Lower limb muscle activity during gait

Electromyographic measurements of walking in high-heeled shoes compared to walking in trainers

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Submitted in partial fulfillment of the requirements for the degree of Bachelor of Science in Physical Therapy

UNIVERSITY OF ICELAND

SCHOOL OF HEALTH SCIENCES

FACULTY OF MEDICINE

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This paper is a final project for a B.Sc. degree in physical therapy. It is prohibited to copy the content of this paper in any way without a consent from the writers.

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Abstract

High-heeled shoes are prominent in today's society and are worn on special occasions as well as with more casual outfits. Certain known kinematic changes occur when walking in shoes with high heels vs. lower heels as well as changes in the centre of mass and a reduction in the base of support. Although altered demands on muscles most likely accompany the kinematic changes, these have not been well documented. Therefore, the main purpose of this study was to identify changes in muscle activity when walking in high-heeled vs. low-heeled shoes. Temporospatial parameters were also observed.

Twenty-four able-bodied women participated in the study. Participants walked on a treadmill with an integrated pressure measuring platform. Data were collected during a single session, where each participant wore four different pairs of shoes with different heel heights (trainers and dress shoes with heels of 3, 6 and 8 cm). Muscle activity in the dominant leg was recorded. The target muscles were: fibularis longus, tibialis anterior, soleus, gastrocnemius, biceps femoris, vastus lateralis, rectus femoris and gluteus medius.

When comparing gait in 8 cm high heels to gait in trainers, a significant increase in muscle activity was seen in the fibularis longus ($p<0,001$), soleus ($p<0,01$), gastrocnemius medialis ($p<0,05$), gastrocnemius lateralis ($p<0,001$) and vastus lateralis ($p<0,001$). The fibularis longus muscle showed the largest increase, as it more than tripled its activity between the two shoe conditions. Step length, step width and foot progression angle decreased as heel height increased and push-off force diminished. The results reflect greater demands for dynamic stability and a less efficient push-off during gait in shoes with increased heel height.

Virkni helstu vöðva neðri útlima í göngu;

Vöðvarafritsmælingar á göngu á háum hælum borið saman við göngu á hlaupaskóm

Ásdís Árnadóttir, Inga Hrund Kjartansdóttir og Sigríður Katrín Magnúsdóttir

Leiðbeinandi: Dr. Kristín Briem

Ágrip

Við það að ganga á háum hælum verða ákveðnar breytingar í hreyfingum um liðamót ganglima samanborið við þegar gengið er á lágbotna skóm. Auk þess verða breytingar á stöðu þungamiðju og undirstöðuflötur minnkar. Kröfur um vöðvavinnu hljóta að sama skapi að breytast, en þær breytingar eru ekki vel þekktar.

Megintilgangur þessarar rannsóknar var því að mæla breytingar á vöðvavirkni lykilvöðva í göngu á háum hælum miðað við göngu á lágbotna skóm, ásamt því að skoða þekktar breytur í göngu.

Þátttakendur í rannsókninni voru tuttugu og fjórar heilbrigðar konur. Hver þeirra gekk í fjórum mismunandi hælahæðum (hlaupaskóm og spariskóm með 3, 6 og 8 cm háum hælum) á meðan EMG mælingar voru gerðar á vöðvavirkni í ríkjandi fæti. Markvöðvar voru fibularis longus, tibialis anterior, soleus, gastrocnemius, biceps femoris, vastus lateralis, rectus femoris og gluteus medius. Þátttakendur gengu á sérhönnuðu göngubretti með innbyggðum þrýstingsnemum.

Þegar ganga á 8 cm hælum var borin saman við göngu á hlaupaskóm kom í ljós marktæk aukning á vöðvavirkni í fibularis longus ($p<0,001$), soleus ($p<0,01$), gastrocnemius medialis ($p<0,05$), gastrocnemius lateralis ($p<0,001$) og vastus lateralis ($p<0,001$). Mest varð aukningin í fibularis longus þar sem hún rúmlega þrefaldaðist. Einnig komu í ljós breytingar á þann veg að skrefvídd, skreflengd og útsnúningur fótar minnkaði almennt með aukinni hælahæð, ásamt því að kraftur fráspyrnu varð minni.

Niðurstöður auka þekkingu okkar á breytingum á göngumynstri og vöðvavinnu sem eiga sér stað við göngu á háum hælum vegna aukins óstöðugleika ökklans og breyttrar afstöðu lykilvöðva liðarins.

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1. Introduction

Tales from all over the world refer to the use of high-heeled shoes for varied purposes. Ancient Egyptian document from 4000 B.C. states the use of high-heeled shoes which could be considered predecessors of today´s high-heeled footwear (Paslawski, 2008). The emperors of China used high heels to appear taller, and centuries ago European women used them to protect their dresses from dirt (Gundaker, n.d.).

In modern society, high heels are mainly used by women for the purpose of a better appearance, since they make legs seem longer and slimmer, which is considered a desirable look. Given the abnormal position the ankle and foot is kept in when wearing high heeled shoes, it is very unlikely that people with good knowledge of the human musculoskeletal system would recommend wearing them. In spite of that, they are very popular amongst women, both as a part of casual and dressy outfits.

Certain known kinematic changes occur in the lower limb when walking in high-heeled shoes. In the sagittal plane, knee flexion is greater from initial contact throughout stance in high-heeled gait compared to low-heeled gait, but less at toe off and during swing. Hip flexion is also less during swing when walking in high heels (Opila-Correia, 1990). However, changes in muscle activity under the aforementioned circumstances have not been well documented.

The main purpose of this study was to measure changes in muscle activity during walking in pairs of shoes with various heel heights, using electromyography. Changes in temporospatial parameters were also to be observed across the different shoe conditions. The data collected in the study will improve our understanding of the altered demands placed on joints and muscles with increasing heel height and may be used in the proposed development of a new prosthetic leg with an adjustable ankle for walking in a variety of shoes with different heel heights.

2. Background

2.1. Anatomy

2.1.1. The hip joint

The hip joint is a synovial joint between the head of the femur and the acetabulum of the pelvis. Three bones form the pelvis; the ilium, the ischium and the pubis. They all contribute to the acetabulum (Drake, 2010). The acetabulum has a horseshoe-shaped cartilage around most of its periphery, while its centre is free of cartilage. The bottom of the peripheral acetabulum is not complete with cartilage but is completed as a ring by the transverse acetabular ligament (Levangie, 2005).

The hip joint has maximum stability because of the deep insertion of the head of the femur into the acetabulum. The acetabulum is deepened with a fibrocartilage ring called the acetabular labrum. The labrum grasps the head of the femur to maintain contact with the acetabulum and to further increase the joint's stability (Levangie, 2005; Magee, 2006).

The hip joint has three degrees of freedom; flexion and extension in the sagittal plane, abduction and adduction in the frontal plane, medial and lateral rotation in the transverse plane. The joint also allows a combined circular movement called circumduction (Drake, 2010; Levangie, 2005).

The primary function of the hip joint is to bear the weight of the head, arms and trunk in static posture as well as in locomotion, such as walking and running (Levangie, 2005).

Ligaments

The hip joint is a structurally stable joint, but is made even more stable by a thick capsule, which is divided into thickened layers forming the iliofemoral, pubofemoral and ischiofemoral ligaments. The iliofemoral ligament is thought to be the strongest ligament in the body and lies anterior to the femoral head, positioned to prevent excessive extension. The pubofemoral ligament lies anteroinferior to the femoral head, prevents excessive abduction of the femur and limits extension. The ischiofemoral ligament lies posterior to the joint and helps stabilize the hip in extension. All three ligaments also limit medial rotation of the femur (Drake, 2010; Levangie, 2005).

In the standing position, the centre of gravity passes behind the centre of rotation of the hip joint. The fibers of all three ligaments spiral around the hip joint and become taut when the

joint is extended. The iliofemoral ligament lies in front of the joint and therefore permits static stance to rely on ligamentous support without supporting muscular contraction (Cailliet, 2003; Drake, 2010).

Muscles

Primary flexors of the hip (Figure 1) are the iliopsoas (formed by the psoas and iliacus muscles), rectus femoris, tensor fascia lata (TFL) and sartorius. The rectus femoris is one of four parts of the quadriceps muscle, the only part that crosses both the hip and the knee joint. It inserts along with the three vasti muscles on the upper aspect of the patella, and lies from there through a common tendon onto the tibial tuberosity. Its role is to flex the hip joint and extend the knee joint. The amount of force it is able to generate at the hip is affected by the position of the knee because it crosses both joints, greater force being generated with the knee held in flexion (Drake, 2010; Levangie, 2005). Primary hip flexors function mainly as mobility muscles in open chain functions, that is to bring the swinging limb forward during gait (Levangie, 2005).
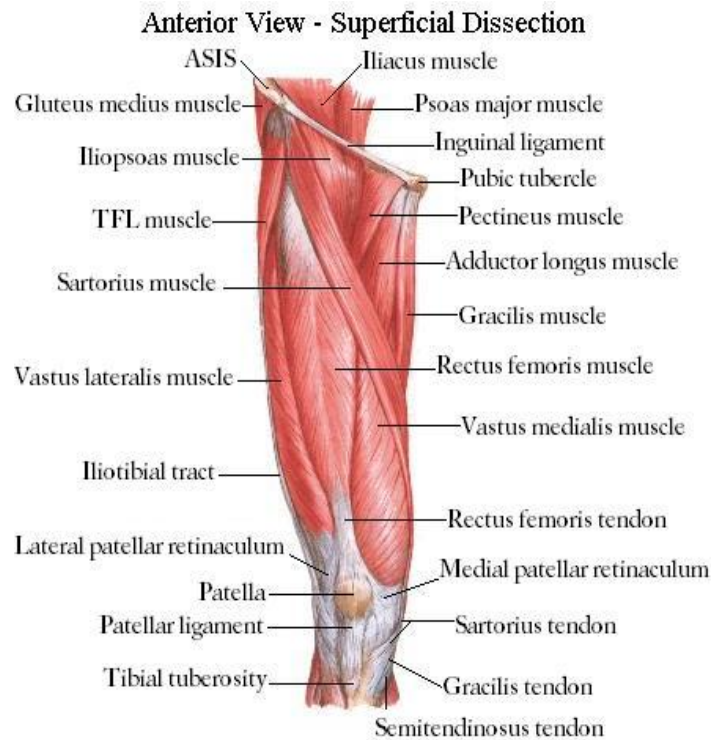
Figure 1. Primary flexors of the hip (FitPro1).

The secondary hip flexors are the pectineus, adductor longus, adductor magnus, and gracilis muscles. These muscles are predominantly adductors of the hip but can all contribute to flexion, depending on the hips position (Levangie, 2005).
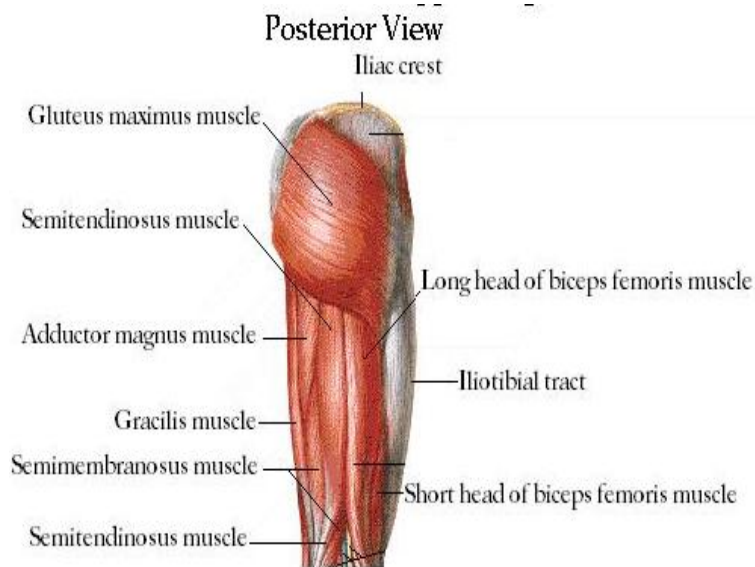
Figure 2. Primary extensors of the hip (FitPro1).

The primary hip extensors (Figure 2) are the gluteus maximus and the hamstrings muscle group. The hamstrings muscle group consists of the short and long head of the biceps femoris laterally, and the semitendinosus and semimembranosus medially (Levangie, 2005). The hamstrings flex the leg at the knee joint and extend the thigh at the hip joint. The short head of the biceps is the only muscle in the posterior compartment of the thigh which does not cross both the hip and knee joints. The long head of the biceps rotates the hip laterally, and when the knee is flexed the biceps can rotate the knee laterally. Together, the semimembranosus and the semitendinosus rotate the hip and the knee medially (Drake, 2010; Levangie, 2005). As biarthrodal muscles, the role of the hamstrings in hip extension is strongly influenced by knee position. (Levangie, 2005).

The primary hip abductors are the gluteus medius and gluteus minimus. The adductor muscle group of the hip includes the pectineus, adductor brevis, adductor longus, adductor magnus and the gracilis (Levangie, 2005).

The lateral rotators of the hip are obturator internus and externus, gemellus superior and inferior, quadratus femoris and piriformis. There are no muscles that have the primary function of producing medial rotation of the hip joint, but the more consistent medial rotators are the anterior portion of the gluteus medius, gluteus minimus and the TFL (Levangie, 2005).

2.1.2. The knee joint

The knee joint is the largest synovial joint in the body, and because its primary function is to move and support the body during both routine and difficult activities, it must fulfill major stability and mobility roles (Snyder-Macker, 2005).

The joint consists of two distinct articulations located within one joint capsule, the tibiofemoral, which is the joint between the distal femur and proximal tibia, and the patellofemoral, which is the articulation between the posterior aspect of the patella and the distal anterior femur (Drake, 2010; Snyder-Macker, 2005).

The tibiofemoral joint is a hinge joint with three degrees of freedom: flexion and extension in the sagittal plane, medial and lateral rotation in the transverse plane, and abduction and adduction in the frontal plane (Snyder-Macker, 2005).

The medial and lateral menisci lie in between the femur and tibia. The menisci are semicircular in shape and are made of fibrocartilage. The menisci have several roles in the knee, serving as shock absorbers, while spreading the stress over the articular cartilage and decreasing cartilage wear. They enhance joint congruency between the two articulating surfaces and improve weight-bearing distribution by increasing their area of contact. They also reduce friction between the tibia and femur and help prevent hyperextension of the knee (Drake, 2010; Magee, 2006; Snyder-Macker, 2005).

The main function of the patellofemoral joint is to create a pulley for the quadriceps muscle and to reduce friction on the quadriceps tendon (Snyder-Macker, 2005).

Ligaments

The knee depends on ligaments and muscles for its strength and stability. The major ligaments surrounding the knee are: the patellar ligament, the medial and lateral collateral ligaments, and the anterior and posterior cruciate ligaments. (Drake, 2010)

The role of the medial and lateral collateral ligaments is to resist valgus and varus stress respectively. The medial collateral ligament (MCL) can be divided into deep and superficial portions which are separated by a bursa. Both parts of the MCL arise from the medial femoral epicondyle and insert onto the medial aspect of the proximal tibia. The deep portion of the MCL is continuous with the joint capsule and is affixed to the medial border of the medial meniscus (Snyder-Macker, 2005).

The lateral collateral ligament (LCL) originates from the lateral femoral condyle and inserts onto the fibular head in a conjoined tendon along with the biceps femoris muscle (Snyder-Macker, 2005).

The two cruciate ligaments, anterior and posterior, are in the intercondylar region of the knee joint and connect the femur and tibia. They are termed cruciate because they cross

each other in the sagittal plane. The main role of the anterior cruciate ligament (ACL) is to prevent anterior displacement of the tibia on the femur and control rotational movement. It also resists hyperextension of the knee. The posterior cruciate ligament (PCL) prevents posterior displacement of the tibia beneath the femur. Both ACL and PCL have a role in preventing varus and valgus stress at the knee (Cooper, 2006; Drake, 2010; Snyder-Macker, 2005).

Muscles

The muscles that cross the knee may be divided into flexor and extensor groups. The extensors of the knee are the four heads of the quadriceps muscles (Figure 1). As stated earlier, the vastii (medialis, lateralis and intermedius) originate on the femur and merge with the rectus femoris into a common tendon which attaches on the upper proximal aspect of the patella. From there, the tendon continues to the tibial tuberosity (Snyder-Macker, 2005).

The primary knee flexors are the hamstrings (semimembranosus, semitendinosus, and biceps femoris), (Figure 2) assisted by the sartorius, gracilis, popliteus and gastrocnemius (Snyder-Macker, 2005).

The two heads of the gastrocnemius (Figure 3), while originating on the lateral and medial distal femur, come together in the proximal part of the leg to form a single muscle belly, which attaches along with the underlying soleus muscle on the calcaneus of the foot, through the calcaneal tendon. The gastrocnemius primarily plantarflexes the foot at the ankle, but also takes part in flexing the leg at the knee joint (Drake, 2010).

2.1.3. The ankle and foot complex

The ankle joint refers to the talocrural joint, which is the articulation between the distal tibia and fibula proximally, and the trochlea of the talus distally. The distal ends of the tibia and fibula are anchored together by strong ligaments to create a deep socket, also called mortise, that the trochlea of the talus fits into, and forms the tibiofibular joint (Drake, 2010; Magee, 2006; Mueller, 2005).

The joint is a synovial hinge joint with mainly one degree of freedom, dorsiflexion and plantarflexion in the approximately sagittal plane (Drake, 2010; Magee, 2006). Also, the joint allows a small side to side movement (Platzer, 2004). The superior surface of the talus is wider anteriorly than posteriorly. In plantarflexion only the relatively posterior

body of the talus is in contact with the mortise and the joint is therefore less stable in that position compared to the neutral or dorsiflexed positions. This leads to an increased number of injuries with the foot in the plantarflexed position (Karlsson, 2006; Mueller, 2005).

The foot complex is divided into the hindfoot, midfoot and forefoot. The tibiofibular and talocrural joints are a part of the hindfoot along with the subtalar joint. The subtalar joint is the articulation between the talus and the calcaneus (Magee, 2006). The joint motion is supination and pronation around an oblique axis (Mueller, 2005). Function of the joint in weight-bearing is to provide shock absorption for the forces imposed by the weight of the body, to adjust the foot to uneven ground, and to maintain contact of the foot to the surface when the leg is at an angle to the surface (Karlsson, 2006; Mueller, 2005).

The midfoot is made of the three cuneiform bones, the cuboid and navicular bones, and the joints between them. Each joint allows only minimal amount of movement, but together they allow significant movement, and enable the foot to adapt to many different positions without stressing each joint too much (Magee, 2006).

The forefoot consists of the five metatarsal bones, the bones of the phalanges, and the joints between them. Each phalange has three bones except for the first digit that has two (Magee, 2006). The role of the forefoot is most prominent during walking where it allows the weight-bearing foot to rotate over the toes during push-off (Mueller, 2005).

The ankle and foot work together, providing a stable base of support (BoS) for the body in weight-bearing postures without excessive muscle activity and energy use, but also act as a lever for push-off during gait (Mueller, 2005).

Ligaments

The main stabilizers at the ankle joint are the medial (deltoid) and lateral ligaments. They maintain contact and congruence between the mortise and talus, and control medial-lateral joint stability, while also providing key support for the subtalar joint (Drake, 2010; Mueller, 2005).

The deltoid ligament is a large and very strong ligament. It originates from the medial malleolus and extends anteriorly to the navicular and talus bones, inferiorly to the calcaneus, and posteriorly to the talus. The ligament is strong and composed of two layers, one deep and one superficial. The ligament resists valgus stress as well as lateral

translation and rotation of the talus. It is infrequently injured (Karlsson, 2006; Mueller, 2005).

Laterally three separate ligaments all arise from the lateral malleolus and help control varus stresses. The calcaneofibular ligament inserts onto the calcaneus, the anterior talofibular ligament passes anteriorly to the lateral talus, and the posterior talofibular runs posteriorly to the talus (Magee, 2006; Mueller, 2005).

Muscles

The muscles in the posterior compartment of the leg are divided into two groups, superficial and deep. The muscles mainly plantarflex and invert the foot, and flex the toes. The superficial group consists of the gastrocnemius, soleus and plantaris muscles. The soleus muscle is a large flat muscle which lies underneath the gastrocnemius, and together they attach onto the posterior calcaneus through a common tendon. They are the strongest plantarflexors of the ankle. The deep group consist of the popliteus, tibialis posterior, flexor digitorum longus and flexor hallucis longus (Drake, 2010; Mueller, 2005).
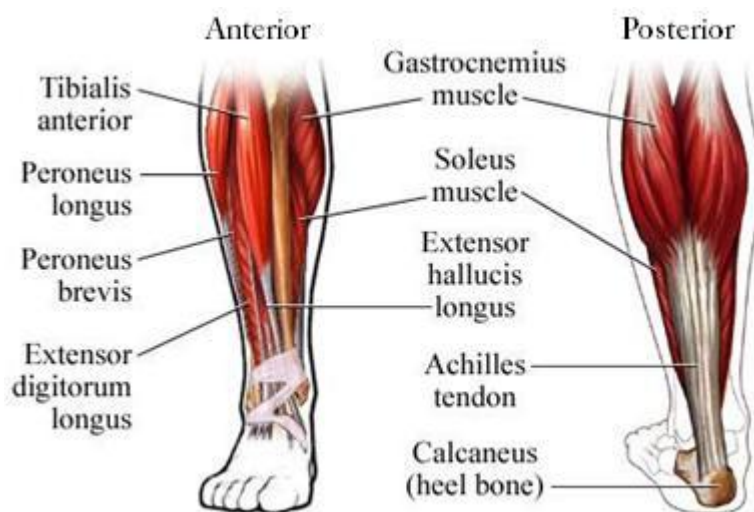
Figure 3. Muscles in the anterior and posterior compartments of the leg (Trainharder.com, 2007-2010).

The muscles in the lateral compartment of the leg are the fibularis (peroneus) longus and brevis. The fibularis longus everts and plantarflexes the foot. The tendon of the fibularis longus, as well as the tendons of the tibialis anterior and tibialis posterior, insert on the undersurfaces of bones on the medial side of the foot. Therefore, they act together to support the arches of the foot (Drake, 2010; Mueller, 2005).

The anterior compartment contains the tibialis anterior, extensor hallucis longus, extensor digitorum longus and fibularis tertius. Together they dorsiflex the foot and extend the toes, The tibialis anterior is also a key supinator of the foot (Drake, 2010; Mueller, 2005).

2.2. Muscle physiology

2.2.1. Action potentials

Muscles are an excitable tissue due to the fact that they receive electrical signals called action potentials (AP), from motor nerves. Consequently, they respond by contracting, producing force and movement (Silverthorn, 2007).

Before and after an AP takes place, the neuron has a resting membrane potential of -70 mV (Figure 4). An AP begins when a stimulus reaches the trigger zone and depolarizes the neurons membrane to its threshold of -55 mV. As the cell depolarizes, voltage-gated Na+ (natrium) channels open and positively charged Na+ ions flow into the cell. This makes the inside of the cell more positive and the AP peaks at about +30 mV. As a result the Na+ channels close. K+ channels open and K+ ions (potassium) move out of the cell to the extracellular fluid. As K+ moves out of the cell, the membrane potential becomes more negative bringing the cell towards its resting potential. When the membrane potential reaches -70 mV again, the K+ channels are not yet closed and so potassium continues to leave the cell. As a result the membrane hyperpolarizes, reaching -90 mV. Once the K+ channels close, retention of K+ and Na+ leak back into the cell, bringing the membrane potential back to -70 mV. (Silverthorn, 2007).

Figure 4. Action potential (mindcreators.com).

When an AP is initiated, it travels down the somatic motor nerve towards the corresponding skeletal muscle. Once it reaches the muscle, the nerve releases a neurotransmitter called Acetylcholine into the neuromuscular junction, which initiates an AP in the muscle fibre. The AP moves across the cell membrane, down the T-tubules,

triggering the release of Calcium, which combines with troponin and initiates a muscle contraction. A single AP in a skeletal muscle fiber evokes a single twitch in the muscle (Silverthorn, 2007).

Action potentials are sometimes called all-or-none phenomena. This is because when a stimulus reaches threshold, a full depolarization occurs. If a stimulus does not reach the threshold an AP does not occur at all. There is no in-between. The signal does not diminish as it travels down the neuron or to another neuron. The signal's amplitude stays the same, allowing it to travel long distances, such as from the spinal cord to a toe or fingertip (Silverthorn, 2007).

2.2.2. Motor units

Each somatic motor neuron innervates a certain number of muscle fibres. This is called a motor unit. All the muscle fibres in a single motor unit function as a whole, so when a motor neuron fires an AP, all the muscle fibres within that motor unit contract. Motor units vary in size and muscle fibre types (Silverthorn, 2007). Muscles controlling large gross movements have bigger numbers of muscle fibres per motor unit (100-1000) than muscles controlling fine movements (usually less than 10). The number of motor units per muscle varies throughout the body (Rash).

The muscle force we use at any given time depends on the task we are performing. To produce force in a single muscle, smaller motor units are recruited first. As the force requirement increases, larger motor units are recruited. The activation frequency is also a factor in producing force. As motor units fire at a faster rate, more force is produced. Therefore, muscle force depends on the recruitment of motor units and their activation frequency (Kamen, 2004).

2.3. Electromyography (EMG)

2.3.1. What is EMG and what does it tell us?

EMG measurements are made with probes known as electrodes. These electrodes can vary in size and shape and be made from different material (Enoka, 2002). The EMG signal is a summation of the AP generated by the motor units within the pick-up area of the electrode being used (Rash). Therefore, as more motor units are activated and their AP frequency increases one would expect the EMG signal to become larger.

The data collected by EMG are useful in determining the relative increase and decrease in muscle activity or the on/off timing pattern of a muscle. They do not tell us whether a muscle contraction is concentric, eccentric, voluntary or involuntary. Furthermore, the EMG signal can neither tell us how strong a muscle is nor if one muscle is stronger than another (Rash).

2.3.2. Different types of EMG

When recording an EMG signal, the electrodes can be placed in the muscle between outside muscle cells (intramuscular/needle/fine wire EMG) or on the skin (surface EMG) (Enoka, 2002). When measurements are performed from within the muscle, recordings can be made from one or more motor units. By doing so, it is possible to monitor the activity of a small unit within the muscle rather than the global activity of the muscle (Kamen, 2004). The advantages of using fine wire electrodes are; more specific pick up area, increased signal band width, isolation of specific muscle parts when testing large muscles and the ability to test deep muscles as well as testing small ones without crosstalk from adjacent muscles being an issue. The disadvantage of using this kind of EMG is that needles must be inserted which causes discomfort. This can increase tightness or spasticity in the muscle which in turn can cause cramping. It is also difficult to place the needle/fine wire in the same area of the muscle each time when performing repeated measures, possibly weeks or months apart, whereby different units may be sampled each time (Rash).

Another way to perform an EMG recording is to place electrodes on the skin over the muscle (surface EMG). That way provides a more global measure of the action potential activity in the underlying muscle (Enoka, 2002). The signal is detected over the skin and depends on the membrane fibre properties, muscle fibre anatomy, location of the motor units, tissue between the fibres and recording location, electrode size and recording montage (Merletti, 2008). A correct placement of the electrodes is very important. They should be placed over active muscle fibres and away from highly tendinous areas (Kamen, 2004) Another important factor is the orientation of electrodes with respect to muscle fibres. The electrode needs to be placed parallel to the muscle fibres to receive a good signal. Not doing so could result in as much as 50% signal reduction (Kamen, 2004; Vigreux, 1979).

Using surface electrodes has certain advantages; they are convenient for movement, easy to apply, cause minimal pain with application and are more reproducible (Rash). The

disadvantages of surface electrodes are that they have a large pick-up area which increases the likelihood of crosstalk from adjacent muscles and makes it hard to record signals from small muscles (Kamen, 2004; Rash). They are not good for recording activity from deep muscle tissue and some evidence suggests that the estimated effective recording area of surface electrodes lies between 10 and 20 mm from the skin surface (Barkhaus, 1994). Therefore a thick layer of adipose tissue could reduce the signal.

Furthermore there is some evidence that superficial motor units may be bigger than deeper motor units. That would mean that surface EMG recording could be biased toward recording activity from larger and more glycolytic motor units (Kamen, 2004; Lexell, 1983).

2.3.3. Frequency and filtering

Most of the EMG signal is contained at a frequency spectrum between 10 Hz and 1 kHz. Little if any falls under or over these values (Kamen, 2004). For the surface EMG signal the upper frequency limit is more bandwidth-limited, with the highest frequency component at about 600 Hz. Fine wire electrodes collect data at a larger band width, ranging from 2Hz-1kHz (Rash).

While recording, the EMG signal can be contaminated by the activity of other sources such as movement of the electrodes and cables, activity in other muscles and electromagnetic radiation in the environment (electric machines, power cords and lights) (Enoka, 2002). These unwanted signals are often referred to as noise. Filters can be used to attenuate certain parts of the frequency spectrum and thereby reduce noise. When using a low pass filter, frequencies above the cutoff are attenuated but lower frequencies remain unchanged. When using a high pass filter, the cutoff is selected so that low frequencies are attenuated but high frequencies remain (Derrick, 2004). Band-pass filters eliminate frequencies above and below certain values. Band-stop filters reduce the signal of a certain frequency or range of frequencies (Enoka, 2002).

To get a clear picture of the data being collected, the Nyquist limit can be used. That involves collecting data at a sampling rate at least twice that of the highest frequency component in the signal. This prevents signals above this frequency from distorting the true signal (Kamen, 2004).

2.3.4. EMG recording

A common way to record whole muscle activity is to use bipolar recordings. Two electrodes as well as a ground electrode are placed on the skin over the muscle. Each measuring electrode has a diameter of about 8 mm and are separated by 1,5 cm. The measurement procedure consists of two main steps. First the voltage difference between the signal detected by the ground electrode and the two measuring electrodes is determined. Then the difference between the voltages measured by each electrode is calculated and amplified. The output is a voltage time signal called EMG (Enoka, 2002). The signal difference between the two measuring electrodes is therefore the object of interest and any signal that is common to both electrodes is greatly attenuated and considered noise (Kamen, 2004; Rash).

2.3.5. Signal processing

In order to use the information from the electrodes the signal needs to be processed (Figure 5). The action potentials measured by the EMG electrodes have a positive and negative phase that fluctuates about a baseline of zero. A mean value is therefore not a valid indicator of the signals amplitude, as the mean value could be close to zero. This is called raw EMG. To compute a meaningful averaged amplitude measured over a time period the signal must be rectified. This is done by converting the negative phases to positive ones and
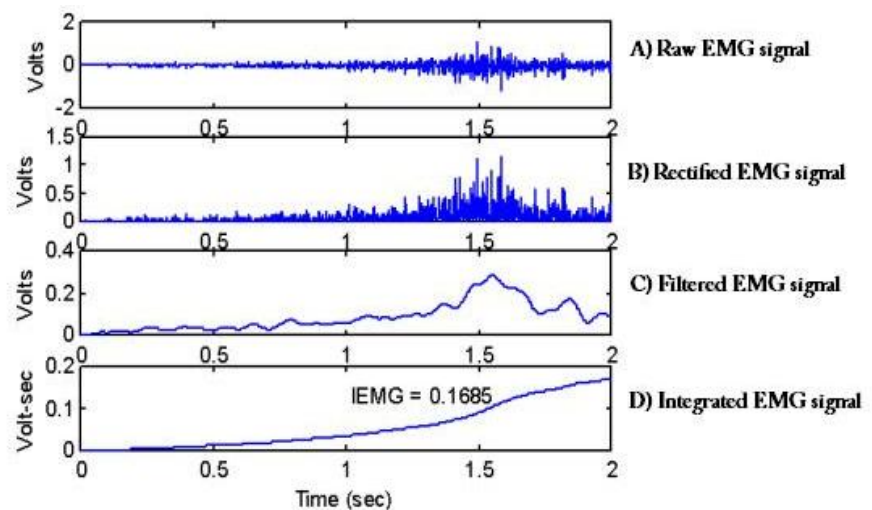
Figure 5. EMG signal processing (School of Biomedical Engineering, Science and Health Systems, 2009).

thereby showing the absolute value of the EMG signal. Next, the EMG can be integrated by smoothing the sharp peaks with a low pass filter. This reduces the high frequency content of the signal. Quantification can then be done by measuring the amplitude of the integrated signal. The outcome of this process is an EMG signal that represents the change in force over a chosen time period (Enoka, 2002; Kamen, 2004).

EMG can be used in different fields such as physical rehabilitation, gait analysis, clinical medicine, biofeedback control, ergonomics, motor control, fatigue analysis and dentistry (Kamen, 2004).

2.4. Kinematics of gait

2.4.1. The gait cycle

During normal gait, a single gait cycle has been defined as the period from initial heel contact until the same heel touches the ground again (Figure 6) (Kaufman, 2006). Each foot goes through two main phases during gait that are expressed as a percentage of the gait cycle; stance phase where the foot is in contact with the ground, and swing phase of the same leg where it is clear off the ground. When walking at a self-selected speed, stance phase takes up about 60% of the gait cycle and swing phase 40%. These phases are further divided into smaller subphases (Shumway-Cook, 2007).
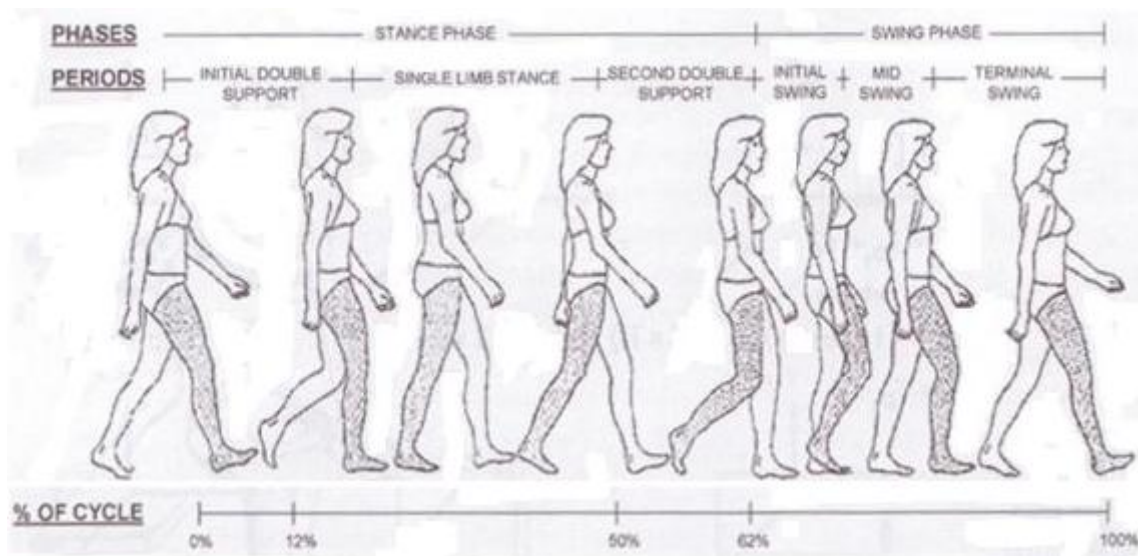
Figure 6. Normal gait cycle illustrating the events of gait (Kaufman and Sutherland, 2006).

Stance phase

The stance phase is traditionally divided into five subphases where each one has a different role. Two periods of double support take place when both feet are in contact with the ground, during the first and last 10% of stance (Olney, 2005).

Initial Contact (IC) refers to the moment the heel touches the ground until foot is flat and takes up the first 7% of stance. The mechanical goal is to position the foot and begin

deceleration. The joint angles in the sagittal plane at IC are approximately 20° flexion of the hip, while the knee and ankle are in neutral position (Kaufman, 2006; Olney, 2005).

Foot flat occurs after IC. Together, IC and foot flat make up a loading response where the body accepts weight, stabilizes the pelvis and decelerates mass. The joint angles in the sagittal plane during foot flat reach approximately 15° flexion of the hip and knee and 5° plantarflexion of the ankle (Kaufman, 2006; Olney, 2005).

When the whole body weight is stabilized over the foot and supported by one leg at 30% of the gait cycle it has reached *midstance*. The joint angles in the sagittal plane are neutral position of the hip, 5° knee flexion and 5° dorsiflexion (Kaufman, 2006; Olney, 2005).

At *terminal stance* the body begins to accelerate mass and the heel leaves the ground. This starts to occur when the gait cycle has reached about 40% of its time. The joint angles in the sagittal plane are approximately 10-20° hyperextension of the hip, and neutral position of both knee and ankle (Kaufman, 2006; Olney, 2005).

Toe off is the last event of the stance phase, occurring at 60% of the gait cycle, when the toes of the relevant foot leave the ground and the leg prepares for swing. The joint angles in the sagittal plane at toe off are approximately 10-20° hyperextension of the hip, 30° knee flexion and 20° plantarflexion (Kaufman, 2006; Olney, 2005).

Although joint motion largely occurs in the sagittal plane during gait, as described above, motion also occurs in frontal and transverse plane. In frontal plane, at the beginning of single support, the contralateral pelvis drops, resulting in a 5° adduction of the supporting limb. The hip then abducts to about 5° until it reaches toe off. The subtalar joint plays an essential part in shock absorption at the beginning of IC where it changes from a 5° inversion to approximately 5° eversion. The motion then reverses to about 10° inversion at the beginning of toe off (Olney, 2005).

In the transverse plane approximately 4° of external rotation occurs in the hip. The talocrural joint rotates from 5° internal rotation at IC to 5° external rotation at foot flat and then inverts again to approximately 7° before toe off. Not many noticeable changes occur in the knee in both frontal and transverse plane through stance. (Olney, 2005).

Swing Phase

Swing phase is traditionally divided into three subphases. *Early swing* starts when the foot clears off the ground and takes over the time between 60-75% of the gait cycle. The joint

angles in the sagittal plane in the beginning of early swing phase are approximately 20° hip flexion, 60° knee flexion and 10° plantarflexion of the ankle. *Mid-swing* takes over the next 75-85% of the gait cycle. During that time the leg accelerates, then decelerates as it passes directly beneath the body. The joint angles in sagittal plane are approximately 30° hip and knee flexion and neutral position in the ankle. *Late swing* closes the last 15% of the gait cycle. The goal of this phase is to decelerate the leg, position the foot and prepare it for contact. The joint angles in the sagittal plane at the end of swing are 30° flexion of the hip and neutral position in the knee and ankle (Kaufman, 2006; Olney, 2005).

In addition to joint movement in the sagittal plane, the hip returns to a neutral frontal plane position right before IC from about 5° of abduction at toe off. The knee abducts about 7° for a brief period of time during mid-swing and then returns to neutral at the end of the gait cycle. From push-off to IC, the subtalar joint moves from 10° to 5° inversion. In the transverse plane, the hip externally rotates until mid-swing and internally rotates from that point to a neutral position before initial contact. The knee joint externally rotates to about 10° from late stance until the end of the gait cycle and turns to neutral at foot flat during the next cycle. During swing, the talocrural joint has small reversals of external and internal rotation (Olney, 2005).

Gait Terminology

Stride length is defined by the distance travelled between two heel-strikes of the same foot and includes all events of one gait cycle. *Step length* is the distance travelled between the heel-strike of one foot to the heel-strike of the contralateral foot. One stride length therefore consists of two steps. *Cadence* is the number of steps taken in a previously decided time frame. *Walking speed* is the distance travelled per unit of time. *Step width* is the distance between the midpoints of both heels during gait. *Progression angle* (Figure 7) represents the angle between the centre of the heel and the second toe (Kaufman, 2006; Olney, 2005).
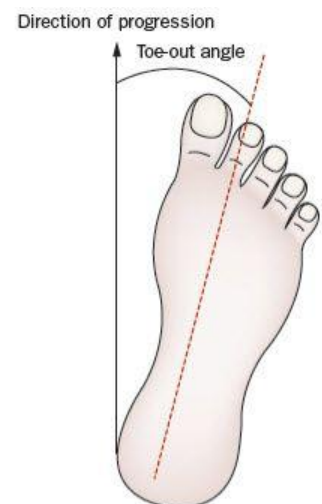
Figure 7. Progression angle (Medscape Education, 2011)

2.4.2. High-heeled gait

High-heeled shoes usually have a narrow toe box and a small heel area that leads to a smaller BoS, as compared to typical shoes. The body´s centre of gravity is also higher in high-heeled shoes and this can lead to a less stable BoS during stance**.**

Studies have shown that women tend to walk slower in high heels and shorten their stride. In addition, the stance phase of gait tends to be relatively longer, especially double support, at the expense of swing phase. Cadence, however usually stays the same (Opila-Correia, 1990).

Walking in high-heeled shoes changes joint movement through the lower extremities. In the sagittal plane, knee flexion is greater from IC throughout stance in high-heeled gait compared to low-heeled gait, but less at toe off and during swing. Hip flexion is also less during swing when walking in high heels (Opila-Correia, 1990).

In the frontal plane, hip abduction is generally less in high-heeled gait compared to low-heeled, which leads to closer alignment of the femur to the trunk. In the transverse plane the foot is more internally rotated during stance and less external rotation is seen in the hip when walking in high heels (Opila-Correia, 1990).

2.5. Kinetics of gait

2.5.1. Muscle activity during gait

When walking, people generate a certain amount of force each time they place the foot on the ground. The ground applies the same amount of force to the foot and this is called ground reaction force (GRF). This force is greater than body weight both early on in stance, while the body is accepting weight, and during late stance, at push-off. In mid-stance, the GRF is lower (Olney, 2005).

Muscles contribute most of the work that is required for normal walking (Anderson, 2003). Most muscles are active at the beginning and end of swing phase, which suggests that the main function of the muscles during walking is to accelerate and decelerate the leg (Boakes, 2006). The rest of the work required for walking is contributed by passive force through joints and bones (Anderson, 2003).

Muscle activity during stance phase

During early stance, the goal of the foot is to slow the body's momentum and prepare for weight bearing. At IC, a deceleration of the limb begins by simultaneously activating the knee extensor and flexor muscles to correctly position the knee before it accepts weight. The hip extensors slow the forward movement of the leg down by eccentrically contract (Boakes, 2006).

During loading response, the ankle dorsiflexors eccentrically contract as the foot reaches the ground. The knee extensors also contract eccentrically as the knee bends, but as the knee extends the contraction changes to concentric. The gluteus medius muscle isometrically contracts in order to stabilize the pelvis (Boakes, 2006).

The body's centre of gravity reaches its highest point during midstance. Along with momentum and passive resistance from bones and ligaments to gravity, nearly all the support is provided by the anterior and posterior parts of the gluteus medius/minimus muscles in an isometric contraction to eccentrically control pelvic drop against gravity in unilateral stance during walking (Anderson, 2003; Boakes, 2006; Drake, 2010). The soleus muscle also has a role in keeping the foot on the floor by eccentrically contracting (Boakes, 2006).

At late stance the body accelerates forward and nearly all the muscle work is generated by a shortening contraction of the ankle plantarflexors (Anderson, 2003; Boakes, 2006). The fibularis longus also transfers weight from the lateral to the medial side of the foot (Mueller, 2005). Just before toe off, hip flexors concentrically contract in order to prepare the leg for swing phase and therefore unloading (Boakes, 2006).

Muscle activity during swing phase

Most of the lower limb muscles are inactive during swing phase and the movement is much like a pendulum, the leg swinging freely. At the beginning of swing, the ankle dorsiflexors contract concentrically to allow the foot to clear off the ground and remain contracted throughout the whole swing phase. At terminal swing, the goal is to decelerate the leg and prepare it for weight acceptance and the hamstrings contract either isometrically or eccentrically in order to slow both hip flexion and knee extension. The contraction in the ankle dorsiflexors changes from concentric to isometric or eccentric (Boakes, 2006).

2.5.2. Muscle activity during high-heeled gait

With increased heel height and plantarflexion of the ankle, the centre of mass moves up and forward. In order to keep balance, the knee flexes more during stance than when wearing low-heeled shoes (Opila-Correia, 1990; Stefanyshyn, 2000). Due to this increased knee flexion, the rectus femoris muscle is more highly activated (Stefanyshyn, 2000). The same research found no difference in the activity of the vastus medialis, biceps femoris or semitendinosus muscles with increased heel height. Whereas no significant changes occurred in the activity of gastrocnemius, fibularis longus or tibialis anterior muscles, increased activity was recorded in the soleus muscle with increased heel height.

Lee, Jeong and Freivalds (2001) studied the biomechanical effect high-heeled shoes have on gait by measuring EMG from the erector spinae muscles in the low back, the tibialis anterior and the vastus lateralis muscles. No increase was found in the peak EMG value of the vastus lateralis when comparing low-heeled gait to 8 cm high-heeled gait. Conversely, Edwards, Dixon, Kent, Hodgson and Whittaker (2008) found greater muscle activity in both the vastus lateralis and medialis in women carrying out a sit to stand task while wearing 5 cm high heels compared to a barefoot condition. No imbalance between the muscles was seen, and increased heel height did not seem to affect the ratio between the vastus medialis and lateralis.

Gefen, Megido-Ravid, Itzchak and Arcan (2002) compared muscular fatigue in regular wearers of high-heeled shoes to non-regular wearers. EMG data was collected from both heads of the gastrocnemius, soleus, tibialis anterior, extensor hallucis longus and fibularis longus muscles. A fatigue test that consisted of 40 voluntary contractions in an open chain, intended to simulate muscle activity during swing phase, and 40 forced contractions in a closed chain intended to simulate muscle activity during stance phase, was used to study the effects of high-heeled gait. The results showed that gastrocnemius lateralis is less able to endure fatigue compared to gastrocnemius medialis in regular wearers of high-heeled shoes, whereas the difference was less with women who were not regular wearers. The fibularis longus was also more vulnerable to fatigue in regular wearers compared to non-regular wearers.

3. Purpose of study and hypotheses

High heels have been worn by both men and women throughout the world for many centuries. The purpose of wearing high heels has been different from time to time, but in today's society long legs are considered beautiful and desirable, and therefore many women choose to wear high heels. Wearing high heels as a part of a nice outfit is something most women take for granted. This however has been a problem for women with a prosthetic leg, since the legs aren't adjustable enough for them to walk efficiently and comfortably in high heels. Moreover, altered firing patterns associated with altered demands on the muscles of the lower extremities with increasing heel height, have not been thoroughly investigated.

Therefore, the purpose of this study is to identify changes in muscle activity when walking in high heels compared to walking in low-heeled shoes. These changes will be analysed with respect to the altered temporospatial parameters observed, with the objective of elucidating the compensatory mechanisms adopted with ambulating in a variety of heel heights during normal gait. The results may be used in the development of new prosthetic legs designed to meet the needs of modern women.

Hypotheses

H1: More muscle activity will be seen in the key muscles crossing the knee joint when walking in high-heeled shoes as compared to low-heeled: the rectus femoris, vastus lateralis and biceps femoris.

Rationale: When walking in high-heeled shoes, the knee is in a more flexed position, therefore it is likely that the rectus femoris and vastus lateralis muscles will need to meet greater demands during the early phases of stance. Given that the plantarflexors will be less effective at push-off, the biceps femoris may demonstrate greater activity due to greater pre-swing demands across the hip and knee.

H2: More muscle activity will be seen in muscles crossing the ankle joint when walking in high-heeled shoes as compared to low-heeled: the fibularis longus, soleus and gastrocnemius medialis/lateralis.

Rationale: When walking in high-heeled shoes, the ankle is put in a plantarflexed position. The gastrocnemius medialis/lateralis and soleus muscles are thereby kept in a more shortened position as compared to walking in low-heeled shoes. They may therefore

demonstrate greater activity in an effort to generate adequate force, in particular during push-off.

With increased ankle plantarflexion, the joint is mechanically less stable. The fibularis longus is an ankle stabilizing muscle and may therefore demonstrate greater muscle activity with increased heel-height.

H3: Less muscle activity will be seen in the tibialis anterior muscle when walking in high-heeled shoes as compared to low-heeled.

Rationale: The tibialis anterior muscle is put in a relatively lengthened state when wearing high-heeled shoes. Initial contact is made by the heel during normal gait. When walking in high-heeled shoes, the demands for dorsiflexion are less, as the hind- and forefoot land more directly on the ground instead of the heel-strike and rollover seen during normal gait in low-heeled shoes. Furthermore, the greater knee flexion during early stance may result in less work from the tibialis anterior in bringing the tibia forward.

4. Methods

The study protocol was approved by the review board at The National Bioethics Committee (VSNb2010110002/03.7) and announced to The Data Protection Authority.

An exploratory repeated measures design was used during the study.

4.1. Participants

Twenty-four able bodied women meeting the inclusion and exclusion criteria volunteered to participate. They were selected on the basis of availability by convenience sampling from the University community and from Össur hf.

Inclusion criteria were women in generally good health and between 20-40 years of age. The exclusion criteria were pregnancy, any neurological or musculoskeletal symptoms in the three months prior to data collection and any dysfunction precluding them from safe ambulation in shoes with up to 8 cm high heels. Obese participants (BMI> 30) were also excluded from the study to increase likelihood of clear EMG signals, as a thick adipose tissue layer can cause excessive filtering and result in a weaker EMG signal.

4.2. Stimulus

Each participant was tested with four pairs of shoes with different heel heights: a pair of trainers and dress shoes with 3 cm, 6 cm, and 8 cm high heels (Figure 8). The high-heeled shoes were provided by Össur hf, and were available in sizes 36-40. Participants wore

Figure 8. Trainers, 3, 6 and 8 cm high-heeled shoes.

their own running shoes which were all very similar in style, heel height and width. The 3 and 6 cm high-heeled shoes were from Gabor and the 8 cm high-heeled shoes were from SixMix. The contact area of the heels was 4,38 cm^2 (2,05*2,14 cm) in the 3 cm, 2,87 cm^2 (1,34*2,14 cm) in the 6 cm and 7,33 cm^2 (3,03*2,42 cm) in the 8 cm high heels. The shape

of the toe-box varied in the high-heeled shoes, that is to say, the 3 and 8 cm shoes had a similar toe-box but it was longer on the 6 cm.

4.3. Equipment

EMG equipment from Kine (Hafnarfjörður, Iceland) was used to record muscle activity (electromyography, EMG) during gait. Snap-on electrodes with an inter-electrode distance of 20 mm were used to collect data from target muscles, and a single heel-switch used to identify heel-strike. Twelve units were available to record, digitize and transmit the signals to a receiver connected to a PC computer with software from Kine; Kine Pro. The sampling frequency was set at 1600 Hz with a signal bandwidth of 16-500 Hz.
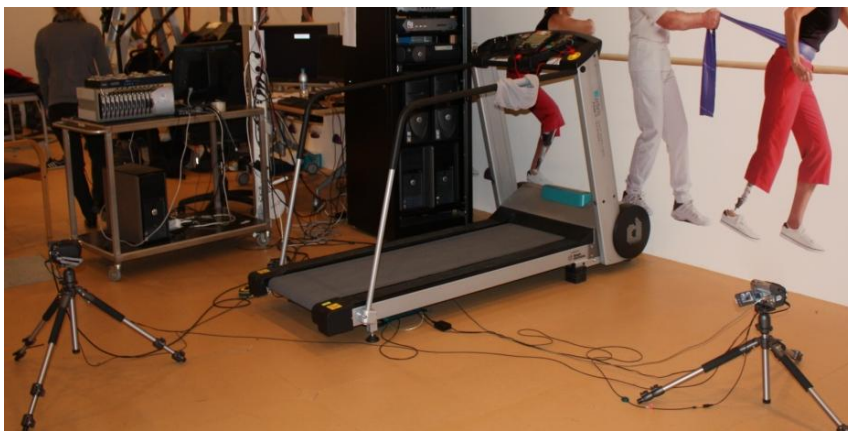
Figure 9. Arrangement of the equipment while recording. The treadmill, two cameras and the EMG equipment.

The participants walked on a FDM-TDM Medical version treadmill (Schein orthopädie service, Remscheid, Germany). An integrated pressure-measuring platform embedded in the treadmill enabled the analysis of pressure distribution, force, time and step parameters, as well as gait symmetry. Over 7100 pressure sensors sampling at 120 Hz covered the platform area of 108.4 x 47.4 cm. Two video cameras were connected and synchronized to the treadmill so that the treadmill and cameras recorded simultaneously (Figure 9).

Figure 10. The pressure sensor and its placement on the aluminium plate.

A thin flat pressure sensor was taped onto an aluminium plate, which was especially cut to fit into the heel area of the shoes. The midpoint of the pressure sensor was placed 22 mm from the posterior edge of the aluminium plate (Figure 10). The pressure sensor was connected to a battery with an EMG

transmitter. A wire connected the pressure sensor to the battery, which was placed in a pouch on the treadmill during each walk.

4.4. Execution

All measurements took place between January 12th and 21st 2011 in the gait laboratory of Össur hf. in Reykjavík, Iceland. Participants received an information letter regarding the purpose of the study, as well as their right to withdraw their consent at any time (Appendix I). They all signed an informed consent form before the measurements took place (Appendix II).

Participants provided information regarding their age, height, weight and shoe size. Their experience of walking in high-heeled shoes was evaluated in terms of hours per week; regular wearers (>7 hours per week) vs. non regular wearers (≤7 hours per week). To determine leg dominance, the women were asked which leg they would prefer to use when kicking a ball (Appendix III).

To determine the preferred walking speed on the treadmill for each participant, they all put on the 8 cm high heels and walked on the treadmill. The benchmark speed was 3,3 km/h but the participants were instructed to walk at the speed that felt most comfortable to them, between 3,1-3,5 km/h. The speed they chose was kept the same during all measurements. This window of speed was decided after the researchers themselves tested different speed settings while walking in 8 cm high-heeled shoes. The 3,3 km/h speed was considered by the researchers to be a normal comfortable walking speed. But given the fact that

Figure 11. Placement of electrodes on a participant during walking.

participants vary in height, they were given an opportunity to decide their own walking speed within this limit.

All participants wore shorts during the measurements. Disposable snap-on electrodes were placed over nine muscles of the dominant leg (Figure 11). To reduce resistance from the

skin during measurements, the skin was cleaned with a cloth wet with rubbing alcohol. Electrodes were placed according to Seniam guidelines (Appendix IV).

To test the signal from each electrode and determine if it was well placed, participants were asked to perform a few simple movements: squat, lift their heels, lift their forefeet, abduct the hip and bend the knee on their dominant side. The size and quality of the signal from each electrode was visually inspected, and in cases where the signal quality seemed poor, a decision was made to reposition the electrode.

The pressure sensor was placed in the heel area under the insole of the shoe. Each participant started off by walking in her own running shoes at her chosen speed for about one minute. A synchronized 10-second recording was made with the EMG and treadmill after they had walked for 20-30 seconds to familiarize themselves with the treadmill and shoes. The procedure was followed for each heel height. A 3-minute break was taken between each measurement while data was being collected from the transmitters and to prevent muscle fatigue.

During measurements researchers had specific roles. One worked on the computer saving data collected, while the other two tended to the participants; obtained their signatures on the consent form, found the right shoe type and size for each participant and placed the electrodes on the selected muscles.

4.5. Processing

Of the 24 women who participated in the study, 2 were excluded for technical reasons. The treadmill didn't record the gait correctly with one subject, and with the other, the files holding the EMG data were incorrectly saved on the computer.

The data were visually inspected after post-processing (raw-data, filtered data, frequency analysis). Subsequently, a decision was made not to use the data collected from the gluteus medius for analysis due to disturbances in the signal in the majority of samples. Other measurements were excluded from the study for the same reason: the fibularis longus in subject 6, the vastus lateralis in subject 24 and the gastrocnemius medialis in subject 8.

A 10-second recording was made for each subject in each heel height. From these recordings, 5 consecutive gait cycles were identified and used for the data analysis. The signal from the pressure sensor intended to pinpoint each heel-strike turned out not to be useful. Therefore, the raw EMG signals from the ankle plantarflexors were used as a

marker to determine the beginning and end of the 5 cycles. A computer software (Matlab;R2007) was used to process the EMG data. The signal was rectified by converting the negative phases to positive ones and thereby showing its absolute value. The signal was then passed through a 30Hz high pass filter, after which the total muscle activity within the 5 gait cycles was calculated by integrating the filtered EMG signal (using a 100-millisecond moving window) (Figure 12).
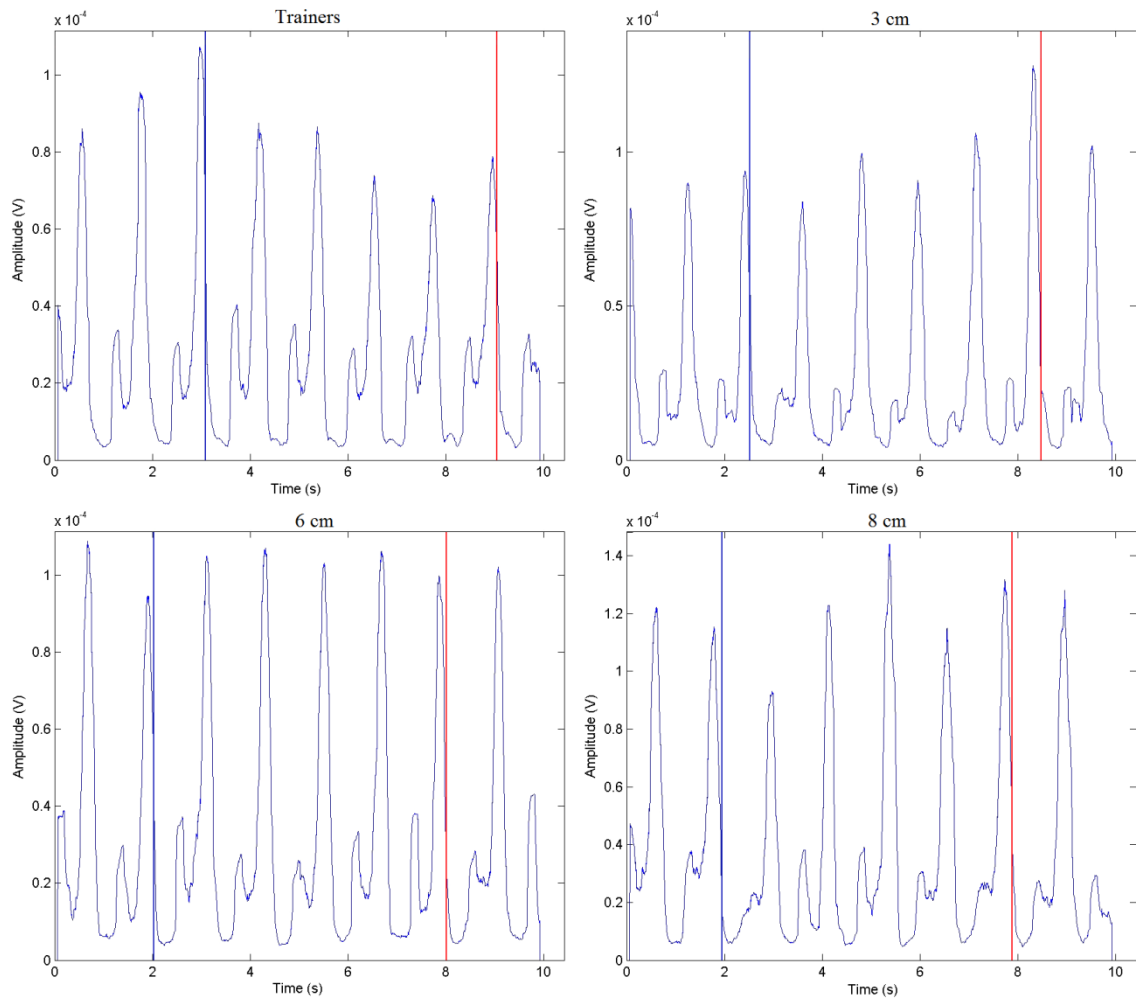
Figure 12. An example of what the EMG measurements looked like across conditions after the Matlab software had finished processing. The figure shows the EMG measurements of the soleus muscle from one of the participants. The blue line marks the beginning of the 5 gait cycles and the red line the end.

Excel was used to calculate the average energy of the muscles' signal for each subject in each of the four shoe conditions. The EMG signal for the three high-heeled conditions was normalized to the signal found when subjects walked in trainers to allow for comparison between muscles and between groups in each heel height.

Temporospatial data was collected by the pressure sensors in the treadmill. Relative factors such as information regarding changes in step width, step length, cadence, progression angle and push-off force were extracted out of the database and used for further analysis.

4.6. Statistical analysis

SPSS (PASW Statisics 18.0) was used for statistical analysis. For each variable of interest a repeated measures ANOVA was carried out to identify differences between the four conditions. High-heel use (regular vs. non-regular wearers) was introduced as a between groups variable. Confidence interval was set at 95% and the level of statistical significance $p < 0,05$. When a significant main effect or interaction was found, post-hoc analyses (t-tests) were used to determine where the difference lay. To identify associations between variables of interest, Pearson's correlation coefficient was used.

Microsoft Office Excel 2007 was used to create tables and figures.

5. Results

5.1. Participants

Descriptive statistics regarding the participants can be seen in Table 1.

Table 1. Participant statistics

	N	Range	Mean	SD
Age [years]	22	20-38	26,55	4,25
Height [cm]	22	159-185	170,77	7,06
Weight [kg]	22	50-84	68,23	8,99
BMI [kg/m2]	22	18,68-28,37	23,38	2,61
Hrs/week	22	0-60	14,91	19,15
Velocity [km/klst]	22	3,1-3,5	3,30	0,13
Shoe size	22	36-40	38,45	1,26

5.2. EMG results

An ANOVA was performed to analyze all the muscle activity found across conditions, and in five of the eight muscles measured, a significant difference in muscle activity was found. An increase in muscle activity was seen in the fibularis longus muscle (F= 11,532; p<0,001), the soleus (F= 14,731; p< 0,001), gastrocnemius medialis (F= 7,638; p= 0,004), and lateralis (F= 11,352; p= 0,001), and vastus lateralis (F= 6,529; p= 0,004) muscles (Figure 13-16 and Figure 18).

No significant difference in muscle activity between conditions was seen in the tibialis anterior (Figure 17), rectus femoris (Figure 19) and biceps femoris muscles (Figure 20).

5.2.1. Fibularis longus

The fibularis longus showed an increase of 219% in average muscle activity, when going from trainers to 8 cm high-heeled shoes (Figure 13).
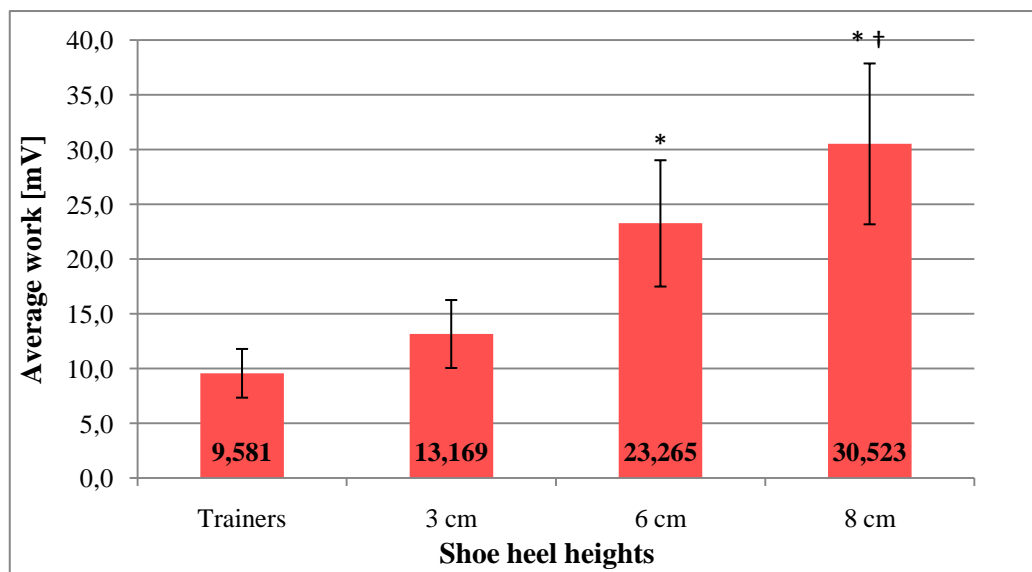

Figure 13. Average (SE) muscle activity of the fibularis longus muscle across conditions. SE= standard error.
* Significantly greater muscle activity when compared to trainers and 3 cm high heels; p< 0,001.
† Significantly greater muscle activity when compared to 6 cm high heels; p= 0,001.

5.2.2. Soleus

The soleus showed an increase of 76% in muscle activity, when going from trainers to 8 cm high-heeled shoes (Figure 14).

Figure 14. Average (SE) muscle activity of the soleus muscle across conditions. SE= standard error.
* Significantly greater muscle activity when compared to trainers and 3 cm high heels; p< 0,01.
† Significantly greater muscle activity when compared to 6 cm high heels; p< 0,03.

5.2.3. Gastrocnemius medialis

The gastrocnemius medialis showed an increase of 47% in average muscle activity, when going from trainers to 8 cm high-heeled shoes (Figure 15).

Figure 15. Average (SE) muscle activity of the gastrocnemius medialis muscle across conditions. SE= standard error.
* Significantly greater muscle activity when compared to trainers; p< 0,05.
† Significantly greater muscle activity when compared to 3 and 6 cm high heels; p< 0,02.

5.2.4. Gastrocnemius lateralis

The gastrocnemius lateralis showed an increase of 132% in average muscle activity when going from trainers to 8 cm high-heeled shoes (Figure 16).
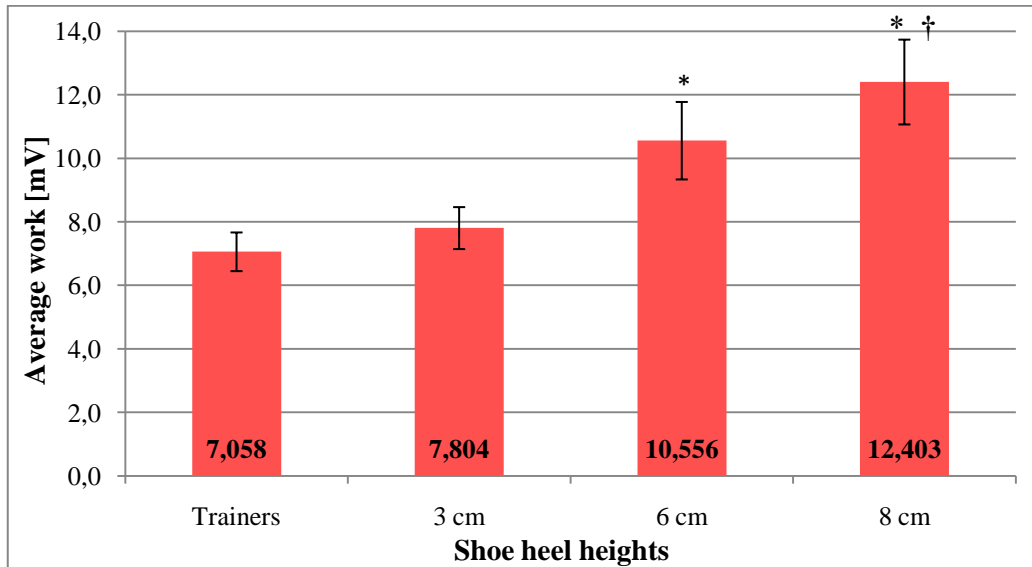

Figure 16. Average (SE) muscle activity of the gastrocnemius lateralis muscle across conditions. SE= standard error.
* Significantly greater muscle activity when compared to trainers and 3 cm heeled shoes; p< 0,001.
† Significantly greater muscle activity when compared to trainers and 3 cm high heels; p≤ 0,03.
Δ Significantly greater muscle activity when compared to trainers; p< 0,005.

5.2.5. Tibialis anterior

The tibialis anterior showed no significant difference in muscle activity between shoe conditions (p= 0,647) (Figure 17).

Figure 17. Average (SE) muscle activity of the tibialis anterior muscle across conditions. SE= standard error. No significant difference was found.

5.2.6. Vastus lateralis

The vastus lateralis muscle showed an increase of 151% in muscle activity, when going from trainers to 8 cm high-heeled shoes (Figure 18).
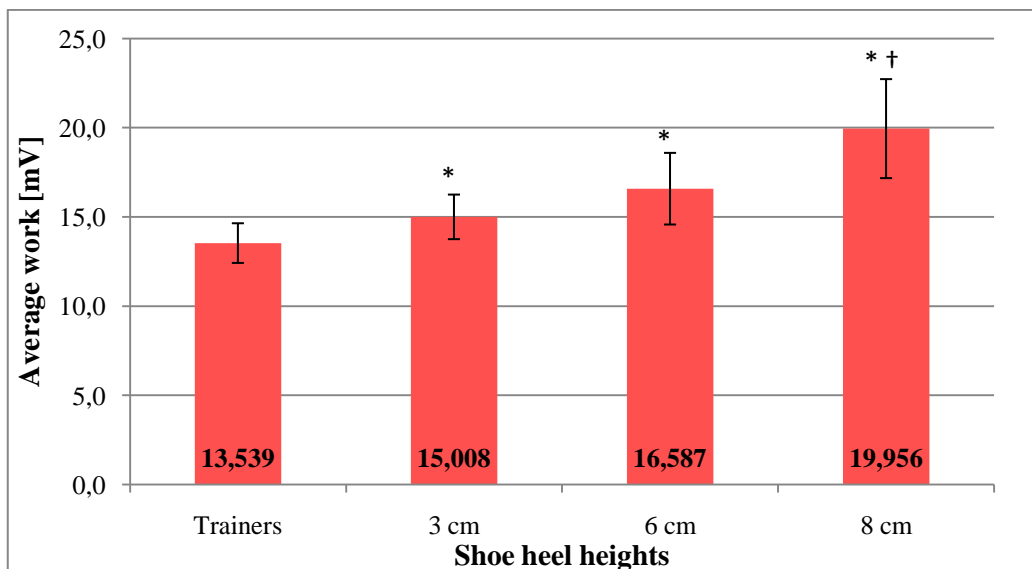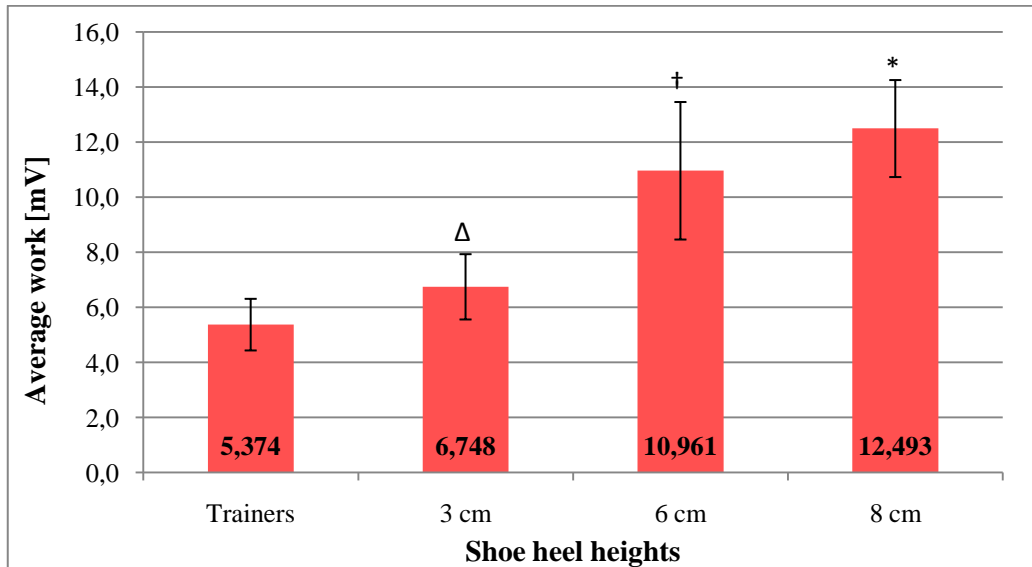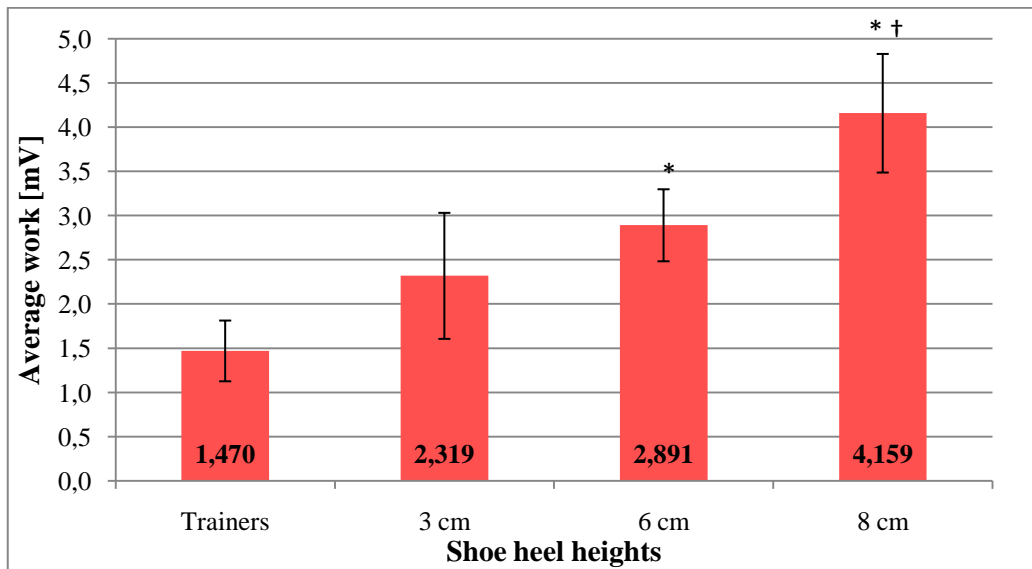

Figure 18. Average (SE) muscle activity of the vastus lateralis muscle across conditions. SE= standard error.
* Significantly greater muscle activity when compared to trainers; $p \leq 0,001$.
† Significantly greater muscle activity than in 6 cm high heels; $p < 0,050$.

5.2.7. Rectus femoris

The rectus femoris muscle showed no significant difference in muscle activity between shoe conditions ($p = 0,599$) (Figure 19).
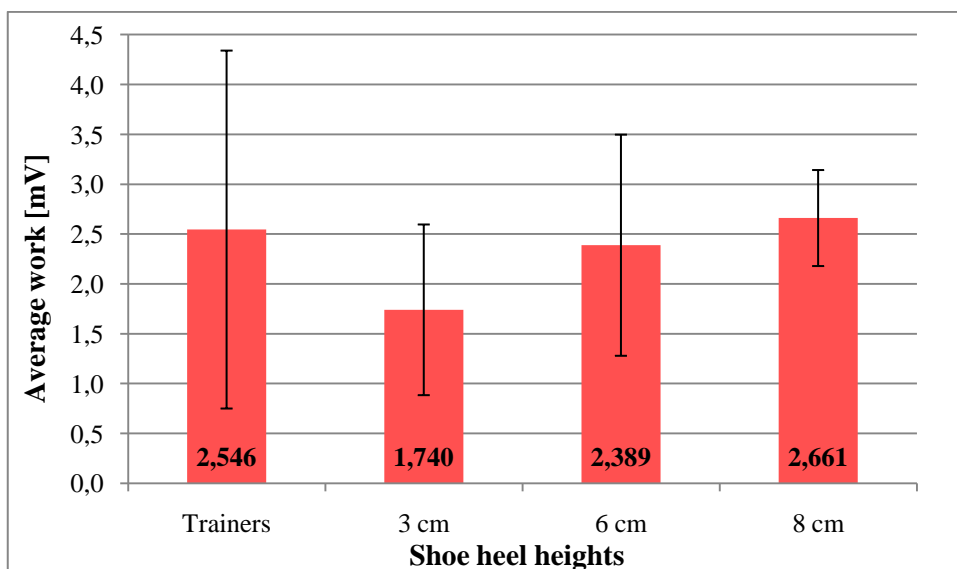
Figure 19. Average (SE) muscle activity of the rectus femoris muscle across conditions. SE= standard error.
No significant difference was found.

5.2.8. Biceps femoris

The biceps femoris muscle showed no significant difference in muscle activity between shoe conditions (p= 0,102) (Figure 20).
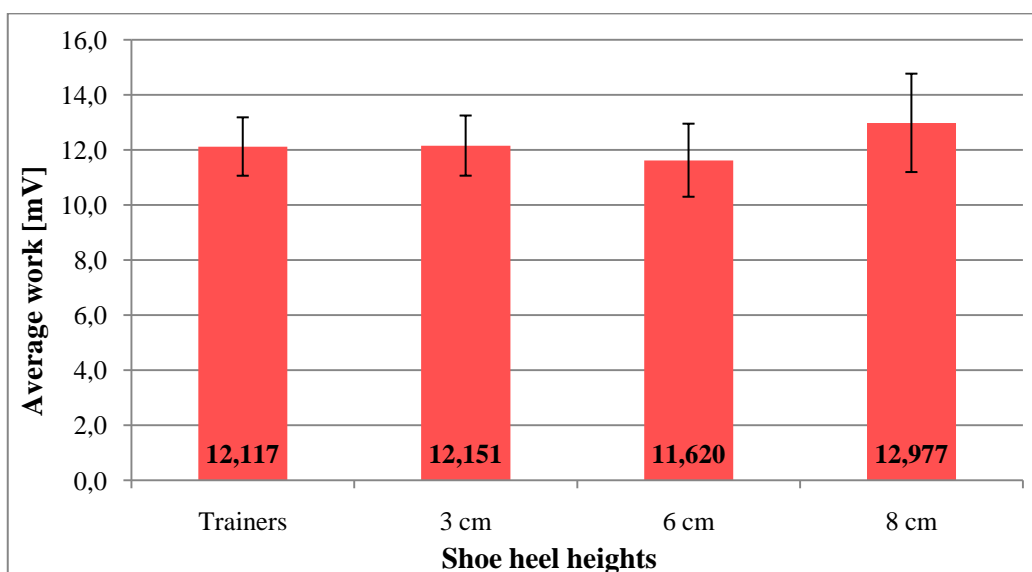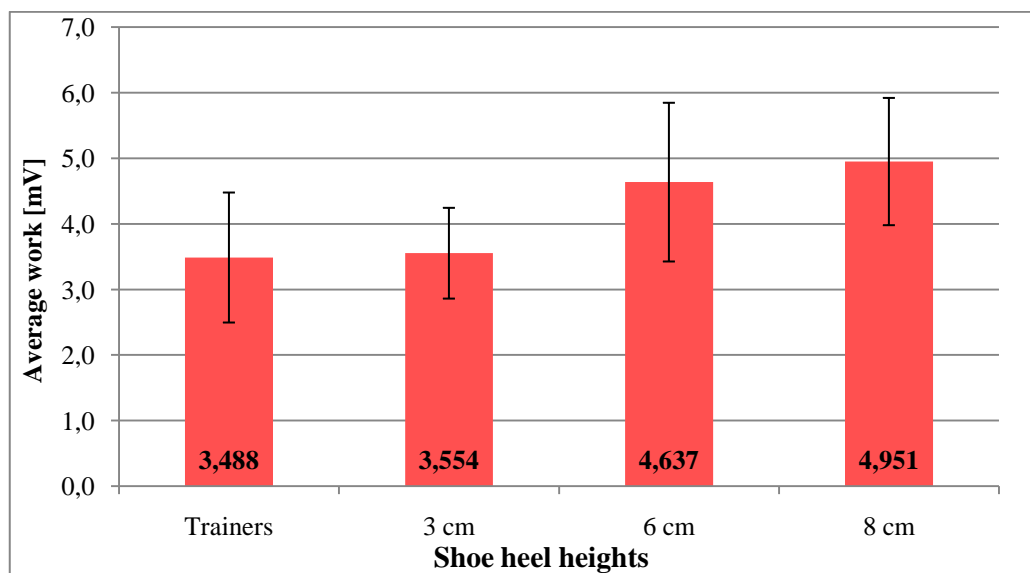

Figure 20. Average (SE) muscle activity of the biceps femoris muscle across conditions. SE= standard error. No significant difference was found.

5.3. Temporospatial results

A main effect of shoe condition was found for step length (F=8,156; p=0,001) (Figure 21), cadence (F=13,470; p<0,001) (Figure 22), step width (F=8,582; p<0,001) (Figure 24) and progression angle (F= 15,760, p< 0,001) (Figure 25).

As heel height increased a certain pattern was apparent where average step width, step length and progression angle generally decreased, while cadence increased.

5.3.1. Step length
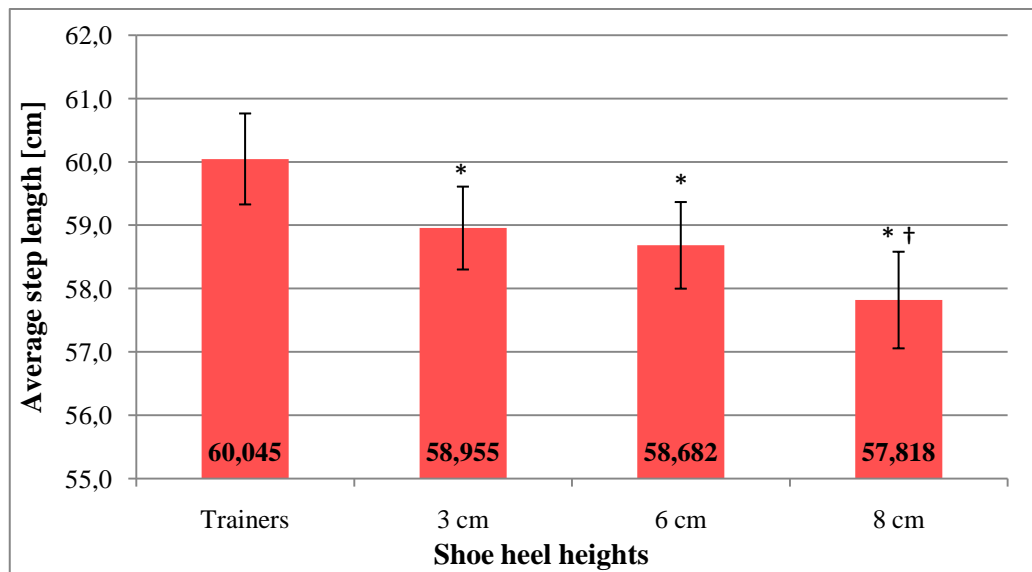
Figure 21. Average (SE) step length [cm] across conditions. SE= standard error.
* Significantly shorter step length when compared to trainers, p<0,02.
† Significantly shorter step length when compared to 3 cm and 6 cm high heels, p<0,05.

5.3.2. Cadence
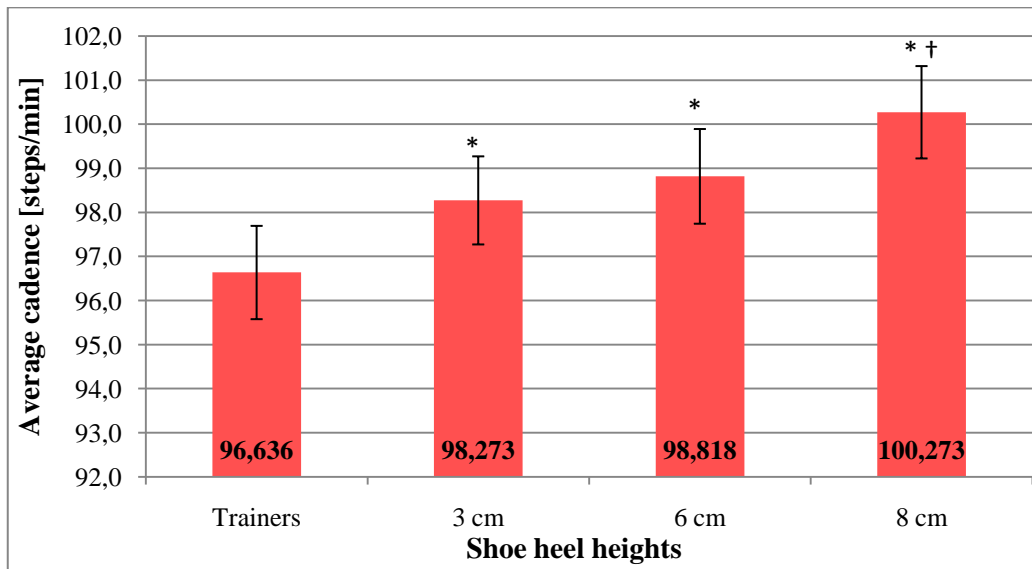
Figure 22. Average (SE) cadence [steps/min] across conditions. SE= standard error.
* Significantly faster cadence than in trainers, $p<0,01$.
† Significantly faster cadence than in 3 cm and 6 cm high heels.

5.3.3. Relationship between step length and cadence

When going from trainers to greater heel heights, an overall significant negative correlation was found between changes in step length and cadence from trainers to 8 cm heels ($r=-0,816$; $p<0,001$): as the heel height increased the steps became shorter and the number of steps per minute increased (Figure 23). This association was significant for each incremental change; from flats to 3 cm ($r=-0,739$; $p<0,001$), from 3 cm to 6 cm ($r=-0,743$; $p<0,001$) and from 6 cm to 8 cm ($r=-0,628$; $p=0,002$).
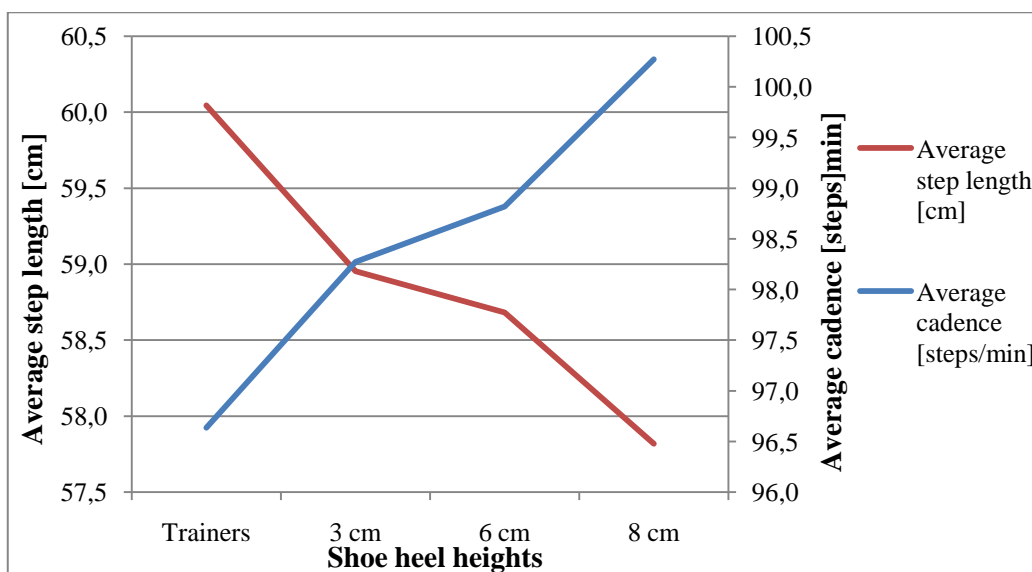
Figure 23. A negative correlation between average step length and cadence across conditions.

5.3.4. Step width
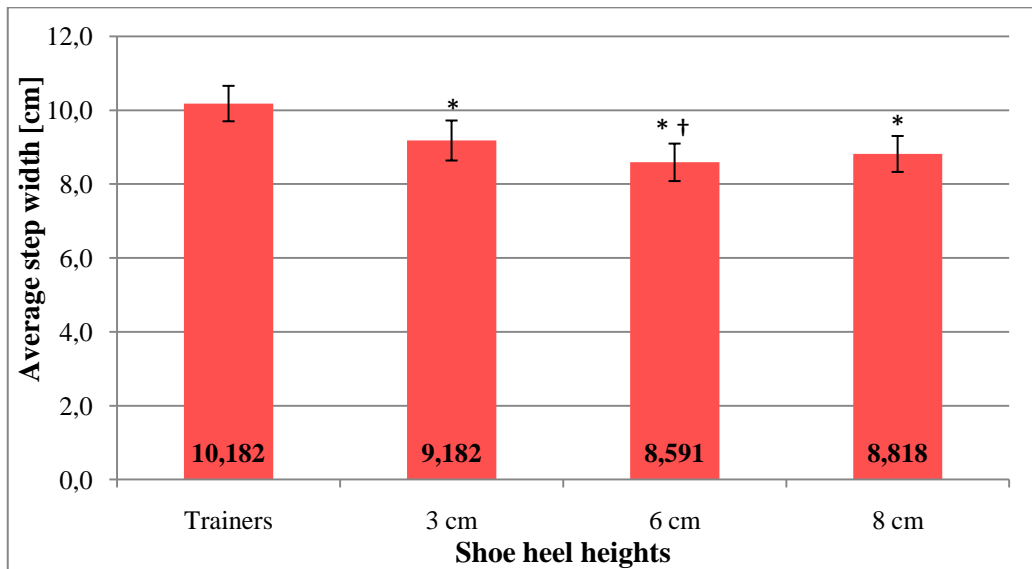
Figure 24. Average (SE) step width [cm] across conditions. SE= standard error.

* Significantly smaller step width compared to flat shoes.

† Significantly smaller step width compared to 3 cm high heels.

5.3.5. Progression angle
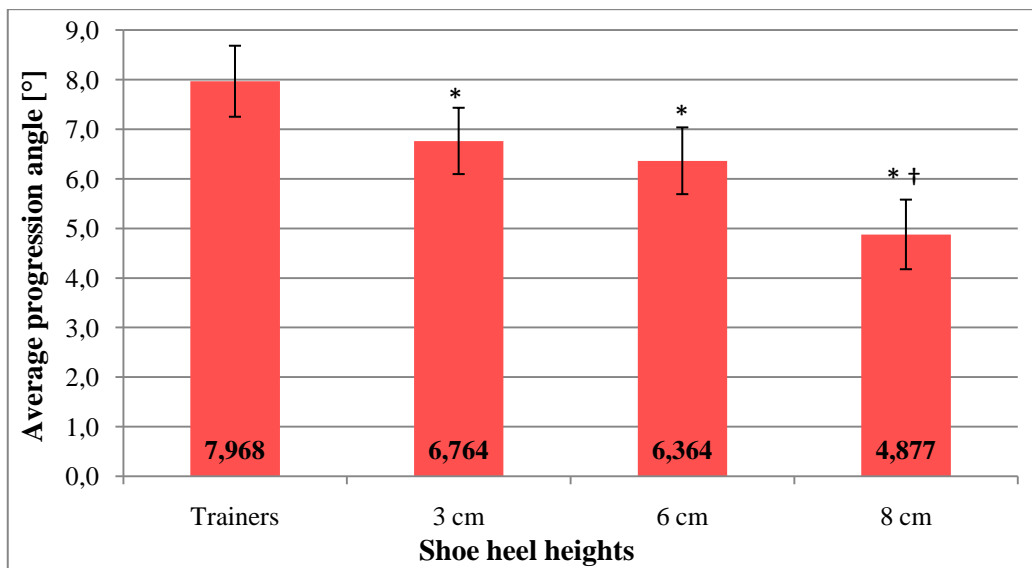
Figure 25. Average (SE) progression angle [°] of the foot in stance phase across conditions. SE= standard error.

* Significantly smaller progression angle compared to flat shoes, p<0,005.

† Significantly smaller progression angle compared to 3 og 6 cm, p<0,003.

5.3.6. Outcome measures compared between regular and non-regular wearers.

When step width and progression angle were compared between regular and non-regular wearers, no interaction between heel-height and groups with respect to step width (regular vs. non-regular: F= 0,260; p= 0,801) was found. Nonetheless a significant difference in average step width was found, in that the non-regular wearers demonstrated overall larger step width compared to regular-wearers (p= 0,019) (Figure 26).

No interaction was found for progression angle between heel-height and groups either (regular vs. non-regular wearers: F= 1,135; p= 0,332), but a significant difference was found in overall average progression angle (p=0,032) (Figure 27).
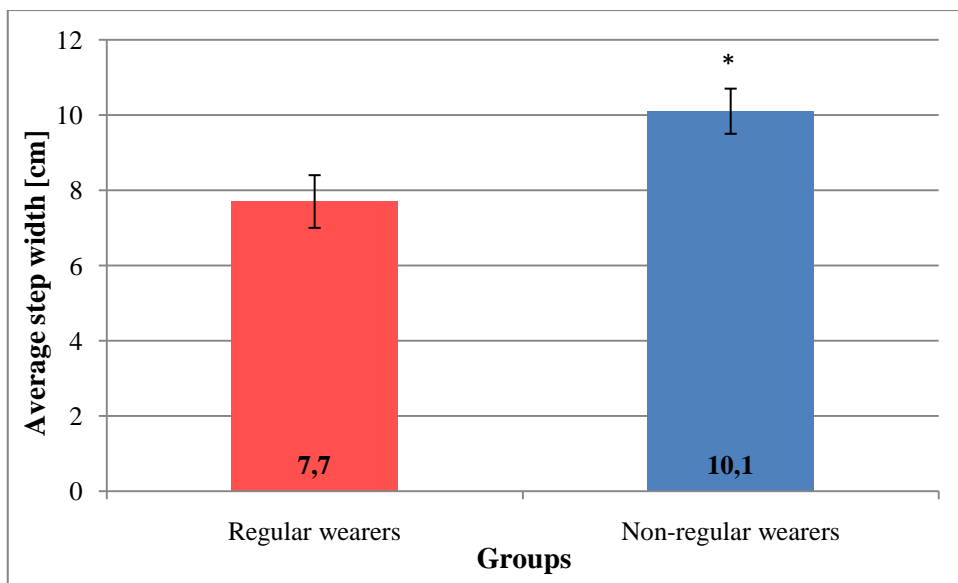
Figure 26. Comparison of overall average (SE) step width [cm] between regular (n= 8) and non-regular (n= 14) wearers (p= 0,019). SE= standard error.
* Significantly larger step width compared to regular wearers.
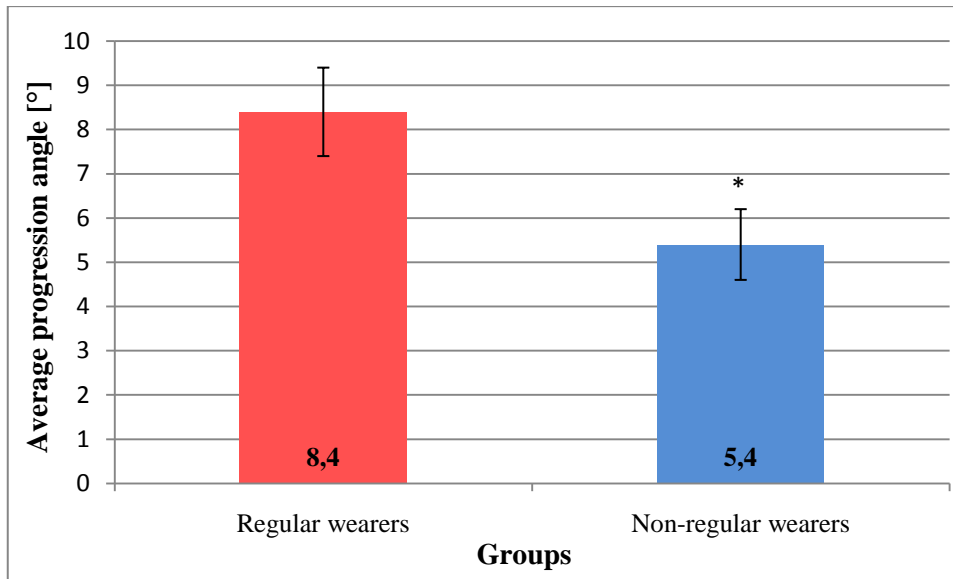
Figure 27. Comparison of overall average (SE) progression angle [°] between regular (n= 8) and non-regular (n=14) wearers. SE=standard error.
* Significantly smaller progression angle compared to regular wearers.

5.4. Push-off force

Significant changes in push-off force were found across conditions (F= 104,691; p<0,001). Post-hoc analysis demonstrated a significant decrease in force when walking in 6 and 8 cm heels compared to trainers, where the force went below body weight (Figure 28).
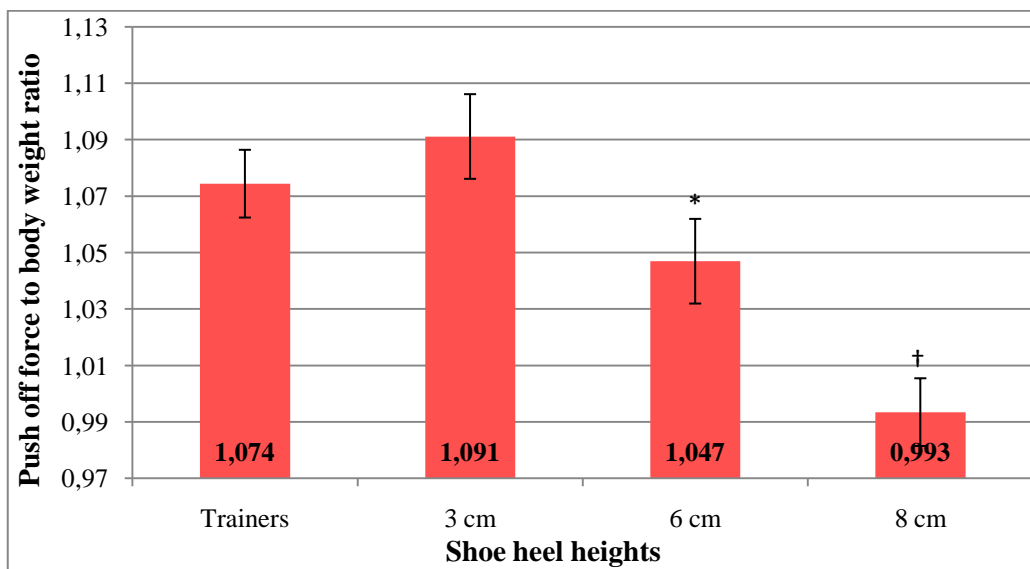
Figure 28. Average (SE) push off force between different heel heights where each number is a ratio of push off force to body weight. SE=standard error.
* Significantly lower push off force compared to trainers and 3 cm high heels; p<0,03.
† Significantly lower push off force compared to trainers, 3 and 6 cm high heels; p<0,001.

6. Discussions

The first hypothesis stated that more muscle activity would be seen in muscles crossing the knee joint. This turned out to be only partially right, where the vastus lateralis was the only muscle showing a significant increase in it´s activity. The results support our second hypothesis, that with increased heel height more muscle activity would be seen in the plantarflexors and fibularis longus. The third hypothesis was refuted as the muscle activity of the tibialis anterior did not change significantly.

6.1. EMG discussion

Incremental increases in heel height significantly affected muscle activation levels of five muscles. Interestingly, they demonstrated two distinct patterns of change in their activity level. Fibularis longus, soleus and vastus lateralis did not significantly increase their activity until the heels became higher than 3 cm, whereas the the gastrocnemius medialis and lateralis, increased their activity as soon as heel height increased. The difference between these two groups can be explained by the fact that only the gastrocnemius muscles cross two joints; the knee and ankle. Given the aforementioned kinematic changes caused by increased heel height, these muscles are put in a slightly shortened state at both joints. This may make it more difficult for them to perform efficiently, even with small increases in heel height.

6.1.1. M. Fibularis longus

One of the main stabilizing muscles of the ankle is the fibularis longus. An increase in its activity is therefore expected when walking on high heels, in part due to less stability at the talocrural joint, but also due to the smaller BoS offered by the shoes. The EMG data supported our hypothesis as a significant increase in the muscles activity was seen between all heel heights (Figure 13).

6.1.2. The ankle plantarflexors

As hypothesized, the ankle plantarflexors significantly increased their activity as heel height increased (Figures 14-16). As stated earlier, these muscles are maintained in a more shortened state as heel height, and thereby ankle plantarflexion, increases. Greater plantarflexion offers less passive stability at the ankle, is a less efficient position for the plantarflexors to work from, and at the same time, the BoS becomes smaller. The resulting

less stable gait, probably explains the change seen in muscle activity. Interestingly, the level of change was not quite the same for each of the three muscles between heel heights. This could mean that although these muscles work together as a whole, their relative workload may vary as heel height increases. All these muscles showed a significant increase in activity when walking on the highest heels (8 cm) compared to walking in trainers. Of these muscles, the gastrocnemius lateralis increased its activity the most. This could both be due to its anatomical location and to its less lateral stability compared to medial at the ankle joint. We therefore conclude that the muscle plays a stabilizing role along with the fibularis longus muscle as heel height increases.

6.1.3. M. Vastus lateralis

The main role of the vastus lateralis muscle is to extend the knee joint. During walking, it works along with other parts of the quadriceps muscle as a shock absorber at the knee, as the lower extremity accepts weight at the beginning of each stance phase of the gait cycle. When walking in low-heeled shoes, the knee fully extends during midstance, and stays passively extended until the leg prepares for swing phase by passively flexing the knee. This means that during normal gait, the vastus lateralis mainly works at the initiation of stance phase. As hypothesized, the muscles activity grew as heel height increased (Figure 18). The relative activation of vastus lateralis when walking in 8 cm high heels was 151% of that found during low-heeled gait. This increase can be explained by the fact that during high-heeled gait, the knee is maintained in a more flexed position during stance phase (Opila-Correia, 1990), and therefore the body has to rely more on muscle work instead of passive support from surrounding joint structure to keep balance and stay upright.

6.1.4. M. Rectus femoris and m. Biceps femoris

Significant difference across shoe condition was not found in the rectus femoris or the biceps femoris muscles (Figure 19-20). Multiple reasons could explain this. Both the biceps femoris and rectus femoris are narrow muscles and can easily move beneath skin during gait, resulting in a movement in and out of the electrodes pick up area. This could also cause crosstalk from adjacent muscle These muscles are biarthrodal and given the kinematic changes that occur through the lower extremity joints, these muscles play an important role across both knee and hip during swing as well as stance. Since there is a change in movement of both the hip and knee, the muscles have an opportunity to decrease work over one joint when it increases in another. Another explanation might be that the

EMG signal does not recognize the difference in concentric, eccentric or isometric contraction: these muscles could alter their muscle firing patterns without the total activity of the EMG signal changing.

6.1.5. M. Tibialis anterior

Hypothesis three stated that less muscle activity would be recorded in the tibialis anterior muscle with increased heel height. The results of the study did not support the hypothesis since no significant difference was found (Figure 17). Intuitively, one might assume that the sustained ankle plantarflexion would call for an altered mechanism for foot clearance during swing, resulting in lower activity in the tibialis anterior muscle. Furthermore, due to greater flexion at the proximal joints at heel strike, it stands to reason that eccentric tibialis anterior contraction from heel strike to foot flat might decrease. However, the average activity did not change, which indicates that the muscle may have assumed a different functional role, possibly in order to stabilize the ankle, thereby maintaining similar levels of activity. Furthermore with increased plantarflexion, the muscle is maintained in a lengthened state, which may alter muscle activity due to a change in the muscle's length-tension relationship.

Finally, one cannot disregard the possibility that no changes in muscle activity were seen simply because there were none. This could be the case for tibialis anterior, as well as the rectus femoris and biceps femoris muscles.

6.2. Temporospatial discussions

Step length decreased as heel height increased, as expected (Figure 21). This result is in accordance with previous studies (Opila-Correia, 1990) and is probably due to greater knee flexion typically seen at initial contact when wearing high-heeled shoes as compared to trainers. Instead of the initial contact being made by the heel, followed by a subsequent rollover, the whole sole touches the ground more simultaneously and the rollover effect is diminished. Therefore, it is not surprising to see the steps becoming shorter with increasing heel height.

Previous studies have shown that when walking in high heels on a self-selected speed, cadence tends to stay the same, resulting in a slower walking speed (Opila-Correia, 1990). In this study, the velocity of the treadmill was maintained the same when walking in

different heel heights. Therefore, when step length decreased, cadence had to increase in order for the participant to keep up with the previously set speed (Figure 22).

Wearing high-heeled shoes challenges balance in general, and the consequences can be seen in many ways. To make up for less balance, a logical response would be to increase the BoS, take shorter steps and slow down the pace. In this study a general shift toward shorter step length and increased cadence was seen. However, contrary to rational thinking, step width decreased as heel height increased (Figure 24). The same results have also been found in previous studies (Opila-Correia, 1990). One can ask oneself why a less stable state should result in a narrower gait and consequently a smaller BoS. A possible explanation is that by placing the foot closer to midline, the centre of mass is placed directly above it. Therefore, swaying and lateral pelvic shift is reduced. Another possible explanation is that this is a vanity related factor. Narrow walk is very prominent in the fashion industry and thought of as appealing. Women could be more self-conscious about their gait when wearing high-heeled shoes, as it is generally considered clumsy, unfeminine and even funny to walk in high heels with a wide BoS. Many comedies have made fun of men walking clumsily in high heels; their BoS is usually wider than when wearing regular shoes and knee flexion is notably increased.

The BoS was not only reduced by the step width but also by the progression angle. When looking at the data it is evident that step width, as well as progression angle, decreases as heel height increases (Figure 24-25). One might think that when the step width is decreased, the progression angle would increase in order to keep the BoS from decreasing.

An interesting pattern was seen when regular high-heel wearers were compared to non-regular wearers. No significant difference was found in interaction between heel-height and groups with respect to step width and progression angle. Nonetheless a significant difference was found overall, where the regular wearers had a narrower gait (Figure 26) and a larger progression angle (Figure 27). Possibly, the regular high-heel wearers made up for smaller step width by increasing their progression angle, and therefore keeping the BoS nearly the same as the non-regular wearers.

6.3. Push-off discussion

Although the ankle plantarflexors showed a significantly greater muscle activity as heel height increased, GRF measured by the treadmill showed a dramatic decrease in push-off force when comparing gait in 8 cm high-heeled shoes to trainers (Figure 28). When

walking in 8 cm high-heeled shoes, the push-off force was less than body weight. We conclude that this low number indicates that the push-off force is extremely limited or close to none. It was surprising to see these results, since, at first glance, they seemed to contradict the EMG data. However, when putting these two factors into context, it seems rational to conclude that even though greater activity is seen in the plantarflexor muscles, their work might not be very effective because of the disadvantaged shortened position. When walking in low-heeled shoes, the ankle plantarflexors eccentrically contract from mid-stance until late stance where the contraction changes to concentric. This lengthening contraction builds up energy that is used during push-off. When the muscle is put in a shortened position, as is the case during high-heeled gait, this energy build up process is less likely to occur and would therefore result in a less efficient push-off. If this conclusion is correct, and the plantarflexors are indeed not working effectively to propel the feet forward, the body must use another way to compensate for the lost force. In such a situation, the hip flexors would have to substitute for diminished push-off, and perform more work in order for the feet to move forward.

Everything regarding high-heeled shoes seems to be related to physical appearance. Most people would agree that high-heeled shoes are less comfortable than low-heeled ones. And, considering the position the feet are maintained in during high-heeled walking, it is an odd fact that anyone would indeed choose to wear them. It is probable that if the appearance factor were not an issue, people in general would not do so.

In the Western world, high heeled shoes are rather common and considered elegant. Women can be seen wearing them almost everywhere - some even wear them all day long.

This common footwear is a fairly strange phenomenon, and the high-heeled gait is most likely something a person habituates to because of society's beauty standards.

7. Limitations

The study took place in a laboratory, which is not a normal everyday surrounding. Participants all knew they were being tested, and may have subconsciously altered their usual gait pattern. Certain environmental factors cannot be overlooked. The walking surface on a treadmill is completely flat and without any bumps or irregularities, which is not always the case in everyday life. Safety bars are present on each side of the treadmill, and they could make people feel more secure while walking.

Previous studies have not found any difference in gait parameters between overground and treadmill walking (Chang, 2009). However, when taking into consideration the fact that the speed is preset and constant, it is likely to affect the muscle work in some way. By having a moving surface, limited push-off might be needed to propel. It may therefore be assumed that the main ankle plantarflexors do not need as much force. As a significant difference was seen in all plantarflexor muscles between conditions in this study, it would be interesting to see if these changes would be the same for overground walking.

The shoes used in this study were not identical. The participants used their own trainers, which were all similar in shape but of different brands. The 6 cm high-heeled shoes had a longer forefoot area than the other heel heights. The heel´s contact area was smallest in the 6 cm high-heeled shoes and largest in the 8 cm high-heeled shoes. It would have been optimal if the all the shoes had had the same shape and contact area.

The electrodes did not make it possible to record muscle activity in muscles lying deeper under the skin, such as the iliopsoas. It would have been optimal to record the activity changes in it but that could only have been possible if needle EMG would have been available.

As stated earlier surface EMG measurements have certain disadvantages. However, this kind of technology offers a good idea of changes in muscle activity and is widely used within many fields.

8. Future studies

- Similar study where the shapes of the shoes and heels are identical between heel heights.
- Similar study where difference in muscle activity of ankle stabilizing muscles is recorded when wearing high-heeled shoes with different heel width.
- Similar study where the pace is self-selected in each heel height.
- Similar study with overground walking.
- Similar study where the data is determined by a time period, instead of a number of gait cycles.
- Similar study where the inclusion criteria is either to have no experience or be a regular wearer to see the difference in muscle activity between the groups.
- Similar study where activity in the erector spinae, gluteus maximus and abdominal muscles are also recorded.
- Similar study where the participants walk in high-heeled shoes for a longer period of time. There are some indications that the fibularis longus muscle has less endurance to fatigue in women who regularly wear high-heeled shoes as compared to non-regular wearers (Gefen, 2002). In our study, the participants only had to walk for under a minute in each heel height. It would be interesting to see if the rise in muscle activity would grow even more after a whole day of walking in high-heeled shoes.

9. Final words

Over the resent years there has been tremendous progress in the development of prosthetic legs. These legs have become more and more advanced, from meeting the demands of normal gait, up to running a marathon. In most developed countries it is considered a normal demand to be able to do most activities people did before their amputation. Therefore, the companies that specialize in making prosthetic legs are always looking for ways to come up with better products and making the life of amputated people closer to normal. One of the latest idea is to make a prosthetic leg with an adjustable ankle designed to be used in different heel heights. In light of the fact that today's prosthetic legs have made it possible for amputees to engage in various sporting activities, it's only a reasonable demand for amputated women to be able to wear high-heeled shoes should they choose to. This study is one part of the preliminary research needed in the developmental process.

Our study aimed at measuring alterations in muscle activity when walking in four different heel heights. The results showed a change in muscle activity in five of the eight muscles measured. The change was most evident in the muscles crossing the ankle joint, with the fibularis longus showing the largest increase. As well as measuring muscle activity, temporospatial changes were observed: progression angle, as well as step length and width decreased, cadence, however, increased with increasing heel height. Push-off force decreased and fell below body weight when walking in the highest heels.

Considering our results, many factors need to be addressed in the development of adjustable prosthetic legs. The manufacturers of these kind of orthopaedic device pride themselves in making outstanding products to improve the quality of life for amputated people. It is only to be expected that these companies will continue to exceed in their product development in the future.

References

Anderson, F. C. and Pandy, M. G. (2003). Individual muscle contributions to support in normal walking. *Gait and Posture, 17*(2), 159-169.

Barkhaus, P. E. and Nandedkar, S. D. (1994). Recording Characteristics of the surface EMG electrodes. *Muscle & Nerve, 17*(11), 1317-1323.

Boakes, J. L. and Rab, G. T. (2006). Muscle Activity During Walking. In J. G. Rose and J. G. Gamble (Ed.), *Human Walking* (3rd ed.) (pp. 103-118). Philadelphia: Lippincott Williams and Wilkins.

Cailliet, R. (2003). Functional Anatomy of the Hip Joint. *The Illustrated Guide to Functional Anatomy of the Musculoskeletal System.* Retrieved February 15th 2011 from: https://catalog.ama-assn.org/Catalog/product/product_detail.jsp?productId=prod170010

Chang, M. D., Shaikh, S. and Chau, T. (2009). Effect of treadmill walking on the stride interval dynamics of human gait. *Gait & Posture, 30*(4), 431-435.

Cooper, R., Morris, H. and Arendt L. (2006). Acute Knee Injuries. In P. Brukner and K. Khan (Ed.), *Clinical Sports Medicine* (3rd ed.) (pp. 460-505). Sydney: McGraw-Hill.

Derrick, T. R. (2004). Signal Processing. In D. G. E. Robertson, G. E. Caldwell, J. Hamill, G. Kamen, and S. N. Whittlesey (Ed.), *Research Methods in Biomechanics* (pp. 227-238). Champaign, IL: Human Kinetics.

Drake, R. L., Vogl, A. W. and Mitchell, A. W. M. (2010). Gray's Anatomy for Students. Philadelphia: Churchill Livingstone -Elsevier.

Edwards, L., Dixon. J., Kent, J. R., Hodgson, D. and Whittaker, V. J. (2008). Effect of shoe heel height on vastus medialis and vastus lateralis electromyographic activity during sit to stand. *Journal of Orthopedic Surgery and Research, 3*(2).

Enoka, R. M. (2002). *Neuromechanics of Human Movement* (3rd ed.). Champaign, IL: Human Kinetics.

Gefen, A., Megido-Ravid, M., Itzchak, Y. and Arcan, M. (2002). Analysis of muscular fatigue and foot stability during high-heeled gait. *Gait and Posture, 15*(1), 56-63.

Gundaker, M. (n.d.). Standing Tall: a short history of the high heeled shoe. In *high heels history 89.* Retrieved August 14th 2010 from: http://hubpages.com/hub/high-heels-history.

Kamen, G. (2004). Electromyographic Kinesiology. In D. G. E. Robertson, G. E. Caldwell, J. Hamill, G. Kamen, and S. N. Whittlesey (Ed.), *Research Methods in Biomechanics* (pp. 163-181). Champaign, IL: Human Kinetics.

Karlsson, J. (2006). Acute Ankle Injuries. In P. Brukner and K. Khan (Ed.), *Clinical Sports Medicine* (3rd ed.) (pp. 612-631). Sydney: McGraw-Hill.

Kaufman, K. R. and Sutherland, D. H. (2006). Kinematics of Normal Human Walking. In J. G. Rose and J. G. Gamble (Ed.), *Human Walking* (3rd ed.) (pp. 33-52). Philadelphia: Lippincott Williams & Wilkins.

Lee, C. M., Jeong, E. H. and Freivalds, A. (2001). Biomechanical effects of wearing high heeled shoes. *International Journal of Industrial Ergonomics, 28*(6), 321-326.

Levangie, P. K. (2005). The Hip Complex. In P. K. Levangie and C. C. Norkin (Ed.), *Joint Structure and Function: A Comprehensive Analysis* (4th ed.) (pp. 354-392). Philadelphia: F. A. Davis Company.

Lexell, J., Henriksson-Larsén, K., Winblad, B. and Sjöström, M. (1983). Distribution of different fiber types in human skeletal muscles: Effects of aging studied in whole muscle cross sections. *Muscle & Nerve, 6*(8), 588-595.

References

Magee, D. J. (2006). *Orthopedic Physical Assessment* (4th ed.). St. Louis, Missouri: Saunders-Elsevier.

Merletti, R. and Farina, D. (2008). Surface EMG processing: Introduction to the Special Issue. *Biomedical Signal Processing and Control, 3*(2), 115-117.

Mueller, M. J. (2005). The Ankle and Foot Complex. In P. K. Levangie and C. C. Norkin (Ed.), *Joint Structure and Function: A Comprehensive Analysis* (4th ed.) (pp. 437-477). Philadelphia: F. A. Davis Company.

Olney, S. J. (2005). Gait. In P. K. Levangie and C. C. Norkin (Ed.), *Joint Structure and Function: A Comprehensive Analysis* (4th ed.) (pp. 517-563). Philadelphia: F. A. Davis Company.

Opila-Correia, K. A. (1990). Kinematics of High-Heeled Gait. *Archives of Physical Medicine and Rehabilitation, 71*(5), 304-309.

Paslawski, M. (2008). From Cavemen to Catwalk. In *Italian Fashion: The History of High Heels.* Retrieved August 14th 2010 from: http://www.lifeinitaly.com/fashion/high-heels.asp.

Rash, G. S. (n.d.). Electromyography Fundamentals. Retrieved February 4th 2011 from: http://myweb.wwu.edu/~chalmers/EMGfundamentals.pdf

Shumway-Cook, A. and Woollacott, M. (2007). *Motor Control: Translating Research into Clinical Practice* (3rd ed.). Philadelphia: Lippincott Williams & Wilkins.

Silverthorn, D. U. (2007). *Human Physiology: An Intergrated Approach* (4th ed.). San Fransisco: Pearson - Benjamin Cummings.

Snyder-Macker, L. and Lewek, M. (2005). The Knee. In P. K. Levangie and C. C. Norkin (Ed.), *Joint Structure and Function: A Comprehensive Analysis* (4th ed.) (pp. 393-436). Philadelphia: F. A. Davis Company.

Stefanyshyn, D. J., Nigg, B. M., Fisher, V., O'Flynn B. and Liu, W. (2000). The Influence of High Heeled Shoes on Kinematics, Kinetics, and Muscle EMG of Normal Female Gait. *Journal of Applied Mechanics, 16*(3), 309-319.

Vigreux, B., Cnockaert, J. C. & Pertuzon, E. (1979). Factors Influencing Quantified Surface EMGs. *European Journal of Applied Physiology, 41*(2), 119-129.

References of figures

Figure 1. FitPro1. Retrieved February 20th 2011 from:
http://www.askthehealthclub.com/worksheets/quads-muscles.html

Figure 2. FitPro1. Retrieved February 20th 2011 from:
http://www.askthehealthclub.com/worksheets/glute-ham-muscles.html

Figure 3. Trainharder.com (2007-2010). Retrieved February 23rd 2011 from:
http://trainharder.com/blogs/pricey/2009/04/11/ims-intramuscular-stimulation/

Figure 4. MindCreators.com (2002). Retrieved April 28th 2011 from:
http://www.mindcreators.com/NeuronBasics.htm

Figure 5. School of Biomedical Engineering, Science and Health Systems (2009).
Retrieved April 20th 2011 from:
http://www.biomed.drexel.edu/labs/biomechanics/emg_analy.htm

Figure 6. Kaufman, K. R. S., D. H. (2006). Kinematics of Normal Human Walking. In J.
G. Rose, J. G. (Ed.), Human Walking (3rd ed.) (p. 40). Philadelphia: Lippincott
Williams & Wilkins.

Figure 7. Medscape Education (2011). Retrieved April 23rd 2011 from:
http://www.medscape.org/viewarticle/736532_3

Figures were adapted by researchers to better suit the study.

Appendix I

Information letter for participants (In Icelandic)

UPPLÝSINGABRÉF

Rafvirkni helstu vöðva neðri útlima kvenna mæld með EMG; samanburður á göngu á háum hælum og berfættri göngu

Kæri viðtakandi,

Fyrirhugað er að hefja ofangreinda rannsókn á vegum sjúkraþjálfunarskors læknadeildar Háskóla Íslands. Rannsóknin er B.S. námsverkefni nema á 4. ári í sjúkraþjálfun. Rannsóknin er hugsuð sem grunnrannsókn fyrir Össur hf. þar sem fyrirtækið hyggst nota niðurstöður til þróunar á nýjum gervifæti.

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Sigríður Katrín Magnúsdóttir, nemi í sjúkraþjálfun við Háskóla Íslands, S: 865 4946

Markmið rannsóknar:

Markmið og tilgangur rannsóknarinnar er að kanna hvernig vöðvavirkni neðri útlima breytist með aukinni hælahæð. Rannsóknir hafa sýnt að ýmsir þættir breytast í göngu kvenna við það að ganga í hælaskóm, svo sem líkamsstaða og hreyfifræði. Minna hefur verið rannsakað hvernig vöðvavirkni breytist við þessar aðstæður, en til þess að hægt sé að

hanna nýja gervifætur sem ætlaðir eru fyrir göngu bæði á lágbotna og háum hælum er þörf á grunnrannsóknum á því sviði.

Þátttöku skilyrði:

Um þægindaúrtak verður að ræða þar sem einungis konur á aldrinum 20-40 ára verða beðnar um að taka þátt. Þátttakendur verða m.a. háskólanemar, vinnufélagar rannsakenda, vinir og vandamenn.

Hvað felst í þátttöku:

Þátttaka í rannsókninni felur í sér að mæta í mælingar á rafvirkni vöðva í neðri útlimum í eitt skipti. Einnig verður hæð og þyngd mæld. Miðað er við að mælingar taki ekki lengri tíma en 60 mínútur.

Lýsing á hvernig rannsóknin fer fram gagnvart þátttakendum og hvað felst í þátttöku:

Hæð og þyngd þátttakenda verða mæld. Elektróður verða festar á húð yfir völdum vöðvum neðri útlimar og rafvirkni þeirra verður mæld. Rafvirkni mælingar fara fram þegar þátttakendur ganga á flatbotna skóm, á 3, 6 og 8 cm hælum.

Áætlað er að þátttakendur verði á bilinu 20-30 og að tíminn sem mælingar fara fram sé ekki meiri en 60 mínútur á hvern þátttakanda. Aðeins þarf að mæta í eitt skipti.

Rannsóknin fer fram í rannsóknarstofu hjá Össur hf. Grjóthálsi 5.

Upplýsingar sem þátttakandi gefur í rannsókninni – trúnaður rannsakenda:

Allar upplýsingar sem þátttakendur veita í rannsókninni, verða meðhöndlaðar samkvæmt ströngustu reglum um trúnað og nafnleynd og farið að íslenskum lögum varðandi persónuvernd, vinnslu og eyðingu frumgagna. Rannsóknargögn verða varðveitt á öruggum stað (tölvukerfi stofnunar (varið með aðgangsorði)) hjá ábyrgðarmanni á meðan á rannsókn stendur og öllum gögnum verði eytt að rannsókn lokinni, nema heimildar verði aflað til að varðveita þau lengur.

Tekið skal fram að þátttakendum er frjálst að hafna þátttöku eða hætta í rannsókninni á hvaða stigi sem er, án útskýringa og án afleiðinga á aðra meðferð.

"Ef þú hefur spurningar um rétt þinn sem þátttakandi í vísindarannsókn eða vilt hætta þátttöku í rannsókninni getur þú snúið þér til Vísindasiðanefndar, Vegmúla 3, 108 Reykjavík. Sími: 551-7100, fax: 551-1444, tölvupóstfang: visindasidanefnd@vsn.stjr.is."

Hvorki fylgir áhætta né ávinningur þátttöku.

Rannsóknin er unnin með samþykki Vísindasiðanefndar og eftir atvikum og tilkynning hefur verið send til Persónuverndar.

Kær kveðja,

með von um góðar undirtektir,

_____   _____

Ásdís Árnadóttir   Inga Hrund Kjartansdóttir

_____   _____

Kristín Briem   Sigríður Katrín Magnúsdóttir

Appendix II

Informed consent form (In Icelandic)

SAMÞYKKISYFIRLÝSING ÞÁTTTAKENDA

Rafvirkni helstu vöðva neðri útlima kvenna mæld með EMG; samanburður á göngu á háum hælum og berfættri göngu

Markmið og tilgangur rannsóknarinnar er að kanna hvernig vöðavirkni neðri útlima breytist með aukinni hælahæð. Rannsóknir hafa sýnt að ýmsir þættir breytast í göngu kvenna við það að ganga í hælaskóm, svo sem líkamsstaða og hreyfifræði. Minna hefur verið rannsakað hvernig vöðvavirkni breytist við þessar aðstæður, en til þess að hægt sé að hanna nýja gervifætur sem ætlaðir eru fyrir göngu bæði á lágbotna og háum hælum er þörf á grunnrannsóknum á því sviði.

Þátttaka í rannsókninni felur í sér að mæta í mælingar á rafvirkni vöðva í neðri útlimum í eitt skipti. Einnig verður hæð og þyngd mæld. Miðað er við að mælingar taki ekki lengri tíma en 60 mínútur.

Ég staðfesti hér með undirskrift minni að ég hef lesið upplýsingarnar um rannsóknina sem mér voru afhentar, hef fengið tækifæri til að spyrja spurninga um rannsóknina og fengið fullnægjandi svör og útskýringar á atriðum sem mér voru óljós. Ég hef af fúsum og frjálsum vilja ákveðið að taka þátt í rannsókninni. Mér er ljóst, að þó ég hafi skrifað undir þessa samstarfsyfirlýsingu, get ég stöðvað þátttöku mína hvenær sem er án útskýringa og án áhrifa á þá læknisþjónustu sem ég á rétt á í framtíðinni.

Mér er ljóst að rannsóknargögnum verður eytt að rannsókn lokinni og eigi síðar en eftir 5 ár frá úrvinnslu rannsóknargagna. Mér hefur verið skýrt frá fyrirkomulagi trygginga fyrir þátttakendur í rannsókninni.

_____

Dagsetning

_____

Nafn þátttakanda

Undirritaður, starfsmaður rannsóknarinnar, staðfestir hér með að hafa veitt upplýsingar um eðli og tilgang rannsóknarinnar, í samræmi við lög og reglur um vísindarannsóknir.

_____

Ásdís Árnadóttir

_____

Inga Hrund Kjartansdóttir

_____

Sigríður Katrín Magnúsdóttir

Appendix III

Question form (In Icelandic)

SPURNINGALISTI TIL ÞÁTTTAKENDA

Aldur:

Hæð:

Þyngd:

Skóstærð:

Með hvorum fætinum sparkarðu í bolta?

Reynsla af göngu á háum hælum:

( ) < 7 klst á viku

( ) > 7 klst á viku

Appendix IV

Electrode placement according to SENIAM guidelines

Muscle	Placement
m. Gluteus medius	At 50% on the line from the crista iliaca to the trochanter major.
m. Rectus femoris	At 50% on the line from the anterior spina iliaca superior to the superior part of the patella.
m. Vastus lateralis	At 2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella.
m. Biceps femoris	At 50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia.
m. Fibularis longus	At 25% on the line between the tip of the head of the fibula to the tip of the lateral malleolus.
m. Tibialis anterior	At 1/3 on the line between the tip of the fibula and the tip of the medial malleolus.
m. Gastrocnemius medialis	On the most prominent bulge of the muscle.
m. Gastrocnemius lateralis	At 1/3 of the line between the head of the fibula and the heel.
m. Soleus	At 2/3 of the line between the medial condyle of the femur to the medial malleolus.