E-mail: eccu@chula.ac.th**

ข้าพเจ้าได้ลงลายมือชื่อไว้เป็นสำคัญต่อหน้าพยาน ทั้งนี้ข้าพเจ้าได้รับสำเนาเอกสารชี้แจงผู้เข้าร่วมการวิจัย และสำเนาหนังสือแสดงความยินยอมไว้แล้ว

ลงชื่อ..
(นางสาวชิดชนก นุตาลัย)
ผู้วิจัยหลัก

ลงชื่อ..
(..)
ผู้มีส่วนร่วมในการวิจัย

ลงชื่อ..
(..)
พยาน

APPENDIX B
PARTICIPANT INFORMATION SHEET

ข้อมูลสำหรับผู้มีส่วนร่วมในการวิจัย

ชื่อโครงการวิจัย ชีวกลศาสตร์ของข้อเข่าขณะเดินในหญิงอ้วน

ชื่อผู้วิจัย นางสาวชิดชนก นุตาลัย ตำแหน่ง นิสิตปริญญาเอก สาขากายภาพบำบัด
สถานที่ติดต่อผู้วิจัย (ที่ทำงาน) ภาควิชากายภาพบำบัด คณะสหเวชศาสตร์ จุฬาลงกรณ์มหาวิทยาลัย
(ที่บ้าน) 42/56 หมู่บ้านพาร์คราชาวดี พระราม 2 ซอย 3 ถนนพระราม2 บางมด จอมทอง กรุงเทพฯ
โทรศัพท์มือถือ 080-073-7269 E-mail Fernnok@hotmail.com

1. ขอเรียนเชิญท่านเข้าร่วมในการวิจัยก่อนที่ท่านจะตัดสินใจเข้าร่วมในการวิจัย มีความจำเป็นที่ท่านควรทำความเข้าใจว่างานวิจัยนี้ทำเพื่อหาความแตกต่างของปัจจัยในการเคลื่อนไหวและแรงที่ข้อเข่าขณะเดินของหญิงไทยที่มีดัชนีมวลกายในระดับต่างกัน กรุณาใช้เวลาในการอ่านข้อมูลต่อไปนี้อย่างละเอียดรอบคอบ และสอบถามข้อมูลเพิ่มเติมหรือข้อมูลที่ไม่ชัดเจนได้ตลอดเวลา

2. โครงการนี้เกี่ยวข้องกับการวิจัยการเคลื่อนไหวและแรงที่ข้อเข่าขณะเดินในหญิงไทยที่มีดัชนีมวลกายในระดับต่างกัน

3. วัตถุประสงค์ของการวิจัย

- เพื่อศึกษาความแตกต่างของ รูปแบบการเดิน องศาการเคลื่อนไหว และ แรงที่ข้อเข่าขณะเดินในสตรีที่มีระดับดัชนีมวลกายต่างกัน
- เพื่อศึกษาความแตกต่างของ รูปแบบการเดิน องศาการเคลื่อนไหว และ แรงที่ข้อเข่าขณะเดิน เมื่อเดินเท้าเปล่า สวมรองเท้า และ สวมรองเท้าเสริมด้วยแผ่นรองในรองเท้า ในสตรีที่มีระดับดัชนีมวลกายต่างกัน
- เพื่อศึกษาความสัมพันธ์ของปัจจัยทางชีวกลศาสตร์ เช่น รูปแบบการเดิน องศาการเคลื่อนไหว และ แรง ต่อระดับดัชนีมวลกายในหญิงไทย

4. รายละเอียดของผู้มีส่วนร่วมในการวิจัย

- ลักษณะของผู้มีส่วนร่วมในการวิจัย เป็น หญิงไทย สุขภาพแข็งแรง ไม่มีโรคประจำตัว

- เกณฑ์การคัดเข้า
  - ไม่มีอาการบาดเจ็บที่ขาซึ่งส่งผลต่อการเดิน
  - ไม่มีประวัติโรคข้อเข่าเสื่อม หรือ ได้รับการรักษาโรคข้อเข่าเสื่อม
  - น้ำหนักเปลี่ยนแปลงภายใน 3 เดือน รวมไม่เกิน 2.5 กิโลกรัม
  - เท้ามีลักษณะปกติ ไม่เป็นเท้าแบน หรือ ความสูงของส่วนโค้งภายในเท้ามากกว่าปกติ
  - มุมข้อเข่าเบี่ยงออกนอก และ เข้าด้านในโดยวัดทางด้านหน้าไม่เกิน 10 องศา
  - เป็นผู้ไม่ออกกำลังกายเป็นประจำ
- เกณฑ์การคัดออก
  - ไม่สามารถเข้าร่วมงานวิจัยจนเสร็จสิ้นกระบวนการเก็บข้อมูล
  - เกิดอาการปวด หรือ อาการอื่นๆ ที่ส่งผลกระทบต่อการเก็บข้อมูล
- จำนวนทั้งสิ้น 80 คน
- วิธีการได้มาซึ่งผู้มีส่วนร่วมในการวิจัย โดยการประชาสัมพันธ์
- เหตุผลที่ได้รับเชิญเข้าร่วมโครงการวิจัย คือ เป็นผู้ที่มีคุณสมบัติตรงตามเกณฑ์คัดเข้าในงานวิจัย
- การแบ่งกลุ่มผู้มีส่วนร่วมในการวิจัยมี 4 กลุ่ม ตามระดับดัชนีมวลกาย กลุ่มละ 20 คน

5. กระบวนการการวิจัยที่กระทำต่อผู้มีส่วนร่วมในการวิจัย นางสาวชิดชนก นุตาลัย (ผู้วิจัยหลัก) จะเป็นผู้ดำเนินการตรวจวัดสัดส่วนร่างกาย ติด reflective marker (อุปกรณ์บอกตำแหน่งด้วยการสะท้อนแสง) บนผิวหนังตามส่วนต่างๆ ของร่างกาย และแจ้งความเร็วการเดินในแต่ละรอบให้ทราบ ณ ห้องปฏิบัติการ Motor control and motion analysis research laboratory ชั้น 1 อาคารจุฬาพัฒน์ 2 ภาควิชากายภาพบำบัด คณะสหเวชศาสตร์ จุฬาลงกรณ์มหาวิทยาลัย ใช้เวลา 3 ชั่วโมง โดยในงานวิจัยจะมีการบันทึกข้อมูล องศาการเคลื่อนไหวของผู้มีส่วนร่วมในการวิจัย เป็นภาพจาก reflective marker (อุปกรณ์บอกตำแหน่งด้วยการสะท้อนแสง) เมื่อเสร็จสิ้นการวิจัยแล้วข้อมูลที่เกี่ยวข้องกับผู้มีส่วนร่วมในการวิจัยจะถูกทำลาย

6. กระบวนการให้ข้อมูลแก่ผู้มีส่วนร่วมในการวิจัย ทำด้วยวิธีการอธิบายจนกว่าจะเข้าใจและหมดข้อสงสัย โดย นางสาวชิดชนก นุตาลัย (ผู้วิจัยหลัก) เป็นผู้ดำเนินการ

7. หากผู้มีส่วนร่วมในการวิจัยไม่อยู่ในเกณฑ์คัดเข้า ผู้วิจัยจะให้ความรู้เกี่ยวกับการดูแลสุขภาพของคนอ้วน หรือ การดูแลข้อเข่าที่เหมาะสมในแต่ละกรณี

8. **ความเสี่ยง**ที่อาจเกิดขึ้นแก่ผู้มีส่วนร่วมในการวิจัย ในการศึกษานี้อาจจะเกิดความล้าจากการเดินหลายรอบ ผู้วิจัยหลักจึงกำหนดให้มีการพักระหว่างการเดินแต่ละรอบ เพื่อลดโอกาสในการเกิดความล้าของกล้ามเนื้อ และเมื่อเสร็จสิ้นกระบวนการเก็บข้อมูล จะแนะนำวิธีการยืดกล้ามเนื้อขาเพื่อลดอาการล้าของกล้ามเนื้อให้กับผู้มีส่วนร่วมในการวิจัย

9. การศึกษาในครั้งนี้เป็นการค้นหาปัจจัยทางชีวกลศาสตร์ที่แตกต่างกันในสตรีที่มีระดับดัชนีมวลกายต่างกัน ซึ่งจะมีประโยชน์ต่อการนำไปสู่การสร้างองค์ความรู้ใหม่ และ ศึกษาแนวทางในการจัดการกับปัจจัยทางชีวกลศาสตร์ที่พบในการศึกษานี้ เพื่อป้องกันการเกิดภาวะข้อเข่าเสื่อมในอนาคต

10. การเข้าร่วมในการวิจัยของท่านเป็นโดยสมัครใจ และสามารถปฏิเสธที่จะเข้าร่วมหรือถอนตัวจากการวิจัยได้ทุกขณะ โดยไม่ต้องให้เหตุผลและไม่สูญเสียประโยชน์ที่พึงได้รับ

11. หากท่านมีข้อสงสัยให้สอบถามเพิ่มเติมได้โดยสามารถติดต่อผู้วิจัยได้ตลอดเวลา และหากผู้วิจัยมีข้อมูลเพิ่มเติมที่เป็นประโยชน์หรือโทษเกี่ยวกับการวิจัย ผู้วิจัยจะแจ้งให้ท่านทราบอย่างรวดเร็ว

12. ข้อมูลที่เกี่ยวข้องกับท่านจะ**เก็บเป็นความลับ** หากมีการเสนอผลการวิจัยจะเสนอเป็นภาพรวม ข้อมูลใดที่สามารถระบุถึงตัวท่านได้จะไม่ปรากฏในรายงาน

13. มีการจ่ายค่าพาหนะ ค่าชดเชยการเสียเวลา

14. หากท่านไม่ได้รับการปฏิบัติตามข้อมูลดังกล่าวสามารถร้องเรียนได้ที่

คณะกรรมการพิจารณาจริยธรรมการวิจัยในคน กลุ่มสหสถาบัน ชุดที่ 1 จุฬาลงกรณ์มหาวิทยาลัย
ชั้น 4 อาคารสถาบัน 2 ซอยจุฬาลงกรณ์ 62 ถนนพญาไท เขตปทุมวัน กรุงเทพฯ 10330
โทรศัพท์ 0-2218-8147 หรือ 0-2218-8141 โทรสาร 0-2218-8147 E-mail: eccu@chula.ac.th

APPENDIX C
SCREENING QUESTIONNAIRE

แบบสอบถามคัดเลือกบุคคลเข้าร่วมงานวิจัยวิทยานิพนธ์
เรื่อง
ชีวกลศาสตร์ของข้อเข่าขณะเดินในหญิงอ้วน

คำชี้แจง : โปรดกรอกข้อมูลและตอบคำถามต่อไปนี้ตามความเป็นจริง ข้อมูลทั้งหมดในแบบสอบถามนี้เป็นข้อมูลเฉพาะบุคคล โดยใช้ประกอบการวิจัยเท่านั้น

1. ชื่อ - นามสกุล ……………………………………….. ……………… อายุ …………... ปี
2. น้ำหนัก ปัจจุบัน………………...…. กิโลกรัม 1 เดือนก่อน...........................กิโลกรัม
   2 เดือน...............................กิโลกรัม 3 เดือนก่อน...........................กิโลกรัม
   6 เดือน...............................กิโลกรัม 1ปี ก่อน.................................กิโลกรัม
   ระยะเวลาที่ดัชนีมวลกายมากกว่า 23.9 กิโลกรัม/เมตร^2…………..……..ปี/เดือน
3. ส่วนสูง ……………..…... เซนติเมตร ดัชนีมวลกาย............................กิโลกรัม/เมตร^2
4. ขาข้างที่ถนัด …………… (สังเกต จากการทดสอบให้เตะบอล)
5. ที่อยู่ปัจจุบันที่สามารถติดต่อได้สะดวก
   ……………………………………………………………………………..…………………
   ……………………………………………………………………………..…………………
   เบอร์โทรศัพท์ …………………………. E-mail address …………………………………...

6. โรคประจำตัว
   - ตรวจสุขภาพประจำปี ในช่วง 1 ปี ที่ผ่านมา ☐ ตรวจ ☐ ไม่ตรวจ
   - ความดันโลหิตสูง ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
   - เบาหวาน ชนิด.. ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
   - โรคหัวใจ ชนิด.. ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
   - ภาวะไขมันในเลือดสูง ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
   - โรคไต ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
   - โรคมะเร็ง ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ

- โรคไทรอยด์ ชนิด.................................. ☐ เป็น ☐ ไม่เป็น ☐ ไม่ทราบ
- อื่นๆ โปรดระบุ..

7. ประวัติเกี่ยวข้องกับฮอร์โมนเอสโตรเจน
   - ประจำเดือนมาเป็นปกติทุกเดือน ☐ ใช่ ☐ ไม่ใช่
   - ปัจจุบันกำลังตั้งครรภ์ ☐ ใช่ ☐ ไม่ใช่
   - เคยผ่าตัดมดลูก ☐ ใช่ ☐ ไม่ใช่

8. ประวัติการใช้ยาในปัจจุบัน
   - ยาลดความดันโลหิตสูง ☐ ใช่ ☐ ไม่ใช่
   - ยาควบคุมระดับน้ำตาลในเลือด หรือ อินซูลิน ☐ ใช่ ☐ ไม่ใช่
   - ยาควบคุมระดับไขมันในเลือด ☐ ใช่ ☐ ไม่ใช่
   - ยาควบคุมระดับฮอร์โมนไทรอยด์ ☐ ใช่ ☐ ไม่ใช่
   - ยาคุมกำเนิด ☐ ใช่ ☐ ไม่ใช่
   - ยาทดแทนฮอร์โมนเอสโตรเจน ☐ ใช่ ☐ ไม่ใช่

9. พฤติกรรมการออกกำลังกาย
   - ท่านเป็นนักกีฬาอาชีพหรือไม่ ☐ ใช่ ☐ ไม่ใช่
   - ท่านออกกำลังกายเป็นประจำ หรือไม่ ☐ ใช่ ☐ ไม่ใช่

   รูปแบบการออกกำลังกาย.................................... นาน.....................ชั่วโมง/นาที

   ความถี่..ครั้ง/สัปดาห์

10. ลักษณะงาน
   - งานในสำนักงาน ☐ ใช่ ☐ ไม่ใช่
   - ยืนมากกว่า 1 ชั่วโมง/วัน ☐ ใช่ ☐ ไม่ใช่
   - ยกของหนัก 25 กิโลกรัม 10 ครั้ง/สัปดาห์ หรือ 50 กิโลกรัม 1 ครั้ง/สัปดาห์ ☐ ใช่ ☐ ไม่ใช่
   - นั่งคุกเข่า หรือ นั่งย่อเขา มากกว่า 1 ชั่วโมง/วัน ☐ ใช่ ☐ ไม่ใช่

11. ประวัติการบาดเจ็บ ระบบกระดูกและกล้ามเนื้อ

   ☐ ปวดหลัง เป็นระยะเวลา …………. วัน / เดือน / ปี

☐ ปวดสะโพก ข้าง ☐ ซ้าย ☐ ขวา เป็นระยะเวลา …………. วัน / เดือน / ปี

☐ ปวดเข่า ข้าง ☐ ซ้าย ☐ ขวา เป็นระยะเวลา …………. วัน / เดือน / ปี

☐ ปวดข้อเท้า/เท้า ข้าง ☐ ซ้าย ☐ ขวา เป็นระยะเวลา …………. วัน / เดือน / ปี

☐ มีเสียงกร๊อบแกร๊บ ภายในข้อเข่า ขณะเคลื่อนไหว

☐ มีอาการข้อเข่าติด หรือ ขยับข้อเข่าได้ยาก หลังจากตื่นนอนตอนเช้า

☐ ปวดกล้ามเนื้อขา ข้าง ☐ ซ้าย ☐ ขวา

เป็นระยะเวลา …………….. วัน / เดือน / ปี ก่อนการทดสอบ

☐ กระดูกหัก โปรดระบุตำแหน่งที่หัก ………….. ....

การรักษาที่ได้รับ ………………………………………………………………

เป็นระยะเวลา …………………………….……. วัน / เดือน / ปี ก่อนการทดสอบ

☐ มีอาการบวม ระบุตำแหน่ง...

12. พฤติกรรมการสวมรองเท้า

- ขนาดรองเท้าที่สวมอยู่เป็นประจำในปัจจุบัน……………………………………...
- ชนิดของรองเท้าที่ท่านสวมอยู่เป็นประจำในปัจจุบัน

☐ รองเท้าส้นสูง.........................นิ้ว ☐ รองเท้าผ้าใบ ยี่ห้อ..........................

☐ รองเท้าแตะ ☐ อื่นๆ โปรดระบุ.............................

-.-

สำหรับผู้วิจัย ☐ ผ่าน ☐ ไม่ผ่าน

ความคิดเห็น ……………………...……………………………………………………………...

ลงชื่อ……………………………………………………………….

วันที่……………………………………………………………...

APPENDIX D

PHYSICAL EXAMINATION FORM

Date………………………………

First name…………………………….. Last name..…………………………………

Age ………………. Years Weight ……………….. kg Height ……….…….. cm

BMI……………...……kg/m^2 Dominate side ☐ left ☐ right

Date of birth ……………………………… Occupation ……………………………

Address ……………………………………………………………………………….

Tel ……………………………………..……………………………………………

Anthropometry

Leg length without shoe during standing (greater trochanter – ground)

Left ………………. cm Right ……………. cm

Knee width (width between medial and lateral femoral condyles)

Left ………………. cm Right ……………. cm

Ankle width (width between medial and lateral malleoli)

Left ………………. cm Right ……………. cm

Foot width (1^{st} – 5^{th} metatarsal head)

Left ………………. cm Right ……………. cm

Foot length (2^{nd} – mid point of heel)

Left ………………. cm Right ……………. cm

Circumference Waist……………….. cm Hip …….……….... cm

Waist to hip ratio……………………………………………..

Knee joint (supine; joint line level)

Left ………………. cm Right ……………. cm

Q angle Left ………… degrees Right ………… degrees

Body composition: Using bioelectrical impedance analysis (BIA)

Percent body fat…………………% Fat mass………….………kg

Lean body weight ……………..…kg Muscle mass………..………kg

Percent visceral fat………………kg Visceral fat……………..….kg

Parameter	Upper extremities		Lower extremities	
	Left	Right	Left	Right
Fat mass (kg)				
Muscle mass (kg)				

Range of Motion

Spine: flexion / extension ………/………/……….

lateral flexion to left / right ………/………/……….

rotation to left / right ………/………/……….

		left	right
Hip:	flexion / extension	………/………/……….	………/………/……….
	abduction / adduction	………/………/……….	………/………/……….
	medial/lateral rotation	………/………/……….	………/………/……….
Knee:	flexion / extension	………/………/……….	………/………/……….
Ankle:	plantar / dorsiflexion	………/………/……….	………/………/……….
	inversion / eversion	………/………/……….	………/………/……….

Special test

SLR test: Left ☐ Negative ☐ Positive ……….. degrees

Rigth ☐ Negative ☐ Positive ……….. degrees

Thomas test: Left ☐ Negative ☐ Positive ……….. degrees

Rigth ☐ Negative ☐ Positive ……….. degrees

Static knee alignment in frontal plan

Left ☐ normal (varus or valgus < 10 degrees) ☐ varus ☐ valgus

Right ☐ normal (varus or valgus < 10 degrees) ☐ varus ☐ valgus

Foot morphology

Left	☐ normal	☐ high arch	☐ flat foot
Right	☐ normal	☐ high arch	☐ flat foot

Navicular Drop Test:

Left	☐	Negative	☐	Positive ……….. mm
Right	☐	Negative	☐	Positive ……….. mm

APPENDIX E
RAW DATA

Table 21 Spatiotemporal parameters in the four groups during walking at self-selected speed (SS), constant velocity (CV), and Froude velocity (FR)

Velocity	Parameter	Normal		Overweight		Obesity I		Obesity II	
		Mean	SEM	Mean	SEM	Mean	SEM	Mean	SEM
SS	Step Width (cm)	9.95	0.70	11.12	0.67	11.49	0.42	12.28	0.47
	Velocity (cm/sec)	116.34	2.21	115.29	1.80	115.96	1.62	113.10	1.80
	Stride Lenght (cm)	127.60	0.91	124.80	1.04	126.63	1.01	124.83	1.27
	Cadence (step/min)	110.63	2.26	110.98	1.39	111.38	1.75	109.70	1.50
	Support Time (% gait cycle)	59.37	0.33	60.07	0.38	60.42	0.22	61.35	0.30
	Non support time (% gait cycle)	40.63	0.33	39.93	0.38	39.58	0.22	38.65	0.30
	Step Length (cm)	62.74	0.52	62.00	0.60	63.42	0.70	62.58	0.92
	Double support time (% gait cycle)	9.68	0.27	10.21	0.39	11.00	0.32	10.82	0.27
CV	Step Width (cm)	10.04	0.53	11.18	0.61	10.73	0.51	11.93	0.53
	Velocity (cm/sec)	119.93	2.41	119.68	1.35	119.45	1.98	118.51	2.43
	Stride Lenght (cm)	127.73	1.51	125.80	0.99	127.45	1.36	125.91	1.64
	Cadence (step/min)	113.89	1.89	114.92	1.17	113.54	1.33	112.43	1.49
	Support Time (% gait cycle)	59.43	0.45	59.68	0.42	60.14	0.33	61.16	0.51
	Non support time (% gait cycle)	40.57	0.45	40.32	0.42	39.86	0.33	38.84	0.51
	Step Length (cm)	63.07	0.63	62.23	0.63	64.28	0.83	62.64	1.24
	Double support time (% gait cycle)	10.75	1.85	12.26	2.05	10.63	0.29	10.65	0.35
FR	Step Width (cm)	9.97	0.58	10.59	0.49	11.28	0.41	11.33	0.38
	Velocity (cm/sec)	114.86	1.88	116.86	1.67	112.41	1.66	109.50	1.91
	Stride Lenght (cm)	126.97	1.38	124.96	1.44	125.73	1.21	123.35	1.25
	Cadence (step/min)	110.02	1.48	112.33	1.57	108.45	1.40	108.24	1.70
	Support Time (% gait cycle)	59.65	0.40	60.79	1.05	60.61	0.25	61.29	0.30
	Non support time (% gait cycle)	40.35	0.40	40.22	0.32	39.39	0.25	38.71	0.30
	Step Length (cm)	62.46	0.90	62.86	0.73	62.04	0.91	60.32	0.76
	Double support time (% gait cycle)	10.30	0.50	10.33	0.44	11.43	0.38	11.34	0.32

Table 22 Spatiotemporal parameters in the four groups during walking at self-selected speed (SS), constant velocity (CV), and Froude velocity (FR) with corresponding statistical findings (Bold indicates significance).

Velocity	Parameter	*P* value in group comparisons and trend							
		Nor vs OW	Nor vs OB I	Nor vs OB II	OW vs OB I	OW vs OB II	OB I vs OB II	F	Trend
SS	Step Width (cm)	0.961	0.388	**0.034***	1.000	0.948	1.000	2.799	**0.046****
	Velocity (cm/sec)	1.000	1.000	1.000	1.000	1.000	1.000	0.602	0.616
	Stride Lenght (cm)	0.403	1.000	0.425	1.000	1.000	1.000	1.684	0.178
	Cadence (step/min)	1.000	1.000	1.000	1.000	1.000	1.000	0.167	0.918
	Support Time (% gait cycle)	0.723	0.120	**0.000***	1.000	**0.030***	0.241	6.900	**0.000****
	Non support time (% gait cycle)	0.723	0.120	**0.000***	1.000	**0.030***	0.241	6.900	**0.000****
	Step Length (cm)	1.000	1.000	1.000	0.943	1.000	1.000	0.692	0.560
	Double support time (% gait cycle)	1.000	**0.026***	0.079	0.482	1.000	1.000	3.611	0.017**
CV	Step Width (cm)	0.864	1.000	0.101	1.000	1.000	0.747	2.105	0.107
	Velocity (cm/sec)	1.000	1.000	1.000	1.000	1.000	1.000	0.088	0.966
	Stride Lenght (cm)	1.000	1.000	1.000	1.000	1.000	1.000	0.522	0.668
	Cadence (step/min)	1.000	1.000	1.000	1.000	1.000	1.000	0.472	0.703
	Support Time (% gait cycle)	1.000	1.000	**0.039***	1.000	0.112	0.607	3.069	**0.033****
	Non support time (% gait cycle)	1.000	1.000	**0.039***	1.000	0.112	0.607	3.069	**0.033****
	Step Length (cm)	1.000	1.000	1.000	0.601	1.000	1.000	1.035	0.382
	Double support time (% gait cycle)	1.000	1.000	1.000	1.000	1.000	1.000	0.324	0.808
FR	Step Width (cm)	1.000	0.313	0.257	1.000	1.000	1.000	1.894	0.138
	Velocity (cm/sec)	1.000	1.000	0.221	0.493	**0.028***	1.000	3.170	**0.029****
	Stride Lenght (cm)	1.000	1.000	0.342	1.000	1.000	1.000	1.307	0.278
	Cadence (step/min)	1.000	1.000	1.000	0.471	0.385	1.000	1.506	0.220
	Support Time (% gait cycle)	1.000	1.000	0.334	1.000	1.000	1.000	1.327	0.272
	Non support time (% gait cycle)	1.000	0.229	**0.003***	0.425	**0.008***	0.847	5.676	**0.001****
	Step Length (cm)	1.000	1.000	0.436	1.000	0.205	0.891	1.805	0.153
	Double support time (% gait cycle)	1.000	0.342	0.481	0.384	0.538	1.000	2.233	0.091

* significantly different between BMI group Bonferroni

** significantly different between BMI group ANOVA

Table 23 Kinematic and kinetic parameters in the four groups during walking at self-selected speed (SS), constant velocity (CV), and Froude velocity (FR) .

Velocity	Parameter	Normal		Overweight		Obesity I		Obesity II	
		Mean	SEM	Mean	SEM	Mean	SEM	Mean	SEM
SS	Peak knee flexion angle (°)	8.549	0.865	12.386	1.031	11.180	1.240	9.424	0.950
	Peak VGRF	1.180	0.034	1.135	0.022	1.105	0.027	1.117	0.007
	Peak external knee adduction moment (Nm/w)	0.334	0.031	0.265	0.019	0.222	0.022	0.261	0.021
	Knee adduction angular impulse (Nms/w)	0.112	0.015	0.097	0.009	0.083	0.009	0.087	0.007
	Peak external knee adduction moment (Nm)	15.645	1.738	19.501	2.020	22.886	2.377	28.089	2.905
	Knee adduction angular impulse (Nms)	5.743	0.800	5.235	0.516	5.444	0.600	6.434	0.517
CV	Peak knee flexion angle (°)	8.965	1.044	11.944	0.948	11.358	1.056	10.599	0.906
	Peak VGRF	1.180	0.035	1.532	0.023	1.120	0.027	1.139	0.009
	Peak external knee adduction moment (Nm/w)	0.352	0.033	0.271	0.024	0.214	0.024	0.255	0.024
	Knee adduction angular impulse (Nms/w)	0.120	0.015	0.086	0.010	0.074	0.010	0.077	0.008
	Peak external knee adduction moment (Nm)	14.584	1.329	16.494	1.076	18.205	1.541	21.821	2.232
	Knee adduction angular impulse (Nms)	5.733	0.823	5.104	0.557	5.342	0.658	6.288	0.608
FR	Peak knee flexion angle (°)	8.001	0.950	11.165	0.991	9.272	0.958	9.768	0.916
	Peak VGRF	1.160	0.036	1.124	0.030	1.091	0.027	1.106	0.008
	Peak external knee adduction moment (Nm/w)	0.325	0.033	0.252	0.023	0.225	0.026	0.258	0.023
	Knee adduction angular impulse (Nms/w)	0.115	0.015	0.089	0.009	0.076	0.011	0.077	0.007
	Peak external knee adduction moment (Nm)	17.664	2.128	20.722	2.344	20.835	2.301	25.306	2.223
	Knee adduction angular impulse (Nms)	5.952	0.829	5.289	0.521	5.147	0.729	6.645	0.573

Table 24 Kinematic and kinetic parameters in the four groups during walking at self-selected speed (SS), constant velocity (CV), and Froude velocity (FR) with corresponding statistical findings (Bold indicates significance).

Velocity	Parameter	*P* value in group comparisons and trend							
		Nor vs OW	Nor vs OB I	Nor vs OB II	OW vs OB I	OW vs OB II	OB I vs OB II	F	Trend
SS	Peak knee flexion angle (°)	0.062	0.450	1.000	1.000	0.274	1.000	2.801	0.050
	Peak VGRF	0.599	0.659	1.000	1.000	1.000	1.000	1.215	0.310
	Peak external knee adduction moment (Nm/ w)	0.453	**0.021***	0.765	1.000	1.000	0.880	3.056	**0.033****
	Knee adduction angular impulse (Nms/w)	0.607	**0.015***	0.084	0.881	1.000	1.000	3.681	**0.016****
	Peak external knee adduction moment (Nm)	1.000	1.000	**0.027***	1.000	**0.006***	**0.002***	5.922	**0.001****
	Knee adduction angular impulse (Nms)	1.000	0.854	1.000	1.000	1.000	0.156	1.835	0.148
CV	Peak knee flexion angle (°)	0.220	0.550	1.000	1.000	1.000	1.000	1.698	0.175
	Peak VGRF	1.000	1.000	1.000	0.956	0.971	1.000	0.994	0.400
	Peak external knee adduction moment (Nm/ w)	0.238	**0.005***	0.317	1.000	1.000	0.806	4.098	**0.009****
	Knee adduction angular impulse (Nms/w)	0.685	**0.024***	0.111	1.000	1.000	1.000	3.345	**0.023****
	Peak external knee adduction moment (Nm)	1.000	0.987	**0.019***	1.000	**0.022***	**0.004***	4.916	**0.004****
	Knee adduction angular impulse (Nms)	1.000	0.873	1.000	1.000	1.000	0.258	1.567	0.204
FR	Peak knee flexion angle (°)	0.130	1.000	1.000	0.988	1.000	1.000	1.881	0.140
	Peak VGRF	1.000	0.489	1.000	1.000	1.000	1.000	1.157	0.332
	Peak external knee adduction moment (Nm/ w)	0.326	**0.025***	0.466	1.000	1.000	1.000	2.610	**0.048****
	Knee adduction angular impulse (Nms/w)	0.572	**0.048***	0.104	1.000	1.000	1.000	2.722	**0.045****
	Peak external knee adduction moment (Nm)	1.000	1.000	**0.013***	1.000	**0.019***	**0.026***	4.056	**0.001****
	Knee adduction angular impulse (Nms)	1.000	1.000	1.000	1.000	0.956	0.724	1.038	0.381

*   significantly different between BMI group Bonferroni
**   significantly different between BMI group ANOVA

Table 25 Kinetic parameters in the three conditions during walking at self-selected speed with corresponding statistical findings

BMI level	Parameter	Barefoot		Shoe only		Shoe with insole		p-values		
		Mean	SEM	Mean	SEM	Mean	SEM	BF vs SH	BF vs SI	SH vs. SI
Normal	Peak VGRF	1.18	0.03	1.16	0.04	1.15	0.04	0.192	0.299	0.248
	Peak external knee adduction moment (Nm/ w)	0.33	0.03	0.3	0.03	0.3	0.03	0.120	0.118	1.000
	Knee adduction angular impulse (Nms/w)	0.11	0.01	0.11	0.02	0.11	0.01	1.000	1.000	1.000
	Peak external knee adduction moment (Nm)	15.65	1.74	13.79	1.29	10.74	1.31	**0.046***	**0.029***	**0.037***
Overweight	Peak VGRF	1.14	0.02	1.12	0.03	1.1	0.03	0.256	0.400	1.000
	Peak external knee adduction moment (Nm/ w)	0.27	0.02	0.25	0.03	0.24	0.02	0.095	**0.043***	0.086
	Knee adduction angular impulse (Nms/w)	0.1	0.01	0.09	0.01	0.09	0.01	0.212	0.247	0.465
	Peak external knee adduction moment (Nm)	19.5	2.02	15.92	1.44	13	1.46	**0.050***	**0.029***	**0.044***
Obesity I	Peak VGRF	1.11	0.03	1.09	0.03	1.09	0.02	0.385	0.754	1.000
	Peak external knee adduction moment (Nm/ w)	0.22	0.02	0.21	0.03	0.2	0.03	1.000	0.732	1.000
	Knee adduction angular impulse (Nms/w)	0.08	0.01	0.07	0.01	0.07	0.01	0.231	0.763	0.467
	Peak external knee adduction moment (Nm)	22.89	2.38	18.26	1.69	13.85	1.9	**0.030***	**0.043***	**0.023***
Obesity II	Peak VGRF	1.12	0.01	1.11	0.01	1.11	0.01	0.053	0.630	1.000
	Peak external knee adduction moment (Nm/ w)	0.26	0.02	0.25	0.02	0.24	0.02	1.000	0.126	0.081
	Knee adduction angular impulse (Nms/w)	0.08	0.01	0.07	0.01	0.07	0.01	0.365	0.295	1.000
	Peak external knee adduction moment (Nm)	28.09	2.91	21.88	2.36	21.83	1.97	**0.038***	**0.029***	1.000

* significantly different between BMI group Bonferroni

Table 26 Kinetic parameters during walking at constant velocity (1.24 m/s) in the three conditions with corresponding statistical findings.

BMI level	Parameter	Barefoot		Shoe only		Shoe with insole		p-values		
		Mean	SEM	Mean	SEM	Mean	SEM	BF vs SH	BF vs SI	SH vs. SI
Normal	Peak VGRF	1.18	0.04	1.16	0.03	1.16	0.03	0.840	1.000	1.000
	Peak external knee adduction moment (Nm/ w)	0.35	0.03	0.33	0.04	0.28	0.02	0.538	**0.025***	**0.040***
	Knee adduction angular impulse (Nms/w)	0.12	0.02	0.11	0.02	0.1	0.01	0.120	0.258	0.845
	Peak external knee adduction moment (Nm)	14.58	1.33	12.74	1.16	10.01	1.14	**0.027***	**0.024***	**0.030***
Overweight	Peak VGRF	1.53	0.2	1.34	0.03	1.13	0.4	0.099	0.099	0.098
	Peak external knee adduction moment (Nm/ w)	0.27	0.02	0.26	0.03	0.25	0.03	0.528	0.213	1.000
	Knee adduction angular impulse (Nms/w)	0.09	0.01	0.09	0.01	0.08	0.01	1.000	0.310	1.000
	Peak external knee adduction moment (Nm)	16.49	1.08	14.09	1.45	13.01	1.43	**0.044***	**0.039***	**0.043***
Obesity I	Peak VGRF	1.12	0.03	1.09	0.03	1.09	0.01	0.122	0.807	1.000
	Peak external knee adduction moment (Nm/ w)	0.21	0.02	0.2	0.02	0.2	0.03	0.373	0.648	1.000
	Knee adduction angular impulse (Nms/w)	0.07	0.01	0.06	0.01	0.06	0.01	0.829	0.583	1.000
	Peak external knee adduction moment (Nm)	18.2	1.54	17.37	1.78	15.15	1.79	0.923	**0.043***	**0.046***
Obesity II	Peak VGRF	1.14	0.01	1.12	0.01	1.11	0.01	0.135	0.997	0.715
	Peak external knee adduction moment (Nm/ w)	0.26	0.02	0.25	0.02	0.24	0.03	1.000	0.914	1.000
	Knee adduction angular impulse (Nms/w)	0.08	0.01	0.07	0.01	0.07	0.01	0.098	0.082	1.000
	Peak external knee adduction moment (Nm)	21.82	2.23	20.94	2.16	18.66	2.24	0.709	0.093	0.930

* significantly different between BMI group Bonferroni

Table 27 Kinetic parameters during walking at Froude velocity in the three conditions with corresponding statistical findings.

BMI level	Parameter	Barefoot		Shoe only		Shoe with insole		p-values		
		Mean	SEM	Mean	SEM	Mean	SEM	BF vs SH	BF vs SI	SH vs. SI
Normal	Peak VGRF	1.16	0.04	1.14	0.03	1.15	0.03	0.266	1.000	1.000
	Peak external knee adduction moment (Nm/ w)	0.33	0.03	0.32	0.04	0.30	0.03	0.725	0.230	**0.027***
	Knee adduction angular impulse (Nms/w)	0.12	0.02	0.12	0.02	0.11	0.01	0.121	0.547	0.060
	Peak external knee adduction moment (Nm)	17.66	2.13	13.24	1.21	12.73	1.21	**0.036***	**0.025***	1.000
Overweight	Peak VGRF	1.12	0.03	1.15	1.40	1.18	0.67	0.999	0.988	1.000
	Peak external knee adduction moment (Nm/ w)	0.25	0.02	0.27	0.03	0.25	0.02	0.297	1.000	0.074
	Knee adduction angular impulse (Nms/w)	0.09	0.01	0.10	0.01	0.09	0.01	0.106	1.000	0.070
	Peak external knee adduction moment (Nm)	20.72	2.34	15.26	1.49	15.18	1.40	**0.018***	**0.014***	1.000
Obesity I	Peak VGRF	1.09	0.03	1.08	0.03	1.09	0.02	0.721	1.000	0.494
	Peak external knee adduction moment (Nm/ w)	0.22	0.03	0.22	0.03	0.22	0.03	1.000	1.000	1.000
	Knee adduction angular impulse (Nms/w)	0.08	0.01	0.08	0.01	0.07	0.01	1.000	0.807	0.518
	Peak external knee adduction moment (Nm)	20.83	2.30	17.29	1.41	13.74	1.30	**0.049***	**0.023***	0.134
Obesity II	Peak VGRF	1.11	0.01	1.11	0.01	1.09	0.01	1.000	0.530	0.194
	Peak external knee adduction moment (Nm/ w)	0.26	0.02	0.26	0.02	0.26	0.03	1.000	1.000	1.000
	Knee adduction angular impulse (Nms/w)	0.08	0.01	0.08	0.01	0.08	0.01	0.522	0.451	1.000
	Peak external knee adduction moment (Nm)	25.31	2.22	23.97	2.21	23.42	2.29	0.315	1.000	0.917

*	significantly different between BMI group Bonferroni