STUDY OF NET JOINT FORCES AT TIBIOFEMORAL JOINT FOR OBESE AND NORMAL SUBJECTS

WAN MOHD RADZI BIN RUSLI

FACULTY OF ENGINEERING

UNIVERSITY OF MALAYA

KUALA LUMPUR

2012

STUDY OF NET JOINT FORCES AT TIBIOFEMORAL JOINT FOR OBESE AND NORMAL SUBJECTS

WAN MOHD RADZI BIN RUSLI

RESEARCH REPORT SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENT FOR THE DEGREE OF MASTER OF ENGINEERING (BIOMEDICAL)

FACULTY OF ENGINEERING

UNIVERSITY OF MALAYA

KUALA LUMPUR

2012

UNIVERSITI MALAYA

ORIGINAL LITERARY WORK DECLARATION

Name of Candidate: **Wan Mohd Radzi Bin Rusli**

Registration/Matric No: **KGL 090005**

Name of Degree: **Master of Engineering (Biomedical)**

Title of Project Paper/ Research Report/ Dissertation/ Thesis ("this Work"):

STUDY OF NET JOINT FORCES AT TIBIOFEMORAL JOINT FOR OBESE AND NORMAL SUBJECTS

Field of study: **Biomechanics**

I do solemnly and sincerely declare that:

(1) I am the sole author/writer of this Work;
(2) This Work is original;
(3) Any use of any work in which copyright exists was done by way of fair dealing and for permitted purposes and any excerpt or extract from, or reference to or reproduction of any copyright work has been disclosed expressly and sufficiently and the title of the work and its authorship have been acknowledge in this Work;
(4) I do not have any actual knowledge nor do I ought reasonably to know that the making of this Work constitutes an infringement of any copyright work;
(5) I hereby assign all and every rights in the copyright to this Work to the University of Malaya ("UM"), who henceforth shall be owner of the copyright in this Work and that any reproduction or use in any form or by any means whatsoever is prohibited without the written consent of UM having been first had and obtained;
(6) I am fully aware that if in the course of making this Work I have infringed any copyright whether intentionally or otherwise, I may be subject to legal action or any other action as may be determined by UM.

Candidate's Signature Date

Subscribed and solemnly declared before,

Witness's Signature Date

Name:

Designation:

ABSTRACT

Walking and stair ascending are categorized as the activity of daily living (ADL). People who cannot perform these activities will constantly need assistants in performing their basic routine. Osteoarthritis is a well known joint disease that reduces the efficiency of the particular joint in performing basic movements. By using motion capture system and 3-Dimensional modeling of human lower limb, the net joint force at the sagittal plane were obtained and can be used in explaining the onset and progression of osteoarthritis at tibiofemoral joint. In this present study, 18 subjects with no history of joint diseases were recruited and 12 of them had a normal body mass index (BMI) and the remaining 6 experiencing obesity. By using the equations of motion, the net joint force at sagittal plane was obtained. The statistical analysis was completed using SPSS (Statistical Statistics for the Social Sciences). Present results indicate that there was a significant different at the net joint force between walking and stair ascending activities, where both normal and obese groups produce 2480.76N and 3562.38N, during stair ascending compare to 1022.5N and 1603.07N for both groups during walking activity. It was also noted that when comparing the normal and obese group, obese group produce higher net joint force at the tibiofemoral joint, where there was 56.7 percentage of difference during walking activity. This is the same with the obese group, 43.6 percentage of difference. In a nutshell, the net joint force is proportional with the body mass index (BMI). As the body mass index (BMI) increase, the net joint force also increase. If we consider that there are no physiological changes at the tibiofemoral joint, then obesity is the biomechanical factors that contribute to the onset and progression of osteoarthritis at tibiofemoral joint.

ABSTRACT

Berjalan dan naik tangga dikategorikan sebagai aktiviti kehidupan seharian. Mereka yang tidak boleh melaksanakan aktiviti-aktiviti ini akan sentiasa memerlukan bantuan dalam melaksanakan rutin asas mereka. Osteoartritis adalah penyakit yang dikenali ramai, dimana ia mampu mengurangkan kecekapan sendi tertentu dalam melaksanakan pergerakan asas. Dengan menggunakan sistem penangkapan gerakan dan pemodelan 3-Dimensi otot bawah tubuh badan manusia, daya bersih sendi pada satah sagittal telah diperolehi dan boleh digunakan dalam menerangkan permulaan dan perkembangan osteoartritis pada sendi tibiofemoral. Di dalam kajian ini, 18 sukarelawan yang tidak mempunyai sejarah penyakit sendi telah diambil dan 12 daripada mereka mempunyai indeks jisim badan yang normal dan baki 6 mengalami obesiti. Dengan menggunakan persamaan gerakan, daya bersih sendi pada satah sagittal telah diperolehi. Analisis statistik telah siap dengan menggunakan SPSS (Statistik untuk Sains Sosial). Keputusan sekarang menunjukkan bahawa terdapat perbezaan yang signifikan pada daya bersih sendi antara berjalan dan aktiviti naik tangga, di mana kedua-dua kumpulan yang normal dan obes menghasilkan 2480.76N and 3562.38N, semasa tangga naik berbanding kepada 1022.5N and 1603.07N bagi kedua-dua kumpulan semasa aktiviti berjalan kaki. Ia juga mencatat bahawa apabila membandingkan kumpulan yang normal dan obes, kumpulan obes menghasilkan daya bersih sendi yang lebih tinggi di sendi tibiofemoral, di mana terdapat 56.8 peratusan perbezaan semasa aktiviti berjalan. Ini adalah sama dengan kumpulan obes, 43.6 peratus perbezaan. Secara ringkasnya, daya bersih sendi adalah berkadar dengan indeks jisim badan (BMI). Dengan peningkatan indeks jisim tubuh (BMI), daya bersih sendi juga meningkat. Jika kita menganggap bahawa terdapat sebarang perubahan fisiologi di sendi tibiofemoral, maka obesiti adalah biomekanikal faktor yang menyumbang kepada permulaan dan perkembangan osteoartritis pada sendi tibiofemoral.

ACKNOWLEDGEMENTS

First and foremost, I wish to extend my deepest gratitude and profound appreciation to Associate Professor Dr. Noor Azuan Abu Osman from the Department of Biomedical Engineering, Faculty of Engineering, University of Malaya for his invaluable guidance, support and encouragement throughout this dissertation. This work would not have been completed if not for his advice and assistance.

To my friends and colleagues who have helped me directly or indirectly, namely Hazli and Ambhigavathi, thank you for your help, ideas and positive inputs.

Further thanks to Universiti Malaysia Perlis (UniMAP) for awarding me the scholarship and study leave to undertake this work.

Last but not least, special thanks to my family especially my parents for their support and love. I would not have completed this dissertation without the understanding from all. Thank you very much.

CONTENTS

LIST OF FIGURES

LIST OF TABLES

1.0 INTRODUCTION

1.1 Problem statement

Malaysia nowadays had been recognized as a developing country due to the rapid phase of the industrialization and urbanization. According to some ministries; statistically, as the citizen of malaysia achieve affluence, they consume a lot of energy, fat and sugar, and this is supported by the rise of the food importation bills (Ismail, 2002). For several years, the fast food industries have been increased steadily in Malaysia (Ismail, 2002). Obesity had been increased due to the fact that the population had changed their lifestyle where unhealthy diet had been practiced. The severe negative effect will attacks those who inherited the metabolic predisposition to fatness (Ismail, 2002). As everybody knows, those who were obesed will tend to have joint diseases such as osteoarthritis (OA).

If ones having OA diseases, they might have the difficulity to perform the activity of daily living (ADL). ADL is the basic activity that we perform in our daily life such as sit to stand, walking, squatting and stair climbing . Those who have obesity will put a large number of load particularly at the lower limb such as knee during walking and stair ascending. There are a lot of research that had been carried out regarding the OA. Many researcher had study what are the major factors that initiate and progressed the OA diseases at the knee joint.

Biomechanics can be used to define the factors that contribute to the initiation and progression of the OA diseasesat the tibiofemoral joint. The difficulty to find the relation between the mechanical factors and the initiation and progression of the OA diseases had inspired us to design the present study.

1.2 Objectives of the study

The purpose of this study is to investigate the net joint force at the tibiofemoral joint for obese and normal subjects during walking and stair climbing. Since the ground reaction force does not give us the value of force at joint; so,this study will collect the kinetics and kinematics data using motion analysis system as well as the force plate system and use inverse dynamic analysis to calculate the net joint force at tibiofemoral joint. The comparison of the net joint force between obese and normal subjects will be conducted. These values of net joint forces are very important as we move to the second objective which is to see the clinical relevance in the study of biomechanics of the tibiofemoral joint in relation to the initiation and progression of OA at the tibiofemoral joint. This information will add to the current knowledge of etiology of knee OA.

1.3 Hypothesis

Obesity is one of the biomechanical factor that can lead to an abnormal loading at the tibiofemoral joint (Anderson et al, 1988). The excessive weight will contribute to the abnormal loading at the tibiofemoral joint while doing daily activities. The weight-bearing joint; in this case is the tibiofemoral joint, cannot withstand the abnormality of load at this particular joint.

From force plate system , only the external forces such as ground reaction force can be obtained and this basic information cannot explained the phenomenon happened at the tibiofemoral joint.The exact internal load that tibiofemoral joint had to hold in any activities cannot be obtained using this system. However by using inverse dynamic equation, the kinetic and kinematic informations obtained from the integrated system of motion analysis and force plate can be usefull in finding the net joint forces at the tibiofemoral joint.

As the obesity can lead to the initiation and progression of OA and for the purpose of the study, I hypothesized that the net joint forces at the tibiofemoral joint is greater than the body weight of the person while doing activities that involves flexion and extension at the tibiofemoral joint. The association of this biomechanical factor toward the initiation and progression of the OA at the tibiofemoral joint is at a linear trend due to the excessive weight which finally increase the load that tibiofemoral joint need to withstand.

1.4 Scope of the study

The first objective in this study is to find the net joint forces at the tibiofemoral joint. This can be achieved by using the motion analysis system where it had been synchronize with the force plate system. This intergrated system will collect the kinetics and kinematics data. Among the data that will be getting by using this system are ground reaction force, diaplacement, velocity, acceleration and centre of pressure (CoP). This data will be feed into the inverse dynamic equation in order to find the net ionternal forces and moments at the tiobiofemoral joint. These procedures will be applied to both obese and normal subjects while walking and stair climbing to facilitate the comparison purpose. The result will be used to support the clinical relevance regarding the association of biomechanical factors toward initiation and progression of OA at tibiofemoral joint.

1.5 Significance of the study

There were many biomechanical factors that can be associated with the OA at tibiofemoral joint. This will be further explained in the next chapter. Current situation that had motivated me to conduct this study are:

1. There were lack of data that had show how obese people were different with the normal people in term of kinetics and kinematics during stair climbing and walking.
2. No study had used the data obtained from the experiment and inverse dynamic calculation in explaining the association of obesity with initiation or onset and progression of tibiofemoral joint OA.

All this were very important because in can help people in term of understanding the cause of their tibiofemoral OA disease.

1.6 Outline of the report

There will be seven chapter throughout this dissertation. In chapter 2, a comprehensive review on the biomechanics of tibiofemoral joint and the relevance of using biomechanics in studying the onset and progression of knee OA is presented. Chapter 3 deals with the method implemented in this study and materials utilized in the investigation. Reasons for each technique and material used will also be stated. Results obtained from the present study will be presented graphically and statistically in Chapter 4. Chapter 5 discusses the overall results in greater depth and finally Chapter 6 concludes the dissertation with suggestion of further work.

2.0 LITERATURE REVIEW

In this chapter, a comprehensive review had been done in order to gain knowledge about this present study. Here, the review start from the biomechanics of tibiofemoral joint that discussed about the anatomy and kinetics of the joint. After that, we proceed with the rational of biomechanical factor that cause the initiation and progression of osteoarthritis (OA).

2.1 Biomechanics of the tibiofemoral joint

In this section the relationships and influences of the anatomy and design of the human knee joint, to kinematics and kinetics are presented.

2.1.1 Design of the joint

The components of the tibiofemoral knee joint can be divided into the tibio-femoral articulation, cruciates and collateral ligaments, menisci and capsular structures. In the tibiofemoral joint the articulation is between the distal end of the femur and the proximal end of the tibia. The medial femoral condyle is larger and more symmetrical than the lateral femoral condyle. The long axis of the lateral condyle is slightly longer than the long axis of the medial condyle and is placed in a more sagittal plane. Also the width of the lateral femoral condyle is slightly larger than the medial femoral condyle at the centre of intercondylar notch. The contact area of the medial plateau is said to be 50% larger than that of the lateral tibial plateau and the articular cartilage of the medial tibial plateau is thicker than that of the lateral tibial plateau. This is relevant because of the larger loads in the medial compartment (Kettlekemp DB 1972). The lack of conformity between the femoral and tibial articulation is augmented by the presence of menisci, which serve as a shock absorber and cushions the load sustained during normal activities. The menisci rest on the articular surface supported by the subchondral plate. Each meniscus covers approximately the peripheral two-thirds of the articular surface of the tibia. The medial menisci are semilunar in shape and the lateral menisci nearly circular. The lateral menisci transmit 75% and the medial meniscus 50% of the load (Walker PS 1975).

The anterior and posterior cruciate ligaments are the prime stabilisers of the knee in resisting anterior and posterior translation, respectively (Noyes FR 1985). The collateral ligaments, menisci and the capsule provide additional restraint to the anterior and posterior movement of the knee, as well as to rotation. The anatomy of the cruciates and the collateral ligaments has been well described in the literature (Arnoczky SP 1983, Jakob and Staubli HU 1992). The articular surfaces hold the two bones apart and resist interpenetration by transmitting compressive stresses across their surfaces, whereas the ligaments hold the two bones together and resist distraction by transmitting tensile stresses along the line of their fibres. The ligaments often act together in limiting motion, sometimes creating primary and secondary ligamentous restraints. These are well described in the literature (Butler DL et al, 1977, Daniel DM et al, 1990).

2.1.2 Tibiofemoral Joint Loads during Activities of Daily Living (ADL)

Kinetics of the tibiofemoral joint looks at the forces and energy that is involved in either maintaining static equilibrium or initiating dynamic activity for the joint. In the knee joint the loading is as much as three times body weight during walking (Morrison JB 1970). Calculated forces in the medial and lateral compartments indicate relatively more loading in the medial compartment (Schipplein OD et al 1991). The design of the tibiofemoral joint is such that the relatively more concave medial plateau provides more congruency for contact with the femoral condyle, and the lateral plateau being convex provides some freedom for condylar mobility over the plateau. Determination of the tibiofemoral contact area is significant in evaluating the weight bearing capacity of the joint. For example, the average area of contact on the medial plateau is 1.6 times greater than the area on the lateral plateau (Kettlekamp 1972). Therefore it is easy to deduce that even though forces in the medial compartment may be larger than in the lateral, the contact stresses may not be different in the two compartments if there is more distribution of forces in the medial compartment due the increased area of contact. These forces are cushioned and accommodated largely by the meniscus and articular cartilage. The distribution of the forces over the area of contact (mainly involving cartilage in the healthy joint) determines the stresses that result. Some of the stresses calculated in previous studies are shown (Table 2.1). Contact stresses are also affected by joint malalignment. A varus malalignment of 30 degrees at the proximal third of the tibia was found to increase medial compartment contact pressures by 101% and decrease the lateral compartment contact pressure by 89% (McKellop et al 1991). Like any multi-support weight bearing structure, the location of the center of gravity will determine the distribution of the forces. With malalignment, and in this case varus deformity, the center of gravity shifts more medially, and so does the center of maximal joint pressure, with even a likelihood of separation of the lateral tibiofemoral joint and "condylar lift-off" (Noyes FR 1992) during maximum weightbearing in walking.

Table 2.1: Some previously derived contact stresses measured in the knee are shown here.

Author	**Year**	**Specimen**	**Joint**	**Joint studied**	**Loading**	**contact area**	**mean contact stress**	**peak contact stress**
Fukubayashi et al	1980	human cadaver	tibiofemoral	tibia	1500N	14.1cm2		
Brown and Shaw	1984	human cadaver	tibiofemoral	femur	3000N		2.6MPa	2.6MPa 8MPa

It was found that varus knee malalignment was a contributory cause to OA from the effects of obesity (Sharma L et al 2000). Increased dynamic loads on the medial compartment as a result of varus malalignment in OA aggravates the problem of excessive loading, and presents the question of whether the malalignment precedes or follows the onset of the disease. In any case many studies on OA wear patterns indicate a higher incidence of degenerative changes in the medial compartment compared to the lateral (Weidow J et al 2002). Another study on Chinese subjects relating the activity of squatting with tibiofemoral OA (Zhang Y et al 2004) also found a significantly higher incidence of medial compartment OA compared with lateral. Incidentally, Chinese subjects were found to have significantly more varus mal- alignment of the lower extremity compared with westerners (Tang WM et al 2000) and also for studies on Asians specifically (Zhang et al 2001) it was estimated that the prevalence of radiographic and symptomatic knee OA in a population–based sample of elderly subjects in China were higher than that reported in the Framingham OA study which looked at a primarily Caucasian population.

2.2 The rationale for a biomechanics approach to investigating the causes and risks of knee osteoarthritis

The primary cause of osteoarthritis (OA) remains vague. Some of the earlier hypotheses and experimental work on the mechanisms involved in the initiation of osteoarthritis come from Radin EL and co-workers, where the fundamental emphasis has been that the cause of OA is due to an "*imbalance between the mechanical stress and the integrity of the tissues"* in absorbing the loads (Radin EL 1987). Bone proliferation in the subchondral area in the cortical envelope of bone just below cartilage (eburnation) or at the margins of the bone (osteophytes), and radiographic joint space narrowing are hallmarks of OA (Kellgren JH and Lawrence JS 1957, Felson DT and Conaghan PG 2004). The clinically useful definition of OA is "the symptomatic loss of significant articular cartilage in a habitual load bearing area of a joint associated with subchondral sclerosis and osteophyte formation" (Radin EL 1995). OA may be a result of tissue injury, damage, or degeneration in relation to mechanical factors, as evidenced by clinical and animal studies of alterations in the mechanical environment of the joint caused by trauma, joint instability, disuse, or obesity (Guilak F et al 2004). The term posttraumatic arthritis (Guilak F et al 2004, Buckwalter JA et al 2004) has also been used to distinguish OA as a disease commonly initiated by damage or injury in the joint.

The evidence that injury or damage produces a significantly higher likelihood for the development of OA is confirmed in prospective studies, where it was reported that this likelihood could be enhanced by as much as 7 times (Wilder FV et al 2002). However there are hardly any studies that show the relative change, if any, in the rapidity of the onset of OA as a result of injury or the type of injury. A prospective study of over 1000 subjects aged about 22 years old and followed- up for over 30 years show that for subjects entering the study without joint injury, the incidence of OA after 30 years was 6%, while for those entering the study with a prior knee injury the incidence increased to 14% by the same time, with clear OA signs manifesting itself earlier for many (Gelber AC et al 2000). However, not much more was reported on the type of injury sustained, or the length of time before OA signs were observed.

In another report it was found that osteonecrosis in the medial tibial plateau presents a significantly high risk for the development of OA in the tibia; this occurred rapidly within a period of one year for many of the subjects studied (Satku K et al 2003). Like its cause, the exact initial onset of OA is difficult to determine and remains vague. Spotting any initial signs is limited to current clinical features mainly of radiographic changes that look for osteophytes and/or joint space narrowing (Satku K et al 2003). From a study of patients with unicompartmental OA, it was shown during intraoperative treatment that the apparently unaffected cartilage was mechanically inferior to normal cartilage, even although clinically, radiologically and morphologically it appeared to be sound (Obeid EM et al 1994). Even in the event that direct visualization of the joint is possible, for example via arthroscopy, there is also no guarantee that the initiation of degenerative changes may be ascertained. Animal studies have shown that macroscopic assessments are insensitive to articular cartilage softening, suggesting that arthroscopic assessments of cartilage status might also perform poorly (Oakley SP et al 2004). This is also indicative that the actual start of the degenerative process, and the time when symptoms first appear, are likely to be different. In a recent review, the pathogenesis and epidemiology of OA were discussed as risk factors for OA and the *symptoms* of OA (Felson DT 2004), the initiation of OA being distinguished from the initiation of symptomatic OA. Symptoms usually present themselves at a later stage after the onset of OA. This is largely due to the reliance on pain as the initial symptom. Cartilage with no nerve endings is usually wornout and it is the bone’s pain fibres, the synovium or other structures that need to be involved before pain is felt. Evidence of this has been reported in previous studies (Conaghan PG et al 2004, Hill CL et al 2001, Felson DT et al 2001). Factors that are known to result in symptoms or the progression of OA are obtained from clinical observation of patients. These are, for example, bone marrow edema, synovitis, and joint effusion (Felson DT 2004). Bone marrow edema lesions in the subarticular bone in patients with knee osteoarthritis identify knees at high risk for radiographic progression (Felson DT et al 2003), and its relation to progression is explained in part by its association with limb alignment.

The initiation of OA and the progression of OA are two processes that may require two different approaches for understanding. In this section the current theories on how OA may be initiated and progressed are described in relation to the different injury and damage modes. The development of OA is discussed as being initiated or propagated via one of several ways. These are based on theories on how cartilage damage may be initiated and propagated from:

I. Subchondral bone changes (Radin EL and Rose RM 1986).

II. Microdamage in the subchondral plate and calcified cartilage that leads to enchondral ossification (Burr DB and Radin EL 2003).

III. Altered mechanics that cause a shift in contact to cartilage that is not conditioned to high loads (Andriacchi TP et al 2004). The common feature is that OA is a consequence of injurious activities acting on a vulnerable joint (Felson DT 2004) and leading to cartilage degeneration as the major clinical sign (Radin EL 1995).

2.3 Modeling of human motion

In this section, we will see how mathematical modeling of human lower limb being developed. Inverse dynamics equation also will be discussed in this chapter. The combination of both techniques, we will have a comprehensive tool that can find the net joint force at the tibiofemoral joint.

2.3.1 Mathematical Modeling (Vaughan et al, 1992)

The mathematical modeling used to model our lower limb is taken from Vaughan and his colleagues (1992). They used the 3D mathematical modeling to model our lower limb. Body segment parameters, linear kinematics and centre of gravity are the main components in their model. Each component will be discussed in details throughout this topic.

2.3.1.1 Body Segment Parameters

When we talk about body segment parameters, it refers to our segment's mass, location of the centre of gravity of each segment and moments of inertia of the segments about three orthogonal axes that pass through the segment centre of gravity. Since everyone having different body segment parameters, the most challenging aspects is how we want to come out with mathematical equations that can compute everyone's body segment parameters.

Anthropometry is the solutions for this problem. Anthropometric measurement is a technique that require less time, inexpensive and safe as well as reasonable accurate. By using anthropometry, everyone's body segment parameters can be summarized in mathematical equations. Figure 2.1 shows how the anthropometric parameters at the lower limb being measured. In total, for lower limb we should have twenty measurements where 18 measurements should come from both leg and the remaining two come from the total body mass and superior iliac spines (ASIS).

The mathematical equations that represent the body segment parameters were derived based on the linear regression equations. Here, we let centre of gravity of each segments later and focused on the mass and moment of inertia for each segments of our lower limb. Multiple linear regressions were performed in order to come out with the mathematical equations that can predict the mass of every segment for the lower limb. Equation 2.1 until 2.2 can be used to represent mass of our thigh, shank and foot.

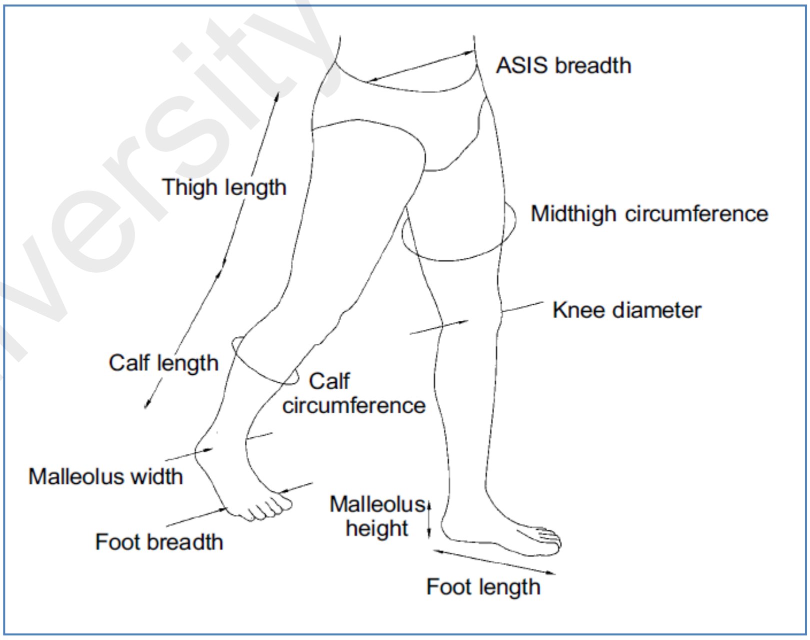


Figure 2.1: Anthropometry of human lower limb.

Mass of calf = (0.0226) (Total body mass)

+ (31.33) (Calf length)(Calf circumference) ^{2}

+ (0.016) (2.1)

Mass of foot = (0.0083) (Total body mass)

+ (254.5) (Malleolus width)(Malleolus height)

(Foot length) + (-0.065) (2.2)

By definition, moment of inertia is a measure of a body's resistance to angular motion. To predict the moment of inertia for each segment, regression methods had been implemented. Each segment should have three equations for moment of inertia where the first equation is rotate about flexion/extension axis, Abduction/Adduction axis and Internal/External rotation axis.

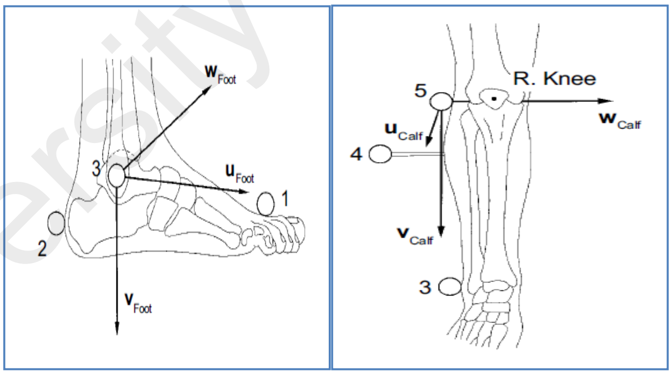
In summary, this section had showed how regression method can be applied in order to come out with mathematical equations that can predict mass and moment of inertia of each segment for everyone.

2.3.1.2 Linear Kinematics

Each segment in our lower limb (foot, shank and thigh) could be treated separately. In two dimensional (saggital plane), the joint centre for shank and foot can be easily found with the assumptions that the joint centre is placed underlying the marker at each joint for every segment. But this method cannot apply in three dimensional analyses. In order to find the joint centre for shank and foot, there are three steps that we need to follow:

1) Select three markers for the segment of interest.
2) Create an orthogonal uvw reference system based on these three markers.
3) Use prediction equations based on anthropometric measurements and the uvw reference system to estimate the joint centre positions.

Figure 2.2 shows the uvw reference system created in both foot and shank. After we had defined the uvw reference system, the prediction equations will be implemented in oreder to find the joint centre positions and equation 4.3 until 4.4 show these equations.



(a) (b)

Figure 2.2: (a) the uvw reference system for foot, (b) the uvw reference system for shank.

pAnkle = pLateral malleolus

+ 0.392(Malleolus height)vFoot

+ 0.478(Malleolus width)wFoot

+ 0.016(Foot length)uFoot (2.3)

pKnee = pFemoral epicondyle

+ 0.000(Knee diameter)uCalf

+ 0.000(Knee diameter)vCalf

+ 0.500(Knee diameter)wCalf (2.4)

Based on equation 2.3 and 2.4, external three dimensional reflective markers and anthropometric data are very useful in determine the position of the joint centre for ankle and knee joint. It should be noticed that equation 2.3 and 2.4 are meant for right foot and shank.

Now that we have the position of the joint centre for both ankle and knee, the next step require us to find the way to determine the orientation of foot and shank in the three dimensional space. Before we proceed, let us finalize a few term:

1) Global reference frame (XYZ) – is the laboratory reference frame.
2) Segment reference frame (xyz) – is the embedded reference frame in each segment.

The segment reference frame (xyz) is originated at the position of the centre of gravity for each segment. Figure 2.3 represent the global and segment reference frame.

Equation 2.5 and 2.6 shows the mathematical equations used to transform from directions of xyz to unit vector **ijk.**

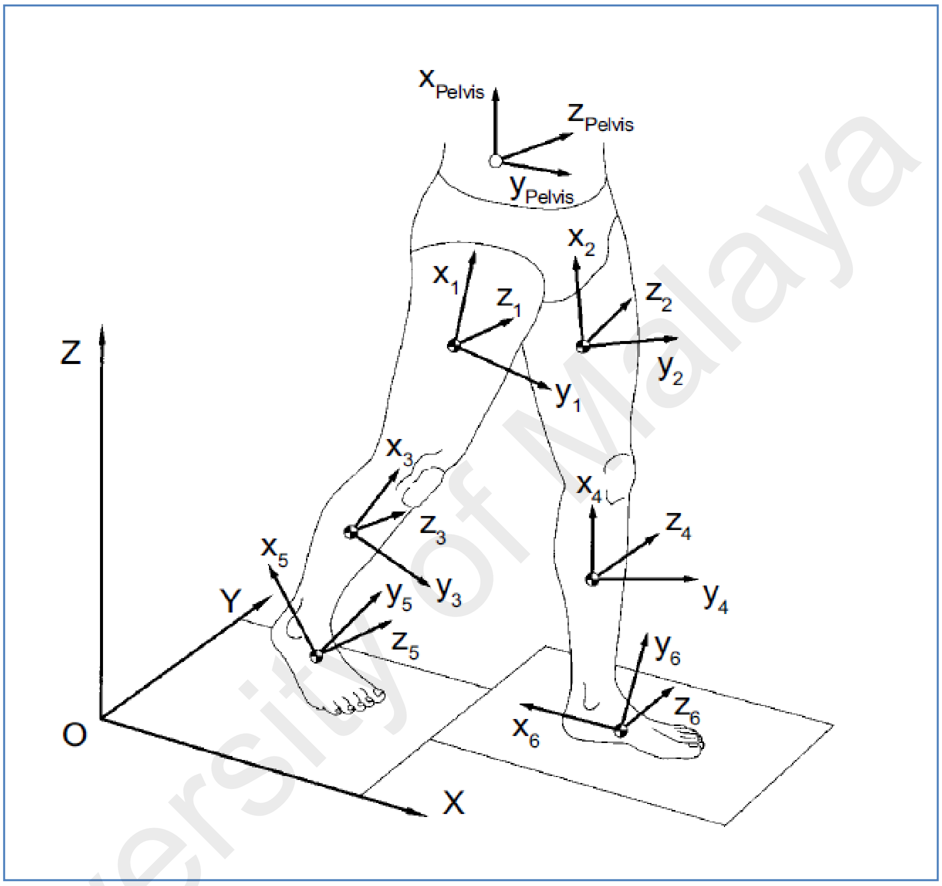


Figure 2.3: The global (XYZ) and segment (xyz) reference frame.

It must be noted that even though the **ijk** vectors are used to define the segmental coordinate system (xyz), it is actually expressed in terms of the global reference system (XYZ).

2.3.1.3 Centre of gravity

In order to calculate the centre of gravity for each segment, we need to sett of informations which are the joint center positions and the body segment parameters data.

Equation 2.5 and 2.6 shows how we can find the position of centre of gravity for shank and foot. These equations can be applied for both legs.

pCalf.CG = pKnee + 0.42 (pAnkle - pKnee) (2.5)

pFoot.CG = pHeel + 0.44 (pToe - pHeel) (2.6)

Now that we had all the necessary kinematics information, we can proceed with the inverse dynamics equation where from this we can find the net joint force at the tibiofemoral joint.

2.3.2 Inverse dynamics equation (Morrison JB 1970)

In this section we want to find the net joint force at the saggital plane of the tibiofemoral joint (2-Dimensional). The free body diagram for the tibiofemoral joint can be senn at Figure 2.4.

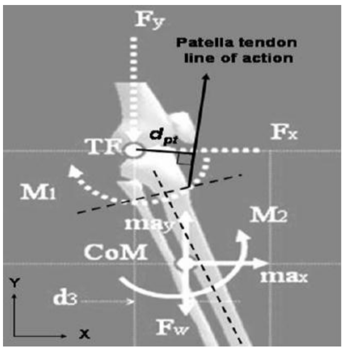


Figure 2.4: Free body diagram at tibiofemoral joint (Morrison JB 1970).

To calculate the tibiofemoral net joint force, the anatomical orientation of the force bearing structures in the knee was required. These structures were simplified to the saggital plane.

$$F_y = F_w + GRF_y - ma_y \tag{2.7}$$

$$F_x = -GRF_x - ma_x \tag{2.8}$$

Based on equation 2.7 and 2.8, we can obtain the net joint force at the tibifemoral joint during walking and stair ascending activities.

2.4 Summary

The role of mechanical factors continues from the initiation of the OA, through to development and progression of the condition. While other ('non-mechanical') factors involved will require their own unique interpretation, and ultimately be integrated to solve the unknowns in OA, the current interest in the mechanical factors should therefore focus on developing more knowledge of the stresses involved in reaching the various critical points or limits that have been described in this review. OA is likened to the failure of a system, and the limits that have to be reached before failure occurs should be ascertained in solving the design problem, one of the limits being the critical stresses that initiate the series of events, be they mechanical or biological, that initiate the degenerative process in the articulating joint. Critical stresses would refer to those from intra-articular joint reactions in relation to specific material properties of the supporting tissues. The biomechanics challenge is to determine these parameters with the specific aim of contributing new and useful knowledge to the larger effort of an integrated approach to resolving the unknowns of OA. The assumptions that provide the basis to the conjectural relationship between critical stresses and osteoarthritis can hopefully be elucidated in the process.

3.0 METHODOLOGY

3.1 Introduction

Methodology of the present study is discussed in this chapter. Figure 3.1 depicted the flow diagram of the methodology implemented. Each part of the flow diagram will be described in the following section. The chapter begins with the description of subjects involved. Statistical analysis is explained at the end of the chapter.

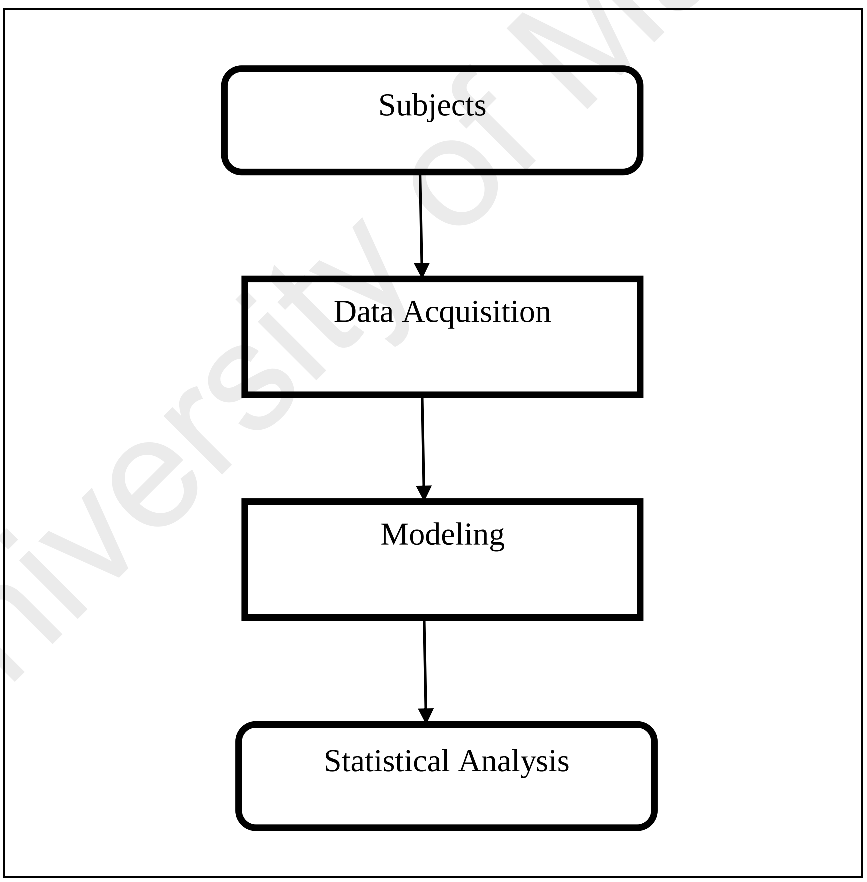


Figure 3.1: Flow diagram of the experimental methods

3.2 Subjects

18 healthy male subjects without any history of joint diseases were recruited. Absence of joint diseases would optimize the work. Subjects had a mean height of 1.6 m, and a mean body mass of 64.87 kg. Body mass index (BMI) of the subjects was also presented. Table 3.1 summarizes the information of the subjects involved in the study.

Table 3.1: Details of the subjects

Subject	Body Mass (kg)	Height (m)	Body Mass Index (kg/m2)
1	63	1.64	23.42
2	60.9	1.57	24.7
3	50.1	1.53	21.4
4	55.3	1.54	23.32
5	53.2	1.53	22.73
6	56.4	1.57	22.88
7	54.4	1.56	22.35
8	58.7	1.61	22.65
9	58.1	1.59	22.98
10	57.5	1.62	21.91
11	58.3	1.63	21.94
12	58.4	1.58	23.4
13	84.3	1.65	30.96
14	83.7	1.67	30.01
15	74.5	1.55	31.00
16	78.6	1.61	30.32
17	82.1	1.64	30.52
18	80.12	1.63	30.16

Note: $BMI = \frac{mass(kg)}{(height(m))^2}$

It is deliberately to select young and healthy subjects to eliminate the possibility of any joint diseases that would affect the results. Furthermore, those selected subjects have the ability to successfully perform the movements required during the experiment. BMI is a measure of body fat based on height and weight that applies to both adult men and women. Table 3.2 indicates the range of standard BMI and their description.

Table 3.2: Standard BMI and description (Reproduced from Henderson, 2005)

BMI ranges	**Description**
< 18.5	underweight
18.5-24.9	normal weight
25.0-29.9	overweight
>30.0	obesity

Each subject was informed of the protocol and risks of this study and was allowed to ask questions or exit the study at any time. The consent letter had been given to all the subjects to be filed.

I, …………………………………………………………………….Identity Card No……….……………………………………
(Name of Subject)

of ………..
(Address)

hereby agree to take part in the clinical research (clinical study/questionnaire study/drug trial) specified below:

Title of Study: Study of net joint forces at tibiofemoral joint for obese and normal subjects.

the nature and purpose of which has been explained to me by researcher; ..……………….…………………………..
(Name & Designation)

and interpreted by ;……………………….................................to the best of his/her ability in
(Name & Designation of Interpreter)

…………………….…………… language/dialect.

I have been told about the nature of the clinical research in terms of methodology, possible adverse effects and complications (as per study protocol sheet). After knowing and understanding all the possible advantages and disadvantages of this clinical research, I voluntarily consent of my own free will to participate in the clinical research specified above.

I understand that I can withdraw from this clinical research at any time without assigning any reason whatsoever.

Date: ……………..……….. **Signature or Thumbprint**

...
(Subject)

IN THE PRESENCE OF

Name ………………………………………..………..…

Identity Card No. ……………………….……….…… **Signature**

...
(*Witness)*

I confirm that I have explained to the subject the nature and purpose of the above-mentioned research.

Date ……………………………. **……………………………………………………**
(researcher)

Figure 3.2: Example of inform consent form filled up by the subjects

3.3 Data Acquisition

There are two main topics that are place under the data acquisition section. The first part discussing the hardware used to capture the motion, ground reaction force and the stair. All the procedures will also being included in this discussion. Second, software that was used in this study will be presented.

3.3.1 Hardware

To start this topic it is better to introduce all the hardware used in this study. All experiments were captured by the same equipments to eliminate any dissimilarity on the data. Seven VICON MX-F20 cameras were used in this study to capture the 3-Dimensional motion of human movement. These cameras were working along with markers which were attached on the human body for the reason that these infra red cameras only detect the reflective markers. For that reason, reflective markers were always placed to represent joints and body segments. Figure 3.3, Figure 3.4 and Figure 3.5 shows the camera, reflective marker and its placement used in this experiment.

Figure 3.3: Vicon MX-F20 camera.

Figure 3.4: Example of reflective marker used in this experiment.

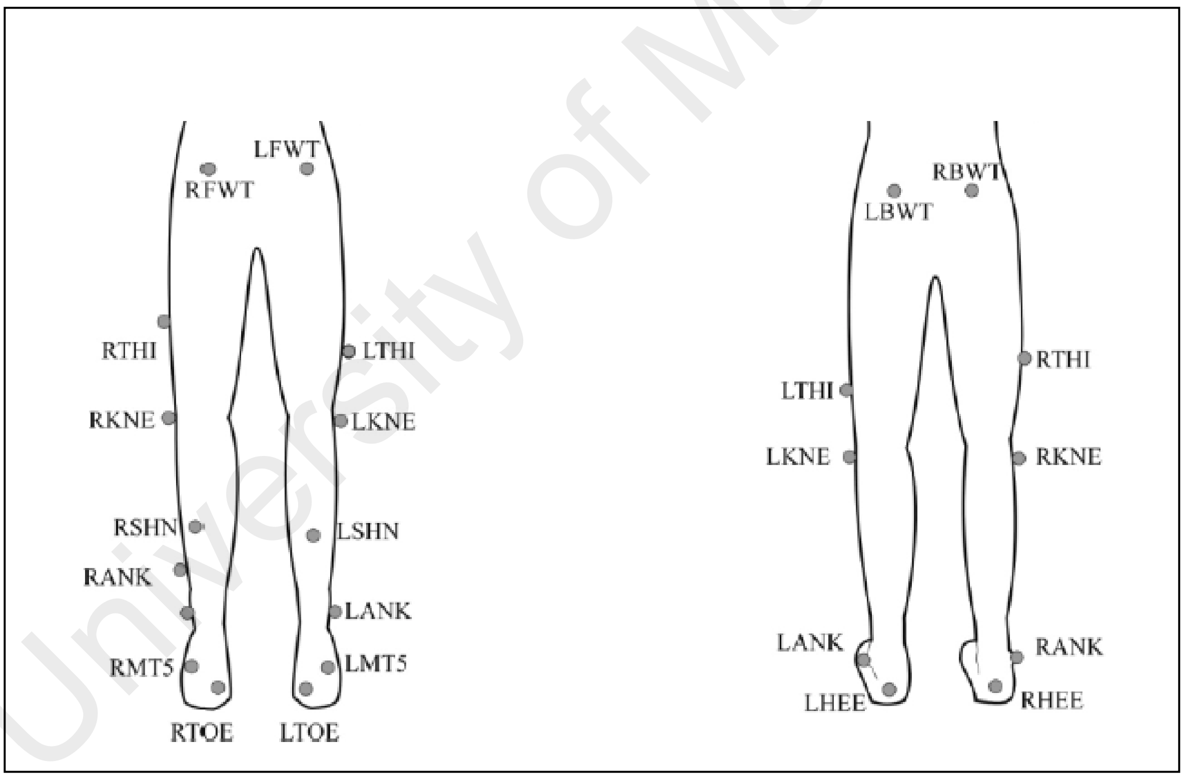


Figure 3.5: Marker placement used in this study.

In this study, the marker placement was applied only on the lower limb. Sixteen markers were used during the experiment and Table 3.3 briefly explained where the marker attachment is. The entire markers placement is based on the manual given by VICON.

Table 3.3: Label for markers placement along with its definition and position.

Marker Label	**Definition**	**Position**
LTHI	Left thigh	Placed on the outside of the thigh below hand swing.
LKNE	Left knee	Placed on the outside of the knee joint.
LSHN	Left shin	Placed on the outside of the lower leg.
LANK	Left ankle	Placed on the bony prominence on the outside of the ankle.
LHEE	Left heel	Placed on the back of the foot.
LMT5	Left 5th metatarsal	Placed on the outside of the foot at the base of the little toe.
LTOE	Left toe	Placed on the tip of the big toe.
RTHI	Right thigh	Placed on the outside of the thigh below hand swing.
RKNE	Right knee	Placed on the outside of the knee joint.
RSHN	Right shin	Placed on the outside of the lower leg.
RANK	Right ankle	Placed on the bony prominence on the outside of the ankle.
RHEE	Right heel	Placed on the back of the foot.
RMT5	Right 5th metatarsal	Placed on the outside of the foot at the base of the little toe.
RTOE	Right toe	Placed on the tip of the big toe.
LFWT	Left Front Waist	Left front waist.
RFWT	Right Front Waist	Right front waist.
LBWT	Left back Waist	Left back waist.
RBWT	Right back Waist	Right back waist.

To acquire the data for ground reaction force, the force transducer known as force plate was used. It will measured the ground reaction force generated by a body who standing still or moving across it. So in this experiment, three Kistler force plate were used in this experiment where, two were used during walking test and the other one was used for stair ascending. Figure 3.6 show the picture of the force plate that was used in this experiment. Force plate that were used in this experiment can acquire the three dimensional components (Fx, Fy, Fz, Mx, My and Mz) of a single equivalent force applied on it.

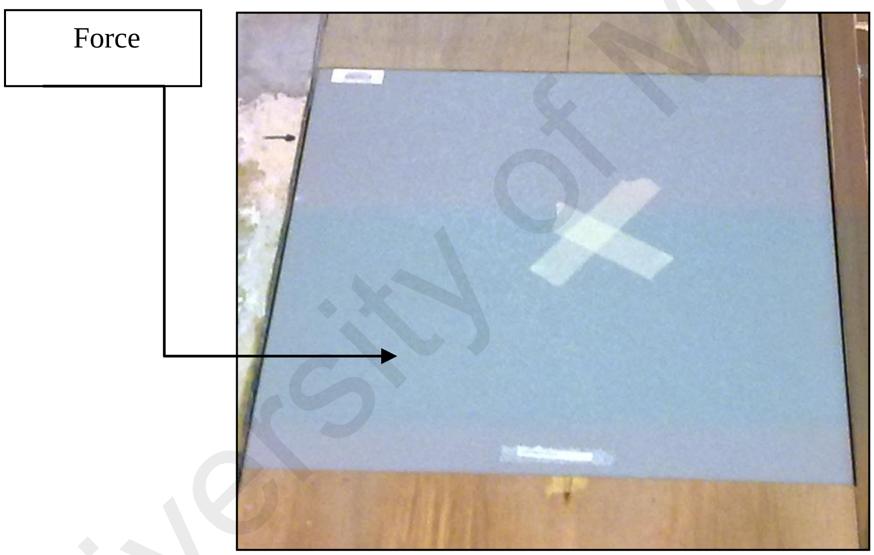


Figure 3.6: A picture of Kistler force plate that was used in this experiment.

The instrumented stair had been made, where one force plate had been placed at the second step of the stair in order to capture the kinetics data from the subject while performing the stair ascending activity. Figure 3.7 shows the instrumented stair used in this experiment. The reason why the force plate was placed at the second step is because we want to eliminate the possibility of drift that might occur when the subjects change from walking to stair ascending.

Figure 3.7: A picture of the instrumented stair.

3.3.2 Software

Two essentials software had been used in this study. The first software is the Vicon workstation where it will collect all the kinetics and kinematics data during the experiment and doing the basic analysis such as acquiring the position and acceleration of markers. We can also acquire the centre of pressure and ground reaction force from it. Vicon workstation play an important role in the analysis process as all the useful information will be collected before the process of inverse dynamic.

The second software that was used in this study is Matlab. All the equations of the inverse dynamics were rewrite in Matlab through the M-file format. By using M-file format it is easily to convert the mathematical modeling of the human motion (inverse dynamics) to the programming language.

3.3.3 Experimental protocol

In this experiment, subjects need to perform two types of activity. In this section, we will explain the experimental protocols according to the types of activity. However, there are general procedures that need to be done before the subjects perform the walking and stair ascending activity. First, anthropometric measurements need to be taken for all subjects. This is a must since we want to record their body mass index (BMI) for future analysis and the anthropometric data for the inverse dynamics equation. Next, subjects need to change their attire to the swimming attire where this is provided for all subjects. After that, reflective markers were attached to the subjects according to the standard marker placement provided by the manufacturer (Vicon). Figure 3.8 shows the subject preparation during the experiment. When all of this was done, subjects were ready to perform the two tasks as described before.

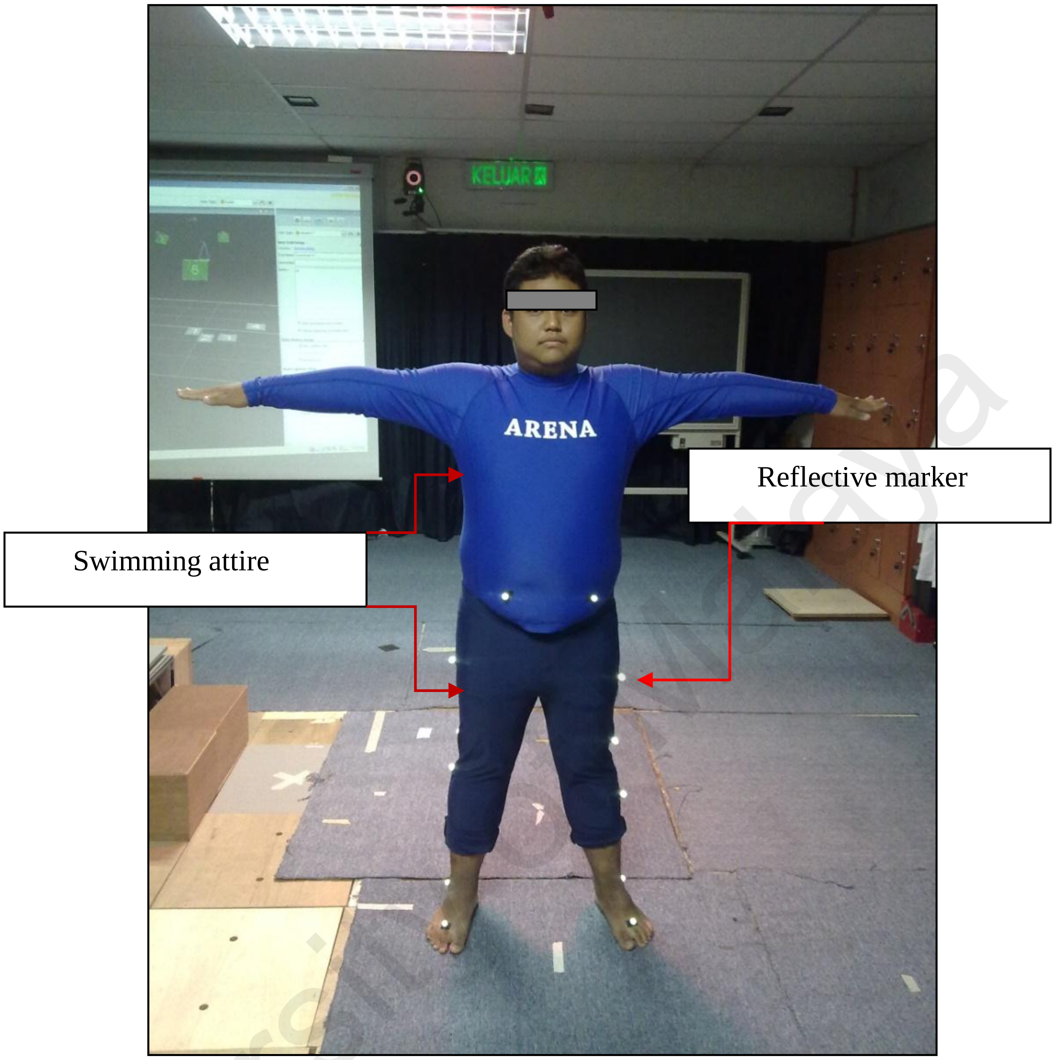


Figure 3.8: Subject preparation.

A. Walking

For walking activity, subjects need to walk along the experimental route as normal as possible. To get a normal gait during the walking activity, the starting point was set away from the force plates and a rubber mat was used to cover it. This is also to eliminate the drift that might occur during the step on the force plate. Subjects will start to walk when the researcher say start and stop walking when they reach the finish line. Subjects need to repeat this procedure for three times. Figure 3.9 shows the subject perform the walking activity.

Figure 3.9: Walking activity.

B. Stair ascending

During stair ascending activity, subjects need to start walking from a distance that had been set away from the first step of the stair. Same as the walking activity, subjects will start walking and stair ascending when the researcher says start. They need to perform this task as normal as possible and need to repeat this procedure for three times. Subjects will stop the stair ascending activity when they reach the final step. Figure 3.10 shows the subject perform the stair ascending activity.

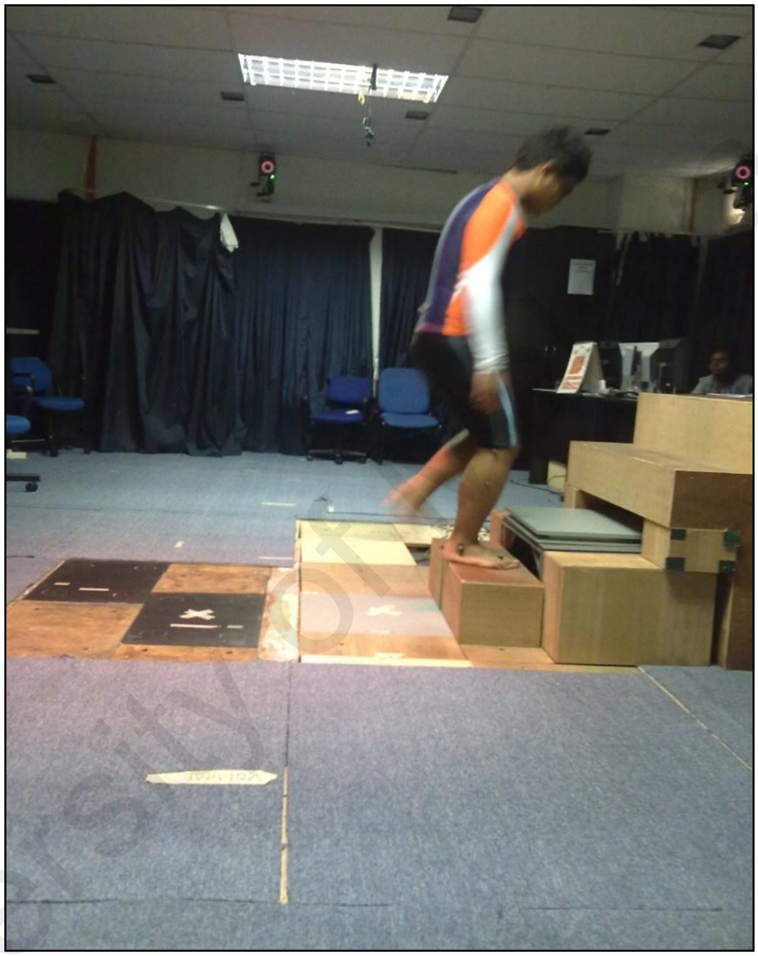


Figure 3.10: Stair ascending activity.

3.4 Inverse Dynamics

In this section, we will discuss about the mathematical modeling of human's lower limb. After that, discussion on inverse dynamic technique will take place because by this technique the net joint force at tibiofemoral joint will be obtained. Mathematical modeling can be in the form of two-dimensional (2D) or three-dimensional (3D). A comparison between 2D and 3D mathematical modeling of human lower limb will be discussed.

Previously, 2D inverse dynamic model was used to determine the net joint force and moment at the sagittal plane (Wu G, 1995). The 2D model is much simpler than the 3D model. A 2D model can be developed by just using one camera and fewer markers used in order to define the position of the 2D joint centre (Winter DA, 1990). There are a few studies that had been done before, to see the differences between 2D and 3D inverse dynamic model (Eng JJ et al, 1995, Ramakrishnan et al, 1987, Cappozzo A et al, 1991). It was reported that the 3D sagittal joint moments were similar with the 2D sagittal joint moments (Eng JJ et al, 1995). Even though there is a small differences at the peak value for joint moments, the overall time course patterns were identical for both inverse dynamic models. The co-ordinate system used at the joint centre is very important as the both inverse dynamic model shows significant differences at the joint moments (Alkajaer T. et al, 2001).

Even though the overall time course patterns is similar for both model, it should be noted that the significant differences in term of values should considered. Therefore, from the above comparison study, it is decided that the human lower limb will be modeled with the 3D mathematical model. Based on the 3D mathematical modeling of human lower limb, the net joint force for the tibiofemoral joint at saggital plane can be obtained.

In chapter 4, the details of the mathematical modeling of human lower limb and the 2D inverse dynamics equation will be presented.

3.5 Statistical Analysis

Mean and standard deviation of the net joint force were calculated. Percentage of changes between net joint force at tibiofemoral joint during walking and stair ascending were also calculated to get the better understanding of the effects of BMI during activity of dailiy living. Correlations between dependent and independent variables were obtained so that the effect of BMI on net joint force can be obtained and proof our hypothesis. Statistical analysis was completed using SPSS (Statistical Package for the Social Sciences) for Windows version 17.0.

3.6 Summary

In this present study, method in acquiring the kinetics and kinematics data for 18 subjects had been discussed. Also in this chapter, the inverse dynamic had been discussed briefly as the details will be on chapter 4. Finally, statistical analysis using computer software had been explained.

4.0 RESULTS

4.1 Introduction

In this chapter, results and data analysis will be presented. It will be in the form of figures, text as well as tables which were derived from the experiment. Results will be focused on the net joint force at knee during both walking and stair ascending and how it associated with body mass index (BMI).

4.2 Measurement of net joint force

Mean and standard deviation of net joint force obtained from the proposed inverse dynamic technique as in the Table 4.1.

Table 4.1: Net joint force obtained from inverse dynamic technique.

Descriptive Statistics

	N	Minimum (N)	Maximum (N)	Mean (N)	Std. Deviation (N)	Variance (N)
Net_Joint_Force_Knee_Walking	18	910.20	1685.09	1216.02	292.42	85513.89
Net_Joint_Force_Knee_StairAscending	18	2392.80	3744.65	2841.30	531.96	282983.78
Valid N (list wise)	18					

During normal gait (walking) the average net joint force at knee is 1216.02N with a standard deviation of +292.42 N and -292.42 N. While during stair ascending the average value is much higher which is 2841.30 N with a standard deviation of +531.96 N and -531.96 N. These values were taken with the assumption that all of the participants placed in the same group. To visualize the contribution of individual group (Normal and Obese) for net joint force, see Figure 4.1 and Figure 4.2.

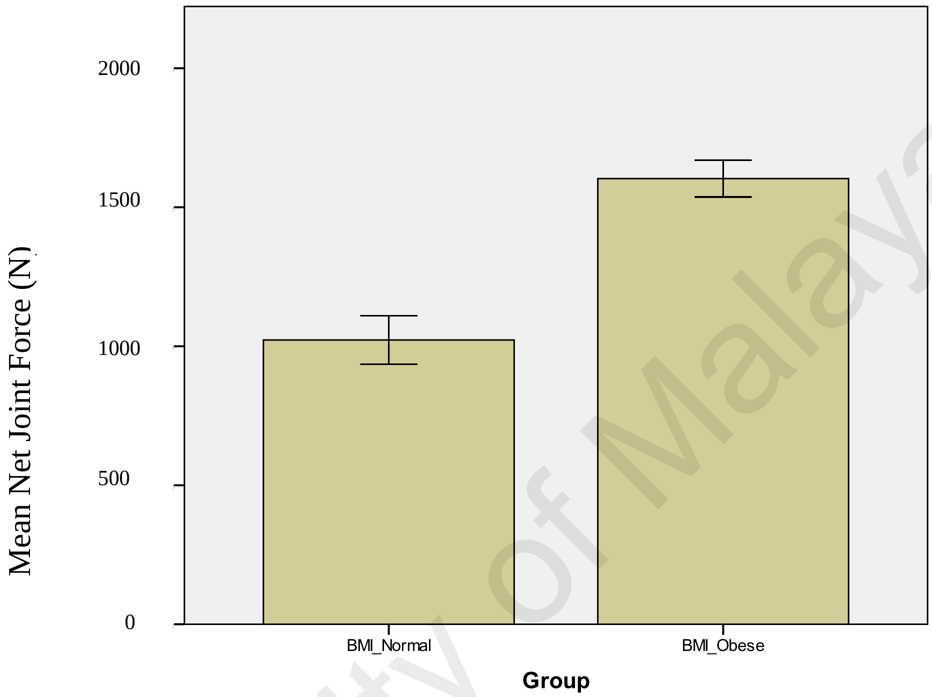


Figure 4.1: Net joint force calculated by using inverse dynamic technique for both normal and obese group during normal gait (walking).

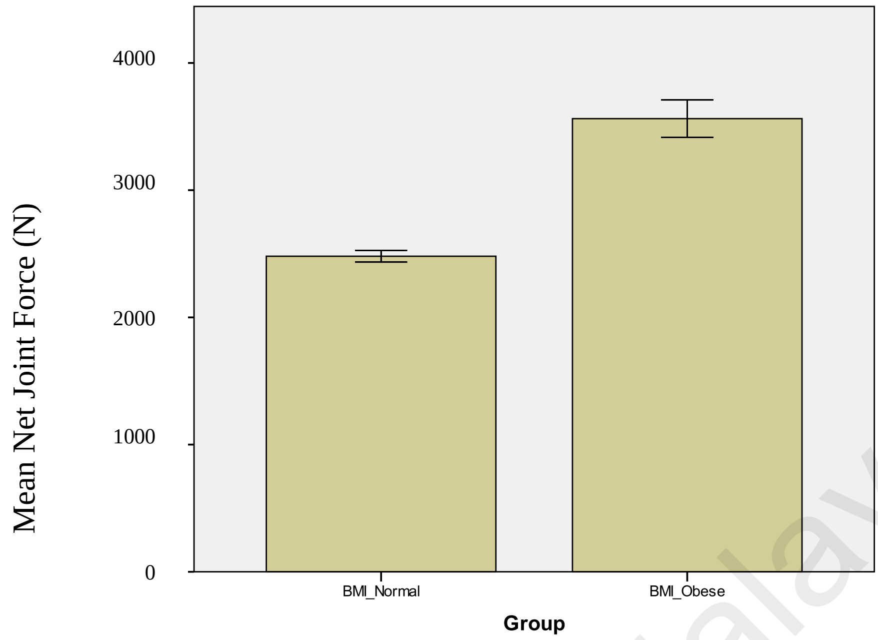


Figure 4.2: Net joint force calculated by using inverse dynamic technique for both normal and obese group during stair ascending.

Two main points that can be depicted from Figure 4.1 and Figure 4.2, obese group generate large value of force compare to normal group in both activity and the value of force is affected by the type of activity being done. Equation 4.1 can be used to see the changes of force at knee between obese and normal group.

$$\frac{Fy\ (Obese) - Fy\ (Normal)}{Fy\ (Normal)} \tag{4.1}$$

The net joint force for obese group is differ from the normal group during both normal gait (walking) and stair ascending, with differences of 56.78% and 43.6% respectively.

$$\frac{Fy\ (stair\ ascending) - Fy\ (walking)}{Fy\ (walking)} \tag{4.2}$$

Using equation 4.2, we can clearly see how the force changes with respect to the activity done for both normal and obese group. The changes of force for normal and obese group with respect to the activity are 142.62% and 122.22% respectively.

4.3 Statistical Analysis

In this study, the net joint force is the dependent variable while body mass index (BMI) is the independent variable. The dependent and independent variables are very important in order for us to see the correlation between both variables. From Table 4.2 and Table 4.3, it was shown that there were strong positive correlations between body mass index (BMI) and net joint force for both walking and stair ascending activities which were statistically significances Pearson Correlation = 0.966, n = 18, $p < 0.05$ and Pearson Correlation = 0.971, n = 18, $p < 0.05$ respectively.

Table 4.2: Correlation between dependent variable (Fy) and independent variable (BMI) during normal gait (walking).

		BMI	**Net_Joint_Force_Knee_Walking**
BMI	Pearson Correlation	1	.966**
	Sig. (2-tailed)		.000
	N	18	18
Net_Joint_Force_Knee_Walking	Pearson Correlation	.966**	1
	Sig. (2-tailed)	.000	
	N	18	18

Table 4.3: Correlation between dependent variable (Fy) and independent variable (BMI) during stair ascending.

Correlations

		BMI	Net_Joint_Force_Knee_StairAscending
BMI	Pearson Correlation	1	.971**
	Sig. (2-tailed)		.000
	N	18	18
Net_Joint_Force_Knee_StairAscending	Pearson Correlation	.971**	1
	Sig. (2-tailed)	.000	
	N	18	18

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

Significant differences between the net joint force for both group during normal gait (walking) and stair ascending were analyzed using ANOVA.

Table 4.4: Test of Homogeneity of Variances

	Levene Statistic	df1	df2	Sig.
Net_Joint_Force_Knee_Walking	.264	1	16	.614
Net_Joint_Force_Knee_StairAscending	4.933	1	16	.041

Table 4.5: ANOVA test.

		Sum of Squares (N^2)	df	Mean Square (N^2)	F	Sig.
Net_Joint_Force_Knee_Walking	Between Groups	1348249.937	1	1348249.94	204.50	.000
	Within Groups	105486.126	16	6592.88		
	Total	1453736.063	17			
Net_Joint_Force_Knee_Stair Ascending	Between Groups	4679551.979	1	4679551.98	570.80	.000
	Within Groups	131172.254	16	8198.27		
	Total	4810724.233	17			

Table 4.6: Robust Tests of Equality of Means

		Statistic[a]	df1	df2	Sig.
Net_Joint_Force_Knee_Walking	Welch	247.28	1	12.60	.000
Net_Joint_Force_Knee_StairAscending	Welch	309.61	1	5.50	.000

a. Asymptotically F distributed.

Before we go to the ANOVA, we must be aware that one of the assumptions of the one-way ANOVA is that the variances of the groups you are comparing are similar. To test this, we use Levene’s Test of Homogeneity of Variance, where Table 4.4 shows the significance value of net joint force for both activities. For normal gait (walking), the significance value is 0.614, which indicate that the assumption of homogeneity of variance is met (0.614 > 0.05). Contrast with the stair ascending activity, it does not met the assumption of homogeneity of variance because the significance value is less than 0.05 (0.041 < 0.05). In a nutshell, group that met the assumption of homogeneity of variance can proceed with the ANOVA analysis, but those who fail to meet the assumption of homogeneity of variance must use Robust Test of Equality of Means to find the significant difference.

Table 4.5 shows the ANOVA test for both during walking and stair ascending. But, need to remember that the significant difference for stair ascending activity cannot be depicted from here due to the fact that it's fail to met the assumption of homogeneity of variance. For normal gait (walking), there was a statistically significant difference between groups as determined by one-way ANOVA (F (1, 16) = 204.50, $p < 0.05$). There was also a statistically significant difference during stair ascending activity where from Table 4.6 it shows that the p value is less than 0.05 ($p < 0.05$).

4.4 Summary

The net joint forces are ranging from 1022.50 N (SD = +87.17, -87.17) and 2480.76 N (SD = +45.78, -45.78) for normal group with respect to both activities. The correlation between dependent and independent variable is very significant for both activities. This is agreed by both test, the Pearson Correlation and ANOVA.

5.0 DISCUSSIONS

There are several factors that contribute to the increasing in the net joint force at knee and one of them is body mass index (BMI). But not to forget, types of activities that been done also affecting the value of net joint force at knee. Among the activity of daily living (ADL) that requires this is sit-to-stand, stair ascending and descending and squatting.

In this chapter, first we will compare the result that had been collected during the experiment. We want to see whether it's agreed with the previous researchers. It is noted that during walking, knee tend to withstand a large value of net joint force which is approximately about 2 times body weight (Zhao D, 2007, Winby CR, 2009). From our study, we had obtained 1.82 times body weight and 2 times body weight for normal and obese group respectively. This is not far behind from the previous result provide by Zhao D. (2007) and Winby CR. (2009) which only had a percentage of changes of 9% and 0% respectively.

From previous study, the value of net joint force at knee was approximately about 4.25 times body weight (Morrison, 1969). In our case, we have 4.07 times body weight and 4.03 times body weight with respect to the normal and obese group (4.24% and 5.18% of changes for both group respectively where it is not far behind from rthe result provided by Morrison (1969)). To see the effect of types of activites on the value of net joint force at knee, we compare the value of loads at knee joint for both walking and stair ascending activity. From these values it is clearly says that types of activity affecting the result of loads at knee joint although the different in term of body mass index among the participants was also included in this statement. However, the normalized value of the net joint force only suitable to discuss the effect of types of activities at the knee joint but does not indicated the correlation between body mass index and net joint force.

It is not enough to just look at the normalized value of the net joint force (Force / Body Weight) in explaining the effect of body mass index (BMI) on the loads at knee joint. For example, during stair ascending the normalized value of net vertical joint force at knee were 4.07 times body weight and 4.03 times body weight with respect to the normal and obese group, where it shows almost no different. This is the same during the normal gait (walking). We need to look at the value of the force itself. Table 5.1 show the data of force itself for both group (normal and obese) during normal gait (walking) and stair ascending.

Table 5.1: Mean and Standard Deviation of net joint force at knee for both groups during walking and stair ascending.

		N	Mean (N)	Std. Deviation (N)	95% Confidence Interval for Mean		Minimum (N)	Maximum (N)
					Lower Bound (N)	Upper Bound (N)		
Net Joint Force (walking)	BMI_Normal	12	1022.50	87.17	967.11	1077.89	910.20	1241.23
	BMI_Obese	6	160307	66.17	1533.63	1672.51	1487.99	1685.09
	Total	18	1216.02	292.43	1070.60	1361.44	910.20	1685.09
Net Joint Force (stair ascending)	BMI_Normal	12	2480.76	45.78	2451.67	2509.85	2392.97	2546.58
	BMI_Obese	6	3562.38	147.05	3408.06	3716.70	3306.65	3744.65
	Total	18	2841.30	531.96	2576.76	3105.84	2392.97	3744.65

From Table 5.1, it shows that the force itself clearly show the huge different in the net joint force for normal and obese group with respect to the same activity. There is 56.78 percentages of changes while performing walking activity. The same goes while performing the stair ascending activity, 43.6 percentages of changes. We can conclude that as the body mass index (BMI) increase, the net joint force at knee also increased, since the correlation between body mass index (independent variable) and net vertical joint force (dependent variable) shows a strongly positive correlations (Pearson Correlation = 0.966, $n = 18$, $p < 0.05$ and Pearson Correlation = 0.971, $n = 18$, $p < 0.05$ with respect to walking and stair ascending).

Furthermore, we want to see the clinical relevance in the study of mechanical factor, which in this case is the body mass index (BMI) related to the knee osteoarthritis (knee OA). In this discussion, we omit the changes in the cartilage properties as here we only study about the loads at the knee. At the weight-bearing joint especially knee, obesity comes as the most preventable risk factor for osteoarthritis. People with a body mass index (BMI) of 27.5 or greater were placed in the group that has the highest possibility on having a musculoskeletal disease (Burton, 1998). To understand how body mass index associated with the onset and progression of knee osteoarthritis, see Figure 5.1 and Figure 5.2. The pivot in the figure represents the properties of cartilage. As there is no changes on the cartilage properties and the loads at the knee become higher (56.78 and 43.6 percentage of changes between normal and obese group of net joint force for both activity, obtained from this study) the risk for cartilage damage become higher. This is agreed with the previous researcher who claims that body mass index (BMI) is associated positively with cartilage defects (Ding, 2005). In every kilograms increase in body weight, there were about 9% to 13% increased in the risk factor for the onset of knee osteoarthritis (Cicuttini, 1996).

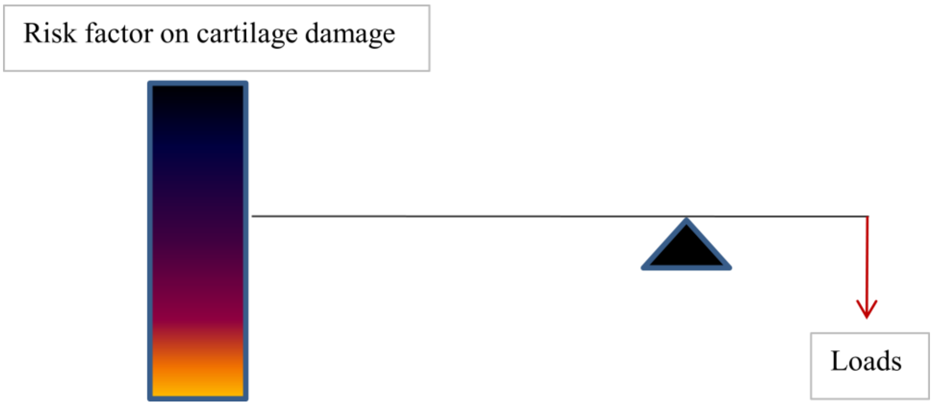


Figure 5.1: Show the balance between loads while doing daily activities and level of risk that associated with cartilage damage.

Figure 5.2: Show the effect of increasing in loads while performing daily activities with the level of risk factor that associated with cartilage damage.

In a nutshell, the loads at the knee will differ with respect to the activities being done during daily life. This is because each activities having different degree of joint flexion and extension. The hypothesis saying that body mass index (BMI) is associated with the onset of knee osteoarthritis is truly proof as increased in body mass index (BMI) will increased the net joint force at knee; therefore, increased the level of risk factor of cartilage's damage.

6.0 CONCLUSION

This study was conducted in order to determine the net joint forces at the tibiofemoral joint. The values of forces at the tibiofemoral joint are very important because based on it we can biomechanically relate it with the joint disease. In tis study, 18 subject had been used where 12 is normal subjects anf 6 is subject with obesity. The experimental procedures made the subject to perform two daily life activities which were stair ascending and walking. The result obtained shows that obese subjects produced bigger force at the tibiofemoral joint compare to normal subjects for both activities. Based on this, it is clearly seen how body mass index (BMI) can bee associated with the onset and progression of knee OA. Another important thing that can be deduced from this tudy is that net joint forces generated at the tibiofemoral joint also been influenced by the types of activities been performed. As for tis study, stair ascending produced bigger net joint force compare to walking activity.

This study is limited to only net joint forces at the tibiofemoral joint. It does not indicates the stress pattern at the tibiofemoral joint. The stress pattern at tibiofemoral joint is very important because based on it we can narrow down the types of OA that the patient had suffered. This information also very beneficial as we can see how the patient altered naturally their gait in order to overcome the pain cause by the knee OA.

In a conclusion, it is appropriate to suggest that obesity can cause an early onset of OA and this can be assessed using inverse dynamic analysis where a non-invansive method were used. Having a normal body mass index will help to overcome this problem.

REFERENCES

Alkjaer T., Simonsen E.B., Dhyre-Poulsen Comparison of inverse dynamics calculated by two and three dimensional mdels during walking. *Gait and Posture* 13 (2001) 73–77.

Anderson JJ, Felson DT, Naimark A, Walker AM, Meenan RF. Obesity and knee osteoarthritis. The Framingham Study. Ann Intern Med. 1988;109:18-24. [PMID: 3377350]

Andriacchi TP, Mundermann A, Lane Smith R, Alexander EJ, Dyrby CO, Koo S. A framework for the in vivo pathomechanics of osteoarthritis at the knee. 2004 Mar; 32 (3): 447-457.

Arnoczky, S.P. (1983): Anatomy of the anterior cruciate ligament. *Clin Orthop, (172): 19-25.*

Brown TD, Shaw DT. In vitro contact stress distribution on the femoral condyles. *J Orthop Res.* 1984;2(2):190-9.

Buckwalter JA, Brown TD. Joint injury, repair, and remodeling: roles in post traumatic osteoarthritis. *Clin Orthop.* 2004 Jun(423):7-16.

Burr DB, Radin EL. Microfractures and microcracks in subchondral bone: are they relevant to osteoarthrosis? *Rheum Dis Clin North Am.* 2003 Nov;29(4):675-85.

Burton WN, Chen CY, Schultz AB, et al. *J Occup Environ Med* 1998;40(9):786–792.

Butler DL, Noyes FR, Grood ES (1977). (In Press)."Ligamentous Contributions to A P Stability of the Knee," [Abstract]. *Symposium on Biomechanics, ASME.*

Cappozzo A. Three-dimensional analysis of human walking: experimental methods and associated artifacts. *Hum Movement Sci* 1991;10:589–602.

Cicuttini FM, Baker JR, Spector TD. *J Rheumatol* 1996;23(7):1221–1226.

Conaghan PG, Felson DT. Structural associations of osteoarthritis pain: lessons from magnetic resonance imaging. *Novartis Found Symp.* 2004;260:191-201; discussion 201-5, 277-9.

Daniel DM, Stone ML: Ch. 3 – Case Studies. In: Daniel DM, Akeson WA, O'Connor JJ: Knee Ligaments: Structure, Function, Injury, and Repair. *New York: Raven Press*, 31-5; 1990.

Ding C, Cicuttini F, Scott F, et al. *Obes Res* 2005;13(2):350–361.

Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? *J Biomech* 1995;28:753–8.

Fazzalari NL, Vernon-Roberts B, Darracott J. Osteoarthritis of the hip. Possible protective and causative roles of trabecular microfractures in the head of the femur. *Clin Orthop.* 1987 Mar(216):224-33.

Felson DT, Chaisson CE, Hill CL, Totterman SM, Gale ME, Skinner KM, Kazis L, Gale DR. The association of bone marrow lesions with pain in knee osteoarthritis. *Ann Intern Med.* 2001 Apr 3;134(7):541-9.

Felson DT, McLaughlin S, Goggins J, LaValley MP, Gale ME, Totterman S, Li W, Hill C, Gale D. Bone marrow edema and its relation to progression of knee osteoarthritis. *Ann Intern Med.* 2003 Sep 2;139(5 Pt 1):330-6.

Felson DT. An update on the pathogenesis and epidemiology of osteoarthritis. *Radiol Clin North Am* 42(1): I -9, *2004*.

Felson DT. Obesity and osteoarthritis of the knee. *Bull Rheum Dis.* 1992;41(2):6-7.

Fukubayashi T, Kurosawa H. The contact area and pressure distribution pattern of the knee. A study of normal and osteoarthrotic knee joints. *Acta Orthop Scand.* 1980 Dec;51(6):871-9.

Gelber AC, Hochberg MC, Mead LA, Wang NY, Wigley FM, Klag MJ. Joint injury in young adults and risk for subsequent knee and hip osteoarthritis. *Ann Intern Med.* 2000 Sep 5;133(5):321-8.

Guilak F, Fermor B, Keefe FJ, Kraus VB, Olson SA, Pisetsky DS, Setton LA, Weinberg JB. The role of biomechanics and inflammation in cartilage injury and repair. *Clin Orthop.* 2004 Jun(423):17-26.

Hill CL, Gale DG, Chaisson CE, Skinner K, Kazis L, Gale ME, Felson DT. Knee effusions, popliteal cysts, and synovial thickening: association with knee pain in osteoarthritis. *J Rheumatol.* 2001 Jun;28(6):1330-7.

Ismail MN. (2002). The nutrition and health transition in Malaysia. *Public Health* Jun;25(6):789e97.

K.N. Laubenthal, G.L. Smidt and *D.B. Kettlekemp*, A quantitative analysis of knee motion duringactivities of daily living. *Phys Ther* 52 (*1972*), p. 34.

Kellgren JH, Lawrence JS. Radiological assessment of osteo-arthrosis. *Ann Rheum Dis.* 1957 Dec;16(4):494-502.

Ramakrishnan HK, Kadaba MP, Wootten ME. Lower extremity joint moments and ground reaction torque in adult gait. Biomechanics of Normal And Prosthetic Gait. ed.Stein JL. 1987;BED-Vol. 4 *ASME Winter Annual Meeting*:87–92.

McKellop HA, Sigholm G, Redfern FC, Doyle B, Sarmiento A, Luck JV Sr. The effect of simulated fracture-angulations of the tibia on cartilage pressures in the knee joint. *J Bone Joint Surg Am.* 1991 Oct;73(9):1382-91.

Morrison JB. Function of the knee joint in various activities. *Biomed Eng* 4:573- 580, 1969.

Morrison, J.B., 1970. The mechanics of the knee joint in relation to normal walking. *Journal of Biomechanics* 3, 51–61.

Noyes FR, McGinniss GH, Grood ES: The variable functional disability of the anterior cruciate ligament-deficient knee. *Orthop Clin North Am* 16: 47-67, 1985.

Noyes FR, Schipplein OD, Andriacchi TP, Saddemi SR, Weise M: The anterior cruciate ligament-deficient knee with varus alignment. An analysis of gait adaptations and dynamic joint loadings. *Am J Sports Med* 20: 707-716, 1992.

Oakley SP, Lassere MN, Portek I, Szomor Z, Ghosh P, Kirkham BW, Murrell GA, Wulf S, Appleyard RC. Biomechanical, histologic and macroscopic assessment of articular cartilage in a sheep model of osteoarthritis. *Osteoarthritis Cartilage.* 2004 Aug;12(8):667-79.

Obeid EM, Adams MA, Newman JH. Mechanical properties of articular cartilage in knees with unicompartmental osteoarthritis. *J Bone Joint Surg Br.* 1994 Mar;76(2):315-9.

Radin EL, Burr DB, Caterson B, Fyhrie D, Brown TD, Boyd RD. Mechanical determinants of osteoarthrosis. *Semin Arthritis Rheum.* 1991 Dec;21(3 Suppl 2):12 21.197)

Radin EL, Parker HG, Paul IL. Pattern of degenerative arthritis. Preferential involvement of distal finger-joints. *Lancet.* 1971 Feb 20;1(7695):377-9. 198)

Radin EL, Paul IL, Rose RM. Role of mechanical factors in pathogenesis of primary osteoarthritis. *Lancet.* 1972 Mar 4;1(7749):519-22. 201)

Radin EL, Paul IL. Response of joints to impact loading. I. In vitro wear. *Arthritis Rheum.* 1971 May-Jun;14(3):356-62.

Radin EL, Rose RM. Role of subchondral bone in the initiation and progression of cartilage damage. *Clin Orthop.* 1986 Dec(213):34-40.

Radin EL, Yang KH, Riegger C, Kish VL, O'Connor JJ. Relationship between lower limb dynamics and knee joint pain. J Orthop Res. 1991 May;9(3):398-405. Erratum *in: J Orthop Res* 1991 Sep;9(5):776. 210)

Radin EL. Osteoarthrosis. What is known about prevention. *Clin Orthop.* 1987 Sep(222):60-5.

Radin EL. Osteoarthrosis--the orthopedic surgeon's perspective. *Acta Orthop Scand Suppl.* 1995 Oct;266:6-9.

Radin ER, Paul IL, Rose RM. Pathogenesis of primary osteoarthritis. *Lancet.* 1972 Jun 24;1(7765):1395-6. 213)

Radin ER, Paul IL, Rose RM. Pathogenesis of primary osteoarthritis. Response of joints to impact loading. 3. Relationship between trabecular microfractures and cartilage degeneration. *J Biomech.* 1973 Jan;6(1):51-7.

Satku K, Kumar VP, Chong SM, Thambyah A. The natural history of spontaneous osteonecrosis of the medial tibial plateau. *J Bone Joint Surg Br.* 2003 Sep;85(7):983 8.

Schipplein OD, Andriacchi TP. Interaction between active and passive knee stabilizers during level walking. *J Orthop Res.* 1991 Jan;9(1):113-9.

Stäubli HU, Jakob RP Anterior knee motion analysis. Measurement and simultaneous radiography. *Am J Sports Med.* 1991 Mar-Apr;19(2):172-7.

Tang WM, Zhu YH, Chiu KY. Axial alignment of the lower extremity in Chinese adults. *J Bone Joint Surg Am.* 2000 Nov;82-A(11):1603-8.

Vaughn L. C., Davis L. B., O'Connor J.. Second Edition: Dynamics of Human Gait. Kiboho Publishers. 1992, ISBN 0-620-23558-6.

Walker, PS, Miegel, RE, Rodger, RF, et al: A compliant interface for total knee replacement arthroplasty. in *Engineering and Clinical Aspects of Endoprosthetic Fixation, Institution of Mechanical Engineers, London*, 1984, pp. 131-134.

Weidow J, Pak J, Karrholm J. Different patterns of cartilage wear in medial and lateral gonarthrosis. *Acta Orthop Scand*. 2002 Jun;73(3):326-9.

Wilder FV, Hall BJ, Barrett JP Jr, Lemrow NB. History of acute knee injury and osteoarthritis of the knee: a prospective epidemiological assessment. The Clearwater Osteoarthritis Study. *Osteoarthritis Cartilage.* 2002 Aug;10(8):611-6.

Winby CR, Lloyd DG, Besier TF, Kirk TB. Muscle and externalload contribution to knee joint contact loads during normal gait. *J Biomech* 2009 Oct 16;42(14):2294e300.

Winter DA. Biomechanics and Motor Control of Human Movement. *New York: John Wiley & Sons*, 1990.

Wu G. A review of body segmental displacement, velocity, and acceleration in human gait. In: *Craik RL, Oatis CA, editors. Gait Analysis: Theory and Application. St. Louis, Missouri: Mosby-Year Book, Inc.*, 1995:205–22.

Zhang Y, Hunter DJ, Nevitt MC, Xu L, Niu J, Lui LY, Yu W, Aliabadi P, Felson DT.Association of squatting with increased prevalence of radiographic tibiofemoral knee osteoarthritis: the Beijing Osteoarthritis Study.*Arthritis Rheum.* 2004 Apr;50(4):1187-92.

Zhang Y, Xu L, Nevitt MC, Aliabadi P, Yu W, Qin M, Lui LY, Felson DT. Comparison of the prevalence of knee osteoarthritis between the elderly Chinese population in Beijing and whites in the United States: The Beijing Osteoarthritis Study. *Arthritis Rheum.* 2001 Sep;44(9):2065-71.

Zhao D, Banks SA, Mitchell KH, D'Lima DD, Colwell Jr CW, Fregly BJ. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *J Orthop Res* 2007