Walking on High Heels Changes Muscle Activity and the Dynamics of Human Walking Significantly

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The aim of the study was to investigate the distribution of net joint moments in the lower extremities during walking on high-heeled shoes compared with barefooted walking at identical speed. Fourteen female subjects walked at 4 km/h across three force platforms while they were filmed by five digital video cameras operating at 50 frames/second. Both barefooted walking and walking on high-heeled shoes (heel height: 9 cm) were recorded. Net joint moments were calculated by 3D inverse dynamics. EMG was recorded from eight leg muscles. The knee extensor moment peak in the first half of the stance phase was doubled when walking on high heels. The knee joint angle showed that high-heeled walking caused the subjects to flex the knee joint significantly more in the first half of the stance phase. In the frontal plane a significant increase was observed in the knee joint abductor moment and the hip joint abductor moment. Several EMG parameters increased significantly when walking on high-heels. The results indicate a large increase in bone-on-bone forces in the knee joint directly caused by the increased knee joint extensor moment during high-heeled walking, which may explain the observed higher incidence of osteoarthritis in the knee joint in women as compared with men.

Keywords: high-heeled shoes, walking, inverse dynamics, EMG

In a historic perspective women started to wear high-heeled shoes more than 400 years ago and for the last 250 years several authorities have warned about the unhealthy aspects of high-heeled shoes (Linder & Saltzman, 1998). Today millions of women wear these shoes on a daily basis and although it has been suggested that the use of high-heeled shoes may lead to an increased incidence of osteoarthritis (Kerrigan et al., 1998, 2001) as seen in women compared with men (Felson, 1988; Davis et al., 1991; Katz et al., 1996), it is still largely unknown how high-heeled walking affects the dynamics of human walking. Eisenhardt et al. (1996) measured pressure distribution under the foot for bare feet versus heel height and found increases in pressure distribution related to heel height. Opila-Correia (1990a, 1990b) studied the kinematics of high-heeled walking and found that high-heeled walking was associated with increased knee joint flexion in the stance phase. Kerrigan et al. (2001) used biomechanical gait analysis and inverse dynamics to evaluate joint loadings during high-heeled walking. In the latter study the peak plantar flexor moment about the ankle joint was reduced with high-heeled shoes. At the knee joint (sagittal plane), Kerrigan et al. (2001) reported an external extensor moment about midstance, which was increased significantly wearing high-heeled shoes. This moment corresponds to a flexor moment speaking about internal muscle moments as calculated by the method of Vaughan et al. (1992). The two extensor moment peaks, as normally seen during human walking in the first and the second half of stance, were unchanged between barefooted and high-heeled walking indicating that knee joint loading could be considered unchanged in the sagittal plane. However, an external varus knee joint moment in the frontal plane was reported to increase. Esenyel et al. (2003) found also a reduced ankle joint moment (internal), an unchanged knee joint extensor moment (internal) in the sagittal plane and an increased varus knee joint moment (external) in the frontal plane. We noticed that use of high-heeled shoes increased varus/valgus moments about the knee joint, the plantar flexor ankle joint moment decreased, but the extensor moment about the knee joint remained unchanged. We also noted that the previous studies used preferred walking speed, in the study of Esenyel et al. (2003) speed was 6% lower.

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during high-heeled walking, while Kerrigan et al. (2001) did not report the actually preferred walking speeds and the subjects were allowed to use their own high-heeled shoes. It has been shown that especially the knee joint moment is highly influenced by walking speed (Holden et al., 1997; Browning and Kram, 2007) and that reduction of the ankle joint plantar flexor moment will affect the knee joint moment directly toward increased extensor dominance (Simonsen et al., 1997). Accordingly, we decided to reinvestigate the joint moments associated with walking on high-heeled shoes but at a fixed and controlled walking speed, and to increase the possibilities for interpretation of the joint moments it was decided also to record EMG from relevant muscles. The hypothesis was that the ankle joint moment would be reduced due to the more plantar-flexed position caused by the high-heeled shoe and further that this reduction would lead to compensatory increased joint moments in the knee joint and/or the hip joint.

Method

Fourteen healthy females volunteered to participate in the experiments. They all gave their informed consent to the procedures, which were approved by the local ethics committee. The subjects were divided into two groups of seven subjects according to frequency of using high-heeled shoes. Group E+ (experienced) consisted of 9 subjects using high heeled shoes 3.8 (1.5–7.0) times per week, while group E– consisted of 9 subjects with less experience using high heeled shoes on average 0.5 (0–1) per week. Mean age of all subjects was 27 (21–38) y, mean body weight 63 (48–85) kg, body height 1.69 (1.58–1.83) m. No statistically significant differences were found between the two groups regarding these bodily measures.

A 10-m-long gait laboratory with five digital video cameras and three recessed force platforms were used to record the gait data. The subjects were taught to walk at 4.0 kph ±10% by immediate feedback on walking speed, which was measured by photocells. Approximately 15 trials were recorded for each condition. Three trials with barefoot walking and three trials with high-heeled shoes were selected for analysis, trials closest to the desired walking speed were chosen.

The shoes had a stiff wooden sole and a 9 cm high heel standing on a surface of approximately 1 cm² (Figure 1). As a sort of sandal, the shoe was strapped firmly to the foot and ankle allowing the reflective markers to remain on the foot and not on the shoe.

Fifteen reflective markers were placed on anatomical landmarks according to Vaughan et al. (1992). Five digital video cameras (Canon MW600) operating at 50 frames per second were used to record the movements. The first two force platforms (AMTI OR6–5–1) were used to measure ground reaction forces in three directions as well as center of pressure in two directions. The third platform was only used to establish the end of the gait cycle. Signals from the force platforms were sampled at 1000 Hz and later reduced to 50 Hz to match the video recordings. Synchronization between video and analog signals was performed by a custom build device, which was triggered by heel strike on the first force platform and output a short electronic signal to the sound track on all five video cameras. The Ariel Performance Analysis System (Ariel Dynamics, San Diego, USA) recognized the synchronization signal. A complete gait cycle and 20 frames (400 ms) before heel strike were digitized. Three trials were analyzed for each subject.

After digitization of the markers, three-dimensional coordinates were constructed by direct linear transformation using the Ariel Performance Analysis System. All coordinates were low pass filtered at 6 Hz by a 4th-order Butterworth filter. The filtered coordinates were input to software written in MatLab together with data from the force platforms. Three-dimensional joint moments were then calculated by inverse dynamics according to Vaughan et al. (1992). Systematic peaks from the time course pattern of joint angles and joint moments in the sagittal plane were extracted. These peaks were easy to identify and have been used previously by, for example, Winter (1988), Pedersen et al. (2004) and Henriksen et al. (2006) (Figure 2). In the following paragraphs all joint moments will be referred to as internal moments.

On a separate day electromyographic (EMG) activity was recorded from eight leg muscles on the right leg: soleus (SOL), gastrocnemius medialis (GAM), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF) and semimembranosus (SM). Bipolar recordings were obtained using surface electrodes (silver-silver chloride) with a fixed interelectrode distance of 2 cm (Multi Bio Sensors, Texas, USA). The electrodes were positioned over the muscles according to

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Figure 1 — The high-heeled shoe with a heel height of 9 cm was used by all subjects.
Before mounting, the skin was shaved and cleaned with pure alcohol. The electrodes were connected to light-weight preamplifiers (input impedance 1 GΩ) each instrumented with an analog to digital converter with 16 bit resolution. The digitized signals were transmitted through thin wires to a small box (weight 70 g) fixed to the back of the subject (MQ16, Marq-Medical, Farum, Denmark). The signals were band-pass filtered (10–1000 Hz) and transmitted wireless using Bluetooth technology to a PC where they were sampled at 1000 Hz and stored for further analysis.

Maximal EMG activity (maxEMG) was recorded during maximal isometric contractions performed in an isokinetic dynamometer (KinCom, Chattex, Chattanooga, TN, USA).

EMG during walking was sampled at 1000 Hz while the subjects were walking on a treadmill (Technogym, Runrace HC1200) at 4 km/h with high-heeled shoes and barefooted, respectively. A foot-switch was placed under the heel and data were sampled for 2 min in each situation.

Fifteen gait cycles were selected from treadmill walking and averaged. Before averaging the EMG signals were full wave rectified and lowpass filtered at 15 Hz to form linear envelopes. The recordings from maximal voluntary contractions were treated exactly the same way and the maximum EMG amplitude was determined for each muscle. These maxEMG values were later used to express EMG during walking relative to maxEMG. The following parameters were extracted from the linear envelopes: peak amplitude, mean amplitude and integrated EMG (IEMG). Onset and offset of activity periods were determined by visual judgment. IEMG was defined as the area under the linear envelope during the period of activity.

The parameters from kinematics and kinetics were entered to a so-called mixed model (repeated measures) (SAS software). The analysis focused in the fixed effects of experience (two levels) and shoe condition (two levels) and their interaction. Three trials were entered for each subject and the level of significance was set to 5%. As 15 gait cycles were averaged for the EMG analysis, these data were tested separately by a nonpaired t test (SPSS software) between groups of subjects and a paired t test between walking with and without shoes, respectively. Level of significance was set to 5%.

**Results**

No significant differences were found between the group of experienced and the group of inexperienced subjects. Accordingly, all data presented are from 14 subjects walking with and without shoes, respectively.

No differences were found between the two walking conditions regarding either the duration of the gait cycle,
the duration of the stance phase, stride length, cadence or velocity. Overground walking velocity was 4.15 km/h with shoes and 4.19 km/h without shoes, a difference of 1%. Stride length during overground walking was 0.68 m without shoes and 0.67 m with shoes. Correspondingly cadence was 1.71 Hz and 1.70 Hz, respectively. On the treadmill stride length was significantly shorter: 0.60 m without and 0.59 m with shoes, while cadence was significantly higher 1.88 Hz and 1.92 Hz, respectively. Joint moments and joint angles for all subjects are illustrated in Figures 3 and 4. The ankle joint moment showed a higher dorsiflexor moment (A1) just after heel strike \((p < .0001)\) and a much lower plantar flexor moment (A2) later in the stance phase during push off \((p < .0001)\) when walking on high-heeled shoes. For the knee joint moment, the first extensor peak (K2) was double as high with shoes than without shoes \((p < .0001)\). The point K3 in the middle of the stance phase is most often negative (flexor dominance) during normal walking. This was also seen in the current study when walking without

![Graphs showing joint moments and angles](image)

Figure 3 — Joint moments and joint angles in the sagittal plane for ankle, hip and knee joint averaged for 14 subjects with three trials per subject. Moments are relative to body weight and the duration of the stance phase is normalized to 100%. Solid lines represent walking on high-heeled shoes and dotted lines represent barefooted walking. Asterisks indicate statistically significant differences.
shoes (Figure 3). However, when walking on high-heeled shoes K3 remained an extensor moment with a statistically significant difference to barefooted walking ($p < .0001$) (Figure 3). In the frontal plane the knee abduction moment increased moderately but significantly ($p < .025$) at peaks K1 and K3 (Figure 4). The same pattern was seen for the hip abduction moment with an additional increase in K2 in the middle of the stance phase while the hip joint angle was unchanged (Figure 4).

As expected the ankle joint angle showed very different values between the two conditions. The ankle angle at heel strike (A0), peaks A1, A2 and A3 were significantly different ($p < .0001$) as the shoe condition forced the foot into a generally more plantar-flexed position (Figure 3). At the knee joint the joint angle at heel strike (K0) was more flexed ($p < .0001$) with shoes and this was also the case for the peak K1 representing peak knee flexion during the first half of the stance phase ($p < .0001$) (Figure 3). No differences in hip joint angles were observed.

A comparison of averaged ensemble EMG between barefooted and high-heeled walking is shown in Figure 5. For TA, SOL, VL, RF, BF and SM, peak amplitude was significantly higher during high-heeled walking. Mean amplitude and IEMG were significantly higher during high-heeled walking except for TA and GAM (Table 1). GAM showed two distinct periods of activity during high-heeled walking (Figure 5), but only the second GAM period located to the stance phase was quantified for IEMG, mean and peak amplitude.

**Discussion**

In the sagittal plane the knee joint extensor moment in the first half of the stance phase increased by approximately 100% from barefooted to high-heeled walking (Figure 3), which supported the original working hypothesis. One reason for this could be that a significant increased flexion of the knee joint was seen in the first half of the stance phase (Figure 3). This is contrary to the study of Esenyel et al. (2003) and Kerrigan et al. (1998), but similar to the results of Opila-Correia (1990b). Increased flexion of the knee joint has been associated with increased knee joint extensor moments (Alkjaer et al., 2003). Another reason could be the controlled walking speed used in the current study as opposed to previous studies. In the study of Kerrigan et al. (2001) self-selected speed was used and not reported numerically, and the internal extensor moment did not increase when wearing high-heeled shoes despite a heel height of 6 cm. In the study of Esenyel et al. (2003) the knee joint internal extensor moment was unchanged, heel height was 6 cm and self-selected walking speed dropped 6% when wearing high heels. In the current study the velocity was only 1% lower on average in the high-heeled condition. Walking speed has been shown to have a significant influence on joint moments in the sagittal plane (White & Lage, 1993) while not in the frontal plane (Kirtley et al., 1985). Holden et al. (1997) demonstrated a substantial influence of walking speed on the knee joint extensor moment as increasing the speed from 3.5 kph to 4.7 km/h (25% increase) almost doubled the knee joint extensor moment. However, the 6% difference in speed observed in the study of Esenyel et al. (2003) is not likely solely to have prevented the knee joint extensor moment from increasing in the high-heeled condition. The unchanged knee joint flexion is a more
Figure 5 — Ensemble averaged linear envelope EMG from 14 subjects (15 gait cycles each). The EMG amplitude is expressed relative to the maximal EMG amplitude measured during isometric contractions. The gait cycle is normalized to 100% and indicated by vertical lines. Solid lines represent high-heeled walking and dotted lines barefooted walking. Several statistically significant differences were observed.
likely candidate. In the current study the large increase in knee joint extensor moment when walking on high heels was not caused by changes in walking speed. The increased knee extensor moment was accompanied by significant increases in EMG amplitude and IEMG in the quadriceps muscle during the first half of the stance phase (Figure 5 and Table 1).

We found that the ankle joint moment decreased significantly from barefooted to high-heeled walking (Figure 3). This supported our initial hypothesis and corroborates the findings of Kerrigan et al. (2001) and Esenyel et al. (2003). Reduced muscle fiber length and a reduced moment arm for the Achilles tendon with respect to the ankle joint at the more plantar-flexed position could be a reasonable explanation. However, the most likely reason could be that the ground reaction vector passed closer to the ankle joint center in the high-heeled condition, thus decreasing the need for a large plantar flexor moment (Simonsen et al., 1997).

In general the EMG recordings showed significantly increased leg muscle activity when walking on high heels as also seen for trunk muscles (Barton et al., 2009). This supports the observation by Ebbeling et al. (1994) that metabolic energy cost increased when walking on high-heeled shoes.

It is worth noticing that small but significant differences were observed between overground and treadmill walking regarding cadence and stride length. These parameters have been shown to influence the sagittal knee joint moment significantly (Umberger & Martin, 2007), however, cadence was varied as much as ±20%. It has been a common approach to study EMG during walking by averaging several step cycles recorded on a treadmill (e.g., den Otter et al., 2004), and comparing to EMG studies of overground walking (Arsenault et al., 1986; Hof et al., 2005), the general patterns look alike, but some differences between the two tasks have been reported. Lee and Hidler (2008) found a lower dorsiflexor moment and knee extensor moment during treadmill walking compared with overground walking. In the frontal plane no differences were reported. Moreover, EMG activity was lower in TA but similar in quadriceps during the stance phase (Lee & Hidler, 2008).

It is also important to notice that the large differences in joint kinematics about the ankle joint with and without shoes will have influenced the EMG signals recorded on the lower leg in some way, however unpredictable and unknown. The kinematics of the knee joint showed much smaller differences, so it is likely that the EMG signals from the thigh muscles are more comparable between the two walking conditions.

The Kerrigan group presents all moments as external moments (1998; 2001), which is opposite to several other groups (e.g., Pedotti, 1977; Winter, 1988; Vaughan et al., 1992; Simonsen et al., 1997; DeVita & Horthobágyi, 2003). However, when the method of calculation is inverse dynamics, both type of moments are exactly the same, it is only a matter of terminology when dealing with the muscular interpretation of the moments. Several authors prefer to call the knee joint moment in the frontal plane for an external adduction moment (Schipplein & Andriacchi, 1991; Esenyel et al., 2003; Henriksen et al., 2006; Foroughi et al., 2009a, 2009b). Accordingly, an external moment attempts to adduct the knee joint, i.e., resist abduction, and this adduction moment is most probably caused by the tibial collateral ligament while the abduction moment may come from the quadriceps muscle although this is an extensor muscle. The reason for this would be that the quadriceps inserts lateral to the joint center located in the middle of the medial femoral condyle as suggested by Schipplein and Andriacchi (1991). In the current study we have chosen to follow the same convention as with the other internal moments, which means that a positive knee joint moment in the frontal plane is termed abductor dominance or a valgus moment. This moment is considered very important regarding joint degradation due to osteoarthritis. If the knee joint is forced into varus position, the lateral compartment of the joint will

Table 1 Mean EMG amplitude (amp), peak EMG amplitude, IEMG, onset and offset

<table>
<thead>
<tr>
<th></th>
<th>Mean Amp High Heels</th>
<th>Mean Amp Barefoot</th>
<th>Peak Amp High Heels</th>
<th>Peak Amp Barefoot</th>
<th>IEMG High Heels</th>
<th>IEMG Barefoot</th>
<th>Onset High Heels</th>
<th>Onset Barefoot</th>
<th>Offset High Heels</th>
<th>Offset Barefoot</th>
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<td>TA</td>
<td>11.8</td>
<td>10.5</td>
<td>25.2</td>
<td>19.3*</td>
<td>27.7</td>
<td>29.4</td>
<td>–39.7</td>
<td>–26.2</td>
<td>10.1</td>
<td>18.0</td>
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<tr>
<td>SOL</td>
<td>26.5</td>
<td>19.3**</td>
<td>59.7</td>
<td>40.9**</td>
<td>60.2</td>
<td>37.3**</td>
<td>1.3</td>
<td>–6.2</td>
<td>59.5</td>
<td>60.3</td>
</tr>
<tr>
<td>GAM</td>
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<td>17.8</td>
<td>44.0</td>
<td>42.3</td>
<td>40.8</td>
<td>35.5</td>
<td>5.1</td>
<td>–5.3</td>
<td>57.9</td>
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<tr>
<td>VM</td>
<td>7.7</td>
<td>3.9**</td>
<td>20.3</td>
<td>14.0</td>
<td>15.7</td>
<td>7.6*</td>
<td>–17.7</td>
<td>–16.8</td>
<td>27.4</td>
<td>34.4</td>
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<tr>
<td>VL</td>
<td>7.5</td>
<td>4.0**</td>
<td>20.6</td>
<td>12.5**</td>
<td>17.2</td>
<td>8.1**</td>
<td>–16.4</td>
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<td>2.5**</td>
<td>14.6</td>
<td>7.9*</td>
<td>14.4</td>
<td>5.3**</td>
<td>–16.6</td>
<td>–12.1</td>
<td>28.9</td>
<td>44.8</td>
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<tr>
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<td>7.0**</td>
<td>22.3</td>
<td>14.8**</td>
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<td>–24.4</td>
<td>–25.1</td>
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<td>14.4</td>
</tr>
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Note. Amplitudes are in percentages of maxEMG. IEMG is in µV·s. Onset and offset are in percentages of the gait cycle.

*Indicates p < 0.05 and **indicates p < 0.01.
“open” and the whole loading will be concentrated on the articular surfaces of the medial compartment (Schipplein & Andriacchi, 1991; Kerrigan et al., 2001; Foroughi et al., 2009a). In the current study the knee joint moment in the frontal plane showed abductor dominance (Figure 4) and increased significantly about 10% when wearing high-heeled shoes. Similar observations have been reported by Kerrigan et al. (1998, 2001, 2003, 2005). This 10% increase may seem moderate compared with the 100% increase in the knee joint extensor moment. However, it has been calculated that 1% increase in the knee joint abduction moment increases the risk of progression of osteoarthritis by 6.46 times (Miyazaki et al., 2002).

The results of the current study showed no differences between experienced and inexperienced subjects regarding high-heeled walking, neither for kinematics or kinetics. This is in opposition to Opila-Correia (1990a, 1990b) who reported a small but significant increase in knee joint flexion during the stance phase of high-heeled walking in inexperienced subjects. The same studies found no differences in the kinematics of the hip joint in the frontal plane corroborating the current study.

In the current study high-heeled walking was compared with barefooted walking, but a realistic alternative for women wearing 8–9 cm high-heeled shoes would probably be to wear shoes with a significantly lower heel. Kerrigan et al. (2005) investigated shoes with 3.8 cm heel height and found 7% increased extensor moment (internal) in young women and 14% increased knee abduction moment (internal) in elderly women. It was concluded that even moderate high-heeled shoes may contribute to development of osteoarthritis in the knee joint (Kerrigan et al., 2005). Shoes have been shown in general to change the foot motion (Morio et al., 2009), but men’s dress shoes and sneakers have been reported not to change the knee joint moment compared with barefoot walking (Kerrigan et al., 2003).

It is concluded that reduction of the net ankle joint moment during push off (high-heeled walking) was probably caused by the remarkably more plantar-flexed ankle joint (position of the foot) thereby decreasing the need for a large plantar flexor moment. The 100% increased knee joint extensor moment could also be caused by the increased knee joint flexion during the stance phase and increased EMG in the quadriceps muscle. The knee joint abduction moment in the frontal plane also increased significantly by app. 10% in the high-heeled condition. The results indicate a large increase in bone-on-bone forces in the knee joint caused by high-heeled walking, which may explain the observed higher incidence of osteoarthritis in the knee joint in women as compared with men.

**References**


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