THE EFFECTS OF ABDOMINAL OBESITY ON THE HEEL STRIKE TRANSIENT DURING WALKING: AN EXPERIMENTAL STUDY.

POSTGRADUATE THESIS

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Contents

Abstract

One important function of the skeletal muscles is to protect the joints from the impact forces that may be applied during certain activities and attenuate the rates of loading. The impact of the foot with the ground during heel strike produces large Ground Reaction Forces (GRF) and excessive uncontrolled loading of the joints from lack of muscular control may predispose the joints to Osteoarthritis (OA). Women are at a higher risk of developing knee OA after menopause compared with men of the same age. One of the main effects of menopause on the female body is an alteration of fat distribution from the pelvis to the abdominal area. Since it is now widely accepted that perimenopausal women are at a higher risk of knee OA, a good indicator for detecting catastrophic changes at the knee joint during gait from the altered fat distribution may be the size or the presence of a Heel-Strike Transient (HST). Since abdominal obesity increases the inward curvature of the spine it is more likely that heel-walking is increased and thus HSTs may become larger, with a deleterious result for the knee joints. The aim of the study was to test whether several external abdominal weights may increase HSTs during walking in healthy individuals and find an association between central obesity and large HSTs which may explain the higher incidence of OA in perimenopausal women.

Ten subjects were solicited from class announcements or emails and screened before participating in the study. The participants walked over a 9.5 walkway instrumented with a Bertec Force plate. The investigator acquired force plate data at a rate of 1000 Hz. The subjects walked at a self-selected low and high speed over the force platform with their dominant foot three times. The Friedman's ANOVA test was used to detect any significant differences between conditions and a Post-hoc analysis with Wilcoxon signed-rank test was conducted for all the variables tested with a Bonferroni correction.

The results showed that the HST, the time taken to reach the first maximum vertical force peak, the contact time and the loading rates changed significantly only when the subjects were instructed to alter their gait speed. However, the first maximum force peak increased significantly when the external weights were placed on the subjects and proportionally with the size of the weight. On the contrary, the second maximum force peak decreased significantly when the external weights were placed on the subjects.

In conclusion, abdominal weight may not affect the size of the HST and rates of loading in healthy individuals. However, it was found that the first maximum force is the only variable that may increase in size when external weights are placed on healthy participants. These findings suggest that muscle strength may attenuate impact forces when extra weight is placed on a subject.

Keywords:

Obesity, Heel-Strike, Abdominal Obesity, External Weight, Gait

Introduction

Gait and Ground Reaction Forces

Gait is defined as a series of rhythmic movements of the lower extremities, resulting in forward movement of the body (Iyer, 2012). The extension of the limbs imparts momentum and the force limbs restrict this momentum when they contact the ground. Heel to toe gait is an essential pattern of movement for each leg, which starts with the initial contact of the heel with the ground, followed by the toes and ends with the heel and then the toes leaving the ground surface (Iyer, 2012).

Human gait has two distinctive phases the stance and the swing phase. During the stance phase the feet are in contact with the ground whereas in the swing phase the lower extremities swing forward in an alternate pattern. The posterior thigh muscles control the swing phase and the heel strike controls forward momentum. From this, it is clear that the stance phase starts at heel strike, and lasts through foot flat and toes strike before ending after toe rise when the swing phase begins (Iyer, 2012).

During gait, when the foot touches the ground surface, it applies a force to the ground and another force is created that is equal and opposite of the force that the foot applies on the ground. The force developed by the ground is called Ground Reaction Force (GRF) (DeLisa, 1998). For example, when a person stands on a surface creates a force on this surface which is equal to the person’s body weight. However, in order for the person to stay motionless and stand on this surface there is a development of an equal and opposite ground reaction force which is exerted on the person. The term reaction derives from Newton’s third law which states that for every action which acts upon a body, there is an equal reaction that must act upon it.

GRF’s are useful markers in the science of biomechanics in order to assess the risk of musculoskeletal injuries. GRF patterns have been used in biomechanics for the assessment of normal or pathological gait (Munro, Miller, & Fuglevand, 1987). It is well known that these forces may increase tremendously during running and can be three times larger than the individual’s weight. Furthermore, repeated development of unusual high GRFs may largely increase the risk of injury and osteoarthritis (Bray, 2014; Cavanagh & Lafortune, 1980).

Obesity

Obesity is a major health problem in most developed countries and it seems that the incidence of the condition is increasing at an alarming rate. Obesity increases the risk of developing several medical conditions such diabetes, OA, cancer, cerebrovascular diseases, respiratory diseases and a plethora of musculoskeletal problems (Hills, Hennig, Byrne, & Steele, 2002).

Studies have shown that ageing is highly associated with obesity and overweight. Ageing is also associated with changes in body fat distribution and more specifically it seems that ageing increases the accumulation of intra-abdominal fat (Ochi et al., 2010). Lovejoy, Champagne, De Jonge, Xie, and Smith (2008) support that, postmenopausal women tend to have more abdominal fat mass than premenopausal women. Estrogens seem to play a role in the distribution of body fat in women and the high concentration of abdominal fat. According to previous research studies, exogenous oestrogen administration in postmenopausal women may result in lower waist-to-hip ratios and less visceral adipose tissue than women without the oestrogen administration (Lovejoy et al., 2008).

Obesity and Osteoarthritis

Osteoarthritis is a degenerative condition that leads to loss of the articular cartilage and degradation of the periarticular bone. Previous research supports that; OA may be caused by abnormal biomechanics of the affected joint, biochemical factors or even genetic predisposition. The biomechanical explanation of Janssen and Mark (2006) supports that, excessive weight places a higher than normal load on the joints during certain activities. This may explain why abdominal obesity increases the risk of OA (Janssen & Mark, 2006).

Research has shown that obesity increases the risk of knee Osteoarthritis (OA). Obese people demonstrate increased GRFs which may explain the effect that obesity has on human joints. Researchers support that obesity increases knee joint loading that leads to activation of chondrocyte mechanoreceptors. This results in the activation of cytokines, growth factors and metalloproteinases. The activation of these factors block matrix synthesis and leads to cartilage degeneration and OA (Bray, 2014).

Messier (1994) supports that; a certain mechanism based on certain factors may result in an increased rate of loading of the joint. Messier (1994) states that repeated uncontrollable impulse loading of the bone increases the risk of micro-fractures that make the bone stiff during the period of healing. It also decreases the period of eccentric tension of the quadriceps, during heel-strike, and creates a mechanism with a combination of rapid muscle fatigue and altered ground reaction forces (Messier, 1994). This mechanism increases the rate of loading on the knee joint as the musculature serves as a shock absorbing mechanism and especially the quadriceps muscle (Hills et al., 2002; Messier, 1994; Syed & Davis, 2000).

Furthermore, in obese people and especially in people with increased concentrations of abdominal fat mass, insulin resistance is highly likely to occur (Carey, Jenkins, Campbell, Freund, & Chisholm, 1996; M. R. Sowers & Karvonen-Gutierrez, 2010). M. R. Sowers and Karvonen-Gutierrez (2010) support that, insulin resistance is associated with muscle mass loss, increased fat mass and as a result higher risk of osteoarthritis onset.

Overweight women are at greater risk of developing knee OA. However, there is no clear mechanism to explain this trend (Syed & Davis, 2000). As stated earlier, the quadriceps may serve as a protective mechanism for the integrity of the knee joint. (Jefferson, Collins, Whittle, Radin, & O'Connor, 1990). However, research is not exhaustive and several authors have stressed the need for more research evidence on this matter (Fisher, White, Yack, Smolinski, & Pendergast, 1997; Syed & Davis, 2000).

Heel-Strike Transient (HST)

At heel strike the vertical impact forces initiate transient stress waves and these travel through the lower extremities up to the kinetic chain (Collins & Whittle, 1989). These transient stress waves are called heel strike transients (Levinger & Gilleard, 2005). HST’s are generated from the sudden impact of the heel with the ground surface. The magnitude of these ground reaction forces are largely depend on gait speed, footwear, hardness of the ground surface and the angle and velocity of the foot (Lafortune & Hennig, 1992; Levinger & Gilleard, 2005; Whittle, 1999). Additionally Whittle (1999) states that, the size of the HST may be influenced even by the mood of the participant.

The HST is seen as short spike of force which usually lasts 10-20 milliseconds, superimposed on the upslope of the GRF, immediately following initial contact of the foot with the ground (Whittle, 1999). Force platforms are mainly used to identify HSTs from the vertical component of the GRF (Collins & Whittle, 1989). Collins and Whittle (1989) suggest that, HSTs have been ignored in previous studies because researchers subjected their data from their studies to low pass filters or ignored HSTs. Also, some researchers sampled at too slow rated and they were unable to detect high-speed HSTs. Finally, some others used force platforms with low resonant frequencies which did not allow them to detect high frequency HSTs (Collins & Whittle, 1989).

HST's may appear in approximately one third of the general population during walking and the reason for their existence in some people is still unclear (Mikesky, Meyer, & Thompson, 2000). However, it has been previously suggested that, the HST is the result of increased rates of loading. Furthermore, previous research supports that there is a positive association of the HST with the aetiology and progression of OA (Collins & Whittle, 1989; Mikesky et al., 2000).

Up to date most studies have tried to identify possible correlations between the existence of the HST and muscular weakness of the lower limb as well the development or existence of OA (Levinger & Gilleard, 2005; Liikavainio et al., 2007). However, conflicting evidence exist in the literature in regards to the above statements. For example, Mikesky et al. (2000) supported through their study that, the HST is much more likely to occur in women with weak leg muscles than in a strength-trained group of women. Also, the study of Jefferson et al. (1990) showed that, paralysis of the quadriceps muscle largely increases the size of the HST. Surprisingly, the study of Hunt et al. (2010) showed that, when walking at a freely chosen walking speed, strength was not found to be associated with the existence or size of the HST in a group of people with knee OA.

Osteoarthritis in women

Many epidemiologic studies have shown that the risk of OA is the same in men and women under the age of 50. However, there is a large increase in the prevalence and incidence of knee OA in women after the age of 50 (Zhang et al., 1998). It is well known that a relationship between hormonal changes from menopause and knee OA exists (Hanna, Wluka, Bell, Davis, & Cicuttini, 2004). This incidence was initially described by Kellgren and Moore (1952) as ‘’menopausal arthritis’’. Although the reason for this incidence remains controversial, it is now widely accepted that the risk of knee OA increases in perimenopausal women (Hanna et al., 2004). However, there is a lack of evidence to suggest that by increasing the levels of oestrogen may also reduce the incidence of knee OA in postmenopausal women (de Klerk et al., 2009; Janssen & Mark, 2006).

Little is known about the effects of fat distribution on impact loading of the foot and on the size of HST. Since there is an alteration of fat distribution in postmenopausal women, along with an increase in the prevalence and incidence of knee OA, it is of significant importance to identify possible mechanisms that may or may not affect joint integrity. Biochemically it is widely know that adipose tissue is not only a passive store of energy, but also an endocrine organ that excretes certain factors that have shown to be destructive for the joint cartilage (Wang et al., 2009). However, this does not explain the fact that only postmenopausal women are at a higher risk for developing knee OA compared with men.

The biomechanical factors are evident at the knee since it is an unstable joint compared with other joints of the lower extremity and the load is disproportionally distributed medially during dynamic activities. Although the association between fat distribution and knee OA has been investigated in previous research studies, there is a lack of consistency on the results (Wang et al., 2009).

Purpose

The purpose of this study was to investigate how an external abdominal weight may affect the GRFs and especially the HST when walking at freely chosen high and low speeds. The aim of this study was to identify a possible mechanism that increases the risk of knee OA in women with abdominal obesity.

Hypothesis

The initial hypothesis of this study was to determine whether there is a relationship between abdominal weight and the HST in healthy adults when walking. The following null hypotheses were tested in this study. There is no significant relationship between increased abdominal weight and HST in a population of healthy adults. There is no significant relationship between increased abdominal weight and the first maximum force peak, the total contact time, the time to first peak and lastly the loading rate.

Delimitations

1. All subjects will be apparently healthy and mobile without a history of any neuromuscular, musculoskeletal, physiological or neurological disease.
2. People with a history of previous surgery on the lower limbs will be excluded from the study.
3. Age will range from 18-50.
4. BMI will range from 18.5-30 kg/m2.
5. Analysis will be performed on the dominant lower extremity of all subjects.

Limitations

1. Accuracy of the Bertec Force Plate
2. Subjects may adjust their normal walking pattern when placing external weights on them

Literature Review

A review of the literature was carried out using SCOPUS as the main research database. The results showed no other studies looking into the effects of abdominal obesity on the size of the HST. Only a few studies have tried to find an association between abdominal obesity or obesity and the GRFs. Furthermore, some researchers have tried to indirectly find correlations with the results of increased GRFs and rates of loading at the knee joint such as the correlations between abdominal obesity and knee OA. Thus, this chapter will review the literature with a focus on similar aspects that make up the title of this study.

Shock-Absorbing Mechanisms

The human body has a number of mechanisms that reduce impact forces. Radin et al. (1973) support that, bony deformation, filtration through cancellous bone, joint motion and muscle lengthening under tension should act as shock absorbers from the human body. The subchondral bone attenuates forces that act inside the joint cartilage and protects the cartilage from degeneration by absorbing impact forces (Radin & Paul, 1970; Radin, Paul, & Tolkoff, 1970). However, these shock-absorbing mechanisms may be lost or become less effective when high rates of loading pass through the joints (Radin & Paul, 1971) and as we age due to the formation of crystal contents in the cartilage that make them less effective (Currey, 1979).

Previous research studies have shown that thigh muscles may attenuate ground reaction forces and reduce the risk of degenerative pathology (Slemenda et al., 1997).

Muscular atrophy that occurs with aging and sedentary lifestyle may therefore negatively affect the protective role of the muscles at the knee joint. Since menopausal transition has been linked with a marked reduction in muscle protein synthesis there is some research evidence to support that this factor could play a significant role on the size of the GRFs in postmenopausal women. However, if this was the only factor for the higher incidence of knee osteoarthritis in women, then the exogenous administration of oestrogens should have had alter this phenomenon in the study of Felson et al. (1987). The results of their study showed that oestrogen administration did not significantly alter the risk and progression of radiographic knee OA. Although the influence that fat distribution has at the knee joint, in postmenopausal women, has been questioned previously by Syed and Davis (2000), to the researcher’s knowledge the interrelationship between the size of the vertical GRF, and especially the HST, and fat distribution has not been adequately investigated.

Obesity and GRF

Previous research in obese adults has shown that increased body weight results in greater GRFs. Browning and Kram (2007) compared the GRF values of 10 obese and 10 normal-weight individuals as they walked on a level, force measuring treadmill in different speeds. They observed that the GRFs were much higher in the obese individuals and were substantially increased in higher walking velocities. Furthermore, Messier et al. (1996) support that BMI was directly related with higher vertical forces and loading rates.

The association of high loading rates and knee damage has been previously investigated by Radin, Yang, Riegger, Kish, and O'Connor (1991). They found that people with knee pain had 37% greater loading rates compared with a group of people without knee pain. Also, the knee pain group demonstrated greater downward angular velocities of the ankle and the shank, frequent violent knee hyperextension, larger impacts of the heel with the ground and weaker eccentric contractions of the quadriceps muscle. It was later suggested by the authors of the study that these parameters reduced the shock absorbing properties of the limb (Radin et al., 1991). Not only muscle strength may influence joint integrity but also knee malalignment may affect load distribution (Hulth & Radin, 1993).

Fat distribution and OA

In the past, several researchers have tried to discover if knee degeneration may be also caused by the way that the body-fat is distributed. Wang et al. (2007) showed that waist circumference and waist-to-hip ratio were positively associated with the presence of cartilage defects. Moreover, large waist circumference from intra-abdominal fat accumulation has been associated with low quality of life and poor physical functions (Han, Tijhuis, Lean, & Seidell, 1998). The study of M. F. Sowers et al. (2008) compared data from a group of women with and without radiographic knee OA. They found that one of the risk factors for radiographic knee OA was waist circumference since it was significantly greater in women with knee OA.

However, further research that has been carried out contradicts the above findings and suggests that people with a lower fat distribution may be more biomechanically disadvantaged than the people with central obesity (Foster et al., 2010). Foster et al.

(2010) support that, lower fat distribution alters gait, causes knee misalignment and increases knee adduction moment. The study of Foster et al. (2010) showed that there is no significant association between body fat distribution and physical functions in people with or without knee OA. Similarly Abbate et al. (2006) assessed fat distribution with DXA scans in 779 women and found no association between body-fat distribution and radiographic knee OA.

When we consider these findings it seems abundantly clear that a relationship may exist between high GRFs and central obesity. In this research it was considered plausible that the influence that central obesity has on human posture, which makes it more lordotic than usual, may have a direct effect on peoples' gait. Increased lordosis may actually increase heel-walking in order to attenuate the vertical downward forces that act on the body from the extra abdominal weight. However, heel walking has been proposed to produce large HSTs which are linked with deleterious degenerative diseases of the knee joint (Levinger & Gilleard, 2005).

The HST and OA

The HST has been questioned by several researchers and it has been proposed that the size of the HST may be influenced by the gait technique of the individual (Collins & Whittle, 1989). Furthermore, it has been proposed that except from quadriceps activation the HST may be eliminated by mid-foot striking or special running footwear (Dickinson, Cook, & Leinhardt, 1985).

The HST has been observed in previous research studies in one third of the American population along with rapid loading rates at heel-strike (Jefferson et al., 1990; Radin et al., 1986). It has been proposed by the investigators that the existence and the size of the HST depend on the ability of the individual to decelerate the lower limb at heel-strike by activating adequately the quadriceps muscle. However, using the floor to decelerate the lower limb may produce large HSTs and potentially damaging forces (Radin et al., 1986).

Although pathology at the knee has been associated with greater HSTs, the study of Levinger and Gilleard (2005) showed that in a group of women with patellofemoral pain syndrome the HST magnitude was lower and occurred later when compared with the control group. It was clarified that the difference was not due to altered gait velocities. This finding is in contrast with previous research studies which showed that knee pain and pathology may increase external vertical forces (Collins & Whittle, 1989; Folman, Wosk, Voloshin, & Liberty, 1986; Simon et al., 1981). This may suggest that the size and presence of the HST may not directly relate to knee pathology or pain but other biomechanical factors may play a role. For example foot pronation has been proposed to act as a shock absorption mechanism that attenuates impact forces (Blake & Ferguson, 1991; Subotnick, 1975).

Proprioception has also been investigated in order to identify whether it can affect the size of the HST in people with injuries. Co, Skinner, and Cannon (1993) investigated the effects of anterior cruciate ligament reconstruction on knee proprioception and the HST. They found that the size of the HST is not affected by the proprioception of the knee. The investigators support that, the subjects with the reconstructed ACL were athletic and that strengthens the notion that heel-strike transient is affected by quadriceps strength.

Thus, based on the evidence above, it seems beneficial to investigate whether there is an association between high rates of loading and HST presence in people with intra-abdominal fat accumulation and whether the abdominal fat changes the vertical GRFs in healthy people. This investigation will allow the investigators to identify possible mechanisms that alter body posture and whether this may in fact be responsible for the higher risk of knee OA in postmenopausal women. In order to assess whether abdominal weight may modify peoples' gait and increase collision forces of the heel with the ground it was considered reasonable to assess the effects of abdominal weight on the size of the HST, maximum vertical forces and loading rates.

Materials and Methods

Subjects

Ten subjects (7 male and 3 female) were recruited for this study from the Department of PE and Sports Science at the University of Thessaly in Trikala. The listserve of the University of Thessaly was utilized in order to send e-mails to the students of the department. Subjects were also recruited from class announcements that took place at the department to ensure that at least 10 subjects would be recruited. All procedures were reviewed and approved from the Internal Ethics Committee of the Department. The subjects selected for this study ranged in age from 18 to 43 years.

Due to the weight carriage portion of the study, a significant number of exclusion criteria were given. Participants were excluded from the study if they had an injury or surgical treatment of the lower limbs or the spine, painful knees and developmental or congenital abnormalities of the lower limbs. Furthermore, subjects with other medical problems that had caused neurological problems, such as paralysis, paresis of the lower limbs or other neurological conditions that had changed their normal gait pattern were also excluded. Lastly, other potential participants were informed that they will not be able to participate if they had suffered from any of the following medical conditions: cerebrovascular disease, hernia, rheumatoid arthritis or any other autoimmune disease that affects normal gait pattern, spinal stenosis and chronic or acute back pain.

The subjects were informed about the exclusion criteria and the investigator explained the procedures that would be employed during the study. Moreover, the subjects were encouraged to ask any questions regarding the procedures or the intentions of the investigator. If they were eligible to participate in the study they were asked to sign a consent form. A copy of the consent form can be found in appendix 1. The subject's age, height and weight were (mean ± st. deviation) 27.5±8.84, 174.6±3.83 and 71.5±6.36 respectively. Demographic data of the participants are listed in table 1.

Subject No.	Gender	Age (years)	Weight (kg)	Height (cm)	BMI	Waist-to-hip ratio
1	Female	25.00	63.00	168.00	22.30	.77
2	Male	43.00	77.00	180.00	23.80	.90
3	Male	18.00	70.00	175.00	22.90	.78
4	Male	24.00	65.00	173.00	21.70	.73
5	Female	24.00	68.00	170.00	24.20	.72
6	Male	28.00	78.00	174.00	25.80	.81
7	Male	28.00	82.00	177.00	26.20	.86
8	Male	18.00	66.00	174.00	21.80	.79
9	Male	24.00	76.00	175.00	24.70	.83
10	Female	43.00	70.00	180.00	21.60	.75

Table 1. Demographic data of subjects

Equipment and procedures

Force plate

In this lab, a Bertec force plate was utilized to measure the GFRs of the subjects. Similar to a scale, a force platform calculates the forces that exerted by the ground in opposition to the weight on it. This force plate is designed to measure the forces in three perpendicular axes and moments and the associated axes. The force plate is a rigid plate that is attached to four instrumented pedestals. When an external load is applied in the force plate, deforms each pedestal at a different amount and that depends on the load direction and location. In this Bertec force plate a strain gauge transducer which is placed in each pedestal measures the deformation that may occur. The forces and moments acting on the plate can be calculated as the force plate geometry and pedestal locations are known. The manufactures of the force plate provides the calibration parameters that are used to convert the voltages measured by the transducers into force and moment measurements (Moir, 2008; Raymakers, Samson, & Verhaar, 2005).

There is also additional equipment that interconnects with the force plate which is essential in order to provide the investigators of the study with the measurements needed. The output from the force plate is digitized and the input that is used by the systems is analog. The digitized output is converted to analog and amplified by a converter. A data acquisition box is used to send digital signals to a computing program. The investigators acquired force plate data at a rate of 1000Hz. The output returned to the system in a text file that included all the necessary measurements and a title in order to know what data have been taken (Moir, 2008; Raymakers et al., 2005).

The biomechanical gait analysis was performed using a 9.5 m level walkway instrumented with a Bertec force plate in order to measure the magnitude and the direction of the GRF applied by the foot to the ground (Collins & Whittle, 1989). The force plates can provide sufficient data for the GRF in three directions. These are: the vertical, the medio-lateral and the anterior-posterior. From these three the vertical is the largest force that can be seen and has been traditionally portrayed as a uniform two two-peak curve. This two peak curves are the result of weight support and the vertical acceleration of the individual’s body (Collins & Whittle, 1989).

The HST may also be seen by using force plates at the initiation of the stance phase. The HST has been previously ignored by several investigators who actually used force plated to collect GRF data. The reasons for this have been described by Collins and Whittle (1989) who state that investigators who collected data from force plates with insufficient high frequency, sampled at a low rate and with excessive low pass filtering were unable to detect the HST.

Experimental Procedure

Testing equipment was calibrated before testing. Weight was measured on the force platform in Newtons and height was measured on a stadiometer. The subjects walked at a self-selected low and high speed over the force platform with their dominant foot three times. The trials were accepted only if the entire foot landed on the force platform and only if the gait pattern was not visibly altered during the trials. The subjects were instructed to walk freely along the walkway without targeting the force plate. Although Liddle, Rome, and Howe (2000) suggest that, subjects should avoid targeting the force plate in order to avoid the creation of unnatural gait patterns, the study of Grabiner, Feuerbach, Lundin, and Davis (1995) showed that visual guidance does not influence GRF variability. Thus, it was not considered essential for the subjects to wear special goggles in order to reduce peripheral vision and eliminate force plate targeting as it has been described by Riskowski, Mikesky, Bahamonde, Alvey Iii, and Burr (2005). The subjects were instructed to fix their gaze forward in level with their eyes and to avoid looking down at the force plates.

The subjects in this study were healthy and fit enough in order to practice their walking trials numerous times and ensure that they were familiar with the walkway dimensions and placement of the force platforms. The initial trials were performed in order to ascertain the appropriate starting point for each subject and ensure that the dominant leg would strike on the force platform. The subjects were asked to perform this procedure three times without the external weight. Then a special device was given that was worn that had enough space to fit 5, 10 and 15 kg plates. The subjects were asked to perform three trials with the 5, 10 and 15 kg external weights that were worn in front of their abdomen. This protocol was performed at a low walking speed which simulated their normal daily relaxed walking and at a higher speed which simulated their rushed walking. The subjects also performed practice trials to ensure that their foot would strike the force plate when the weights were given and ascertain the appropriate starting point (Liddle et al., 2000).

The time taken to reach the force plate without the external weight and with each external weight was manually controlled to detect any alterations in gait velocity. An independent observer was always present to ensure that measurements were taken under the same environmental conditions (Liddle et al., 2000). In order to accept the data taken the observer ensured that the dominant foot entirely contacted the force plate, subjects did not target the force plate and did not altered their gait pattern in order to reach the force plate, and sampling time was adequate to record data for the total foot contact period (Liddle et al., 2000).

The investigator collected the force plate data at a sampling rate of 1000 Hz and the force plates were activated manually, after instructing each subject to start his/her walking trial at a specific starting point. The software provided numerous data that were transferred to Microsoft Excel 2007 and for each valid trial the programme produced a force curve graph. The GRF data generated by the subjects were normalized to body weight for the trials that the external weight was not included. For the trials that the external weight was included, the investigator normalized the data to body weight plus the weight of the external load.

Each trial was coded for subject number and for the existence and size of the external weight. Mean readings were taken for several parameters and these are: the HST, the first maximum force peak (FMFP), the second maximum force peak (SMFP), the time taken for the FMFP to occur and the total contact time as detailed by Liddle et al. (2000) (Fig. 1). Also, the time to first peak along with the loading rate of the first peak maximum were calculated by dividing the FMFP by the time to FMFP as described by Roberto Bianco (2011) and Liddle et al. (2000).

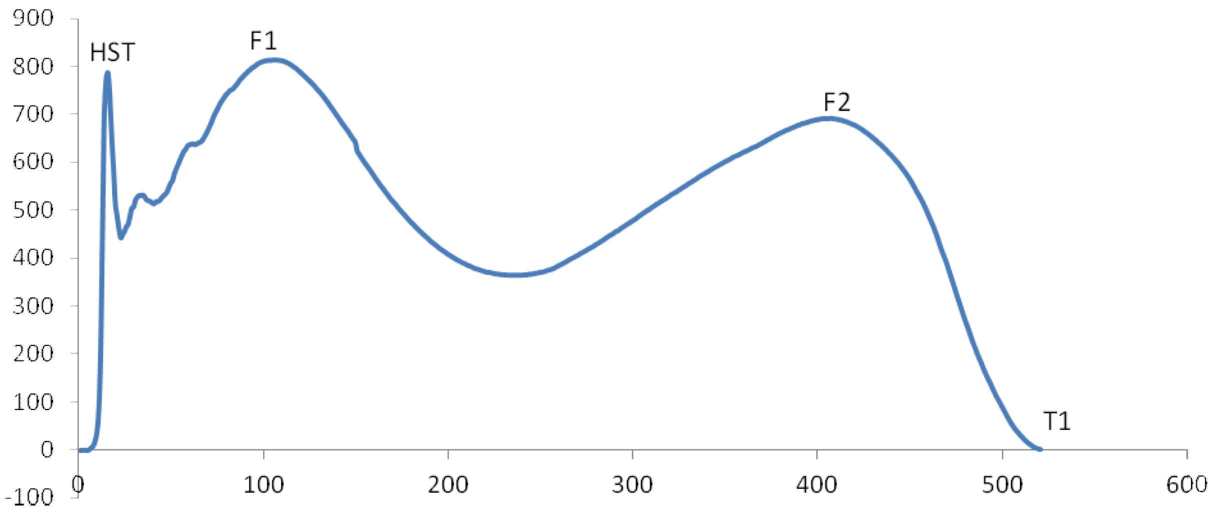


Figure 1. Force/time curve demonstrating the HST, the first maximum force peak (F1), the second maximum force peak (F2) and the total contact time (T1).

Experimental Design and Statistical Analysis

The aim of this study was to discover statistically significant differences in the GRF components, the total contact time and the loading rates of the HST and the first maximum force peak between the several alterations of the external weight that was given and between the different walking velocities. A total of 240 trials were analyzed. Data were exported to MS Excel 2007 in order to derive the dependent variables needed. A graph was generated for each trial and the values were obtained from each graph. The mean value of the three accepted trials was calculated and compared with the appropriate statistical test using SPSS 18 for Windows.

The dependent variables were calculated and defined as follows: The HST was defined as the vertical force recorded prior to the usual, expected peak vertical force derived from heel strike, calculated via the force plate. A specific way to identify HSTs is not available in the literature and most researchers use their own rational to measure HSTs (Hunt et al., 2010). The maximal vertical force recorded after the obvious HST (if the HST is present), calculated via the force plate. The total contact time as it was evident by the graph, calculated via the force plate. The loading rates were calculated by dividing the maximal vertical force by the time to the maximal vertical force (Puddle & Maulder, 2013).

In order to choose the appropriate test for comparison for the several different trials the normality of the data was tested with the Kolmogorov-Smirnov Test. Means and standard deviations were also calculated. The one-way repeated measures analysis of variance (ANOVA) test was found to be the appropriate test for each dependent variable. This test was used to discover statistical significant differences between the external weights and then between the different velocities for the HST, the first maximum force peak, the total contact time and the loading rates. Differences were considered significant at the 5% level.

Results

Normality Tests

Many statistical tests require the response variables being analyzed to be normally distributed. Normality tests indicate whether appropriate statistical tests are being utilized. From the Kolmogorov-Smirnov test it was found that the HST data, the first vertical peak maximum and the contact time were normally distributed. However, the time to first peak, the loading rates and the second peak maximum were not normally distributed.

Friedman's ANOVA

Since the data did not appear to be normally distributed, a nonparametric test was used. The Friedman's ANOVA is the nonparametric alternative to the one-way ANOVA with repeated measures. The Friedman's ANOVA was used to evaluate differences across several different measurement conditions for each variable. The Friedman's ANOVA test showed a statistical significant difference at the HST in the 8 different conditions measured ($p = 0.000$). Post-hoc analysis with Wilcoxon signed-rank test was conducted for all the variables tested with a Bonferroni correction applied, resulting in a significance level set at $p < 0.00625$.

HST statistical analysis

The mean and standard deviation show that there was mainly an increase at HST when gait velocity increased (table 1). The statistical significant difference (p=.000) was due to the difference in gait velocity (table 2).

HST Condition	Mean ± St. Deviation
HSTSL 0	0.79 ± .18
HSTL 5	0.78 ± .18
HSTSL 10	0.79 ± .16
HSTL 15	0.71 ± .31
HSTFA 0	1.08 ± .16
HSTFA 5	1.09 ± .20
HSTFA 10	1.06 ± .18
HSTFA 15	1.08 ± .20

Table 1. HST Characteristics (Mean ± Standard Deviations), (HSTSL–HST Slow, HSTFA–HST Fast)

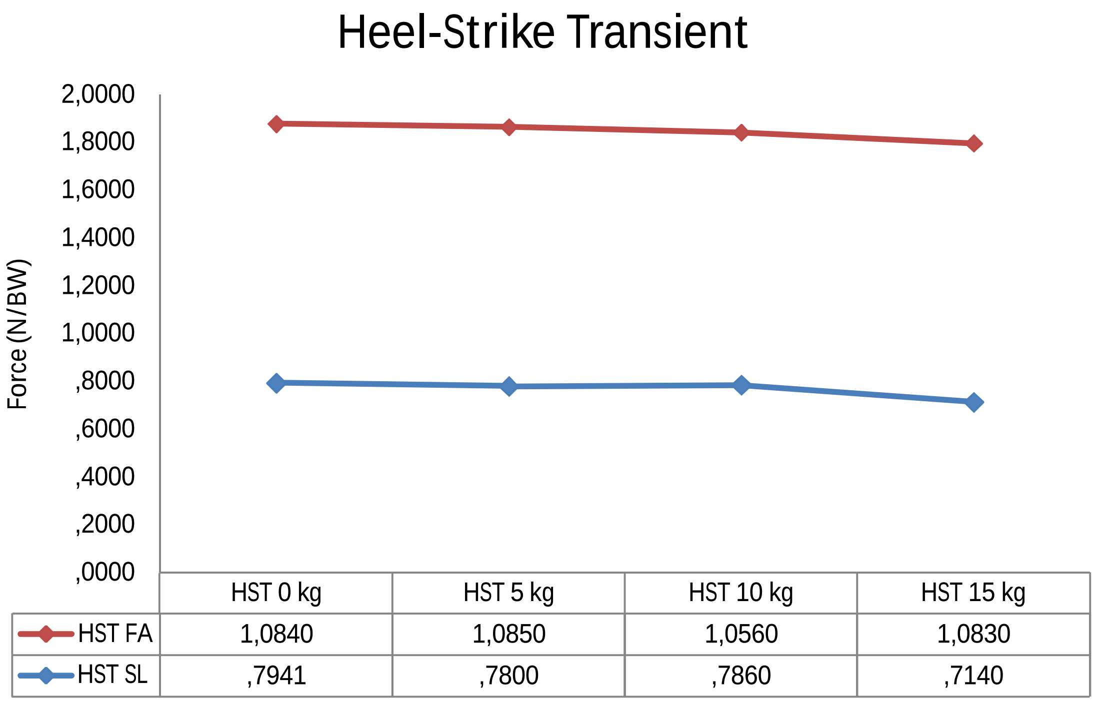


Diagram 1. The Mean Heel-Strike Transient

HST comparison	Asymp. Sig (2-tailed)
HSTSL0-HSTSL5	0.357
HSTSL0-HSTSL10	0.959
HSTSL0-HSTSL15	0.646
HSTSL0-HSTFA0	0.005
HSTSL0-HSTFA5	0.005
HSTSL0-HSTFA10	0.005
HSTSL0-HSTFA15	0.005
HSTSL5-HSTSL10	0.233
HSTSL5-HSTSL15	0.953
HSTSL5-HSTFA0	0.005
HSTSL5-HSTFA5	0.005
HSTSL5-HSTFA10	0.005
HSTSL5-HSTFA15	0.005
HSTSL10-HSTSL15	0.507
HSTSL10-HSTFA0	0.005
HSTSL10-HSTFA5	0.005
HSTSL10-HSTFA10	0.005
HSTSL10-HSTFA15	0.005
HSTSL15-HSTFA0	0.005
HSTSL15-HSTFA5	0.005
HSTSL15-HSTFA10	0.005

HSTSL15-HSTFA15	0.005
HSTFA0-HSTFA5	0.213
HSTFA0-HSTFA10	0.283
HSTFA0-HSTFA15	0.798
HSTFA5-HSTFA10	0.475
HSTFA5-HSTFA15	0.760
HSTFA10-HSTFA15	0.575
Significant at p=0.00625	

Table 2. Results of Wilcoxon Signed Ranks Test (HSTSL - Heel-strike Transient Slow Gait 0kg, 5kg, 10kg, 15kg), (HSTFA - Heel-Strike Transient Fast Gait 0kg, 5kg, 10kg, 15kg)

First Maximum Force Peak

The mean and standard deviation for the first maximum force peak shows that there was a gradual increase for the FMFP as the external weight was increasing along with the gait velocity (Table 3). The statistical significant difference (p=.000) was evident in several comparisons (table 4).

Vertical GRF Max Condition	Mean Vertical GRF Max ± St. Deviation
GRF Max SL 0	1.19 ± .10
GRF Max SL 5	1.24 ± .10
GRF Max SL 10	1.27 ± .11
GRF Max SL 15	1.30 ± .13
GRF Max FA 0	1.39 ± .13
GRF Max FA 5	1.47 ± .13
GRF Max FA 10	1.50 ± .18
GRF Max FA 15	1.51 ± .19

Table 3. Vertical GRF Max Characteristics (Mean ± Standard Deviations) (SL-Slow, FA-Fast)

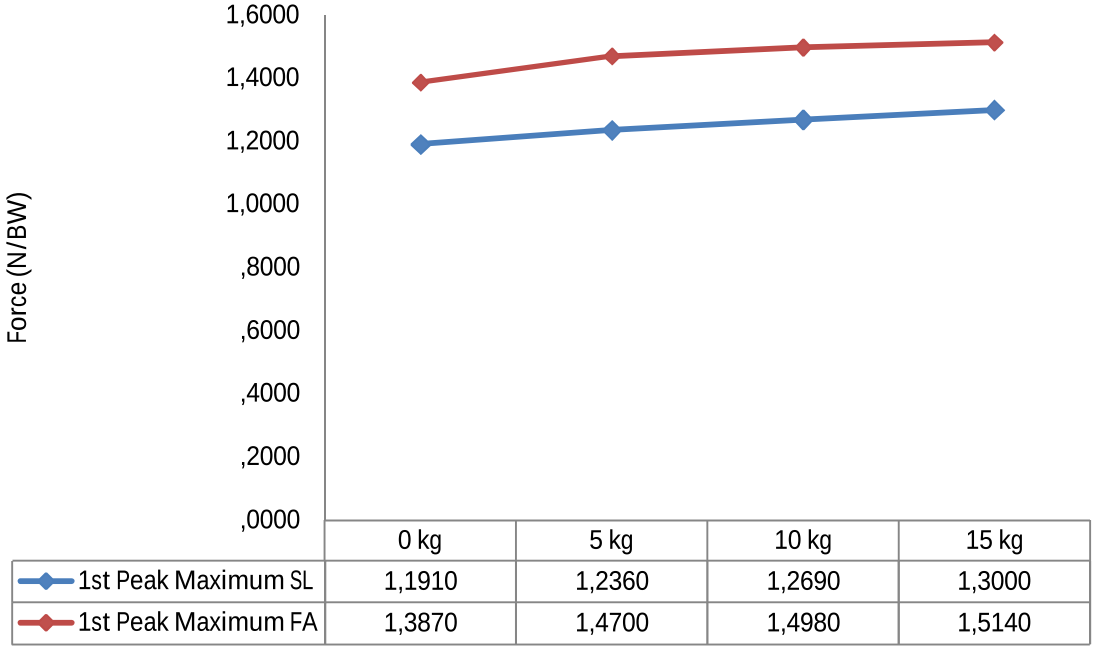


Diagram 2. The 1st Vertical Force Peak Max (SL-Slow, FA- Fast)

GRF Max Comparison	Asymp. Sig (2-tailed)
GRFMaxSL0 – GRFMaxSL5	0.046
GRFMaxSL0 – GRFMaxSL10	0.005
GRFMaxSL0 – GRFMaxSL15	0.005
GRFMaxSL0 – GRFMaxFA0	0.005
GRFMaxSL0 – GRFMaxFA5	0.005
GRFMaxSL0 – GRFMaxFA10	0.005
GRFMaxSL0 – GRFMaxFA15	0.005
GRFMaxSL5 – GRFMaxSL10	0.052
GRFMaxSL5 – GRFMaxSL15	0.011
GRFMaxSL5 – GRFMaxFA0	0.007
GRFMaxSL5 – GRFMaxFA5	0.005
GRFMaxSL5 – GRFMaxFA10	0.005
GRFMaxSL5 –	0.005

GRFMaxFA15	
GRFMaxSL10 – GRFMaxSL15	0.138
GRFMaxSL10 – GRFMaxFA0	0.007
GRFMaxSL10 – GRFMaxFA5	0.005
GRFMaxSL10 – GRFMaxFA10	0.005
GRFMaxSL10 – GRFMaxFA15	0.005
GRFMaxSL15 – GRFMaxFA0	0.033
GRFMaxSL15 – GRFMaxFA5	0.005
GRFMaxSL15 – GRFMaxFA10	0.007
GRFMaxSL15 – GRFMaxFA15	0.009
GRFMaxFA0 – GRFMaxFA5	0.005
GRFMaxFA0 – GRFMaxFA10	0.005
GRFMaxFA0 – GRFMaxFA15	0.007
GRFMaxFA5 – GRFMaxFA10	0.332
GRFMaxFA5 – GRFMaxFA15	0.139
GRFMaxFA10 – GRFMaxFA15	0.553
Significant at p=0.00625	

Table 4. Results of Wilcoxon Signed Ranks Test (GRFMaxSL – GRF Maximum Slow Gait 0kg, 5kg, 10kg, 15kg), (GRFMaxFA – GRF Maximum Fast Gait 0kg, 5kg, 10kg, 15kg)

Second Maximum Force Peak

The mean and standard deviation for the SMFP (Table 5) and the statistical significant difference (p=.000) was evident in several comparisons (table 6). In this comparison the difference in the GRF was not due to gait velocity. Table 6 shows that there were significant differences due to the external weights in the SMFP and not due to gait velocity. Interestingly, in contrast with the HST and the FMFP, adding more weight and increasing gait velocity did not increase the SMFP but resulted in a small decrease in most subjects.

2nd Maximum Force Peak	Mean 2^{nd} Maximum Force Peak ± St. Deviation
SL 0	1.15 ± .09
SL 5	1.15 ± .10
SL 10	1.11 ± .07
SL 15	1.06 ± .09
FA 0	1.18 ± .14
FA 5	1.12 ± .14
FA 10	1.05 ± .13
FA 15	1.00 ± .13

Table 5. Vertical GRF Max Characteristics (Mean ± Standard Deviations)

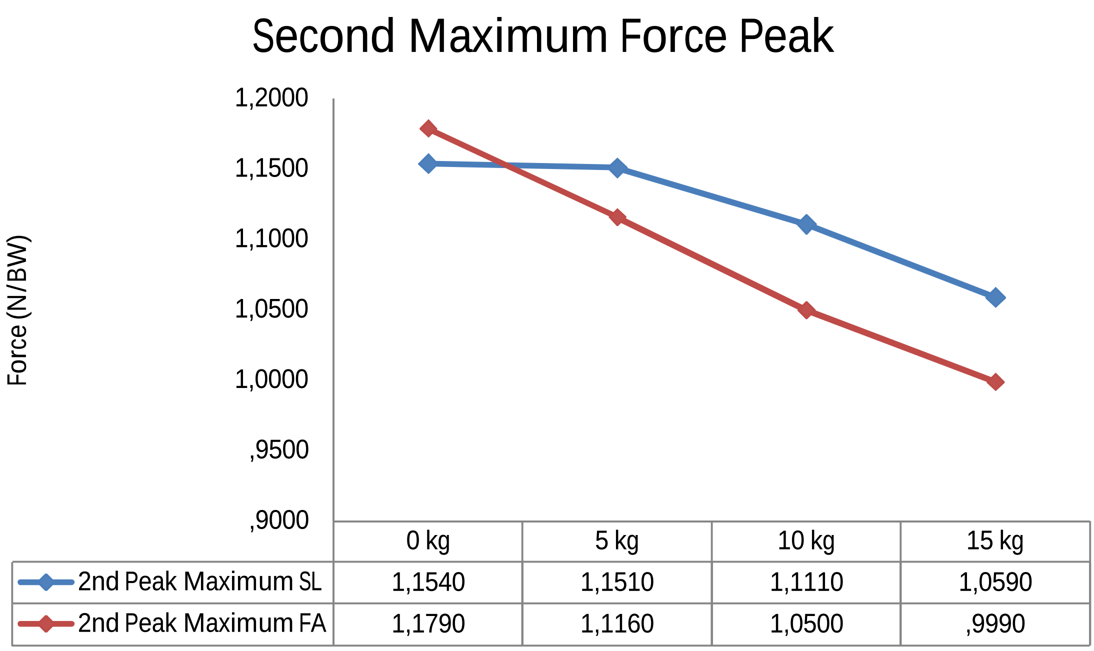


Diagram 3. The 2^{nd} Vertical Force Peak (SL-Slow, FA- Fast)

2^{nd} Maximum Force Peak comparison	Asymp. Sig (2-tailed)
SL0 – SL5	0.918
SL0-SL10	0.052
SL0-SL15	0.007
SL0-FA0	0.507
SL0-FA5	0.343
SL0-FA10	0.021
SL0-FA15	0.005
SL5-SL10	0.021
SL5-SL15	0.005
SL5-FA0	0.202
SL5-FA5	0.307
SL5-FA10	0.012
SL5-FA15	0.005
SL10-SL15	0.005
SL10-FA0	0.075
SL10-FA5	0.720
SL10-FA10	0.044
SL10-FA15	0.012
SL15-FA0	0.028
SL15-FA5	0.074
SL15-FA10	0.720
SL15-FA15	0.066
FA0-FA5	0.005
FA0-FA10	0.014
FA0-FA15	0.009
FA5-FA10	0.041
FA5-FA15	0.047
FA10-FA15	0.032
Significant at p=0.00625	

Table 6. Results of Wilcoxon Signed Ranks Test for the 2^{nd} Vertical Peak Maximum (SL-Slow, FA-Fast)

Time to First Peak

The time taken to reach the first maximum vertical force peak was significantly reduced only when the subjects increased their gait velocity. Again, the statistical significant reduction (p=.000) of the mean time to first peak was mainly due to an increase in the gait velocity.

Time to First Peak Condition	Mean Time to First Peak ± St. Deviation
SL0	.1267 ± .04397
SL5	.1383 ± .02329
SL10	.1333 ± .01673
Sl15	.1350 ± .01854
FA0	.1002 ± .02728
FA5	.1076 ± .01855
FA10	.1077 ± .02281
FA15	.0979 ± 02514

Table 7. Time to First Peak (Mean ± Standard Deviations)

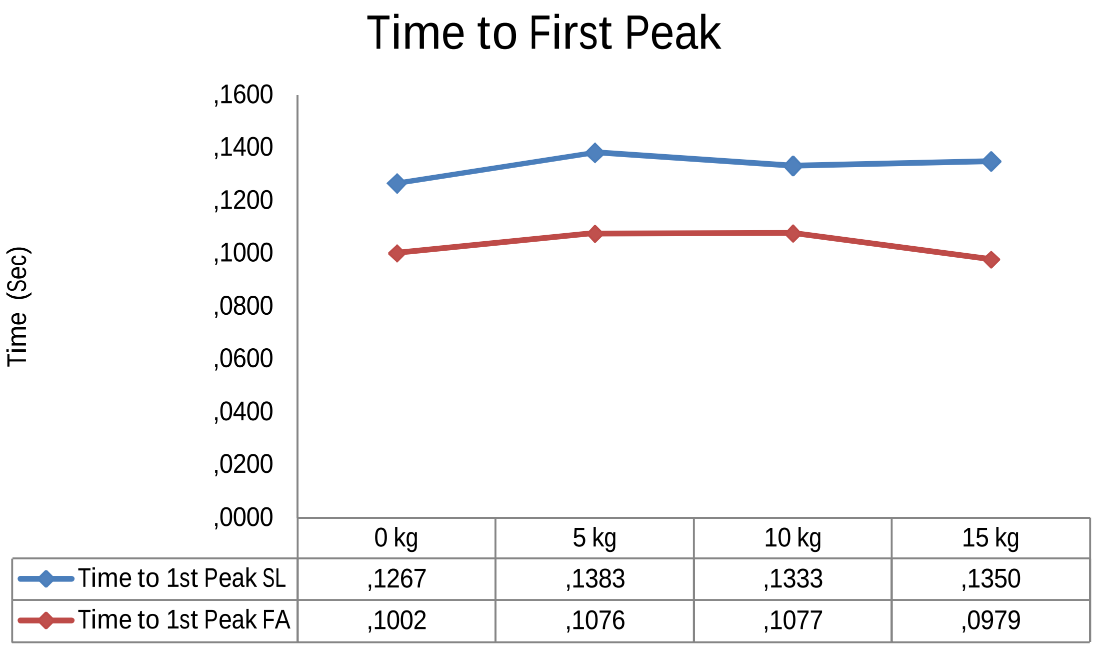


Diagram 4. The Time to First Peak (SL-Slow, FA- Fast)

Time to First Peak Comparison	Asymp. Sig (2-tailed)
SL0 – SL5	.906
SL0-SL10	.878
SL0-SL15	.959
SL0-FA0	.009
SL0-FA5	.074
SL0-FA10	.074
SL0-FA15	.074
SL5-SL10	.308
SL5-SL15	.445
SL5-FA0	.005
SL5-FA5	.009
SL5-FA10	.007
SL5-FA15	.005
SL10-SL15	.575
SL10-FA0	.005
SL10-FA5	.007
SL10-FA10	.005
SL10-FA15	.005
SL15-FA0	.005
SL15-FA5	.005
SL15-FA10	.005
SL15-FA15	.005
FA0-FA5	1.000
FA0-FA10	.889
FA0-FA15	.374
FA5-FA10	.575
FA5-FA15	.139
FA10-FA15	.037
Significant at p=0.00625	

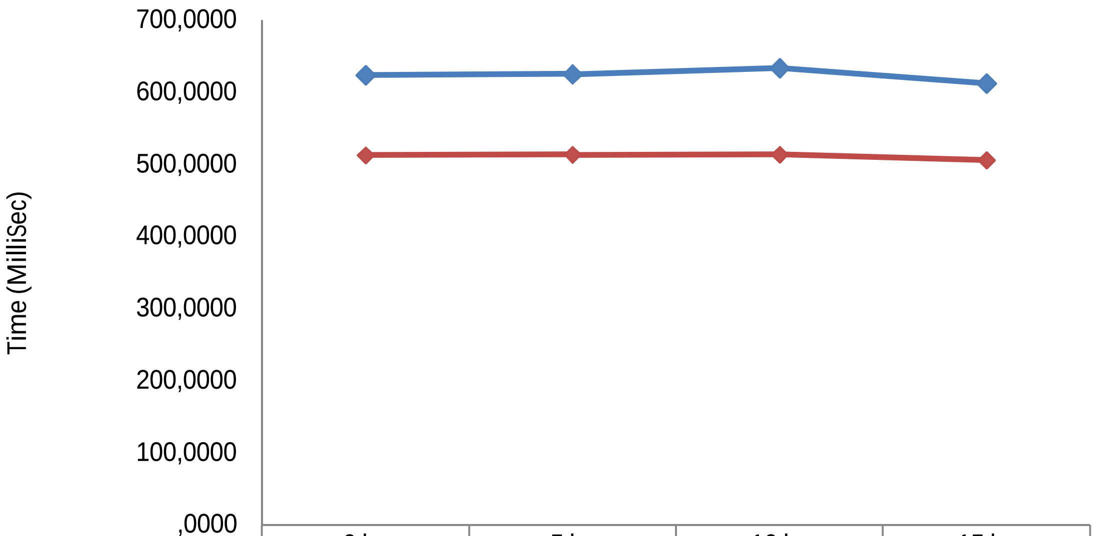
Table 8. Results of Wilcoxon Signed Ranks Test for the Time to 1st Peak (SL-Slow, FA-Fast)

Contact Time

The Contact time of the foot with the force plate increased significantly (p=.000) only when the subjects altered their gait velocity. The total contact time is a good estimate of alterations in gait velocity (Liddle et al., 2000). In this case the total contact time did not changed significantly when the external weights were placed on the subjects.

Contact Time Condition	Mean Time to Contact Time ± St. Deviation
SL0	624.14 ± 45.95
SL5	625.26 ± 43.57
SL10	633.82 ± 61.29
SL15	612.66 ± 36.87
FA0	513.11 ± 40.17
FA5	513.74 ± 42.69
FA10	513.94 ± 43.86
FA15	506.17 ± 40.08

Table 9. Contact Time (Mean ± Standard Deviations)



	0 kg	5 kg	10 kg	15 kg
Contact Time SL	624,1460	625,2640	633,8230	612,6560
Contact Time FA	513,1140	513,7470	513,9440	506,1740

Diagram 5. Contact Time (SL-Slow, FA- Fast)

Contact Time Comparison	Asymp. Sig (2-tailed)
SL0 – SL5	.721
SL0-SL10	.308
SL0-SL15	.241
SL0-FA0	.005
SL0-FA5	.005
SL0-FA10	.005
SL0-FA15	.005
SL5-SL10	.646
SL5-SL15	.059
SL5-FA0	.005
SL5-FA5	.005
SL5-FA10	.005
SL5-FA15	.005
SL10-SL15	.022
SL10-FA0	.005
SL10-FA5	.005
SL10-FA10	.005
SL10-FA15	.005
SL15-FA0	.005
SL15-FA5	.005
SL15-FA10	.005
SL15-FA15	.005
FA0-FA5	.799
FA0-FA10	.721
FA0-FA15	.575
FA5-FA10	.878
FA5-FA15	.241
FA10-FA15	.241
Significant at p=0.00625	

Table 10. Results of Wilcoxon Signed Ranks Test for the Contact Time (SL-Slow, FA-Fast)

Loading Rate

The loading rate changed significantly when gait velocity changed (p=.000). Table 11 shows that the loading rate was actually greater when the subjects walked on the force plate without any external weights in the slow and fast gait condition.

Loading Rate Condition	Mean Loading Rate ± St. Deviation
SL0	13.40 ± 14.58
SL5	8.78 ± 3.26
SL10	9.64 ± 1.71
SL15	9.78 ± 1.99
FA0	16.12 ± 9.81
FA5	13.97 ± 3.26
FA10	14.61 ± 4.40
FA15	16.68 ± 6.19

Table 11. Loading Rate (Mean ± Standard Deviations)

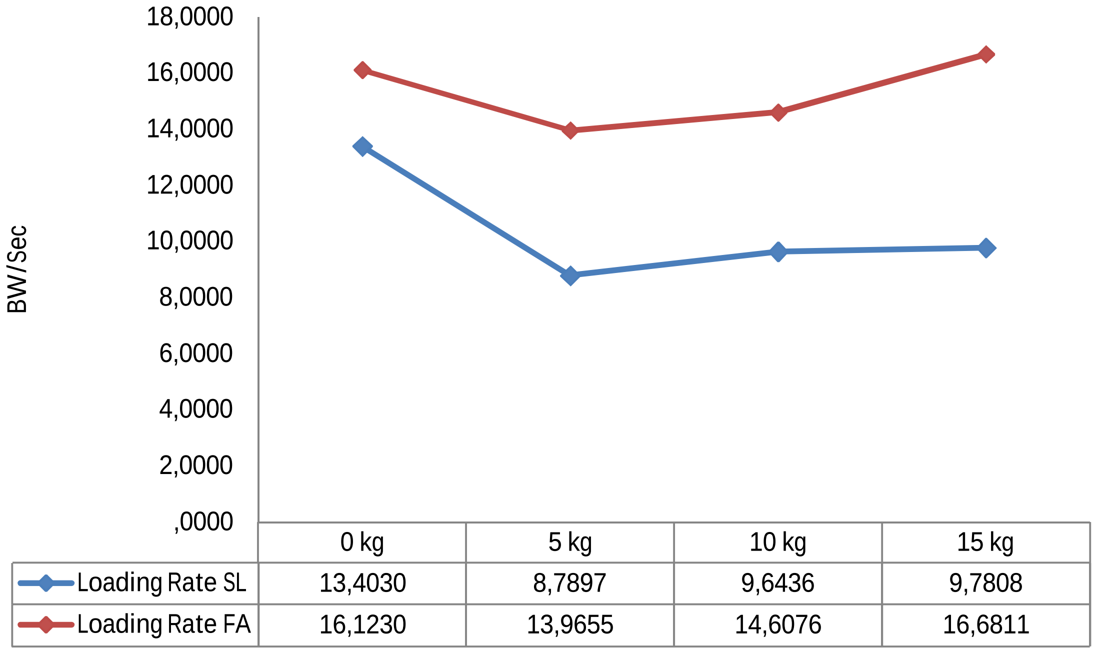


Diagram 6. Loading Rate (SL-Slow, FA- Fast)

Loading Rate Comparison	Asymp. Sig (2-tailed)
SL0 – SL5	.959
SL0-SL10	.508
SL0-SL15	.386
SL0-FA0	.074
SL0-FA5	.074
SL0-FA10	.074
SL0-FA15	.074
SL5-SL10	.333
SL5-SL15	.139
SL5-FA0	.005
SL5-FA5	.007
SL5-FA10	.007
SL5-FA15	.005
SL10-SL15	.959
SL10-FA0	.005
SL10-FA5	.005
SL10-FA10	.005
SL10-FA15	.005
SL15-FA0	.005

SL15-FA5	.005
SL15-FA10	.005
SL15-FA15	.005
FA0-FA5	.333
FA0-FA10	.203
FA0-FA15	.139
FA5-FA10	.285
FA5-FA15	.047
FA10-FA15	.051
Significant at p=0.00625	

Table 12. Results of Wilcoxon Signed Ranks Test for the Loading Rate (SL-Slow, FA-Fast)

Discussion

The findings of this study show that the size of the HST is not affected by an external weight that imitates the accumulation of abdominal fat weight. In this study most subjects were young and active as most of them were PE students, PE teachers or athletes. This finding strengthens the previous line of thought which supports that muscles of the thigh are the main shock absorbers. In this study the subjects were healthy and without any known degeneration of the knee cartilage. It has been previously supported by Brandt, Radin, Dieppe, and Van De Putte (2006) that, quadriceps strength produces a breaking action that is essential for knee deceleration just before heel-strike. Therefore, quadriceps weakness may increase the size of the HST and increase impact loading of the knee.

The results of this study show that the subjects were able to contract their quadriceps muscles adequately even when external weights were placed on them in order to absorb the shock produced during ambulation (Brandt et al., 2006). Furthermore, due to the fact that the subjects were able to practice their gait technique with the external weights many times they were able to prepare their muscles appropriately in order to activate the appropriate reflexes and absorb the energy of the impact by lengthening the muscles around the knee joint (Jones & Watt, 1971).

Although the HST and the loading rates were not significantly different when the external weights were placed on the subjects, the size of the FMFP was significantly increased when external weights were placed on the subjects. This finding is similar with the findings of Messier et al. (1996) who noticed that absolute peak vertical GRF increases in direct proportion with BW. However, in this study the investigators recruited people with a medical diagnosis of knee OA. Thus, this study provided no information about the size of the GRF in obese people without knee OA. Furthermore, Messier et al. (1996) showed that the statistical significance was only apparent in the absolute GRFs. In this current study the statistical significance was evident even after normalizing the GRFs for bodyweight.

Previous research has shown that the size of the GRF may be influenced by walking speed (Browning & Kram, 2007). However, in this study gait speed did not change when the external weights were placed on the subjects as the contact time was statistically unchanged (Tongen & Wunderlich, 2010; Weyand, Sternlight, Bellizzi, & Wright, 2000). Also the size of HSTs did not change significantly at any point with the external weight-placement something which is strongly influenced by gait speed (Whittle, 1999). It should be especially noted that, significant alternations in gait velocity would have created statistically significant differences in foot contact time (Tongen & Wunderlich, 2010). Moreover, previous research studies have also found that obese individuals demonstrate larger vertical GRFs when compared with normal-weight individuals (Browning & Kram, 2007). However, in their study the comparison was made with absolute and not with normalized GRFs. Still, the statistical significance in the current study was evident with and without normalizing the GRFs.

Interestingly, the SMFP (that is the vertical GRF during the propulsive phase of gait) decreased when the external weights were added and then decreased even more when gait velocity was increased in contrast with the FMFP. In some cases there was a statistical significant difference in the SMFP between conditions irrespective of gait speed. It has been observed by Došla et al. (2013) that, the SMFP during the propulsive phase, which is the active part of gait, tends to be larger than the FMFP during the absorption phase, which is the passive part of gait. In this current study the addition of the external weight resulted in the development of larger GRFs during the passive part of gait that may be seem detrimental for the integrity of the knee joint. Sol (2001) has shown that heel contact produces larger forces that are easily transmitted to the knee joint. In contrast, landing with the metatarsal heads may attenuate the GRFs that may reach the knee joint and reduce the risk of knee degeneration. Thus, protecting the knee joint by strengthening the knee musculature may somehow protect the joint; however, the alteration of the GRF distribution may have a damaging effect in the long-term.

The loading rates increased significantly only when gait velocity increased. It is evident from this study that placing external weights in a group of healthy and relatively active people may increase the loading rates during walking only when altering gait speed. It seems that the absence of knee OA and similarly quadriceps muscle weakness may not affect the loading rates of the lower limbs when adding external weights. This finding is also supported by Messier, Loeser, Hoover, Semble, and Wise (1992) who state that people with knee OA and muscle weakness tend to have higher loading rates. Furthermore the study of Radin et al. (1991) showed that people with pre-osteoarthritic changes may have distinct HSTs and as a result higher loading rates. Radin et al. (1991) further showed that, the pre-osteoarthritic group demonstrated shorter periods of eccentric contraction of the quadriceps with the aid of an EMG. This further strengthens the line of thought which supports the shock absorbing properties of the quadriceps muscle.

Moreover, obese people exhibit lower limb weakness and especially of the quadriceps. Several studies have shown that obese individuals have lower torque values when normalized to bodyweight. However, knee flexion strength is not significantly different between obese and non-obese individuals (Capodaglio et al., 2009; Hulens, Vansant, Claessens, Lysens, & Muls, 2003; Radin et al., 1991). Russell and Hamill (2010) explain that obese individuals may increase their quadriceps muscle strength in order to adapt to their gradual increase in bodyweight and be able to support it; however, they still avoid to rely on their quadriceps as they walk. Furthermore, obese individuals tend to adopt a gait pattern that reduces knee flexion and the eccentric contraction of the quadriceps and as a result limit the shock absorbing properties of their lower limb muscles (Messier et al., 1996).

Quadriceps avoidance during ambulation is often noticed in people with knee OA and especially in those knee OA sufferers who are also obese (Taylor, Heller, Bergmann, & Duda, 2004). This gait pattern transfers significant amount of impact shock to the knee joint which reduces the thickness of the cartilage and joint space. This mechanism also makes the ligaments of the knee more lax (Mikesky et al., 2000). Also, their ligamentous laxity may be reduced only by the muscles of the thigh and if their weak their loading rates may increase by 21% when compared with people with stronger muscles (Fisher et al., 1997). It should be noted that higher loading rates may be seen irrespective of bodyweight or knee joint integrity in people with weak lower limb muscles (Mikesky et al., 2000).

Limitations

A limitation of this study is that the investigator is not aware of any possible altered rear-foot motions that may have been used by the subjects in order to reduce the vertical impact of their lower limbs. According to Levinger and Gilleard (2005), the everted rearfoot posture in their experimental group reduced the impact force during ambulation and resulted in a smaller and delayed HST. Since the first and second peak maximum were statistically different in multiple weight conditions but with the same speed, it is unknown whether the subjects used different multi-joint compensatory strategies in order to maintain the size of the HST unchanged when the external weights were added. Several compensatory strategies have been suggested by other investigators such as altered knee flexion between trials, trunk position, altered hip motion or a combination (Brechter & Powers, 2002; Dillon, Updyke, & Allen, 1983; Nadeau, Gravel, Hébert, Bertrand Arsenault, & Lepage, 1997).

Conclusion

The findings of this study suggest that increased abdominal fat may not affect the HST and rates of loading in a group of healthy people. However, the first peak maximum may increase in size with the addition of extra weight on the abdominal area and then increase even more as gait velocity increases. Also, the second peak maximum may decrease in size when extra abdominal weight is added and then decrease even more as gait velocity increases. It should be noted that changes in the first and second peak maximum may not be due to altered gait velocity. The results of the study suggest that strong thigh muscles and especially the quadriceps may act as shock absorbers and protect the knee joints from large impact forces. However, it is unknown whether compensatory strategies of the foot or the upper joints of the lower limbs and spine may affect the size of the vertical GRFs. Suggestions for future work includes using video cameras in order to record kinematic data and assess whether other joints or body posture may change in a similar study.

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Appendix 1

ΠΑΝΕΠΙΣΤΗΜΙΟ ΘΕΣΣΑΛΙΑΣ
ΤΜΗΜΑ ΕΠΙΣΤΗΜΗΣ ΦΥΣΙΚΗΣ ΑΓΩΓΗΣ ΚΑΙ ΑΘΛΗΤΙΣΜΟΥ

Έντυπο συναίνεσης δοκιμαζόμενου σε ερευνητική εργασία

Τίτλος Ερευνητικής Εργασίας: Τα αποτελέσματα της κοιλιακής παχυσαρκίας στο Heel-Strike Transient (HST) κατά τη βάδιση: Μια πειραματική μελέτη.

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1. **Σκοπός της ερευνητικής εργασίας**

Σκοπός της μελέτης είναι η διερεύνηση της επίδρασης της κοιλιακής παχυσαρκίας στις δυνάμεις αντιδράσεις του εδάφους και πιο συγκεκριμένα στην εμφάνιση ή και αύξηση του HST σε υγιή άτομα.

2. **Διαδικασία**

Οι συμμετέχοντες θα αξιολογηθούν 1 φορά στο εργαστήριο εμβιομηχανικής του ΤΕΦΑΑ για περίπου μία ώρα. Στην μοναδική επίσκεψη θα γίνει αξιολόγηση του Δείκτη Μάζας Σώματος (ΔΜΣ). Εάν είναι στα φυσιολογικά όρια θα ζητηθεί από τους συμμετέχοντες να περπατήσουν στην επιθυμητή ταχύτητα πάνω σε δύο δυναμοδάπεδα 3 φορές. Έπειτα θα τους ζητηθεί να τοποθετήσουν ένα βάρος 5 κιλών στην περιοχή της κοιλιάς και να ξαναπερπατήσουν 3 φορές. Αυτό θα επαναληφθεί άλλες 3 φορές με βάρος 10 και με βάρος 15 κιλών. Ο ερευνητής θα καταγράψει όλες τις προσπάθειες.

Κίνδυνοι και ενοχλήσεις

Θα σας ζητηθεί να αξιολογήσετε την ικανότητα σας να περπατήσετε με το βάρος που θα δοθεί. Εάν θεωρείτε οτι είναι αρκετά μεγάλο για να το φορέσετε ή φοβάστε για πιθανό τραυματισμό τότε θα σταματήσουμε τις προσπάθειες και θα καταγράψουμε τις υπόλοιπες προσπάθειες.

Εάν έχετε κάποια από τις ακόλουθες παθήσεις παρακαλώ ενημερώστε τον ερευνητή:

Νευρολογικά προβλήματα, παράλυση ή οποιαδήποτε πάθηση έχει επηρεάσει την βάδιση σας

Καρδιοαγγειακά προβλήματα που δεν σας επιτρέπουν να σηκώσετε βάρος η να κουραστείτε έστω και λίγο

Ρευματοειδής αρθρίτιδα η άλλες παθήσεις του ανοσοποιητικού συστήματος που επηρεάζουν την βάδιση σας

Σπονδυλική στένωση

Χρόνια η οξεία οσφυαλγία

Εάν πάσχετε έστω και από μια από τις παραπάνω παθήσεις τότε θα σας ζητηθεί να μην λάβετε μέρος στην έρευνα αυτή.

Εάν όχι τότε μπορείτε να συμμετάσχετε στην έρευνα αυτή. Δεν υπάρχει κανένας κίνδυνος τραυματισμού κατά τη διάρκεια των δοκιμασιών. Παρ’ όλα αυτά υπάρχει πρόβλεψη πρώτων βοηθειών και εκπαιδευμένο προσωπικό για κάθε ενδεχόμενο

.

3. Προσδοκώμενες ωφέλειες

Με την συμμετοχή σας θα λάβετε πολλές πληροφορίες για την βάδιση σας και εάν υπάρχει κίνδυνος να αναπτύξετε οστεοαρθρίτιδα γόνατος στο μέλλον με κάποια μεταβολή στο βάρος ή στην κατανομή του βάρους σας. Επίσης θα λάβετε δωρεάν αποτελέσματα από αξιολογήσεις που στο εμπόριο κοστίζουν > 100 ευρώ. Η διερεύνηση των επιδράσεων του βάρους στις δυνάμεις αντίδρασης του εδάφους ίσως αποτελέσει τη βάση για την ανάπτυξη νέων μεθόδων που θα μειώσουν τον κίνδυνο ανάπτυξης οστεοαρθρίτιδας αλλά και θα ενημερώσουν τους επαγγελματίες υγείας καθώς και το κοινό για τον κίνδυνο αύξησης του βάρους στην υγεία των αρθρώσεων.

4. Δημοσίευση δεδομένων – αποτελεσμάτων

Η συμμετοχή σας στην έρευνα συνεπάγεται ότι συμφωνείτε με την μελλοντική δημοσίευση των αποτελεσμάτων της, με την προϋπόθεση ότι οι πληροφορίες θα είναι ανώνυμες και δε θα αποκαλυφθούν τα ονόματα των συμμετεχόντων. Τα δεδομένα που θα συγκεντρωθούν θα κωδικοποιηθούν με αριθμό, ώστε το όνομα σας δε θα φαίνεται πουθενά.

5. Πληροφορίες

Μη διστάσετε να κάνετε ερωτήσεις γύρω από το σκοπό ή την διαδικασία της εργασίας. Αν έχετε οποιαδήποτε αμφιβολία ή ερώτηση ζητήστε μας να σας δώσουμε διευκρινίσεις.

6. Ελευθερία συναίνεσης

Η συμμετοχή σας στην εργασία είναι εθελοντική. Είστε ελεύθερος-η να μην συναινέσετε ή να διακόψετε τη συμμετοχή σας όποτε το επιθυμείτε.

7. Δήλωση συναίνεσης

Διάβασα το έντυπο αυτό και κατανοώ τις διαδικασίες που θα ακολουθήσω. Συναινώ να συμμετάσχω στην ερευνητική εργασία.

Ημερομηνία: __/__/__

Ονοματεπώνυμο και υπογραφή συμμετέχοντος	Υπογραφή ερευνητή