DOES THE INCREASE IN BODY WEIGHT CHANGE THE KNEE AND ANKLE JOINT LOADING IN WALKING AND RUNNING?

Claire HY Wu

Master Thesis in Biomechanics

Claire HY Wu

Spring 2015

Department of Biology of PhysicalActivity

University of Jyväskylä

Supervisors:Janne Avela and

Juha-Pekka Kulmala

ABSTRACT

Wu Han Yu, 2015. Does the increase in weight changes the joint load in ankle and knee in walking and running? Department of Biology of Physical Activity. University of Jyväskylä. Master's Thesis in Biomechanics. 72 pp

Previous obese studies showed that there is a correlation of weight to joint load in the knee joint during walking. In load-carrying studies, there are several studies but no consistent findings in the knee joint loads in walking and running.

Our aim was to investigate how the joint load changes in body weight gaining, using added vest that was 10% of the body weight to mimic the conditions. We used Vicon eight-camera system and five force platforms with inverse dynamics to calculate the peak ground reaction forces(GRFs), joint angle, joint moments, and patellofemoral contact forces(PFCF), patellafemoral stress(PFS), peak patellofemoral tendon force(PTF) and peak Achilles tendon force (ATF).

The results showed that in walking, there were significant increases in joint loadings in both knee and ankle as the weight increased. There were significant increases in PFCF, PFS, PTF and ATF ($p<0.001$). This may be due to the significant increases in knee flexion ROM, ankle dorsiflexion ROM maximum ($P<0.05$) and added weight. The relationship between the weight increase and PFCF was 1:1, in PTF 1:2, in peakATF 1:3 and in PFS 1:0.05. During running, the relationship between body weight increase and PFCF change is 1:4, in peakPTF 1:4, in peakATF 1:2 and in PFS 1:0.1. Our results indicated that the knee joint loading increased proportionally with increased weight and in running, the loading increased in double, suggesting the speed also affected the loadings. This implies that as human gains weight, there might be a higher risk of getting knee osteoarthritis, especially with one that has major recreational sports in running. Our study also suggests that weight reduction is crucial for those that are overweight to prevent osteoarthritis since weight is directly correlated to joint loads.

Keywords: body weight, human gait, walking, running, joint loading

CONTENTS

ABBREVIATIONS

GRF = Ground reaction force

GRFver = Vertical ground reaction force

GRF AP= Anterior-posterior ground reaction force

GRF ML = Medio-lateral ground reaction force

OA= Osteoarthritis

ADL= Activities of daily living

BMI = Body mass index

BW = Body weight

sEMG = Surface electromyography

N = Newtons

kN = KiloNewtons

IEMG= Integrated EMG

TA = Tibialis anterior

GA = Gastrocnemius

MPL = Multi-planar loading

AL = Axial loading

AT = Achilles tendon

COM=Center of mass

ROM= Range of motion

PFCF = Patellofemoral joint contact force

PTF = Patella tendon force

PFS = Patellofemoral joint stress

PTF = Patella tendon force

ATF = Achilles tendon force

1. INTRODUCTION

Osteoarthritis (OA) is the most common form of arthritis and with high prevalence and incidence with increase in age. Among adults in the United States, people elder than 30 years old, 6% were affected by symptomic knee OA and 3% in hip OA (Jordon et al 2000; Felson et al 1988; Yelin et al 1998). Knee and hip OA has great functional impact, and is frequently associated with disability in activities of daily living (ADL) involving lower extremity functions such as walking, transferring, and using the bathroom (Zhang & Jordan 2010). OA is a chronic disease with multifactorial etiology, which includes modifiable and non-modifiable risk factors. Modifiable risks include physical activity, diet, weight control, smoking etc. The identification of high risk subsets of the population, through risk factor stratification, may provide important insights particularly with regard to disease prevention.

Felson (1996) suggested that being overweight accounts for 33% of OA in women. Among many obese studies (Susan et al 1999; Felson et al 1988), over weight is an important modifiable risk factor in development of OA in hand, hip and knee. From the study by Susan et al (1999), after controlling for estrogen use, smoking status, height, and health care use, they found that body weight was a predictor of incident osteoarthritis of the hand, hip, and knee joint load and weight. In overweight patients that there were relationship between changes in body weight and knee joint loads, with 1:2 (Messier et al 2005) to 1:4 (Aaboe et al 2011). In another study by Felson et al (1992), within 10 years, decrease in 2 BMI decrease the development of OA by over 50%. In the hip joint and ankle joint, there were no studies that showed the relationship between changes in weight and joint loadings. Weight loss programs may prevent disease, especially in the knees, and those who are overweight are at high risk of disease progression and are likely to have a progressive disease course.

From the increase of 1:2 to 1:4 ratio of weight changes and joint loadings, weight changes is correlated with joint loadings; therefore, we would like to see whether the changes is also evident in normal healthy adults. This could answer the question of whether people gained weight after getting osteoarthritis or they gained weight then the increase in joint loads leads to osteoarthritis. It would be also intriguing to investigate the differences in walking and running to see whether the weight gain influence more on walking or running. We could also understand how 10%weight gain affected the joint loads. This study differs from other load-carrying studies since most studies (Browning et al, 2007;Holt et al, 2003) used more than 20% BW added on the subjects. The reason to select 10%BW is that it is a reasonable numbers in increase in weight. We hypothesized that there would be correlation between the weight changes and joint loadings in both walking and running, especially in the knee joints. There might also be correlation in ankle joint. The correlation might be higher in running than walking.

2. HUMAN GAITS

Human gait is the medical term for locomotion through the movement of human limbs. Human gait is defined as bipedal, biphasic forward propulsion of center of gravity of the body, where there are alternate movements of the lower extremity (Winter. 1984) . Gaits can be categorized by the speed, such as walking, jogging, running and sprinting. The followings are detailed description about walking gait cycle and running gait cycle.

2.1 Walking

Human gait in walking can be analogized as a wheel. The cyclic pattern of movement is repeated over and over. This assumption of cyclic pattern is the description for walking as a single cycle. (Vaughan. 1992) Figure 1 shows the rotating wheel of the cyclic nature of forward progression. As the wheel moves from left to right. the spoke rotates in clockwise. When the spike returns to its original position. the cycle is complete(Goswampi. 2008).

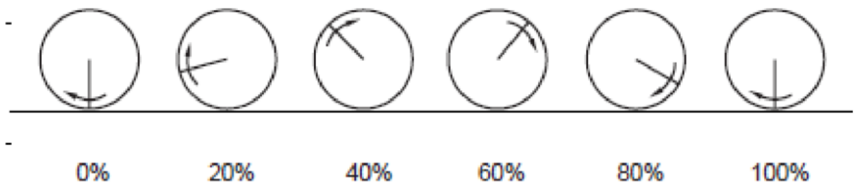

FIGURE 1. Analogy of the rotating wheel to the human gait (Goswampi. 2008)

There are two main phases in the gait cycle: stance phase and swing phase. During stance phase, the foot is on the ground, whereas in swing phase the foot is not in contact with the ground(Vaughan. 1992). Figure 2 shows that a single cycle of a normal adult and the classic and new gait terms. The cycle begins when one foot makes the contact with the ground and follow swing phase which the feet is in the air. The cycle ends with the heel that contacts the ground.

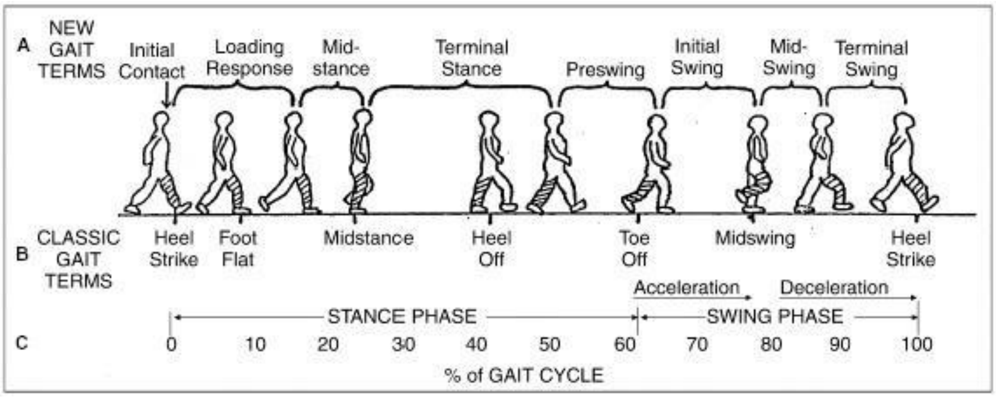

FIGURE 2. The normal gait cycle of an adult with new gait, classic gait and cycle percentage terms (Vaughan. 1992).

Because the stance phase in walking is longer than 50% of the gait cycle, there are two periods of double support when both feet are on the ground. The stance phase could be divided into three separate phase, depending whether both of the feet are on the ground: first double support (when both feet are on the ground), single limb stance (when one foot is swinging and another is in contact with ground) and second double support (when both feet are again on the ground). (Novacheck. 1998) In normal healthy adults, gait is a natural symmetry between both sides of the lower extremity; however, in pathological gait, such as osteoarthritis, hip arthroplasty or hemiplegia etc. asymmetry pattern often occurs. (Inman et al. 1953; Olney et al. 1996)

2.1.1 Muscle activation in stance phase

Muscle activity can be measured by surface electromyography (sEMG). Based on the electrophysiology of the ionic charges at the muscle fiber membrane, electrode recorded all the sum of the action potential. (Winter. 2009) Through processing of the raw EMG signal, muscle activity can be “seen” after full-wave rectification, linear envelope, integration of linear envelope and a simple binary threshold. (Figure 3) The details of processing can be referred to Winter (1979) book or other commercial sources like Noraxon, Delsys etc.

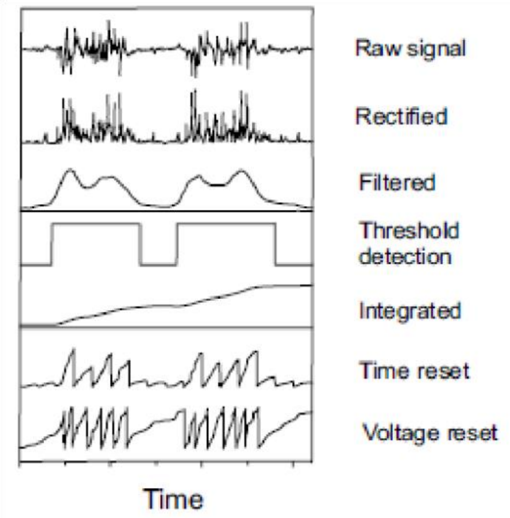

FIGURE 3. Common methods for processing the EMG signal (Winter 1979)

The central nervous system controls many muscles at the same time and figure 4 shows the muscle activation throughout the gait cycle from heel strike through stance phase towards the next heel strike (Bethodl. 1975). The figure shows only one side of the lower extremity. During mid stance phase, muscles included gastrocnemius, peroneus brevis, peroneus longus, soleus and gluteus contributes to the stability of the knee and ankle. (Hunt 2001; Schipplein and Andriacchi. 1991)

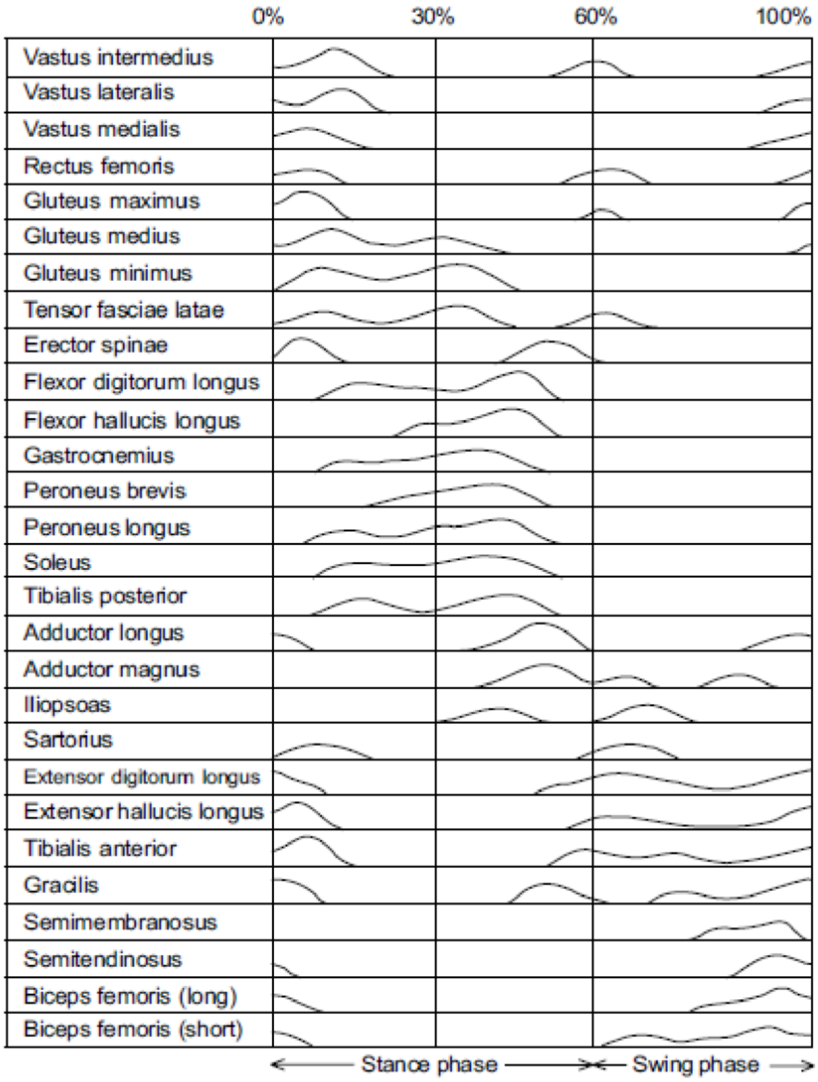

FIGURE 4. Normal EMG pattern for major muscles in the lower extremities plotted to the gait cycle of stance phase and swing phase. Muscles are listed in the order of activation in the gait cycle (Bethodl. 1975).

At the ankle joint, Hunt et al (2001) concluded that the demands on the controlling muscles are greatest prior to foot flat and after heel rise. From heel contact to 10% stance, tibialis anterior restrained foot plantarflexion, and eversion between 10% stance and foot flat. Activity in peroneus longus was consistent with its role in causing eversion after heel contact, then as a stabilizer of the forefoot after heel rise. Activity in peroneus brevis suggested a role in restraining lateral rotation of the leg over the foot late in stance phase.

2.1.2 Joint load in stance phase

The maximum resultant force across the human hip joint during walking is from 2.5 to 5.8 times body weight, measured in vivo. (Rydell. 1966; English. 1979) The load sharing between these structures gives the knee its dynamic stability because the knee is stabilized by the simultaneous action of soft tissues, dynamic muscle forces and external loads (Schipplein and Andriacchi. 1991).

Kuster et al (1997) found the estimates of knee joint loadings for normal subjects from kinematic and kinetic measures is at maximum tibiofemoral compressive force at an average load of 3.9 times body-weight (BW). Muscle forces contributed 70% during level walking whereas the ground reaction forces contributed only 30%. Morrison et al (1970) and Harrington et al (1976) showed that the tibiofemoarl joint load at 3 and 3.5 body-weight multiples respectively. Figure 5 indicates the mean tibiofemoral joint loadings of 6 female subjects from Kuster et al (1997) study.

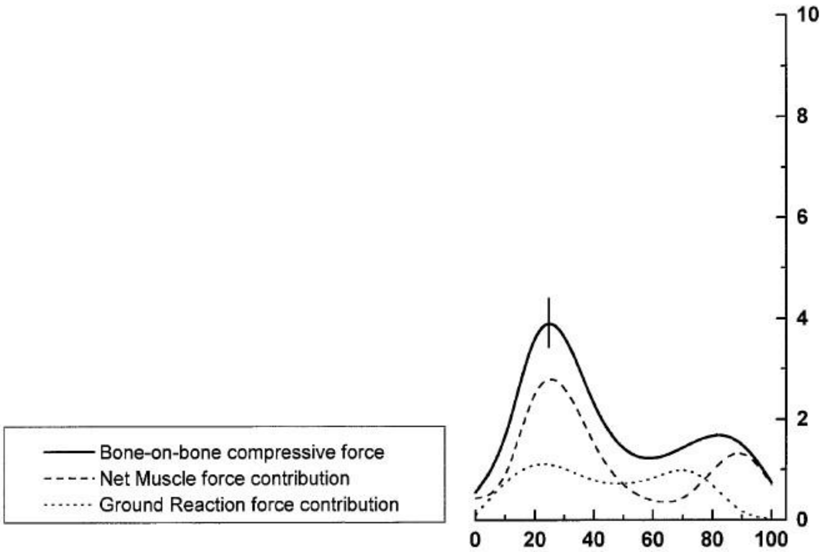

FIGURE 5. Mean tibiofemoral joint loadings of 6 female subjects in level walking. The values in Y axis are the multiples of body weight (BW) that normalized to 100% stance time. X axis is the gait cycle. Heel strike at 0% and toe-off at 100% stance phase. Bone on bone compressive force showed high correlation with the net muscle force (Kuster et al.1997).

In walking, patellofemoral joint load is in the range of 0.7 to 2 times BW (Collier et al. 1991). Recent research showed higher loadings. at 1.3 to 1.8 times BW(Nisell. 1985). In the Achilles tendon force, the joint loading was ranging from 2.6 kN to 5.6 kN in walking with different speeds (Komi et al, 1987). Figure 6 shows the Achilles tendon forces in 4 different speeds in X,Y and Z lines.

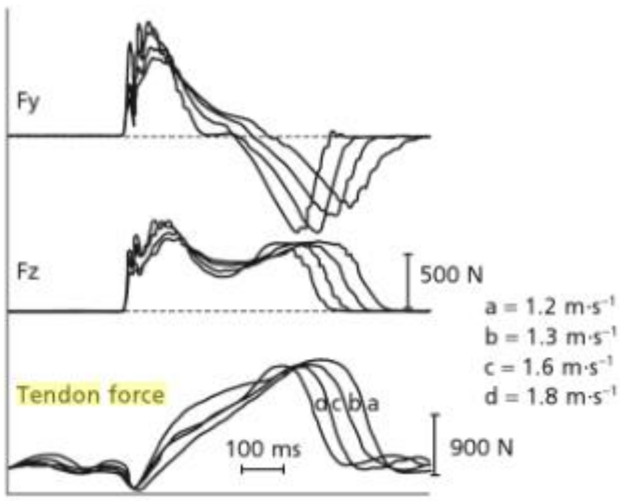

FIGURE 6. Achilles tendon force curve during walking at different speeds. In Fz and Fy, the upward increase in the curve reflected the ground reaction force in the beginning of the locomotion (Komi et al, 1987).

2.2 Running

The gait cycle in running is similar to the gait cycle in walking. There are no periods when both feet are in contact with the ground. In other words, there is no double stance phase. (Rodgers 1988) In running, toe off occurs before 50% of the gait cycle is completed. Both feet are instead airborne twice during the gait cycle, one at the beginning and one at the end of swing, which is referred to as double float. The timing of toe off depends on the running speed. As the speed increases, stance phase shortens. The length of stance phase progressively decreased from 62% for walking to 31% for running (Novacheck 1998).

Harrison et al (1987) reported at running, maximum ankle-joint reaction force at 8.97BW for compressive forces and 4.15BW for shear forces. The vertical reaction force was about 2.5 to 3 times larger in running compared to walking (Rodgers 1988).

In Subnotnick (1985) study, they showed 2 to 5 times BW at different phases of running. At initial contact, there is a large spike in force, later gradually decrease then builds up to 3 times BW in push off (propulsion) phase. Variables that affect the vertical GRF are the touchdown velocity of the heel, position of the foot and lower leg angle before contact.

2.2.1 Muscle activation in stance phase

Studies showed that EMG activity increases with running as compared with walking. Miyashita et al (1971) reported that integrated EMG (IEMG) activity of the TA and GA increases exponentially with increasing speed. Ito et al (1985) reported that with increasing running speed, the IEMG increased during swing but remained the same during the stance phase.

As the speed increases, EMG in quadriceps muscle group and hamstring group increase. The calf muscles that normally function during the mid-stance phase in walking became a late swing phase muscle and were active through the first 80% of stance phase. (Mann. 1980) Figure 7 shows the muscles activity in running of hamstring, hip extensors, rectus quadriceps, calf muscles and anterior tibialis.

FIGURE 7. Muscle activity is shown in solid bar related to gait cycle. At initial contact, there is greater number of active muscle groups and at toe off there is a lack of muscle activation (Mann. 1980).

In the beginning of the stance phase, muscles are most active in anticipation and just after foot contact. The quadriceps and rectus femoris appear from late swing to mid-stance for ground contact and absorb the shock from the impact. The hamstrings, hip extensor and calf muscles act to decelerate the momentum of the tibia before initial contact. The anterior tibialis provides clearance in swing (concentric contraction) and eccentric control the lowering of the foot on the ground in stance phase. (Novacheck 1998)

2.2.2 Joint loads in stance phase

During running, in hip joint, Cole et al (1996) reported loading of 49.8 ±22.4BW. In the knee joint, the peak patellofemoral joint compressive force normalized to subject body weight was 5.6 ± 1.3 BW with self-selected speeds (Flynn et al, 1995). Kulmala et al (2013) reported 5.1 ±1.1BW in patellofemoral compressive force and 13.0 ±2.8 in patellofemoral stress.

In Cole et al (1996) study, in running, the mean and standard deviation of the rate of joint loading were 49.8 ±22.4 BW at the ankle. There was considerable variability loading of the joints between subjects, this may be due to individual differences in activating the muscles in the lower extremity in running (Cole et al, 1996).

In Giddings et al (2000) model, the predicted peak loads for the Achilles tendon was 7.7 BW during running. The total joint contact, Achilles tendon, and plantar fascia and plantar ligament forces all contribute to the loads on the calcaneus during gait.

3. USING INVERSE DYNAMICS TO CALCULATE JOINT MOMENTS IN GAIT ANALYSIS

3.1 Basis of inverse dynamics approach

Inverse dynamics is an analysis techniques often used in biomechanics and gait labs. The aim of inverse dynamics is to determine forces or torques needed to produce kinematic motions. (Winter 2009) Human movements can be captured with motion capture system of the body and limbs, such as Vicon, OptiTrack and MoCap etc. Force plates record the ground reaction forces (GRFs). Inverse dynamics use the link-segment model that includes foot, shank and thigh segments with joints located at the ankle, knee and hip. In this approach, in order to obtain the external joint moments, the lower body segments were modeled as rigid bodies connected by pin joints. Embedded principal coordinate systems were assigned to the pelvis, thigh, shank and foot segments. The mass, inertial parameters and the mass center location were driven by a model described by Hinrichs (1985).

Joint moments can be calculated using GRFs. the path of center of pressure, inertial forces and body segment weights and simple dynamics principles (Newton-Euler equations). The equations are known as follows (Vaughan et al. 1992).

Newton (linear): F = M x A (Force = mass x linear acceleration)

Euler (angular): M = I x α, (Moment = mass moment of inertia x angular acceleration)

Figure 8 shows the inverse dynamic approach in the rigid body from the four components in the movement chain---electromyography, anthropometry, displacement of segments, ground reaction forces. Figure 9 shows the mathematical symbols. In order to understand human gait, the main key is integration---integration of different components that help researcher to obtain a deeper understanding of gait. Many gait labs and analysis measure one, two or four of these components. The net moment and powers about the joints can be reconstructed from these components.

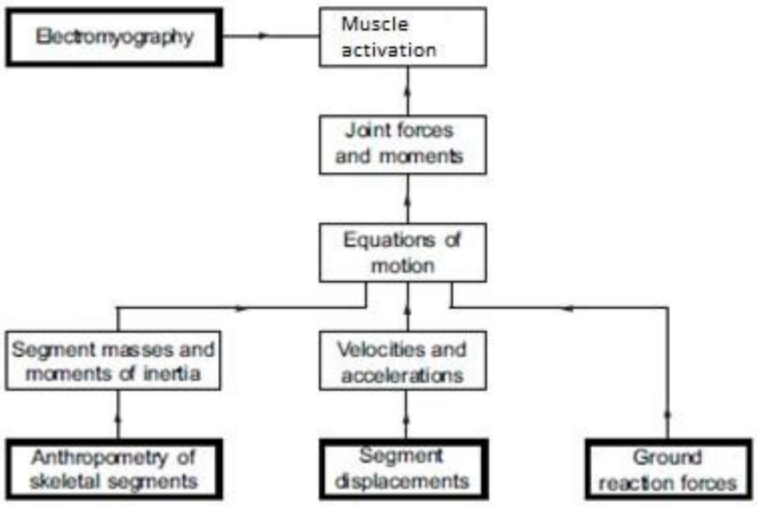

FIGURE 8. The components of inverse dynamics from gait could calculate the joint forces and moments (Vaughan et al. 1992).

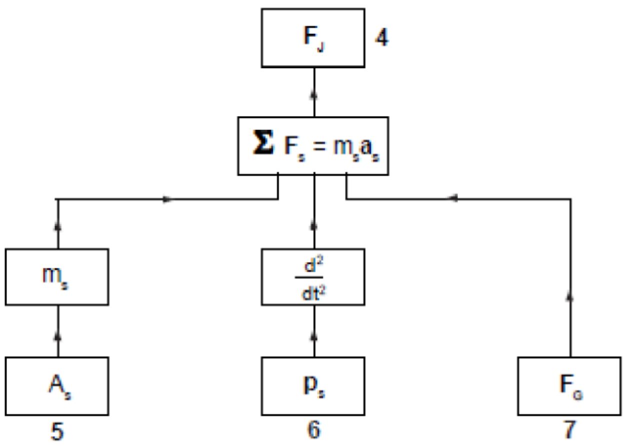

FIGURE 9. The inverse dynamic approach in rigid body expressed in mathematical symbols. A_S as the anthropometry of the body segments. Ps is the segment displacements. Ground reaction forces F_G are used with the segment masses and accelerations in the equations of motion which are solved in turn to give resultant joint forces and moments F_J (Vaughan et al. 1992).

Simulation models could be studied in 3 different versions with (Vaughan et al, 1992)
1.Rigid sagittal plane model with the total body mass in four segments (trunk, thigh, lower leg, foot). Muscles forces and soft tissue movements were not included
2. Rigid model with muscles: masses muscles group were added
3. Non- rigid model with muscles: the model incorporate muscles and soft tissue movement

The rigid body model is based on several assumptions (Winter. 2009):

1. That the joints are frictionless
2. There is generalized. Uniform and/or concentrated mass distribution
3. There is no co-contraction of agonist and antagonist muscles
4. The air friction is minimal

The principal moments of inertia of the thigh, lower leg and foot were calculated based on cadaver data. Figure 10 shows the four skeletal segments and nine soft tissues segments in the model.

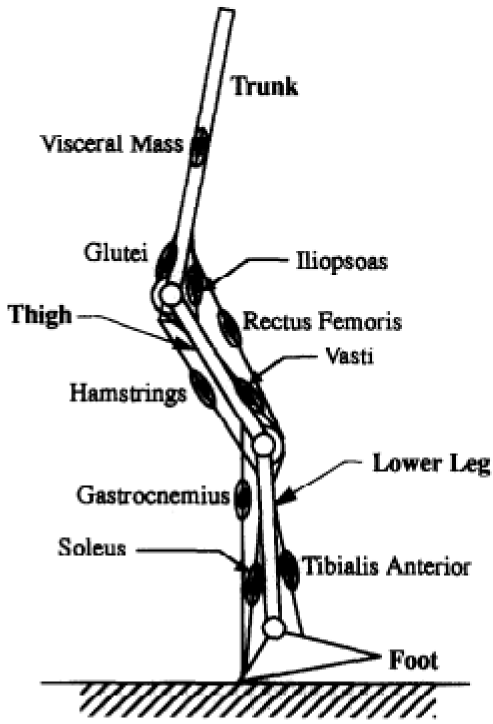

FIGURE 10. The model is based on the four skeletal segments and nine soft-tissue segments (Zatsiorsky, 1983)

From the study be Van Eijen (1985), the non-rigid model was the best model to predict the external loading in the hip, knee and ankle. In rigid sagittal plane model, the patellofemoral joint is assumed and simplified to a 2-dimensional mechanism in the sagittal plane. Femur and patella are considered to be 2 rigid elements. Friction between the articulating surfaces is neglected. The model applies primarily to static situations(Figure 11) (Radin and Paul. 1972). However, with multiplane movement and taken into account of muscle forces in different directions, different models had been assumed to calculate the mechanism between the femur and patella (Powers et al. 1998) Figure 12 shows the studies that estimate the multi-planar loading (MPL) and axial loading (AL)at knee flexion 0 and 90 degrees. From Powers et al (1998) study, axial loading of the extensor mechanism underestimates contact pressure at 0” and overestimates contact pressure at 90” of knee flexion when compared to multi-plane loading. Additionally, loading of the individual vastius intermedius appears to have an effect on patellar kinematics.

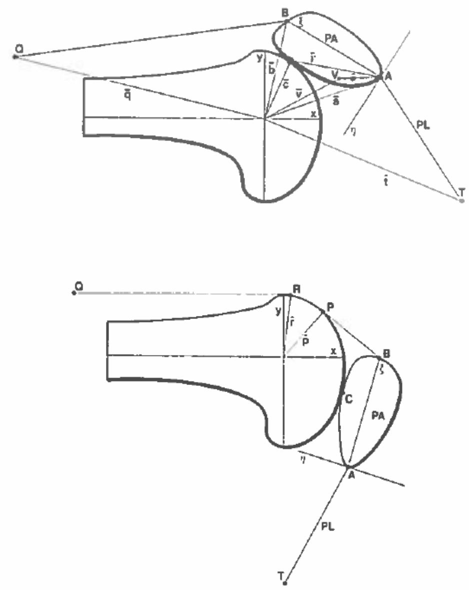

FIGURE 11. Simple model of the patellofemoral joint: model of the patellofemoral joint in knee flexion (Radin and Paul. 1972).

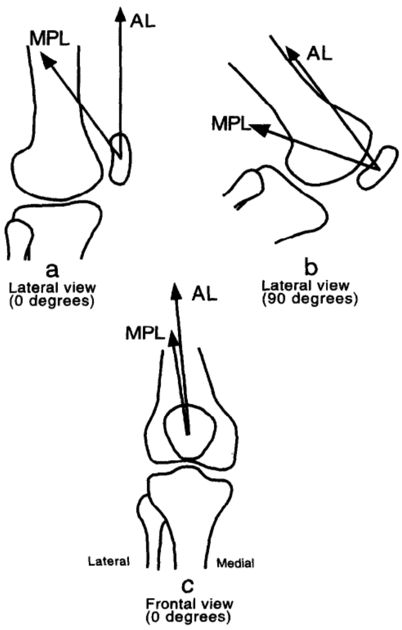

FIGURE 12. Estimation of the direction and magnitude of the resultant force vector on the sagittal and frontal plane of both the multi-planar loading (MPL) and axial loading (AL) at knee flexion 0 and 90 degrees (Powers et al. 1998)

3.2 Limitations of inverse dynamics approach

The limitation of inverse dynamics is mostly from the assumptions of the technique that there may be friction at the joint (e.g. in arthritis) and the distribution of mass in the segment is not uniform and certainly not concentrated at one point and estimating the joint center of rotation is inclined to error (Holden & Stanhope. 1998)

1. The typical models (e.g. Helen Hayes) used rely heavily on anthropometry to define the hip joint center
2. The joint center of rotation may also (and often does) move during motion. especially at the knee (Manel et al. 2000)

3. Some models (e.g. Cleveland model and six degree of freedom model used at National Institute of Health using marker triads) make less assumptions in this respect measurement error (Holden et al. 1997)

4. the worst of these tends to be inaccuracies in co-alignment of the force platform and motion analysis system (Gill and O'Connor. 1997)

For example, there may be marker motion on the skin, especially "wand" type markers on sticks, motion at the skin-bone interface and marker tracking is sometimes contaminated by errors due to interpolation when markers go missing and data from some frames is lost. Also, body segment parameters (anthropometry) are approximations and generalizations. In very thin or overweight people, children and patients with atrophied limbs may have different proportions. (Jansen, 1989) Inverse dynamics can only determine the net moment and power, co-contraction of antagonistic muscles will cancel out, such as in spastic conditions such as cerebral palsy and stroke.

3.3 Other methods of calculating joint loads

The methods available to estimate joint loading in vivo include cadaver measurement and modeling. (Colet et al, 1996) In cadaver measurement, human legs or animal models were taken into measurement. In modeling, different models are taken into account to simulate the in vivo condition in movements. Simulation model provides a good understanding of different mechanisms including muscles, ligaments and other soft tissues that influence joint loading.

3.3.1 Cadaver measurements of patella tendon force

Motion simulators are created to simulate motions for human and animal knee cadavers to measure the forces. Figure 13 and 14 show different motion simulators that study the knee flexion and joint loading in different activities. (Takano et al, 2009; Chao et al, 2007)

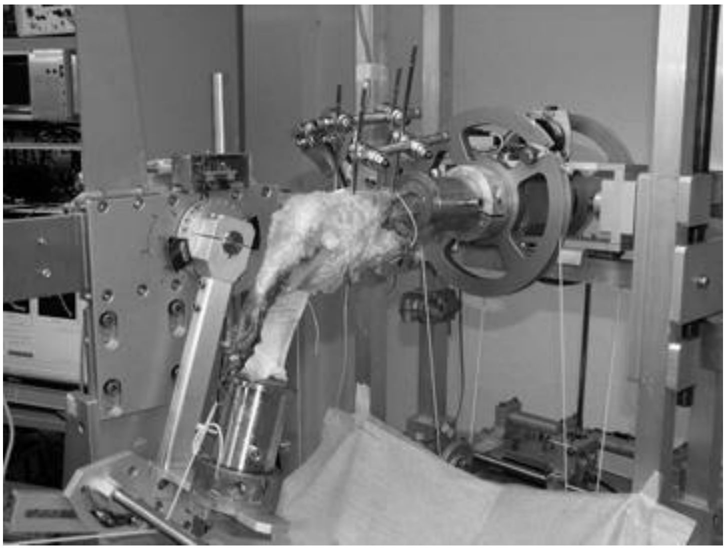

FIGURE 13. This is the knee motion simulators. The knee joint is accurately and repeatedly moved from 0 to 180 flexion angle in six degrees of freedom. The tibiofemoral and patellofemoral motion and joint reaction forces can be continuously measured. The similar force vector is produced by two in assumption of the muscular forces is mainly generated with the quadriceps and rectus femoris muscles (Takano et al, 2009).

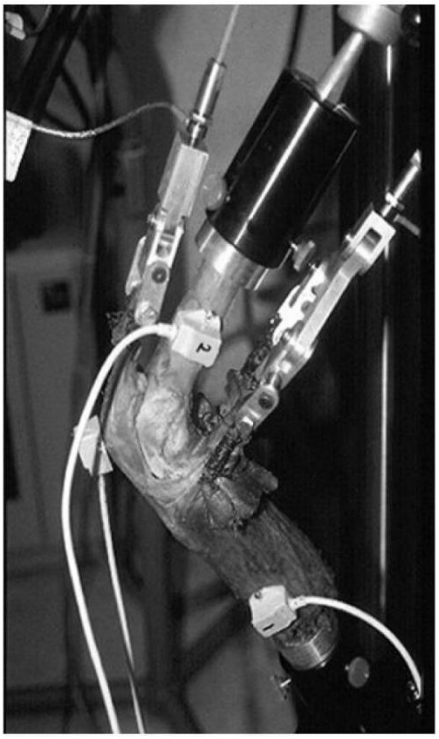

FIGURE 14. The simulator is to study of knee flexion and joint loading under squatting activity. Independent loads are applied to the simulated hip joint, the medial and lateral hamstrings tendons and the quadriceps tendon using hydraulic actuators. The tendons are secured to the loading actuators using cryo-clamps (Chao et al, 2007).

In Van Eijen et al (1986) model, morphological parameters were obtained from later view radiographs of autopsy knees in different flexion angles (Figure 15).Radiographs and macroscopic examination of the specimens were measured from cadavers from the human legs. Identified in the radiographs, the patellar ligament length, patellar length, moment arm length and the femoral condylar width were measured. (Insall & Salvati, 1971) The femur was fixed in a clamp that the lower leg could move freely in the horizontal plane. A force parallel to the femoral shat was applied to the quadriceps tendon to ensure the patella was firmly pressed in all conditions of knee flexion. During flexion, the direction of the quadriceps tendon relative to the femur changed, thus, the moment arm changed as well.

Prior to clamping, muscles were wrapped in saline moistened gauze to prevent drying during testing, and to evenly distribute the force of the clamp through the tissue. Clamping was performed such that the muscle fibers were perpendicular to the clamp itself. This ensured even loading of all muscle fibers. The vastus intermedius and rectus femoris were clamped together since the direction of the resultant force vectors of these muscles with respect to the patella are similar. (Powers et al. 1998)

Muscle clamps were attached to the loading cables so that only tensile forces were applied. This assisted in preventing the muscle tissue from pulling apart during loading. However, ligament, medial and lateral articular facets, articular cartilage deformation has not been considered in the model by Van Eijen; therefore, the joint pressure distribution cannot be assumed to increase with the compression forces. Direct measurements of internal loads usually require the dynamic knee simulator and limited numbers can be investigated in human cadavers.

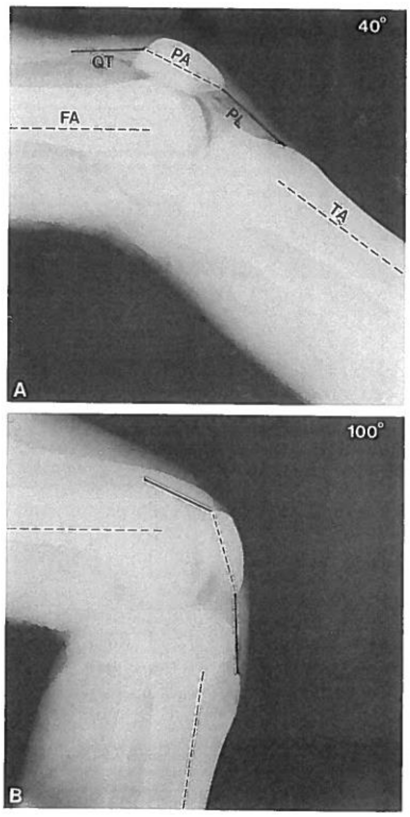

FIGURE 15. Lateral view radiographs of one knee at knee flexion 40° and 100° (Powers et al. 1998).

3.3.2 Cadaver measurement for patellofemoral stress

The patellofemoral stress is calculated from the patella tendon force divided by the contact area between the patellofemoral joint. The function of joint incongruity is to allow the articular surfaces to come out of contact at light loads so that the cartilage may be exposed to synovial fluid for the purposes of nutrition and lubrication. (Greenwald& O'Connor, 1971) At large loads, the distribution of cartilage thickness ensures that a state of hydrostatic pressure is achieved in order that cartilage, with a large fluid content, may transmit large pressures without flow and consequent loss of its integrity.

The contact area within the tibiofemoral joint was first investigated using radiography methods (Kettelkamp and Jacobs, 1972) and casting techniques (Walker and Hajek, 1972). These studies helped identify the meniscus as load distributing structures of the knee.
The casting method used silicone rubber (Dow Corning, 3110 RTV4) in liquid form is poured around one bone of the joint, followed immediately with load application (Stormont et al., 1985). The joint is then kept under load until the silicone rubber sets (setting time around 4 minutes). After removal of the load, contact areas can be determined from the location and size of holes in the rubber cast.

Fuji film method was to insert Fuji film through bilateral rectinacular incisions directly between the articulating surfaces of the joint by retracting the patella. Knee extensor forces were produced by stimulation of the femoral nerve while the tibia was held in position by a restraining bar. The resulting patellofemoral joint contact pattern was recorded directly onto the Fuji film. This technique has enabled comparison of anterior cruciate ligament intact and transected joints as well as early stage osteoarthritic and contralateral joints. Changes in the patellofemoral contact profile shape and location relative to the retropatellar surface resulting from anterior cruciate ligament transection were observed. Strips of sealed film were prepared 1×10 cm^2 to allow multiple measurements per strip. The strips were inserted into the patellofemoral joint in an anterior/posterior direction, so that the film had to negotiate only one curvature and crinkle artifacts could be avoided. The patellar cylinder can be seen in figure 16.

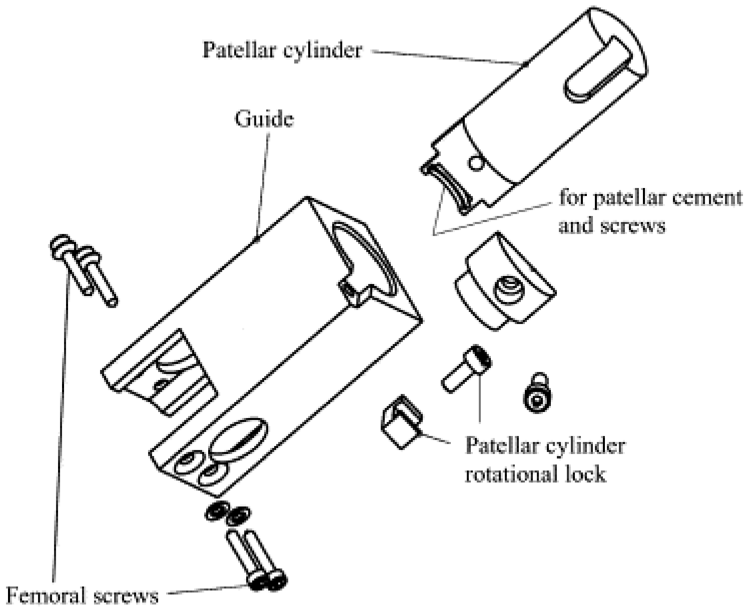

Figure16.The patellar cylinder for the measurement of contact is on the Fuji film. The direction of force application along the cylinder being approximately perpendicular to the patellofemoral contact area (Fukubayashi and Kurosawa. 1980).

Fukubayashi and Kurosawa (1980) combined the casting method with Fuji pressure sensitive film to measure both contact area and pressure distribution in the tibiofemoral joints of degenerated and healthy knees with and without meniscus. Degenerated knees were found to have a larger contact area when compared to healthy joints and the pressures experienced by the cartilage increased significantly when the meniscus was removed.

Ahmed and Burke (1983) and Ahmed et al. (1983) measured static pressure distribution in the tibio- and patello-femoral joints using a micro-indentation transducer. The stress distribution within the patellofemoral joint was found to be dependent upon the degree of action of the various components of the quadriceps muscle group. When using any technique to measure contact area and pressure distribution in an articulating joint, its disruption of the natural contact mechanics of the surfaces must be considered

3.3.3 In vivo measurement of Achilles tendon force

The loading of Achilles tendon has been characterized by the magnitude of the peak ATF. (Komi et al, 1986). Direct measurement of in vivo tendon and ligament loads remains a challenge with ethical issues and technical difficulties. However, there are advantages for direct in vivo measurement, it provides continuous recordings that is immediately available for inspection. Also, several experiments can be performed in one session and the movements are natural. (Woo et al, 2008)

3.3.3.1 Buckle transducer

Buckle transducer (Figure 17) initiated with animal experiments, which took into account many factors: transducer design, surgical operation procedures, and duration of implantation. The first human experiment utilized an E form transducer implanted around the AT under local anesthesia. (Komi et al, 1987) The transducer was kept in situ for 7 days, and on the 8th day recordings were made on simple plantar flexion movements and during slow walking. Later development of the measurements can be made immediately after operation and the measurement lasted 2-3 h. The transducer was immediately removed after the measurement.

The AT transducer can be calibrated by placing the subject in a prone position on to a calibration table. A pulley system with calculated weight was used to dorsi-flex the ankle. The measurement usually encompassed the EMG recordings, force platforms or oscillating ergometers to investigate the Achilles tendon force in different movements Taking into consideration the geometrical arrangement of the AT transducer, axis of rotation, and the pulley system, the exact values of AT forces could be calculated. This method is quite invasive and may receive objections from the ethical committee. There are not many tendons that could be selected for measurements since the size of the buckle is large. However, the method produces important peak to peak force and rate of force development that can describe the loading characteristics of the tendon in normal locomotion.

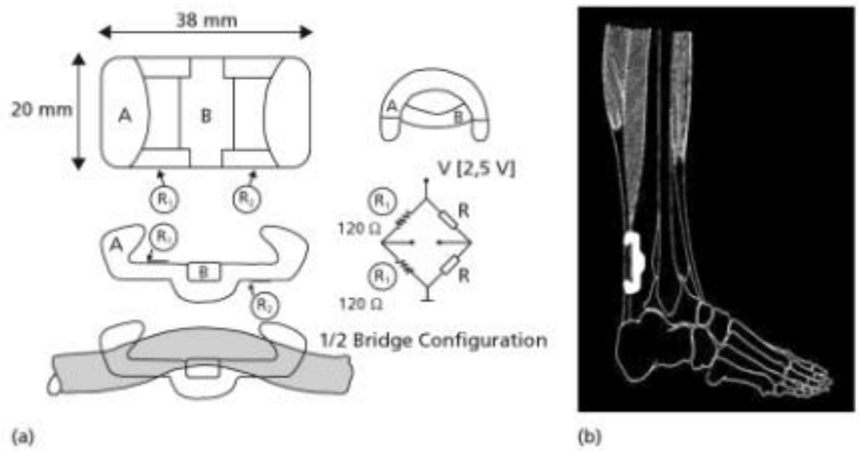

FIGURE 17. (a) schematic presentation of the buckle transducer in human subjects that investigated running, jumping and different vigorous conditions. A is the main buckle frame. B is the cross-bar. The lower part is in situ with the bending of the Achilles tendon. (b) Schematic presentation of the buckle transducer implanted around the Achilles tendon (Komi et al, 1987)

3.3.3.2 Optic fiber technique

The optic fiber technique (Figure 18) as a transducer is based on the light intensity modulation when the fiber is compressed inside the tendon. (Komi et al, 1996) Power from battery supplied to the light emitting diode. When the light entered the photodiode receiver, it could be converted to analog signal and later to the force signal with the ankle ergometer (Nicol et al, 2003). Movements could be performed with the insertion of the optic fiber. Before insertion of the optic fiber, anaesthetic cream was put on the skin. Needle with the fiber was passed through the Achilles tendon. (Finni et al, 1998) This technique has been used in normal walking (Ishikawa et al, 2005b), running (3-5 m/s) and long jump (Kyrolainen et al, 2005), hopping (Finni et al, 2001a), isometric plantarflexions (Arndt at al, 1998), passive dorsiflexion stretches (Nicole & Komi, 1998) and submaximal squat jump and countermovement jump (Finni et al, 2000). The technique could be reproduciple provided the research team has enough experiences in the application. (Komi et al, 2008)

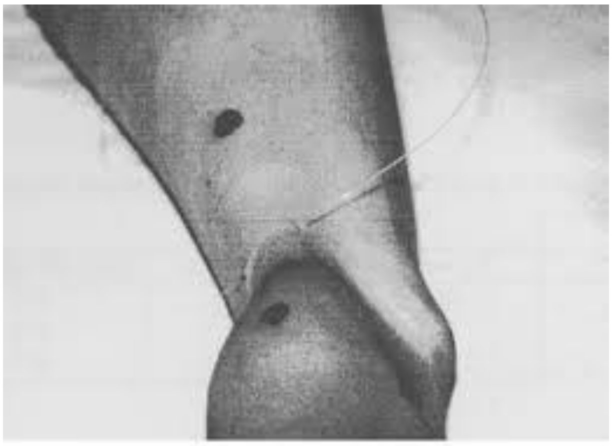

FIGURE 18. Optic fiber was inserted in the Achilles tendon, 3 centimeters above the calcaneus (Arndt et al, 1998).

4. LOAD-CARRYING BIOMECHANICS

4.1 Kinematic and kinetic changes

In walking, several investigations have reported no significant differences in sagittal plane joint kinematics while carrying external loads that range from 10% to 64% of an individual's body weight (Browning et al, 2007;Holt et al, 2003). Other reports have found an increased amount of knee flexion and ankle dorsiflexion while carrying external loads that range from 20% to 40% of the individual's body weight (Kinoshita et al, 1985; Quesada et al, 2010). Harman et al (2000) found an increase in maximum knee angle in loading response phase. In Tilbury-Davis and Hooper study (1999), load carriage up to as much as 64% of subjects' body weight has little effect on sagittal plane gait motion.

Kinoshita (1985) and Harman (1992) have reported differences in the ground reaction forces in horizontal and vertical and lower limb joint forces are increased in load-carrying.With load carrying, knee excursion and compression of the center of moment did not change in the stance phase. There were significant increases in joint angles in the sagittal plane at the hip, knee and ankle.

The majority of research suggests that the increase in vertical and anteroposterior GRFs is directly proportional to the applied load. These studies suggest that 1 kg of added load equates to approximately a 10 N increase in force. (Tilbury-Davis and Hooper, 1999; Kinoshita et al, 1985; Harman et al, 2000) When the weight of the pack is included in the calculations of ground reaction force as a percentage of total weight (body + pack), the mean values are almost identical showing that the increase in ground reaction force is proportional to the pack weight.

Peak forces change with loads as low as 20 kg and the forces necessary for balance increase significantly when any load is carried. Higher loads do not cause further significantly increases in impact forces. (Tilbury-Davis D.C, R.H. Hooper, 1999).

However, there is a suggestive threshold of 40 kg in load-carrying, the increases in propulsive and vertical impulses increase significantly (Kinoshita 1985). Below this there are no significant differences between the loaded and loaded conditions. When the load exceeded 40 kg, a compensating mechanism of reduced speed was initiated for the stance limb, giving protection from the effects of excess loading.

Training (or task experience) may influence the kinetic and kinematic responses to carrying loads (Littlepage, Robinson & Reddington, 1997; Vasta,Rosenberg, Knott & Gaze, 1997). In clinical weight vest study by Griffin et al (2003), about 7 to 10% of the subjects' body weight improved static body balance.

There are very little studies regarding load-carrying in running. To our best knowledge, there are only 2 studies on this. The vertical ground reaction forces increased linearly with gait speed up to about 60% of the subjects' maximum speed. At higher speeds, vertical forces remained constant at approximately 2.5 times body weight (Keller et al, 1996). Changes in the ground reaction forces have found to be insignificant in Schiffman (2004) study.

4.2 Muscle activity changes in load carrying

In load carrying studies by Ghafari et al (2009), muscle activity in hip extensors (gluteus maximus, hamstring) and knee extensors (rectus femoris) increased during the stance phase. Ankle plantarflexors muscle activity are increased as well, in which Park et al (2014) showed higher EMGs of the medial gastrocnemius, Stephens & Yang (1999) showed increase in soleus muscles. Previous analyses of gait (Shelburne et al, 2006) have found that the muscular stabilization against external adduction moments is predominantly provided by the quadriceps during early stance and the gastrocnemius afterwards.

EMG of quadriceps muscles in load-carrying are mixed in findings, with these muscle groups showing either no increase (Harman et al, 1992; Norman, 1979; Park et al, 2014), or significant increase with load. (Knapik et al, 1996) Under heavier backpack loads, the gastrocnemius muscles are more active than in normal walking. The increase in EMG activity with load is pronounced when the load mass exceeds 30-40 kg (Harman et al, 1992; Norman, 1979).

5. OVERWEIGHT AND OSTEOARTHRITIS

Overweight people are at high risk of osteoarthritis in knee and also in the hips. (Felson. 1996). Although obesity is strongly related to knee OA, some adults adapt to excessive weight and reduce knee-joint torques, knee-joint forces in walking (Messier et al. 2005) Nevertheless, biomechanical joint stress could be the pathogenesis and progression of knee OA (Felson. 1988)

In Messier et al study (2005), there was a significant direct association between follow-up body mass and peak follow-up values of knee compressive force, resultant force, abduction moment and medial rotation moment. Reduction of 9.8 N (1 kilograms) was associated with reductions of 40.6 N in compressive forces and 38.7 N in resultant forces. From the study, there was a 1:4 ratio between weight loss and joint load. In Aaboe et al (2011) study, every 1 kg in weight loss, the peak knee load was reduced by 2.2 kg. Thus, 1 kilograms reduction in body weight was related to more than twice the reduction in peak knee force at a given walking speed.

The external knee adduction moment is defined as the torque that tends to adduct the knee during gait. It is related to the distribution of forces between the medial and lateral compartments of the knee joint. Increased external knee adduction moments are indicative of increased loads on the medial compartment of knee. (Schipplein and Andriacchi. 1991)

Messier et al (2005) showed that after weight loss, there was significant direct association between decreases in internal abduction moment at the knee, which contributed to the lower joint loads at knee. The result was as Schipplein and Andriacchi had proposed. (1991) However, there was no direct relationship between weight loss and knee flexion/extension internal moments. In previous study by Schipplein and Andriacchi (1991), it was found that in OA patients, comparing with healthy adults, there were higher knee peak external extension and flexion moments for knee stability.

In Aaboe et al (2011) and Messier et al (2011) study, 10% weight loss in an overweight and obese osteoarthritic population showed positive changes in the lower knee joint compressive loads during walking compared to low (5% weight loss) and no weight loss groups. The decrease in knee joint compressive forces was mostly because of reductions in hamstring co-contraction during the initial portion of the stance phase.

Excessive subtalar pronation is also common in obese adults (Messier. 1994). Obese gait was characterized by a greater magnitude and rate of rearfoot eversion when compared with non-obese subjects. There is a great association between rearfoot eversion and higher BMI values. (Wearing et al. 2006) The study also speculated that the excessive rearfoot movement associated with obesity might place additional strain on musculotendinous structures of the lower limb, therefore increasing the possibility of injury. Weight loss in this obese population results in a reduction in the lateral rotation moment.

A study showed that the risk of hip osteoarthritis in BMI 30-35 was twice as great as the risk in those with BMI<25. (Juhakoski et al. 2009) High percentage of patients with end-stage hip OA are overweight, including younger adults and those with symptoms of 3–6 months' duration. (Marks et al. 2002) Nevertheless, there are no studies that show the relationship between the reduction in weight and reduction in hip joint load.

6. PURPOSE OF THE STUDY

Overweight has been reported as one of the most important risk factors for knee OA. During walking, the relationship between weight gain and joint load in knee joint varied in 2 studies, between 1:2 (Messier et al 2005) to 1:4 (Aaboe et al 2011). In the hip joint and ankle joint, there are no studies that showed the relationship between joint load and weight. Most obese studies focus on the knee joint, since the epidemiology of osteoarthritis is highly related to obese. The study (Juhakoski et al 2009) also showed that between BMI over 25, the joint load significantly correlated with the weight loss. In load-carrying studies, there are inconsistent results in the joint loads in walking and running, some found a proportional increase in increased carry loads and joint loads, some found no correlations.

Therefore, the purpose of the study is to investigate the relationship between 10% body weight increase and joint loadings in knee and ankle in walking and running. We used several tools to quantify joint loadings such as joint moments, patellofemoral tendon force(PTF), Achilles tendon forces(ATF), patellofemoral compressin forces(PFCF), patellofemoral stress (PFS). This could give a possible insight into how the weight changes affect the knee and ankle moments, PTF, ATF, PFCF and PFS. Our hypothesis is that there will be correlation between weight changes and those parameters in healthy populations in both walking and running.

7 METHODS

7.1 Subjects

Eleven male subjects (mean ± standard deviation. mass 76.32±8.97 kg, height 177.10±7.97 cm, BMI 24.29±1.77) voluntarily participated in the study. They all provided informed consent and were performed in accordance to the Helsinki declaration. The subjects were allowed to withdraw from the study at will.

The subjects performed shod running along a 6.2-m track at preferred speed. They perform shod running with vest at 10% of their body weight at preferred speed on the 6.2 m track. Later the subjects walked along the track at preferred speed.

7.2 Experimental procedures and measurements

The experimental setup is shown in figure 19. The filming took place in an indoor running track. Eight-camera system (Vicon T40. Oxford. UK) and 5 force platforms, total length 5.7 meters (AMTI. Watertown. MA. USA) were used to record marker positions and GRF of 3 directions synchronously, including anteriorposterior, mediolateral and vertical GRF. Eight-camera system recorded at 300 frames per second while GRF data recorded at 1500 frames per second. Each subject had 15 (2.5 cm diameter) spherical reflective markers attached to the lower body. Bilateral placement of 14 retroreflective markers (table 1) and 1 sacrum marker were carried out according to Plug-in Gait lower body model (Vicon. Oxford. UK). Figure 20 and 21 show the plug-in gait lower body model from the back and lateral view. Anthropometric measurements such as height, weight, leg length, and knee and ankle diameters were taken before the experiment.

Two photocells were set up at the beginning and end of the track to control the velocity between trials for running at 3.5-4.5 m/s and walking at preferred speed. Subjects ran without vest then with vest at 10% more of their body weight. The walking trials were also conducted without vest and with vest (Figure 22).

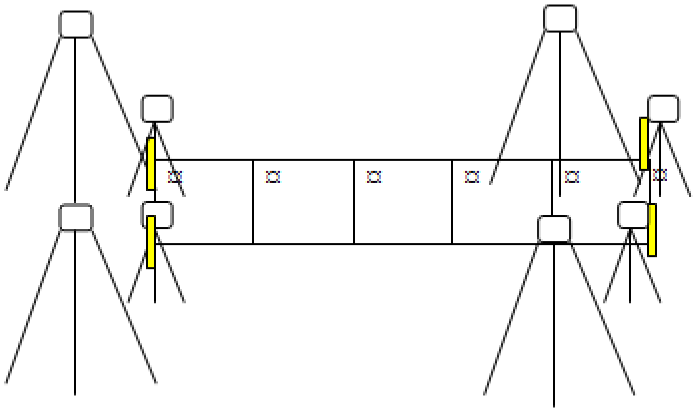

FIGURE 19. Eight-camera system. 2 photocells and 5 force platforms set up

TABLE 1 Names of the marker placement

NUMBER	MARKER PLACEMENT
1	right second metatarsal head
2	right posterior calcaneus
3	right lateral malleolus
4	right tibial tubercle
5	right femoral epicondyle
6	right greater trochanter
7	right anterior superior iliac spine (ASIS)
8	left second metatarsal head
9	left posterior calcaneus
10	left lateral malleolus
11	left tibial tubercle
12	left femoral epicondyle
13	left greater trochanter
14	left anterior superior iliac spine (ASIS)
15	sacrum

FIGURE 20. Plug-in gait lower body model from the back view

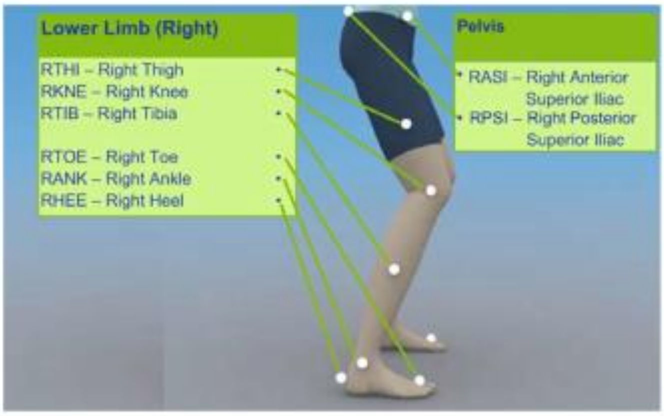

FIGURE 21. Plug-in gait lower body model from the later view

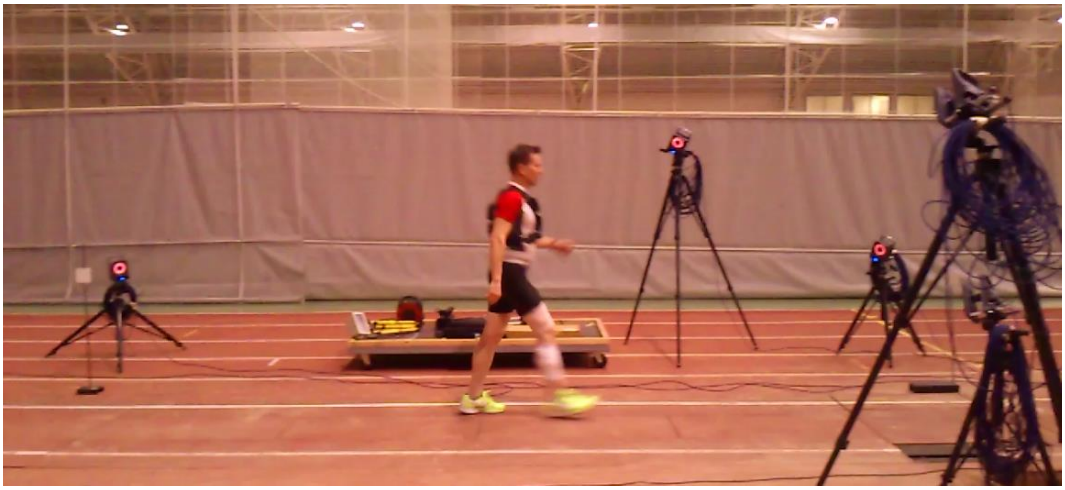

FIGURE 22. One of the subjects was performing a walking trial with vest

7.3 Running and walking analysis

Marker trajectories and GRF data were respectively low-pass filtered using a fourth-order Butterworth filter with cutoff frequencies of 12 and 50 Hz. Three successful ground contacts of the right leg were selected for the analysis. GRF data were exported into the Signal software v.4.1 (Cambridge Electronic Design. Cambridge. UK).

Plug-in Gait model (Vicon Nexus v1.7. Oxford Metrics) was used for kinematic and kinetic analyses. During initial ground contact and toe-off, calculation of the position of the center of mass (COM) was carried out to calculate cadence, step length and width and COM–heel distance in the anterior–posterior direction. Inverse dynamics approach was used to calculate hip, knee and ankle joint angles (range of motions) and internal joint moments (N/m/kg) during the stance phase of running and walking were determined across three successful force plate contacts of the right leg.

The kinematic and kinetic data were timely normalized(0%-100%) and averaged across several or within one trial in the Polygon software. Gait cycles were normalized to stance phase. All the joint angles and GRF data were exported to a specifically-designed Excel file (Microsoft 2010) template for processing.

7.4 Patella tendon force (PTF)

The patella tendon force was calculated based on the knee joint angle from the equation of Powers et al (1998), X indicated the joint angle

$$y = -0.0616X^2 + 6.3619X + 76.486$$

7.5 Patellofemoral joint contact force (PFCF)

A sagittal plane model of the patellofemoral joint was developed to utilize input from sagittal plane mechanical analysis of the lower extremity and patellofemoral joint contact area from the MRI. An overview of the model is in Figure 23. Input variables for the model algorithm included knee joint flexion angle, knee extension moment and patellofemoral joint contact area obtained from the data collection and quadriceps lever arm and a constant (k) obtained from van Eijden et al (19875, 1987) that used two dimentional model of the patellofemoral joint.

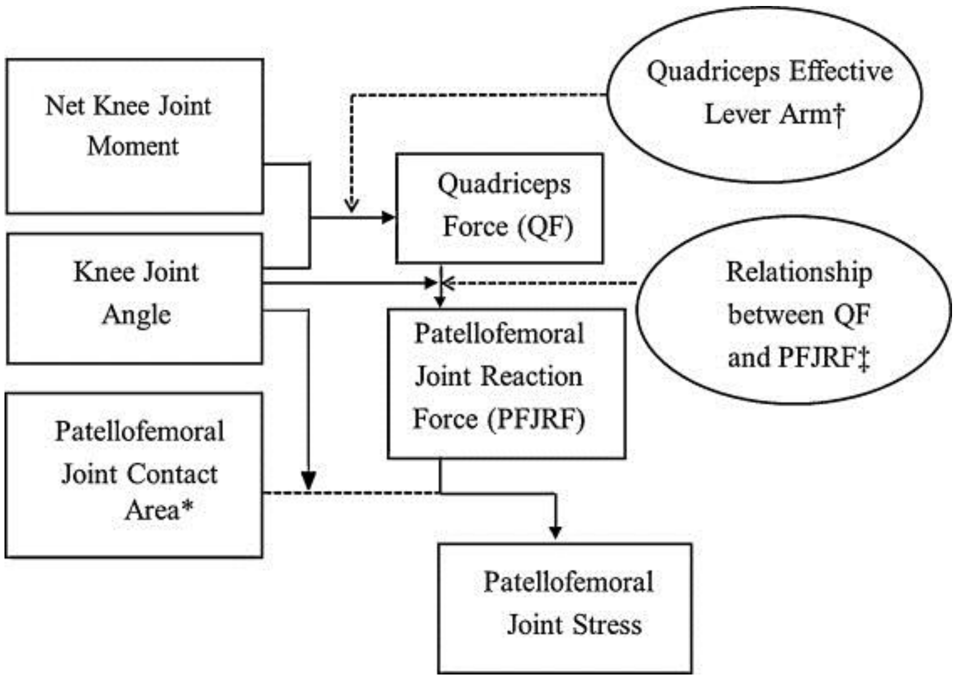

FIGURE 23. Figure of the flow chart of patellofemoral joint model. (Van Eijden et al. 1986)

Later quadriceps effective moment arm calculated according to Van Eijden -85 & 86 (A mathematical model for patellofemoral joint) the equation was below

Later quadriceps effective moment arm (x)=$(0.00008X^3-0.013X^2+0.28X+0.046)$

Quadriceps force = (knee moment/effective moment arm)

Patellofemoral joint contact force (PFCF) is the forces in the knee between patella and femur bone. The medial-lateral component of the contact force exhibited substantial variability among knees. The direction of this force (medially or laterally directed) varied among knees and, in some knees, changed direction as a function of flexion angle. The point of application on the patella of the resultant contact force migrated superiorly from 20 to 90 deg flexion. About 90 deg flexion this point tended to migrate inferiorly. Patellofemoral contact force (PFCF) could be calculated as the product of the quadriceps force (Fq) and a constant (K): PFCF=FqK

The constant K was estimated for tibiofemoral joint angle (x) from the nonlinear equation of the curve fitting to the data of van Eijden et al (1985)

$$k(x)=0.462+0.00147X^2-0.0000384X^2/ (1-0.0162X+0.000155X^2-0.000000698X^3)$$

(Bredeweg & Buist. 2011)

7.5 Patellofemoral joint stress (PFS)

Patello femoral joint stress was calculated from PFCF divided by contact area of the knee joint. According to Ho et al (2012), a polynomial curve fitting algorithm was used to estimate the contact area. The patellofemoral joint contact area estimation was based on the in vivo data of Powers et al.(1998) Table 2 shows the average pressure and contact area obtained from seven knee flexion angles with multi-planar and axial loading. Seven different contact areas (83.140.227.236.235.211.199 mm2) were reported for seven knee flexion angles (0.15.30.45.60.75.90 degrees) respectively to obtain the continuous contact are from 0 to 90 degrees. Patellofemoral joint stress was then obtained by dividing the patellofemoral joint reaction force by the utilized contact area for the knee flexion angle corresponding to the patellofemoral joint reaction force value. Possible increase in PFJ stress may be related to increased joint reaction force and decreased contact area.

TABLE 2. In different knee flexion angel the average pressure and contact area from the in vivo study (Powers et al.1998) There was no significant differences between the

axial loading (rectus femoris loaded in the frontal plane) and multiple plane loading conditions (individual components of the quadriceps loaded along their respective fiber directions in both frontal and sagittal planes)

Patellofemoral joint pressures and areas, mean (standard error)

	Knee flexion angle						
	0	15	30	45	60	75	90
Average pressure (MPa)							
AL	0.62 (0.14)	1.00 (0.10)	0.78 (0.07)	0.55 (0.02)	0.58 (0.03)	0.63 (0.06)	0.75 (0.05)#
MPL	0.75 (0.16)*	1.10 (0.09)	0.80 (0.06)	0.63 (0.04)	0.59 (0.02)	0.61 (0.05)	0.62 (0.04)
Contact area (cm^2)							
AL	0.76 (0.22)	1.19 (0.14)	1.71 (0.17)	1.95 (0.19)	2.40 (0.10)	2.12 (0.07)	2.20 (0.13)
MPL	0.83 (0.19)	1.40 (0.14)	2.27 (0.15)*	2.36 (0.24)	2.35 (0.20)	2.11 (0.13)	1.99 (0.15)

AL = Axial loading; MPL = Multi-planar loading. * indicates MPL condition significantly greater than AL condition ($P < 0.05$); # indicates AL condition significantly greater than MPL condition ($P < 0.05$).

7.6 Achilles tendon force (ATF)

Achilles tendon force was obtained by diving the plantarflexion moment (Ma. from inverse dynamics) by the estimated Achilles tendon lever arm (La), where a is the ankle angle.

$$\mathrm{ATF} = M_a / L_a$$

$$L_a = -0.5910 + 0.08297a - 0.0002606a^2$$

(Daoud et al. 2012)

7.7 Statistical analysis

The two-tailed independent T test was carried out for each participant in kinetics and kinematics between normal and vest conditions. Absolute values were calculated for normal and vest conditions in both walking and running. Paired T test was also carried out in knee and ankle joint loadings between walking and running. Statistical significance was set at a level of p less than 0.05. Pearson correlation was carried out to see the changes in body weight and changes in PFCF, PTF, PFS and ATF in both walking and running. The correlation strength is according to Dancey& Reidy (2004). Statistical significance was set at a level of p less than 0.05.

8. RESULTS

8.1 Walking

8.1.1 Stride parameters

The average speed in normal walking was 1.45±0.35m/s. The average walking speed with vest was 1.45±0.30m/s

8.1.2 Walking kinematics

In walking trials, with increasing weight by 10%, all the ROM from the lower extremity increased. There are significant increases in knee flexion ROM from 33.13±3.54 to 34.75 ±4.2 ($p<0.001$)(Figure 24) and ankle dorsiflexion ROM maximum increased from 11.83±5.02 to 15.45±5.87($p<0.001$)

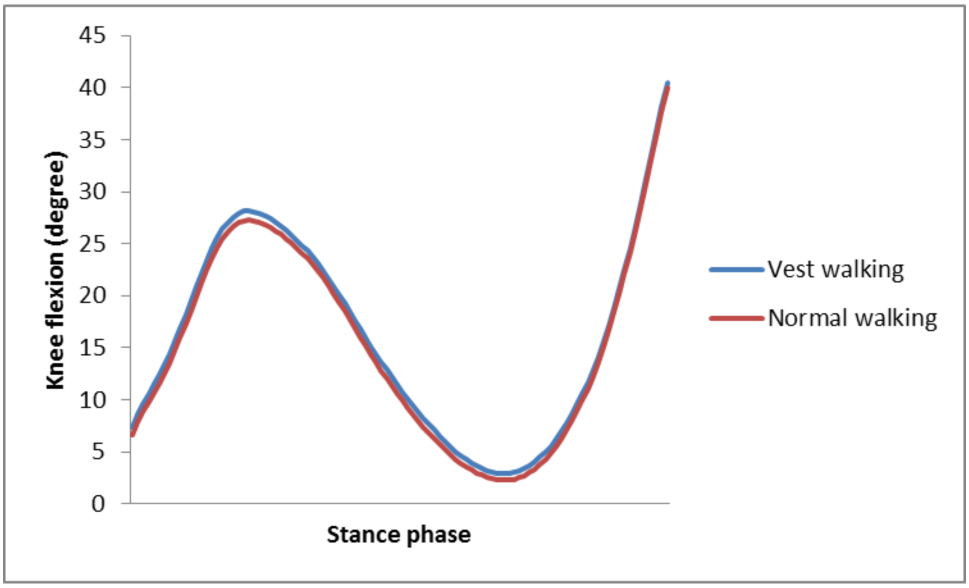

FIGURE 24. Knee flexion angle in the stance phase in normal and vest conditions.

8.1.3 Walking ground reaction forces

Ground reaction forces reflect the summation of the mass acceleration of all body segments. Ground reactions forces were normalized to body weight for normal trials, and normalized to increased weight (body + added vest) , in order to reflect the amount to body weight. During walking, normalized and absolute peak ground reaction forces among every participant increased in vest condition. There were no significant increases in normalized and absolute peak ground reaction forces. See figure 25 for the absolute values between groups.

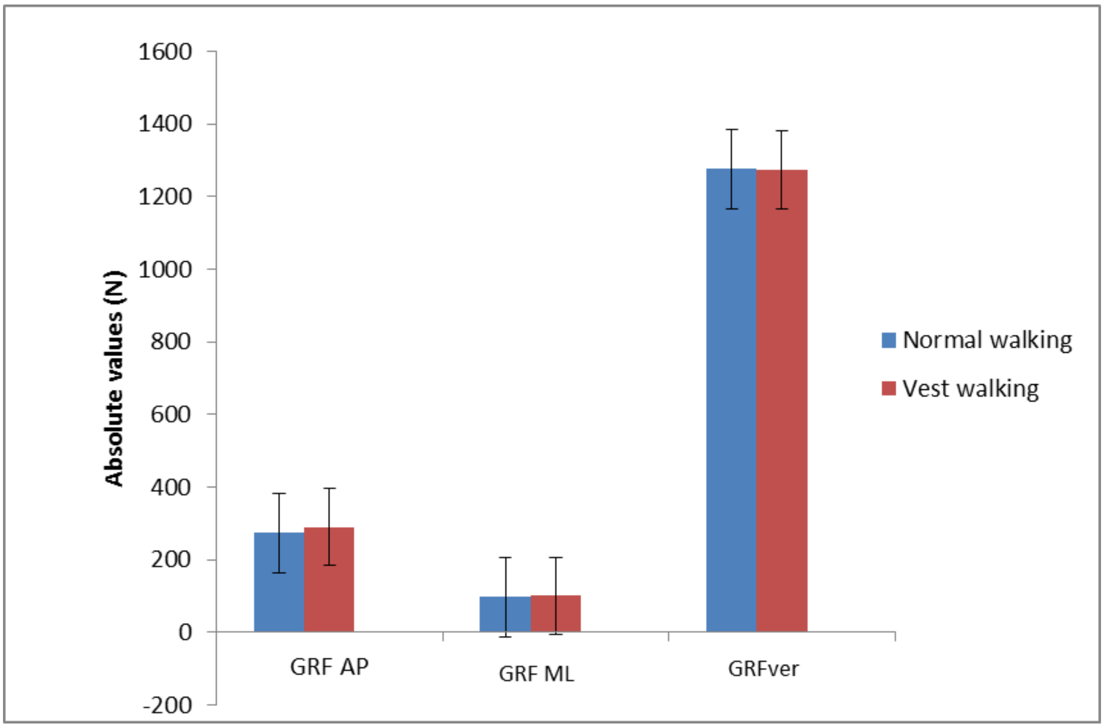

FIGURE 25.Average of absolute GRF in peak AP GRF, ML GRF and vertical GRF. There were increases in all absolute GRF.

8.1.4 Walking joint moments

In walking trials, all subjects showed increase in hip, knee and ankle moments in walking. In both normalized and absolute values, the values increased in vest conditions. In normalized value, all values increased but not significant (table 3). In absolute values, there are significant increases with increasing weight in knee flexion moment and ankle dorsiflexion moment. (table 4) increased from 76.32±33.33 to 104.12±56.55 ($p<0.05$) and ankle dorsiflexion moment increased from 68.36±6.06 to 33.05±7.99 ($p<0.05$)

TABLE 3. Knee and ankle joint moment changes in walking. Data are presented in mean (Nm/kg)± (standard deviation) The data are normalized to body weight

	Normal condition	Vest condition	p value
Knee flexion (Nm/kg)	0.96 (0.38)	1.24 (0.65)	0.13
Knee adduction(Nm/kg)	0.85(0.27)	0.95 (0.38)	0.24
Knee rotation (Nm/kg)	0.23 (0.06)	0.22 (0.08)	0.39
Ankle dorsiflexion(Nm/kg)	1.49 (0.16)	1.68 (0.38)	0.17
Ankle eversion(Nm/kg)	-0.34 (0.11)	-0.35 (0.13)	0.53

TABLE 4. Knee and ankle joint moment changes in walking. Data are presented in absolute mean (Nm/kg) ± (standard deviation)

	Normal condition	Vest condition	p value
Knee flexion (Nm)	76.32 (33.33)	104.12 (56.55)	0.04*
Knee adduction(Nm)	61.77 (25.09)	78.63 (32.59)	0.11
Knee rotation (Nm)	15.94(3.91)	18.14(6.06)	0.29
Ankle dorsiflexion(Nm)	116.52 (21.96)	141.96 (38.34)	0.03*
Ankle eversion(Nm)	-26.48 (7.94)	-29.34 (10.65)	0.89

*P<0.05

8.1.5 Patellofemoral joint and Achilles tendon force

In patellofemoral joint, patellofemoral joint stress (PFS), patellofemoral joint contact forces (PFCF) patellotendon force (PTF) and Achilles tendon force increased in vest conditions. Table 5 shows the values normalized to body weight in the PFCF, PTF and ATF. There were significant increases ($p<0.001$) in PFJ, PFCF, PTF and PFS between normal and vest walking. PFS increased from 3.92±1.65 to 4.33±1.81 ($p<0.001$) (Figure 26).Figure 27 shows PFCF, peakPTF and peakATF in the normal and vest conditions of the joint loading in walking.

TABLE 5. Joint loading in walking. Data are presented in mean (degrees)± (standard deviation)

	Normal condition	Vest condition	p value
PFCF (BW)	1.09 (0.48)	1.21 (0.5)	P<0.001***
peakPTF(BW)	2.08 (0.74)	2.30 (0.82)	P<0.001***
peakATF(BW)	31.31 (3.29)	31.22(3.35)	P<0.001***

*** P<0.001; BW = body weight

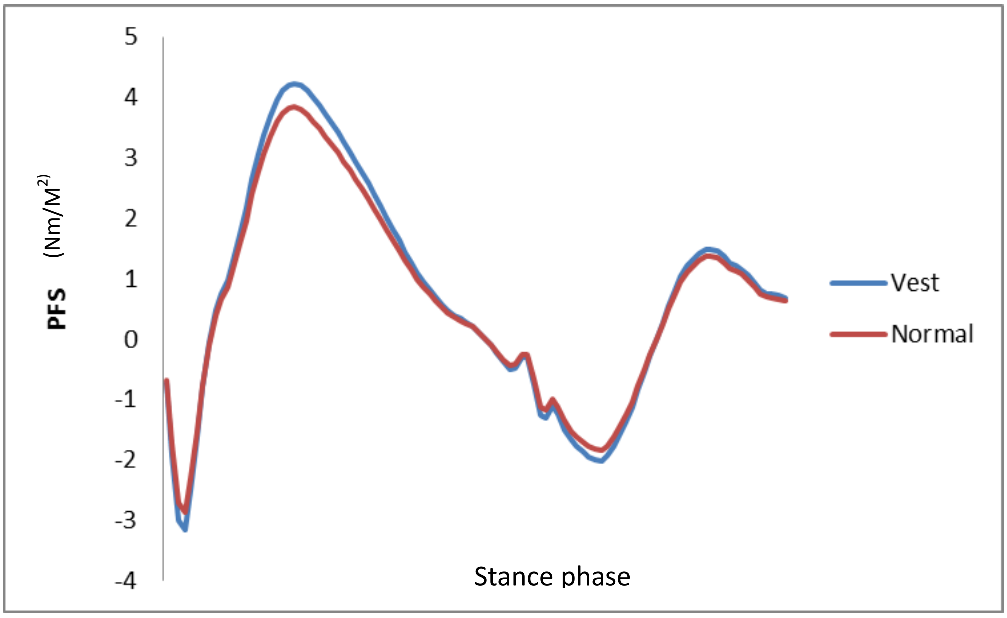

FIGURE 26. Patellofemoral stress of both normal and vest walking in the whole stance phase.

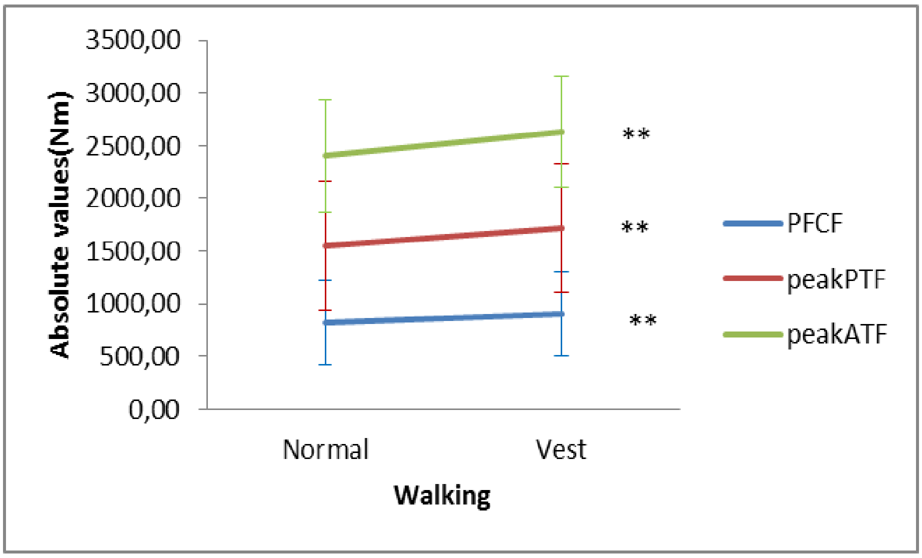

FIGURE 27. Average values of 11 subjects in walking joint loadings. PFCF , peakPTF and peakATF all increased significantly.

***p value<0.01*

8.1.6 Correlation between body weight and joint load

There was correlation(r=0.32 to 0.82, p<0.05), between change in body weight and changes in PFCF peakPTF, peakATF and PFS. When there was 1N increase in body weight, there were 1.2 N increases in PFCF, 2.14N increase in peakPTF, 2.83 N increase in peakATF and in PFS around 0.05N.

8.2 Running

8.2.1 Running stride parameters

The average running speed in normal condition was 4.03±1.26 m/s. The average running speed with vest was 3.95±1.47 m/s.

8.2.2 Running kinematics

In running trials, with increasing weight by 10%, all the ROM from the lower extremity increased. There are no significant increases in the ROM. Figure 28 shows the knee flexion angle in stance phase during running.

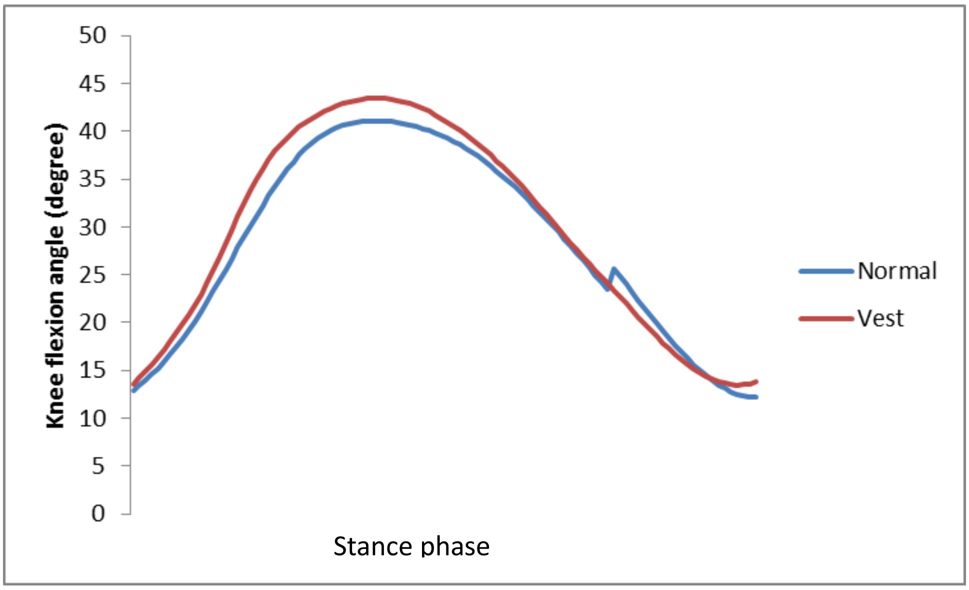

FIGURE 28. Knee flexion angle in stance for normal and vest conditions in running stance phase.

8.2.3 Running ground reaction forces

During running, normalized ground reaction forces among every participant increased in vest condition. There were no significant increases in normalized values. However, in absolute values, there were significant increases in absolute values in peak anterior posterior GRF ($p<0.0001$. See figure 29 for the absolute values between groups.

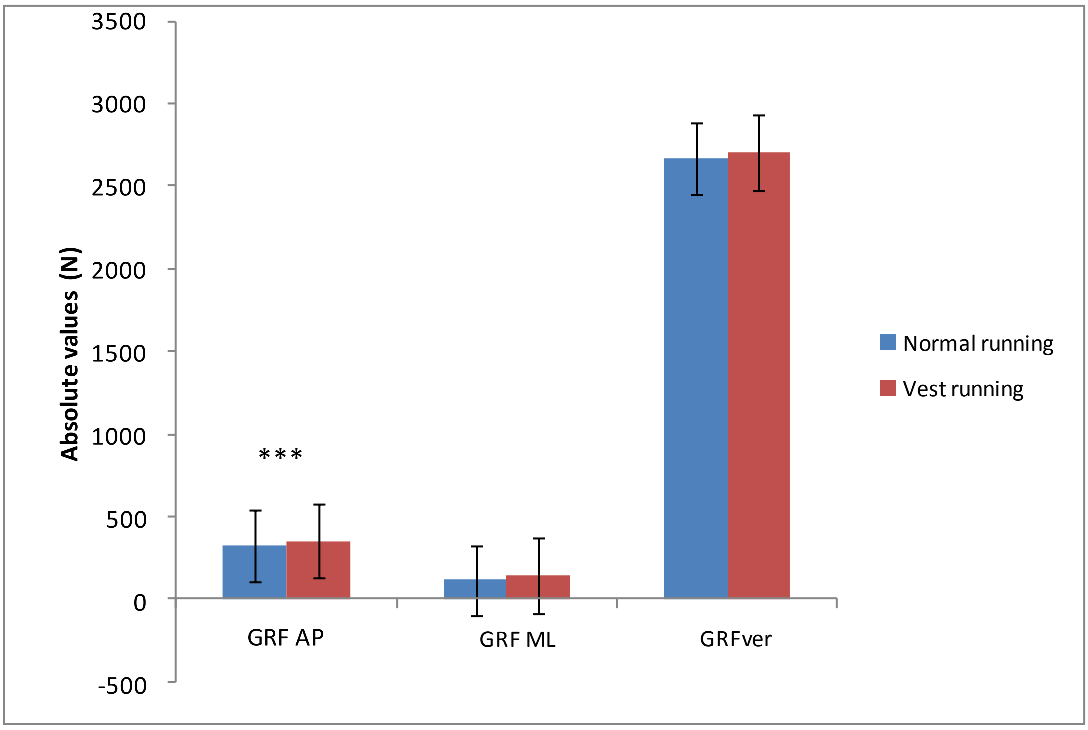

FIGURE 29.Average of absolute GRF in peak anterior posterior GRF, mediolateral GRF and vertical GRF. There were significant increases in absolute GRF AP.

****p value<0.001*

8.2.4 Running joint moments

In running trials, the normalized values of all subjects showed general increase in hip, knee and ankle moments in vest running. There was a significant increase in knee adduction moment ($p<0.05$) and hip rotation moment ($p<0.001$). The increase in knee adduction angle was by 1.08%, in knee adduction moment was by 1.07% and in hip rotation moment by 1.17%.

In absolute values, knee and ankle joint moments increased in the lower extremity in vest conditions. Table 6 shows the increases in lower extremity joint moments.

TABLE 6. Knee and ankle joint moment changes in normalized to body weight in walking. Data are presented in mean (N/kg)± (standard deviation)

	Normal condition	Vest condition	p value
Knee flexion(N/kg)	2.57 (0.91)	2.59 (0.88)	0.67
Knee adduction(N/kg)	1.99 (0.64)	2.14(0.69)	0.01*
Knee rotation (N/kg)	0.21(0.07)	0.22 (0.10)	0.55
Ankle dorsiflexion (N/kg)	3.26 (0.34)	3.22(0.34)	0.27
Ankle plantarflexion(N/kg)	-0.25(0.11)	-0.27(0.11)	0.29

*P<0.05

TABLE 7. Knee and ankle join moment changes in running. Data are presented in absolute values (Nm) in mean (degrees)± (standard deviation)

	Normal condition	Vest condition	p value
Knee flexion(Nm)	196.08 (75.19)	217.71 (79.23)	<0.001***
Knee adduction(Nm)	154(63.22)	182.45(77.89)	<0.001***
Knee rotation (Nm)	18.61(15.09)	17.93(15.56)	0.119
Ankle dorsiflexion(Nm)	248.84 (36.44)	270.10(42.91)	<0.001***
Ankle plantarflexion(Nm)	-18.63(8.09)	-22.51(9.16)	0.054

*** P<0.001

8.2.5 Patellofemoral joint and Achilles tendon force

During running, PFCF, peak PTF and PFS increased in vest running. There were no significant differences in the increase between two conditions (p>0.05). Table 8 shows the normalized values to body weight. In the ankle joint, vest running has higher peak ATF than normal condition. PFS increases from 12.13±5.81 to 13.08±5.68(p>0.05).. See figure 31 for the joint loadings and joint forces.

TABLE 8. Patellofemoral Compression force, peak patellofemoral tendon force, peak Achilles tendon forces in running. Data are presented in normalized values to body weight in mean (degrees)± (standard deviation)

	Normal condition	Vest condition	p value
PFCF(BW)	3.44 (1.460)	3.89 (1.45)	0.097
peakPTF(BW)	5.69 (2.69)	6.15 (2.71)	0.626
peakATF(BW)	63.48 (9.28)	59.72 (5.55)	0.097

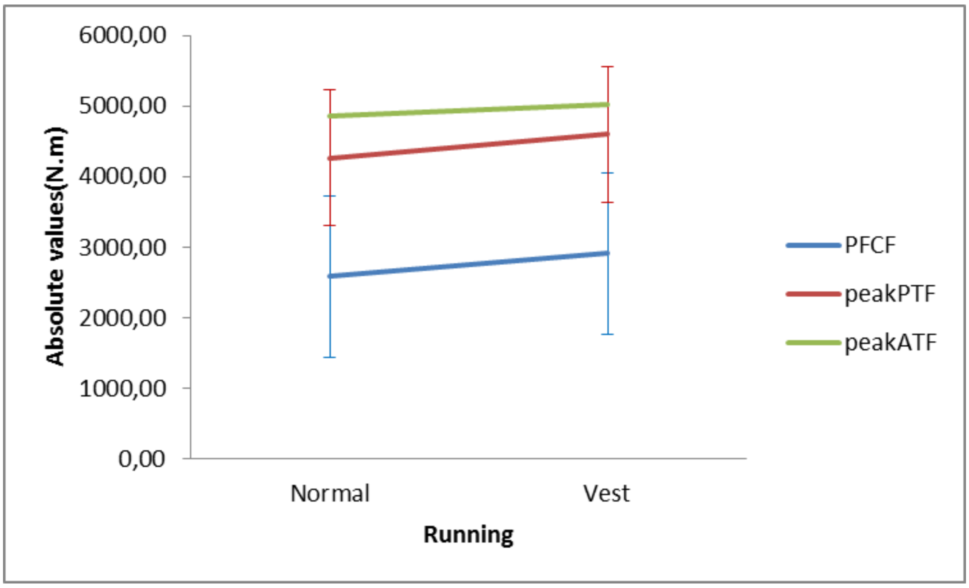

FIGURE 30. During running, PFCF, peakPTF and peakATF increases in vest conditions.

8.2.6 Correlation between body weight and joint load

There was no significant correlation (p>0.05) between body weight and knee, ankle joint loading. During running, the relationship between body weight increase and PFCF change is 1:4, in peakPTF 1:4, in peakATF 1:2 and in PFS 1:0.1.

8.3 Comparison between walking and running in specific condition.

In normal conditions, joint loading increased in running. There were significant increases between walking and running in PFCF, peakPTF and peakATF(p<0.001). In vest conditions, there were significant increases between walking and running in PFCF, peakPTF and peakATF(p<0.001). Figure 32 and 33 showed the normal and vest conditions respectively.

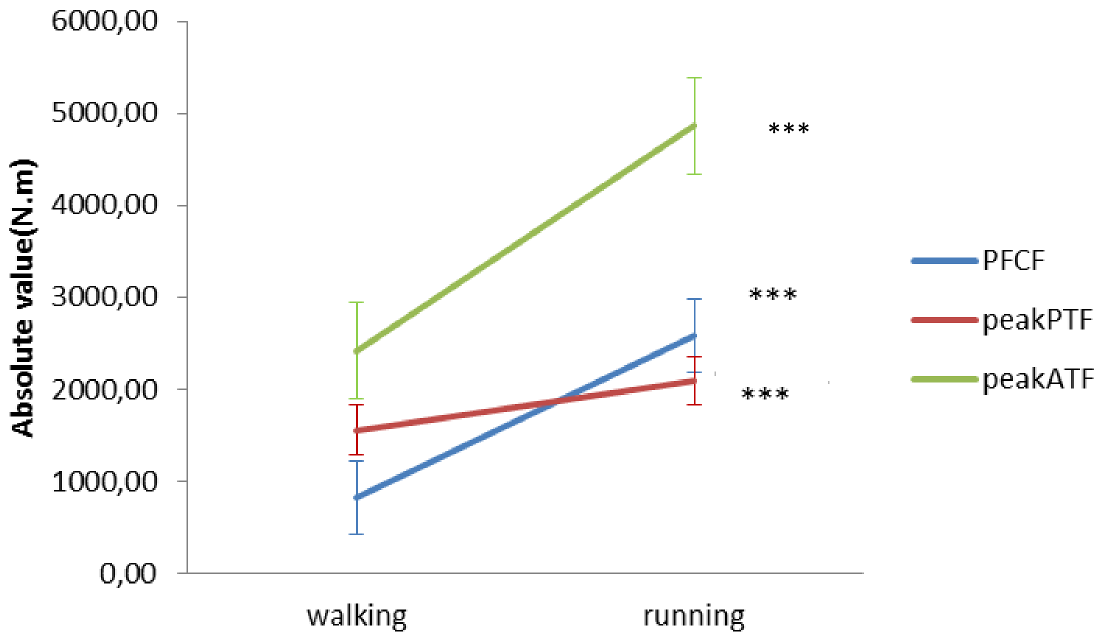

FIGURE 31. In normal conditions, running has higher PFCF, peakPTF and peakATF

P<0.001; BW = body weight

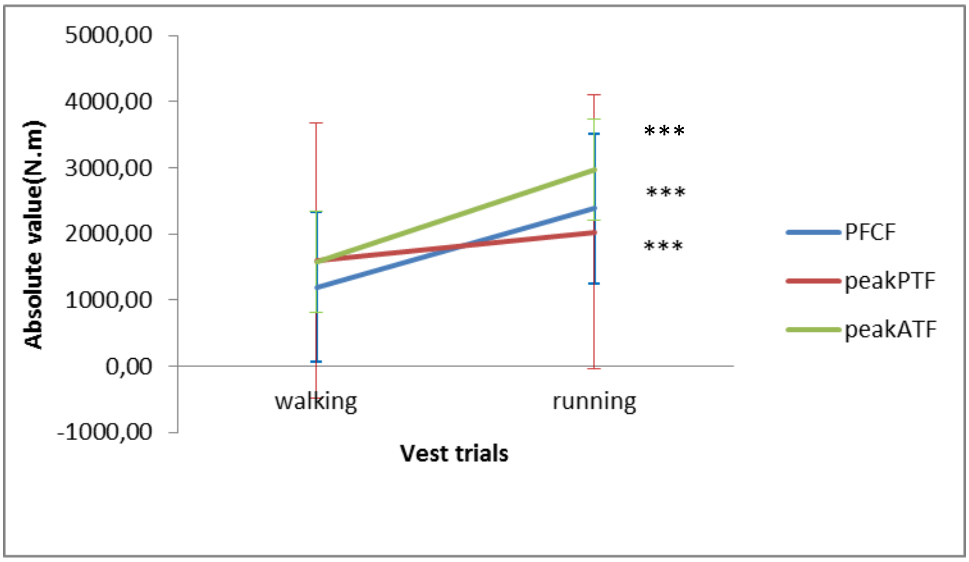

FIGURE 32. In vest conditions, running has higher PFCF, peakPTF and peakATF.

*** P<0.001; BW = body weight

9. DISCUSSION

The primary aim of this study was to calculate and compare the joint moments, ground reaction forces, patellofemoral joint contact forces(PFCF), patellotendo force (PTF), patellofemoral stress (PFS) and Achilles tendon forces (ATF) respectively in normal trials and with added weights (vest trials) in walking and running. There were several researches that investigate the kinetics and kinematics changes in added loads on waist or back (Tilbury-Davis& Hooper. 1999; Griffin et al. 2003). However, there have been no previous researches that investigate the relationship between 10% increase in weight in healthy populations in both running and walking using this experimental method.

9.1 Relationship between weight and ground reaction forces

In the study, we showed both the normalized values and absolute values. Although normalizing the values to body mass in the parameters of interest allows comparisons for two conditions, it distracts the attention of actual loads placed on the joints. According to Ding et al (2005), joint articulating surface area does not scale with body mass. Therefore, we suggest using absolute values to reflect the actual loads on the lower body joints.

During walking, our results showed increases in vest conditions (weight increase) in absolute values. The proportion of body weight and ground reaction forces were 0.36xBW in peak anterior-posterior (AP) GRF, 0.13 peak mediolateral (ML) GRF and 1.67 peak vertical GRF. The results were in line with Messier et al (1996) and Browning et al (2007). In Messier et al (1996), absolute peak AP GRF was 0.14 x body weight and peak vertical GRF was 1.03 x body weight. Absolute peak AP and ML GRF increased proportionally with body weight. Messier et al (1996) found 0.15xBW in absolute peak and 1.02xBW in peak AP GRF

In Tilbury-Davis& Hooper study, 20kg and 40kg of loadings were carried on the participants' back in walking. The study design was using the same weight in different individuals, the load weight ranged from 20% to30% (20kg) and 47%to 64% (40kg) increase in individual. There were no significant increases in normalized values (to weight) Moreover, Kinoshita (1985) also found GRF in vertical, ML and AP increased by increasing loads. This is in line with our research, suggesting that the weight increase is not the main factor affecting GRF.

During running, we also found increases in both normalized and absolute values in peak vertical GRF , anteriorposterior GRF and mediolateral GRF. From our results, the peak vertical GRF is by 3.7 x body weight (BW), anteriorposterior GRF by 0.48x BW and mediolateral GRF by 0.18x BW among normal and added weight conditions. There are no studies in obese that investigate in the running biomechanics in obese group since that brisk walking has been the recommended protocol for weight loss (Browning et al, 2011)

In GRF, the relationship between weight and mediolateral, anteriorposterior GRF did not change much in walking and running; however, the relationship changed in vertical GRF, in that there is an almost 4 times increase in body weight comparing running to walking, suggesting that in higher speed, weight increased has a bigger impact on GRF.

9.2 Relationship between weight and knee, ankle moment

During walking, we found out there were significant increases between normal and vest walking in knee flexion moment and knee adduction moment. This might be due to the significant increase in knee flexion ROM. In load carrying studies, Tilbury-Davis & Hooper (1999) showed no such differences in knee flexion with load carriage in experienced military males. Other studies showed increased knee flexion (Kinoshita 1985). The difference between the studies may be explained by the experience among the subjects since that the muscles in the lower extremity were strengthened to maintain a "similar" gait motion despite the carried load. (Tilbury-Davis & Hooper, 1999) that knee flexion during mid-stance helps maintain stability by keeping the body's centre of mass lower (Harman et al, 2000) Our results were in line with the obese studies (Messier et al, 2005; Schipplein and Andriacchi, 1991) as the weight increased, knee flexion increased and knee flexion moment, knee adduction moment also increased, suggesting that in walking kinetics, the main compensation for increased weight is by increase in knee flexion, rather than hip or ankle flexion.

We showed preliminary results in running with weight gaining by 10%BW gain. All joint moments increased in the lower extremity in vest conditions. There were significant increases in hip flexion, hip extension, rotation moment, knee rotation and knee flexion.

There were also significant increases ROM in hip rotation and ankle maximum dorsiflexion. In lower extremity ROM, there were significant increases in knee adduction, knee rotation, and flexion. These results were different than walking. As the weight increased, human altered gait by hip rotations, knee flexion and ankle dorsiflexion, using all the lower extremity joint kinetics adapt to the weight gain.

9.3 Relationship between weight increase and joint loading

We found that during walking, the relationship between body weight increase and PFCF change is 1:1, in PTF 1:2, in peakATF 1:3 and in PFS 1:0.05. The increase in the knee loading might be due to the increase of knee kinetics and kinematics and the weight increase, since there was a significant increase of knee flexion angle. During walking in obese patients, Aaboe et al (2011) found that 13.5% weight loss showed 7% reductions of knee PFCF. In other words, the correlation between body weight increase and PFCF change is 1:0.5. Comparing to obese that there were 1:2 to 1:4 relationship between weight gain and joint loadings in walking, we suspected that the increase in joint loading doubled or more in higher BMI individuals. Messier et al (2005) found that the correlation is 1:4. The differences between obese and healthy populations may be that the muscle strength and muscle activity is different among two populations. The activation ratios of the gastrocnemius to tibialis anterior were also statistically higher in knee OA than in healthy adults and healthy young adults, with large variations between OA subjects, duration of muscle activity in vastus lateralis, hamstring, tibialis anterior and gastrocnemius also increased (Hortobágyi et al, 2005; Childs et al, 2004).

Achilles tendon forces (ATF) has not been studied before in load carrying studies and obese populations. The increase in ATF may be due to the increase in ankle moment and weight increase. The impact of the weight appeared to affect more on the ankle joint than the knee joint loadings.

To our best knowledge, there were no studies that investigate the joint load in increase of 10% weight in running. We found the correlation between body weight increase and PFCF change is 1:4, in peakPTF 1:4, in peakATF 1:2 and in PFS 1:0.1. As the kinetics in knee and ankle both increased significantly, the joint loading might be related to the increase in joint ROM, joint moments and muscle activation in running.

9.4 Speed and joint loading

In both normal and vest conditions, our results showed that the PFCF, peakATF and peakPTF and PFS are significantly higher in running, suggesting that the speed increase affected the joint loadings. The speed seemed to have higher impact on PFCF in that the relationship of weight changes and PFCF is 4 times in running. In peakPTF and PFS, comparing walking to running, the ratio was twice higher in running. However, in peakATF it was slightly lower in the increase weight/peakATF ratio in running. This could be explained by the slight smaller peak ankle dorsiflexion in running trials. The peakATF was in line with other studies (Burdett, 1981; Giddings et al, 2000) with peakATF ranging from 2-6 x body weight in running in both normal and vest conditions. In normal and vest conditions, peakPTF ranged from 4-7 x body weight from our studies and also other studies with similar speeds (Burdett, 1982)

9.5 Limitations of the study

In weight gain, people usually increase fat tissue in different areas, for instance in arm, thighs or ventral area. In our study, vest wearing only represented the increase of fat around the ventral area, which might not fully present the condition of weight gaining in healthy populations. Our subjects were all male young students; we might make a generalization of the results to vest walking and running in female students. Gender needs to take into considerations since there are studies showing the differences in muscle activity in genders and elderly populations. There were studies (Kerrigan et al 1998; Decker et al 2003) showing that in walking, females had significantly greater hip flexion and less knee extension before initial contact, greater knee flexion moment in pre-swing, and greater peak mechanical joint power absorption at the knee in pre-swing. As in running(Ferber et al 2003; Chumanov et al 2008), female recreational runners exhibit significantly different lower extremity mechanics in the frontal and transverse planes at the hip and knee during running compared to male recreational runners

In the study we could gain possible insight in vest running or vest walking for training normal people or athletic training, even though the weight is less than those that are actually used in the training conditions. Task experience and previous training in carrying heavy loads may influence the kinetic and kinematic responses to loads (Littlepage, Robinson & Reddington,1997; Vasta, Rosenberg, Knott & Gaze. 1997). This also needs to take into considerations since our subjects were not familiar with the loaded vests, it might influence the gait pattern in walking and running

9.6 Suggestions for further study

In future research, experimental design could involve diet control and weight gain through foods to investigate the relationship between weight gain and joint loads among healthy subjects. Measurement of electromyography (EMG) of muscle activity could be incorporated into an inverse dynamic study of the lower body. By incorporating the EMG, as the subjects gain weight, an understanding of the muscle activation patterns and timing change could be further investigated to obtain the roles of muscles in the lower extremity, abdominal muscles in generating the resultant joint moments and forces, patellofemoral stress, patellofemoral contact forces, patellotendon force and Achilles tendon force. We should also note that there are differences in the lower extremity kinematics, especially in running between male and female; therefore, with female subjects in the study, we might have different results in the joint loading in two conditions. Therefore, for further studies, both genders should be included to understand the relationship between weight gain and joint loadings.

10. CONCLUSIONS

Overweight is associated with osteoarthritis, specifically in the patellofemoral joint. Weight loss programs such as brisk walking, incline walking was recommended for overweight people. Previous studies showed in overweight patients that there were relationship between changes in body weight and joint loads, with 1:2 and 1:4 ratios in changes of the loading. However, there were no studies investigating from the healthy populations of how weight gaining changes the loading in the lower extremity. The relationship between weight changes and joint loadings in healthy population would be very important since this could give possible insights into whether weight gaining is affecting loading and whether there are differences in walking and running.

We used a novel method in simulating 10% increase in weight in healthy populations with vest in both walking and running. We used absolute values to report because it showed the actual loading in the lower extremity. From our study, the knee joint loadings increased with weight proportionally In walking, ratio between knee joint loading per increase weight doubled or more in higher BMI individuals (especially in obese patients). During running, knee joint loading increased in double, suggesting the speed also affected the loadings. The speed seemed to have higher impact on PFCF in that the relationship of weight changes and PFCF is 4 times in running. In peakPTF and PFS, comparing walking to running, the ratio was twice higher in running. In a nutshell, our study implies that as human gains weight, there might be a higher risk of getting knee osteoarthritis, especially with those that have major recreational sports in running. Our study also suggests that weight reduction is crucial for those that are overweight to prevent osteoarthritis since weight is directly correlated to joint loads.

11. REFERENCES

Aaboe. J.. Bliddal. H.. Messier. S. P.. Alkjaer. T.. & Henriksen. M. (2011). Effects of an intensive weight loss program on knee joint loading in obese adults with knee osteoarthritis. Osteoarthritis and Cartilage. 19(7). 822-828.

Arellano, C., Layne, C., O'Connor, D., Scott-Pandorf, M., & Kurz, M. (2009). Does load carrying influence sagittal plane locomotive stability? *Medicine Science in Sports Exercise*,*41*(3), 620.

Ahmed, A. M., & Burke, D. L. (1983). In-Vitro of Measurement of Static Pressure Distribution in Synovial Joints—Part I: Tibial Surface of the Knee. *Journal of biomechanical engineering*, *105*(3), 216-225.

Ahmed, A. M., Burke, D. L., & Yu, A. (1983). In-vitro measurement of static pressure distribution in synovial joints—Part II: retropatellar surface. *Journal of biomechanical engineering*, *105*(3), 226-236.

Arndt, A. N., Komi, P. V., Brüggemann, G. P., & Lukkariniemi, J. (1998). Individual muscle contributions to the in vivo achilles tendon force. *Clinical Biomechanics*, *13*(7), 532-541.

Astephen. J. L.. Deluzio. K. J.. Caldwell. G. E.. & Dunbar. M. J. (2008). Biomechanical changes at the hip. knee. and ankle joints during gait are associated with knee osteoarthritis severity. *Journal of Orthopaedic Research*. *26*(3). 332-341.

Ateshian, G. A., Kwak, S. D., Soslowsky, L. J., & Mow, V. C. (1994). A stereophotogrammetric method for determining in situ contact areas in diarthrodial joints, and a comparison with other methods.*Journal of biomechanics*,*27*(1), 111-124.

Bechtol. C.O. (1975). Normal human gait. In J.H. Bowker & C.B. Hall (Eds.). Atlas of orthotics:American Academy of Orthopaedic Surgeons (pp. 133-143). St. Louis: Mosby.

Bredeweg. S.. & Buist. I. (2011). No relationship between running related injuries and kinetic variables. *British Journal of Sports Medicine*. *45*(4). 328-328.

Browning, R. C., Modica, J. R., Kram, R., & Goswami, A. (2007). The effects of adding mass to the legs on the energetics and biomechanics of walking. *Medicine and science in sports and exercise*, *39*(3), 515.

Browning. R. C.. & Kram. R. (2007). Effects of obesity on the biomechanics of walking at different speeds.*Medicine and Science in Sports and Exercise*.*39*(9). 1632.

Burdett, R. G. (1981). Forces predicted at the ankle during running. *Medicine and Science in Sports and Exercise*, *14*(4), 308-316.

Chao, E. Y., Armiger, R. S., Yoshida, H., Lim, J., & Haraguchi, N. (2007). Journal of Orthopaedic Surgery and Research.*Journal of Orthopaedic Surgery and Research*,*2*, 2.

Childs, J. D., Sparto, P. J., Fitzgerald, G. K., Bizzini, M., & Irrgang, J. J. (2004). Alterations in lower extremity movement and muscle activation patterns in individuals with knee osteoarthritis.*Clinical biomechanics*,*19*(1), 44-49.

Chumanov, E. S., Wall-Scheffler, C., & Heiderscheit, B. C. (2008). Gender differences in walking and running on level and inclined surfaces. *Clinical biomechanics*, *23*(10), 1260-1268.

Clark. A. L.. Herzog. W.. & Leonard. T. R. (2002). Contact area and pressure distribution in the feline patellofemoral joint under physiologically meaningful loading conditions. *Journal of biomechanics*. *35*(1). 53-60.

Csintalan, R. P., Schulz, M. M., Woo, J., McMahon, P. J., & Lee, T. Q. (2002). Gender differences in patellofemoral joint biomechanics.*Clinical Orthopaedics and Related Research*,*402*, 260-269.

Cole, G. K., Nigg, B. M., van Den Bogert, A. J., & Gerritsen, K. G. M. (1996). Lower extremity joint loading during impact in running. *Clinical Biomechanics*, *11*(4), 181-193.

Collier. J. P.. MCNAMARA. J. L.. SURPRENANT. V. A.. Jensen. R. E.. & Surprenant. H. P. (1991). All-polyethylene patellar components are not the answer. Clinical orthopaedics and related research. 273. 198-203.

Dancey, C., & Reidy, J. (2004). Analysis of differences between two conditions: the t-test.*Statistics without Maths for Psychology: Using SPSS for Windows Essex, England: Prentice Hall*, 206-236.

Daoud. A. I.. Geissler. G. J.. Wang. F.. Saretsky. J.. Daoud. Y. A.. & Lieberman. D. E. (2012). Foot strike and injury rates in endurance runners: a retrospective study. *Med Sci Sports Exerc*. *44*(7). 1325-34.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Steadman, J. R. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics*, *18*(7), 662-669.

Ding. C.. Cicuttini. F.. Scott. F.. Cooley. H.. & Jones. G. (2005). Knee structural alteration and BMI: a cross-sectional study.*Obesity research*. *13*(2). 350-361.

Ehlen, K. A., Reiser, R. F., & Browning, R. C. (2011). Energetics and biomechanics of inclined treadmill walking in obese adults.*Med Sci Sports Exerc*,*43*(7), 1251-9.English. T. A. & Klivington. M. (1979) J. Biomed. Eng. 1.111-115.

Ewing (Ed.).J.W. In Articular Cartilage and Knee Joint Function: Basic Science and Arthroscopy. Raven Press. New York (1990). pp. 93–101

Ferber, R., Davis, I. M., & Williams Iii, D. S. (2003). Gender differences in lower extremity mechanics during running. *Clinical biomechanics*, *18*(4), 350-357.

Felson. D. T.. Anderson. J. J.. Naimark. A.. Walker. A. M.. & Meenan. R. F. (1988). Obesity and knee osteoarthritisThe Framingham Study. Annals of internal medicine. 109(1). 18-24.

Felson, D. T., Zhang, Y., Anthony, J. M., Naimark, A., & Anderson, J. J. (1992). Weight loss reduces the risk for symptomatic knee osteoarthritis in women: the Framingham Study. *Annals of internal medicine*,*116*(7), 535-539.

Felson, D. T. (1996). Does excess weight cause osteoarthritis and, if so, why?. *Annals of the rheumatic diseases*, *55*(9), 668.

Finni, T., Komi, P. V., & Lukkariniemi, J. (1998). Achilles tendon loading during walking: application of a novel optic fiber technique. *European journal of applied physiology and occupational physiology*, *77*(3), 289-291.

Finni, T., Komi, P. V., & Lepola, V. (2000). In vivo human triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump. *European journal of applied physiology*, *83*(4-5), 416-426.

Finni, T., Ikegaw, S., Lepola, V., & Komi, P. (2001). In vivo behavior of vastus lateralis muscle during dynamic performances. *European Journal of Sport Science*, *1*(1), 1-13.

Flynn, T. W., & Soutas-Little, R. W. (1995). Patellofemoral joint compressive forces in forward and backward running.*Journal of Orthopaedic & Sports Physical Therapy*,*21*(5), 277-282.

Fukubayashi, T., & Kurosawa, H. (1980). The contact area and pressure distribution pattern of the knee: a study of normal and osteoarthrotic knee joints. *Acta Orthopaedica*, *51*(1-6), 871-879.

Giddings, V. L., Beaupre, G. S., Whalen, R. T., & Carter, D. R. (2000). Calcaneal loading during walking and running. *Medicine and science in sports and exercise*, *32*(3), 627-634.

Goswami. A. (2008. December). Kinematic and dynamic analogies between planar biped robots and the reaction mass pendulum (RMP) model. InHumanoid

Robots. 2008. Humanoids 2008. 8th IEEE-RAS International Conference on IEEE. 182-188

Greenwald, A. S., & O'Connor, J. J. (1971). The transmission of load through the human hip joint.*Journal of biomechanics*,*4*(6), 507-528.

Griffin. T. M.. Roberts. T. J.. & Kram. R. (2003). Metabolic cost of generating muscular force in human walking: insights from load-carrying and speed experiments.*Journal of Applied Physiology*. *95*(1). 172-183.

Harrington IJ.(1976) A bioengineering analysis of force actions at the knee in normal and pathological gait. Biomed Eng. 11:167-72.

Harman, E., Hoon, K., Frykman, P., & Pandorf, C. (2000).*The effects of backpack weight on the biomechanics of load carriage*. Army research institute of environmental medicine and military performance IV

Hills. A. P.. Hennig. E. M.. Byrne. N. M.. & Steele. J. R. (2002). The biomechanics of adiposity–structural and functional limitations of obesity and implications for movement. *Obesity Reviews*. *3*(1). 35-43.

H.S. Gill. J.J. O'Connor (1997). A new testing rig for force platform calibration and accuracy tests. Gait& Posture 5 (3)

Ishikawa, M., Komi, P. V., Grey, M. J., Lepola, V., & Bruggemann, G. P. (2005). Muscle-tendon interaction and elastic energy usage in human walking. *Journal of applied physiology*, *99*(2), 603-608.

Jensen RK. (1989) Changes in segment inertia proportions between 4 and 20 years. J Biomech; 22(6-7):529-36.

Jordan JM, Kington RS, Lane NE, Nevitt MC, Zhang Y, Sowers MF. Systemic risk factors for osteoarthritis. In Felson DT. Osteoarthritis: new insights. Part 1: The disease and its risk factors. Ann Intern Med 2000; 133: 637-9.

J.P. Holden. J.A. Orsini. K.L. Siegel. T.M. Kepple. L.H. Gerber. S.J. Stanhope(1997). Surface movement errors in shank kinematics and knee kinetics during gait. Gait & Posture 5 (3): 217-227.

J.P. Holden. S.J. Stanhope(1998). The effect of variation in knee center location estimates on net knee joint moments. Gait & Posture 7 (1):1-6.

Keller, T. S., Weisberger, A. M., Ray, J. L., Hasan, S. S., Shiavi, R. G., & Spengler, D. M. (1996). Relationship between vertical ground reaction force and speed during walking, slow jogging, and running. *Clinical Biomechanics*, *11*(5), 253-259.

Kerrigan, D. C., Todd, M. K., & Croce, U. D. (1998). Gender differences in joint biomechanics during walking normative study in young adults. *American journal of physical medicine & rehabilitation*, *77*(1), 2-7.

K. Manal. I. McClay. S. Stanhope. J. Richards. B. Galinat (2000). Comparison of surface mounted markers and attachment methods in estimating tibial rotations during walking: an in vivo study. Gait & Posture 11 (1)

M. Brown. S. Stanhope (1995). Preventing Implementation And Tracking Errors From

Becoming Clinical Judgement Errors In Functional Movement Analysis. Gait & Posture 3 (2)

P.S. Fairburn. R. Palmer. J. Whybrow. S. Fielden. S. Jones (2000). A prototype system for testing force platform dynamic performance. Gait & Posture 12 (1)

Giddings, V. L., Beaupre, G. S., Whalen, R. T., & Carter, D. R. (2000). Calcaneal loading during walking and running. *Medicine and science in sports and exercise*, *32*(3), 627-634.

Han. S. K.. Federico. S.. Epstein. M.. & Herzog. W. (2005). An articular cartilage contact model based on real surface geometry. *Journal of Biomechanics*. *38*(1). 179-184.

Ho. K. Y.. Blanchette. M. G.. & Powers. C. M. (2012). The influence of heel height on patellofemoral joint kinetics during walking. *Gait & posture*. *36*(2). 271-275.

Horita, T., Komi, P. V., Nicol, C., & Kyröläinen, H. (1996). Stretch shortening cycle fatigue: interactions among joint stiness, reflex, and muscle mechanical performance in the drop jump. *European journal of applied physiology and occupational physiology*, *73*(5), 393-403.

Hortobágyi, T., Westerkamp, L., Beam, S., Moody, J., Garry, J., Holbert, D., & DeVita, P. (2005). Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis. *Clinical Biomechanics*,*20*(1), 97-104.

Hunt. A. E.. Smith. R. M.. & Torode. M. (2001). Extrinsic muscle activity. foot motion and ankle joint moments during the stance phase of walking. Foot & Ankle International. 22(1). 31-41.

Inman. V. T.. & Eberhart. H. D. (1953). The major determinants in normal and pathological gait. The Journal of Bone & Joint Surgery. 35(3). 543-558.

Ito A. Fuchimoto T. Kaneko M. (1985) Quantitative analysis of EMG during various speeds of running. In Winter DA. et al (eds): Biomechanics IX-B. Champaign. IL. Human Kinetics Publishers Inc. 301-306

Juhakoski. R.. Heliövaara. M.. Impivaara. O.. Kröger. H.. Knekt. P.. Lauren. H.. & Arokoski. J. P. A. (2009). Risk factors for the development of hip osteoarthritis: a population-based prospective study. Rheumatology. 48(1). 83-87.

Kellis, 1998. Quantification of quadriceps and hamstring antagonist activity.*Sports Medicine*,*25*(1), 37-62.

Kinoshita, H. (1985). Effects of different loads and carrying systems on selected biomechanical parameters describing walking gait.*Ergonomics*, *28*(9), 1347-1362.

Knapik, J., Harman, E., & Reynolds, K. (1996). Load carriage using packs: A review of physiological, biomechanical and medical aspects.*Applied ergonomics*, *27*(3), 207-216.

Komi, P. V., Salonen, M., Järvinen, M., & Kokko, O. (1987). In vivo registration of Achilles tendon forces in man. I. Methodological development. *International journal of sports medicine*, *8*, 3-8.

Komi, P. (Ed.). (2008). *The Encyclopaedia of Sports Medicine: An IOC Medical Commission Publication, Strength and Power in Sport* (Vol. 3). John Wiley & Sons.

Kyröläinen, H., Avela, J., McBride, J. M., Koskinen, S., Andersen, J. L., Sipilä, S., ... & Komi, P. V. (2005).

Effects of power training on muscle structure and neuromuscular performance. *Scandinavian journal of medicine & science in sports*, *15*(1), 58-64.

Kuster. M. S.. Wood. G. A.. Stachowiak. G. W.. & Gächter. A. (1997). Joint load considerations in total knee replacement. Journal of Bone & Joint Surgery. British Volume. 79(1). 109-113. Littlepage, G., Robinson, W., & Reddington, K. (1997). E€ ects of task experience and group experience on group performance member ability and recognition of expertise. Organisational Behaviour and Human Decision Processes, 69, 133.

Marks. R.. & Allegrante. J. P. (2002). Body mass indices in patients with disabling hip osteoarthritis. Arthritis research.4(2). 112-116.

Mann. R. A.. Moran. G. T.. & Dougherty. S. E. (1986). Comparative electromyography of the lower extremity in jogging. running. and sprinting. *The American Journal of Sports Medicine*. *14*(6). 501-510.

Messier. S. P.. Gutekunst. D. J.. Davis. C.. & DeVita. P. (2005). Weight loss reduces knee-joint loads in overweight and obese older adults with knee osteoarthritis. Arthritis & Rheumatism. 52(7). 2026-2032.

Messier. S. P.. Legault. C.. Loeser. R. F.. Van Arsdale. S. J.. Davis. C.. Ettinger. W. H.. & DeVita. P. (2011). Does high weight loss in older adults with knee osteoarthritis affect bone-on-bone joint loads and muscle forces during walking? Osteoarthritis and Cartilage.19(3). 272-280.

Morrison. J. B. (1970) The mechanics of muscle function in locomotion. J. Biomech. 3. 431- 451.

Nicol, C., Kuitunen, S., Kyröläinen, H., Avela, J., & Komi, P. V. (2003). Effects of long-and short-term fatiguing stretch-shortening cycle exercises on reflex EMG and force of the tendon-muscle complex. *European journal of applied physiology*, *90*(5-6), 470-479.

Novacheck. T. F. (1998). The biomechanics of running. *Gait & posture*. *7*(1). 77-95.

Nisell. R. (1985). Mechanics of the knee: a study of joint and muscle load with clinical applications. Acta Orthopaedica.56(S216). 1-42.

Olney. S. J.. & Richards. C. (1996). Hemiparetic gait following stroke. Part I: Characteristics.Gait & Posture. 4(2). 136-148.

Powers. C. M.. Lilley. J. C.. & Lee. T. Q. (1998). The effects of axial and multi-plane loading of the extensor mechanism on the patellofemoral joint. *Clinical Biomechanics*. *13*(8). 616-624.

Polcyn A, Bensel C, Harman E, Obusek J, Pandorf C, Frykman P.Effects of weight carried by soldiers: Combined analysis of four studies on maximal performance, physiology, and biomechanics.Technical Report No TR-02/010, MA: US Army Research Institute of Environmental Medicine; 2002: 1–64.

Quesada, P. M., Mengelkoch, L. J., Hale, R. C., & Simon, S. R. (2000). Biomechanical and metabolic effects of varying backpack loading on simulated marching. *Ergonomics*, *43*(3), 293-309.

Rodgers. M. M. (1988). Dynamic biomechanics of the normal foot and ankle during walking and running. *Physical therapy*. *68*(12). 1822-1830.

Runhaar. J.. Koes. B. W.. Clockaerts. S.. & Bierma-Zeinstra. S. M. A. (2011). A systematic review on changed biomechanics of lower extremities in obese individuals: a possible role in development of osteoarthritis. *Obesity reviews*. *12*(12). 1071-1082

Rydell. N. W. (1966) Acta. Orthop. Scand. Suppl. 88. 9.

Salsich, G. B., Brechter, J. H., Farwell, D., & Powers, C. M. (2002). The effects of patellar taping on knee kinetics, kinematics, and vastus lateralis muscle activity during stair ambulation in individuals with patellofemoral pain.*Journal of Orthopaedic & Sports Physical Therapy*,*32*(1), 3-10.

Schipplein. O. D.. & Andriacchi. T. P. (1991). Interaction between active and passive knee stabilizers during level walking. Journal of Orthopaedic Research.9(1). 113-119.

Schiffman JM. Load carriage: quiet stance, postural sway and balance recovery. Natick, MA: Natick Soldier Systems Center; 2004 [Pre-sentation].

Shelburne, K. B., Torry, M. R., & Pandy, M. G. (2006). Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait.*Journal of orthopaedic research*,*24*(10), 1983-1990.

Stephens, M. J., & Yang, J. F. (1999). Loading during the stance phase of walking in humans increases the extensor EMG amplitude but does not change the duration of the step cycle.*Experimental brain research*,*124*(3), 363-370.

Stormont et al., 1985; DM Stormont, BF Morrey, K. An, JR Cass; Stability of the loaded ankle—relation between articular restraint and primary and secondary static restraints. Am. J. Sports Medicine, 13 (1985), pp. 295–300.

Subotnick. S. I. (1985). The biomechanics of running implications for the prevention of foot injuries. *Sports Medicine*. *2*(2). 144-153.

Tilbury-Davis. D. C.. & Hooper. R. H. (1999). The kinetic and kinematic effects of increasing load carriage upon the lower limb. *Human Movement Science*.*18*(5). 693-700.

Van Eijden. T. M. G. J.. De Boer. W.. & Weijs. W. A. (1985). The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion-extension angle. *Journal of biomechanics*. *18*(10). 803-809.

Van Eijden. T. M. G. J.. Kouwenhoven. E.. Verburg. J.. & Weijs. W. A. (1986). A mathematical model of the patellofemoral joint. *Journal of biomechanics*. *19*(3). 219-229.

Vasta, R., Rosenberg, D., Knott, J. A., & Gaze, C. E. (1997). Experience and the water level task revisited:Does expertise exact a price? Psychological Science, 8, 336±339.

Vaughan. C. L. (1983). Biomechanics of running gait. Critical reviews in biomedical engineering. 12(1). 1-48.

Vaughan. C. L.. Davis. B. L.. & O'connor. J. C. (1992). Dynamics of human gait(pp. 15-43). Champaign. Illinois: Human Kinetics Publishers.

Wearing. S. C.. Hennig. E. M.. Byrne. N. M.. Steele. J. R.. & Hills. A. P. (2006). The biomechanics of restricted movement in adult obesity. Obesity reviews. 7(1). 13-24.

Winter DA. Moments of force and mechanical power in jogging. J Biomech 1983; 16 (1): 91-7

Winter. D. A. (1984). Kinematic and kinetic patterns in human gait: variability and compensating effects. Human Movemen t Science. 3(1). 51-76.

Winter. D. A. (2009). Biomechanics and motor control of human movement. John Wiley& Sons.

Yelin E. The economics of osteoarthritis. In Brandt K, Doherty M, Lohmander LS, Eds. Osteoarthritis. New York: Oxford University Press, 1998: 23-30.

Zatsiorsky V, Seluyanov V. The mass and inertial characteristics of the main segments of the human body. In: Matsui H, Kobayashi K, eds. Biomechanics VIII-B. University Park Press, Baltimore, 1983; 1152-9

Zhang, Y., & Jordan, J. M. (2010). Epidemiology of osteoarthritis. *Clinics in geriatric medicine*, *26*(3), 355-369.